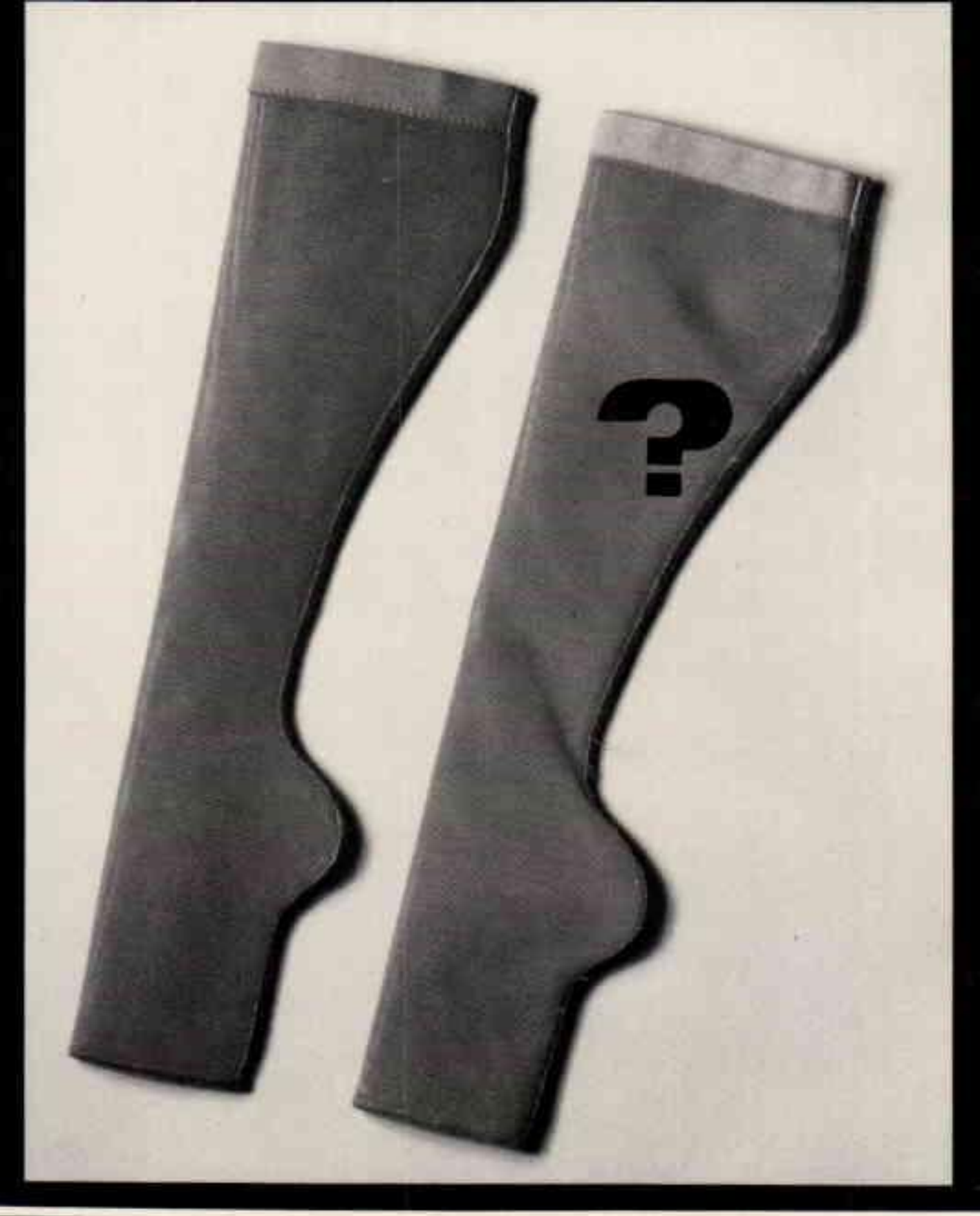
Summer 1982
Volume 36
Number 2

Orthotics and Prosthetics

Journal of the American Orthotic and Prosthetic Association



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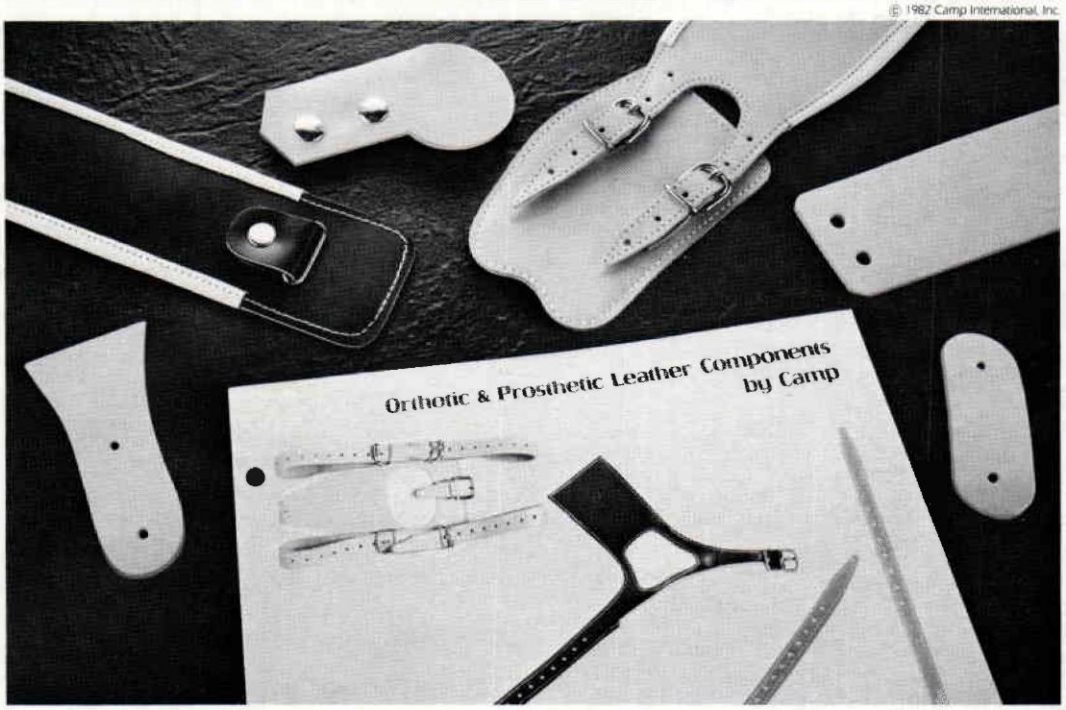

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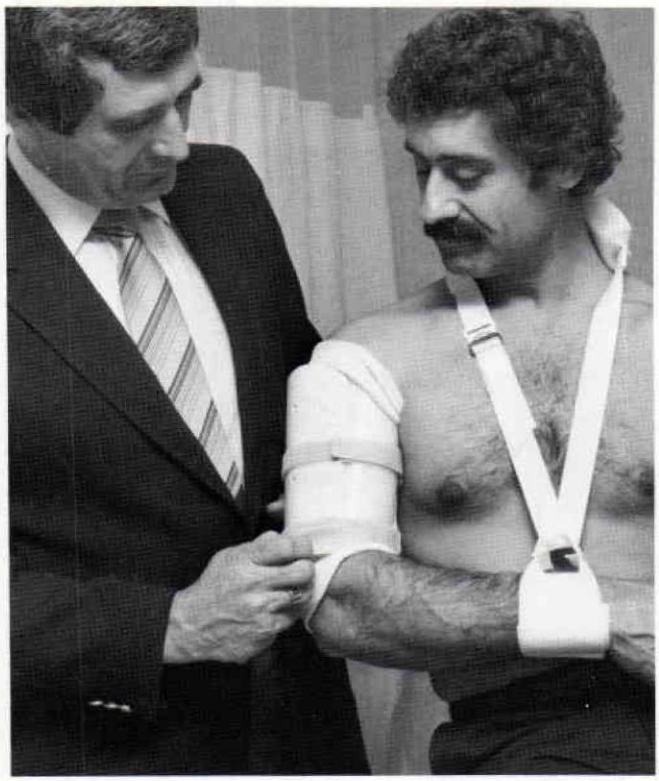

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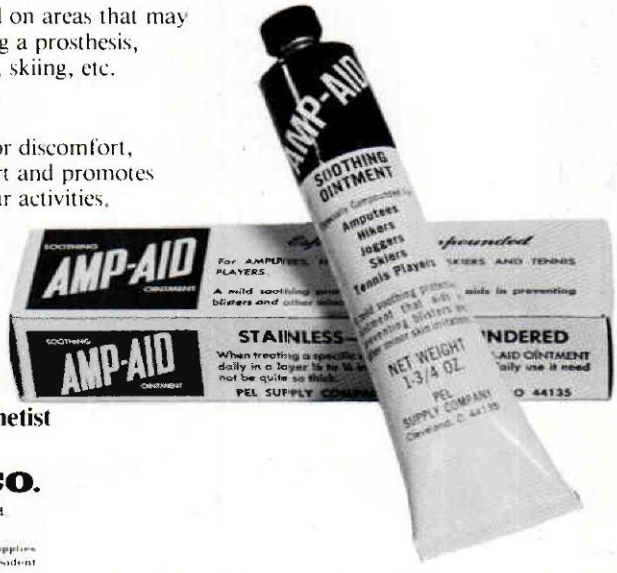

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1982, November 12-13, Third Southern California Course on the "Current Concepts—the Diagnosis and Care of the Patient with Neuromuscular Disorder," Rancho Los Amigos Hospital, Downey, California.

1982, November 12-13, Workshop, sponsored by Freeman Manufacturing Company, Century Airport Inn, Atlanta, Georgia.

1982, December 5-8, American Medical Association's Meeting of the House of Delegates, Fountainbleu Hilton, Miami, Florida.

1982, December 6-8, Department of Orthopaedics and Rehabilitation, University of Miami Medical School PostGraduate Course "New Technology in Orthopaedics and Rehabilitation," Sheraton Bal Harbour, Miami Beach, Florida.

1982, December 10-11, Florida Association of Orthotists and Prosthetists Hands On Computer Workshop, Hilton Inn Florida Center, Orlando, Florida.

1983, January 26-30, AAOP Annual Meeting, Hyatt Islandia, San Diego, California.

1983, February 17-19, Seating the Handicapped Child," International Seating Symposium, Instructional Resources Centre, University of British Columbia, Vancouver, British Columbia, Canada.

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1983, September 5-9, The IV World Congress of the International Society for Prosthetics and Orthotics, Imperial College of Science and Technology, London, England.

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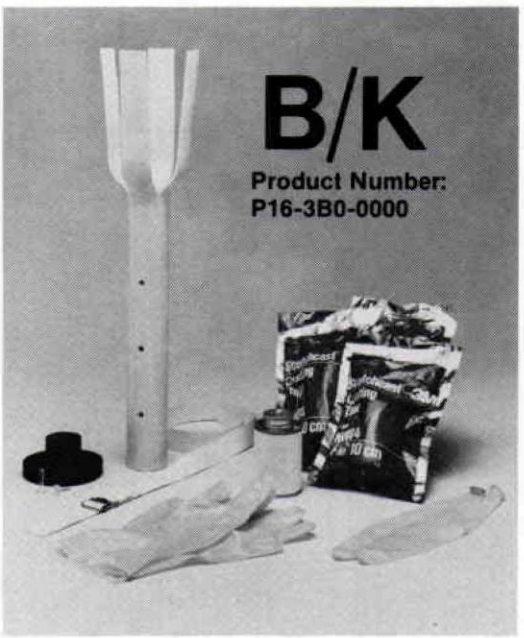

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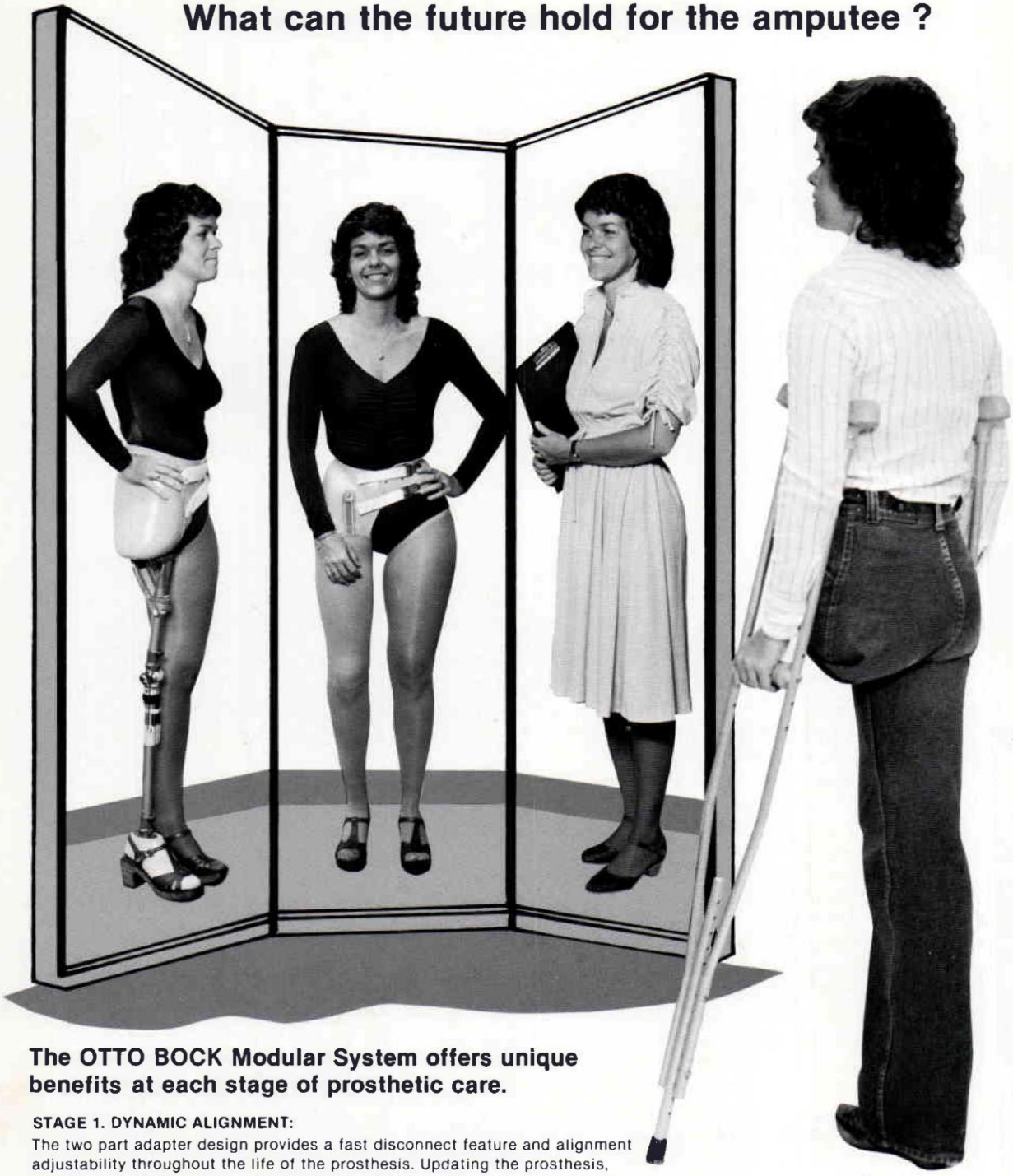

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A Modular Below Elbow Prosthesis For Children

John Hodgins, C.P.O.
Richard Sullivan, M.D.
Sudesh Jain, M.D.

INTRODUCTION

In the history of prosthetic design and development, the ability to extend the life of a child's prosthesis has only been partially solved. The ideal design would include the ability to increase both the prosthetic length and girth without having to completely replace the prosthesis. The parameters for length of the upper limb have been discussed in detail by Lustid and Keats[1], Wallis[2], and Meredith[3].

Exoskeletal lower limb prostheses utilizing SACH feet can be readily lengthened by incorporating an extension block between the ankle and the foot. Additional length compensation can be planned by making the prosthesis long at the outset and compensating for this added length by the addition of a lift on the opposite shoe. Then as growth occurs, the removal of the lift equalizes the length resulting from growth and augments the life of the prosthesis.

Girth changes can be achieved by removing the soft insert in the below-knee prosthesis and in addition building up or relieving the socket wall to accommodate shrinkage or swelling. The adding or removal of the residual limb socks could also aid in the process of girth compensation.

In the case of the upper limb prosthesis, the problem has proven not to be as simple or as easily solved.

The modular prosthesis developed by the U.C.L.A. Child Amputee Prosthetic Project was the first upper limb prosthesis that attempted to attend to the problem of linear growth through the use of modular components[4]. It failed to explain how they would attend to the other problem secondary to growth, that of changes in girth. In response to this need, the Prosthetics Department of the Kessler Institute for Rehabilitation in West Orange, New Jersey has developed an adjustable upper limb prosthesis that will accommodate both the circumferential and longitudinal growth of the child's residual limb. This article presents the concept, design and fabrication steps of this unique approach.

CONCEPT

The rationale of this prosthesis is to accommodate length and girth changes due

to the normal development of a child, both on the amputated and non-amputated side, since both must be considered routinely by the physician and the prosthetist as the child matures. Changes in the girth and length of the residual limb requires prosthetic alterations during the life of the prosthesis. Longitudinal growth in the contralateral arm requires extension of the prosthetic arm.

The prosthesis presented in this article allows an overall increase in the forearm portion of two-and-one-half centimeter (2.5 cm.). This significant lengthening is achieved without the need of adding any external component to the wrist unit. It is a modular component design that is fabricated as part of the original prosthesis. It allows changes in length to be carried out simply without replacement of the endoskeletal components.

Conventional methods for extending the life of a child's prosthesis have been to either fit the patient initially with extra ply socks, to incorporate a socket within a socket (triple wall), or a combination of the two. As the child grows, the ply of socks can be reduced and eventually the inner socket can be removed. Continued growth of the residual limb cannot be accommodated by the prosthesis, so usually a new prosthesis is required.

There are compromises that have to be accepted with this approach. Extra ply socks in hot climates will not enhance wearing comfort. A triple wall socket will add additional weight to the prosthesis. Also, the outer socket is just a larger replica of the inner socket and may not exactly reflect actual growth changes in the residual limb. All standard approaches to accommodate residual limb growth can be incorporated into this modular prosthesis. However, one of the unique features about this prosthesis is that the original socket can easily be removed and a new custom molded socket can be attached to the existing superstructure. This can be done simply by heating the proximal portion of the forearm and allowing it to expand and shape itself over the new, enlarged socket.

The prosthesis can also function as a temporary prosthesis by using the endoskeletal components to connect the socket to the wrist unit[5]. The temporary prosthesis (Fig. 1) would provide the recent traumatic amputee with a functional prosthesis until volume changes in the residual limb have been stabilized.

In summary then, the concept is to allow the prosthetist to lengthen and increase girth of the prosthesis. It is possible using this design to augment the life expectancy of a pediatric prosthesis during the growth years.

DESIGN

This is a modular, polypropylene, below-elbow prosthesis with supracondylar suspension attached to a figure 9 harness. Polypropylene has been used in the past in a limited capacity for the fabrication of certain upper limb prostheses. We have selected polypropylene as the choice material in the fabrication of the below elbow prosthesis because it can be easily thermoformed over a plaster model and offers a hard, durable surface that is easy to clean. This material is more malleable and less brittle than polyester or acrylic resins and also provides a surface resistant to high impact surface fractures. Another advantage is that polypropylene is inert at ambient temperatures which would eliminate the risk of dermatological allergic reactions at the socket interface.

Since polypropylene can readily be thermoformed, socket relief adjustments can be performed simply by removing the socket and heating the particular area enough to form the proper relief.

The prosthesis is composed of the following modular components (Fig. 2):

- Polypropylene forearm
- Polypropylene socket
- DELRIN* socket receptacle with ½-20 internal threads
- A nylon ½-20 threaded rod
- A DELRIN* friction wrist unit
- A DELRIN* retainer for cable housing
- A DELRIN* and polyethylene cross bar assembly

The average weight of the prosthesis assembled, but without the terminal device,

*DELRIN—E.I. DuPont De Nemours trade name for polyacetal resin.

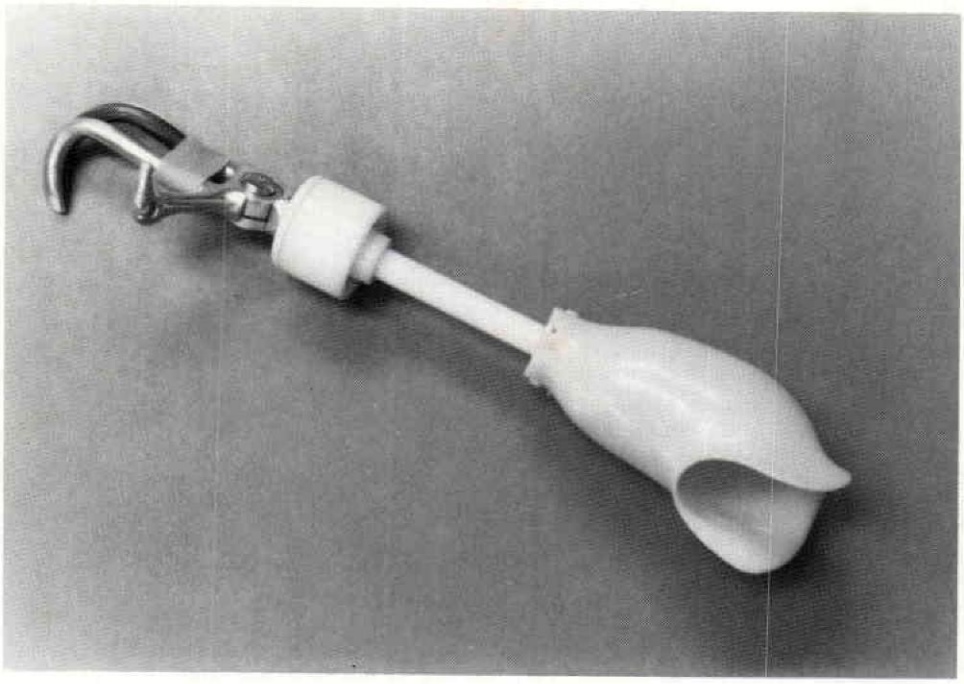

Fig. 1—Temporary below elbow prosthesis using nylon rod and Delrin wrist unit.

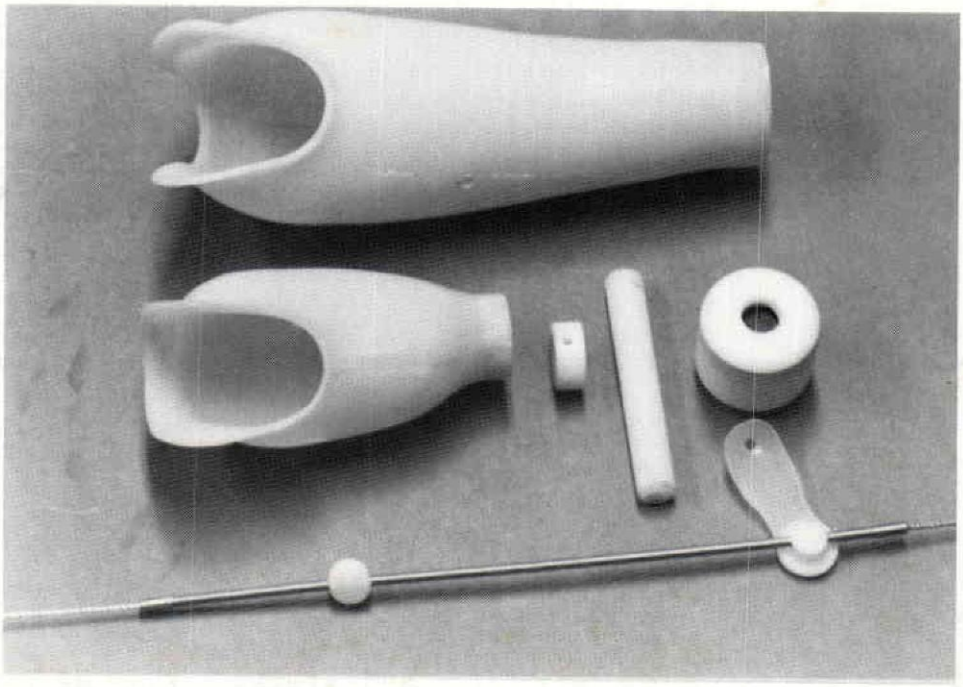

Fig. 2—Component parts for the Modular Below Elbow Prosthesis.

is between 140-165 grams. The forearm weight can be as light as 47 grams; the socket 36 grams; the wrist unit 32 grams.

There is an endoskeletal nylon ½-20 threaded rod which is connected to a friction wrist unit and a socket receptacle at the distal portion of the socket. These endoskeletal components provide the prosthesis with structural integrity. In addition, they allow the lengthening adjustments of up to 2.5 centimeters.

The polypropylene forearm provides a protective, rugged cover for the endoskeletal components since it is primarily fabricated for use by the pediatric age group. In addition, the cover provides attachment points for the control assembly components.

ALIGNMENT

The casting of the residual limb is carried out with the development of the positive mold in the usual manner. Anterior and lateral mid lines are drawn on the plaster model representing the long axes of the residual limb in both the frontal and sagittal planes (Fig. 3). A hole is made at the distal end of the mold where the two lines intersect and an alignment receptacle is inserted. A mock wrist unit attached to a temporary rod is then positioned in the receptacle (Fig. 4). The required length of the prosthesis is established by sliding the wrist unit to the proper position on the rod. The method of alignment is similar to the method used by Otto Bock Industries for aligning below-elbow myoelectric prostheses. Further alignment adjustments, if necessary, can be carried out by changing the angle of the receptacle.

FABRICATION

Once the alignment is achieved, the temporary rod and the mock wrist unit are removed. One-eighth inch thick polypropylene is draped and vacuum-formed over the receptacle and model. Care must be taken at this point to assure that the seam is on the anterior surface of the model. The ridge created on the socket during the vacuum forming process is sanded flush with the surface and is then burnished with a heat gun.

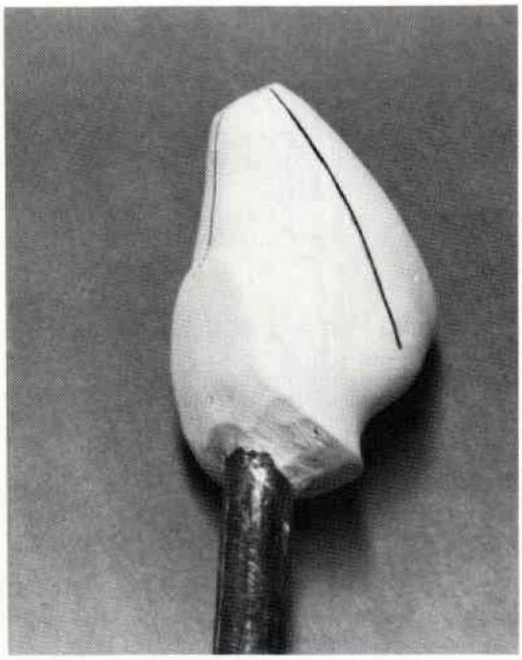

Fig. 3—Anterior and lateral lines, drawn on the positive impression model, are used to establish the alignment of the prosthesis.

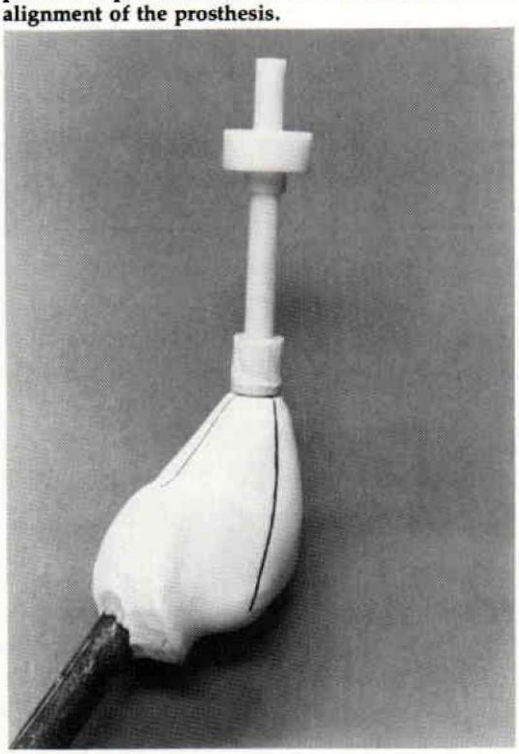

Fig. 4—Alignment receptacle positioner on the positive impression model, with the alignment rod and mock wrist positioned in place.

The mock wrist unit and rod are once again inserted into the receptacle and a forearm is shaped using clay. When the proper shape of the forearm is obtained, the rod is removed and a plaster model of the forearm is made using an alginate impression. Once dried, a one-eighth inch polypropylene sheet is draped and vacuum formed over the model making sure the seam this time is on the posterior surface. The seam is smoothed in the manner previously described. The forearm section and socket are removed from the plaster models.

ASSEMBLY

The alignment receptacle is then removed from the socket and replaced by the definitive socket receptacle. The receptacle has a ½-20 internal thread and is securely fastened to the distal portion of the socket. A nylon ½-20 threaded rod is screwed into the socket receptacle and is fastened so that no further rotation takes place.

The proximal posterior portion of the forearm is lowered three centimeters to allow enough expansion for it to fit easily over the socket.

After the forearm is in place, the friction wrist unit is inserted into the distal portion of the forearm and is threaded onto the rod until the desired length is obtained.

Assembly is then completed by fastening the forearm to the proximal medial lateral portion of the socket using two nylon screws for fixation. The proper terminal device is then attached (Fig. 5).

DISCUSSION

The prosthesis, once prepared, is used by the child after proper training.

As growth occurs in the contralateral limb, the prosthesis is returned to the library for lengthening which is achieved by adjusting the length of the threaded rod gradually up to its maximum (Fig. 6).

Continued residual limb growth that cannot be accomodated by minor socket adjustment can be achieved by replacing the inner socket without any alterations to the external appearance of the prosthesis (Fig. 7). The increased size is achieved following a second casting for the fabrication of the new inner socket (Fig. 8).

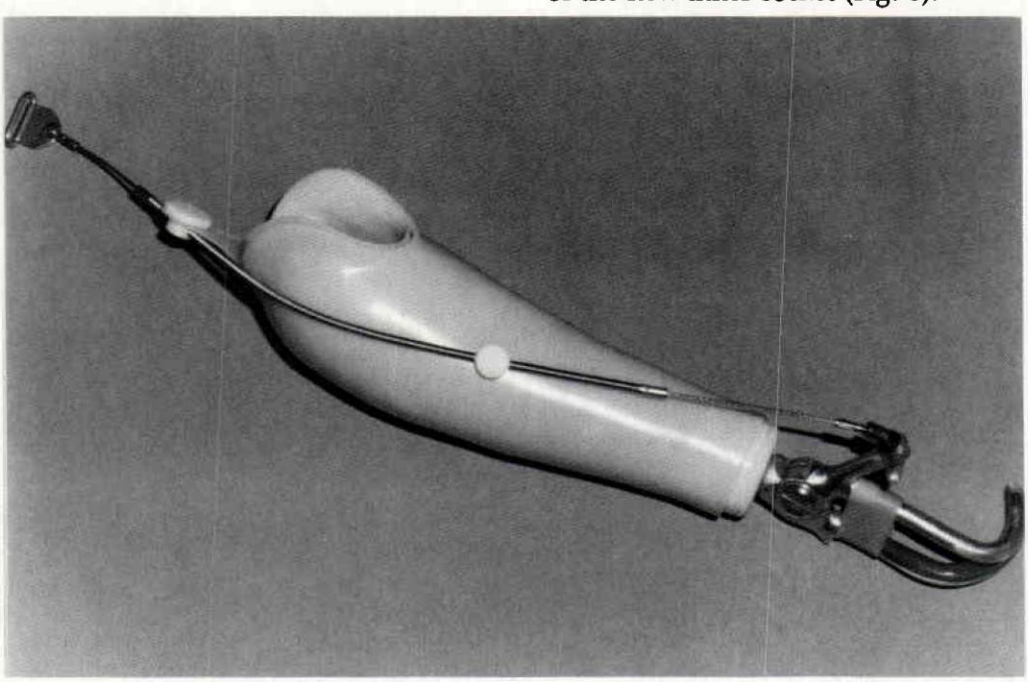

Fig. 5—Modular Below Elbow Prosthesis completely assembled.

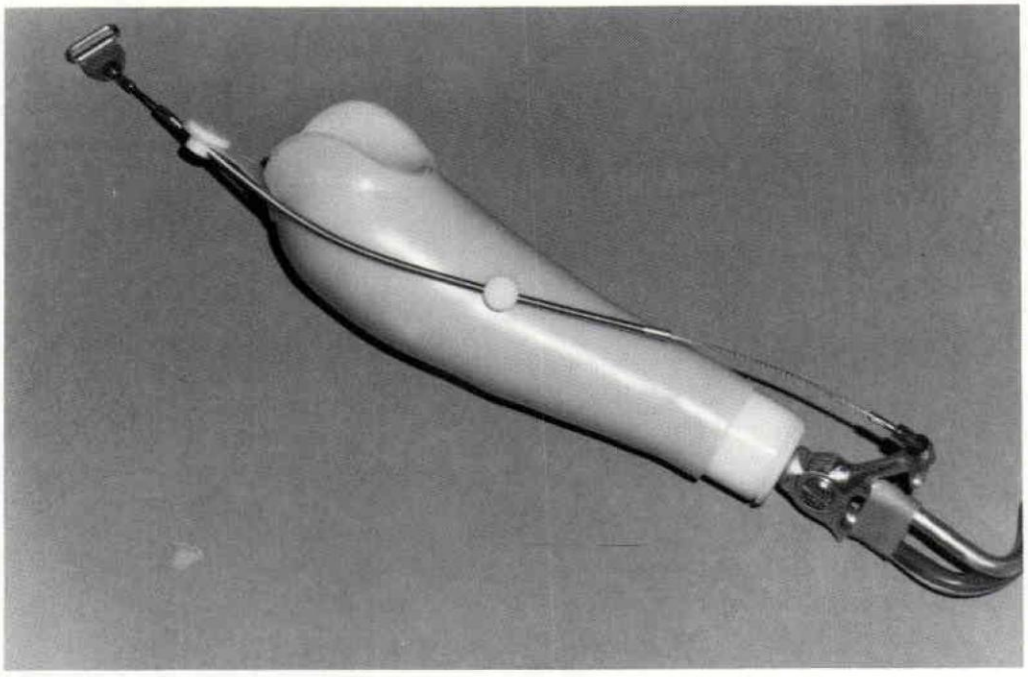

Fig. 6—Modular Below Elbow Prosthesis with the adjustable wrist unit in the extended position.

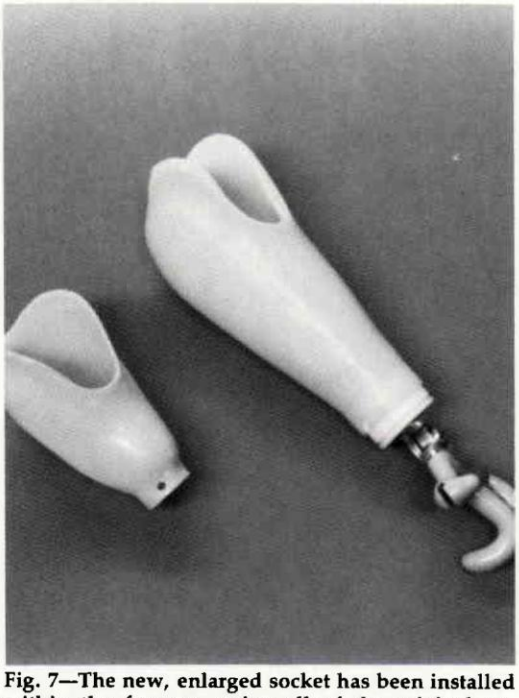

Fig. 7—The new, enlarged socket has been installed within the forearm, using all of the original endoskeletal components. The original socket is also shown.

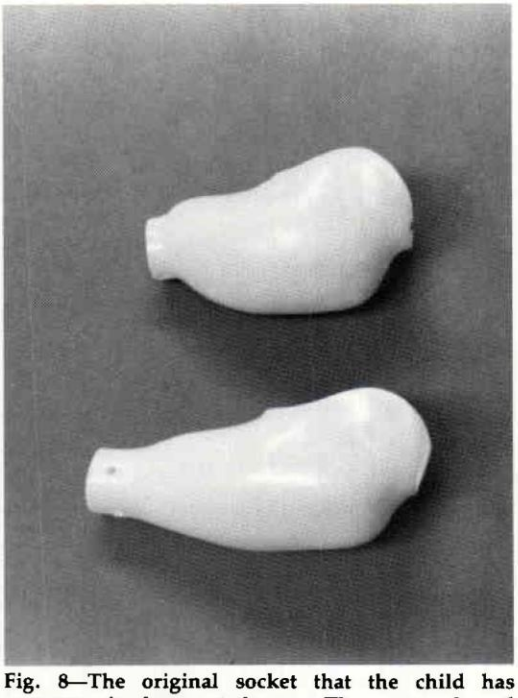

Fig. 8—The original socket that the child has outgrown is shown at the top. The new, enlarged socket that will be attached to the existing superstructure of the prosthesis is shown at the bottom.

All the above changes can, in our view, significantly extend the life of the average pediatric below-elbow prosthesis. The required changes can be carried out economically and with minimal alteration to the superstructure of the prosthesis.

SUMMARY

A new concept in below-elbow prosthetics is presented that is modular and allows increase in length and girth which significantly extends the life of the pediatric age prosthesis.

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Juvenile Amputees Classification and Revision Rates

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INTRODUCTION

The care of the juvenile amputee presents special problems for the physical therapist, prosthetist, and orthopedic surgeon. Optimum surgical treatment and prosthetic fitting demands careful planning and frequent clinical and radiographic followup, as growth alters the size and configuration of the residual limb. Growth of the residual limb and soft tissues may be disproportionate, resulting in bony overgrowth. This problem, as stated by Lambert[1], represents the most common unfavorable sequela of the surgical treatment of the juvenile amputee. Parents of these children and third party payers often request information from medical personnel concerning the anticipated nature and frequency of surgical and prosthetic modification due to bony overgrowth. Ideally, a classification of the juvenile amputee by type, age and bones involved, and an analysis of patients with such classifications, would help to answer these questions.

PURPOSE

The purpose of the study is to present a classification of juvenile amputees, and to describe the relationship of classification, treatment, bones involved, and patient age at the time of amputation to the necessity for surgery for bony overgrowth (Fig. 1).

REVIEW OF LITERATURE

Aitken[2], Lambert and Pellicore[3] describe the bony overgrowth in the juvenile amputee as appositional bone growth independent of ephiphyseal growth. Aitken[5] documented the order of overgrowth frequency as: humerus, fibula, tibia, femur, and tibia/fibula. Once the bone pierces the skin, treatment of choice is bony excision and soft tissue closure. Pellicore[3] demonstrated that, in addition to the specific bone involved, age at amputation and occurrence of transection through bone lead to recurrence of bony overgrowth. According to him the bones involved in order of frequency were: tibia, humerus, fibula, femur, and tibia/fibula. He and Aitken concluded that prevention of overgrowth can best be accomplished by joint disarticulation.

Many children with acquired amputations need revision, particularly of the humerus and fibula. True congenital am-

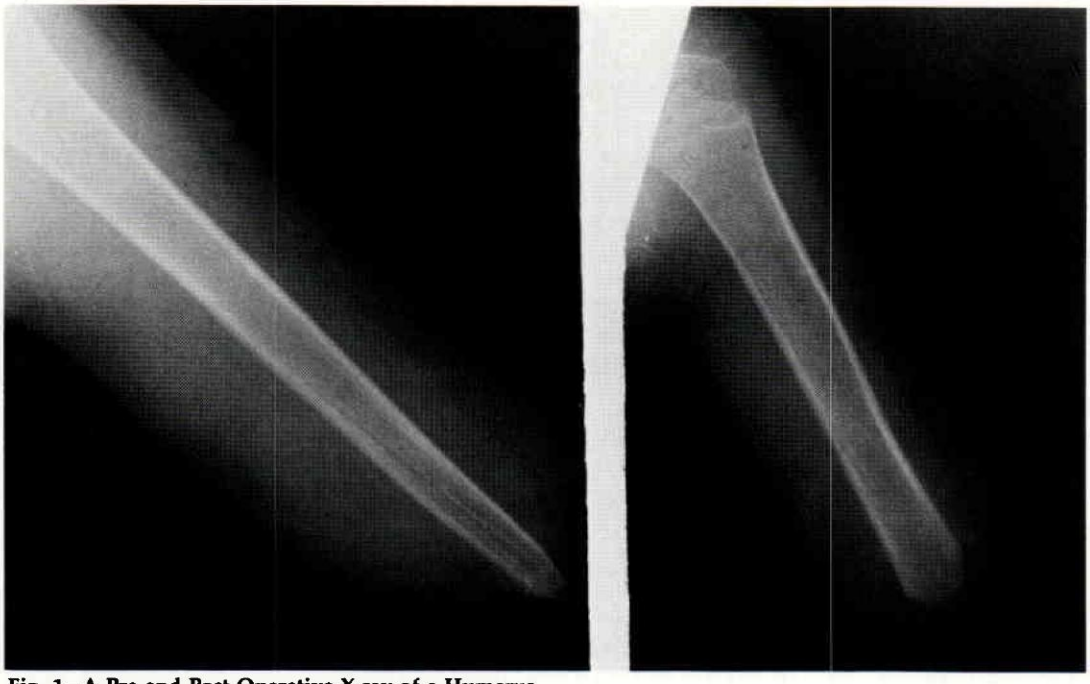

Fig. 1—A Pre and Post Operative X-ray of a Humerus.

putations which need revision, although rare, have been reported by Lambert and Aitken[7], although they did not describe these cases.

METHOD

We identified juvenile amputees in two ways. First, by computer search of "limb reduction deficiencies" in the Medical Record Department of the University of Iowa Hospitals and Clinics from 1965 through 1981. These patients had at least one hospital admission during that period, though not necessarily related to the amputation. Additionally, we identified congenital amputees by limb and level from the Congenital Hand Project, which extended from 1960 to 1980. The charts of all patients were reviewed. For inclusion in the analysis, the initial amputation or first revision had to be completed prior to skeletal maturity. For each patient, we documented the following: age at amputation, etiologic factors, bones involved, level, and the nature and number of operations on bone. The data were tabulated and graphed for analysis.

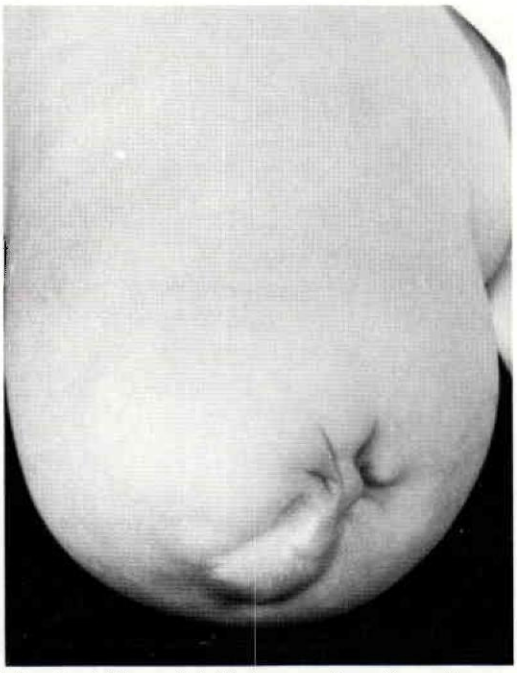

Fig. 2—A Type II BK Amputation, Secondary to Constriction Ring Syndrome.

RESULTS

One hundred twenty patients with major limb deficiencies or amputations prior to skeletal maturity were classified to the following five-category system:

Type	Description	Number of Cases
I	Acquired amputations (infection, trauma, etc.)	26
II	Congenital amputations through long bones.	8
III	Congenital deficiencies surgically converted by amputation through bone.	3
IV	Congenital deficiencies surgically converted by disarticulation.	4
V	Congenital deficiencies treated non-surgically with prostheses.	79
		120

The distinction between Type II and Type V amputations is important. Type II includes true congenital amputations through long bone and may result from such defects as constriction ring syndromes (Fig. 2). Type V includes terminal transverse deficiencie, in which vestigial appendages are usually present (Fig. 3).

Type I cases range at amputation from 3 months to 14 years. The humerus overgrows most frequently and requires as many as six revisions, demanding a total of 14 procedures on four patients. The fibula was revised in three cases, the tibia/fibula twice, and the tibia alone once. Cases acquiring amputation after age 12 needed no revision, regardless of the bone involved. Only seven cases are skeletally mature, and only one of these occurred at the above-elbow level. Of those skeletally mature, the only case requiring revision was for a tibia and fibula on separate occasions in a below-knee acquired at age 10. The revisions were done at ages 15 and 16 respectively.

Fig. 3—A Classic Congenital BE, with vestigial hand.

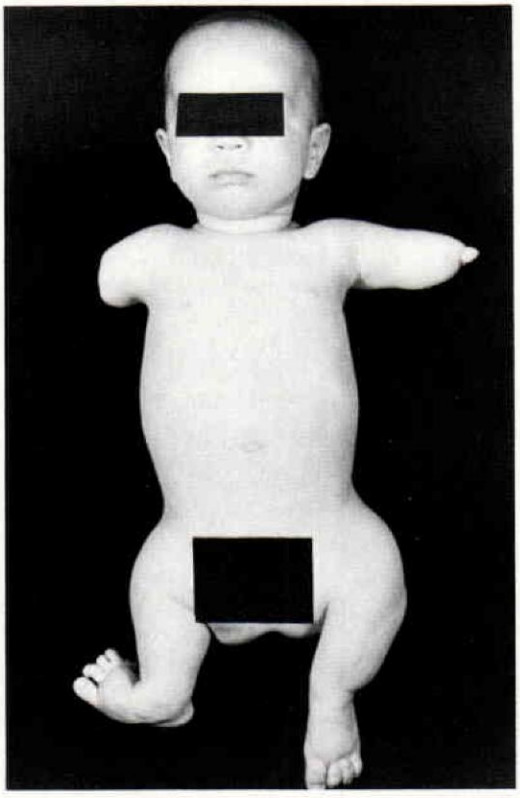

Fig. 4—A Type II Congenital Quadramembral, who had two revisions of the right humerus.

Type II cases, true congenital amputations, are defined as amputations through the long bones, present at birth. We excluded the classic terminal transverse below-elbow deficiency, with vestigal hand or nubbins, and all other congenital limb deficiencies not through the long bones. We included these (PFFD, phocomelias and amelias) in Type V. Of the eight Type II cases, seven underwent at least one revision. One fibula was revised five times, and two humeri were revised twice. Of interest was one patient (Fig. 4) who had bilateral PFFD and bilateral above-elbow amputations, one longer than the other. In 20 years, he underwent two revisions for the right and no revisions for the left.

Type II congenital, through bone, amputations do overgrow. The humerus, the most common bone involved, required six revisions in four patients. Two tibia/fibulae were revised a total of four times: three in one case and one in another (Table 5).

Since all patients in Type II had at least three limbs involved and had at least simple syndactylys, this group appears to respond differently than the classical limb deficiency or the patient with only one limb involved. From our data, patients with congenital amputations of the humerus and fibula react much like Type I or Type III, even though they appear to be true congenital amputations.

Type III cases are those in which surgical conversion of congenital limb deficiency involved a cut-through bone. All are at the below-knee level (Table 4). Two were congenital pseudarthroses of the tibia and fibula, and one was a congenital absence of the fibula. Like the type I amputations, Type III require bony revision. Of the below-knees, two needed revision of the tibia, one at age 12 and one at age 15, and one required revision of the fibula. All amputations were done prior to age 5 and, predictably, acted similar to Type I.

Type IV encompasses congenital deficiencies converted surgically by disarticulation. PFFD and single or double ray feet without a hindfoot are examples. Surgical conversion enhances prosthetic fitting without concommitant loss in function. No bony overgrowth occurred in this

Type I by Level and Cause

Level			Cause		
AE	=	7	Farm Equipment	=	11
BE	=	4	Lawn Mower	=	5
WD	=	2	Infection	=	6
BK	=	13	Burn	=	3
			Motor Vehicle	=	1
		26			26

Table 1—Type I by Level and Cause.

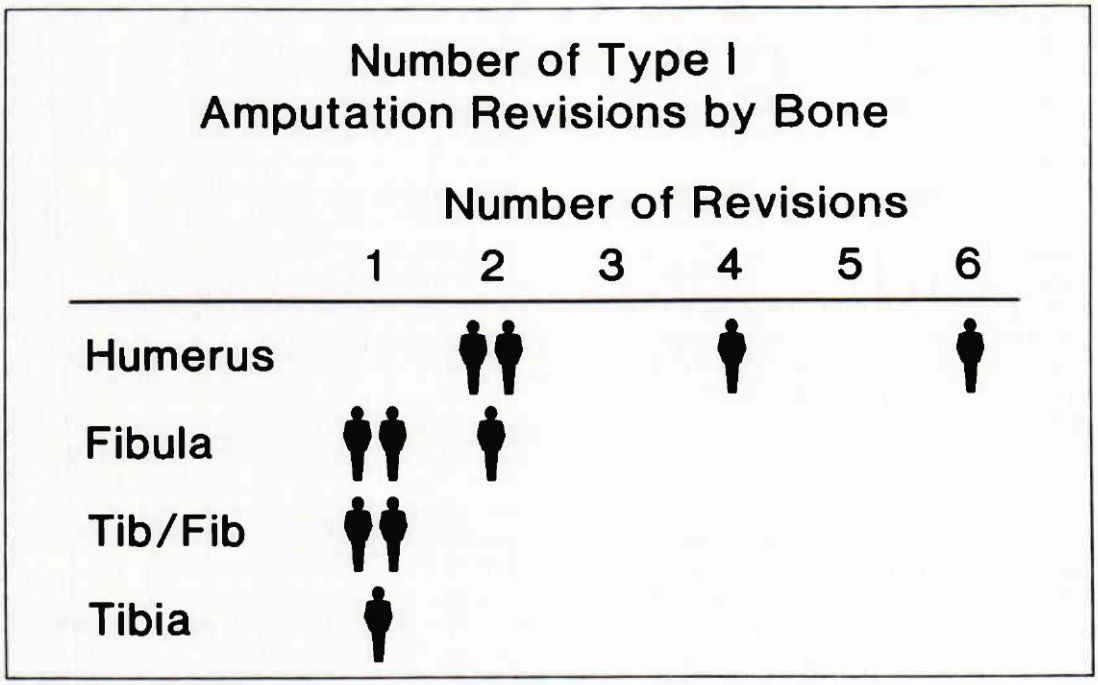

Table 2—Number of Type I Amputation Revisions by Bone.

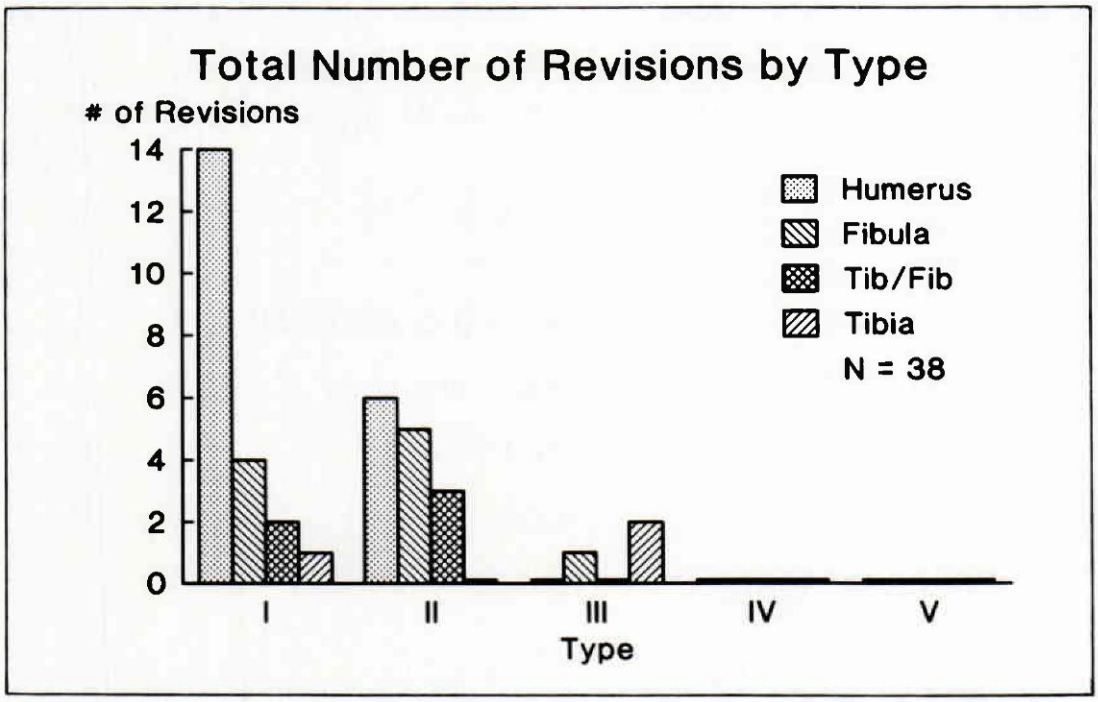

Table 3—Total Number of Revisions by Type.

Total Cases by Type and Level

Level	Type I	II	III	IV	V	Total
AE	7	5	0	0	0	12
BE	4	0	0	0	33	37
WD	2	0	0	0	0	2
AK	0	0	0	0	0	0
BK	13	2	4	0	0	19
AD	0	0	0	0	0	0
All Others	0	0	0	4	46	50
Total	26	7	4	4	79	120

Table 4—Total Cases by Type and Level.

Type II Congenital Amputations Through the Long Bones

Level	F/U Years	Number of Revisions	Age, in Years, at Revision Humerus	Fibula	Tib/Fib
AE	8	1	7		
AE	8	2	8, 10		
AE	3	1	2		
BK	14	3			2,4,14
(L)AE	20		3, 7		
(R)AE		2(R)			
BK	7	1			5
BK	9	5		1,3,6,7,9	

Table 5—Type II Congenital Amputations through the Long Bones.

group.

Type V comprises the largest sub-group of these congenital deficiencies, treated non-surgically by prostheses. Type V includes the classic terminal transverse below-elbow amputation which occurs more often in females, and on the left side[6]. A vestigial hand or nubbins are usually present. In review of 33 unilateral cases with no other abnormalities, we found no case of bony overgrowth. Since no bony overgrowth occurred, we believe that these are true limb deficiencies, and therefore belong in our Type V classification.

DISCUSSION

After analysis of this group of juvenile amputees, we concluded that for prognostic purposes the proposed classification system is valuable. The known tendency of acquired amputations (Type I) to require revision, sometimes multiple revisions, is confirmed. Twenty-six patients underwent 21 revisions. The analysis indicates that congenital transverse amputations (Type II) through long bones frequently require revision, occurring 14 times in seven patients. This confirms the opinions of Aitken and Lambert. For prognostic purposes, the transverse congenital amputation (Type II) should be considered an entity distinctly different than non-surgically treated congenital terminal transverse deficiencies (Type V), in which revisions were not required.

When congenital deficiencies were treated by amputation through long bones (Type III), they behaved, relative to surgical revision, as a congenital amputation through long bone. The congenital deficiency treated by amputation through long bone changed from an entity in which revision was unlikely to one in which revision was nearly predictable. If the congenital deficiency was surgically treated by disarticulation instead of amputation (Type IV), revision was not necessary, suggesting that, when possible, disarticulation is the preferred surgical procedure.

Our data cannot be used to accurately describe relative incidence for the five amputee types. The data and classification system does provide guidelines by which other investigators may further explore the question of surgical revision in juvenile amputees, especially in patients with congenital amputations of long bones.

CONCLUSIONS

Surgical revision of the juvenile amputee occurs in both congenital and acquired cases. The younger the patient at the time of amputation, the more likely the need for revision. In our series, the humerus was most often revised, followed by the fibula, tibia/fibula, and tibia. The data supports the frequent need for revision for bony overgrowth in true congenital amputations through the long bones, particularly through the humerus and fibula. It also reaffirms the contention that through-bone amputations should not be done if disarticulation is possible.

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A Review of the Failures In Use of the Below Elbow Myoelectric Prosthesis

Sondra Millstein
Hanna Heger
Gordon Hunter

INTRODUCTION

The myoelectric program at the Ontario Workmen's Compensation Board Hospital and Rehabilitation Centre started in the middle 1970's. The "Russian Arm" from the Rehabilitation Institute of Montreal (RIM) was used initially, and now the Otto Bock 6 volt system (117 patients) is used predominantly. The total number of below elbow myoelectric fittings as of September, 1981 was 128. Each patient is fitted with a cable-operated prosthesis with a hook and hand and continues to possess and use this prosthesis after being fitted with the myoelectric prosthesis.

Our early experience was previously reported in the *Journal of Bone and Joint Surgery*.[2] A number of patients have subsequently been found to either use the myoelectric prosthesis on rare occasions only or have rejected it. A review of this small group of patients was undertaken in order to determine why these patients did not use their prosthesis, and to hopefully provide us with information to predict unsuccessful users and, therefore, contraindications to fitting a below elbow amputee with an expensive myoelectric prosthesis.

MATERIAL AND METHODS

Twenty-five out of 128 patients who had been fitted with the below elbow myoelectric prosthesis were included in this review. Five of these patients had voluntarily returned their myoelectric prosthesis; 14 had indicated rejection or limited use of the prosthesis at follow-up Amputee Clinic Reviews and six had failed to return for review following supply of the prosthesis.

Twenty-two of 25 patients were reviewed with a mean follow-up time between myoelectric fitting and review of 33 months, (range of 2 to 57 months). The remaining three patients could not be contacted.

A standardized questionnaire was used in a personal or telephone interview with each patient, which evaluated the functional use of the myoelectric prosthesis in activities of daily living, vocational and avocational areas, and of the amputee's degree of acceptance in terms of function, cosmesis and comfort. The amount of time during which the prosthesis was actually

worn was asked, its use and reliability were assessed, and the problems the patient encountered with the prosthesis were investigated. As each patient possesses both the cable-operated and myoelectric prostheses, a comparison was made between the two. Interpretation of the results is necessarily guarded because of the small sample of 22 patients.

RESULTS

The mean age at the time of review was 41 years, the oldest patient was 67 and the youngest 29. Six patients were bilateral and 16 were unilateral below elbow amputees. The dominant side was injured in 63 per cent of the cases. Twenty-one patients were male and one was female.

The patients were divided into two groups; one group (ten patients) defined as limited users, wore their myoelectric prosthesis infrequently. The other group (12 patients) defined as non-users, were not wearing the myoelectric arm.

PROSTHESIS WEAR AND USE

Data was collected by asking each group of patients the number of hours they wore each prosthesis on an average day during the week and on the weekend (Tables I and II). Of the group of ten limited users, all wore the myoelectric prosthesis on the weekend; although the frequency and amount of time varied a great deal during the week, based to a large extent on their social activities.

The 12 non-users had completely rejected the myoelectric prosthesis, indicating a rejection rate of 9 per cent. This figure is compatible with the rejection rate of 8 per cent for the population of 59 Workmen's Compensation Board amputees previously studied by Northmore-Ball et al (1980). In comparison, the Swedish study of trained and untrained amputees reported by Herberts et al (1980) indicated a rejection rate of 44 per cent and 77 per cent respectively.[1]

The patients were used as their own controls and were asked to answer from memory the same questions for the time period before they were fitted with their myoelectric hand (Tables III and IV).

During the week, the time the hook was used remained relatively constant for both groups. On the weekend, while the hook use remained constant for the non-users of myoelectric prostheses, there was a decrease for the limited users, which indicated they were now using the myoelectric prosthesis for some of the time instead of the hook.

The mechanical hand appears to have been seldom used, except on social occasions, and with the advent of the electric hand, it appears to have become obsolete.

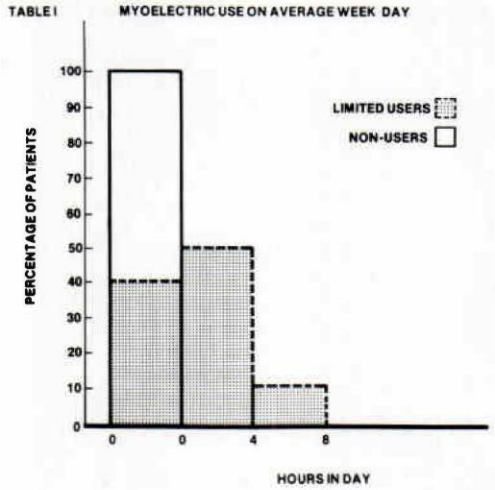
Several factors possibly affecting the patient's use of the electric hand were investigated, and the factors that were significant were type of job and leisure activities, the daily needs of bilateral amputees and the need for a cosmetically acceptable and comfortable prosthesis.

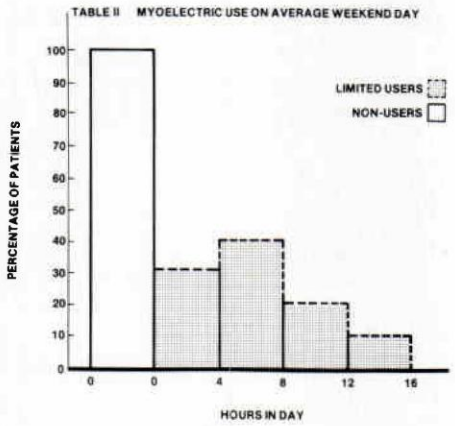
VOCATIONAL USE

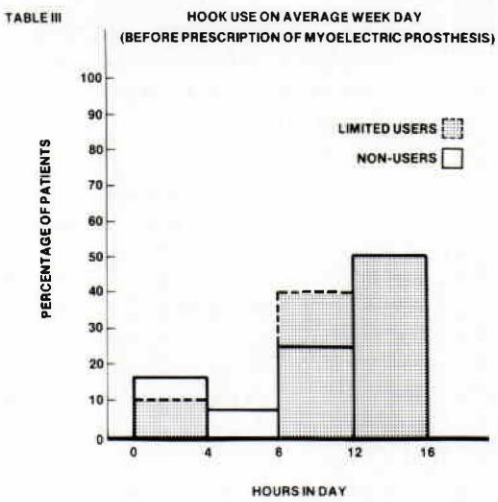
Although there was 100 per cent rejection of the myoelectric prosthesis at the work site by both groups, 84 per cent of the patients used their hook at work. The following factors determined the criteria for not using the myoelectric prosthesis at work: the cleanliness of the work environment, the manipulative skill required in handling tools, the force required to lift and carry heavy objects, and the degree of exposure to the public. The previous study by Northmore-Ball et al[2] confirmed that those patients who used the electric hand predominantly at work tended to have office type jobs. Most of the patients in both our groups were unable to use their prosthesis because of the physical demands of their jobs as mentioned.

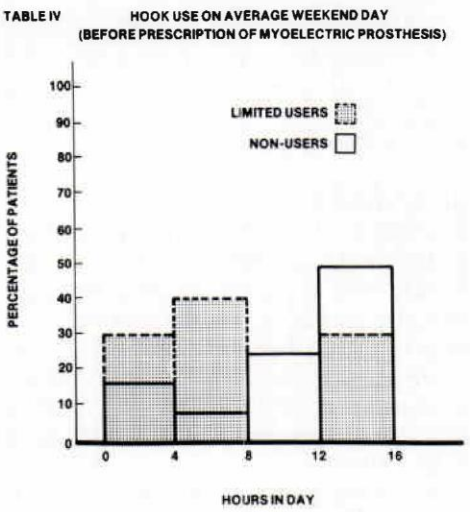
RECREATIONAL USE

All the limited users indicated that the prosthesis was useful at social events, i.e. eating, drinking and carrying light objects. This group places a high cosmetic value on the myoelectric prosthesis when in contact with the public. Although half the patients wore their myoelectric prosthesis passively, they commented that the possibility of using the prosthesis functionally was important to them in a social setting.









BILATERAL BELOW ELBOW AMPUTEES

All six of the bilateral amputees in the myoelectric program fell into the group under study; two were in the group of limited users and four in the non-users' group. Two were supplied with electric wrist rotators, one in each group. Bilateral amputees require a high level of function from their prostheses, and all overwhelmingly indicated that because the myoelectric prosthesis had very little functional value for them, they preferred to use their hooks. They complained about a lack of precision with two myoelectric hands which was possible with two hooks. To provide bilateral below elbow amputees with a more functionally valuable prosthesis requires further evaluation of wrist rotators and the myoelectric hook.

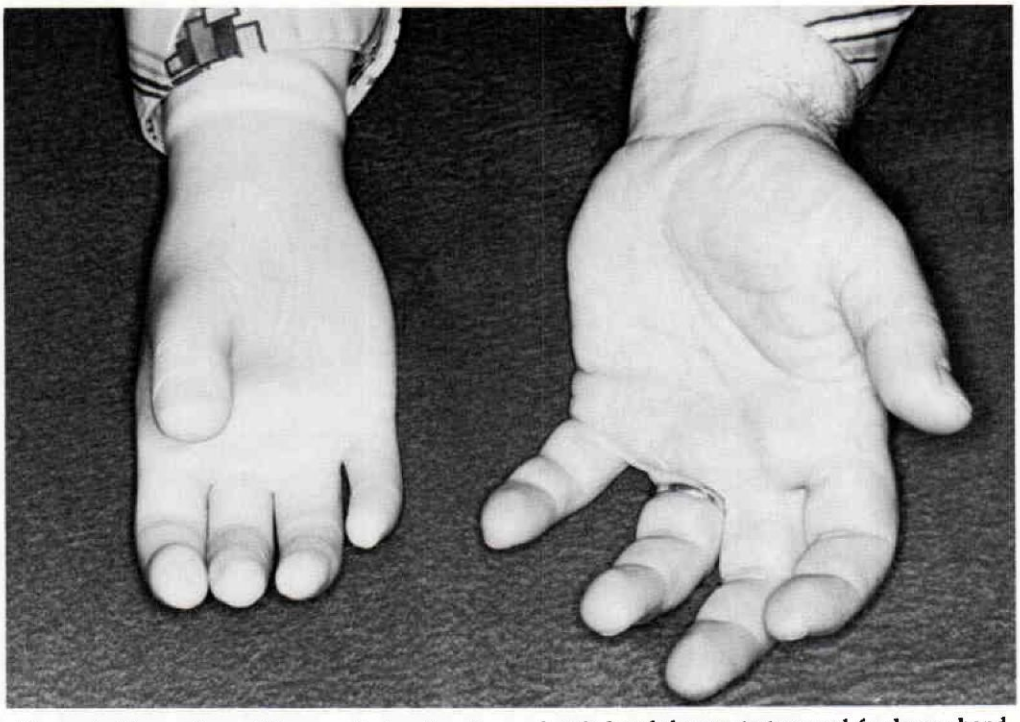


Figure 1—Comparison of large male hand and myoelectric hand demonstrates need for larger hand size.

COSMESIS

Eighty per cent of the limited users expressed a strong concern with cosmesis; whereas only about 33 per cent of the non-users expressed a similar concern. The group of non-users were comfortable with their hooks in public, and even felt that the hook was as good looking as the myoelectric hand. The majority of patients felt that the myoelectric prosthesis was acceptable to the public and to their families, but a few were dissatisfied with the color of the glove and some patients were dissatisfied with the small size of the hand. They felt that its smaller size in comparison with their own larger hand made it appear "feminine". At present, the standard sizes of the adult myoelectric hand are seven and one-quarter inches (18.4 cm.) and seven and three-quarter inches (19.7 cm.). There is an obvious need for a larger hand to better approximate the hand sizes in most men, which average eight to eight and a half inches (21.6 cm.) (Fig. 1).

Two patients with a transcarpal amputation requested a myoelectric prosthesis. In order to allow room to accommodate the prosthetic components, they had to be fitted with a prosthesis that resulted in a considerable length of discrepancy. It becomes evident that a suitable level of amputation is necessary to achieve good cosmesis (Fig. 2).

COMFORT

More than half of the patients of each group were very aware of the weight of the myoelectric prosthesis and its tiring effect. Although the actual difference in weight between the two types of prostheses is small (Table V), the weight is accentuated by the suspension used in the myoelectric prosthesis. To anticipate this problem of weight intolerance, a preliminary check with a weighted socket may be indicated for some patients.

Approximately three-quarters of the limited users and non-users complained of discomfort in the elbow area. Close ques-

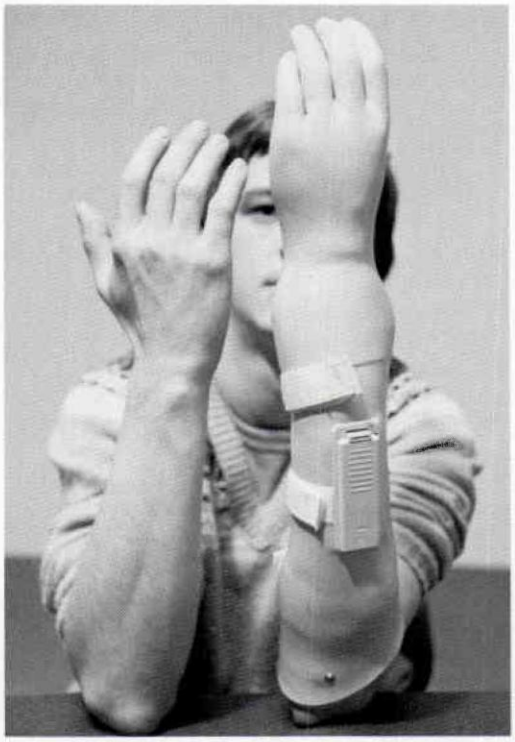


Figure 2—Unsatisfactory appearance of transmetacarpal amputation fitted with myoelectric prosthesis.

tioning of the patients, however, revealed that the discomfort in this specific area did not appear to be a key factor in their rejection of the myoelectric prosthesis.

Although three-quarters of the patients in both groups stated they did not mind the harness of their cable-operated prosthesis, most of the limited users preferred the comfort of the myoelectric prosthesis because of the absence of a harness.

REASONS FOR NON-USE AND REJECTION

For the limited users, the most common reason for not using the myoelectric hand at work, home or during leisure activities was fear of damaging either the prosthesis itself or its cosmetic glove. Approximately half the patients found that the glove was much too easily torn or cut, while almost all of them found that it could be easily ruined by grease, newsprint, ball point ink, carbon paper and wet blue jeans.

Another reason for non-use was instability of the socket for heavy work due primarily to the type of suspension used. The Muenster socket requires particularly skillful fitting and any fluctuations in

TABLE V

COMPARATIVE WEIGHTS OF UPPER LIMB PROSTHESES

CABLE OPERATED PROSTHESIS	
CABLE OPERATED (STANDARD) BELOW ELBOW PROSTHESIS WITH STAINLESS STEEL HOOK (DORRANCE NO. 5)	2 LB. (908 gm.)
CABLE OPERATED (STANDARD) BELOW ELBOW PROSTHESIS WITH MECHANICAL HAND	2 LB. 7 OZ. (1107 gm.)
MECHANICAL HOOK (SPLIT HOOK)	7 - 13 OZ. (198 - 368 gm.)
MECHANICAL HAND (DORRANCE NO. 4)	1 LB. 2.5 OZ. (525 gm.)
MYOELECTRIC PROSTHESIS	
COMPLETE BELOW ELBOW MYOELECTRIC PROSTHESIS WITH MYOELECTRIC HAND	2 LB. 6 OZ. (1078 gm.)
COMPLETE BELOW ELBOW MYOELECTRIC PROSTHESIS WITH MYOELECTRIC HOOK	2 LB. 5 OZ. (1050 gm.)
MYOELECTRIC HAND	1 LB. 3.5 OZ. (553 gm.)
MYOELECTRIC HOOK	1 LB. 2.5 OZ. (525 gm.)
WITH ELECTRIC WRIST ROTATION UNIT - ADD	5.1 OZ. (145 gm.)

residual limb size will affect the socket fit and may cause instability.

The reason given most frequently for rejection of the myoelectric prosthesis by the group of 12 non-users was its poor functional value. These patients were especially happy with the precision of the hook and the strength of the cable-operated prosthesis for heavy work. They found the prosthesis comfortable, did not mind the harness and did not have a strong cosmetic need. This may, in part, be due to the fact that one-third of the non-users were bilateral amputees.

CONCLUSIONS

It is our policy to give below elbow amputees the opportunity to be fitted with a myoelectric prosthesis six to twelve months after the injury, and we have come to realize that amputees will use both the hook and the myoelectric prosthesis for various purposes, rather than one prosthesis exclusively.

As a result of this review, the following are now used as relative contraindications to fitting a below elbow amputee with a myoelectric prosthesis:

- Bilateral below elbow amputation,
- Lack of concern with the cosmetic appearance of a hook or the discomfort of a harness,
- Requirement of a prosthesis with strength, precision and durability at work or leisure,
- Too long a residual limb (transcarpal/metacarpal amputation) unless the amputee is prepared to undergo revision surgery so that a prosthesis of satisfactory length can be fitted.

Acceptance of aids by the handicapped is a very complex process in which individual psychological, technical and socio-economic factors interact. Almost all of our reviewed patients had accepted the standard prosthesis and used it at work, home and in leisure activities. The fact that all these myoelectric non-users still use a hook would suggest that their rejection of the myoelectric prosthesis is based on limitations of the prosthesis, rather than the psychological influences of a device for the handicapped.

The myoelectric hook may prove to be the solution for those patients who find that the myoelectric hand does not provide them with enough function to be of value.

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Use of Electrical Stimulation in Prosthetics for the Control of Pain

Drew A. Hittenberger, C.P.

INTRODUCTION

Electrical stimulation as a method of pain control has become increasingly popular because of its rate of success and non-invasive technique. Its use is well accepted in the medical profession, but is limited in the field of prosthetics due primarily to the lack of familiarity and knowledge of the system.

The purpose of this article is to explain the uses and possible applications of electrical stimulation in prosthetics to reduce pain. In particular, the uses of electrical stimulation and its incorporation into definitive prostheses will be discussed. The treatment of one patient will also be reviewed to illustrate some of the results obtained.

PAIN CONTROL

Pain is defined as "a more or less localized sensation of discomfort, distress, or agony resulting from the stimulation of specialized nerve endings." While one is able to define pain in terms of general sensation, knowledge of how it works is still not completely understood. Currently there are three theories regarding the explanation of pain.

The "specificity theory" holds that pain is the result of stimulation of specific peripheral nerves and once stimulated, transfers itself in much the same way as hot and cold impulses. The impulses are either on or off with no degree of variation. The nerve ends are considered individual "end organs" from which specific impulses are relayed directly to the brain. While this theory does explain the transmission of impulses, it fails to explain the interpretation of pain by the brain in terms of intensity and the existence of pain in unidentifiable locations.

The "pattern theory" on the other hand suggests that pain exists as a stimulus to be interpreted by the brain. This could explain the various degrees of pain. It suggests that pain is not a specific stimulation in a specific area but rather the result of intensity and frequency of non-specific nerve endings.

The "gate control theory" (Melzack & Wall, 1965) suggests that pain is based on the function of the substantia gelatinosa

"T" cells which operate as transmission cells to enable the transfer of pain impulses to the dorsal horn. It is believed that the afferent impulses go through either large or small fibers to be controlled by the substantia gelatinosa as the "T" cell and it is this "T" cell that excites or inhibits pain based on ascending or descending impulses.

It is theorized that electrical stimulation creates an electrical impulse within the nerve fibers to block the pain impulses in the "T" cells. While there are several theories on how this is accomplished, there is no question that electrical stimulation can reduce pain. Since the gate control theory numerous studies have been conducted (Shealy 1968, Hymes 1973, Indeck and Printy 1975) and each resulted in the same conclusion: electrical stimulation reduced or eliminated pain in 60-80% of those patients tested.

TRANSCUTANEOUS ELECTRICAL NERVE STIMULATION

Transcutaneous electrical nerve stimulation (TENS) is a simple form of electrical stimulation. To briefly discuss its use, each unit consists of a small stimulator where wire leads are connected to a pair of flexible carbon electrodes. Electrodes are placed on the patient's skin with gel and an electrical impulse is transmitted through the electrodes by means of the stimulator. The electrical impulse can be controlled by the stimulator unit and the location may be changed by moving the position of the electrodes.

The success of the unit depends on its ability to stimulate large nerve fibers with low strength current. To do this, rate, amplitude and pulse width of the impulse can be adjusted. The most effective impulse pattern is referred to as "square pulse" where the impulse actually pulses by going on and off. Other patterns can be modified to peak in amplitude but while this accomplished the same effect it may result in an unpleasant sensation on the skin.

In addition to impulse control, the system's success depends primarily on the placement of electrodes. The electrodes are to be positioned primarily over the painful area and if no relief is achieved, should be placed over the trigger point or as close to the peripheral nerve as possible. Electrode gel should be used between the skin and electrodes to ensure good conductivity. Electrodes should not touch one another as this eliminates the electrical flow through the body. They should be held in place with tape or self-adhesive backing.

Once the electrodes are placed on the skin, the unit should be turned on slowly until the patient feels a tingling sensation. Caution should be used because of the possibility of discomfort. The unit should be turned up to increase the tingling sensation until it envelopes the area in question to eliminate pain. Again the success or failure of the system usually depends on electrode placement and several attempts may have to be made to find their optimum location. The period of treatment can range from 30 minutes every other day to 24 hours a day depending on the type and location of pain.

PATIENT DATA

The patient, a 41 year old male, sustained a crush injury to the right fifth finger resulting in a rupture of the profundus tendon. The patient was taken to the operating room for a tendon graft. During the procedure, the patient's ulnar neurovascular bundle was lacerated. The patient went on to develop Sudeck's atrophy (neurovascular dystrophy) of the right hand and considerable stiffness secondary to prolonged immobilization. The patient never had neurologic return of the ulnar nerve and developed a chronic pain syndrome of the hand later diagnosed as causalgia (defined as "burning pain often accompanied by trophic skin changes, due primarily to injury of the peripheral nerve"). After a total of five operations, each resulting in a worsening effect, the pain increased with severe burning and cold sensations. Finally, with

total loss of function about the hand and wrist, in combination with causalgic pain, the hand was amputated at the below elbow level to relieve the pain and improve function.

Post operatively, as the wound healed, the patient continued to have a large amount of pain that over subsequent months progressed proximally up the arm into the shoulder and upper neck region. At this point causalgia was diagnosed due to the type of pain and its persistence even after amputation. The patient described the pain as burning, starting in the residual limb and radiating proximally, often accompanied by profuse sweating. It increased with changes in temperature and altitude, and yet remained unpredictable in duration and intensity.

The patient was given multiple medications including narcotics and tranquilizers that only resulted in minimal relief. He underwent a total of eight stellate ganglion blocks and attended pain clinics each with no success. To complicate the situation further, the patient sustained a broken humerus which severely limited his shoulder motion on the prosthetic side. Prosthetic fit was successful because his residual limb was normal and well developed. Due to the increased radiating pain and limited range of motion, the patient's physical and emotional condition declined. Every known form of pain control had been tried with little or no success.

EVALUATION

TENS has been tried before around the shoulder and neck region with little success so it was decided to reexamine the patient concentrating our efforts around the medial distal end of his residual limb from which the pain radiated.

The same procedures were followed on this patient as with the TENS patients. A Medtronic unit and four electrodes were used. The reasoning for using the four electrodes was to surround the residual limb with as much current as possible. Since the patient's pain originated on the medial aspect of his residual limb, 2 large pads (2x3 inches) were placed on the medial and lateral sides and 2 smaller pads (2x2 inches) on the anterior and posterior sides (Fig. 1). Once the gel and electrodes were in place, the device was turned on and the patient allowed to adjust the intensity of stimulation to a comfortable level. With considerable repositioning of the electrodes, to maximize comfort, the patient soon noticed a marked decrease in the amount of radiating pain. He also noticed when he turned the unit off that within a two to four minute period the pain progressively returned to the level before treatment. During the treatment the only pain remaining was the dull pain about his shoulder, a pain which he said he could easily live with. He stated that this was "the first time I have ever been without pain."

Encouraged by the initial success of the system it was decided to incorporate the electrodes into a prosthesis so that he could wear his prosthesis and still experience the benefits of the TENS unit. To do this, a check socket was fabricated. The electrodes were then incorporated into the socket to check the system to see if it worked (Fig. 2). It was discovered that this

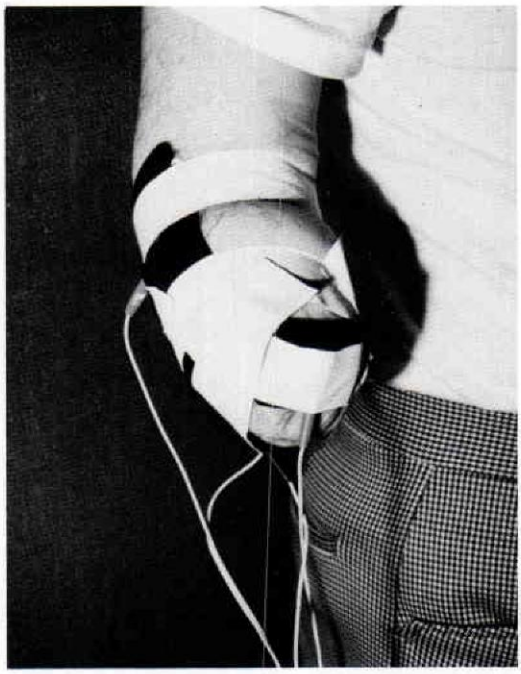

Fig. 1—Initial tests were conducted by temporarily securing the electrodes to the patient's skin.

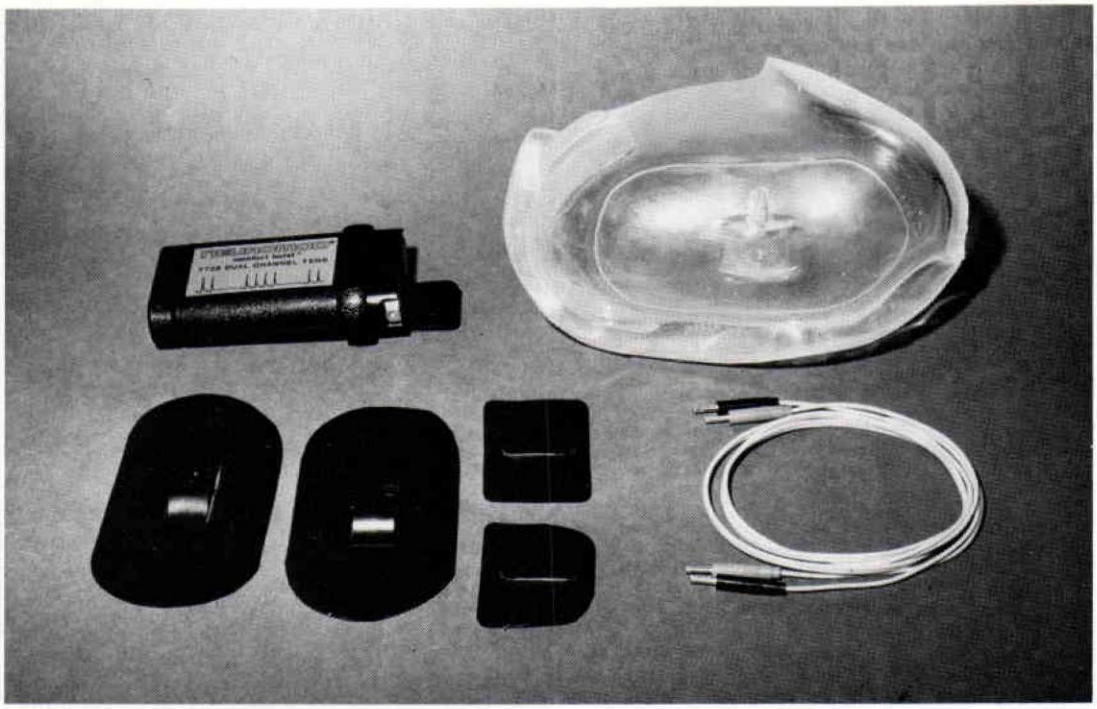

Fig. 2—The system included a medtronic TENS unit, four carbon electrodes, four electrode wires and a clear Surlyn socket.

system was not only possible, but also very successful in reducing the patient's pain; so successful that the electrodes were incorporated into the definitive prosthesis.

FABRICATION

Initially a negative impression was taken for a Muenster type socket. The positive model was modified and a Surlyn check socket fabricated to evaluate inital fit. Once a good fit was established, the existing socket was filled with plaster which was later removed to give a positive model (Fig. 3). The carbon electrodes were trimmed as desired (Figure 4) and glued to the positive model using rubber cement. The model was powdered and Surlyn pulled over the model and electrodes to form a socket (Fig. 5). This clear socket contained built-in recesses for electrode placement. To remove the socket, the plaster was broken out with careful attention not to damage the electrodes. Once removed, the edges of the socket were smoothed and holes drilled so that lead wires could attach to the electrodes on the inside of the socket (Fig. 6).

With the socket and electrode placement complete, electrode gel was placed on the inside of the electrodes to help conduct the electrical impulse and assist the patient in its application by lubricating the skin (Fig. 7 and 8). The electrodes were held against the skin by the socket. Because their location on the residual limb was the same, the unit produced consistent results.

The socket was tested for a period of one week before a final definitive prosthesis was fabricated. During the trial period the patient needed to have the unit on for a 24 hour period because of his pain. Due to this dependency, it was elected to have the patient wear just the socket with embedded electrodes at night, and a definitive prosthesis with electrodes would be made for day use.

The definitive prosthesis was made of acrylic resin to eliminate any allergic reaction. During fabrication, the electrodes were again cemented to the model and a PVA bag was pulled over it. Conventional

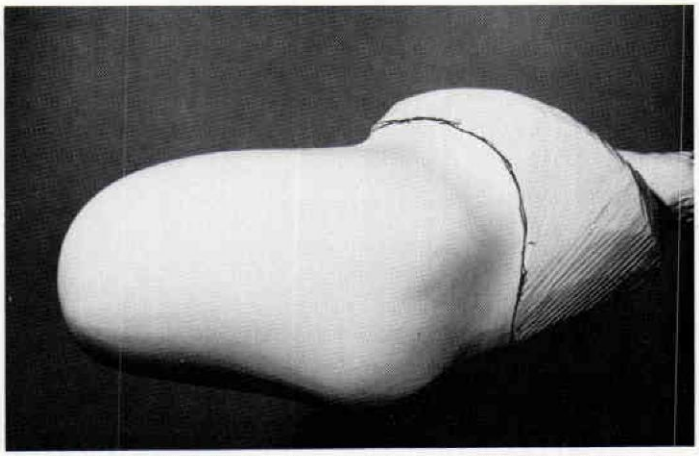

Fig. 3—The positive impression model was modified to obtain a total contact socket.

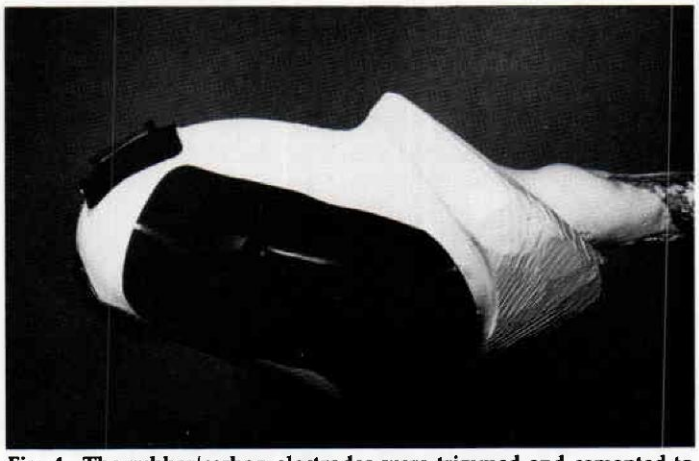

Fig. 4—The rubber/carbon electrodes were trimmed and cemented to the socket as desired.

Fig. 5—Surlyn/Thermovac plastic is vacuuformed over the positive impression model and electrodes.

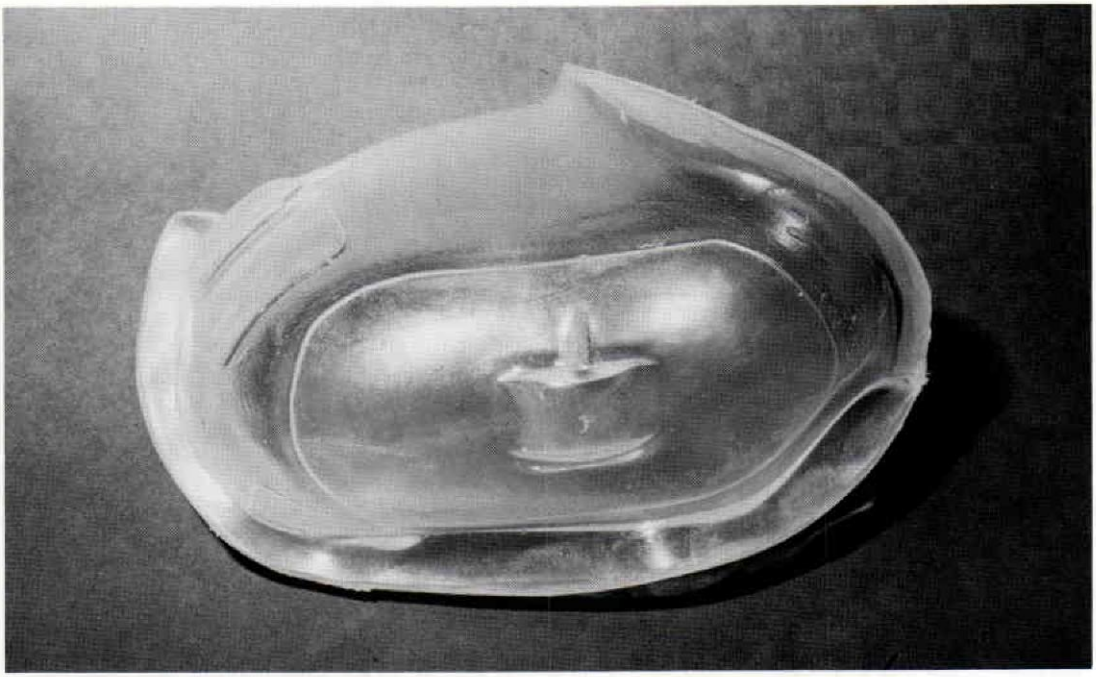

Fig. 6—The positive impression model and electrodes are removed to facilitate the finishing of the socket, and the drilling of electrode wire holes (¼").

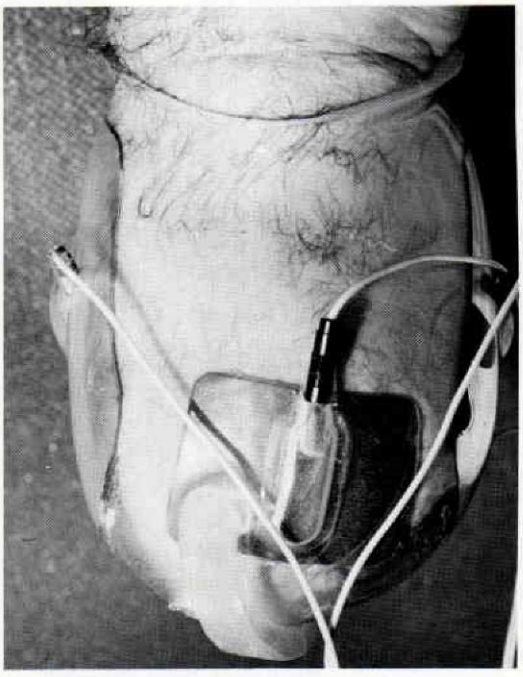

Fig. 7—The carbon electrodes were inserted inside the socket and electrode wires attached.

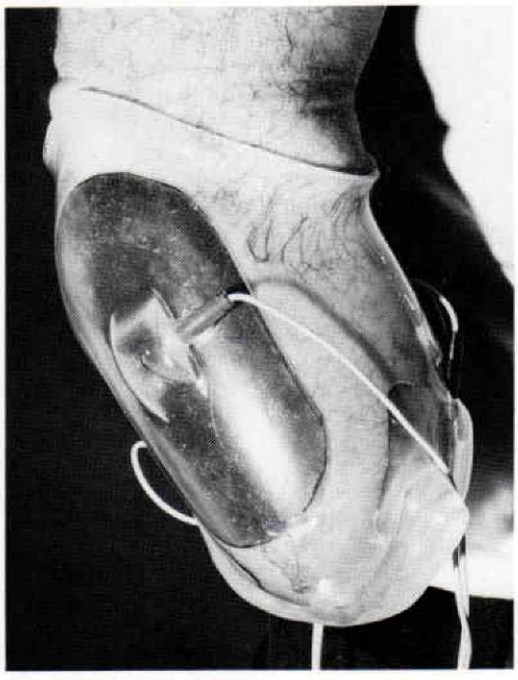

Fig. 8—Electrode gel was placed on the inside of the electrodes to ensure good conductivity.

laminating and alignment procedures were followed. The cast was broken out, electrodes removed from the socket, and lead wire holes drilled through the socket. The electrodes were reinserted and attached to the wire leads. This system worked as well as before and in combination with the socket at night proved to be an exceptional system for the relief of pain.

CONCLUSION

Incorporating a TENS unit directly into the socket of a prosthesis is a useful tool for decreasing pain. It is a system that can be tested for a predetermined amount of time and, when successful, be incorporated into a definitive prosthesis to produce the same results.

So far it has been used on upper extremity Muenster and above knee suction sockets with a large degree of success, and it is hoped that through continued development could be used on other types of sockets. With its initial success it is felt the system will find continued applications in prosthetics and give the patient an alternative means of reducing pain.

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ACKNOWLEDGEMENTS

Dr. Ernest M. Burgess,[1] Morris Dodge,[2] and the staff at the Prosthetic Research Study Center are thanked for their participation in the development and preparation of this report.

Drew Hettenberger, CP is the principal investigator for the Prosthetic Research Study Center and Dodge and Lundquist are located in Seattle, Washington.

Survey: Current Orthotic Practices In Occupational Therapy

Mike Lohman, O.T.R., C.O.

INTRODUCTION

Occupational Therapists concern themselves with normal daily life tasks. When neurological orthopedic or developmental dysfunction impedes purposeful activity, occupational therapy attempts to eliminate the barrier and enhance performance. This process might involve construction of corrective or facilitative orthoses. But construction of orthoses constitute only one facet of the multiple roles an occupational therapist must fulfill. Orthotists are experts in the construction of all types of orthoses, as their sole role. In some instances, both professions are involved in providing similar services. The extent to which this is the case was one component of a survey to evaluate prevalent clinical practices in orthotics. Although the survey and article refers to orthotics in general, the major area of orthotics effected concerns the upper limb.

METHODOLOGY

A survey questionnaire designed to evaluate prevalent clinical practices in orthotics was mailed to centers offering affiliations in physical dysfunction to University fo Kansas seniors in occupational therapy. One-hundred and one centers in 25 states were sent surveys; 50 centers in 22 states returned completed surveys.[3] While not geographically representative, the respondents were widely distributed.

This article will examine data collected related to; (1) use of materials, (2) frequency that various health professionals initially suggest, fabricate/fit, and collaborate on orthoses; (3) frequency that upper, lower and spinal orthoses are regularly constructed; (4) percentage of pre-fabricated orthoses utilized; (5) percentage of staff time devoted to orthotics. Additional information compiled by this survey, but not significant to this article include; types of affiliation offered; incidence of diagnostic treatment; and frequencies of specific anti-spasticity, contracture-reduction and fracture-bracing systems.

Frequencies and cross-tabulations were analyzed with the aid of Statistical Program for the Social Sciences (SPSS) batch system on Honeywell conversion. The term frequencies implies that similar types

of descriptive statistics are calculated. The term cross-tabulations relays a comparison of joint frequency distribution of two variables.[4]

RESULTS

The regularity with which different types of materials are used by occupational therapists is illustrated in Figure 1. Low temperature thermoplastics including Orthoplast/Polyflex and Polyform/Kaysplint are utilized most frequently in the construction of orthoses. Aquaplast, another low temperature thermoplastic, is utilized significantly less frequently. This may be due to its perceived short shelf life. High and medium temperature thermoplastics, as well as aluminum, spring steel, plaster and Plastazote are employed less often. However, it is important to note that roughly 80 percent of the responding therapists do utilize these advanced materials occasionally in orthotic construction.

The process of referral, evaluation, implementation and follow-up are central to the rehabilitation team model. Figures 2-4 denote the frequency with which responding therapists perceive themselves and other health professionals in the delivery of orthotic services. Figure 2 represents the frequency with which various professionals initially suggest orthotic treatment. Data gathered appears to reflect that occupational therapists and physicians are primary instigators of orthotic management for the upper limbs. Figure 3 indicates that occupational therapists identify themselves and orthotists as primarily responsible for the fabrication/fit of orthoses. This graph illustrates a sharp division among occupational therapists. Roughly half of those responding fabricate/fit 75% of the orthoses required by their institution; whereas, over one-third fabricate/fit less than 5%. Figure 4 represents how often professionals col-

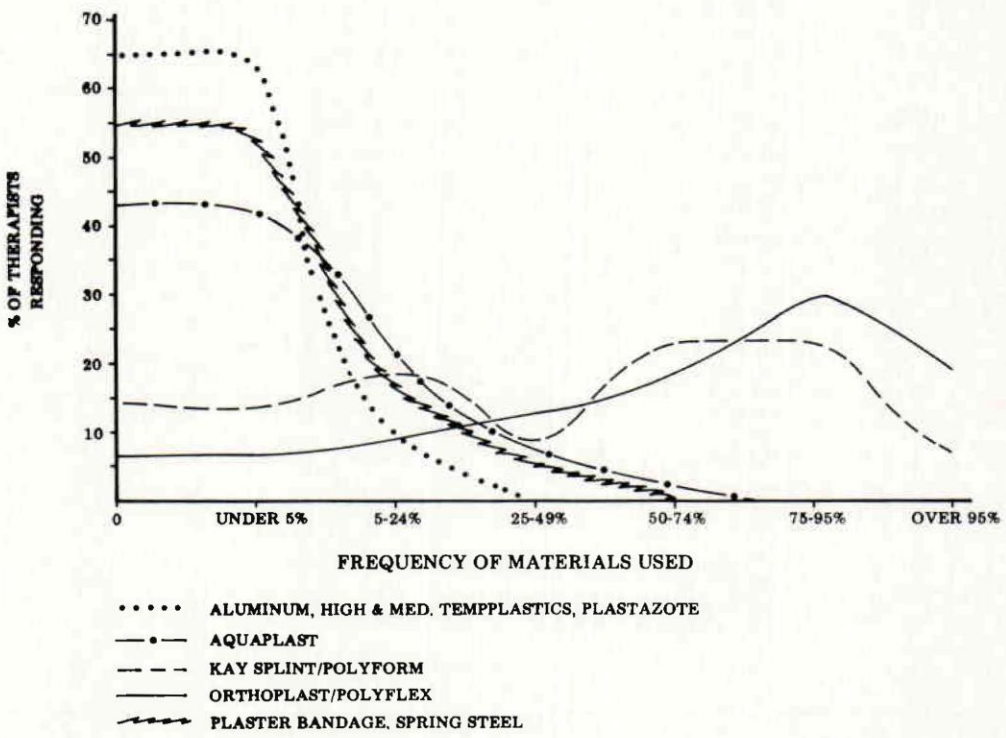
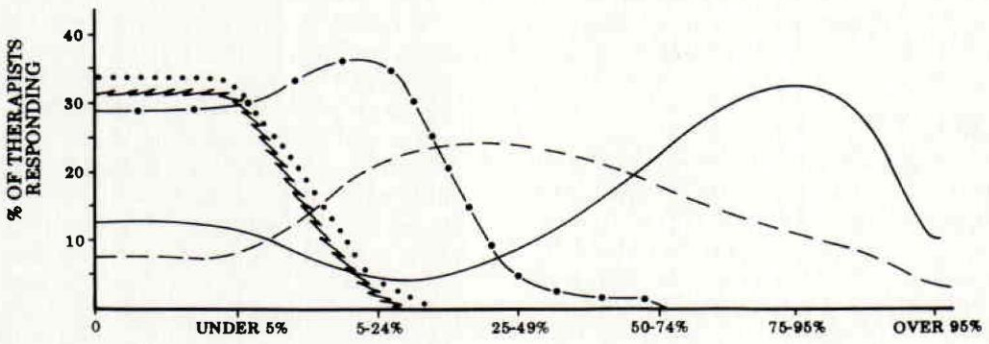


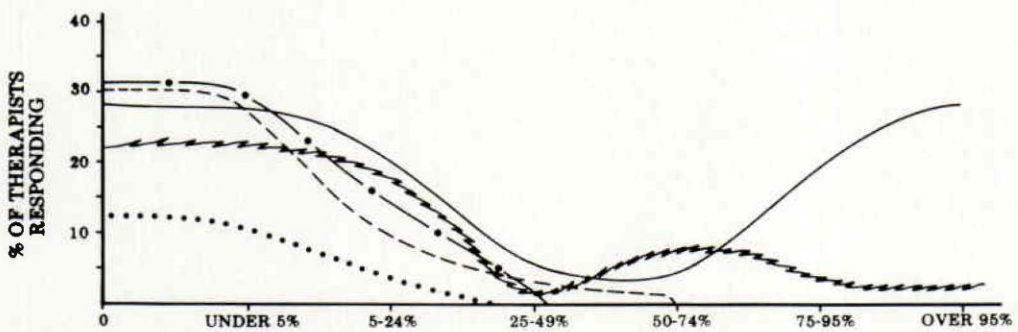
FIGURE I



FREQUENCY OF PROFESSIONALS INITALLY SUGGESTING ORTHOTIC MANAGEMENT

• • • • NURSES
—— OCCUPATIONAL THERAPISTS
ORTHOTISTS
—•— PHYSICAL THERAPISTS
– – – PHYSICIANS

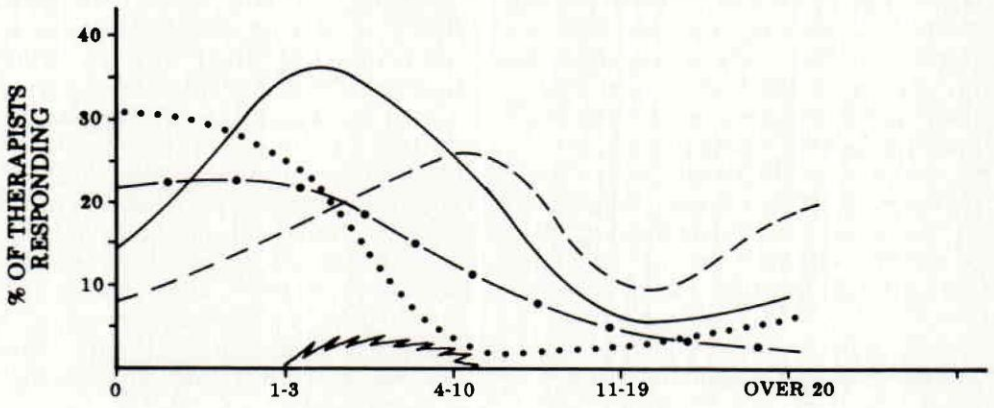
FIGURE II



FREQUENCY OF PROFESSIONALS FABRICATING ORTHOSES

• • • • NURSES
—— OCCUPATIONAL THERAPISTS
ORTHOTISTS
—•— PHYSICAL THERAPISTS
– – – PHYSICIANS

FIGURE III



FREQUENCY THAT RESPONDING THERAPISTS COLLABORATE WITH OTHER PROFESSIONALS ON FABRICATION/FIT OF AN ORTHOSIS IN A TYPICAL MONTH

• • • • NURSES
—— OCCUPATIONAL THERAPISTS
~~~ ORTHOTISTS
—•— PHYSICAL THERAPISTS
– – – PHYSICIANS

FIGURE IV

	NUMBER OF ORTHOSES*	0	1–3	4–10	11–19	OVER 20
UPPER EXTREMITY	RESTING HAND ORTHOSES	0%	38%	34%	14%	6%
	COCK-UP HAND ORTHOSES	44%	28%	6%	4%	0%
	DYNAMIC HAND ORTHOSES	8%	4%	20%	46%	12%
	TENODESIS HAND ORTHOSES	42%	34%	6%	4%	0%
	OVERHEAD SLINGS	26%	16%	30%	12%	4%
	ARM TROUGHS	8%	16%	32%	16%	12%
	BALANCED FOREARM ORTHOSES	42%	20%	26%	2%	0%
	DYNAMIC SUBLUXATION SLINGS	38%	28%	12%	2%	2%
	CONVENTIONAL SUBLUXATION SLINGS	44%	20%	8%	4%	8%
UPPER & LOWER EXTREMITY	ANTI-SPASTICITY ORTHOSES	10%	44%	32%	8%	0%
	EDEMA CONTROL ELASTIC GARMENTS	2%	44%	34%	6%	4%
	FRACTURE ORTHOSES	44%	26%	12%	0%	2%
	CONTRACTURE REDUCTION ORTHOSES	16%	48%	18%	0%	4%
	LOWER EXTREMITY ORTHOSES	58%	18%	4%	0%	0%
	SPINAL ORTHOSES	68%	8%	6%	2%	0%

*FABRICATED/FIT BY RESPONDING THERAPISTS IN A TYPICAL MONTH.

FIGURE V
~~~

laborate in the process of delivery of orthotic services. This consultation could occur during referral, evaluation, fabrication/fit, check-out, training or follow-up. Orthotists are singularly absent from this graph. This lack of communication between professionals involved in the management of the orthotic client indicates that a potential exists for duplication of services, failure to provide services at all, or otherwise poorly coordinated services.

Figure 5 reflects the frequency that responding therapists construct upper, lower and spinal orthoses. Predictably, the greatest activity is related to the upper extremity. Approximately one-fifth are involved in spinal and lower limb orthotics. But a clear majority are regularly delivering orthotic services that encompass both upper and lower extremity.

More than half of the responding therapists utilize pre-fabricated orthoses less than 5 percent of the time (Fig. 6).

In Figure 7 more than one-quarter of those answering indicated they spend more than half their time on orthotics. Cross-tabulations of this sub-group reveals a dramatic increase in the use of advanced materials, i.e., aluminum, high temperature thermoplastics, plaster and Plastazote, when compared to the overall survey. Similar significant gains occur in construction of spinal and lower extremity orthoses. Over 90% of this sub-group are involved in contracture-reduction, anti-spasticity, edema-control and fracture bracing. Only 8% regularly collaborate with orthotists.

DISCUSSION

It is readily apparent that this survey reflects only the perception of occupational therapists. Other health-care professionals might respond very differently to a similar questionnaire. The academic education for occupational therapists in orthotics is largely limited to low

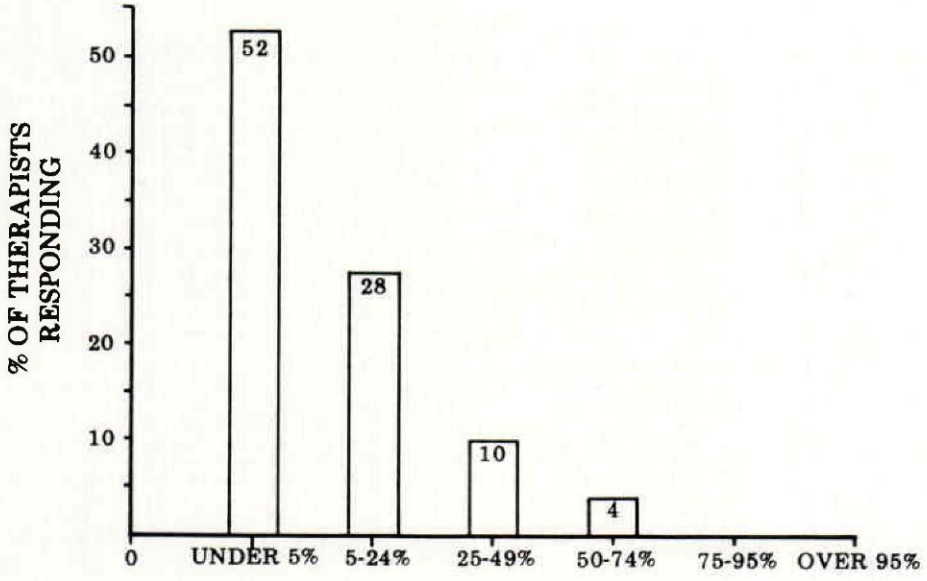


FIGURE VI

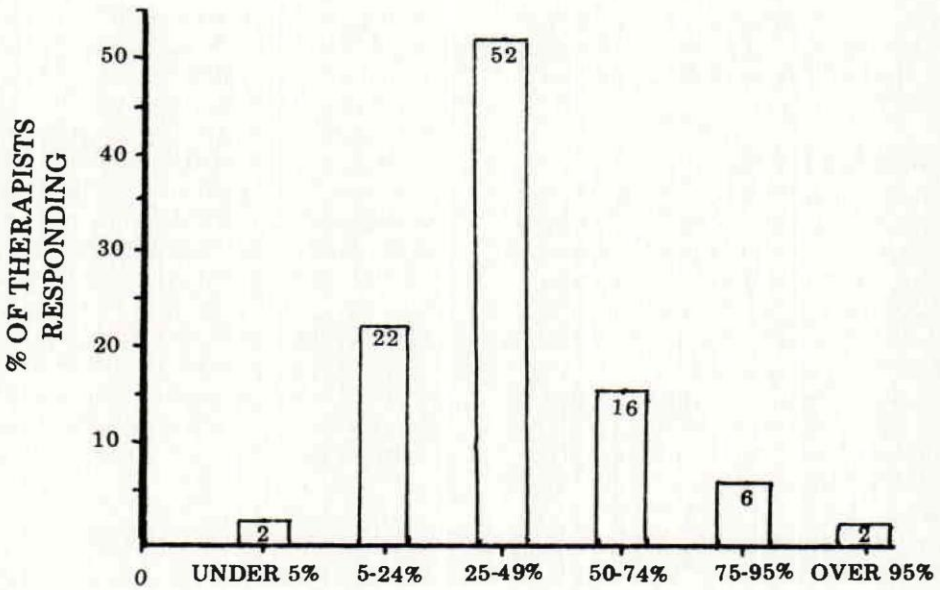


FIGURE VII

temperature thermoplastics and the upper extremity.[5,6,7] Yet clinicians in the field report significant involvement in a broad spectrum of orthotic materials and products. Obviously, of these skills are not acquired at the basic professional level, they are cultivated elsewhere. This learning process could occur through post-graduate courses, continuing education workshops or on-the-job training. Regardless, there is a potential possibility that a sub-specialty of occupational therapy is developing that parallels the role of the orthotist. The interrelationship of roles extends beyond the simplistic division of interim or definitive orthotics.

There are several possible explanations for this phenomenon. Most orthotists are independent and not directly linked to the staff of a particular institution. As such, contacts with any institution are limited to occasional clinics and consultations. While many referrals do pass directly to the orthotist from the physician, other health professionals who are intimately involved in implementing therapy may lack knowledge of an orthotists' skill.

The influence of third-party reimbursement may also contribute to the changing status quo. Selection of which professional provides orthotic services is often dependent upon the coverage of the client's particular policy. Plan A may reimburse therapy but not orthotic management. Plan B may cover both, but only under strict restrictions. The vast majority of medical insurance plans cover only the cost of the orthotic product and not related orthotic services. As such, orthotists are unable to charge for evaluation, training, and follow-up related to the orthotic process. Reimbursement is entirely dependent on delivery of an end-product. Thus, if the orthotist is financially inhibited from providing orthotic services, the staff occupational therapist may be called upon, not because the occupational therapist's are necessarily the most appropriate; but because they are readily accessible and have a basic knowledge of biomechanics,

materials, and fabrication techniques. Most importantly occupational therapists are reimbursed for services as well as the end-product.

CONCLUSION

Dramatic changes have altered the relationship between occupational therapists and orthotists. It is no longer appropriate to define roles in terms of interim vs. definititve or upper vs. lower orthotics. Additional research must be compiled before any meaningful delineation of roles can be ascertained. It is apparent that occupational therapists are being called upon to provide orthotic services they were not academically prepared to perform. Equally significant, due to the orthotic profession's detached status and the reimbursement system, orthotist's are not an integral part of the rehabilitation team. Some method must be found to bring these professionals in close collaboration with each other in the delivery of orthotic services. The potential development of competing, overlapping roles—which can only result in higher medical costs and increased confusion to the consumer—must be avoided.

Education in both fields needs to stress the unique skills both professions inherently offer. Increased emphasis must focus on regular collaboration between occupational therapists and orthotists. If orthotists are to retain a major role in upper limb orthotics, it may be necessary to engage in aggressive efforts to establish collaborative and consultative relationships with occupational therapists. When occupational therapists receive referrals for lower limb or spinal orthoses, they should be encouraged to collaborate with orthotists. Orthotists need to increase their visibility at local institutions in clinical settings. A stronger interface between these related professions could also be promoted through joint workshops, research projects, inservices, publications and promotional displays. Progress in these critical areas of joint cooperation should alleviate the prospect of an increasingly competitive relationship.

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AZ-1	ILL-3	MINN-1	WI-3
CA-6	IA-2	MO-5	
CO-1	KS-5	OH-1	
CN-1	NE-2	OR-1	
DC-1	NJ-1	TN-1	
DEL-2	NY-4	TX-5	
FL-3	ND-1	UTAH-1	

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ACKNOWLEDGEMENTS

The writer wishes to express appreciation to Dorothy Anne Elsberry, Associate Professor of Occupational Therapy for assistance in compiling statistical data. Graphics were generously provided by Harvey Hofstetter.

Kinematic Comparison of the BiCaal Orthosis and the Rigid Polypropylene Orthosis in Stroke Patients

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A rigid ankle-foot orthosis (AFO) is commonly applied to the lower extremity to stabilize the limb and correct gait deviations caused by neuromuscular disease. The rigid design is useful in correcting plantar flexion deformities resulting from moderate or severe spasticity or to stabilize the ankle of a flaccid limb. In addition to sagittal plane stability a rigid ankle-foot orthosis is capable of supporting the ankle in the frontal plane, correcting existing varus or valgus. In the past the bichannel adjustable ankle locking (BiCAAL) orthosis has been the traditional and most commonly used orthosis with a rigid ankle design (Fig. 1). The adjustable double action joint mechanism enables the clinician to set the AFO in the optimal position.[1 7]

Widespread use of plastic polymers in lower extremity orthotics has occurred in the last decade.[8 12] Orthoses fabricated with these materials are lighter, more comfortable, and more cosmetic. The typical plastic AFO is flexible and allows more motion than the BiCAAL orthosis. It, therefore, is less effective in stabilizing the ankle and foot when rigid support is desired. Although rigid plastic AFOs are in use, their effectiveness is questionable because they still retain some flexibility and are not adjustable.

The purpose of this paper is to report on a kinemetric evaluation of the rigid polypropylene orthosis compared to true conventional metal BiCAAL design.

METHOD

Patients already fitted and accustomed to a BiCAAL AFO were fitted with a rigid ankle polypropylene AFO. Kinematic gait characteristics were studied to make an objective comparison of walking performance in the two orthoses.

EQUIPMENT AND INSTRUMENTATION

Orthoses

The rigid polypropylene AFO was custom fitted over a positive mold of the subject's leg. Conventional vacuum forming techniques were used to fabricate the orthosis from a sheet of polypropylene (Fig. 1).[9 10] Rigidity was obtained in two ways: by using a thick sheet of polypropylene,

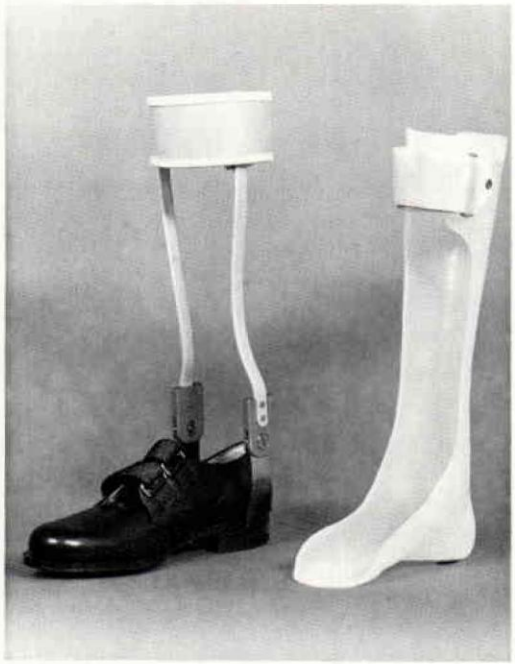

Fig. 1—The BiCaal and Rigid Ankle Polypropylene Orthosis.

and by extending the trim lines anteriorly at the ankle. Three-sixteenths inch polypropylene was used initially in contrast to the more commonly used ⅛ inch material used in most conventional orthoses. The medial and lateral trim lines were extended anteriorly to the apex of the malleoli to serve as supporting side struts. The footplate of the orthosis extended just proximal to the metatarsal heads.

The insertion of the polypropylene orthosis inside the shoe usually required a shoe size one-half size larger and the next width wider that previously worn. To eliminate the purchase of two pairs of shoes, once a shoe was fitted on the polypropylene side, an inlay was placed in the shoe of the second limb.

The ankle position of the polypropylene orthosis was set so that with the patient wearing a shoe with his normal heel height (1.3 to 1.9 cm), the tibia extended several degrees backwards from the vertical position. This slight plantar flexion position was chosen because patients with ankle instabililty secondary to paresis of the triceps surae generally walk in one of two abnormal gait patterns. In some, the tibia collapses forward during the stance phase and the ankle is excessively dorsi-flexed. There is compensatory knee flexion and greater than normal quadriceps contract is necessary to stabilize the knee. Most patients quickly learn a second gait pattern. The knee is purposefully hyperextended just after heel contact before significant limb loading occurs. Instead of excessive forward tibial rotation the tibia extends backward and the ankle is in plantar flexion. Because the patient's center of gravity passes anterior to the axis of the knee joint the knee is stabilized by the posterior joint capsule so excessive quadricaps activity is not required.

If an attempt is made to stabilize the tibia in the true vertical position with an orthosis there may be a tendency for the tibia to rotate either forwards or backwards against the shank cuff, since some ankle motion occurs within the orthosis. Because most patients perceive forward tibial rotation to be unstable, orthosis was positioned in slight plantar flexion to ensure the unstable tibia would always be thrust backward.

The BiCAAL AFO was of standard design.[1,6] The ankle joints had anterior and posterior channels enabling adjustment of the ankle in the sagittal plane. The uprights were attached via a steel stirrup to an oxford style shoe containing a steel shank extending to the metatarsal heads (Fig. 1).

Foot Switch System and Level Walkway

An insole foot-switch system provided quantitative information about foot-floor contact patterns on both the involved and sound limbs.[13] The insole contained four switches; under the heel, great toe, and the heads of the first and fifth metatarsals. Each switch consisted of a Pressex® module which when compressed, completed a circuit. Each switch transmitted a different voltage for output on a strip chart recorder. Analysis of the combined output voltage enabled identification of the different combinations of weight bearing.

All foot-floor contact information was transmitted to a receiver by a radio telemetry system and recorded on analog tape and a visicorder. The most representative pattern of contact was determined for each subject and used for comparison as subjects walked in the rigid polypropylene and BiCAAL AFO's.

All walking was done on an indoor level 15 meter walkway. The middle six meter segment, with a light beam emitter to mark the patient's entry and exist at either end was used for data collection. The light beam triggered a signal superimposed on the foot switch record to enable a computation of gait velocity.

Electrogoniometers

Electrogoniometers described in a previous report[13] measured the ankle and knee rotation. The knee goniometer was positioned in front of the joint. It consisted of a double parallelogram linkage secured by cuffs above and below the knee. Joint rotation drove a potentiometer attached to the proximal arm of the linkage. This enabled direct measurement of knee position. Cables supported by an overhead track above the walkway transmitted the output.

The ankle electrogoniometer was constructed similarly to the knee goniometer. It was secured above and below the ankle joint and recorded the angular motion occurring between the tibia and foot in the sagittal plane.

PROCEDURE

Data from the foot switches and electrogoniometers was received during the middle six meters of the fifteen meter walkway. The first test session involved walking at free velocity in the BiCAAL AFO. One warm-up run was permitted.

A similar second test session using the rigid polypropylene orthosis was scheduled a minimum of two weeks after the fitting to allow the patient to become accustomed to the new orthosis. They were instructed to wear the new orthosis exclusively during this time.

Subjects

Each of the 17 hemiparetic patients selected for the study could walk safely in a prescribed BiCAAL AFO without the aid of a cane, crutch or walker and had worn the orthosis for an average of 28 months (range: 3 to 132 months) prior to fitting with the rigid polypropylene AFO. Their ages ranged from 23 to 70 years (mean: 53 years, Table 1). Each subject had unilateral hemiparesis of varying degrees, ten were right and seven were left hemiplegic. All

Table 1
PATIENT DATA

Subject	Sex	Age (Years)	Hemiparesis	Time Since Onset[a]	Time in BiCAAL AFO[b]
1	Female	70	Left	7	6
2	Male	60	Right	36	31
3	Male	53	Left	7	5
4	Male	63	Left	5 Years[b]	2.5
5	Male	53	Right	20	17
6	Male	43	Left	7	6
7	Male	53	Right	7	5
8	Female	55	Right	11 Years[b]	36
9	Male	23	Right	10	7
10	Female	30	Right	15	13
11	Male	65	Left	6	4
12	Female	64	Left		4
13	Male	61	Right	9	6
14	Female	57	Left	5	4
15	Female	58	Left	8.5 Years[b]	6.2 Years
16	Male	70	Right	8	4
17	Female	23	Right	3	3

[a]At time of testing, in months.
[b]Date of initial episode.

Table 2
Average Stride Characteristics Generated by Subjects as they Walked in the Polypropylene and BiCAAL AFO's at Free Cadence. N = 16[a]

Stride Characteristic	Polypropylene AFO		BiCAAL AFO	
Velocity (cm/sec)	57±	(34)	51±	(24)
Cadence (Steps/min)	80	(21)	77	(17)
Stride Length (meters)	.80	(.3)	.76	(.2)
(%) Normal Single Limb Support Time	49	(25)	51	(20)
(%) Gait Cycle Swing	39	(7)	48	(6)
(%) Gait Cycle Total Double Support Time	36	(13)	26	(7)
(%) Gait Cycle Initial Double Support Time	16	(5)	10	(5)
(%) Gait Cycle Terminal Double Support Time	21	(12)	16	(6)

[a]Subject 5 dropped out of study prior to second test session.

were able to follow instruction for participation in the study.

Ankle stability depends on the presence or absence of selective muscle strength, patterned motion, spasticity in the triceps surae, proprioception, tactile sensation and ankle range of motion. The participants were grouped according to their degree of spasticity to determine how this component of tibial control affected ankle position. Of the sixteen persons, five had moderate to severe plantar flexion spasticity. In these latter persons the orthosis was worn to stabilize the ankle and prevent excessive forwards or backwards ankle rotation.

RESULTS

Stride Characteristics

The average values for velocity, cadence, and stride length in the group did not differ when using the rigid polypropylene orthosis or the BiCAAL orthosis (Table 2). The average velocity in the rigid polypropylene AFO was 57±34 cm/sec and in the BiCAAL AFO 51±24 cm/sec. Stride lengths were .80±.30 meters in the polypropylene AFO and .76±.24 meters in the BiCAAL AFO. Cadence in the polypropylene AFO averaged 80± steps/minute and 77±17 steps/minute in the BiCAAL AFO.

There was a significant difference (P□.05) in the involved limb swing time. Expressed as a percent of the gait cycle it averaged 48±6 percent in the BiCAAL and 39±7 percent in the polypropylene AFO. No difference (P□.05) was found in the amount of time spend in single limb stance. Double support time, however was significantly greater (P□.05) with the rigid polypropylene AFO (36±13 percent of the gait cycle) than with ambulation in the BiCAAL AFO (26±7 percent of the gait cycle).

When the subgroups were compared, the degree of clinical plantar flexion spasticity had no consistent effect on the stride characteristics in either of the orthoses.

Knee and Ankle Joint Rotation

Both AFO's permitted some motion at the ankle (Figure 2). Similar ankle joint rotation patterns were noted for the two orthoses with the polypropylene AFO demonstrated a slightly greater flexion range throughout the gait cycle. At initial contact both orthoses were in plantar flexion with a gradual dorsiflexion deflection occurring in mid-stance and reaching a peak in terminal stance. A rapid increase in ankle plantar flexion occurred at the end of terminal stance after heel rise and continued throughout the swing phase of gait. Maximum plantar flexion occurred at initial contact. The only statistical difference (P□.05) in ankle posture between the two AFO's existed at toe-off. Ambulation in the rigid ankle polypropylene AFO yielded an average of 4±5 degrees of plantar flexion, while in the BiCAAL orthosis the average was 1±5 degrees of dorsiflexion.

Significant differences (P□.05) in the amount of knee motion occurring between

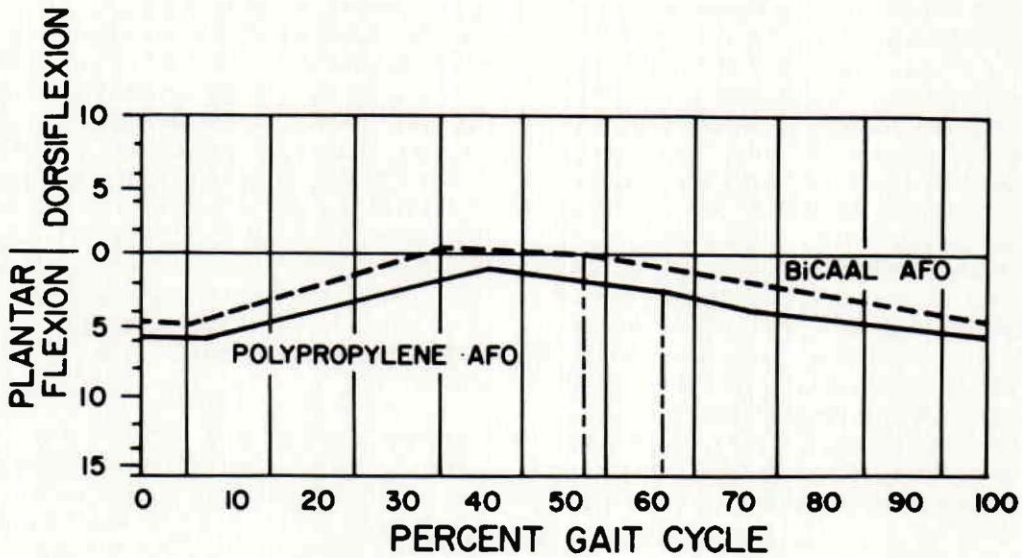

Fig. 2—The mean ankle joint position during free cadence walking with the polypropylene and BiCAAL AFO's.

ambulation in a polypropylene and a BiCAAL AFO were found at initial contact during free velocity walking. In the rigid ankle polypropylene AFO the participants demonstrated greater knee extension at initial contact (polypropylene AFO 6±8 degrees of knee flexion, BiCAAL AFO 10±7 degrees of knee flexion). A trend of more mean knee extension in the polypropylene AFO existed throughout the gait cycle although only statistically significant at initial contact.

Knee and ankle range of motion throughout the gait cycle were not influenced by the degree of clinical plantar flexion spasticity.

Foot-Floor Contact Patterns

The foot-floor contact patterns at initial contact were influenced by the degree of ankle plantar flexor spasticity and the type of orthosis. Fifteen of 16 subjects initially contacted the floor with the heel using the polypropylene AFO. While using the BiCAAL AFO only nine had normal heel first contact patterns. The differences were most apparent in the group of patients with excessive plantar flexion spasticity. Four of five subjects with moderate to severe plantar flexor spasticity had heel first foot-floor contact pattern when they wore the polypropylene orthosis. None of these subjects had this pattern in the BiCAAL AFO. In the group of 11 patients with absent to minimal plantar flexion spasticity and triceps surae paresis, all persons had heel first initial contact patterns when the polypropylene AFO was worn. Nine of 11 persons had this pattern in the BiCAAL AFO.

When the polypropylene AFO was worn, seven of sixteen persons demonstrated first and fifth metatarsal or first, fifth and great toe foot-floor contact patterns at terminal stance. In the BiCAAL AFO, these characteristic patterns were not noted. When the subgroups of plantar flexion spasticity, were analyzed those persons with absent to minimal increases in tone had four, and those with moderate to severe tone had three of these patterns. No trend existed between the degree of clinical spasticity and the foot-floor contact patterns at terminal stance. The patterns appear to be more dependent on the type of orthosis rather than the degree of plantar flexion tone.

Patient Preference

Of the 17 patients who began this study, three chose not to wear the polypropylene AFO after the initial two week period of accommodation. One subject complained of a burning sensation on the foot and could not tolerate the total limb contact provided by the plastic orthosis. Another stated he preferred the BiCAAL orthosis because he felt unstable while wearing the polypropylene AFO. His velocity and single limb support reflected his feelings of instability. In the BiCAAL orthosis the patient demonstrated a greater percent normal single limb support time (BiCAAL 58 percent; polypropylene 30 percent) and a greater free velocity (BiCAAL 58 percent; polypropylene orthosis had difficulty putting on his shoe over the AFO.

DISCUSSION

The results indicate that a rigid polypropylene orthosis can provide the same ankle stability as a BiCAAL orthosis. No significant differences were noted with respect to gait velocity, cadence, or stride length when either orthosis was used to correct plantar flexion deformity resulting from spasticity or to stabilize the flaccid ankle in the paretic limb.

Patients walking in the polypropylene orthosis had a shorter swing time and spend a greater percentage of the gait cycle in double limb support than when using the BiCAAL orthosis. One possible explanation for the reduced swing time is the light weight of the rigid polypropylene orthosis, 250-400 grams, versus the BiCAAL orthosis weight of 800-900 grams (weight varies slightly because of differences in shoe size, style and amount of material for different sized patients).[12] A more normal pattern of foot-floor contact was observed in patients wearing the rigid polyproylene AFO at initial contact and at terminal stance. When the polypropylene AFO was worn the involved limb was advanced at a faster rate creating greater momentum. This afforded the patients a better opportunity to achieve terminal knee extension and a heel first initial contact foot-floor contact pattern. Significantly greater knee extension at initial contact in the polypropylene AFO as compared to the BiCAAL AFO at free cadence walking demonstrated this factor. Since the velocity of walking is similar in both orthoses, this decrease in swing time in the polypropylene AFO enables this group to devote a greater percentage of the gait cycle to the stance phase of gait. This is reflected in greater initial and terminal double support times. This increase in balance assist time could account for the greater number of normal foot-floor contact patterns occurring during terminal stance in the polypropylene AFO.

Despite the fact that the patients had worn the BiCAAL orthosis a minimum of three months prior to entry into the study, the majority (14 to 17) preferred the rigid polypropylene orthosis. Since our data indicates the rigid polypropylene orthosis can provide the same degree of rigidity as the BiCAAL orthosis, its obvious superiority in terms of weight, cosmesis, and the ability of the patient to interchange shoes with the same heel height makes it the more preferable for most patients for chronic usage.

In our current clinical practice the BiCAAL orthosis is usually prescribed as the initial orthosis for patients with excessive plantar flexion deformity secondary to spasticity or ankle instability in a flail limb. The adjustable ankle feature allows the clinician to determine the optimum ankle position which varies among patients. This feature is also useful if the patient's neurologic condition has stabilized and the optimum ankle position has been determined, the rigid polypropylene orthosis is used as the permanent treatment of choice.

SUMMARY

The lighter polypropylene AFO promotes more efficient advancement of the involved limb, allowing a greater percentage of the gait cycle to be devoted to the stance phase of gait. It was functionally equivalent to its counterpart, and was aesthetically acceptable to the majority of the

stroke patients who participated in this study.

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Augmented Feedback Spinal Orthosis
An Introduction and Preliminary Report

Larry Mortensen, C.O.

INTRODUCTION

The Milwaukee* cervical-thoracic-lum bar-sacral orthosis has been the standard non-operative treatment for scoliosis in growing children since 1945 (Fig. 1). End result studies of the use of the Milwaukee orthosis have shown that it prevents curve progression in the majority of curves between 20 and 45 degrees.[1] The thoracolumbar-spinal orthosis originating in Boston and adapted elsewhere has shown similar results to those with the Milwaukee orthosis.[2] [3] The efficacy of the Milwaukee orthosis was thought to be due to an active shift of the trunk away from the thoracic pad within a restricted space where the occiput was aligned over the sacrum.[4]

In order to help enforce the lateral trunk shift exercise within the orthosis, a tactile stimulator of the trunk with a plastic CTLSO was designed. The stimulator signals at preset intervals when the patient is to shift the trunk away from it and shut off the stimulator by the shift. In this way, exercise is enforced while the patient is erect and awake. By incorporating the major features of the Milwaukee orthosis with the all-important active voluntary exercise of lateral trunk shift, it is hoped to improve the results of treatment by achieving reduction of the spinal curvature. The mechanisms of reinforcing the active exercise in the orthosis is termed "augmented feedback."

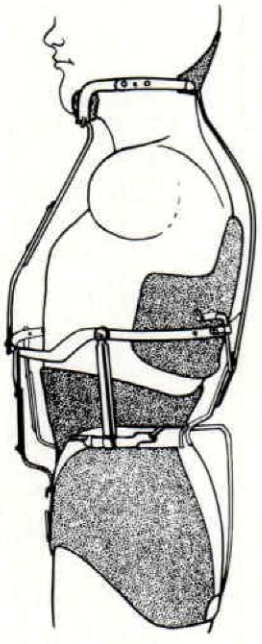

Fig. 1—The 1946 Milwaukee orthosis was constructed of leather and steel, which consisted of a pelvic portion, steel super-structure and a mandibular-occipital component (later found to cause orthodontic problems and eliminated). This orthosis was used as management for post operative protection at that time.

*An orthotic system which was conceived in Milwaukee, WI in 1945 by Walter P. Blount, M.D., John H. Moe, M.D. and through the technical pioneering efforts of Mr. Richard Bidwell, C.O. This orthotic approach has evolved technically through the years, but the biomechanical principles are very sound to this date.

The project goals were as follows:

- redesigning, formulating and fabricating an orthotic system incorporating a TLSO plastic component with a modified Milwaukee superstructure.
- using a refined negative casting procedure and positive model modification.
- determining the practicality of a refined orthosis in terms of correction, spinal realignment, comfort and cosmesis.
- determining the effect of the augmented feedback in improving and maintaining correction of the scoliosis during the spinal growth spurt in children.
- interfacing an electronic mechanism to maximize a lateral trunk shift exercise with an orthosis.
- inducing the patient to apply a corrective musculoskeletal action, thereby obtaining optimal therpeutic function in maintaining and limiting curvature of the spine through a dynamic force system.

PATIENT SELECTION

Twenty patients were selected for fitting of the augmented feedback spinal orthosis (AFSO); all had idiopathic scoliosis and growth potential with a Risser* sign (Iliac crest apophysis ossification) of three or less. All curvatures (except in two patients) were greater than 20 degrees. Patients 6 and 13 had curvatures of 19 and 14 degrees respectively. Both had been followed by observation; only one had demonstrated curve progression.

All patients had curvatures measured by the Cobb method† on standing anterior-posterior and initial standing lateral radiographs of the spine. Moire‡ polaroid photographs were made of the trunk as additional documentation. Radiographic examinations and Moire photographs were used in follow-up examinations.

Flexibility was determined by a posterior-anterior radiographic projection of the spine when manual pressure was applied to the apex of the curve and counter pressure against the opposite lateral aspect of the pelvis.[5] Active correction of the major curve was documented with a radiograph of the spine taken when the patient actively shifted the trunk away from the apex of the major curve. (Figs. 2a, 2b, and 3a x-ray; 3b, 3b x-ray).

Growth was assessed by interpretation of Iliac crest apophyseal ossification according to the Risser method, Tanner‡ stages of growth and serial measurements of the patient's height. Radiographic examinations to determine skeletal ages were performed in some patients but were of no value in most patients because all

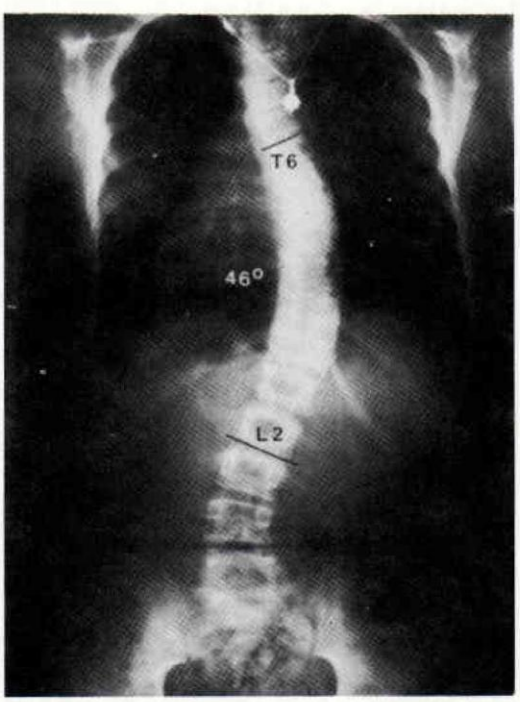

Fig. 2a—Anterior-posterior films are taken to determine curve patterns and degree of curvature. Also taken as a follow up to determine whether the patterns are progressional.

This radiographic film follow up indicated a progressional curve.

*Risser sign: is used to determine bone growth potential; a radiographic film is used to determine Iliac crest apophysis ossification.

+ Cobb Method: a method to determine the length of a curve and to denote degree value of that curve.

‡Moire topography is a non-radiographic approach using a photographic method, and instrumentation, for scoliosis screening.

‡Tanner stages of secondary sexual characteristic: a clinical method to be used to determine growth potential related to osteo-maturation.

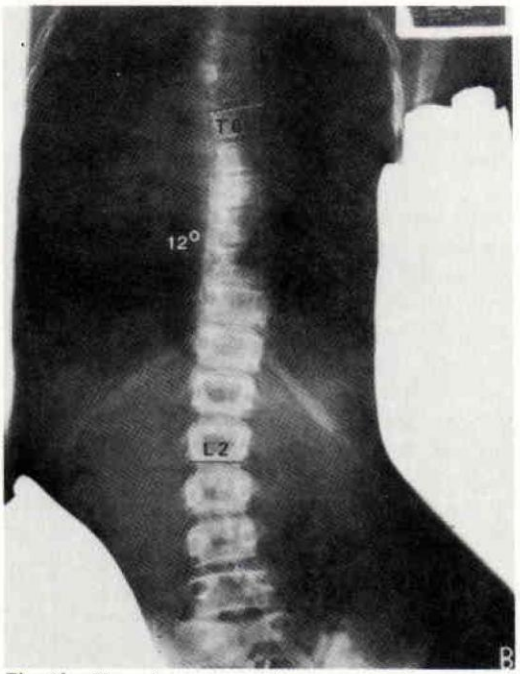

Fig. 2b—Use of a prone push posteroanterior film of the spine, in which manual pressure is applied, determines spinal flexibility in patients with scoliosis.

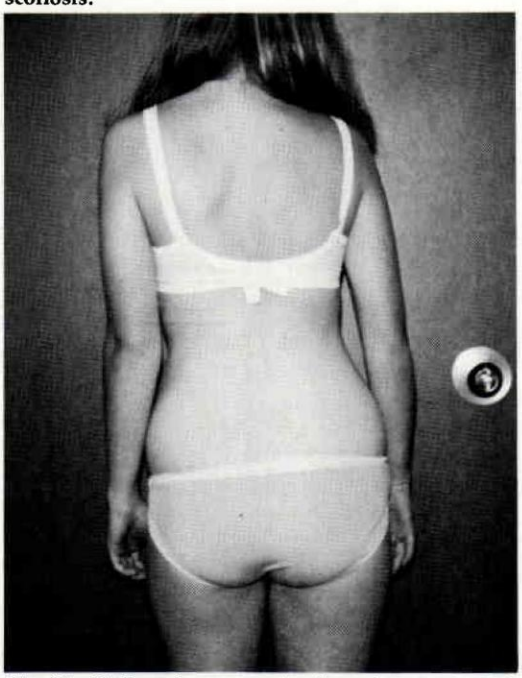

Fig. 3b—Patient demonstrating lateral musculoskeletal movement away from the apex of the right thoracic curve to the concave side of that curve.

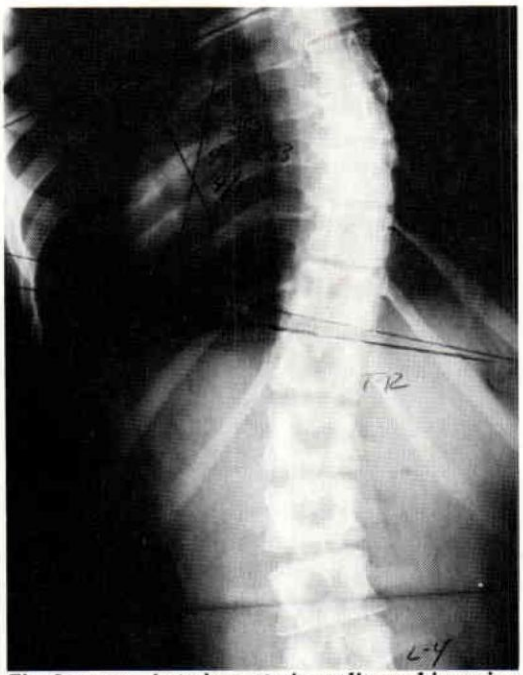

Fig. 3a x-ray—Anteriorposterior radiographic projection of the spine of patient indicating right thoracic curve from T-5 to T-11 with a degree value of 36 degrees and a left lumbar curve from T-12 to L-4 of 29 degrees.

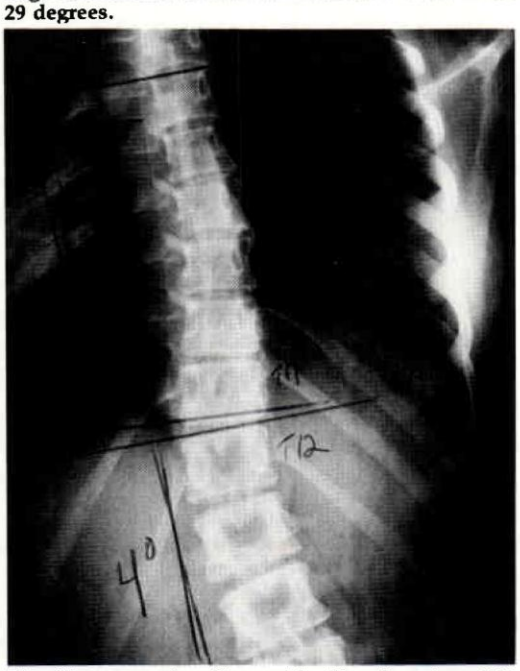

Fig. 3b, x-ray—A radiographic film denoting shift movement indicating flexibility and muscular correctional control patient can exhibit. Note; right thoracic now 7 degrees and left lumbar, now 4 degrees.

had obvious growth potential in consideration of their mean chronological age of 11.6 years (range 8.6 to 15 years), and the Risser sign less than three. In addition, growth was demonstrated during the follow-up by measurements of their height (Table I).

Following these examinations, the patient was scheduled for an orthotic evaluation and casting procedure at the Rehabilitation Engineering Center (REC). Medical records and radiographs were reviewed to formulate the approach, modifications to the casting procedure and orthotist-patient communications.

An integral part of our orthotic treatment is to have open communication between orthotist, patient and parent(s), as well as developing the patient's psychological aspect of a positive attitude in relation to acceptance of any type of orthosis, stressing the importance of body awareness and presenting a positive image through posture. There is, naturally, an initial difficult adjustment period.[6]

NEGATIVE IMPRESSION PROCEDURE

A Risser casting frame is used for all of our spinal casting procedures (Fig. 4). The patient lies supine on the belt of the frame to reduce gravitational effects on the spine, thus allowing external manipulation of the trunk by applying pressure to the negative plaster impression to create a realignment effect based upon the apex of the curve.

Surface anatomy is used to determine hand positioning on the plaster impression. For example, if the patient we evaluate has an S-shaped curve, consisting of a right thoracic curve from T-5 to T-12 with an apex of T-9 and a left lumbar curve from T-12 to L-4 with an apex of L-3, surface anatomy can be more specific in relation to placement of external force (Fig. 5). The thoracic pressure area will be positioned laterally on the negative impression at a right angle to the xyphoid process (which indicates position of T-9 vertebra) so hand

Table I
FOLLOW-UP AFSO AFTER 12 MONTHS

Patient #	Age Mos.	Risser Sign	Curve Location	Degrees	AFSO Use Mos.	Growth Cms.	Degrees at follow-up	Increase or decrease degrees	Result
01. BF	161	161	0	T5-11	34	27	17.5	29	– 5
			T6-10	31					
2. LD	150	1	T11-L4	34	12	14.0	37	+ 8	–
3. EJ	155	0	T5-L2	24	22	13.0	38	–14	+
			T12-L4	27			26		
4. AS	157	3	L2-3	27	13	12.5	26	*0*	*0*
			T7-12	*20*			25		
5. AQ	*174*	*0*	*L1-4*	*18*	*27*	*4.0*	25	+ 5	–
			T6-9	19			17		
6. AW	114	0	T10-L3	24	18	22.5	21	0	0
			T6-11	36			35		
7. KK	167	1	T12-L4	22	14	14.5	25	0	0
8. SW	127	0	T7-L2	29	20	19.5	20	– 9	+
			T4-9	30			23		
9. CL	150	0	T10-L3	27	12	16.0	25	– 7	+
			T5-11	35			21		
10. MM	192	1	T12-L5	24	18	11.0	16	–14	+
			T5-11	40			50		
11. KB	132	0	T12-L4	40	19	6.0	50	+10	–

—Increase or decrease in curve of less than 5 degrees not significant.
—Result grading: + = improvement, – = worse, 0 = unchanged.

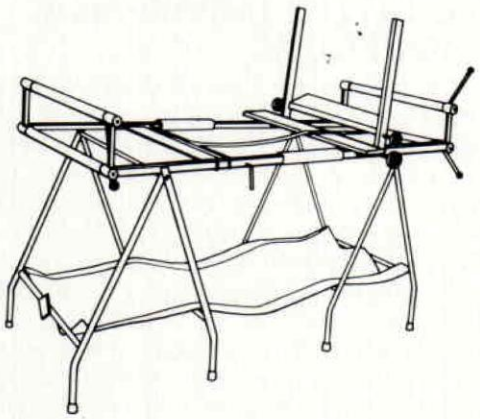

Fig. 4—Risser casting frame used for all our spinal casting procedures. Patient lies supine on the belt in the middle of the frame. This method allows us to limit gravitational forces acting on the spine.

position can be related to the apex of the curve on the convex side. This pressure is a lateral to medial force for realignment of the spine with somewhat of an anterior lift reducing the rotational aspect of that curve (Fig. 6). The hand placement for the left lumbar curve would be on the convex side of the curve at a right angle to the position of the umbilicus, which would indicate the apex of the third lumbar vertebra. The hand is placed posteriorly and laterally to the para-spinal muscles with a lateral-medial pressure realigning the lumbar spinal segment.

The casting procedure requires three people. Upon determination of the hand placement, the patient is positioned on the casting frame. Plaster bandage is applied, and the following steps are taken until hardening of the impression:

- One person controls pelvic rotation and lateral movement of the pelvis in relation to the transverse and frontal planes (Fig. 7). This person places his hands underneath the buttocks and a person from the opposite end, near the head, instructs that member to lift either the left or right hand, insuring that the buttocks are parallel to a horizontal surface. This allows us to maintain a symmetrical positioning of the pelvis within the cast.
- Then, as described, pressures are applied to the curve(s) until the plaster impression is dried to a point where

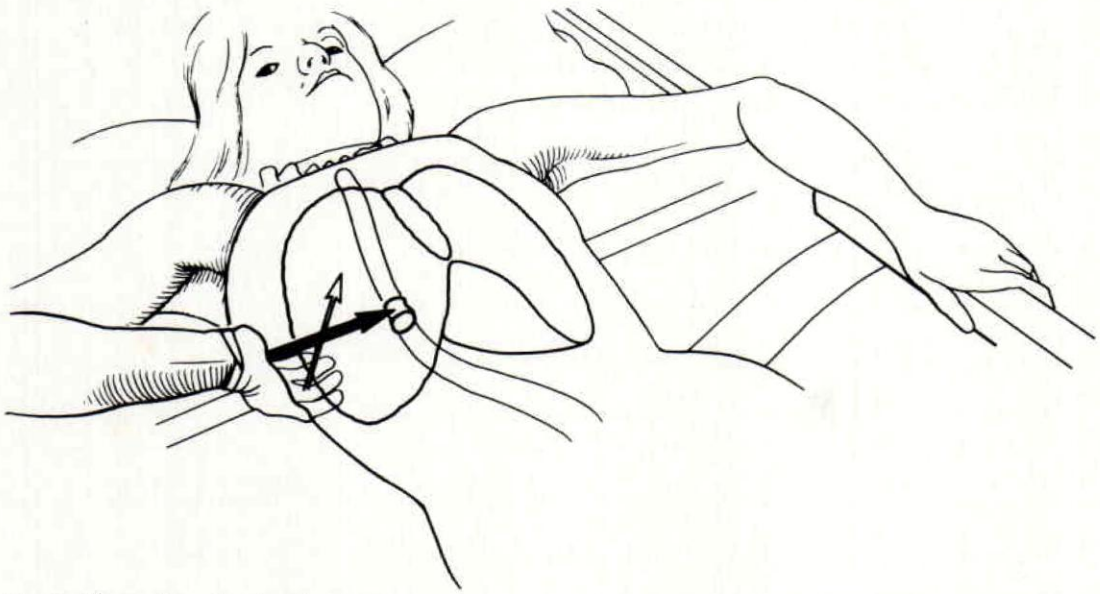

Fig. 5—Illustration of patient in plaster cast on Risser frame. Upon determination of hand placement pressure applied to the cast is sufficient to cause a lateral movement of the thoracic curve area from the convex to the concave side. A visual evaluation is made by a slight off set of the trunk in reference to the mid-sagittal line. This pressure is maintained until the plaster hardens.

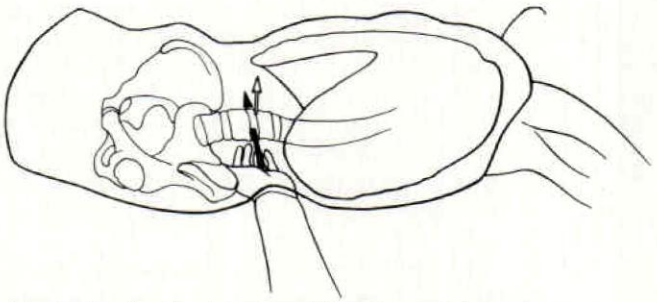

Fig. 6—Lumbar hand position. The palmer surface of the hand is placed near the paraspinal muscles in the lumbar area exerting a medial-lateral pressure from the convex to the concave side of that curve on the plaster cast as in Fig. 5, thoracic curve.

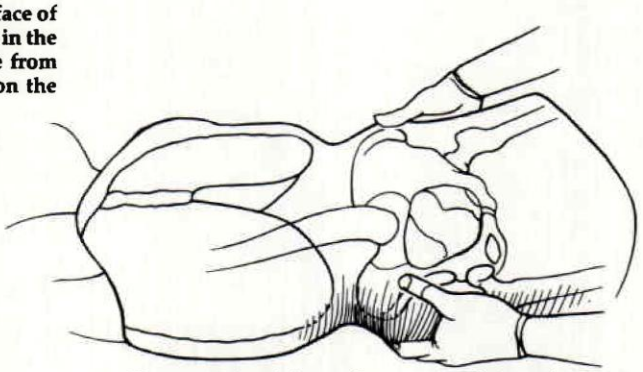

Fig. 7—Maintaining a firm grasp of the pelvis, as a stabilizing function, during the casting procedure establishes a symmetrical positioning of the pelvis in the cast where manipulative realignment pressures can be applied superiorly.

the patient can be lifted and stood erect on the floor.

- Metal washers are plastered near the apex of the external hand impressions on the plaster cast so they appear flat on the radiograph.
- A radiograph is taken of the patient in in the plaster impression for evaluation of the casting procedure to determine the amount of flexibility of the curve(s) and the amount of correction or realignment attained through the casting procedure.

The washer on the radiograph denotes the distance to the lateral aspect of vertebral bodies, the apex of the pressure of the negative hand impression in reference to the apex of the curve, whether the apex of that pressure needs to be moved superiorly or inferiorly, and the position and the angle of the ribs in regards to the apex of the curve in the thoracic area.

The lumbar washer placement gives the same indications as in the thoracic pressure area: the distance from the apex of washer to the lateral aspect of the vertebral bodies and the distance from the washer to para-spinal muscles in relation to the transverse processes in the lumbar area.

ORTHOTIC DESIGN BLUEPRINT

By drawing a horizontal line across the iliac crests on the film, a vertical ninety degree line can be drawn through the apex of each metal washer from the iliac crest line. By drawing right angle (reference to vertical line) horizontal lines on the radiograph through the apex of each metal washer to the lateral aspect of the lumbar (Fig. 8) and thoracic vertebrae, measurements are taken to determine positive model modification.

Lumbar Blueprint Layout

- From the horizontal line, we can determine the distance from the apex of the washer to transverse processes and the lateral edge of the lumbar vertebral bodies.

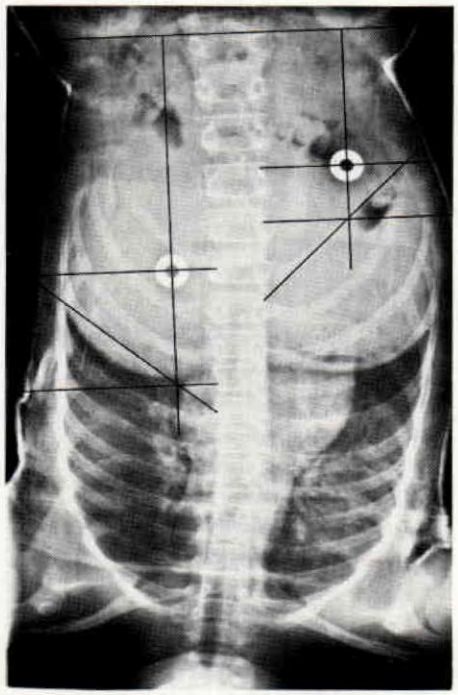

Fig. 8—Illustration of radiographic "blueprint" layout; a radiographic projection of patient in the negative plaster cast.

- From the vertical line, a vertical superior measurement is taken to identify the location and angle of the eleventh and twelfth ribs and an inferior measurement indicates location of the iliac crest of the pelvis. When modifying the positive model, a pressure relief can be built up, thus maintaining applied forces to the apex and transverse processes of the lumbar vertebrae. This reduces the amount of pressure being exerted on the ribcage (eleventh and twelfth ribs superiorly) and Iliac crest inferiorly.

Thoracic Blueprint Layout

The distance from the apex of the washer to the thoracic vertebral bodies is determined. The horizontal line is used as a reference to measure the angle of the ribs to the apex of the thoracic curve in relation to the apex of the pressure being applied.

Upon completion of the evaluation of the radiograph, the washer location is transferred to the positive model as a point of reference for positive model modifications.

The positive model basically reflects what is determined by the casting procedure, using the radiograph as a blueprint. That radiograph gives indications of maintaining what is achieved through the casting procedure. This method of positive model modification takes an asymmetrical positive model and realigns specific areas, which creates a symmetrical model maintaining asymmetrical pressure areas.

FABRICATION

The feedback orthosis consists of a TLSO component, vacuum-formed from a thermoplastic called Surlyn.* Surlyn was the plastic of choice because of its clarity, which enhances the cosmetic appeal of the device (Fig. 9). Due to the flexure of the material, it was necessary to provide more structural support to the plastic orthosis. Incorporating polypropylene rod crest components† into the vacuum-forming process reinforced the unity of the orthosis. The crest components are used for suspension and the foundation for forces can be applied.

The trimlines of the TLSO are similar to that of the Boston module† and Milwaukee orthosis. The inferior-anterior trimlines are cut to allow for hip flexion and sitting comfort. Superior-anterior trimlines are contoured along the base of the breast (Fig. 9a). On the convex side of the orthosis (right thoracic curve pressure area), the trim lines are high, allowing maximum distribution of pressure over a wider area (Fig. 10). The pressure pad area is concave or elliptical in shape to provide a fulcrum at the apex of the curve (Fig. 9a). This will promote a slight lateral spinal adjustment superior to the apex. The superior concave trimlines are similar to the convex side, left higher to place a counter pressure superior

*"Surlyn": trade name used by E. I. Du Pont de Nemours and Company (1007 Market Street, Wilmington, DE 19898), which is an Inomer resin produced by Du Pont Company. The plastic that is used in the drape forming process is Surlyn™, which is purchased from the United States Manufacturing Company under their trade name "Thermo-Vac Clear Plastic."

†The polypropylene "model #9493 Crest Pieces," which are used in the fabrication process, are purchased from Orthomedics, Inc. They are reheated to a pliable state and recontoured to each positive model.

†The Boston modular system: developed at Children's Hospital Medical Center, Boston, Mass., by Bill Miller, C.O., et. al.

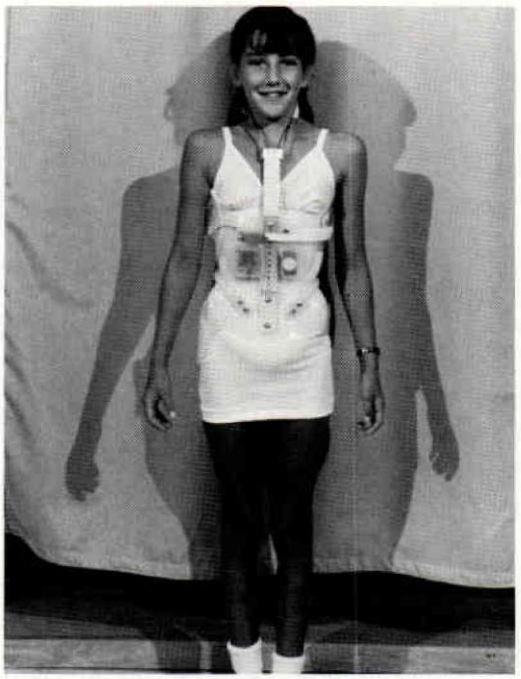

Fig 9—Anterior view of patient wearing Augmented Feedback Spinal Orthosis (A.F.S.O.).

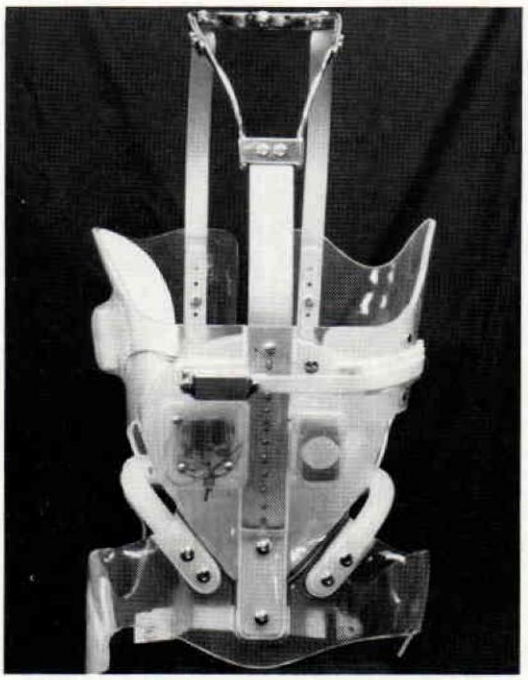

Fig. 9a—Anterior view of the A.F.S.O.

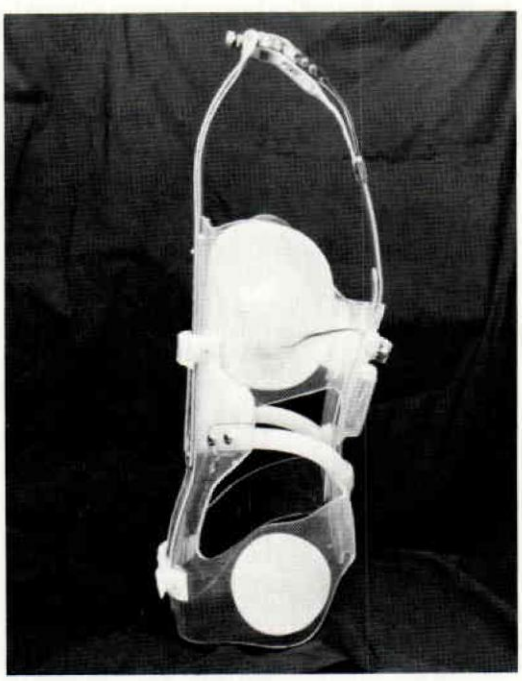

Fig 10—Right lateral view.

to the apex of the convex pressure (Fig. 11 and 11a). On the concave side, the posterior-superior trimline is cut out the permit movement of the scapula (Fig. 12 and 12a). The posterior inferior trimlines are trimmed approximately 2.5 cm above a hard surface to aid sitting comfort (Fig. 12a). The lateral inferior trimlines on the convex side of the thoracic curve are left low, approximately 2.5 cm to 3 cm. below the apex of the greater trochanter. This extends the lever arm of the thoracic pressure and incorporates another firm point of pressure in the force system (Fig. 10). On the lateral inferior concave side, the trimlines are high above the trochanter to allow lateral and pelvic motion (rotation). Large fenestrated areas are trimmed away to provide sufficient ventilation and heat adjustment to plastic for a growing patient (Fig. 11 and 11a).

Attached to the TLSO are standard and para-spinal bars. Superiorly attached to the superstructure is a redesigned cervical component* modified to remove the mandibular portion (Fig. 9 and 9a).

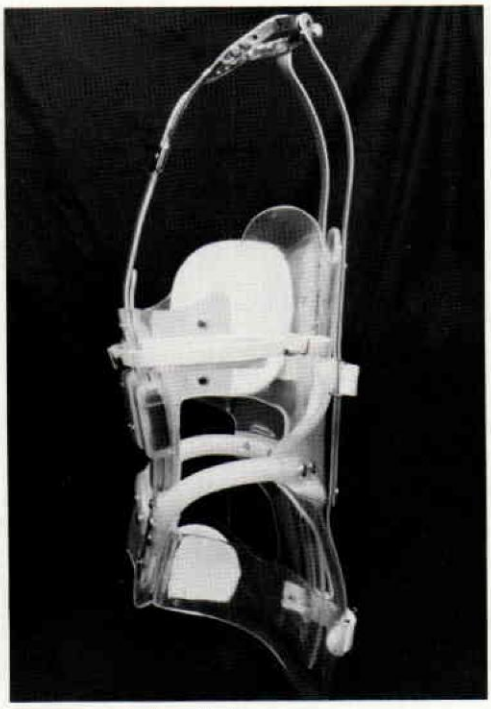

Fig. 11—Left lateral view.

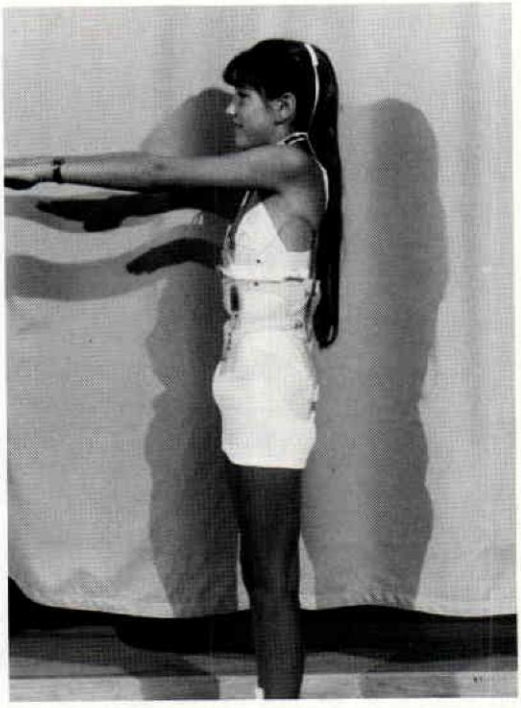

Fig. 11a—Left lateral view of A.F.S.O. on the patient.

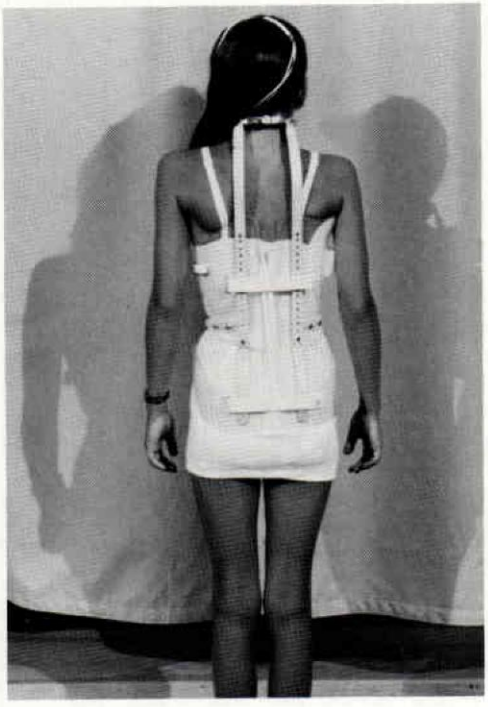

Fig. 12—Posterior view.

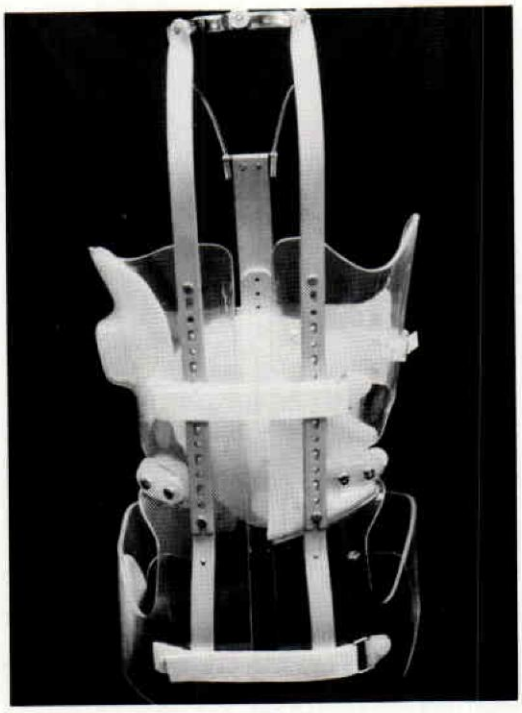

Fig. 12a—Posterior view of A.F.S.O. on the patient.

The purpose of the cervical component in this orthotic design is to control lateral movement of the head and cervical vertebrae, and maintain vertical alignment over the pelvis. The cervical ring is positioned low on the neck to make it more cosmetically appealing.

The TLSO has minimal foam interface. To areas having an extreme amount of correctional pressure, five millimeter medium density foam pads are glued in place to absorb some of the pressure being applied (Fig. 10 and 12). A large oval pad is placed laterally in the thoracic area, a triangular shaped pad is placed in the lumbar area and a circular pad is glued in place at the apex of the greater trochanter. Using a minimal amount of interfacing reduces the bulkiness of the orthosis, maintaining a more intimate fit.

On the abdominal portion—to the right side of the anterior bar—is the location of the electronic mechanism, which is a sophisticated timer (Fig. 13). This timer will activate the tactile stimulator every 45 minutes. The stimulator is a small electric motor with an offset flywheel that provides a vibratory tactile sensation. The stimulator is laterally located on or near the apex of the curve. Above the electronic unit is the pull switch*; its function through the strap attachment to the left side will discontinue the stimulator when the patient holds the lateral movement away from the convex side (Fig. 13a). The strap and strap attachment to the concave side of the curve are adjustable to minimize the exursion of the pull switch.

AUGMENTED FEEDBACK MECHANISM USAGE

The augmented feedback mechanism is interfaced into the orthosis to encourage lateral musculoskeletal movement away from the tactile vibratory stimulus on the convex side to the concave (Fig. 14).

*The headpiece, which includes the occipital brackets and throat frame mandible brakcets and superstructure are purchased from Fillaeur Orthopedic. The cervical component is modified by removing the mandible portion and the tangs on the occipital brackets.

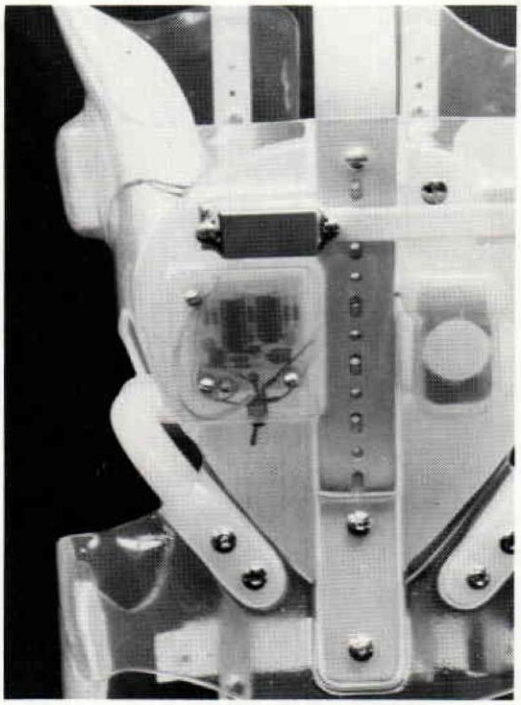

Fig. 13—Anterior abdominal view of the electronic section of the orthosis.

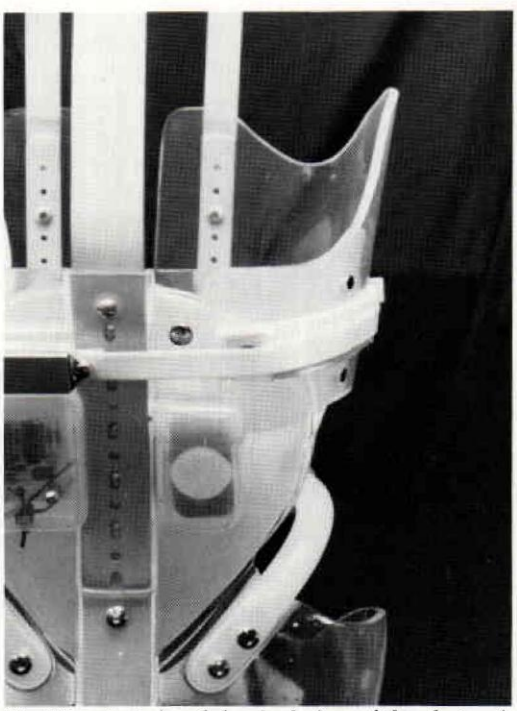

Fig. 13a—Anterior abdominal view of the electronic section.

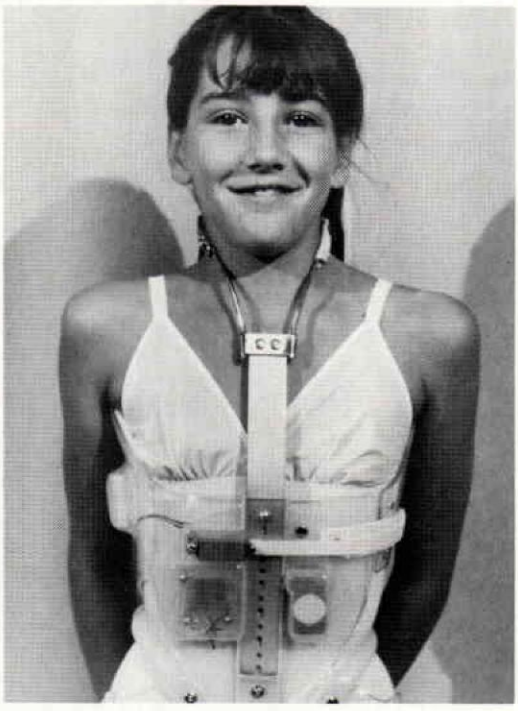

Fig. 14—Anterior close-up of the orthosis on the patient.

Through this mechanism, the paraspinal muscles on the convex side of the curve can be conditioned by performing an isotonic (or dynamic) exercise that occurs within the orthosis during the day hours. This conditioning and conscious effort by the patient to comply with the exercise program and regimentative wearing of the orthosis will, hopefully, have a beneficial effect upon the curve(s) by limiting further progression.

EXERCISE EVENT

Patients are instructed to wear their orthosis for 22 out of 24 hours. When donning their device, they secure the cervical component and then the velcro straps sufficiently enough to feel pressure within the orthosis. The two posteriorly-placed straps should have an equal amount of tension without incurring discomfort. A torso liner is worn beneath the orthosis.

The operation of the electronic system has two functions: the daily timed exercise event and the ten minute daily exercise.

The timed exercise event occurs approximately every 45 minutes. The patient is instructed as follows:

- When the stimulator's vibratory action is felt, you must keep your head and neck directly aligned over the pelvis, keeping your shoulders level (highly stressed when you are fitted with a TLSO not involving superstructure).
- Stand erect or, if sitting, make yourself taller.
- Move away from the sensation and hold that position for six seconds; this will discontinue the stimulator.

The ten-minute exercise is done in front of a mirror. The parents are involved in this segment of the exercise program to reinforce the child's efforts.

- Placing a magnet over an area on the electronic unit opens a switch and overrides the fixed time event, activating the stimulator.
- The patient would perform the same movement as instructed, relaxing for a few seconds and repeating for ten minutes.

Follow-up visits are made approximately every three months. All activities are encouarged in the orthosis with the exception of contact sports.

This report has described the protocol, casting, design and use of the Augmented Feedback Orthosis. This design has evolved over the last three years. In the past, more of our patients were fitted with TLSO's that with CTLSO's. Recently, we began using more CTLSO's (with superstructure), as the cervical component restricts occipital motion in relation to shifting exercise localizing the shift within the curve area exhibiting a more finite shifting pattern.

RESULTS

Eleven patients who were fitted with AFSO have been followed 12 months or longer. Of these 11, four major curves have decreased five degrees or more, four were unchanged, and three have progressed more than 5 degrees (8.5 and 10 degrees) (Table I).

Table II
FOLLOW-UP AFSO BEFORE 12 MONTHS

Patient #	Age Mos.	Risser Sign	Curve Location	Degrees	AFSO Use Mos.	Growth Cms.	Degrees at follow-up	Increase or decrease degrees	Result
			T6-12	24			19	− 5	
12. BQ	130	0	L1-4	22	3	0	13	− 9	
13. LW	150	0	L1-4	14	3	0	13	− 1	+
			T5-10	27			21		
14. KT	180	1	T11-L4	31	8	5.5	31	− 6	+
			T7-12	26			3	−23	
15. DK	153	2	T11-L3	16	6	2.5	23	− 7	+
			T5-11	27			24	− 3	
16. KO	169	2	T12-L4	27	6	7.0	22	− 5	+

—Increase or decrease in curve of less than 5 degrees not significant.
—Result grading: + = improvement, − = worse, 0 = unchanged.

Five patients have been followed less than 12 months. In all, the curvatures have decreased. One patient, age 14, was treated because of demonstrable progression of scoliosis during a period of observation (Table II). Four patients were eliminated from the study, four because they did not return after the orthosis was applied and one because indicated surgical treatment was delayed. When she was first seen, her curvature from the sixth thoracic to the first lumbar vertebrae was 52 degrees. She had severe congenital heart disease and her parents thought her to be a poor operative risk. Her curve increased to 57 degrees in six months. At this time, she was deemed a reasonable operative risk.

Of the 16 patients included in this report, only one eventually refused to wear the AFSO (Patient DK). Her reasons were interference with her social life, even though her major curve decreased 23 degrees in seven months.

Table III
AFSO HOURS OF WEAR PER DAY, RISSER SIGN, RESULT

Patient #	Hourse Wear Per Day	Risser Sign	Result
	23	0	
1. BF	22	1	+
	12	3	
	22	2	
2. LD	12	3	−
	22	2	
3. EJ	12	4	−
4. AS	16	3	0
	18	0	
5. AQ	8	4	−
6. AW	22	0	0
7. KK	22	1	0
8. SW	22	0	+
9. CL	22	0	+
	22	1	
10. MM	8	2	+
	5	4	
11. KB	12	2	−

DISCUSSION

Does duration of wear in 24 hours affect the outcome? Table III delineates the duration of daily use of the orthosis and the result to date. From this preliminary data, it appears that the duration of use to 20 hours per day or less adversely affects the correction.

We are sufficiently encouraged by the addition of augmented feedback to the principles of the Milwaukee brace in obtaining more than mere holding of a curve to possible correction that we have continued to offer it as a correcting device. It is likely that the best results will be obtained in idiopathic scoliosis when the major

Table 4
ASFO PATIENTS FOLLOWED AFTER
12 MONTHS (12-27 MONTHS)
RESULT RELATED TO GROWTH
(total cms. during treatment)

Patient #	Height Gained Cms.	Result
1. BF	17.5	+
2. LD	14.0	–
3. EJ	13.0	–
4. AS	12.5	0
5. AQ	4.0	–
6. AW	22.5	0
7. KK	14.5	0
8. SW	19.5	+
9. CL	16.0	+
10. MM	11.0	+
11. KB	6.0	–

– + = curve decreased 5 degrees or more; – = increased 5 degrees or more; 0 = no change in curve.
—The above data on growth shows that most patients had rapid skeletal growth while in the orthosis. Patients 5 and 11 who had negative results had the least amount of gain in height recorded. The number of patients is too small to obtain statistically significant values; however, 50% of the patients who grew in height in a range of 11 to 19.5 cms. during the period followed had decreases of 5 degrees or greater in their spinal curves.

curve is 30 degrees or less. To date, no orthotic or electrical spinal stimulation has demonstrated significant decrease in scoliosis in the growing children on whom these methods were used.*

The etiology of idiopathic scoliosis remains an enigma and, furthermore, we do not know which cases will progress and which will not. Some current evidence suggests that scoliosis is a defect of the postural righting mechanisms under central nervous system control. If so, methods to bring improved postural control by feedback mechanisms seem logical if the spinal curve can be treated early enough. This poses a problem, however, because of the large number of minor curvatures 5-20 degrees that may not progress at all.

Application of orthotics to all such minor curves would result in massive over treatment and illusionary good results.* We hope that our research program on the postural mechanisms in scoliosis will define at an early age those children whose curves less than 20 degrees are likely to progress.

ACKNOWLEDGEMENTS

Much of this research was made possible through the National Institution of Handicapped Research (NIHR), grant number G997995817, under the Department of Education and the Orthotic Department of the Rehabilitation Engineering Center, Children's Hospital at Stanford.

The contributions of Mr. Bill Hastings and Ms. Leslie Roberts in their fabrication of the prototype I electronic system and the design of the prototype II electronic component, and the fine illustrations by Mr. Bayard H. Colyear, III, are greatly appreciated.

The project director would like to thank Ms. Carrie Beets for her assistance in gathering data for this preliminary report.

Last, but not least, we would like to thank all orthotic personnel and, especially, the patients.

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Clinical Experiences with Triple Amputees

Timothy B. Staats, M.A., C.P.

INTRODUCTION

This paper explores the possibilities of rehabilitation for patients with multiple amputations. Three cases are presented involving bilateral above knee amputations with unilateral above elbow amputations. The cooperative effort of the patients who participated in this project demonstrates that the avenue to rehabilitation may be difficult, but that anything is possible with perseverance and desire.

The difficult task of providing prosthetic care for the bilateral above knee amputee is seldom underestimated. The additional insult of an above elbow amputation enhances the problems and ultimate abilities for the return of the patient to independent living. The entire rehabilitative process must be carefully evaluated and planned if hope for success is anticipated. The patient must be highly motivated. It is important not to have preconceived ideas of what the abilities of individual patients may be in spite of the extent or number of amputations.

An aggressive positive approach to the prosthetic fitting and component selection may be crucial to the success of the triple amputee. Clinical experience has shown that no physical experience the amputee goes through is comparable to actual walking on full length prostheses. To spend time and energy working on a less than satisfactory end such as "stubbies," results in and leads to only more energy expenditure and emotional stress of the patient when the next echelon prostheses are fitted. It seems to be far more successful to give the greater challenge to the patient from the beginning and pull back to more conservative choices if, after a reasonable period of trying, the patient cannot achieve the primary goal of walking with full length prostheses. By coupling the initial strong efforts of the patient with prosthetic components that offer the best chance for function and progress the patient can know that everything that can be done is being done. It must never be "too much work or effort." Extra assistance from physical therapists, more than one prosthetist or trained assistants may be necessary to cast, fit, stand and walk the patient and to protect him during all stages of walking.

Rising to a standing position and sitting from a standing position position may require more "feel" than understanding for the patient. It is most helpful when the patient can see the tasks being performed. Possibly the photo sequences in this paper may be of some assistance to that end. When standing or sitting many functions must be performed without error. Rather than dwelling on instructions of how to do tasks it may be better to gather around the patient, protect him from falling and to fully assist the patient in the tasks until he establishes the coordination and strength for the maneuvers.

A fine balance of aggressive training with an acute sensitivity to the physical and mental endurance of the patient is essential. Do not accept negative attitudes. Realize however, that each appointment is extremely stressful to the multiple amputee. The first time the patient is assisted to the standing position he is filled with anticipation, fear and some pain. Concentrate on solving one or two problems with each standing effort. Try to direct the patient to concentrate with you on the problem solving effort. It is important to direct the fitting process, particularly when the patient may stand for only a few minutes. Determine, quickly and accurately, what must be done to improve fit and alignment. Then focus on the patient to question him about the comfort and kinesthetic or positional feelings he is experiencing. If the patient can remember and recall the balanced feeling experience when standing, it can be practiced mentally and improve his performance. Do not overwork the patient during the early fitting stages; thirty to forty-five minutes will be exhausting for most patients at first. Dizzyness and heavy perspiring are not uncommon reactions when the patient stands for the first time.

When assisting the bilateral above knee/ unilateral above elbow from the seated position to the standing position two or even three people may be needed to assist. Care must be taken to keep the patient from sliding away from the chair or from allowing the knee mechanisms to flex. As the patient sits the same assistance must be intensified as loss of balance will quickly increase falling velocity and force when balance control is lost. As the patient gains strength and coordination, assistance is only gradually reduced. Generally, even experienced patients are protected while trials are being performed between parallel bars.

CASE PRESENTATION NO. 1: J.B.

The following case studies present a number of prosthetic and physical techniques that have been utilized to allow the triple amputee to sit, stand, walk, fall and rise from a fall.

Patient J. B. is a congenital amputee (Fig. 1) with a right hip disarticulation, left proximal femoral focal deficiency, a right above elbow amputation and pincer deformity of the left upper extremity. He was fitted bilaterally with Kolman safety knees (Fig. 2) with an essentially standard left

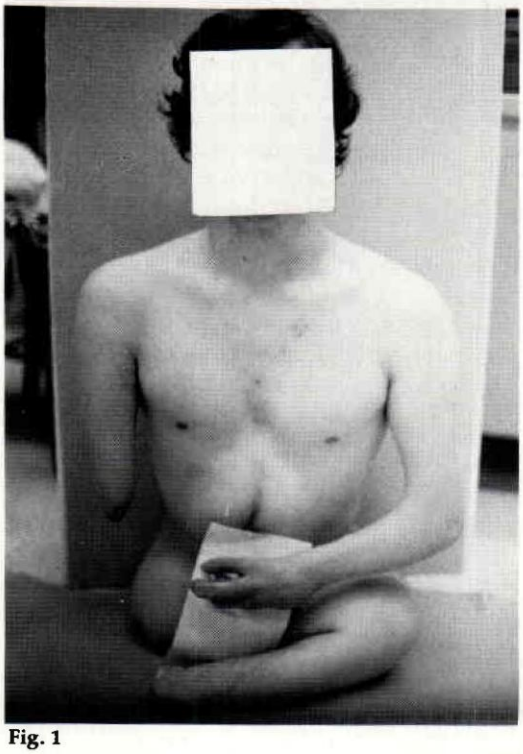

Fig. 1

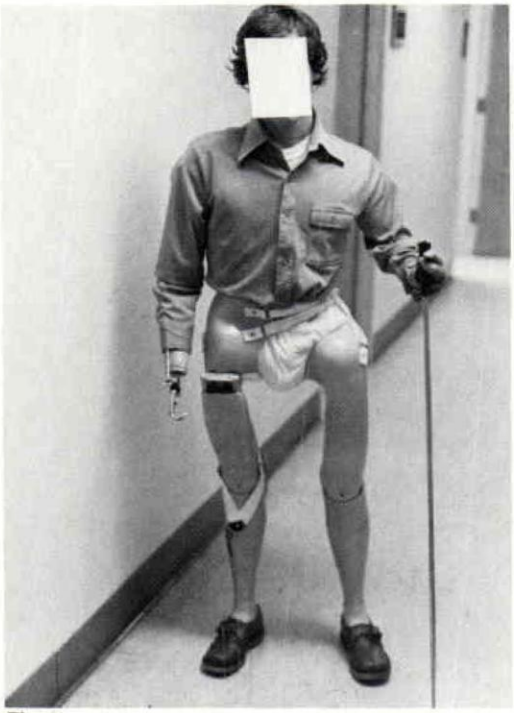

Fig. 2

hip disarticulation prosthesis, a non-standard right above knee prosthesis, both lower extremity prostheses used S.A.C.H. feet. His right above elbow is fitted with a standard above elbow prosthesis. The lower extremity prostheses (Figs. 3 and 4) required auxilliary suspension systems consisting of light weight velcro closures. These straps aid suspension of the prostheses and prevent rotation during the swing phase of the gait cycle. The hip disarticulation prosthesis required a flexible belt rather than a rigid structure so as not to interfere with the P.F.F.D. above knee side. Hip extension assist straps are seen in figure 5, to prevent excessive hip flexion on the hip disarticulation prosthesis.

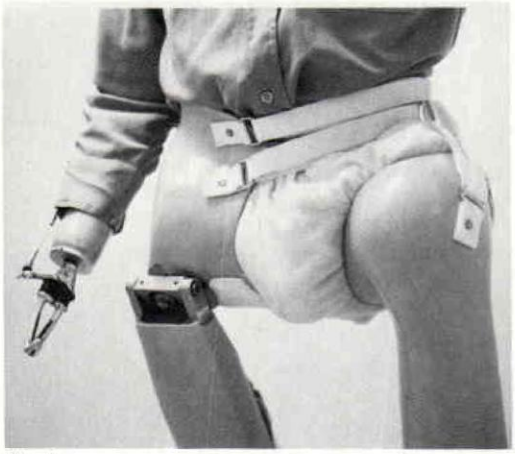

Fig. 3

Rising From A Seated Position

To rise from the seated position the subject necessarily utilized his above elbow amputation prosthesis locked in extension to act as the primary power to raise himself up. His "sound arm" elbow did not possess enough strength to assist. As weight was born on to the Kolman locking knees the weight actuated the locking mechanisms thereby affording stability and safety in standing. Forward steps are initiated (Fig. 6) by rocking to the hip disarticulation side and extending the right P.F.F.D. side forward with a flipping motion. The hip disarticulation forward step (Fig. 7) follows with a transfer of weight to the P.F.F.D. side and noticeable lateral trunk bending

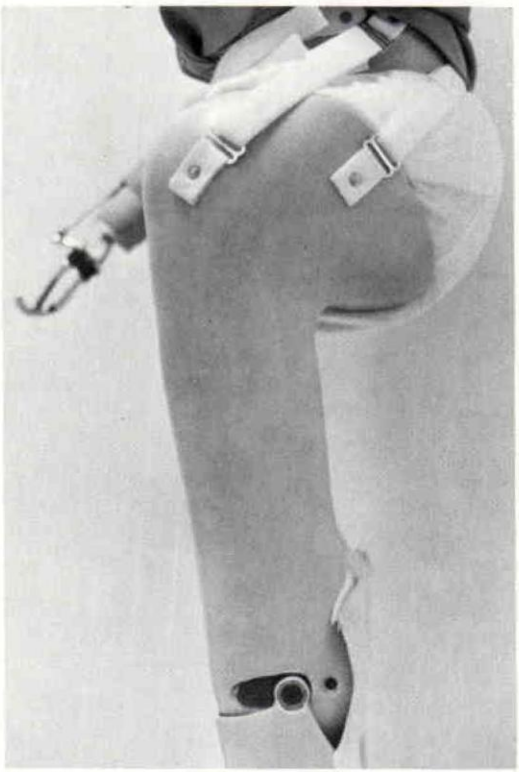

Fig. 4

Fig. 5

Fig. 6

Fig. 7

Fig. 8

to that side. To sit J. B. rocks forward over the toes of the prosthesis while reaching for the chair with the locked and extended above elbow prosthesis. He then rolls around that arm until locating on the chair seat. Because the knee units are weight actuated locks, the prostheses will not unlock until he is fully seated. J. B. is shown in street clothes (Fig. 8) on his departure from the laboratory. Acceptable cosmesis was achieved. With the extent of his disabilities the wide base gait was not considered inappropriate.

CASE HISTORY NO. 2: R.H.

R. H. is a recent bilateral above knee and unilateral left above elbow amputee as a result of a power line accident. Figure 9 shows the socket designs that were fitted to this patient. Notice that the medial brims are less than ½'' wide and are carefully smoothed and rolled. The posterior medial corners of the sockets have been reduced and rounded to prevent pinching should the sockets be applied in external rotation or a misstep be taken.

R. H. was fitted with Hydra-Cadence hydraulic knee mechanisms with quadrilateral total contact plastic suction sockets. The plaid pattern was chosen by the patient for a final lamination finish. The polyester plaid material was laminated using clear acrylic resin with one or two layers of white nylon stockinette as backing.

Fig. 9

To rise from the seated position R. H. places his hand at the back of a supported chair (Fig. 10). The prostheses are placed about 18" apart. As he extends his arm he simultaneously forces his knees into extension (Fig. 11). He then pushes himself into a balanced standing position. The Hydra-Cadence knee units provide plantar flexion in the ankles allowing the knee centers to be easily maintained in a stable position during standing. As a balance point is reached (Fig. 12) R. H. takes a small forward step with the left prosthesis to bring both feet together. R. H. uses a cane for added support particularly when walking down inclines and rough surfaces. R. H. is presently the owner-manager of a machine shop working full time wearing his prostheses. For very active chores in the shop and at home "stubbies" were fabricated after the walking techniques on the full length Hydra-Cadence prostheses had been mastered. Figure 13 shows that the basic alignment of the stubbies can be duplicated from the full length prostheses. Wood blocks (Fig. 14) were laminated to the socket and ground to shape and alignment. Vibram soles from hiking boots were epoxied and nailed to the distal ends to prolong wear.

CASE HISTORY NO. 3: K.K.

The third patient K. K. is a bilateral above knee and unilateral left above elbow amputee. He is presently wearing bilateral quadrilateral total contact sockets, one suction, one partial suction with a light silesin belt. He is fitted with the Hydra-Cadence knee mechanisms. His basic alignment is no different than that of a unilateral above knee amputee (Fig. 15). A plumb line dropped from the ischial tuberosity falls with the foot slightly outset. The base of his gate measured between the heels is approximately 3½'' to 4''. Figure 16 illustrates the A-P foot position is essentially normal with the plumb bob position falling about 1'' anterior to the breast of the heel.

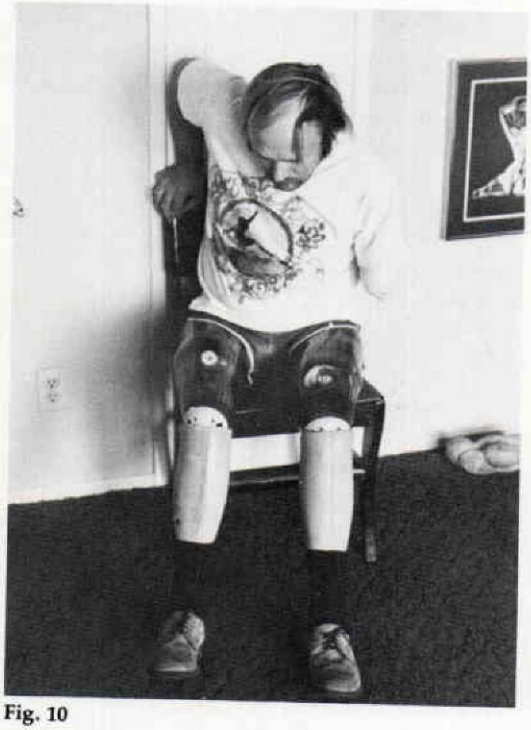

Fig. 10

Fig. 11

Fig. 12

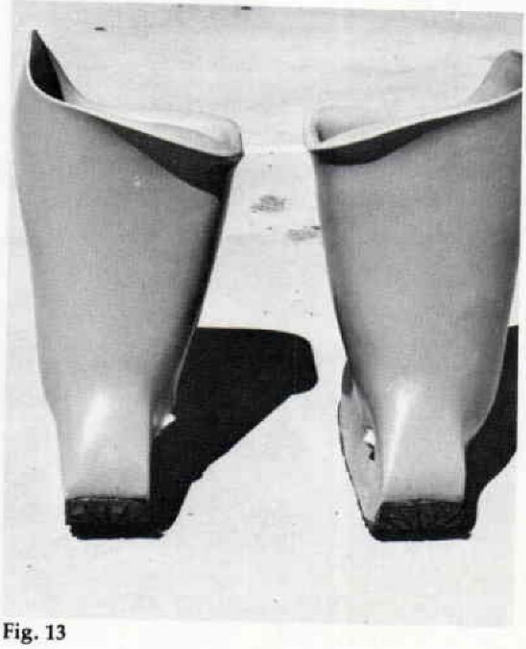

Fig. 13

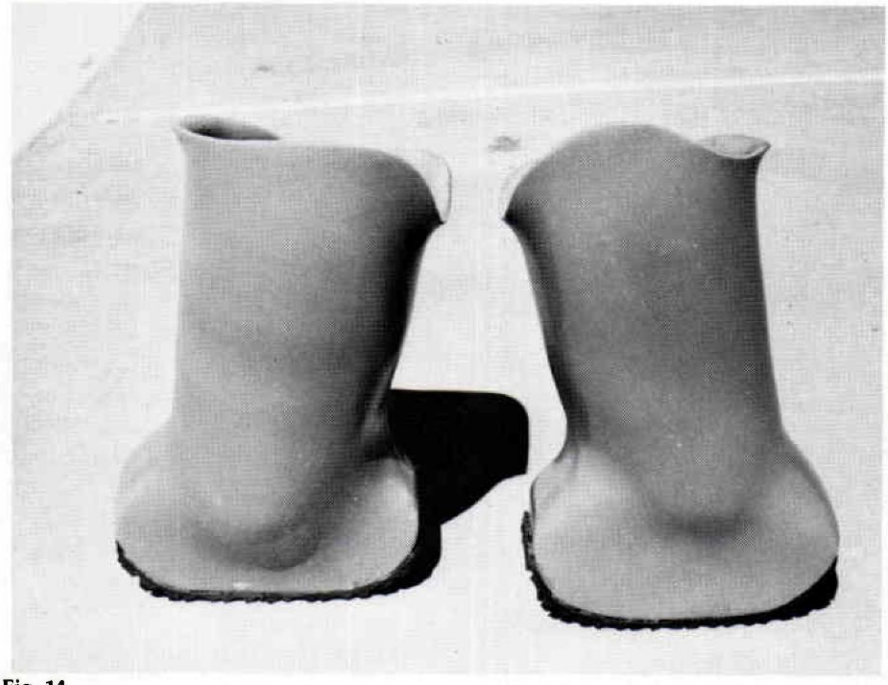

Fig. 14

Fig. 15

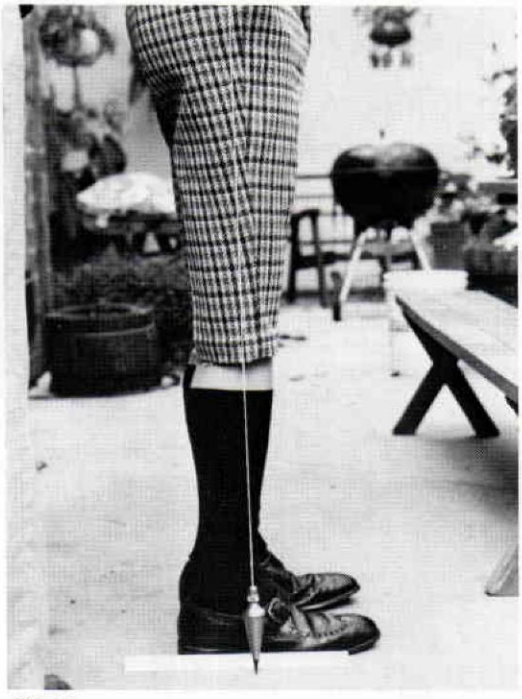

Fig. 16

Sitting Technique

Figure 17 begins a sequence of photographs of a sitting technique called the "roll and sit" technique. The patient approaches the chair face forward and places a hand on the side of the chair that it will be on when he is seated. Notice that the Hydra-Cadence units are in full extension and that the right leg is slightly forward of the left leg. In Figure 18 K. K. swings the left leg behind him crossing the right side and thereby aligning his feet in front of the chair. This is done while rolling to the lateral side of the right foot. Both knees are still extended. In Figure 19 the weight of his body transfers to the contact arm and is counter balanced by the still extended right prosthesis. The left prosthesis begins to flex, offering some guidance and positioning control by the fact that the unit's foot is on it's medial border. By extending the left residual limb against the force of the bending knee some support is afforded. This support is rather instantaneous. By pivoting his arm K. K. can lower himself safely to the seated position in Figure 20. Exceptional arm strength is required to perform this sitting sequence in a controlled manner.

A second sitting technique is used more often by bilateral above knee amputees.

Fig. 18

Fig. 17

Fig. 19

The patient positions himself in front of the chair but facing away from the chair. He is turned so that his good arm is closer to the chair. K.K. suggests pointing the hand at the chair and carefully judging its position. As he puts it "you only have to miss the chair once to understand why it is important to know its location." The hand is placed firmly on the chair (Fig. 21). By leaning slightly backwards the stability of this system now has three points.

Notice that the above elbow amputation aids in counterbalancing the weight of the body as it begins to lead towards the chair (Fig. 22). The right Hydra-Cadence unit is carefully flexed, leaving the full body

Fig. 20

Fig. 21

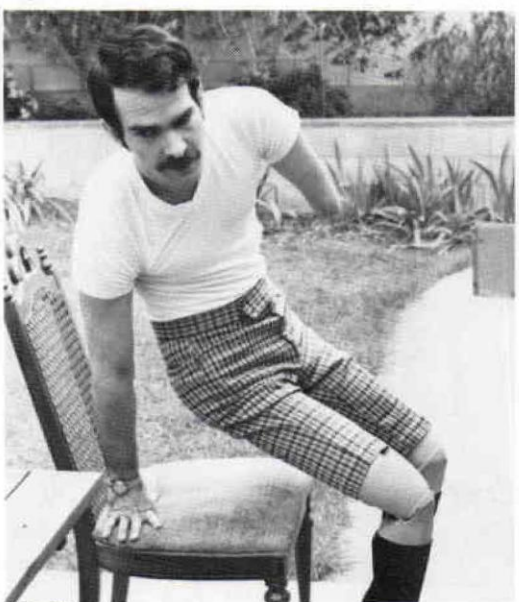

Fig. 22

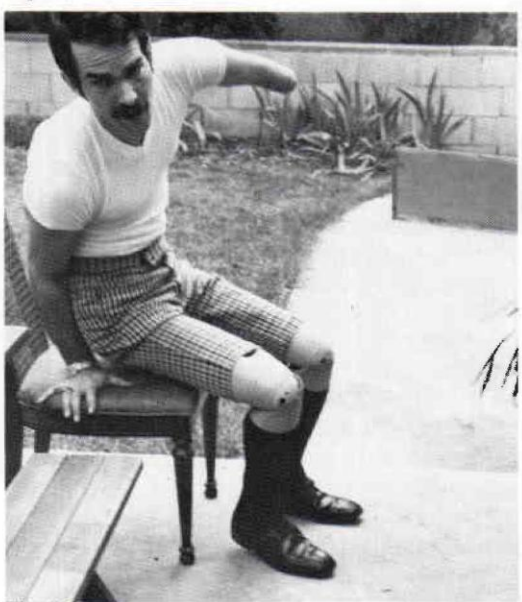

Fig. 23

weight on the left and the arm. When all weight is shifted to the arm and the right knee is flexed, the body system will quickly settle to the chair surface. A majority of body weight shifts to the arm. Notice the distance the torso is from the arm. Arm position and grip is crucial as the body descends. At this point neither Hydra-Cadence can aid in guiding the body to the chair. Any attempt to do so might shift the body weight forward thus throwing the subject to the floor. The buttocks swing to contact the arm and then slides down the wrist until contact is made with the chair seat (Fig. 23).

Rising From Seated Position

Careful positioning of the body is important to rise from the seated position in a chair that is not backed up against a solid wall or object. Force must be directed down through the chair rather than in a manner that would cause the chair to slide.

To rise from the chair, the hand is positioned directly behind the buttocks (Fig. 24). Once the body is lifted it is crucial that the position be such that an aborted standing to sitting can be safely completed. By extending the hips (Fig. 25) at the same time as the body is lifted, the knee units can be locked into extension, first the left side (Fig. 26) followed by the right side (Fig. 27). When the left side knee unit is locked into extension the safety of this standing sequence is greatly enhanced. When both units are in full extension the fingers are extended to push the torso forward balancing the body over the prostheses. K. K. now stands erect (Fig. 28) and takes a short step forward to counteract against the momentum generated during standing.

Controlled Falling

Falling to the ground can be a most devastating mishap to the bilateral above knee patient. For the triple amputee a fall presents even more of an intensified problem. It is important that the patient learn how to fall and how to rise to the standing position.

Fig. 24

Fig. 25

Fig. 26

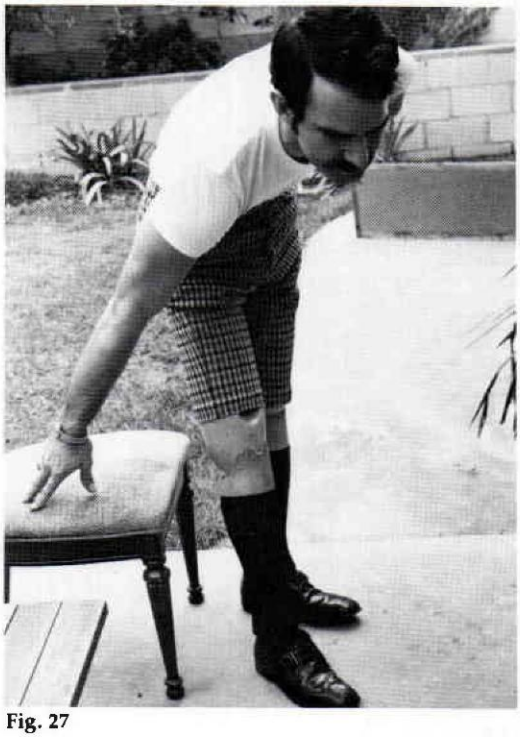

Fig. 27

Fig. 28

A demonstration of controlled falling begins by placing the right foot in front of the left (Fig. 29). As he begins to topple to the ground he bends forward at the hips to reach for his landing position with the sound arm (Fig. 30). The left knee unit is popped into flexion. By falling to the side the chances of being hurt are lessened as are the possibilities of damaging the prostheses. As the fall continues (Fig. 31) and downward velocity increases K. K. rolls around backwards so that the shock of the fall is absorbed by the buttocks and slowed appreciably by the shock absorbing action of the arm catching the fall and gradually bending at the elbow in a controlled manner (Fig. 32). It may not always be possible to control a fall. If a fall begins and any control is available at all, an attempt to twist or roll to the side should be made rather than straight down or backwards falling.

Rising From The Ground

Once on the ground or the floor, a far greater problem is how to rise to a stand-

Fig. 29

Fig. 30

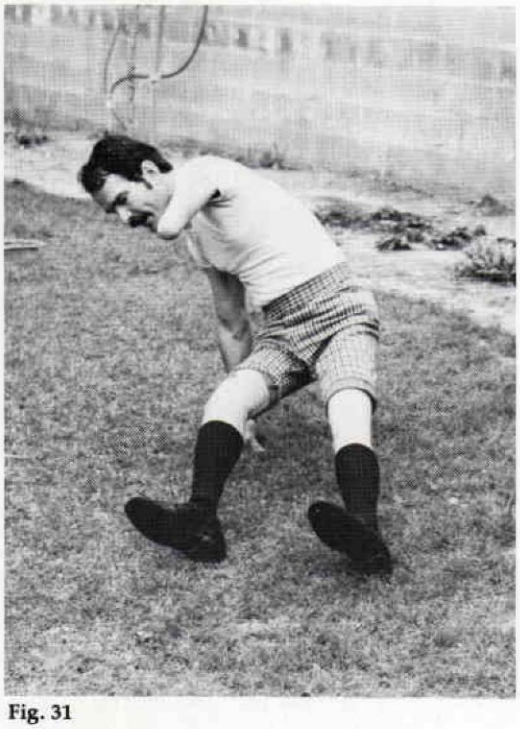

Fig. 31

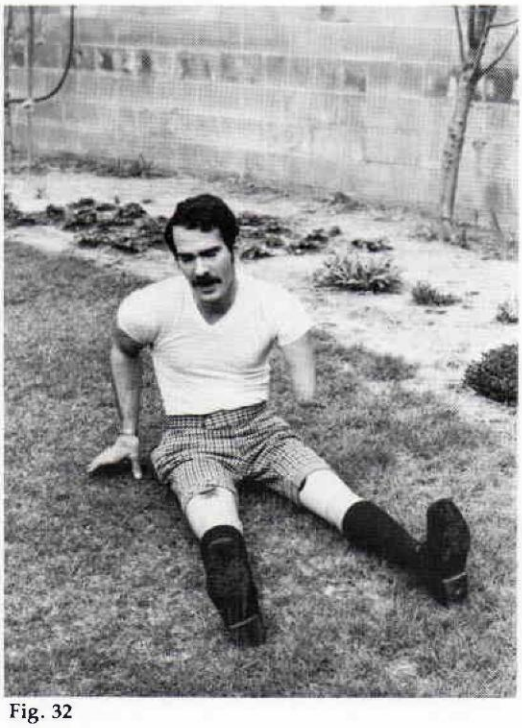

Fig. 32

Fig. 33

ing position unaided. The starting position is taken by spreading the prostheses about three feet apart. The arm is held next to the body and the elbow is bent (Fig. 33). The torso raised slightly off the ground. The pelvis is rolled to the side so that the Hydra-Cadence units are on their sides and are stable regardless of whether or not they are extended. The pelvis is now raised 1½ to 2 feet off the ground.

Fig. 34

By using a rocking, hoisting, bouncing, body english motion, K. K. hops his hand 10 to 12 inches closer to the feet position (Fig. 34). this motion is repeated and the torso is now flexed fully at the hips. By pushing forcefully but in a controlled manner the balance position over the center of gravity is achieved (Fig. 35). K. K. then returns to the upright standing position (Fig. 36). In this short sequence of photo-

Fig. 35

Fig. 36

Fig. 37

Fig. 38

Fig. 39

graphs it is easy to appreciate the agility, coordination and practice it takes to develop this technique.

An alternate technique to rise from the ground is also available to the triple amputee. First, roll on to the stomach (Fig. 37). By performing a "one-arm push-up" while forcing the body weight down through the prostheses the entire body can be raised without the knees buckling. K. K. leans forward then backward raising his buttocks higher in the air and hops his hand to a new position (Fig. 38). The process may be repeated one more time so that he is able to rock into the standing position (Figs. 39 and 40).

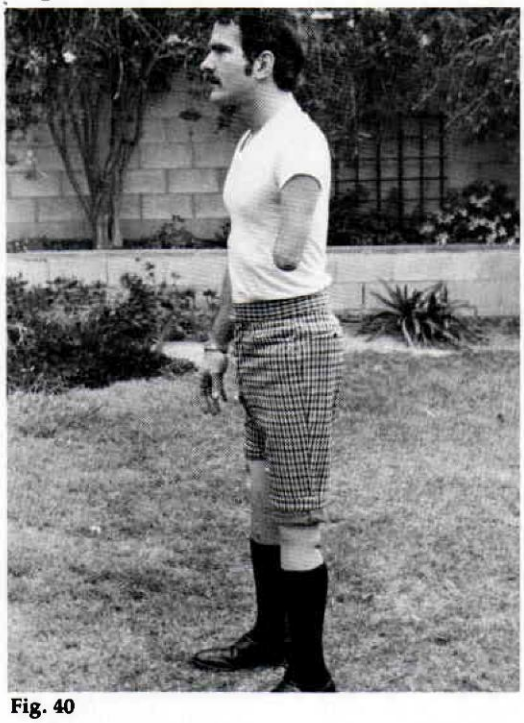

Fig. 40

Ascending and Descending Steps

Step climbing can be accomplished by the bilateral above knee amputee, unilateral above elbow only with great difficulty.

Begin by placing one foot on the step with the front of the heel catching the leading edge of the step (Fig. 41) by grasping the door jamb and pulling the body weight forward (Fig. 42) he is able to climb

the step to the position shown in Figure 43. A similar procedure is used to descend a single step (Fig. 44). K. K. grasps the door and lowers his left extended prosthesis to the lower level.

While it is not usually recommended for the bilateral above knee to descend stairs "step over step" K. K. has perfected a technique for descending at least curb height obstacles (Figs. 45 and 46). This sort of maneuver must only be attempted when tremendous confidence, not only in one's own physical abilities, but also when the knowledge and understanding of the functional capabilities of the knee mechanisms has been mastered.

The choice in this case of the Hydra-Cadence knee units required careful consideration and discussion with the patient. There are advantages and disadvantages to every knee unit. The patient must accept the facts about certain aspects of his prosthesis so that an understanding of problems and situations can be handled judiciously.

While the Hydra-Cadence is somewhat

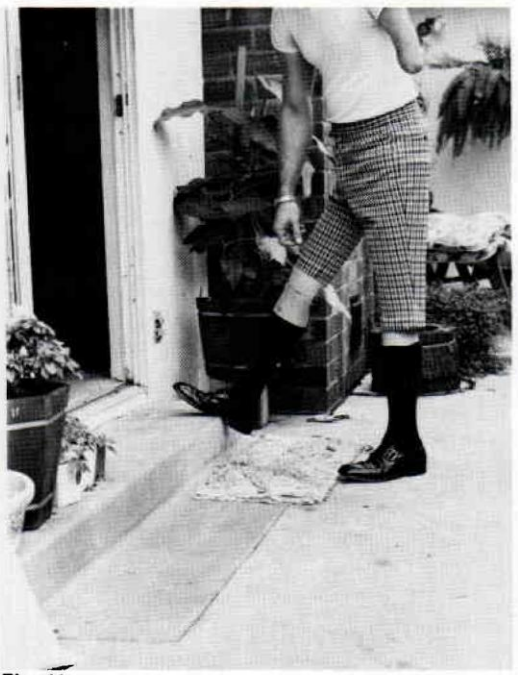

Fig. 41

Fig. 42

Fig. 43

Fig. 44

Fig. 45

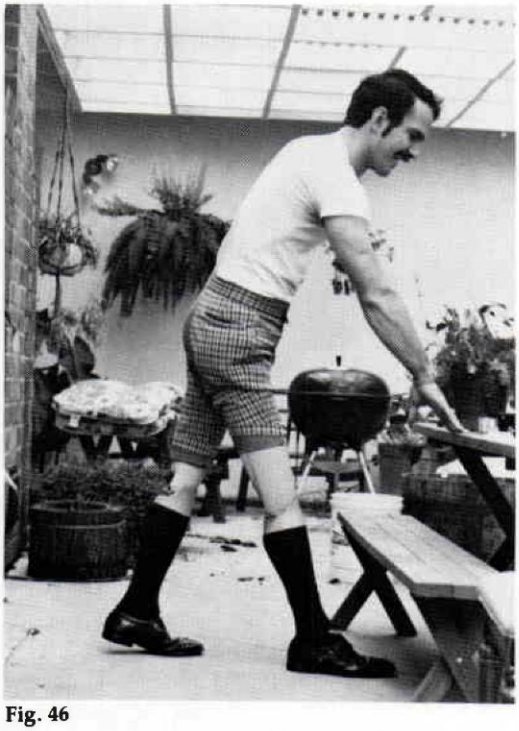

Fig. 46

heavier than other knee units, it is more sophisticated in the functional capabilities it provides to the patient. Many improvements have been made to the Hydra-Cadence knee mechanism but is a relatively complex system. As with any complicated system a certain amount of servicing may be necessary. The patient must be aware that servicing is normal. Most patients are willing to accept the trade-offs of weight, function, service and complexity because they are able to live more normally when well fitted.

ACKNOWLEDGEMENT

Special thanks is given to Barbara Brown for her assistance in the preparation of this manuscript.

Tim Staats, CP was formerly Techinical Director at U.S. Manufacturing Company. He currently is Director of Prosthetic and Orthotic Education and Research at the University of California at Los Angeles.

A Comparison of Pistoning Forces in Orthotic Knee Joints

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John Schmidt

INTRODUCTION

Motion of the knee joint is much more complex than flexion about a simple hinge or fixed axis of rotation. Varying amounts of displacement, internal-external rotation, and abduction-adduction may accompany flexion-extension[4 5]. Knee motion may be visualized as the rotation of the femur about a series of three-dimensional instantaneous axes rather than a single fixed axis [1 8 9]. Knee motion with proper joint stability is a result of the complex interaction of the ligamentous and capsular structure surrounding the joint, the geometry of the opposing articular surfaces, and the musculo-tendinous activity across the joint. The path or locus of the continuously moving instantaneous axis will vary with the activity being performed, the amount of inherent laxity in an individual's knee, variations in how individuals perform similar functional activities, and whether or not joint instability exists due to ligamentous or capsular insufficiency.

Motions intrinsic to orthotic knee joint designs are much less complex than corresponding natural knee motion. In most cases, orthotic flexion occurs about a fixed axis of rotation as allowed by simple hinges[3], or about a single pattern of axes (polycentric) prescribed by a variety of mechanisms such as the interaction of two cam surfaces, two geared surfaces, or on a roller in a slot design[6]. Note that the polycentric designs just mentioned are still more simplified than natural knee motion because such joint mechanisms provide one single locus of axes for all possible functional conditions, whereas the locus of instantaneous axes of the knee varies with different functional conditions.

An orthosis should hypothetically allow a full, unrestricted range of normal joint motion to occur except at appropriate limits of motion where orthotic constraints are intentionally introduced to provide extra stability required to compensate for soft tissue insufficiency. When an orthosis is applied to the knee, however, a conflict occurs as the orthosis attempts to force the knee to follow its simplified motions. Since this is prevented by natural joint structure (ligaments, muscles, articular

surfaces), unwanted constraint forces are also generated in the orthosis at suspension points causing the interface components to migrate or piston over the limb segments during the supposed unrestricted range of normal motion (usually flexion-extension). Note again that some orthotic constraints are beneficial when they compensate for lacking stability (such as medial-lateral force for correction of varus or valgus), yet those which arise from conflict of joint motions are detrimental when they occur during unaffected normal phases of knee motion. Clinical consequences of these unwanted, over-constraint pistoning forces are a possible obstructed range of normal motion, or discomfort due to skin irritation from increased pressure over bony prominences at suspension points.

Many orthotic designs[2], some of which have been mentioned above, have evolved in an attempt to minimize the constraint forces and migration effects. However, no objective evaluation has been undertaken to demonstrate the relative efficacy of these designs.

OBJECTIVE

The objective of this study was to develop an experimental method to quantitatively evaluate the relative efficiency of knee orthosis joint designs. The comparison was based upon the tendency of the orthotic joints to cause migration or pistoning during motion. The pistoning tendency was quantified by designing a transducer which measured the portion of the orthotic constraint force which was directed parallel to the sidebars which attach the joints to the orthotic interface components. The tests were performed on a human subject with normal knee laxity. The rationale for this is that since no soft tissue instability exists in the subject, any constraint forces that occur will reflect only the conflict between the simple orthotic and more complex anatomical knee motions, thus an indication of the migration tendency and efficacy of the orthotic knee joint designs.

METHODS AND MATERIALS

Interface components were fabricated for the human subject. Orthotic knee joint designs, including several commercially available types, as well as a design from our laboratory, could be interchangeably attached to the interface components. The pistoning force transducers, which were developed for this study, were installed on both the medial and lateral orthotic joint sidebars, just proximal to the joint. Pistoning constraint forces were then experimentally measured for each joint design as the subject performed a series of functional activities.

Figure 1—Commercially available orthotic joints tested: posterior offset (left), single axis (middle), polycentric with geared articulating surfaces (right).

Description of Orthotic Joints

The three commercially available joint designs tested were 1) the single axis hinge, 2) the posterior offset hinge, and 3) the polycentric hinge (FIg. 1). The polycentric design provides motion about a single path of axes as prescribed by two geared surfaces which are mechanically constrained[6]. The posterior offset design is based upon a single axis approximating the location of the sagittal radius of curvature of the posterior femoral conyles, which articulates with the tibia in flexion[3].

The orthotic knee joint designed in our laboratory can be referred to as the "optimal single axis" joint. "Optimal single axis" refers to the optimal location and orientation of a single axis of rotation for a range of motion such that flexion about

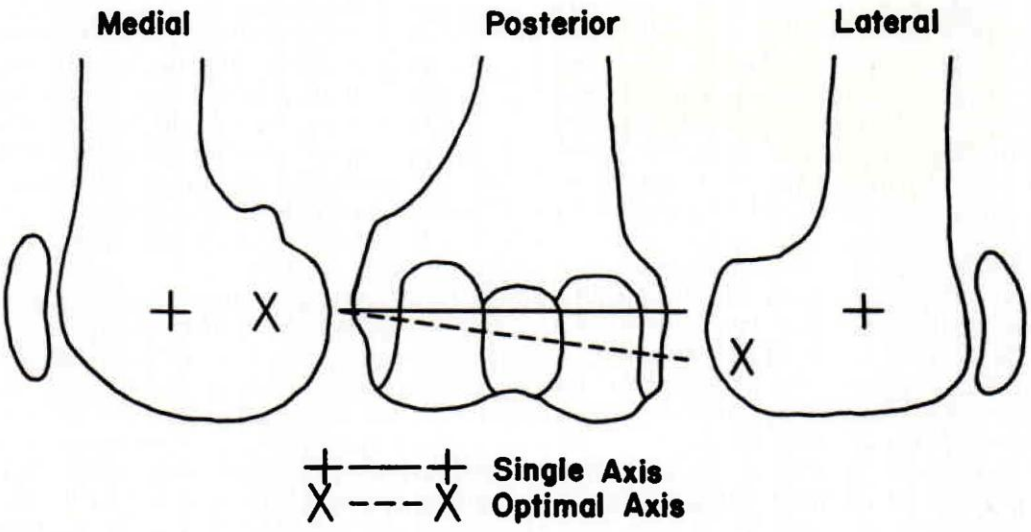


Figure 2—Position of single axis hinges defining the "optimal single axis of rotation" for extension to 90° flexion, were predicted to be posterior and distal to the traditional placement of single axis hinges.

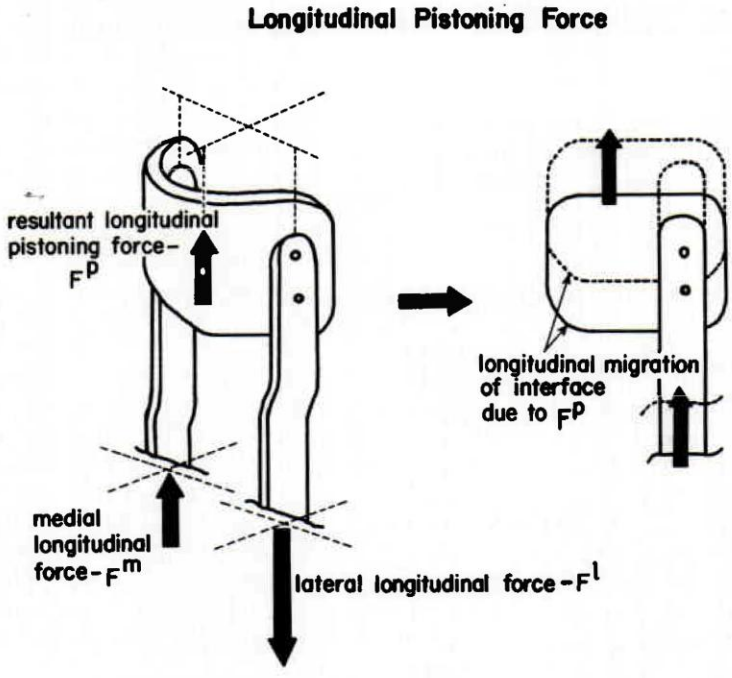


Figure 3—Resultant longitudinal pistoning force (left) is the vector sum of the medial and lateral sidebar forces measured by the pistoning transducers (either tension or compression), and causes the pistoning or migration of the interface (right).

this single axis most closely approximates the equivalent complex anatomical motion. Thus, this "new" design is actually a method which predicts the optimal placement of single axis hinges. The formulation of this optimal single axis is based upon a mathematical model of the knee[7], and can be conveniently custom incorporated into the fabrication technique for individual orthotic devices for particular ranges of motion, if desired. Details of the optimal single axis procedures will not be described in this report since it is not the major objective of this study. The average of this data will be used to locate the optimal single axis on the human subject in this study (Fig. 2). When compared with the traditional placement of single axis hinges on an orthotic interface (bisect) distance between adductor tubercle and medial joint space; bisect distance between patella and popliteal region) the average optimal single axis for the six subjects was found to be

medial side of knee—21 millimeters posterior
0 millimeters proximal-distal
lateral side of knee—26 millimeters posterior
9 millimeters distal

This assymmetrical axis was located in the study by contouring single axis hinges at the appropriate locations just described.

Description of Pistoning Force Transducer

Migration or pistoning clinically manifests itself as the longitudinal "riding up and down" of the orthotic components over the lower limb segments. An orthotic constraint force is potentially three-dimensional in nature; that is, it can have components in the longitudinal (superior-inferior), anterior-posterior, and medial-lateral directions (Fig. 3). Constraint forces in the anterior-posterior and medial-lateral directions, along with the movements which they create (sagittal, transverse, and torsional), are predominantly present at times when the orthosis provides useful compensation for soft tissure instability. Constraint forces in the longitudinal direction, on the other hand, primarily lead to the unwanted pistoning of interface components during normal phases of motion. Constraints in the other directions are probably present but insignificant in this case. Measurement of longitudinal sidebar constraint forces, and their subsequent vector addition (taking into account whether they act in tension or compression), quantifies the resultant pistoning constraint force acting at the orthotic interface (Fig. 3), which gives rise to component migration.

A transducer was designed (Fig. 4) which 1) measures the longitudinal pistoning constraint force in each sidebar (and differentiates between tension and compression), 2) is insensitive to moments created by extraneous components of the constraint force, 3) is unobtrusive during functional activities, and 4) can be interchangeably attached to orthotic sidebars.

The lightweight aluminum transducers (Fig 4) were approximately 3½ inches long, with a reduced cross section in the middle portion, and two recesses (one on each end) used to securely clamp the transducer to the orthotic sidebars. Four semiconductor stain gauges were installed on each transducer, two in the transverse plane and two in the sagittal plane of the reduced cross section. The gauges were placed at the bottom of the notches (Fig. 4 and 5) so that they would be as close as possible to the neutral axis in each cross section, thus minimizing the effects of orthotic constraint components other than the longitudianal pistoning force (torsional, sagittal and transverse bending moments, if present at all).

To further cancel out these extraneous effects and to ensure that migration comparisons are based solely upon the longitudinal portion of the pistoning constraint, one strain gauge from each cross section was placed in series, and formed opposite arms of the Wheatstone bridge circuit (Fig. 6), amplifying the longitudinal strain approximately four times, and cancelling out the response due to out-of-plane movements.

The lead wires from the strain gauges (and the cable containing these wires) were stress relieved (Fig. 4 and 6) in sev-

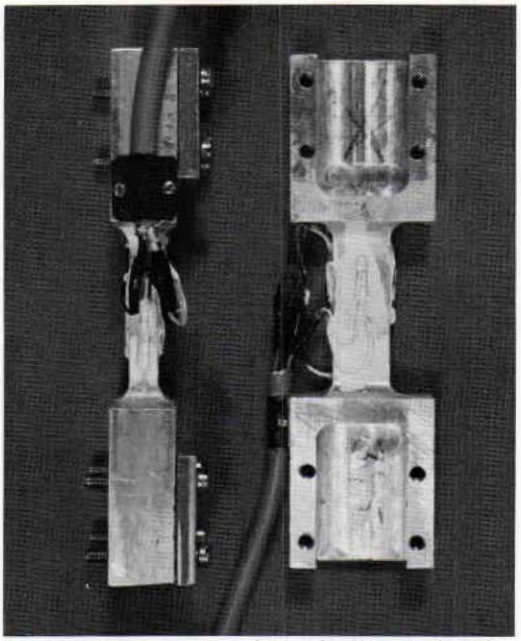


Figure 4—Two views of an aluminum pistoning force transducer-semiconductor strain gauges are located in mid-section notches, and leadwires from gauges are collected in a stress-relieved cable.

eral places, so that tension on the cable during testing would not harm the gauges. The gauges were also protected and waterproofed by several layers of gauge coat (Fig. 4). The strain gauges and Wheatstone bridge were connected to a Beckman strip chart recorder.

The pistoning transducers could be installed on both medial and lateral sidebars, could be easily interchanged among the joint designs tested, and their presence on the evaluation orthosis was unobtrusive and did not inhibit normal knee motion in any way. The transducers were installed as follows (Fig. 6). A 1½ inch section of each sidebar was removed just proximal to the joint mechanism. The recesses on both ends of the transducer (Fig 4) were designed to securely accommodate the remaining ends of the sidebar, so that the transducer takes the place of the sidebar section removed (Fig. 7). The neutral axes of the reduced transducer cross section were in direct line with the neutral axes of the orthotic sidebars, thus preventing moments inherent in the system. The top of the clamps were screwed down so that

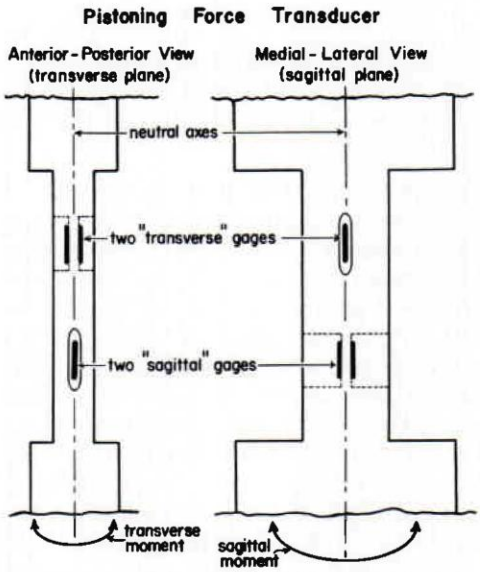


Figure 5—Sketch indicating placement of the four strain gauges in notches on either side of the neutral axes, minimizing the effect of transverse, sagittal, and torsional moments.

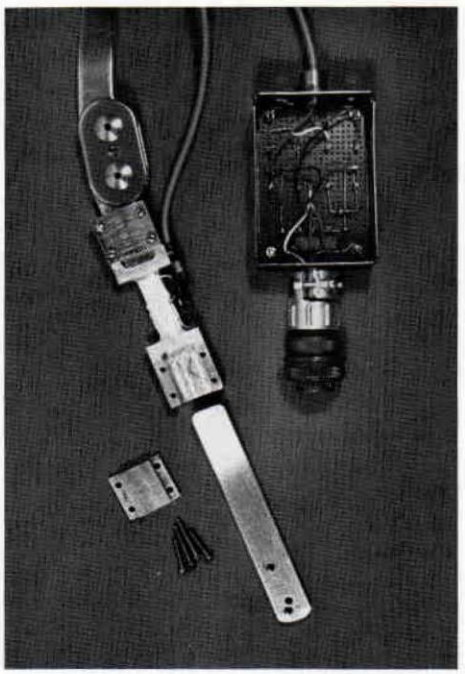


Figure 6—Transducers can be interchangeably incorporated into sidebars, proximal to hinge mechanisms. A section of each sidebar was removed. The recess and clamp at both ends of the transducer, securely accommodate the sidebar geometry so that the transducer takes the place of the removed sidebar section. Note the Wheatstone bridge circuitry at the bottom left.

the sidebars did not move within the recesses. Each sidebar was prepared in an equivalent manner, so that the transducers could be interchangeable among the joint designs.

The pistoning force transducers were directly force calibrated before and after each test procedure, so that the voltage output of the strain gauges could be related to an equivalent amount of force (newtons/millivolt). The calibrations were performed by hanging a series of weights from a calibration bar secured to one end of the transducer. During the test procedure, a zero force baseline was determined, so that tension or compression in the transducers indicates whether the longitudinal sidebar forces were directed distally or proximally. A series of bench tests were also performed prior to testing, to demonstrate that the pistoning transducers do cancel out extraneous constraint forces and movements, so that pistoning comparisons will be valid.

EXPERIMENTAL PROCEDURE

The evaluation orthosis (Fig. 7) was fabricated as follows. A plaster impression was taken of the right lower limb of one human subject. The positive plaster impression was modified, with emphasis given to the parallel buildups on both sides of the knee to ensure that orthotic joints were parallel. Sidebars containing the orthotic joint designs were contoured to the positive plaster impression, so that hinges were located as previously described. Proximal and distal orthotic interface components were fabricated by vacuum-forming high density polyethylene over the positive plaster impression. The interface components were adequately suspended proximally in the medial femoral area, distally at the medial tibial flair, and circumferentially in the thigh and calf region by broad straps composed of gum rubber with a leather backing. The interface components were capable of accepting interchangeable joint designs with accompanying sidebars.

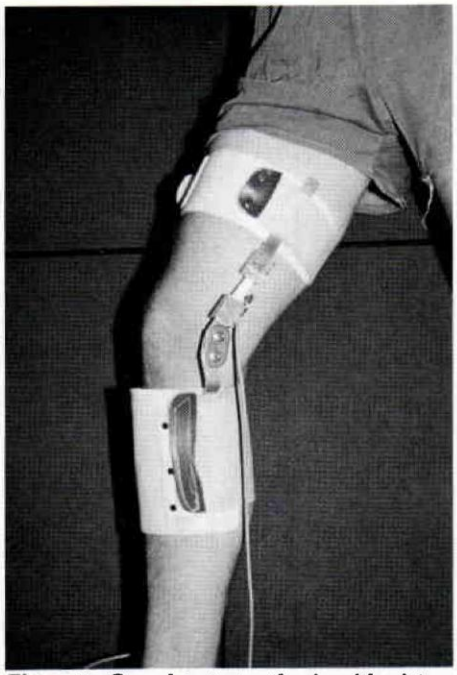


Figure 7—Complete test orthosis with pistoning transducer incorporated into polycentric hinge sidebar.

An initial force calibration of the pistoning transducers was performed, and then the transducers were installed on the medial and lateral sidebars of one of the orthotic joint designs (Fig. 6). The joints and sidebars were attached to the appropriate locations on the interface components, and the evaluation orthosis was applied to the knee of the subject (Fig. 7). The medial and lateral longitudinal pistoning constraint forces were recorded during the following activities (refer to Fig. 8A-E):

(1) Unloaded Flexion—non-weight bearing flexion-extension (and hyperextension), with forces recorded at flexion angles indicated on Fig. 8a;
(2) Loaded Flexion—repeating the flexion-extention range in (1), with the addition of the weight bearing condition (partial deep knee bends);
(3) In/Out of Chair—beginnng in a standing position, the subject sat in a chair, then got out of the chair—this activity was repeated twice (Fig. 8D);
(4) Level Walking—forces were measured from heel strike to heel strike of the limb containing the evaluation or-

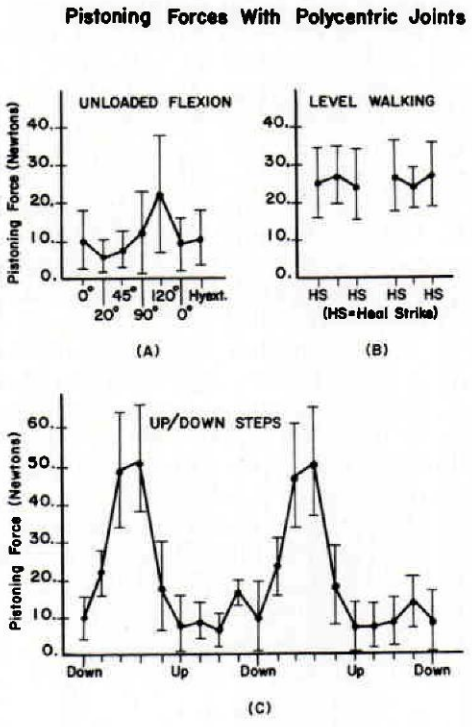
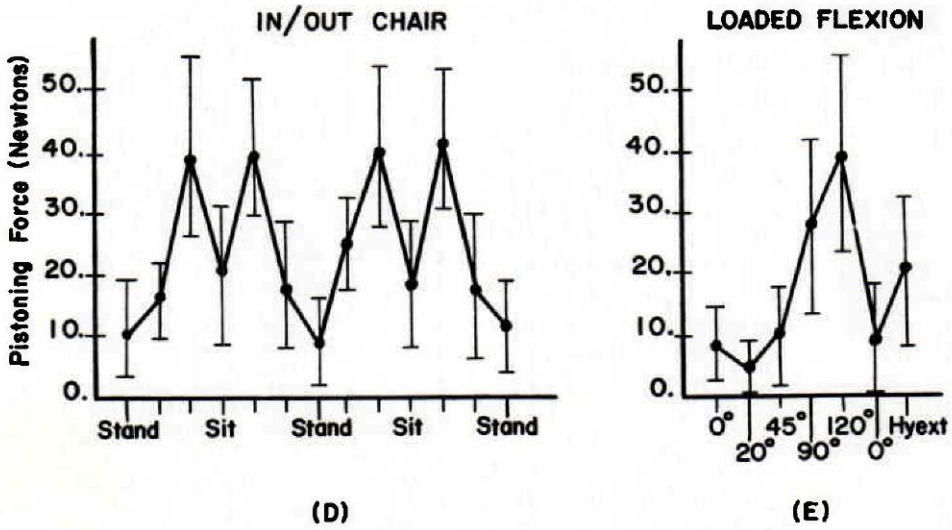


Figure 8—Mean and standard deviation (newtons) of absolute magnitude of resultant pistoning forces for 15 repeated applications of the test orthosis to the subject's lower limb during the following functional activities: (A) unload flexion, (B) level walking, (C) up/down steps, (D) in/out chair, (E) loaded flexion.

thosis, during two consecutive gait cycles (note that this motion contains both stance and swing phases of gait);

(5) Up/Down Steps—beginning in a standing position on the floor, the subject stepped up on a 16 inch height, then returned to the floor, leading with the evaluation orthosis limb in both cases—this maneuver was repeated twice (Fig. 8C).

Note that flexion angles in (1) and (2) above were measure by a simple handheld goniometer. The evaluation orthosis was then removed from the subject's limb, and place on a laboratory bench so that a zero force baseline could be recorded for the isolated, unloaded state of the transducer-sidebar complex. After several minutes, the evaluation orthosis was reapplied to the limb of the subject, and the above test sequence repeated. Pistoning constraint force data was recorded for a total of fifteen orthosis applications for each joint design, the purpose being to estimate the portion of the constraint force due to the variations in interface component fit among repeated applications of the orthosis by the same subject. Once known, this error can be averaged out, leaving only the pistoning constraint force due to the



particular joint design. After testing of one of the four joint designs, both transducers were again removed and force calibrated, with the calibration constants (newtons/millivolt) being taken as the mean of the individual pre- and post- test calibration trials. The entire test procedure above was repeated for each of the orthotic joint designs; polycentric hinges, and single optimal axis hinges.

RESULTS

A typical set of data is demonstrated in Fig. 8 summarizing the pistoning constraint forces for a particular joint design, this one being for the polycentric hinge. The standard deviations in the figure indicate the scatter in forces due to the reapplication of the evaluation orthosis fifteen times, as previously described. The force values in Fig. 8 are all positive as they represent the absolute value of the magnitude of the result longitudinal pistoning force.

Note that with unloaded and loaded flexion in Fig. 8A, E, force levels were recorded at particular flexion angles. In the remaining three functional activities (Fig. 8B, C, D), the observer recorded only the "end points" of the activity, such as "stand-sit" for getting in and out of a chair. For a particular activity, however, the resultant pistoning outputs for all four joint designs were equivalently shaped, indicating that certain physiologic conditions repeatedly occurred during the activity, such as the shift in body weight or the tension in muscular groups. Since no electromyographic or photographic record of the activities were obtained, the activities were not defined in more detail than the end points. When reducing the data, however, forces representing some of the obvious equivalent, un-named physiologic conditions were included. They are indicated by the force values other than the end points defined in Fig. 8B, C, D).

For the purpose of comparing the four orthotic knee joint designs, the data as typified in Fig. 8 was further reduced to the means and standard deviations of all the recorded resultant pistoning constraint forces for each activity, for each joint design, as presented in Figs. 9-13. For each of the functional activities, these mean resultant pistoning forces were normalized over the four joint designs, as indicated by the bar graphs at the top of Fig. 9-13. The tables at the bottom of Figs. 9-13 present the statistical significance of the difference between the means (and standard deviations) of the resultant pistoning forces for each of the joint designs, during the particular activity. This significance is based upon the calculation of standard or Z scores. To compare two joint designs, follow the respective row and column to the appropriate square. NS indicates that there is no statistical significance between the resultant pistoning forces generated by the two orthotic joint designs during the

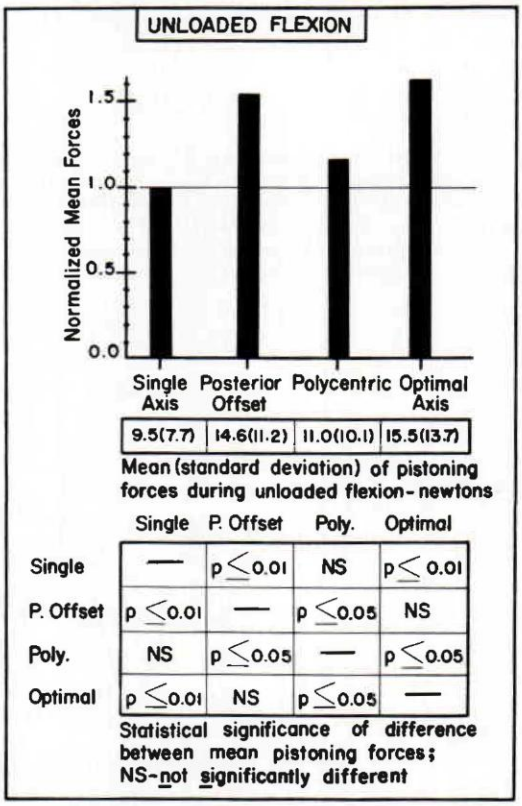


Figure 9—Single axis and polycentric hinges produce statistically significant lower pistoning forces than the posterior offset and optimal designs, during unloaded flexion.

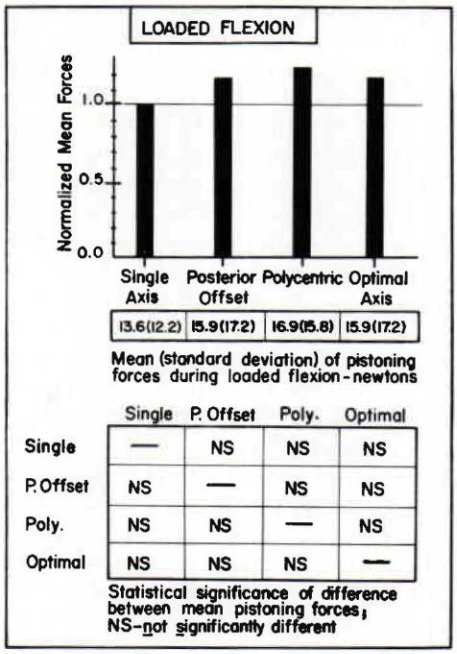


Figure 10—Single axis hinges give rise to the lowest mean pistoning forces during flexion, but the difference between the four joint designs is not statistically significant.

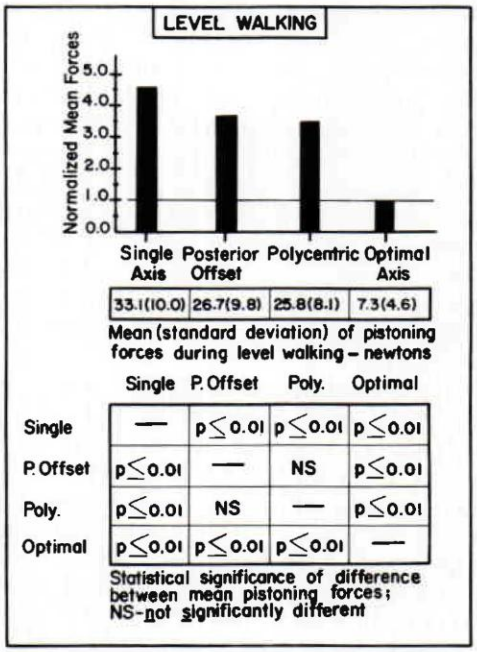


Figure 11—Optimal single axis hinges produce the lowest mean pistoning forces during level walking (statistically significant at p 0.01), followed by the posterior offset and polycentric hinges.

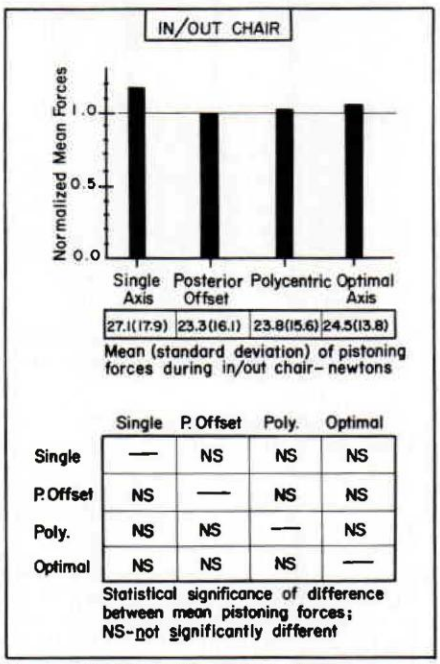


Figure 12—Posterior offset hinges generate the lowest mean pistoning forces with getting in/out chair, but the difference between the four joint designs is not statistically significant.

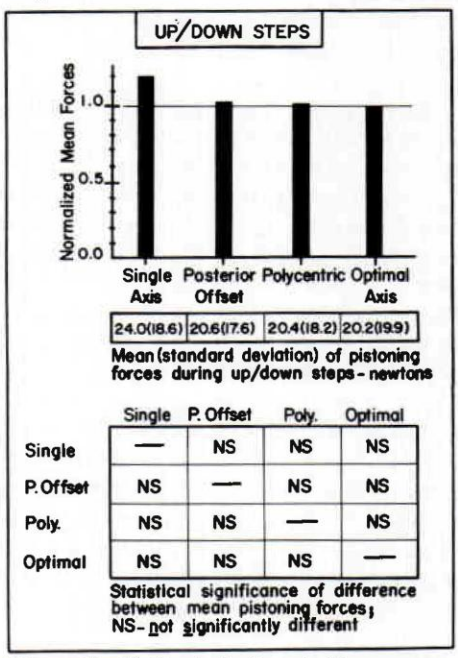


Figure 13—Optimal single axis hinges cause the lowest mean pistoning forces during up/down steps, however, the difference between four joint designs is not statistically significant.

particular activity. When $p \pm 0.01$ appears, a one percent level of significance exists; that is, there is a probability of .99 that the pistoning forces produced by two joint designs are significantly different (in 99 out of 100 cases), or a probability of .01 that they are not different ($p \div 0.01$). A $\pm$ 0.05 indicates a 5 percent level of significance. An overall comparison of the four joint designs with respect to the above parameters, based upon the combination of forces for all five activities, is presented in Fig. 14.

Fig. 9 demonstrates that the single axis and polycentric hinges produce statistically significant lower pistoning constraint forces than the posterior offset and optimal designs during unloaded flexion.

The data in Fig. 10 indicates that single axis hinges give rise to the lowest mean resultant pistoning forces during loaded flexion, however, the difference between the four joint designs is not statistically significant.

Optimal single axis hinges (Fig. 11) produce the lowest mean pistoning forces during level walking (significantly different than the others designs at the one percent level), followed by the posterior offset and polycentric hinges.

Fig. 12 shows that posterior offset hinges generate the lowest mean pistoning constraint forces with getting in/out of a chair, however, the difference between the four joint designs is not statistically significant.

Optimal single axis hinges (Fig. 13) cause the lowest mean pistoning forces during up/down steps, but again, the difference between the four designs is not statistically significant.

When resultant pistoning forces are combined for all five functional activities (Fig. 14), the optimal single axis hinges generate the lowest mean pistoning forces, followed by the polycentric and posterior offset designs, although there is no statisically significant difference between the three.

DISCUSSION

Several interesting observations are evident from the data. First, there are differences between the resultant mean pistoning constraint forces due to the orthotic joint designs in each of the activities tested. However, no one orthotic joint consistently generated lower pistoning forces in all of the activities. Also, in most cases, the differences in mean pistoning forces that do exist between joint designs are not significant. For example, when resultant pistoning constraint forces were combined for all the activities tested (Fig. 14), the optimal single axis hinges produce the lowest mean pistoning force, followed in order by the polycentric and posterior offset designs. However, there is no statistically significant difference between the pistoning forces due to these three orthotic joints. This implies that based upon the criterion of pistoning tendency, no one of these three orthotic knee joint designs offers an advantage over the others.

For most of the functional activities tested, there was greater variation in the

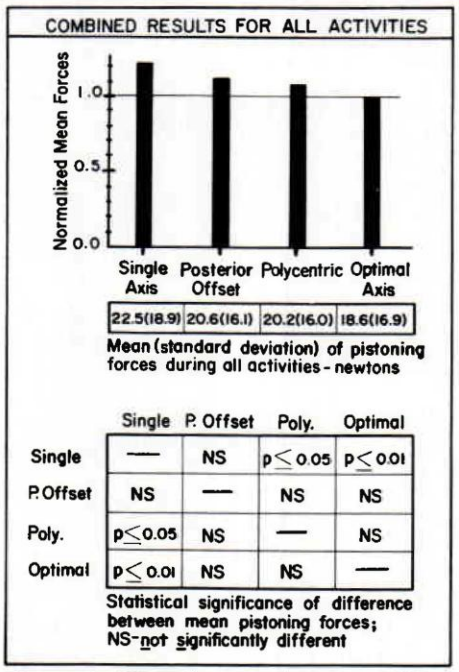


Figure 14—When force results are combined for all five functional activities, the optimal single axis hinges generate the lowest mean pistoning forces, followed by the polycentric and posterior offset designs, although there is no statistically significant difference between the three.

pistoning constraint forces due to the fifteen reapplications of the orthosis with a particular joint design, than there was between the pistoning forces due to the four orthotic joints during a specific activity. This is particularly evident from force results during loaded flexion (Fig. 10), getting in/out of a chair (Fig. 12), and climbing up/down steps (Fig. 13). This implies that for many functional activities, it does not matter which of the four orthotic joint designs tested are incorporated into an orthosis, since any advantage of one over another is masked by the wide variation in pistoning tendency due to daily reapplication of the orthosis.

Another observation apparent from the data is that resultant mean pistoning constraint forces vary with the functional activity performed, for a given orthotic joint design. As previously described, the pistoning constraint is caused by the conflict or mismatch between complex anatomical and simplified orthotic kinematics. The data indicates that the mismatch varies with the activity performed. This implies that no single-axis-of-rotation orthotic knee joint dedign (single axis, posterior offset, and optimal single axis), as well as a polycentric design with a single prescribed axis path, can match the variety of kinematical axis pathways which occur during functional activities closely enough to consistently reduce the pistoning contraint. This observation indicates the need for development of a semiconstrained orthotic knee joint which is able to imitate or allow a variety of normal knee motions to freely occur, yet provide appropriate stability for different single or combined ligamentous insufficiencies.

The results of this study provide some guidelines for orthosis design; however, minimizing the pistoning constraint is but one of several design goals or criteria for a knee orthosis. For example, a primary design criterion would be that the orthotic device provide knee stability when necessary to compensate for injured, healing or absent ligaments. This criterion could be evaluated by measuring relative joint motion in an unstable knee both before and after application of an orthosis design. The goal of minimizing the pistoning constraint force, as described in this report, provides a guide to one aspect of knee orthosis design, provides insights into the biomechanics of natural knee and orthotic knee joints, and is hopefully a first step in the direction of rational design and evaluation of knee orthoses.

ACKNOWLEDGEMENTS

The authors gratefully acknowledge Linda Armstrong, Michael Shoemaker, and Terry Supan, C.P.O., for their contributions in the preliminary phases of this study.

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This work was supported by Research Grant No. 23P55898 from the National Institute of Handicapped Research, Department of Education, Washington, D.C. 20202.

Prosthetic Implications With The Diabetic Patient

Lawrence R. Lange, B.S., C.P.O.

INTRODUCTION

Diabetes mellitus, though a common disorder in man, is seldom understood in its surgical as well as prosthetic implications. In this paper there will be presented a prosthetic view of the impact of the disease. Various surgical as well as non-surgical procedures will be discussed and this will be followed by their respective prosthetic appliances and complications.

Diabetes mellitus affects five percent of the world's population. An additional twenty five percent are carriers of the disease. It is, in addition, the third leading cause of blindness and the seventh leading cause of death.

There are three main types of diabetes and their names relate to the period of onset. Growth-onset or juvenile type is charaterized by its development in childhood, but may appear anytime. Onset is abrupt and leaves its victim very dependent on insulin to survive. This type of diabetes is usually difficult to control and is referred to sometimes as "brittle." The second type of diabetes is mature-onset and usually develops, as the name implies, later in life. It frequently occurs after the age of forty and affects more women than men. The patient still produces insulin, but there is either less produced or the need for insulin becomes too great. The patient is not ketosis prone and is usually very stable. There may or may not be a need to take insulin and most cases are well controlled by a diet low in carbohydrates and sugars. The last major classification is the nonhereditary diabetic. This patient is diabetic due to a decrease or absence of insulin production because of damaged or obliterated cells that produce the insulin. It may be due to disease or trauma. Diseases affecting the thyroid, adrenal, or pituitary glands may also effect the production of insulin and indirectly cause diabetes.

Complications of the disease are numerous, and include: hyperglycemia, insulin allergy, or resistance, insulin edema, susceptibility to infection, diabetic retinopathy (for which there is no prevention), vascular complications, and neuropathy. The last complication listed is of major concern to the prosthetist and needs further discussion.

The neuropathic diabetic usually exhibits one or more of the following disorders: parasthesias, pain, weakness, paralysis, and there is some evidence of autonomic pathology.[10,1] The evidence shows that the autonomic disorders may cause vasoconstriction and decreased blood flow to the extremities. If the blood sugar levels are maintained at or near normal levels, any neuron damage or axonal degeneration can be diverted, but once the process has begun, there is no removal of the damage done.[1] For there to be any correction or halting of the disease process, there must first be a diagnosis.

DIAGNOSIS

The diagnosis of diabetes mellitus is achieved by evaluating the clinical signs. The most easily recognized is the high blood sugar concentration and glycosuria, sugar in the urine. This along with the patient's complaints of constant thirst, hunger and polyuria complete the clinical picture. Breakdown of the body's fat stores produces a high level of circulating ketone bodies. Excretion of ketones in the urine leads to an unfavorable acid-base ratio and further dehydration. (Fig. 1) From this background information it is now possible to look at some of the more intricate workings of the pancreas and its product, insulin.

ETIOLOGY

The pancreas is a large lobulated gland resembling the salivary glands in structure. It is situated in the abdominal cavity just distal to the stomach and runs the length of the large intestine. The pancreas is the sit of both exocrine and endocrine functions. Pancreatic juice, excreted by the Acinar cells, is inactive until mixed in the

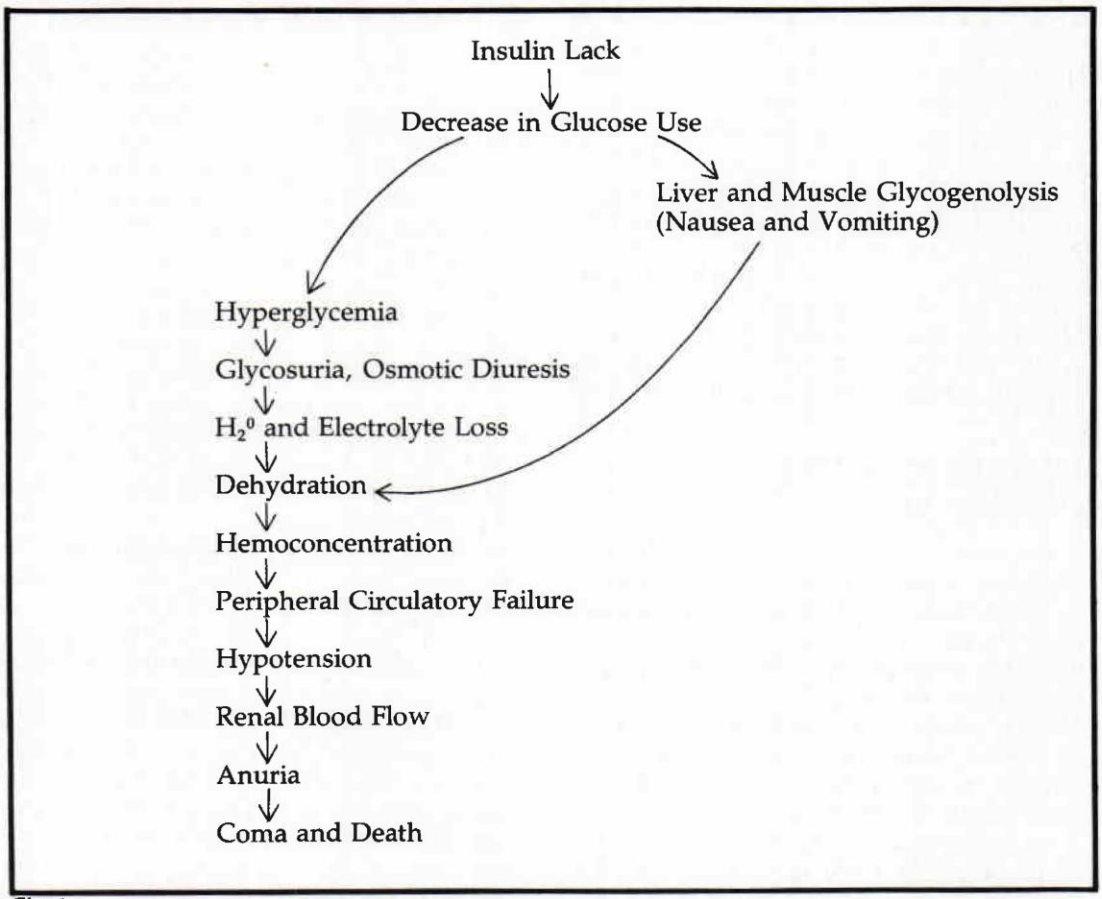

Fig. 1

digestive system and is the exocrine half of the products derived from the pancreas. The endocrine part is composed of two hormones, glucagon and insulin. Glucagon, an antagonist of insulin, is produced in the alpha cells of the pancreas, whereas insulin is produced in the beta cells. Both alpha and beta cells are located within the Islets of Langerhans. Tumor, trauma, or other diseases of the pancreas could lead to an insulin deficient state in the patient. This would be classified under the nonhereditary type of diabetes as mentioned previously.

Insulin, a fifty one amino acid protein was isolated in 1921 by doctors Best and Banting. By 1922 the hormone was available to the diabetic and allowed the well controlled patient a seventy five percent to full normal life span. It is secreted by the beta cells of the pancreas in response to a high glucose concentration in the circulating blood stream. Insulin acts on the blood sugar level by traveling through the bloodstream to its target cells, muscle and adipose. Here it attaches to a receptor structure on the plasma membrane and facilitates entry of the glucose into the cell. Insulin indirectly promotes the anabolic processes, glycogen, fatty acid, and protein synthesis, within the target cells. Glucagon and epinephrine, which act as antagonists to insulin by signalling that the blood sugar is low, support catabolic processes such as gloconeogenesis and gycogen breakdown.

DIABETIC FOOT PROBLEMS

The most conservative treatment for the diabetic with foot problems, such as ulcers or other pressure problems, is the use of special shoes. These shoes are designed to make the neuropathic diabetic bear weight properly on the sole of his foot. It is the diabetic's lack of sensitivity in the plantar surface of the foot that prevents him from bearing weight properly and eventually leads to ulceration and/or gangrene.

According to one study,[6] patients presenting a foot with no sensation and pressure sores on the sole of their feet show absent sural nerve action potentials in ninety five percent of the cases. Inexcitability of small foot muscles secondary to neuropathy leads to foot deformity and poor weight distribution with resultant pressure necrosis of the anesthetic skin. When damage to the foot is more extensive and can not be treated conservatively, an amputation with or without reconstructive vascular surgery may be indicated. These cases will be treated in order of severity.

AMPUTATION SURGERY

The diabetic in need of surgery undergoes special surgical risks due to the early arteriosclerotic state, vascular and neuropathic complications, low resistance to infection and slow healing rate. Surgical, as well as anesthetic, stress on the system may lead to hyperglycemia. The diabetic patient is also prone to the development of cholecystitis, cholelithiasis, and cancer of the pancreas. Assuming that the neuropathy is the major cause for the need for surgery, the coexistence of arterial disease could cause the development and persistence of an ulceration, and, in this way, the two processes are synergistic in their action.[3] The longer the two processes are allowed to operate unchecked, the more severe the damage. Twenty five to fifty percent of patients receiving amputations are diabetics with peripheral vascular disease.[14] The use of a Doppler ultrasound device or mercury in a silastic blood pressure cuff is effective in determining the level of amputation to be performed.

The least severe of the amputations for localized gangrene of the toes is amputation of the affected toe without use of a prosthesis. In a study of fifty three patients with vascular complications, more than fifty percent of the patients were diabetic. It was determined that if there was a palpable pedal pulse or functioning vascular reconstruction that there was a good chance for successful healing following the amputation. However, by a median time of thirteen months, a major amputation was necessary for fifty percent of

the patients. By three and one half years almost seventy five percent came to a major amputation. It was concluded that the long range outlook for the toe amputee was poor and would remain so until a more effective method of distal vascular reconstruction could be found.[11]

Transmetatarsal amputation is the next level proximally. The theory behind this amputation is that, with the diabetic patient, arteriosclerosis affects small vessels more than it does large vessels, and that the gangrene will respond more favorably to this conservative level. This site, as with the toe amputation, allows the amputee to walk without the use of a prosthesis and must be considered one factor in selecting this level of amputation. Localized gangrene of the toes along with controlled infection and good collateral circulation are other reasons for the selection of this site. The study used in this paper reported a range of success from 44 to 83 percent. Of the 433 patients studied, five died within days post-operatively, 69 died within two years and, of 336 survivors, only 63 percent had a well healed amputation. The failures in healing related to factors including: length of time the patient had been diabetic, and the level of palpable pulse. Fifty percent of the patients without a pulse palpable distal to the femoral artery failed to heal. Only four percent, however, with a palpable pedal pulse failed to heal.

The preceeding amputations are all considered conservative. The major surgical amputations should now be considered. Until 1942 the mortality rate for major limb amputations approached that recorded for open heart surgery, about 44 percent. Fifty-one to 66 percent mortality rates were not uncommon in those cases where spreading gangrene or cellulitis were present.[16] Gangrene was almost always the major cause of amputations amongst diabetics. Post operative deaths among diabetics were more than twice those of non-diabetics. Midthigh amputations had a two times higher mortality rate than that for midleg amputations. Sepsis was the major mortality factor in postoperative diabetics. Mortality rates were lowered only when surgical procedures were modified to adapt to the very infection prone diabetic patient. Adequate control of diabetes must be maintained, insulin reactions must be avoided, and procrastination must not occur. If sepsis is controlled, the diabetic patient need not have a lower success rate than that for the nondiabetic.

The Symes amputation affords a lower level of amputation as well as providing the patient with both a better psychological outlook and an increased energy efficient gait over amputations more proximal. With the use of the Doppler ultrasound flow meter, the circulation can be mapped out and a resulting ninety three to ninety eight percent success rate has been achieved by William Wagner, M.D. at Rancho Los Amigos Hospital. The use of a two stage amputation and a drainage pump to remove pooled blood and fluids from the wound also increase his success rate.

The below-the-knee amputation is performed more often for the diabetic vascular patient than for the ischemic nondiabetic. The reasons for this are that the area of the midleg is very well vascularized and is much more viable in the diabetic vascular patient than in the nondiabetic patient. This is demonstrated by a much higher success rate with the diabetic as opposed to the nondiabetic. The below-knee level is selected by the surgeon if he believes the leg will heal, the patient has a good chance of walking again, and the proximal two thirds of the skin are viable. A good popliteal pulse is associated with a high success rate, and in one study, 77 percent of the diabetic patients received and walked with a prosthesis.[2] All the diabetic deaths postoperatively showed cardiovascular or renal complications with an average survival time of 30 months.

The above-the-knee amputation is one of the most radical amputations performed for diabetes. The reasons for using this level are varied and include: failed lower level amputation, rapidly spreading gangrene, procrastinating on either the part of the surgeon or the patient. With the above-the-knee amputation there was

recorded in one study a 76 percent primary healing rate and a nine percent morbidity rate. Seventy five percent of these amputees received and walked with the ischial weight bearing prosthesis with or without some soft tissue bearing. Neither tourniquet nor antibiotics were used either during surgery or postoperatively. This appeared to have no significance in the success of failure rates. Four failures in diabetic patients were resolved with a hip disarticulation amputation. Diabetics showed a 53 percent primary healing rate and 18 percent morbidity. Ischemic lesion of the remainig limb developed in 66 of 283 patients, usually due to the bilaterality of the disease. Many of these had walked with a prosthesis and all 66 required an amputation of the second limb.

POSTSURGICAL MANAGEMENT

Postsurgically, wound healing depends on many factors including: the restoration of local circulation, formation and deposition of collagen, which involved protein as a building block. Both of these processes are diminished in the diabetic due to the patients early arteriosclerosis and their decreased protein production. Insulin appears to have an effect on the collagen phase of healing, in that the well controlled diabetic produces much more protein than the uncontrolled or poorly controlled diabetic. It is therefore most important to have the diabetic well under control both prior to and after surgery. The protein requirement for normal healing is one gram per kilogram body weight.[18] Though this is a small ratio, to the diabetic who is poorly controlled it is virtually unobtainable. Hemoglobin content of the blood needs to be maintained at or near 12 grams per 100 milliliters of blood for proper healing to be initiated.[18] Respiratory, nervous system, musculoskeletal system, or mental disease also act to retard the healing rate. The healing of the residual limb must be already in process in order for prosthetic intervention to be successful. Preprosthetic care is essential and, if undertaken correctly, can assure the patient of an optimal chance of ambulating again.

It is most important that care be taken to prevent contractures. In the above-knee amputee, hip flexion, abduction, or external rotation contractures are the most common. Development of these are enhanced by fibrous tissue development commonly found in the ischemic. Contractures may also be preoperative in nature. These may be the result of conservative treatment to unload the affected limb in the surgeon's desire to postpone the amputation (eg. sitting in a wheelchair to relieve the site of ulceration on the plantar surface of the foot). Contractures require the skills of a therapist and may improve somewhat with the use of a prosthesis.[9]

SKIN PRESSURE PROBLEMS

Mild chronic trauma may be tolerated by an insensitive diabetic without concern until actual skin breakdown occurs. High shear forces or excess pressure, positive or negative, are bad for the amputee and especially so for the diabetic. Coupled with decreased subcutaneous tissue, these forces may result in a major problem for the amputee. Body positioning, local tissue environment, and activity levels may also contribute to the problem and should be distinguished from breakdown related to direct pressure from a poorly contoured socket.

Diabetics usually remain active and ulceration usually comes from repetitive loading of short duration and of skin pressure levels from 30 to 300 pounds per square inch or more.[15] A very tight socket may lead to continuous pressure problems and will create edema as will a popliteal pressure in the below-knee amputee of 25 millimeters mercury. The insensitive diabetic patient at the start of the day can tolerate loading as can a normal person, but when tissues reach their threshold, the insensitive patient has no way of knowing and does not change his gait to reduce the loading. There is a need to make the patient aware when threshold pressures have been reached, and pressure transducers are under research to find an answer to this problem. Tests have also been done to show the effects of pressure

on sites of infection as might occur over an ulcer in a diabetic patient.[15] Results showed that an infection site was more likely to remain localized if pressure was absent that it was if pressure was applied. Pressure over an infected site was also more often the cause of ulceration or abcess than was either pressure or infection alone. Finally, severe pressure applied to an area before infection resulted in pus formation whereas pressure applied after infection did not result in pus formation. It was concluded from these tests that pressure applied continuously over the freshly infected site eliminating access of blood and leukocytes, white blood cells, meant delayed localization and inflammation. A total contact cast or socket with correctly applied tension was actually found to be beneficial for diabetic ulceration healing in another survey and demonstrates that there is a boundary.[15]

PROSTHETIC TREATMENT

The prosthetic implications for the major amputation levels and complications to prosthetic fitting can now be discussed. At the Symes level, there are usually no contrary implications to a prosthesis. (This site is only chosen if the physician believes the patient is going to walk again, as the below knee level is more functional if the patient will always be wheelchair bound.) A permanent prosthesis is usually prescribed of either the bladder or Canadian Symes variety.

The considerations are similar with the below-knee amputee. The knee joint is maintained for a more energy efficient gait. The patient is also expected to walk with a prosthesis in most of these cases. A temporary, or preparatory, prosthesis may be provided for many reasons; the patient may desire to walk as soon as possible or the second limb may be threatened and the physician desires to have the patient gain some prosthetic experience before the loss of the second limb. In these cases, a rigid dressing may be fit immediately after surgery. The doctor may desire to have the patient in an ischial weight bearing prosthesis if healing has not occurred after two months and there is no thought of reamputation at a more proximal level. The residual limb also may not have atrophied enough for a definitive prosthesis to be fitted. In these cased the fitting is not as urgent as the previous situations. In all these cases though, some fit and alignment is sacrificed for early ambulation. A definitive prosthesis may or may not be considered depending on a number of factors including the patient's ability to manage himself on the prosthesis.

With the above-knee amputee, a temporary prosthesis is often encouraged but there are more problems at this level than at more distal levels. It is basically a non-energy efficient gait and, if the patient is debilitated in any way, could be an even greater hinderance to mobility than crutch walking. In no case, and at no level was ischemia in the second limb considered a contraindication to prescription of a prosthesis.

In the geriatric patient, of which many are diabetic, changes in functional ability were more due to increased disease processes than prosthesis use.[8] Eighty-eight percent of patients in one study were successful prosthesis wearers with 15½ percent needing a second amputation within six to 31 months following the first amputation. Sixty six percent of the patients requiring a second amputation were diabetics and functional loss was determined not to be due to prosthesis wearing. In fact a prosthesis was thought to be beneficial in a majority of the cases, even when vascular changes were noted in the second leg. Special care must be taken with the geriatric diabetic, as he may develop cardiovascular problems due to his disease process and an increased energy load on his system (eg. above-knee or hip disarticulation prostheses) may prohibit use of or decreased use of a prosthesis for mobility.

CONCLUSION

A look at the fate of the second limb in the diabetic is in order. The outlook is not positive in any respect and must present the prosthetist with a challenge to provide the diabetic amputee with the best possible service.

Of 67 diabetics studied, only twenty survived.[19] Only five of the 47 deceased had lost their second limb and all of these five died within five years of their second amputation. Within two years after their first amputation, 49 percent were deceased and within five years, 67 percent. The authors stated, "Arteriosclerosis complications will probably kill the patient before he loses his second leg."

SUMMARY

This report presented a look at the types, causes, and complications associated with diabetes as well as the prosthetic outlook for patients following amputation. While most of the material is not new, it does focus the subject and provide background that should be considered by the prosthestist who is confronted with the end result of the disease on a daily basis. An understanding of the disease process is encouraged and hopefully will assist in providing a better service to the patient.

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Temporary Below-Knee Prosthetics

Robert S. Feldman, R.P.T.
Chris Herman, C.P.

INTRODUCTION

In recent decades amputee management has advanced significantly. With the advent of blood flow studies and new surgical procedures, more knee joints are being saved. Physical therapists, using advanced facilitation techniques such as proprioceptive neuromuscular facilitation (PNF), are better able to restore limb strength, range of motion, and coordination. Advances in the field of prosthetics include the development of ischial weight bearing suction sockets, patellar tendon bearing (PTB) below knee sockets, supracondylar suspension, hydraulic knee joints, and multiaction feet.

The prosthetics field has advanced not only in its mechanical and engineering aspects, but in its rehabilitative aspect as well. Among contributions to better amputee management in this area are the development of immediate post-operative prosthetics (IPOP), and the use of temporary prosthetics. The fact that the use of IPOP and adjustable temporary above knee prostheses hastens the patient's return to a more functional lifestyle has been well documented. However, temporary BK prostheses have not been used to their full advantage. Given a more functional, better fitting temporary prosthesis, BK amputees would be better able to use their definitive prosthesis to its full potential.

This article will review some of the literature which traces developments made in the field of temporary BK prothetics; and will set forth criteria for the fitting of BK amputees with functional temporary prostheses.

REVIEW OF THE LITERATURE

In 1965, Hammontree et al,[1] reported on a temporary BK prosthesis being used on a group of geriatric patients. Their clinical team was attempting to provide, "an inexpensive, temporary walking leg, that will not only give the elderly BK amputee an adequate socket for the purpose of limb shrinkage and shaping, but will also get the geriatric patient back on his feet within three to four weeks after operation." The authors also suggested the possible use of the prosthesis as a diagnostic tool with

regard to further prosthetic rehabilitation.

The prosthesis used in Hammontree's study incorporated a plaster socket, either the Northwestern BK pylon or an aluminum pylon, SACH foot, and cuff suspension plus waist belt. Despite drawbacks to this system, it did prove successful in its purpose, and a great deal about temporary systems was learned.

In 1976, Charles H. Pritham et al,[2] reported on the use of thermoplastic components in temporary prostheses. Their prosthesis utilizes either a standard plaster of paris or a plastic PTB socket that is laminated into pieces of galvanized pipe strapping which is attached to a PVC nylon and ankle plug. The use of a plastic PTB socket with a pylon that can be continually adjusted has undoubtedly contributed to the success of this system. The authors report that with only a few exceptions, their patients have been able to wear the temporary prosthesis up to and exceeding six months.

The importance of the use of temporary BK prosthetics was further illustrated by William A. Tosberg, C.P.O., in an article entitled "Temporary Prostheses"[3]. He states, "Following long hospitalization, the patient may be debilitated. The limb may be edematous and painful. Motivation may be impaired. It is therefore essential that the physician has a choice of prescription. All former criteria which determined the type of limb indicated are no longer valid for this type of patient. If a standard prosthesis is prescribed for such a patient, he may never be able to utilize its full potentials."

CRITERIA

The authors of this article agree with Mr. Tosberg that a standard definitive prosthesis is not always the best choice of prescription for a first prosthesis. Temporary prostheses, if made correctly, can be worn until the residual limb completely stabilizes. Thus, when the patient receives his definitive prosthesis, he will be ready to achieve his full rehabilitative potential. While the advantages of a functional temporary prosthesis for the new BK amputees have been acknowledged, criteria with which to fit them have never, to our knowledge, been established. Based on our study of pertinent literature, and our direct experience with temporary prostheses, we now propose the following four criteria to be used in fitting temporary BK prostheses:

- Temporary prosthetic sockets should be a PTB custom fit.
- The temporary prosthesis should be capable of continual adjustment.
- The weight of a temporary prosthesis should be similar to that of a definitive prosthesis.
- The temporary system should meet indications set forth by existing pathological and physical conditions.

Criteria 1

Temporary prosthetic sockets should be a PTB custom fit. There are two important advantages to a PTB custom fit; it distributes pressure in the correct places, and it protects bony areas. As a result, the PTB custom fitted temporary prosthesis is better able to shape and shrink the residual limb than standard plaster of paris sockets.

Criteria 2

The temporary prosthesis should be capable of continual adjustment. In order for an amputee to attain the most function possible from a prosthesis, the prosthesis must be mechanically correct. In a new amputee, the biomechanical situation is constantly changing. As gait improves and the residual limb matures, the prosthesis must be adjusted to accommodate the change. For example, once an amputee becomes more accustomed to walking in a prosthesis, it is often possible to further inset the foot, thus less energy is consumed and ambulation further enhanced.

Criteria 3

The weight of a temporary prosthesis should be similar to that of a definitive prosthesis. When the amputee is ready to wear a definitive prosthesis, the transition should be as smooth as possible. The definitive prosthesis must be lightweight

without compromising strength. The temporary prosthesis should come as close as possible to matching that weight. The exception to this is the patient who would use the temporary leg as the definitive prosthesis, in which case an ultralight BK prosthesis is the choice of prescription.

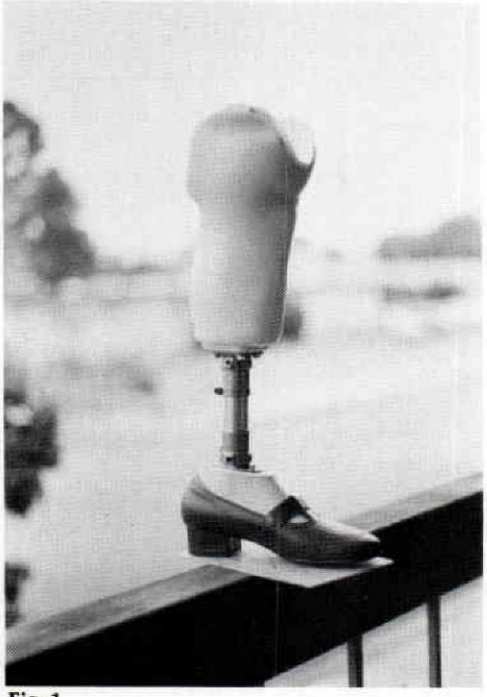
Criteria 4

The temporary system should meet indications set forth by existing pathological and physical conditions. When fabricating the definitive prosthesis, the pathological reason for amputation and the resulting physical condition of the residual limb must be taken into consideration. As the main purpose of temporary prostheses is to prepare the amputee to make the best possible use of the definitive prosthesis, it is essential that these considerations be recognized. The temporary system should be versatile, allowing the use of the same foot and suspension as will be used in the definitive prosthesis.

METHOD

The temporary BK prosthesis used in our facility was developed by Chris Herman C.P. and consists of a custom made 3/16 inch polypropylene PTB socket attached to an Otto Bock endoskeletal pylon (Fig. 1).

By modifying a positive model taken for each patient, we meet the first criteria. Table 1 represents 14 of 17 BK amputees fit at our facility at the time this study began. Because our patient population is small and not all of these amputees have received a definitive prosthesis, statistically significant data can not yet be determined. However, analysis to date indicates that 78.6% of these patients have increased their number of socks worn by five ply or more, and 57.1% by at least 10 ply. Thus, amputees wearing a temporary PTB socket have been able to significantly shrink in volume, maintain a PTB fit, and not lose one day of ambulation time. When shrinkage necessitates a new socket, it can be replaced, when ready, in a matter of minutes during an office or home visit.



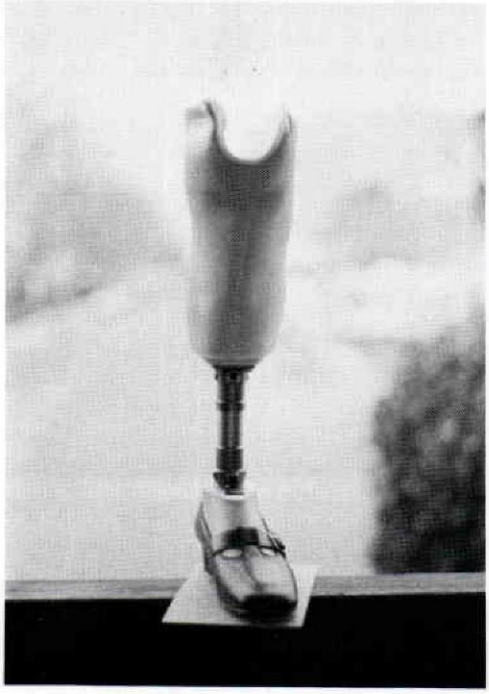


Fig. 1

The second criteria, that of adjustment, is met by the use of a Bock endoskeletal pylon. This system has adjustments at the socket attachment plate and ankle plug, therefore accurate linear and angular adjustments are possible. This system was also used at our facility on the three patients not included in table 1. All three patients were put in temporary BK prostheses for unique reasons. Two are bilateral amputees, and the other presented with a 45° flexion contracture. In the case of the bilateral amputees, having a means of adjustment proved invaluable as they became more accustomed to ambulation on two prostheses. The amputee with a 45° flexion contracture underwent intensive physical therapy to stretch the hamstring muscles. Although this degree of contracture often makes ambulation with a prosthesis difficult, the prosthesis was a valuable tool in maintaining the work of the therapist. The contracture eventually was reduced to 20°. Furthermore, the constant alignment and height adjustments required were easily made with no loss in rehabilitation time.

Criteria number three is met through the combination of a lightweight polypropylene PTB socket and an Otto Bock endoskeletal pylon system. We have found the weight of this combination closely approximates the weight of a conventional exoskeletal BK prosthesis.

In satisfying the final criteria, the Bock pylon and plastic socket have proven versatile in meeting various prosthetic indications. There have been no limitations in choosing the type of prosthetic foot desired since the pylon is designed to accommodate just about all prosthetic feet on the market today. The 3/16 inch polypropylene is strong and durable enough to sustain supracondylar or suprapatellar cuff suspension.

DISCUSSION

The vast majority of amputees receiving temporary BK prostheses present with extremely edematous residual limbs at the time of casting. This is due to a combination of factors including: 1) The trauma of a major surgical procedure, 2) the reluctance of many physicians to use rigid dressings on new amputees, 3) a lack of activity after surgery, and 4) the improper

Patient	Months in System	Equivalent Increase in Number of Sock Ply While in Temporary Prosthesis
1	8	14
2	3	4
3	5	5
4	7	25
5	2	6
6	3	27
7	6	26
8	5	10
9	6	0
10	4	2
11	4	12
12	6	12
13	5	13
14	5	8

Table I

use of ace wraps and shrinkers. The resulting large circumferential discrepancy between mid-patellar tendon (MPT) and the levels distal to MPT, create difficulties not only in the casting for, but in the donning and doffing of the temporary prosthesis. The following modified procedures have proven useful in working with an edematous residual limb:

- The placement of a shrinker under either the cast sock or tube gauze. This helps distribute the residual tissues evenly, allowing the cast to be more easily removed. Many amputees who wear the shrinker under regular prosthetic socks have felt it to be beneficial in hastening residual limb maturity.
- The addition of a double layer of pelite or neoprene glued to the proximal posterior aspect of the insert. When this is done prior to lamination, it can be used to help the patient slip the insert past the narrow anterior-posterior dimension.
- The fabrication of the insert and socket over a three ply sock. This helps to insure a comfortable fit for those patients suspected of little or no limb shrinkage prior to prosthetic wear.
- The drilling of a small hole in the bottom of both the insert and socket. This allows trapped air to escape as the bulbous distal end squeezes past the MPT circumference, making donning easier.
- The placement of a piece of dacron, to be used as tabs, around the medial and lateral sides of the insert. This helps the amputee in doffing a prosthesis with supracondylar suspension.

CONCLUSION

In a study done by O.D. Parker,[4] a comparison was made between amputees fitted with temporary BK prostheses as a first prosthesis, and those immediately fitted with definitive prostheses. The amputees fitted with the temporary prosthesis passed final checkout in the clinic 9.4 months earlier than those who received permanent prostheses. Furthermore, temporary prosthetic wearers received the final permanent prosthesis an average of 8.3 months before those treated in the conventional manner.

As a result of the success we have had with temporary BK prostheses, and our research into studies such as those reviewed above, we have begun to fit all our new BK amputees with temporary prostheses.

In an article published in *Clinical Orthopaedics*, C. Leslie Mitchell, M.D. wrote, "Clearly, in spite of the impressive advances that have been made toward solution of problems of prosthetics, large areas remain for exploration . . . And a tremendous gap yet remains between what an amputee can do and what he should be able to do."[5] The goal of modern prosthetics is to enable the amputee to achieve maximum function with his prosthesis. The correct and diligent use of temporary BK systems enhances the possibility of the amputee reaching this goal.

ACKNOWLEDGEMENT

We wish to acknowledge the assistance of Linda Ann Feldman, M.L.S., for her editorial contributions.

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Technical Note:

A Reinforcing Technique In Orthotics and Prosthetics

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Laird David, R.T.(O)

Since the advent of sheet plastics and its utilization in the field of Orthotics and Prosthetics, many new innovative devices have been developed. The plastic custom molded ankle foot orthosis (AFO) and knee ankle foot orthosis (KAFO) come to mind quickly when referring to such devices.

However, along with these advanced techniques in the field have come certain areas of frustration. The problem of rupturing of the plastic device in high stress areas is one of the problems, but there are others, such as memory retention of the plastic sheet after the vacuum forming process and the plastic being flexible during certain phases of gait when flexibility is not desired.

Obviously, discontinuing the use of sheet plastics in orthotics and prosthestics is not being suggested, but only that certain areas need to be improved.

Several techniques of reinforcing have been used in the plastic rigid ankle foot orthosis (AFO). One type which utilizes metal struts at the ankle was developed at Rancho Los Amigos Hospital.[1] Also, similar to the Rancho style, is one which uses carbon composite inserts instead of metal.[2] These systems are extremely effective when rigidity of the ankle is desired. However, there are other areas which can be reinforced quickly and effectively using the left over portions of polypropylene.

The orthotist-prosthetist at his discretion determines the location, thickness, and width of the plastic strip desired. Recommended steps in actual fabrication are stated below using the overlapping procedure.

- Take the necessary steps and prepare the modified positive mold for drape vacuum forming process. NOTE: It will be assumed by the reader that all processes described in this article refer to the drape molding process.
- Decide where the strips of plastic will be located and measure exact circumference. Cut the strip only as long as the circumference.
- Select the width and thickness desired for the strip.
- Bevel edges lengthwise if desired for a more cosmetic effect.

- Measure and cut the main piece of polypropylene.
- Spray a parting agent on the non-beveled side of strip.
- Place pieces of material on the plate with the parting agent down. Place beveled side of the strip up. (Fig. 1)
- After the plastic has been heated to its proper forming state, remove metal sheet from the oven.
- Pick up strip of the heated plastic and place where previously marked. Twist end into place and work quickly (use powder on gloves) (Figs. 2a and 2b).
- Secure large sheet and form over strip as quickly as possible without moving the extra strip. Note: It is advisable that two knowledgeable people attempt this process (Figs. 3a and 3b).
- After sealed and air is evacuated, remove excess material from the mold. Process is now completed (Fig. 3b).

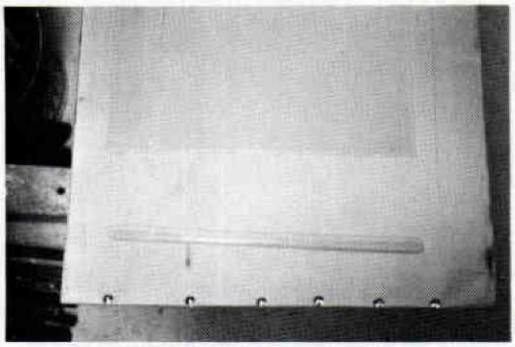
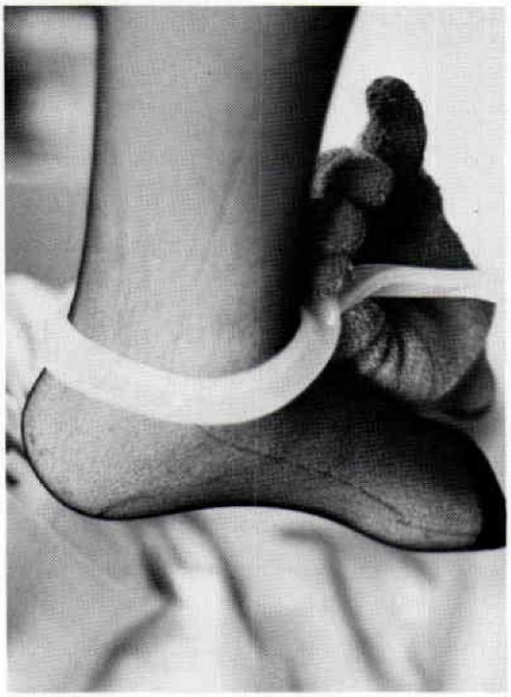
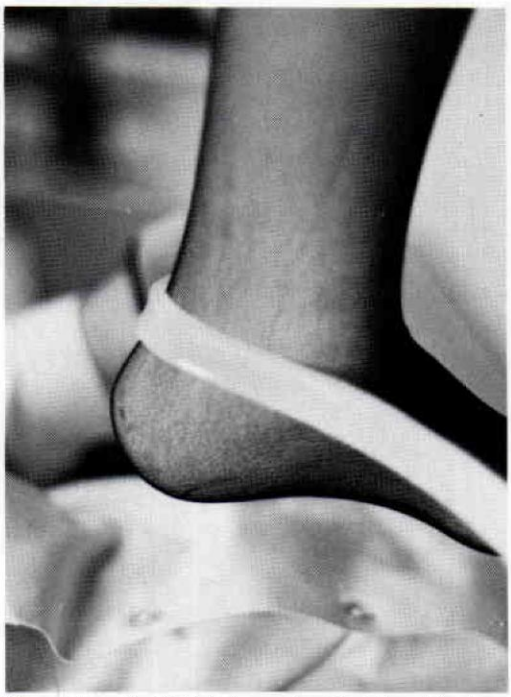


Fig. 1—Main sheet of plastic with 7/8" x 3/16" strip with beveled side up. Note: beveling the strip makes a more cosmetic look to the finished device.

Advantages:
* Cosmetically acceptable
*More durable
*Inexpensive as far as material cost
*Lightweight compared to metal reinforcement
*Inhibits spreading of the plastic away from positive mold after the forming process





Figs. 2a and 2b—Strip is placed over apex of the medial and laterial malleoli. It is important that the strip does not move during the forming process.

Fig. 3a—Large sheet is placed over heated strips on a posterior section of a floor reaction orthosis.

Fig. 3b—Complete forming process on an ankle foot orthosis.

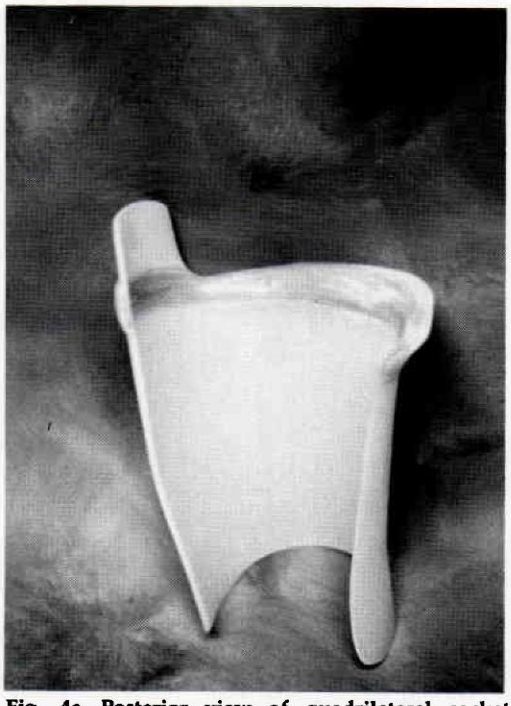


Fig. 4a—Posterior view of quadrilateral socket. Darkened area shows area of concentrated reinforcement.

Fig. 4b—Anterior view of quadrilateral socket (posterior section).

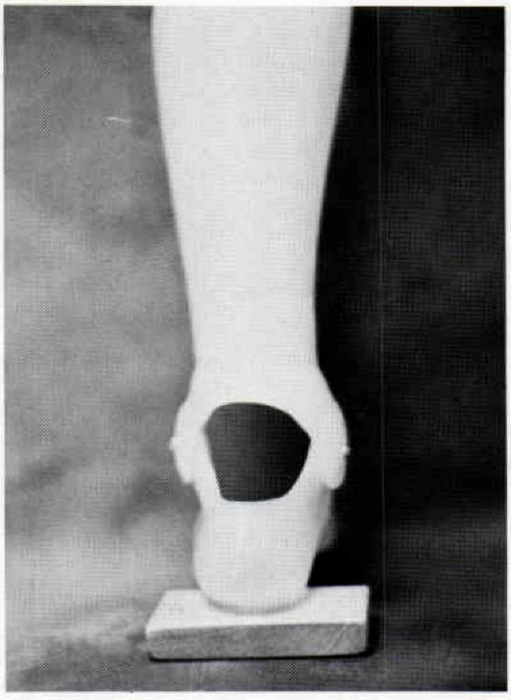


Fig. 5—Reinforcement over and articulated AFO.

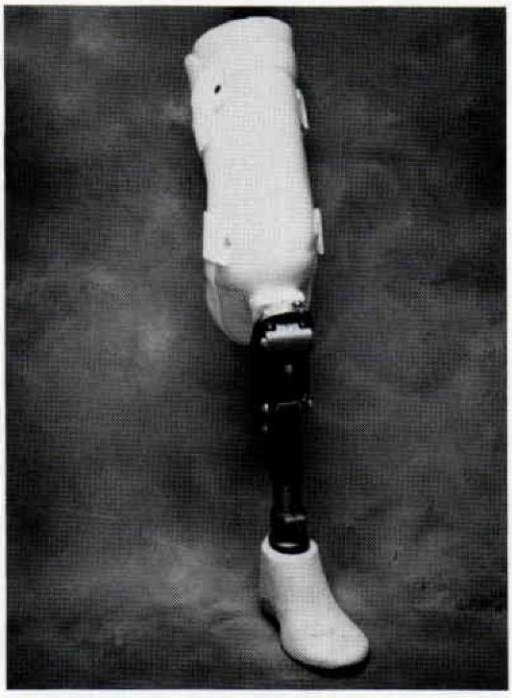


Fig. 6a—Reinforcement of an anterior section of a temporary prosthesis.

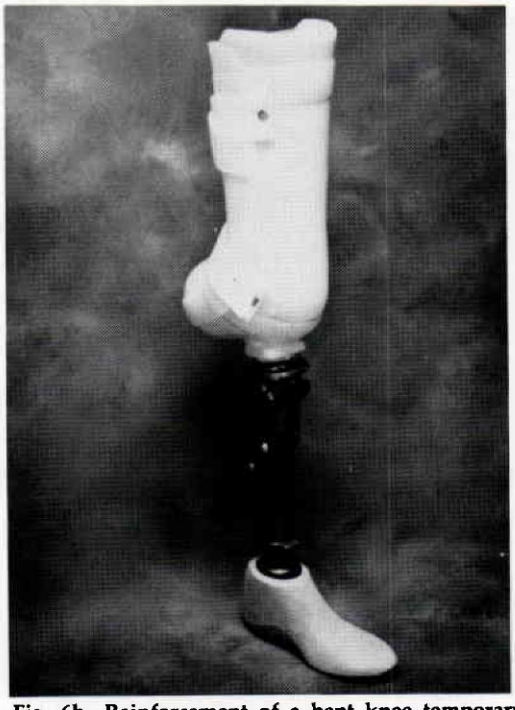


Fig. 6b—Reinforcement of a bent knee temporary prosthesis. One layer is distal and the other is proximal.

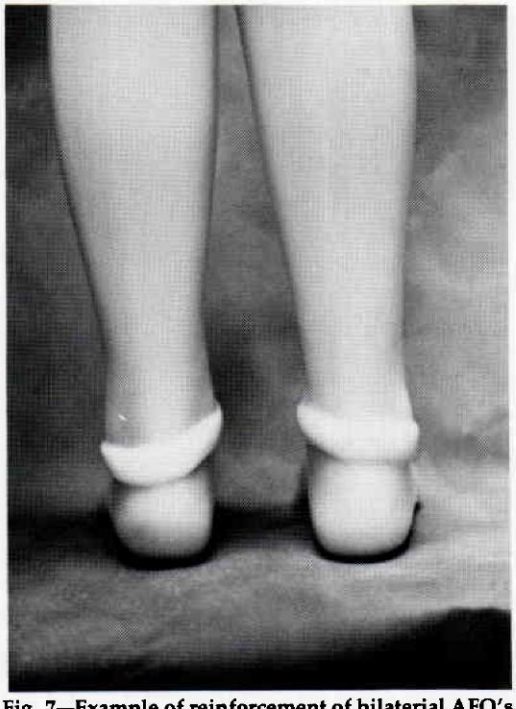


Fig. 7—Example of reinforcement of bilaterial AFO's for a patient afflicted with ALS.

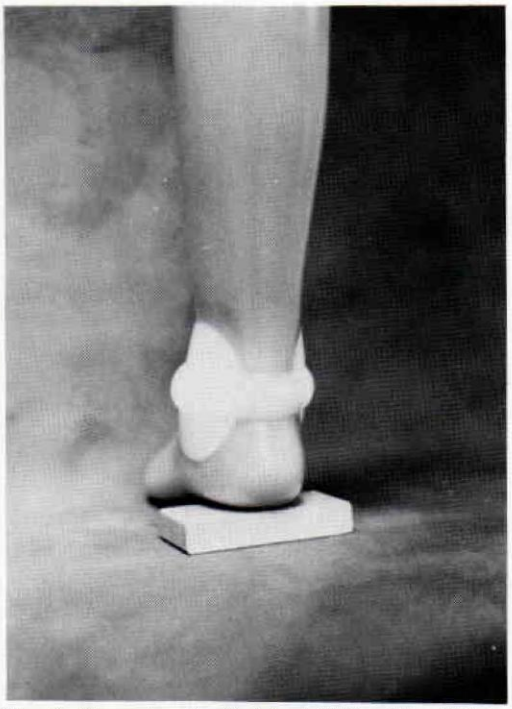


Fig. 8—Example of a finished AFO with a ¼" beveled reinforcement for a CVA patient.

*Plastic may be removed sooner after vacuum forming

Disadvantages:

*More skill with a greater risk of error in forming process

*More difficult to change M-L of the orthosis

*Only adaptable when wrapping circumferentially

*More preparation before vacuum forming

For the past three years the overlapping reinforcing technique has been utilized on many different devices. Some of these other than the ankle complex of the plastic molded ankle foot orthosis (AFO) have been: the calf of the leg in a single upright knee ankle foot orthosis (KAFO) and ankle foot orthosis (AFO), the gluteal area of a quadrilateral socket in prosthetic and orthotic cases (Figs. 4a and 4b), over knee and ankle hinges of KAFO and AFO (these hinges were both plastic (Fig. 5) and metal articulations). Anterior forms have been reinforced in orthotics and used to attach plates for below knee and above knee pylons (Figs. 6a and 6b).

We must emphasize that we do not use this technique in all cases, nor are we suggesting that anyone else do so. We are also not suggesting that we have solved all of our problems by using this technique. But, what we are proposing is a system which might be helpful to orthotists and prosthestists who work extensively with sheet plastics.

REFERENCES

[1]Darrell R. Clark, Thomas R. Lunsford, Reinforced Lower-Limb Orthosis-Design Principles, *Orthotics and Prosthetics*, (32:2:35-45, 1978)

[2]Carlton Fillauer, C.P.O., A New Ankle Foot Orthosis With A Moldable Carbon Composite Insert, *Orthotics and Prosthetics*, (35:3:13-16, 1981)

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New Publications

Reviewed by Charles Pritham, C.P.O.

Lower-Limb Orthotics including Orthotist's Supplement

1981 Revision. 524 pages, $35.00

Lower-Limb Prosthetics including Prosthetist Supplement

1982 Revision. 557 pages, $35.00

Both published by and available from:

Prosthetics and Orthotics
New York University
Post-Graduate Medical School
317 East 34th Street
New York, N.Y. 10016

Due to their many similarities in function and design, these two books are considered together. They are intended to be used by all members of the clinic team and thus include information pertinent to the prescribing physician, therapist and prosthetist/orthotist. Covered are such topics as pathology, mechanics, biomechanics, gait, gait deviations, components, prescription guidelines, training, functional evaluation with the device (checkout), and fabrication. As such, they provide a common body of knowledge for the team members and can be recommended to new or inexperienced personnel as basic references. They should also prove useful as texts for inservice training when supplemented by lectures from the various team members and pertinent readings.

More specifically they provide the basic body of knowledge needed by a prosthetist/orthotist. They contain the information used at New York University in the teachings of fitting and fabrication of Below-Knee and Above-Knee prostheses, Foot, Ankle Foot, and Knee Ankle Foot Orthoses. The material should be familiar to all, as wherever possible, sections of now out of print texts are reproduced where they will suffice. The instructional material presented has been well tested in numerous courses at New York University and the two volumes can be used with confidence in instructing technicians and fledgling prosthetist/orthotists. While the companion volumes cover plastic orthoses, they do not treat such relative rarities as Symes, Knee Disarticulation, Hip Disarticulation/Hemipelvectomy, or Knee Orthoses in the sort of detail necessary to fit and fabricate such devices. They are considered in their theoretical aspects. Supracondylar and Supracondylar/Suprapatellar Suspension of Below-Knee Prostheses receives similar treatment, despite their importance in modern prosthestic practice.

In short, the two books are useful additions and could be used with confidence in instructional courses, or in preparing for certification, provided it is remembered they are basic texts.

A Manual for the Congenital Unilateral Below-Elbow Child Amputee

Rhoda Weiss-Lambrou, M.Sc., O.8.(C). The American Occupational Therapy Association, Inc., 1383 Piccard Drive, Rockville Md. 20850.
105 pages, index, and 6 appendices.

This book is described by the author as a guide to the non-expert therapist occasionally confronted with such a patient, and is an outgrowth of her thesis project. It is a result of a comprehensive review of the literature and has been reviewed and commented on by therapists experienced with the topic. As such it may not necessarily be the last word on the subject, but it is a comprehensive and uncontroversial treatment of the problem and should prove useful to all members of the clinic team. The book covers: limb deficiencies and etiology, prosthetic prescription and components, evaluation, and training.

While we as prosthetists/orthotists may quibble about the use of the term "Checkout" and may question the distribution of certain responsibilities to the occupational therapist and not to the prosthetist, in general this is a very useful book. It would behoove the prosthetist to have it available, not only for his own reference, but also, for the therapists, expert and non-expert.

Advanced Prosthetics/Orthotics In Cancer Management

Principal author, H.R. Lehneis, Ph.D., C.P.O. Institute of Rehabilitation Medicine, New York University Medical Center, 400 East 34th St., New York, New York 10016.
114 pages, 3 appendices.

This report describes the results of a course of clinical investigation undertaken over a period of some two years and intended to enhance the Prosthetic/Orthotic treatment of those patients who undergo high-level amputation, or resection as a result of cancer. The initial half of the monograph, that dealing with hip disarticulation and hemi-pelvectomy, essentially was published in totu in the June 1981 issue of *Orthotics and Prosthetics*. It includes much food for thought and any prosthetist would do well to review the principles of socket design put forth. The second section describes in very limited detail preliminary attempts to develop prefabricated modular orthoses intended for rapid application after local resections. This aspect of the project was greatly hampered by lack of suitable patients, but several intriguing possibilities were developed. A third section describes the use of Myo-electric Control to enhance the function of prostheses for high-level upper-limb amputations. A very small number of patients were treated and modified electronic circuitry based on designs of Hans Schmidl, of Italy, was utilized.

Aside from the purely technical details involved in the development of improved prostheses and orthoses the intent of the project was to develop a model for multidisciplinary treatment of the patient as a whole; addressing not just physical needs but psycho-social ones as well. The report is quite candid in discussing the relative lack of success in this area, primarily due to the low number of patients seen. If nothing else, however, the subsequent

lack of statistics and the reliance on brief summary statements is refreshing.

In any event, those involved in the multi-disciplinary treatment of these patients should find the material of interest. Prosthetist/Orthotists involved in such teams would do well to review the material and to call it to the attention of their co-workers.

The Limb Deficient Child

ed. Y. Setoguchi, M.D., and Ruth Rosenfelder, R.P.T., 1982. Charles C. Thomas, 2600 South First Street, Springfield, Illinois. 324p., Index and 6 appendices.

This book is the result of the Child Amputee Prosthetics Project's (CAPP) many years of experience with the subject. The CAPP clinic team has long been considered by many as pre-eminent and more importantly has functioned for years as a stable cohesive unit with very little turnover. The book reflects these facts not only in the opinions put forth by the various members of the team but also, in the extensive and well organized bibliography.

In any such book it is tempting to devote the bulk of the space to a review of the hardware and gadgetry available. It is particularly welcome that while the authors have not ignored the prosthetic armamentarium they have devoted much of their attention to such less easily addressed topics as therapy and management of educational and psychological problems. It is in this respect that the many years of experience of the CAPP team are most valuable. The treatment of the child amputee is considered by level, discipline, and chronologically. A practitioner of a particular discipline may feel that his is slighted, but the aim has been to provide a balanced overview. The bibliography provides ample opportunity for further exploration. Among the interesting aspects of the books, are brief histories of CAPP and of the Thalidomide episode.

This is a very interesting and valuable book, and one well worthwhile purchasing.

Introducing Houston, Texas, A City of Firsts

Houston is the city that gave America its first domed stadium, the Astrodome, its first astronauts, and its first artificial heart transplant. Oil made the city boom, its port made it international, and the NASA LBJ Space Center as well as the Texas Medical center made it a scientific "brain pool," attracting thousands of scientists and engineers to this fifth largest city in America.

Houston's cultural heritage gives the city four flavors—western, cosmopolitan, Mexican and Creole—each contributing to the cultural, entertainment, and dining styles available here.

After dark, Houston is a festival—for wining and dining! The city's proximity to rich Texas cattle country and the Gulf of Mexico is immediately evident in the number of first class steak and seafood restaurants to be found. But the city is also a melting pot of international cuisines: French, Chinese, Japanese, Indian, German, and of course, Mexican. Then there are the clubs and night spots where you can enjoy disco or country western or jazz or Las Vegas style groups or—horseshoe pitching.

If you're bringing the kids along, Houston has lots of amusement centers. At Astroworld Six Flags there are over eleven theme parks, 65 acres with more than 100 shows, restaurants and rides including the Texas Cyclone, and Thunder River, which simulates white water rafting. Then there's the Harmann Park Zoo, 42 acres featuring over 200 varieties of birds and thousands of animals. And if ice skating is what they want, you can take them to Galleria—even in summer.

Not to be missed is the Burke Baker Planetarium where dome-projected celestial shows transport the visitor to other planets and galaxies of outer space. Adjoining the Planetarium are the museums of Natural Science and Medical Science.

The Planetarium is located in Hermann Park, which also houses an outdoor theater, rose gardens and a Chinese Pavilion.

Houston is a picture postcard city. Within its scope are rolling ranchlands and grazing cattle, sandy white beaches at nearby Galveston, and the Alabama Coushatta Indian Reservation at Big Thicket. It has dazzling contemporary architecture, exciting sports, a Fabulous nightlife, one of the best medical centers in the world, and the pleasure and honor of hosting AOPA this fall.

Highlights of Houston

Friday morning, October 22, climb aboard for an in-depth tour of Houston! Visit Rice University, the Texas Medical Center, Sam Houston Park, the Heritage Homes, and the Civic Center with its Alley Theater.

Feast your eyes on the magnificient mansions of River Oaks and Herman Park's famous Fragrant Gardens.

Take a relaxing break with a delicious luncheon at the Brownstone. Uniquely furnished with fine antiques and an art gallery upstairs, the Brownstone promises a perfect ending to a busy morning and to recharge you for the busy afternoon.

Preliminary Program

Monday, October 18
9:00 a.m.–5:00 p.m.
AOPA Executive Committee Meeting
1:30 p.m.–5:00 p.m.
AOPA Board of Directors Workshop

Tuesday, October 19
9:00 a.m.–5:00 p.m.
Exhibit Set-Up
9:00 a.m.–5:00 p.m.
Business Procedure and Data Committee
9:30 a.m.–5:00 p.m.
AOPA Board of Directors
11:00 a.m.–5:00 p.m.
Registration
Noon–5:00 p.m.
Bus to NASA
7:00 p.m.–9:30 p.m.
Presidents' Welcome Reception

Wednesday, October 20
8:00 a.m.–5:00 p.m.
Registration
9:00 a.m.–10:00 a.m.
Continental Breakfast
10:00 a.m.–5:00 p.m.
Exhibit Hall Opening
10:30 a.m.–12:30 p.m.
Ladies Auxiliary Meeting
12:30 a.m.–2:00 p.m.
AOPA Past Presidents' Lunch
1:00 p.m.–5:00 p.m.
Scientific Session

Thursday, October 21
8:00 a.m.–5:00 p.m.
Registration/Hospitality
9:00 a.m.–5:00 p.m.
Exhibits
9:00 a.m.–Noon
Scientific Session
9:00 a.m.–1:00 p.m.
F-19 Committee Meeting
9:00 a.m.–5:00 p.m.
JEC Meeting
1:00 p.m.–5:00 p.m.
Business Seminar
1:00 p.m.–5:00 p.m.
Free Shuttle Bus to the Galleria Mall
3:00 p.m.–5:00 p.m.
AOPA Suppliers Meeting
6:00 p.m.–1:00 p.m.
RODEO!

Friday, October 22
8:00 a.m.–5:00 p.m.
Registration/Hospitality
8:00 a.m.–10:00 a.m.
Editorial Board Meeting
9:00 a.m.–1:00 p.m.
Exhibits
9:00 a.m.–2:00 p.m.
Business Workshop
9:00 a.m.–2:30 p.m.
Scientific Session
9:30 a.m.–4:00 p.m.
Tour of Houston
10:00 a.m.–1:00 p.m.
COPE/UCOPE
1:00 p.m.–
Exhibit Tear Down
2:30 p.m.–5:00 p.m.
AOPA Annual Business Meeting

Saturday, October 23
8:30 a.m.–1:00 p.m.
Registration
9:00 a.m.–5:00 p.m.
ABC Board of Directors Meeting
9:00 a.m.–3:00 p.m.
Scientific Session
9:00 a.m.–5:00 p.m.
Hospitality
10:00 a.m.–1:00 p.m.
Ladies Auxiliary Breakfast Meeting
(with photographs)
11:30 a.m.–1:30 p.m.
ABC Report Luncheon
6:30 p.m.–11:30 p.m.
Concluding Banquet

Scientific Program

Stand-N-Go: A Self-propelled Standing Wheelchair for Children
M. Silverman, CO; O. Silverman, CO; F. Netznik

Lower Limb Biomechanics and Orthotics of the Child With Cerebral Palsy
J.M. Carlson, MS, CPO

Custom Molded Shoes for the Diabetic
E. Schultz, DPM

Photographic Principles of Use to the Orthotist/Prosthetist
W. Eversmann, MD

Concepts, Practices, and Experiences with Functional Bracing of the Tibial Fracture
W. Racette, CPO

Sports and Upper Extremity Prosthetics
B. Rodacy; R. Dick

The Mini-Grip: A New High-performance Child's Prosthetic Hand
B. Radocy; R. Dick

The Science of Ergonomics: Promoting the Development and Maintenance of Amputee Neuromuscular Systems with Prosthetic Design and Therapy
B. Radocy; R. Dick

A New Combination of Old P.T.B.-S/C Suspension System
A Reid, CPO

Substitutes For Plaster of Paris in Temporary Prostheses
M. Stills, CO; S, Kapp, CP; D. Snelson, CPO; V. Mooney, MD

Impact of New Casting Tapes on Prosthetic Fabrication
J. Seery; J. Leal, CP; J. Malone, MD

The Lerman Multiligamentus Knee Control Orthosis
M. Lerman, CO; J. Schwartz, CO; M. Schwartz, CO

Mastectomy Prosthetics-Post-Surgical Care
L. Klemmt, CO

The Orthotic Management of Fractures
M. Stills, CO; V. Mooney, MD

Investigation and Application of the PPT Material in Orthopaedic Technology
Dr. W. Kuehnegger, CO

Ultralight Above Knee Prosthesis
H. Lawall, Jr.; T. Vachranukunkiet, MD

Ionomer Resin Check Sockets With Definitive Application
S. Kapp, CP; V. Mooney, MD; D. Snelson, CPO, M. Stills, CO

Test Socket Materials: Uses and Fabrication
A. Hrynka, B.S. Ed., CPO

Solutions To Some Problems With Knee-Disarticulation Prostheses
D. Snelson, CPO; S. Kapp, CO; V. Mooney, MD; M. Stills, CO

Clinical Trial of the "Prehensile Hand"
W. Winter, MD

Adjusting to Every Aspect of Amputation
S. Lane

Experiences With the Handicap National Ski Championships
A. Finnieston, CPO

An Elevating Concept for Prosthetics
G. Fessenden, CP

The Contour-U Custom Total Seating System
O. Silverman, CO; M. W. Silverman, CO; J. Torossy, CO

The "No Post Op Cast" Post Op Body Jacket
H. Shufflebarger, MD; D. Henkel, CO

The Ultracentric Knee Joint: A New State of the Art
R. Foster, CPO; C. Foster, MD

The Custom Cuirass Ventilator
G. Hecht, CO

Total Non-weight Bearing Orthosis for Neurotrophic Plantar Foot Ulcer
T. Vachranukunkiet, MD; H. Lawall, Jr.

The Miami Prefabricated Fracture Brace System
J. Zagorski, MD; J. Schenkman, MD: A. Finnieston, CPO; R. Elliott, CPO

Static Rotational Control Cervical Orthosis for the Treatment of Congenital Muscular Torticollis and Associated Plagiocephaly and Hemihypoplasia
Barry W. Townsend, CPO

Terminal Transverse Congenital Limb Deficiency of the Forearm
Donald Shurr, LPT; Reginald R. Cooper, MD; Joseph Buckwalter, MD; and William Blair, MD

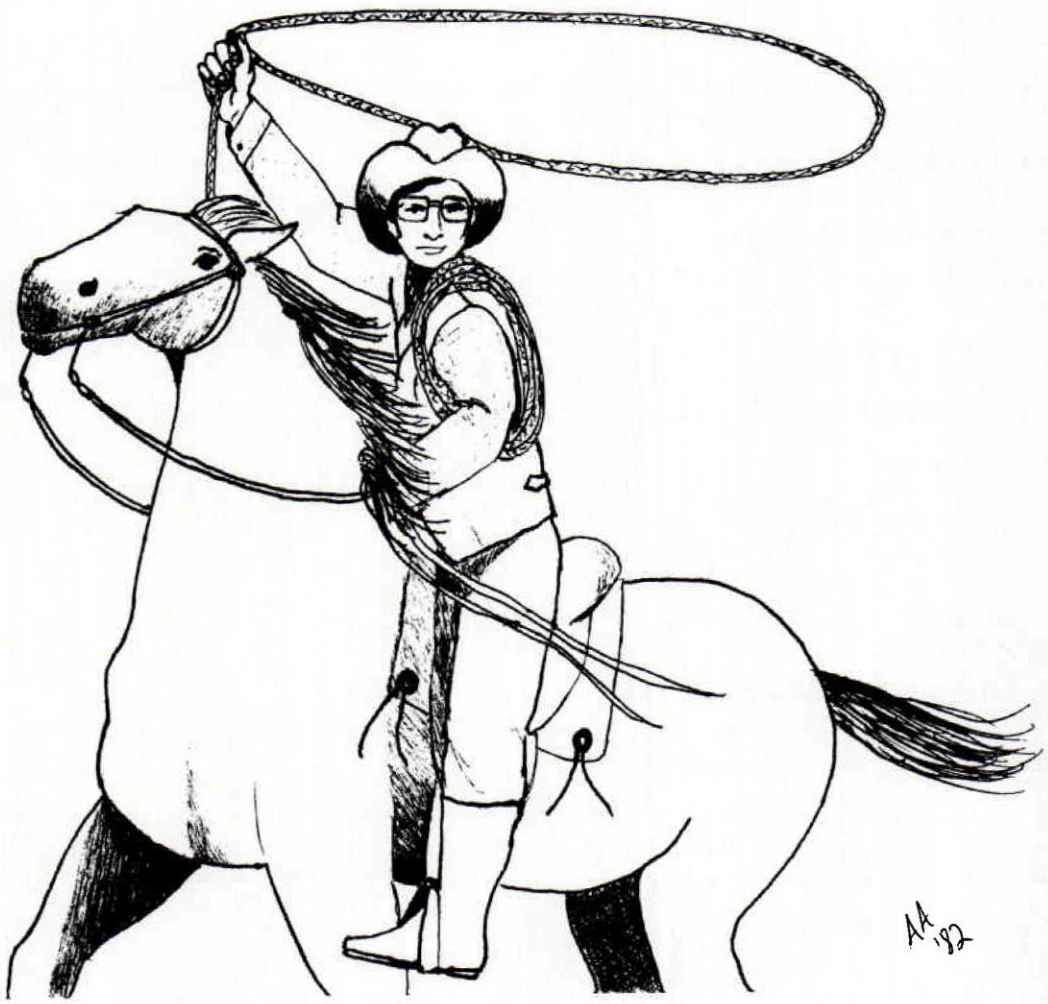
Assembly '82

RODEO!

On Thursday, October 21, several busloads of AOPA Assembly participants will converge on Simonton, Texas, for one of the most action-packed, thrilling evening's worth of entertainment available anywhere. There is nothing in the world of sport that can compare with this uniquely American concoction. Nowhere else — not in Spanish bullfighting, nor Afghan polo, nor English steeplechasing — do men and animals compete with and against each other in so many different and often dangerous ways. Many rodeo events grew out of competitions designed to showcase the skills that cowboys had learned in the normal course of their work but, when men began pitting themselves against bulls and wild horses, they went far beyond the safer challenges of beating the clock or besting each other. They took the test of riding skills to the farthest imaginable extreme when they added to the unpredictability, power, and rage of huge animals with sharp hooves and horns. It would be tough to find a harder test of courage in any other sport than the one bull and bronc riders face.

Luckily, the folks who run the Round-Up Rodeo have planned a few lulls in the heart-stopping action so that you'll be able to calm down with some Texas Bar-B-Que and country western music. Then it'll be back to the arena for more of the pagentry, excitement and just palin fun that make up a rodeo.

This event promises to be one of the best AOPA has ever planned. Don't miss it!



Shamrock Hilton
Houston, Texas
October 19-23

AOPA 1982 National Assembly

Advance Registration Form

american orthotic & prosthetic association

Please indicate the name(s) as you would like them to appear on badge(s).

Nickname(s) if desired _____ CP CPO CO

_____ MD Other

_____ CP CPO CO

_____ MD Other

( ) AOPA Member ( ) Active ( ) Affiliate ( ) Associate ( ) Institutional

Name of firm or affiliation _____

Address of firm _____

Home address _____

Paid registration includes name badge, entrance to exhibit areas, scientific sessions, Closing Banquet and Business Seminar. No refunds or exchanges on ticket functions after September 15, 1982.

Paid Registrations	**Price**	**Number**	**Cost**
Employee of AOPA member firm	$175.00		
Employee of Non-AOPA member firm	200.00		
Spouse or Child over 18	95.00		
Student	120.00		
Physician	175.00		
Emeritus practitioner or child under 18 (see item 5)	0.00		
Additional Functions			
Ladies Auxiliary Breakfast	$ 11.00		
Presidents' Welcome Reception	12.00		
City Tour/Luncheon	23.00		
Rodeo	25.00		
TOTAL			

Make checks payable to AOPA National Assembly and send to:
AOPA National Assembly, 717 Pendleton Street
Alexandria, VA 22314

STRIKE IT RICH AT THE 1982 NATIONAL ASSEMBLY HOUSTON, TEXAS October 19-23

1982 AOPA National Assembly
Advanced Registration Form
Additional Information

1. Registrants are responsible for making their own reservations with the Shamrock Hilton Hotel using the special postage paid reservation cards. Note that reservations can only be assured if these cards are received at the Hotel by September 17, 1982.

2. On-site registration and function ticket prices will be more expensive.

3. No refunds or exchanges will be made after September 15, 1982.

4. All items must be paid in advance. There will be no post-Assembly invoicing.

5. Emeritus practitioners and children under 18 are entitled to a free badge which allows admittance to all non-pay business seminars, scientific sessions and exhibit areas. Function tickets must be purchased separately.

6. Registration information and payment must be received by September 15, 1982 to be considered for pre-registration. Registration forms received thereafter will be returned.

Please fill out:

Is this the first AOPA Assembly for any of the persons registering?
( ) yes ( ) no If yes, please list names: _____

_____

_____

For those persons who have attended more than one assembly, please list names and how many years attended if possible.

_____

_____

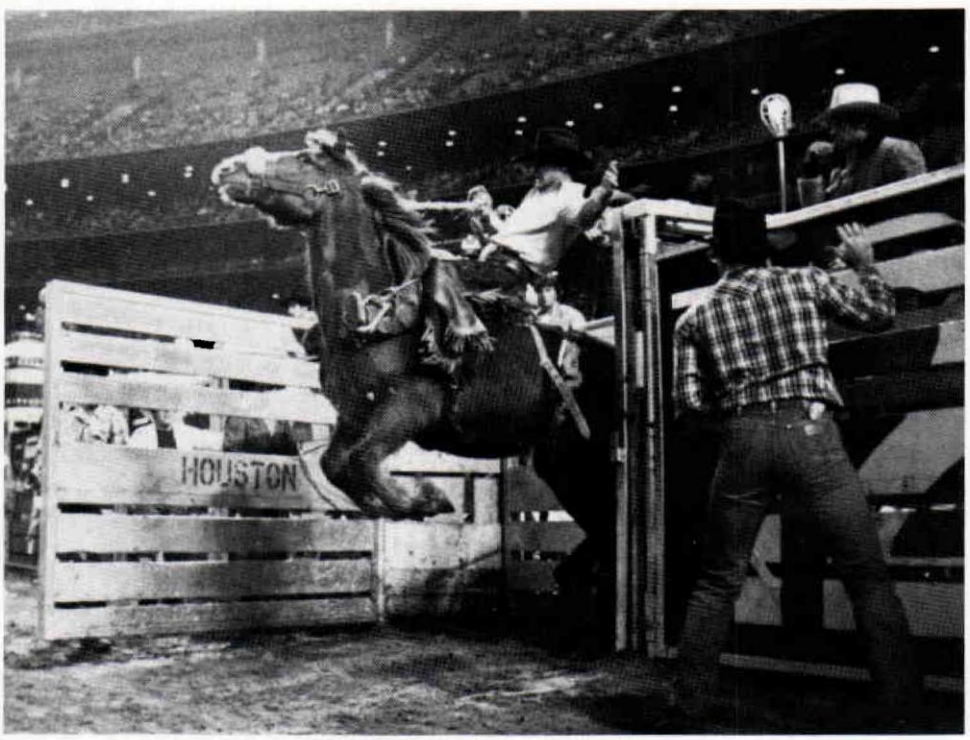
_____

SEND YOUR REGISTRATION IN EARLY AS REGISTRATIONS RECECEIVED AFTER SEPTEMBER 15, 1982 WILL BE CHARGED AT THE MORE EXPENSIVE ON SITE RATES.

Exhibitors

Listed are those firms which have already contracted to exhibit at the 1982 Assembly.

Apex Foot Products
Acor Orthopaedic
Alden Shoe Company
American Prosthetics, Inc.
ATCO Surgical Supports
Auto Soler
Becker Orthopedic Appliance Company
Bell Horn
Bremer Brace of Florida
CAMP International, Inc.
Carapace, Inc.
Cascade Orthotic Supply
CASH Manufacturing Company
Classique, Inc.
Comfit Shoes
Comfort Sock Manufacturing Company
Davis Grosse, Inc.
DAW Industries, Inc.
Durr-Fillaver Medical, Inc.
Feiner Brothers/Elkay Supply
Florida Brace Corporation
Frankel Shoe Machinery
Freeman Manufacturing Company
Gingher, Inc.
Gottfried Medical, Inc.
Hosmer Dorrance Corporation



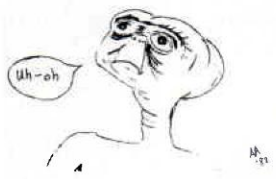
Exhibitors (continued)

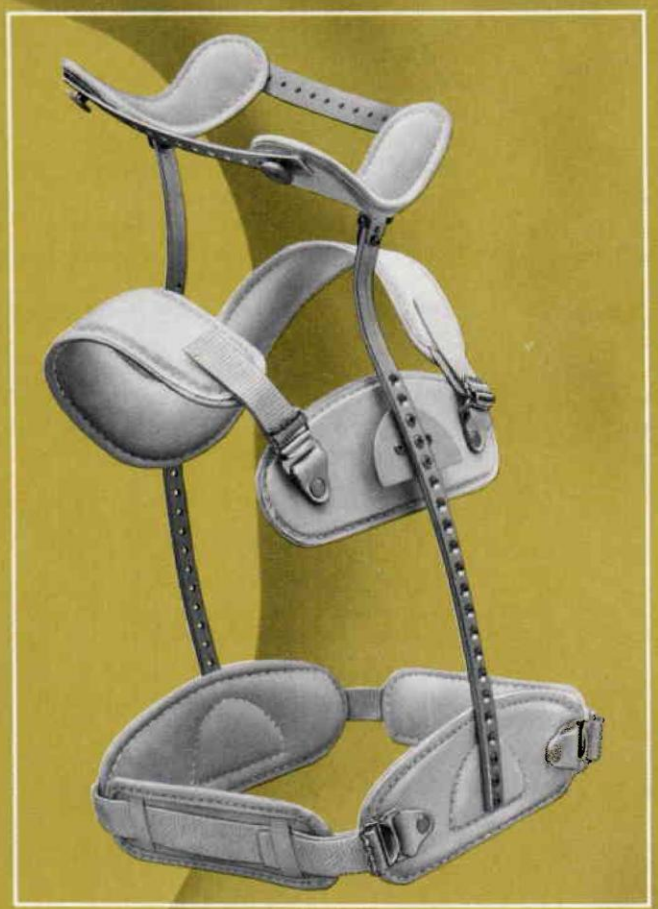
Listed are those firms which have already contracted to exhibit at the 1982 Assembly.

Kingsley Manufacturing Company
Knit-Rite, Inc.
Lehde Brown Orthopedic Company
Maramed Precision Corporation
M.J. Markell Shoe Co.
May Marie Company
Medical Center Prosthetics
Jevy Miller I.D. Shoes
Miller Shoe
P.W. Minor and Son, Inc.
Motion Control
Northeast Paramedical Industries, Inc.
The Ohio Willow Wood Company
Orthomedics, Inc.
Orthopedic Systems, Inc.
Orthotic Prosthetic Laboratory
Otto Bock Orthopedic Industry, Inc.
Pel Supply Company
Pin Dot Products
Pope Brace
Professional Technology, Inc.
Rehabilitation Technical Components, Inc.
Roloke Company
Sabel Shoe Company
Scott Orthotic Labs, Inc.
Shamrock Medical, Inc.
Southern Prosthetic Supply
Sports Supports, Inc.
Sutton/Landis Shoe Machinery
Therapeutic Recreation Systems, Inc.
Tri Hawk Corporation
Truform Orthotics and Prosthetics
Tru-Mold Shoes, Inc.
United States Manufacturing Company
U.S. Orthotics
Washington Prosthetic Supplies

See Ya In Houston!
Shamrock Hilton
October 19-23

Don't get left behind! There's still time to sign up before the Assembly blasts off in Houston. Send in your registration today!





J-22 Our Two-Post Orthosis (J-22) provides indicated stabilization of the cervical and upper thoracic regions with optimum patient comfort. Prescribe it by name, "Florida Brace J-22," to be sure your patient receives the correct orthosis with the results you want. Most ethical dispensing orthotists throughout the country can supply and fit the J-22 to your prescription in a matter of hours. Florida Brace Corporation. P.O. Box 1299, Winter Park, Florida 32789.

Florida Brace Corporation

Classified Ads

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Orthotist-Prosthetist

A unique opportunity exists for a qualified and experienced Orthotist-Prosthetist at the King Faisal Specialist Hospital and Research Centre in Riyadh, Saudi Arabia. This 250 bed tertiary referral facility, with 150 bed minimal care unit opened in June, is one of the leading medical centers in the Middle East.

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Interested candidates should forward resumes to: Kathleen Langan, Personnel Consultant, Hospital Corporation of America-International Division, P.O. Box 550, Nashville, Tn. 37202. AN EQUAL OPPORTUNITY EMPLOYER.

HCA
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Orthotic-Prosthetic facility for sale in Pacific Northwest—only facility in area. Excellent opportunity for certified individual. Reply to AOPA Box 68205, 717 Pendleton Street, Alexandria, VA 22314.

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Orthotist—Certified or Board eligible, needed for Elizabethtown Hospital for Children and Youth. Elizabethtown is a unique, 45-bed rehabilitation hospital with an active in-patient and out-patient program. Orthopedic Appliance Department currently exists with two prosthetists, one orthotist and CPO director, three assistants and three technicians. The Department is involved with scoliosis, amputee, pediatric orthopedic, brain injury, spinal cord injury, adaptive seating/adaptive equipment, cerebral palsy, and spinal bifida programs. Elizabethtown offers a competitive salary and fringe benefit package and is an affirmative action/equal opportunity employer. Interested orthotists should send resumes to Mr. James Sweigart, CPO; Director, Orthopedic Appliance Department; Elizabethtown Hospital for Children and Youth; Elizabethtown, Pennsylvania 17022.

C.P.O. or C.O. with prosthetic experience desired for new progressive practice. Excellent salary and benefits with possibility of business partnership. Send resume to AOPA Box 68202, 717 Pendleton Street, Alexandria, VA 22314.

Chief, Prosthetic & Orthotic Lab—Major South Florida teaching hospital has an opening for a Chief Prosthetist/Orthotist to manage a modern, progressive P & O Lab within a 100-bed Rehab Center. Certification required; Bachelors degree in a health related field and experience preferred. Excellent salary and benefits package. Send detailed resume to: Jay Flynn, Employment Office, Park Plaza LL 301.

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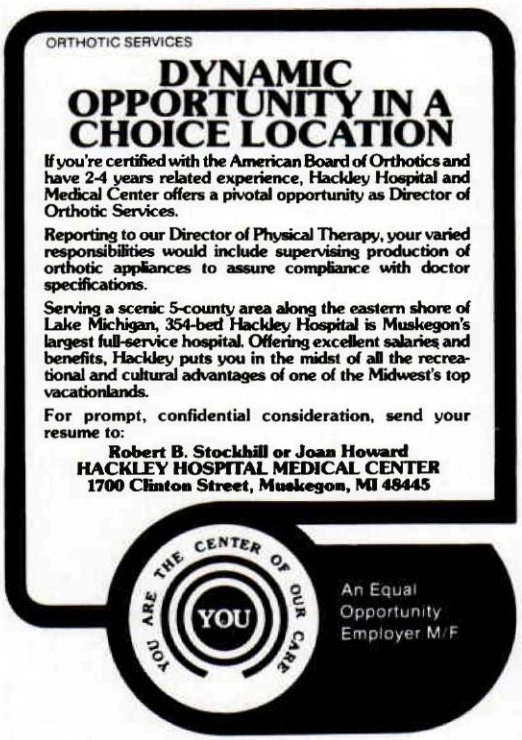

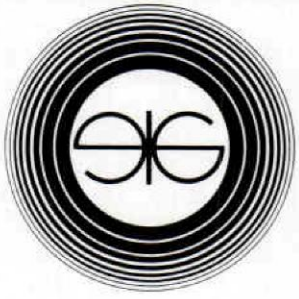

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Prosthetist/Orthotist—Lincoln Institute of Health Sciences, a College of Advanced Education situated in Melbourne, requires a Prosthetist/Orthotist. The primary responsibilities of the position include teaching prosthetics and/or orthotics in undergraduate and continuing education courses, and involvement in clinical programmes. Applicants should have clinical and/or practical experience in prosthetics and/or orthotics and preferably will have completed appropriate courses of training. Teaching experience would be an advantage. The successful appointee is expected to commence as early as possible in 1983. The appointment will be offered either on a continuing or limited term basis.

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Third Southern California Course on the "Current Concepts — the Diagnosis and Care of the Patient with Neuromuscular Disorder" will be held at Rancho Los Amigos Hospital, Downey, California, November 12–13, 1982. For further information, please contact: John Hsu, M.D., Children's Orthopaedic Service, Rancho Los Amigos Hospital, 7601 E. Imperial Highway, Downey, Ca., 90242, USA.

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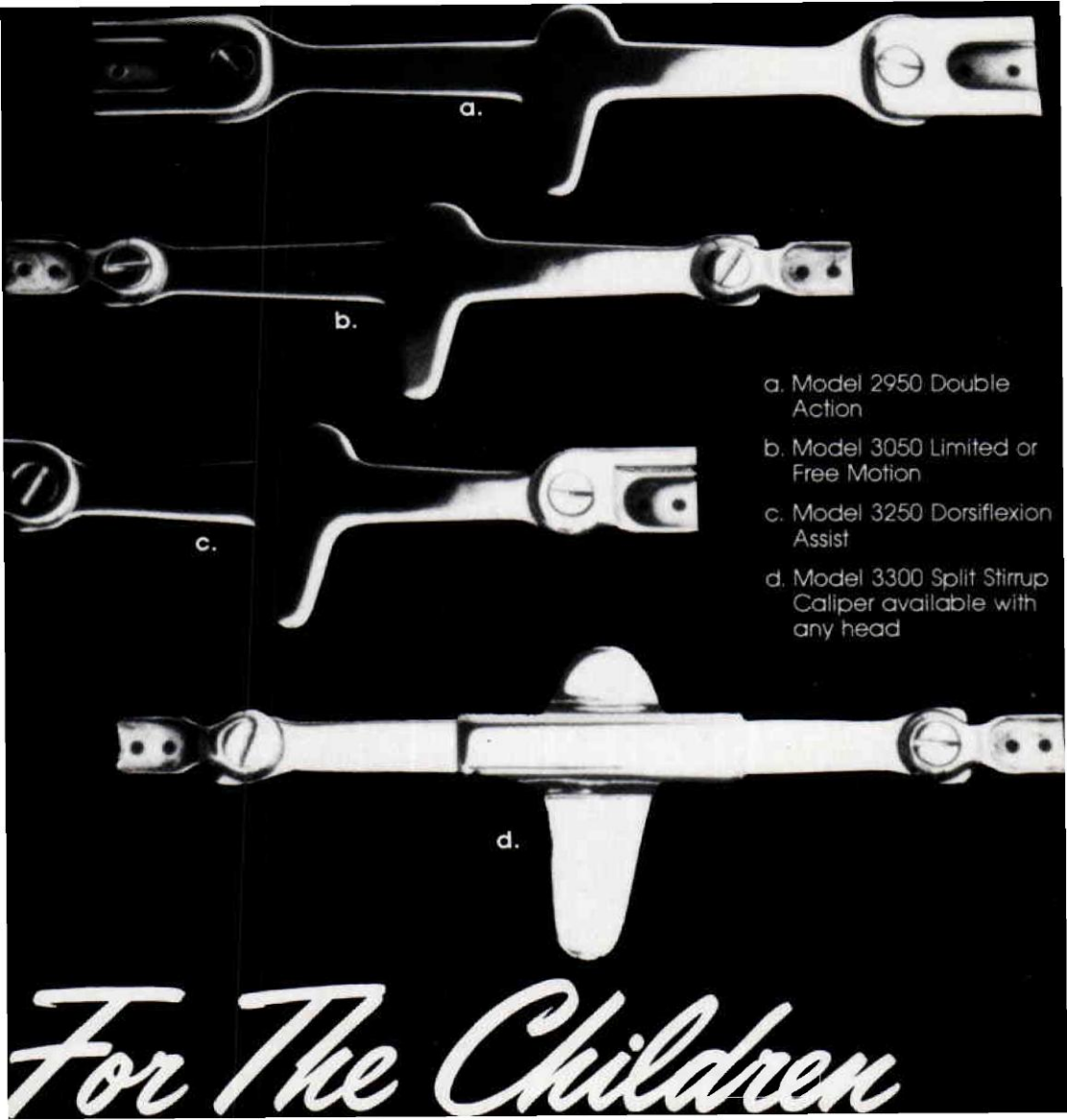

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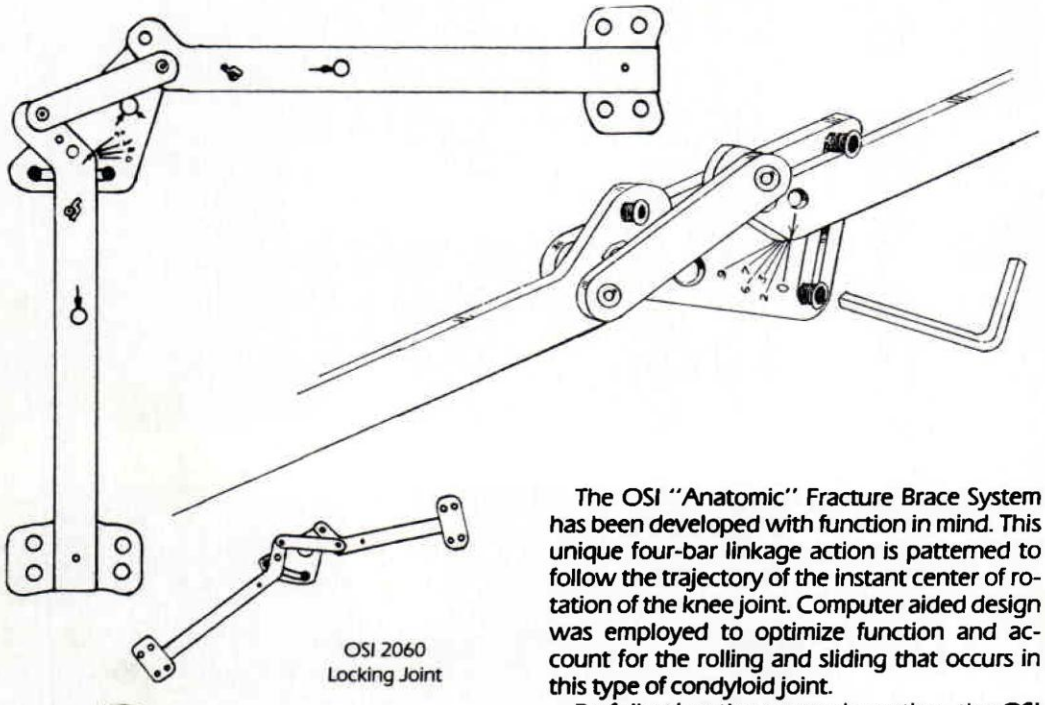

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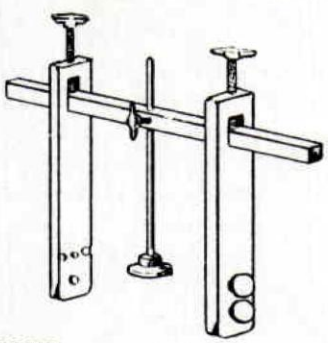

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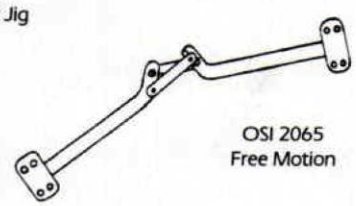

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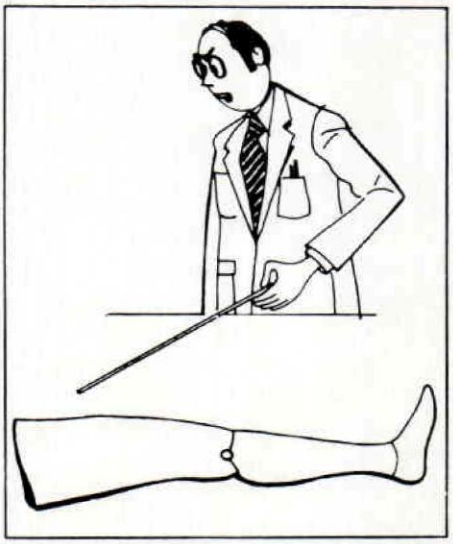

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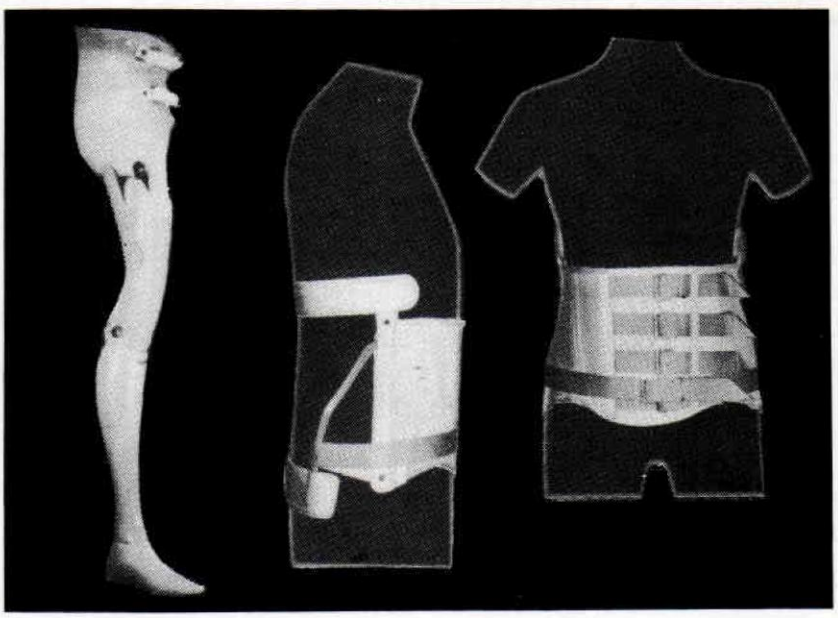

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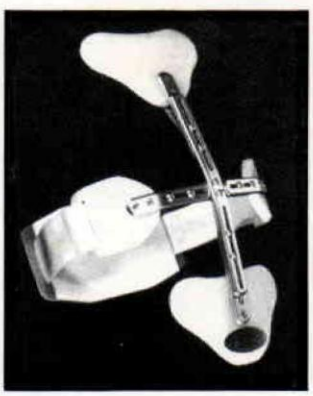

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