

MARCH 1975

Orthotics and Prosthetics



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Orthotics and Prosthetics is issued in March, June, September and December. Subscription price, payable in advance, is \$10.00 a year in the U.S. and Canada. Rate elsewhere is \$11.00 a year. Single issues, \$3.00 each. Publication does not constitute official endorsement of opinions presented in articles. The Journal is the official organ of the publisher, The American Orthotic and Prosthetic Association in collaboration with the American Academy of Orthotists and Prosthetists, and serves as the U.S. organ for Interbor. All correspondence should be addressed to: Editor: **Orthotics and Prosthetics**, 1444 N St., N.W., Washington, D.C. 20005. Telephone, Area Code 202, 234-8400.

Orthotics and Prosthetics is indexed by Current Contents/Clinical Practice.



(US ISSN 0030-5928)

Volume 29, No. 1

March 1975

CONTENTS

EDITORIAL	1
A DYNAMIC ORTHOTICS SYSTEM TO ASSIST PELVIC EXTENSION: A PRELIMINARY REPORT <i>John Glancy and Richard E. Lindseth</i>	3
SOME NONSTANDARD PROSTHESES FOR CHILDREN <i>Yoshio Setoguchi</i>	11
THE USE OF POLYPROPYLENE IN AN UPPER-LIMB ORTHOSIS <i>Gary Zamosky</i>	19
LOWER-LIMB MODULAR PROSTHESES: A STATUS REPORT <i>A. Bennett Wilson, Jr.</i>	23
A MODIFIED PROSTHETIC FOOT FOR PILOTS <i>Harry N. Hughes and Gene Helmuth</i>	33
A PATELLAR-TENDON-BEARING ORTHOSIS <i>Bert R. Titus</i>	35
NEW PUBLICATIONS	41
<i>Prostheses and Rehabilitation After Arm Amputation—Bender</i>	
<i>The Control of Upper-Extremity Prostheses & Orthoses—Herberts et al.</i>	
<i>Manual on Dynamic Hand Splinting with Thermoplastic Materials—Malick</i>	
METRICATION	43



Index to Advertisers

ALDEN SHOE CO.	XVII
BECKER ORTHOPEDIC APPLIANCE CO.	XVIII
CAMP INTERNATIONAL	IX, X
C. D. DENISON CO.	IV
IRVING DREW CORP.	XVIII
FILLAUER ORTHOPEDIC	V
FIWAY MANUFACTURING CO.	VI
FLORIDA BRACE CORP.	VII
JOHNSON & JOHNSON	VIII
JAMES R. KENDRICK CO.	XXI
KINGSLEY MANUFACTURING CO.	XI
KNIT-RITE, INC.	XII
PEL SUPPLY CO.	XX
STRIDE RITE CORP.	XV
SUTTON SHOE MACHINERY CO.	XVI
TRUFORM ANATOMICAL SUPPORTS	XIX
UNITED STATES MFG. CO.	XIII, XIV
Classified Advertisements	XXII

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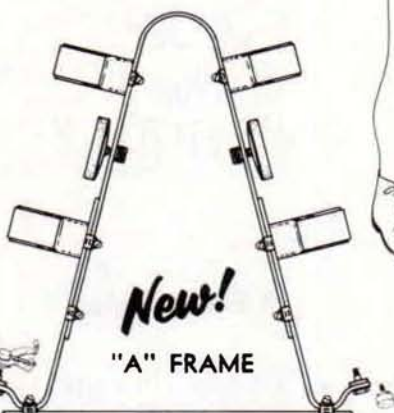
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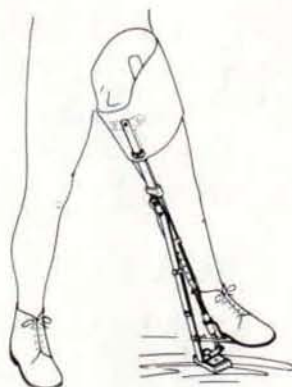
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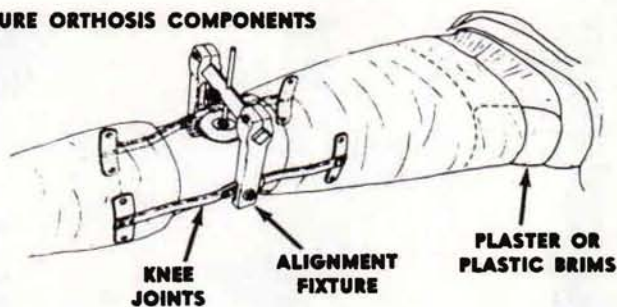


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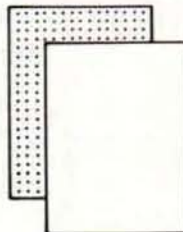
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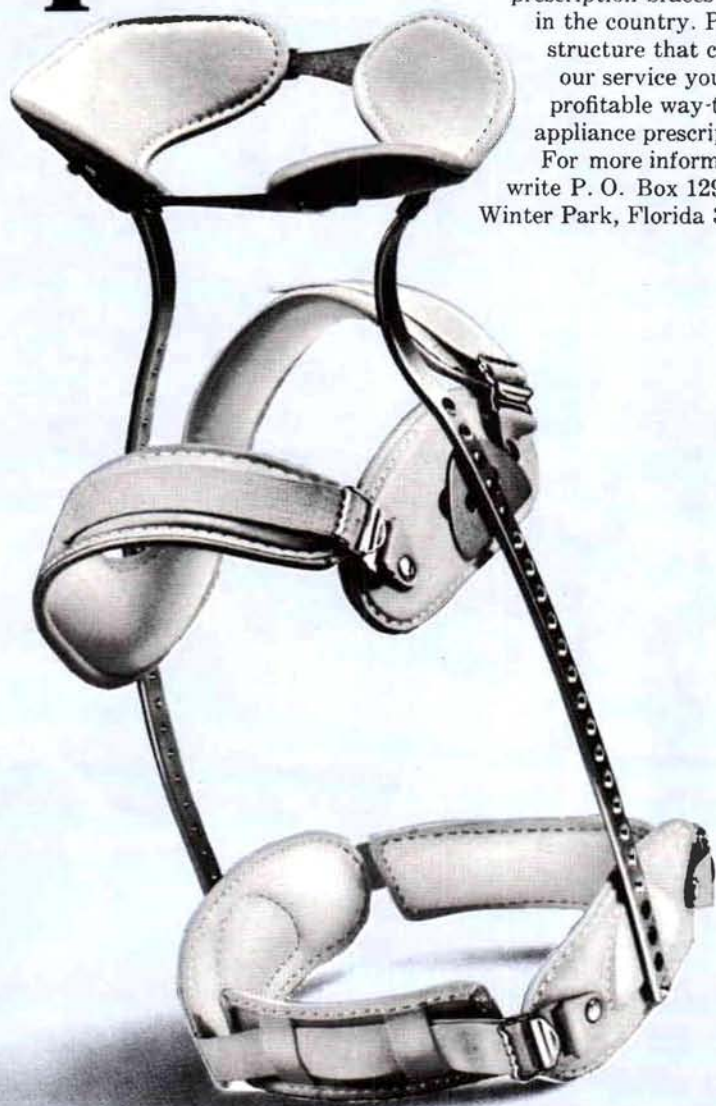
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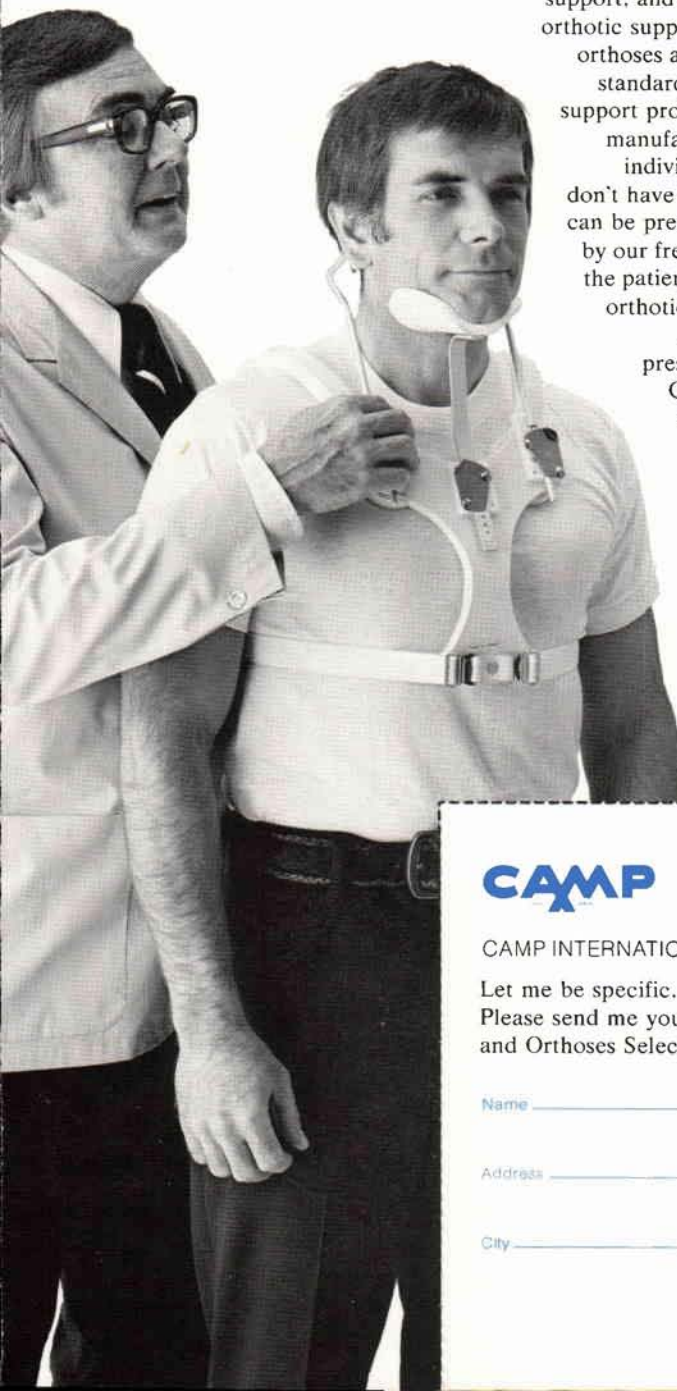
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A CHANGE AT THE HELM

As continuing readers of "Orthotics and Prosthetics" have likely noticed, it has experienced a number of major changes in recent years. Most obviously, the cover and general appearance of our journal and the type of articles which it publishes have improved significantly. These have moved "Orthotics and Prosthetics" into the mainstream of paramedical publications in the U.S. and the world at large.

Less obvious are the personnel shifts which publications like this one undergo from time to time. In 1972, A. Bennett Wilson, Jr. became Technical Editor. Actually, in a larger sense, he assumed full responsibility for the scientific content of the journal, and under his administration its stature improved. In recognition of his efforts, Mr. Wilson was made Editor in 1973.

Also in 1972, the American Orthotic and Prosthetic Association, publishers of "Orthotics and Prosthetics," established an Editorial Board whose purpose was to work with the editor in reviewing prospective articles for publication, resolving disputes among competing authors, and otherwise advising the editor on the tone and direction of the journal. This concept, followed by most other scientific and medical publications, proved successful.

A large measure of that success properly should be credited to the first Chairman of the Editorial Board, Bert R. Titus, C.P.O. During 1972-74 Mr. Titus, who is associated with the Duke University Medical Center, Department of Prosthetics and Orthotics, provided sound leadership in steering the journal away from its appliance oriented format and in the direction of a professional publication for fostering better standards of care to the orthopedically disabled. Mr. Titus also sought to capitalize more directly on the scientific contributions of professional practitioners who were members of the American Academy of Orthotists and Prosthetists. Soon after its formation in 1970, the Academy was invited to assist in monitoring the technical contents of AOPA's journal, a task which it has performed well. The next major step was to involve the Academy on the Board of Editors, and this was successfully accomplished in 1974.

Succeeding Mr. Titus as Editorial Board Chairman for 1975 will be Mr. Alvin Muilenburg, C.P.O., a name not unfamiliar to orthotists, prosthetists, and medical practitioners throughout the U.S. A Past President of AOPA in 1967-68, Mr. Muilenburg, too, has been very active in both the professional and business aspects of orthotics and prosthetics. He has presented a number of scientific papers before AOPA audiences and can be counted on to be a strong supporter of "O&P's" mission.

A new member of the Editorial Board this year will be Henry F. Gardner, Immediate Past President of the Academy, who fills the seat formerly occupied by H. Blair Hanger of Northwestern University. Mr. Gardner brings to his new responsibilities an extensive background in prosthetic services through his work with the Veterans Administration Prosthetics Center in New York City. He has a dynamic attitude toward progress, which will serve the journal well.

In closing, on behalf of AOPA and our Academy, we wish to sincerely thank Mr. Titus and Mr. Hanger for their services on the Editorial Board and their many helpful contributions. We also say "Welcome" to new Chairman Muilenburg and to new Board member Gardner. And to our editor Ben Wilson, we say "Well done," and extend our best wishes for continually improving a successful journal.

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A DYNAMIC ORTHOTICS SYSTEM TO ASSIST PELVIC EXTENSION: A PRELIMINARY REPORT^{1,2}

John Glancy, C.O.³

Richard E. Lindseth, M.D.³

Myelomeningocele has been described as "the most complex, treatable congenital anomaly consistent with life" (2). About five years ago, we began to focus our attention on musculoskeletal deformities that occur after birth. First priority was given to the child with lower lumbar paraplegia (L3 through L5) because these children have the potential to walk indoors and outdoors. Yet, few of these youngsters are still capable of independent ambulation when they reach adulthood. Throughout their growing years, a losing battle is waged against pathomechanical changes that affect the musculoskeletal system, resulting in flexion deformity of the hip and excessive lordosis of the lumbar spine.

Flexion deformity of the hip develops in the L3-level myelomeningocele child because all of the active muscles, iliopsoas, rectus femoris and sartorius, are located anterior to the hip joint and both of the normal hip extensors, the gluteus maximus and the hamstrings, are paralyzed, and thus there is no dynamic extending force to resist anterior rotation of the pelvis when the patient is in the upright position.

The youngster with a lesion at the L4 level, in addition, has innervated semitendinosus or gracilis muscles or both. Unfortunately, the presence of the medial hamstring is of little or no use in preventing hip flexion. When this muscle contracts to prevent unwanted hip flexion (pelvic forward rotation), it inadvertently flexes the

knee. The knee's bending is then opposed by the rectus femoris, which, in turn, flexes the hip, thus negating whatever extension force the semitendinosus may have on the pelvis.

Although at the L5 level of paraplegia the medial hamstrings are fully innervated, the gluteus maximus is paralyzed and the same muscle imbalance exists as is the case of the L4 paraplegic. However, although the stronger force of the hamstring muscles improves the anteroposterior balance across the hip joints and prevents the development of fixed hip flexion contractures, a dynamic or functional hip flexion posture is developed in order to accommodate for the excessive lordosis that is essential for these children to achieve balance.

Despite surgical procedures such as posterior transfer of the iliopsoas (which appears to prevent the deformity when the child is recumbent) to correct the flexion deformity, we have found such procedures to be inadequate in preventing hip flexion and excessive lordosis when the child stands. Therefore, we feel that an external orthosis is necessary to control unwanted hip flexion and excessive lordosis. Past experience with external orthoses of our own design as well as designs of others led us to the following conclusions:

1. Pelvic control can be maintained by passive "prepositioning" when normal range of motion of the hips and lumbar spine has been maintained prior to application of the orthoses.

2. Control of the pelvis cannot be achieved by external means unless there is a mechanical connection between the pelvic component and the thigh components of an orthosis.

3. Pelvic control for the "low-level" myelomeningocele child should be limited to rotation in the anteroposterior plane only. Rotation in the transverse and mediolateral planes is useful and should not be inhibited by an appliance.

¹This project is supported by a grant from the James Whitcomb Riley Memorial Association.

²Portions of this report were presented at the Workshop on Mobility Aids for Patients with Spina Bifida, held at the Ontario Crippled Children's Centre, Toronto, Canada, September 12-14, 1974. The Workshop was sponsored by the Committee on Prosthetics Research and Development of the Division of Medical Sciences, National Research Council.

³Indiana University/Purdue University at Indianapolis School of Medicine, 1100 West Michigan Street, Indianapolis, Indiana 46207.

4. No orthotics system, now known, is available for reduction or prevention of an increase in hip flexion contractures when they exist at the time the orthosis is applied. (This is true as to both the short and the long term.)

5. No orthoses currently available provide a dynamic force that creates an extension moment about the hip axis, which is essential in the prevention of hip flexion contractures in the L3 and below myelomeningocele child.

6. An orthotics system designed to provide an extension moment about the hip joints should allow motion in those planes in which the child has control so that the overly confining features of contemporary orthoses can be avoided.

7. The orthotics system must offer stability

and guidance to motion that is the product of muscle activity. It must not inhibit such activity.

Our latest prototype orthotics system, evolved from five earlier designs, is shown in Figure 1. We believe that this type of system meets the needs of the growing, low-level myelomeningocele patient by providing freedom of motion for the child's immediate use while, at the same time, providing protection from hip flexion deformity and excessive lordosis throughout the growing years. The system consists of:

1. A polypropylene thoracic "girdle" with a movable pelvic section. The adjustable pelvic section of the thoracic unit permits passive control of the lumbar region by prepositioning and maintaining an optimal relationship between pel-

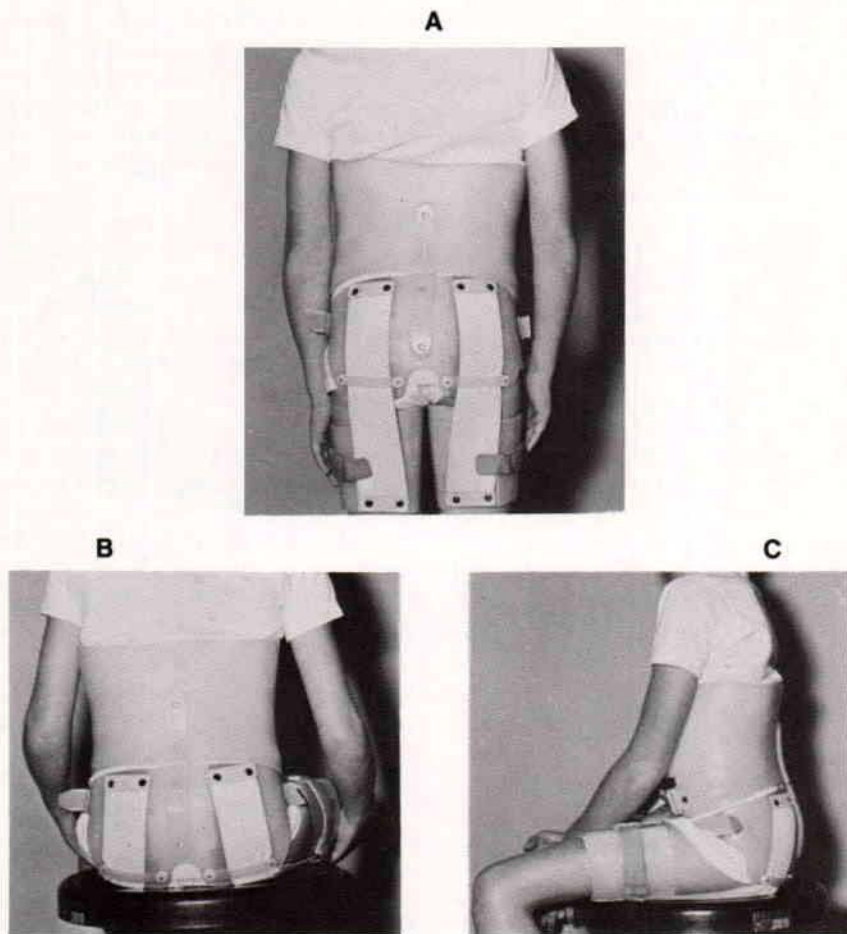


Fig. 1. The Dynamic Orthotic System to Assist Pelvic Extension: A. Posterior detail in standing position. B. Posterior view in the sitting position. C. Lateral view in the sitting position.

vis and thorax. Both sections are lined with $\frac{1}{4}$ -in. Plastazote.

2. Bilateral, quadrilateral-type polypropylene thigh cuffs lined with $\frac{1}{8}$ -in. thick Plastazote.

3. Bilateral woven elastic panels that provide the force that produces the extensor moment about the hip joints.

At their proximal ends, the elastic panels are riveted just below the upper edge of the movable pelvic section of the thoracopelvic unit. The distal ends of the two woven elastic panels are riveted to the posterior side of the quadrilateral cuffs. In this arrangement, the force is transferred from one side to the other by the polypropylene pelvic section. The total weight of the orthotics system is 655 g for the average five-year-old child. The thoracopelvic unit weighs 440 g, and each of the quadrilateral thigh cuffs, including its elastic panel, weighs 107.5 g.

DISCUSSION

The development of plastics has opened up new design possibilities for orthotics devices within the past few years. Major advantages include reduction in weight, better cosmesis, and a more intimate fit that permits an efficient application of the three-point pressure system to the trunk. For example, a polypropylene body jacket, lined with Plastazote and Silastic and utilizing the Milwaukee brace technique for the waist and abdomen (1,5) increase intra-abdominal pressure (6), makes it possible for the patient with insensitive skin to receive excellent support day after day, free from pressure sores. These materials are waterproof and impervious to body excretions (Fig. 2).

We have taken advantage of the characteristics of polypropylene to fatigue to build into spinal jackets an adjustable pelvic panel that permits adjustment to the wearer's optimum position of balance in the anteroposterior plane and also gives the wearer freedom to rotate his pelvis in the transverse plane and thus to walk with a gluteus medius sway. These features are obtained without loss to the efficiency of the three-point pressure system necessary to prevent excessive lordosis. The pivotable pelvic band of the Williams brace (7) permits "prepositioning" of the lumbar curve. It is this feature of the Williams brace that we have incorporated into our thoracopelvic design in the form of a molded adjustable pelvic section (Fig. 3).

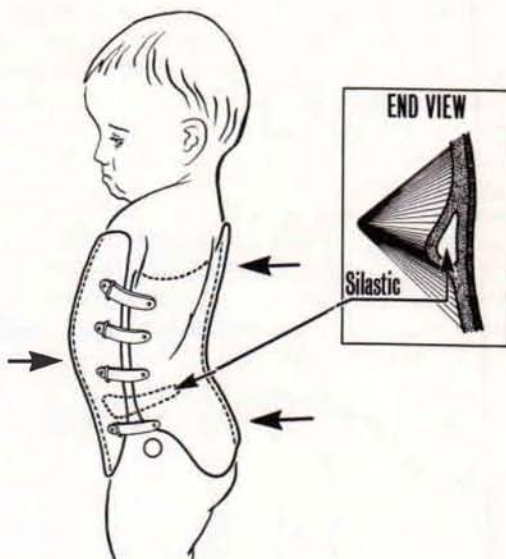


Fig. 2. Earlier polypropylene thoracopelvic unit showing incorporation of Silastic for insensitive skin. Note the natural contour of the outer surface.

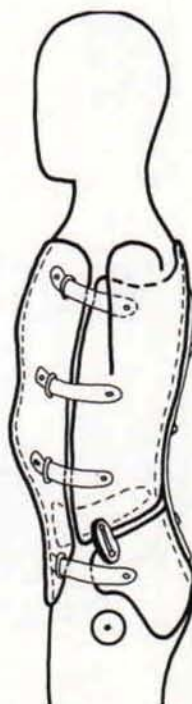


Fig. 3. Present model of a polypropylene thoracopelvic unit showing molded, adjustable pelvic section.

In order to show the importance of pelvic control to the overall biomechanical problems of the L3, 4, and 5 myelomeningocele child as they relate to balance and gait, it is necessary to refer to previous work (4,6) done with these children related to the foot/ankle complex and knee (Fig. 4). The beneficial effect of the SA braces to the myelomeningocele child's foot/ankle complex and knees is evident and gratifying, but it is equally evident that no benefit toward pelvic control could be attributed to them. On the contrary, an uneasy feeling persists that serious further deformity to the lumbar spine may be a direct

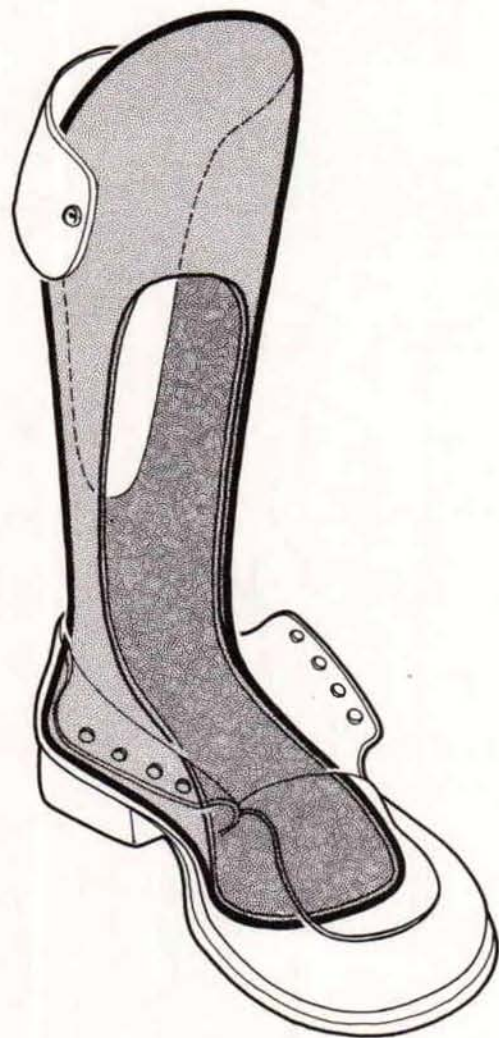


Fig. 4. Polypropylene solid-ankle (SA) below-knee orthosis with Plastazote lining.

result of long-term use of any type of bracing that stabilizes the ankle and knee joints to facilitate more efficient ambulation, unless a means of controlling the pelvis can be devised (Fig. 5) since the

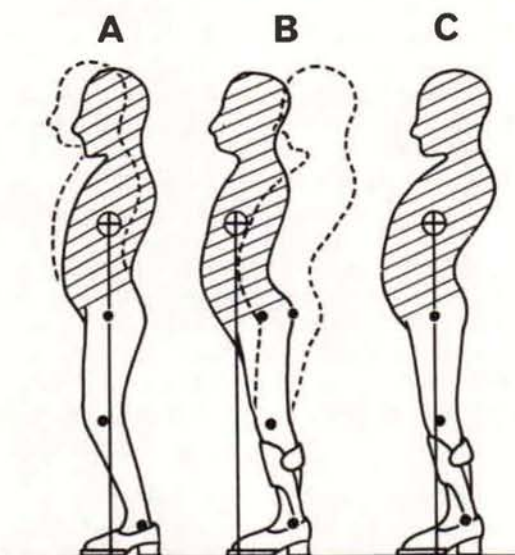


Fig 5. Effects of stabilizing the foot/ankle complex and knee upon AP balance of the trunk when hip flexion contractures are present. Solid-ankle orthosis is shown. However, any type of bracing produces similar effects: A. Solid-lined figure shows posture necessary to balance without bracing. B. That knee flexion alone cannot accommodate for hip flexion contractures is demonstrated. It can be seen that forward flexing of the thorax to restore balance, without active trunk extensors, is not a practical alternative. C The only anatomic option left is to increase lordosis. (⊕ represents C.G. of trunk only.)

mere act of standing without bracing also leads to deformity of the lumbar spine (Fig. 6).

Protection for the hip is provided by our design by permitting a "preload," or extension force, to be applied by introducing slight stretch to the elastic panels when the unit is placed on the patient. This provides a mild stretching force upon the hip flexors and quadriceps muscles to prevent the gradual contracture of the muscles when the child is standing at ease. Furthermore, whenever the wearer flexes either his leg or his pelvis, the force provided by the woven elastic panels is increased which, in turn, increases the extensor moment about the hip joints. This increase in force is proportionate to angular changes, thereby aiding control of motions about the hip joint.

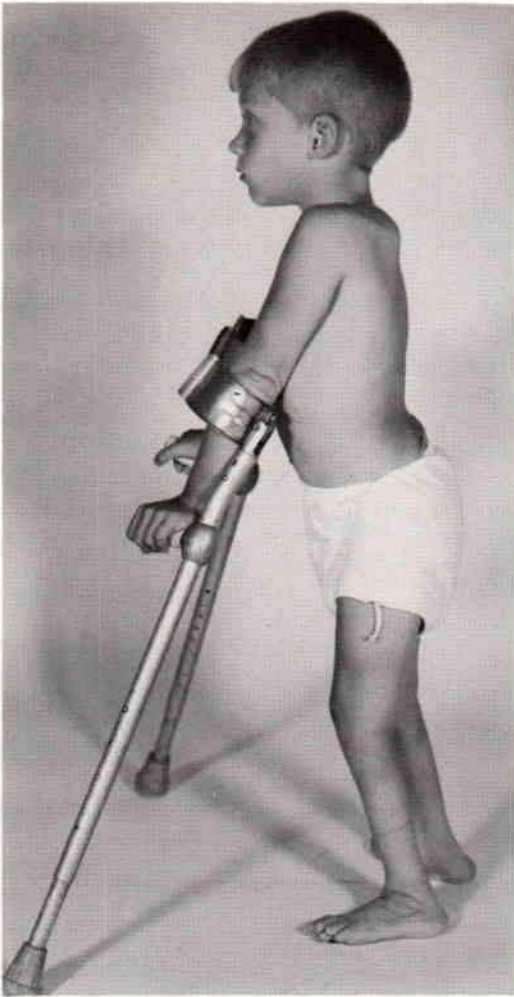


Fig 6. Effects of standing, without bracing, upon the lumbar spine. Note the proportion of body weight being supported through the arms and crutches. Without the crutches, the child must assume an excessive lordotic posture in order to bring and maintain his trunk over a much smaller base of support.

A typical load-elongation curve for a 10-in. long piece of the elastic panel is illustrated in Figure 7. When the leg is in 90 deg. of flexion, the elastic panel will stretch from a minimum preloaded condition to an extension of approximately 4 in. This will create a force of nearly 6 lb, according to Figure 8, acting on the lower part of the thigh cuff. The moment about the hip is

$$M_{\text{hip}} = F r$$

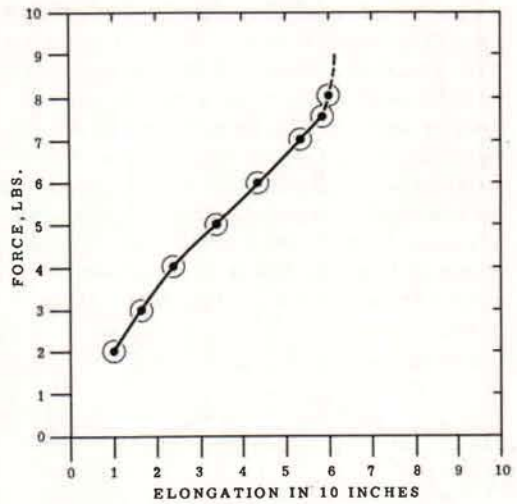


Fig. 7. Typical load-elongation curve for elastic panel. The one-inch reading represents the preload.

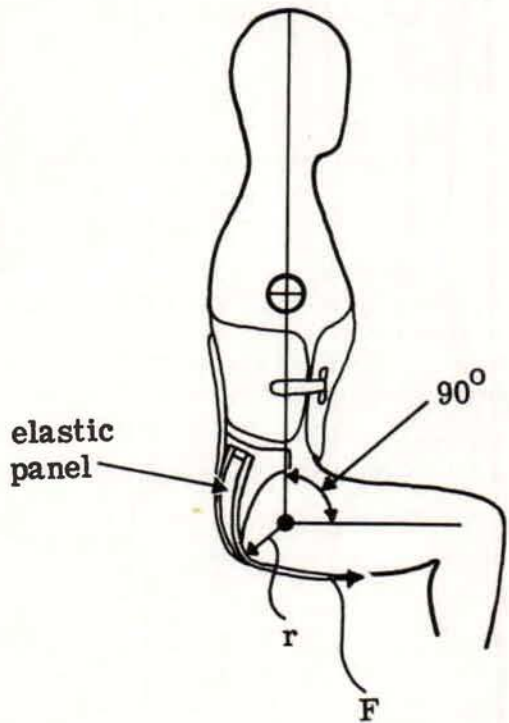


Fig. 8. Schematic illustration showing moment at the hip created by the force "F" in the elastic panel.

where "r" is the radial distance between hip axis and elastic panel. When "r" is 2 1/2 in., the moment about the hip is 15 in. lb. Such a moment provides a steady influence by providing resistance to the active anterior muscles that cross the hip joints. However, the magnitude of the moment is very small compared to the moment due to the weight of the trunk. To illustrate this, consider a subject bending over at the waist, as shown in Figure 9. When the value of the moment, M_R , is that required to maintain equilibrium,

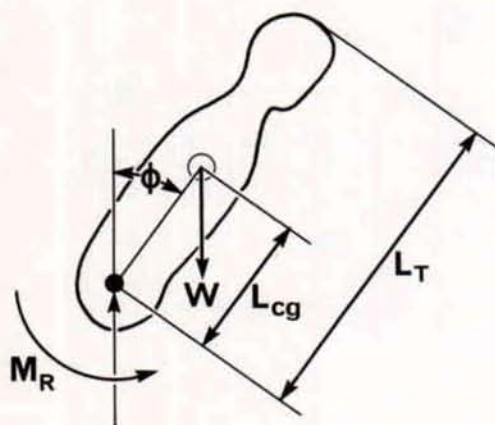


Fig. 9. Free body diagram of trunk.

$$M_R = F \cdot r = W L_{cg} \sin \phi \quad (2)$$

From W. T. Dempster (3): $W \approx 0.5 \cdot W_B$ where W_B is the total body weight and $L_{cg} \approx 0.4 \cdot L_T$ for adults. Assuming these ratios are also true for children, the solution to equation 2 becomes:

$$M_R = (0.4)(0.5)W_B L_T \sin \phi \quad (4)$$

As an example, for a subject with

$$W_B = .32 \text{ lbs}$$

$$L_T = 19 \text{ lbs}$$

gives

$$\begin{aligned} M_R &= (.2)(19)(32) \sin \phi \\ &= 121.6 \sin \phi \end{aligned}$$

As the subject bends, the angle ϕ increases and the restoring moment necessary to maintain equilibrium increases. For this example:

ϕ	M_R (inch pounds)
5°	10.57
10°	20.93
15°	31.37
30°	60.80

which far exceed the moment created by the elastic panel, except in the near vertical positions.

Similarly, the force from the elastic panel is easily overcome by the muscle action, because it is small compared to the forces that the muscles can generate. It appears that the magnitude of extensor moment provided by the elastic is sufficient to check involuntary forward flexion (sway in an anterior direction) of the trunk in the upright position, and thus contributes substantially to the anteroposterior balance in the lower lumbar paraplegic child. We believe that such a contribution is possible with our present design.

SUMMARY

An orthotics system has been developed which provides a dynamic extensor moment to the pelvis. Its purpose is twofold: 1) to prevent the occurrence of hip contractures, excessive lumbar lordosis, and knee contractures that predictably develop in the L3, 4, and 5 level myelomeningocele child, and 2) to improve gait and make physical activities in general easier by making the action of an incremental extensor moment to the pelvis reciprocal. The background and rationale to the system's development is outlined in this preliminary report.

The authors wish to express their gratitude to B. M. Hillberry, Ph.D., of Purdue University School of Mechanical Engineering, for his force computations and free-body analysis which have contributed so much to our understanding.

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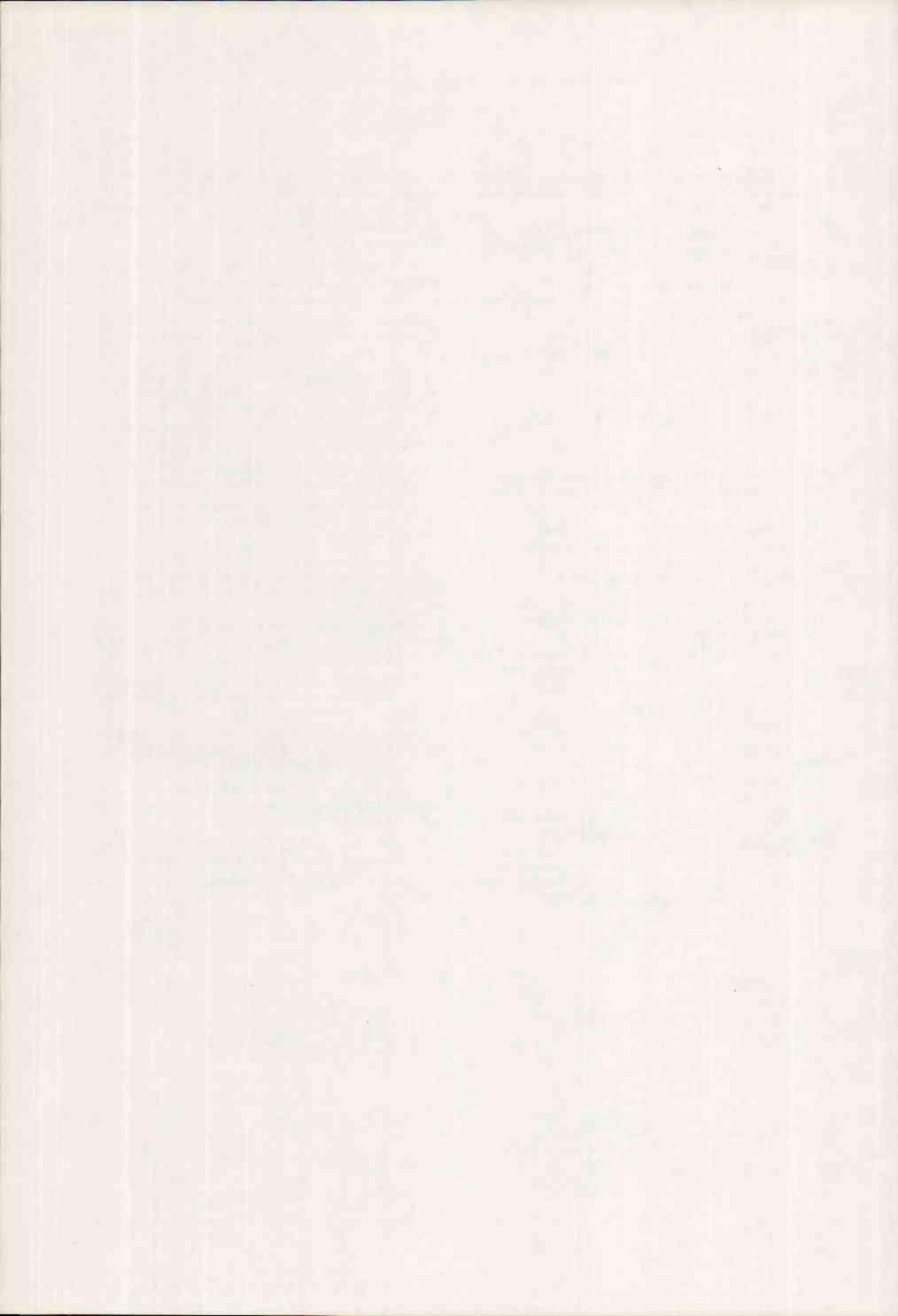
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SOME NONSTANDARD PROSTHESES FOR CHILDREN

Yoshio Setoguchi, M.D.¹

At the UCLA Child Amputee Prosthetics Project (CAPP), amputee patients are seen from birth until 21 years of age for complete rehabilitation management. Because of the restriction relative to the age of patients, the Project does not see as many patients who have acquired amputations as those who are born with limb deficiencies, but prosthetics devices developed for the more common congenital limb deficiencies involve fitting techniques that are applicable to patients with acquired amputations.

Three major areas of changes in prosthetics fitting are discussed here: modular upper-limb prostheses, partial-foot prostheses, and the flexible-wall socket with new applications.

MODULAR PROSTHESES

A review of the history of prosthetics research and development of the past few years shows that considerable emphasis has been placed on the concept of modular prostheses. It is also apparent that interest at most major prosthetics centers is concentrated on development of "cosmetic" prostheses that retain the basic functional characteristics of the previous "standard prostheses." The Child Amputee Prosthetics Project has also undertaken development in the area of "modular" prostheses or, as we prefer to call them, "unitized" prostheses. Since the major case load of both congenital and acquired amputees at our clinic consists primarily of children who have upper-limb involvement, it was natural for us to focus on this group of patients. In 1967 a study was undertaken for providing improved function for this group of infant children while providing better cosmesis.

When we analyzed the standard fitting techniques used for infant amputees (Fig. 1), it was obvious that fitting these children with

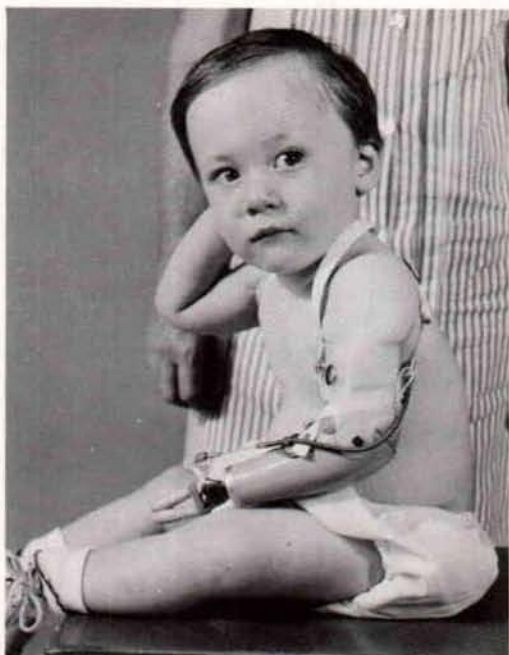


Fig. 1. Standard prosthesis and harness for a child with a below-elbow amputation.

scaled-down adult prostheses did not take into consideration the fact that the body proportions of infants are not miniaturizations of adults, and the activities and movement patterns were not the same as those of an adult. Adults may bend, lift, and reach, but never go through the contortions of an infant. It was felt that if the infant functions differently from the adult, and his body proportions are also different, his prosthetics requirements must be different.

A considerable amount of data was gathered relative to the growth, development, and activities of children. These observations, together with specific functional criteria, suggested a radically new approach in prosthetics design. The standard prosthesis is constructed on an exoskeletal concept, in which the supporting framework is the external shell and the interior is

¹Medical Director of the Child Amputee Prosthetics Project, University of California, Los Angeles. The Child Amputee Prosthetics Project is supported by a grant from the Department of Health, Education, and Welfare, Maternal and Child Health Service.

hollow to the point where the terminal device is attached. In contrast, the unitized prosthesis is hollow only in the socket area which fits over the patient's stump, and the remainder is constructed on an endoskeletal concept in which the basic structure is internal and other materials fill the space around it to complete the arm (Fig. 2).

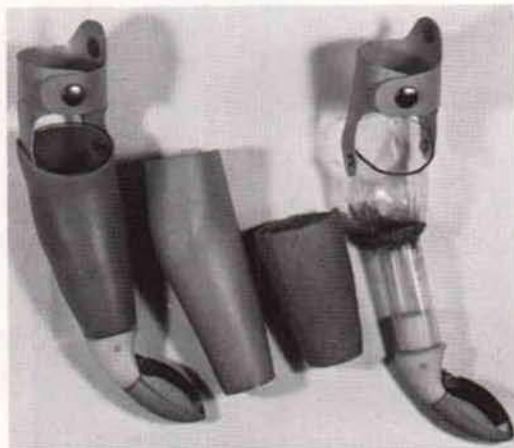


Fig. 2. Experimental modular prosthesis.

Certain functional criteria were considered in the design and development of the prostheses. First and foremost was that for the child to actively incorporate the prosthesis into his use pattern, the prosthesis had to be light. Furthermore, the soft forearm was used for the function it would provide rather than for improved cosmesis.

Some of the major needs of infant below-elbow amputees are the ability to use the prosthesis for stabilizing and supporting objects, for clasping and also for crawling, for stabilizing himself, and for pushing himself up from the floor. To provide the infant with a prosthesis which would permit him to do these things, a soft-textured forearm with a frictional surface that would also enable him to hold a variety of shapes and textured objects in his arms was incorporated in the design. Such a prosthesis, then, would give the child much more function than he had enjoyed previously.

With this criteria in terms of materials and function as a basis for the development of a new prosthesis, a modular prosthesis was designed.

The basic socket configuration remains essentially unchanged, but because we were considering an entirely new prosthetic concept, various transparent thermoplastic materials, such as Lexan, were tried. The purpose of the clear socket was to allow the prosthetist and therapist to observe the actual stump-socket relationship as the patient uses the arm. The transparent socket concept remains a desirable feature, but because of problems with production (cost, as well as fabrication) and lack of strength once fabrication was accomplished, the clear plastic socket has not yet been successful clinically (Fig. 2).²

Because of the difficulty experienced with the clear plastics, we reverted to the polyester-laminate socket. To this socket is attached a flared cup of polyvinylchloride (PVC) or other thermoplastic material to fit over the end of the socket, to provide a receptacle approximately 1 in. long to receive a central tubing of PVC or Lexan (Figs. 3, 4, and 5). The tubing is cut to the

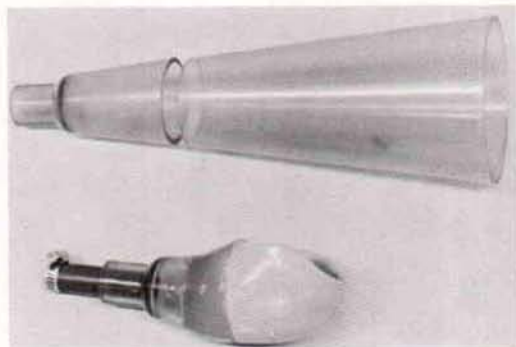


Fig. 3. Use of a flared plastic cup for attaching forearm section to the socket.

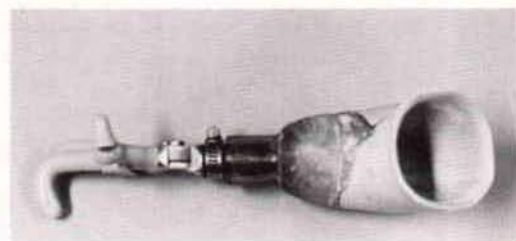


Fig. 4. Complete assembly using the flared tube technique.

²This work was carried out before the successful application of vacuum-forming techniques to prosthetics and orthotics. *Ed.*

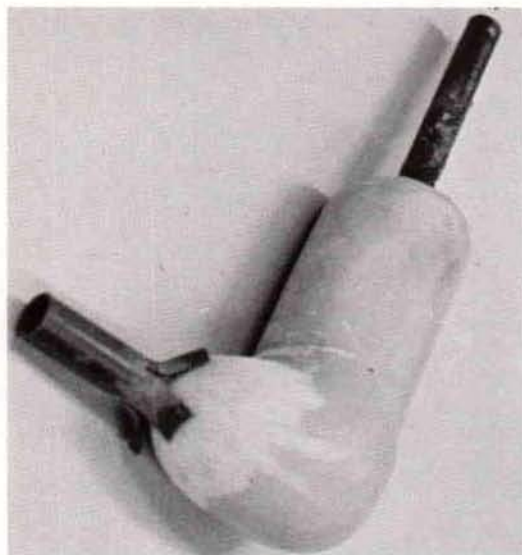


Fig. 5. Attachment of PVC forearm to socket. The PVC tube is threaded at the distal end to receive a terminal device.

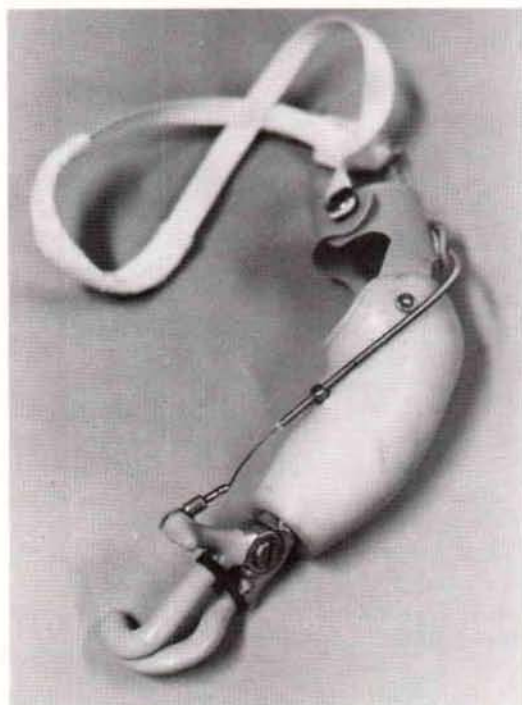


Fig. 6. Completed modular prosthesis for below-elbow case. Note the plastic cuff.

appropriate length and a ring clamp is attached to provide friction when the terminal device is attached. The forearm is then covered by a light-weight polyurethane foam. This material is shaped and covered by a PVC sleeve. To this is applied a standard ortholen or polyethylene full cuff (Fig. 6).

The unitized prosthesis is constructed largely of plastics that have sufficient strength and durability to survive approximately three years of wear. Plastics were used because of additional advantages such as light in weight, soft appearance, and minimal maintenance. Furthermore, the expense is less when components are produced in fairly large quantities.

For infant below-elbow amputees fitted with a nonactivated terminal device, a harness that would provide direct counterforce to supply vertical suspension of the prosthesis was necessary. The harness must not fall off the shoulder, cause undue axillary pressure, or require critical fit in the chest area. In addition, it should allow maximum shoulder motion (Fig. 7). To meet these requirements a modified chest-strap harness was designed. The harness consists of a chest strap, an over-the-shoulder strap, axillary cord, and swivel tab. The over-the-shoulder strap and axillary cord together form a suspension loop which fits on the amputated side. The swivel tab slides on the axillary cord permitting free arm movement. The swivel tab snaps to the cuff and is the only attachment point to the harness for suspension.

The end result of this development was a modular prosthesis which provided the infant amputee, below-elbow type, with a prosthesis which was light in weight, had a soft appearance, with a frictional sleeve surface, and a much simplified but more effective harness that required minimal, if any, maintenance.

Our clinical trials have shown that the infant amputees readily accept this type of prosthesis and use it immediately in their daily activities. The functional advantages of the modular prosthesis are compatible with their needs as physical and mental development occurs.

Because of the excellent acceptance of the below-elbow type of modular prosthesis, the Child Amputee Prosthetics Project has also developed a unitized prosthesis for infants who have above-elbow and shoulder-disarticulation amputations. Again, the criteria of light in

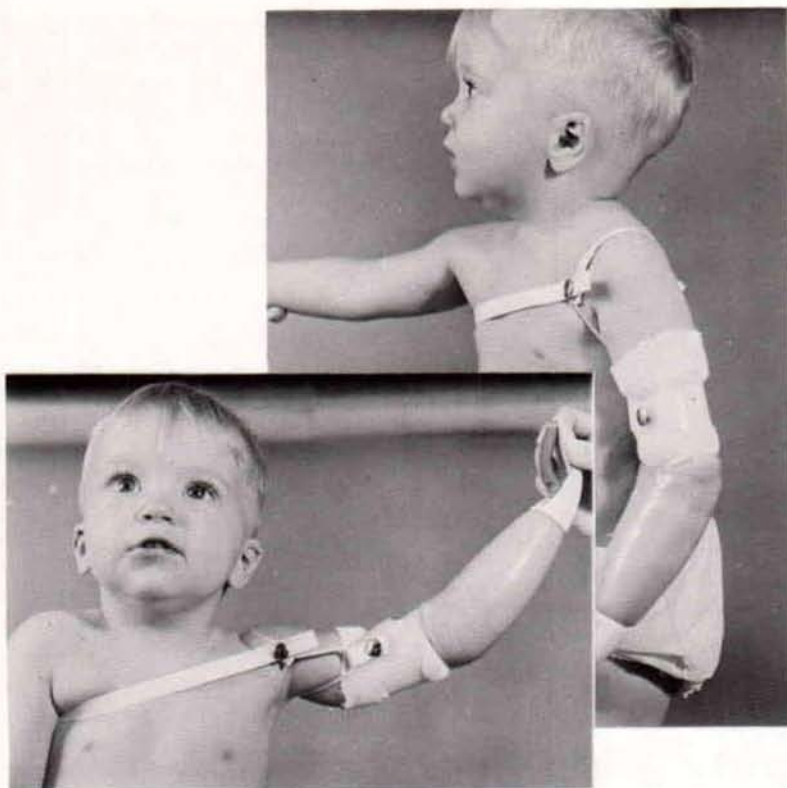


Fig. 7. Modified chest-strap harness for young below-elbow amputee.

weight, a soft appearance, and function for clasping and leaning were considered. The modular above-elbow and shoulder-disarticulation prostheses utilize a manually positioned, six-position elbow joint, three sizes of threaded PVC tubes, ring clamps, and a fixed 90 deg. angle (Fig. 6).

These parts can be combined in a variety of ways to attach the CAPP plastic wrist unit³ which is laminated to the socket for use as a turntable or a shoulder joint (Fig. 8). Preliminary results with this type of prosthesis show that the modular system is extremely versatile for the infant amputee who has a high-level amputation. Since there are no firm guidelines for prostheses segment lengths for young patients with extensive limb deficiencies, especially shoulder-disarticulation types, it is often necessary to estimate the lengths that will be needed. Once a conventional-type prosthesis is fabricated,

changes are not possible. With a modular type the therapist and prosthetist, after the prosthetist has made the socket, have an opportunity to work with the child using interchangeable endoskeletal arm segments prior to the time the prosthesis is covered (Figs. 9 and 10). This is very helpful because it is possible to find the best combination of segment lengths for the patient with a minimum cost.

Another advantage of the modular prosthesis for these youngsters is that when the child has grown linearly, or when functional needs change, it is possible to change segment lengths without providing an entire new prosthesis. These high-level modular prostheses seem to provide additional functional advantages. The prostheses are used for gross clasping, and some children use the soft textured cover to stabilize themselves in leaning and pushing. One side benefit from this type of prosthesis has been that the parents of children fitted with modular prostheses report a positive reaction to the general appearance of the prosthesis (Fig. 11).

³Manufactured by Hosmer/Dorrance, P.O. Box 37, Campbell, California 95008.

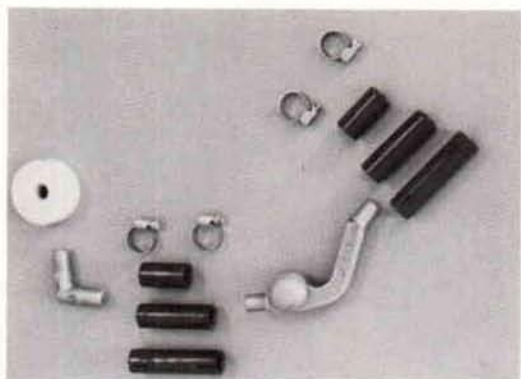


Fig. 8. Parts for assembly of modular prostheses for young above-elbow and shoulder-disarticulation amputees.

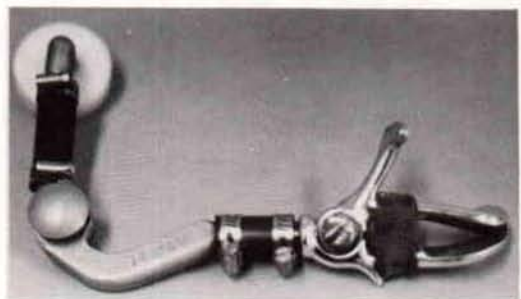


Fig. 9. Basic parts for the shoulder-disarticulation prosthesis assembled, but without socket and cosmetic treatment.

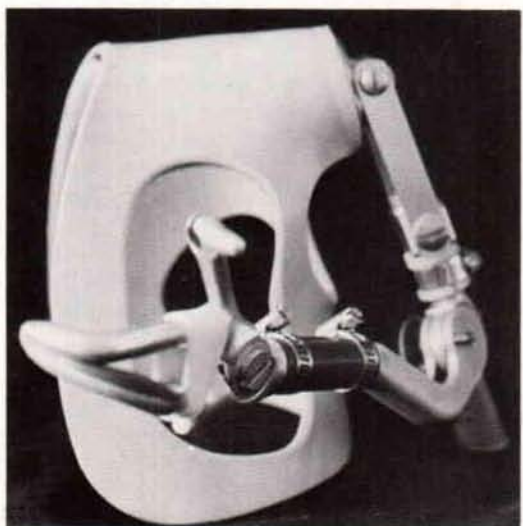


Fig. 10. Modular prosthesis for shoulder-disarticulation amputee before cosmetic treatment.



Fig. 11. A shoulder-disarticulation amputee with functional modular prosthesis.

PARTIAL-FOOT PROSTHESIS

Experience at UCLA with children who have partial-foot amputations indicates that some major problems besides those of normal wear are: outgrowing the prosthesis very frequently, mechanical breakdown, wear and tear on normal shoes, or a combination of all of these conditions. Children in almost all of their waking hours engage in strenuous activities, thus subjecting the partial-foot prosthesis to the most severe kinds of stress. It is not uncommon for these patients to require new prostheses in less than a year after the fitting, and this need places a considerable financial burden on either the parents or the funding agency. Therefore, a design study was undertaken to develop a partial-foot prosthesis which would reduce the costs of time and material and yet provide function and cosmesis equal to or superior to that of existing partial-foot prostheses.

The design criteria were: 1) fabrication should be by existing methods, utilizing materials normally available in prosthetics facilities, 2) fabrication time should be reduced by simplification of procedures, 3) function should be satisfactory, and 4) cosmesis should be improved as compared to hard-shelled or metal frame leather-type devices.

The prosthesis as developed is made of laminated RTV Silastic.⁴ Reinforcement is provided in several key areas to provide the strength necessary for optimal function. In the sole area a belting material as well as a strip of spring steel is added. Over the dorsum of the foot, and extending medially and laterally over the instep area, Dacron webbing is used for additional reinforcement (Fig. 12). Over this reinforcement a final



Fig. 12. Dacron tape in place on RTV Silastic laminate to provide reinforcement for prosthesis for amputation through the forefoot.

lamination is made using the same RTV Silastic material. In order to provide easier donning, a cut is made, usually on the medial side, and these two flaps are then closed using either a "zipper" or lacing. A stocking is worn over the prosthesis.

This fitting has provided excellent function (Fig. 13) for a variety of partial-foot patients, including transtarsal, Lisfranc type, and Chopart amputations. Both function and cosmesis appear to be considerably improved compared to conventional designs. Patients and parents have accepted this technique unanimously and positively. An improved gait pattern has also been evident in these patients.

FLEXIBLE-WALL SOCKET

Another area of major concern, particularly for the acquired amputee, was the patients who have Syme's amputations. Two methods of fitting had become relatively standard, and although both

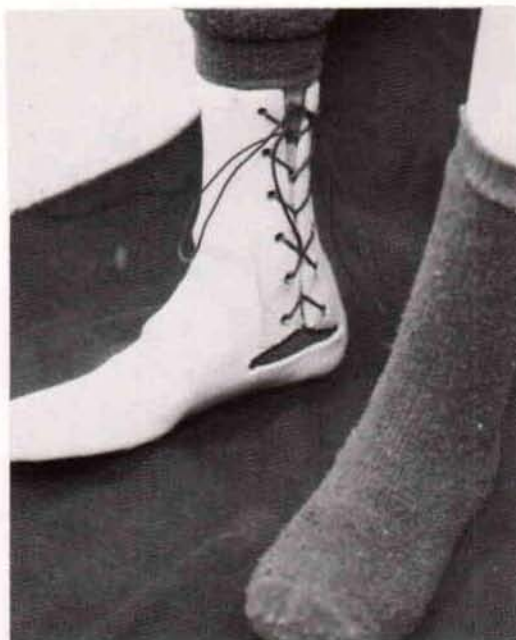


Fig. 13. Completed prosthesis for amputation through the forefoot.

were functionally adequate, appearance was poor. When the bulbous end of the Syme's amputation was not too different in circumference as compared to the narrow portion of the lower leg, a long below-knee prosthesis without a window was fabricated, and suspension provided by a suprapatellar cuff. Those patients who had a large bulbous distal end were fitted with a conventional Canadian-type Syme's prosthesis. Both designs leave much to be desired in terms of cosmesis. Particularly, the young girls were concerned about appearance. A new type of prosthesis was developed for this group of patients, using a technique that had been designed and developed previously for patients who have proximal femoral focal deficiencies (PFFD) and have been converted surgically to functional above-knee amputees (Fig. 14).

Since 1967 the Child Amputee Prosthetics Project has been fitting PFFD patients with a socket that has a flexible inner wall (Fig. 14). The flexible layers of the socket extend from the bottom of the socket to at least the level at which the bulbous end can pass through freely. This satisfies the function of the window in a standard

⁴Dow Corning Corporation, Midland, Michigan 48640.



Fig. 14. Socket for prosthesis for amputee with a proximal femoral focal deficiency, sectioned to show flexible inner liner.

Syme's prosthesis and provides room for expansion of the flexible wall as the bulbous end is inserted into the socket.

Once the stump is inserted fully, the flexible wall closes around it, giving a total-contact fit. Now it is possible to use the bulbous portion of the stump to provide suspension and thus make it possible to eliminate side joints and the pelvic belt. Because of the success of the flexible-wall socket when fitted to the PFFD patients, the concept of a flexible-wall socket was tried on a patient who had an acquired Syme's amputation (Figs. 15 and 16). The results have been very satisfactory. Suspension has been excellent and the patients feel more comfortable without the patellar cuff. Cosmesis is greatly enhanced by this design. This technique is not revolutionary or new in concept, and many other clinics have similar sockets, but we have made a modification in the fabrication of the inner lining. To avoid some



Fig. 15. Patient with Syme's amputation and prosthesis.

of the problems inherent with the use of RTV Silastic material, such as crumbling with aging and retention of odor from perspiration, we have used a completely elastic polyester resin for the fabrication of the inner lining. This material allows the parents or the child to clean the socket adequately without fear of damaging the inner lining.

This same bladder-wall or flexible-wall concept has now been applied to patients who have wrist-disarticulation stumps with bulbous ends. The concept of flexible-wall sockets that provide excellent suspension, while eliminating the need for windows or cuff or other type of harness suspension, has been a tremendous asset in providing functional prostheses with good cosmesis.



Fig. 16. Patient shown in Figure 15 in action.

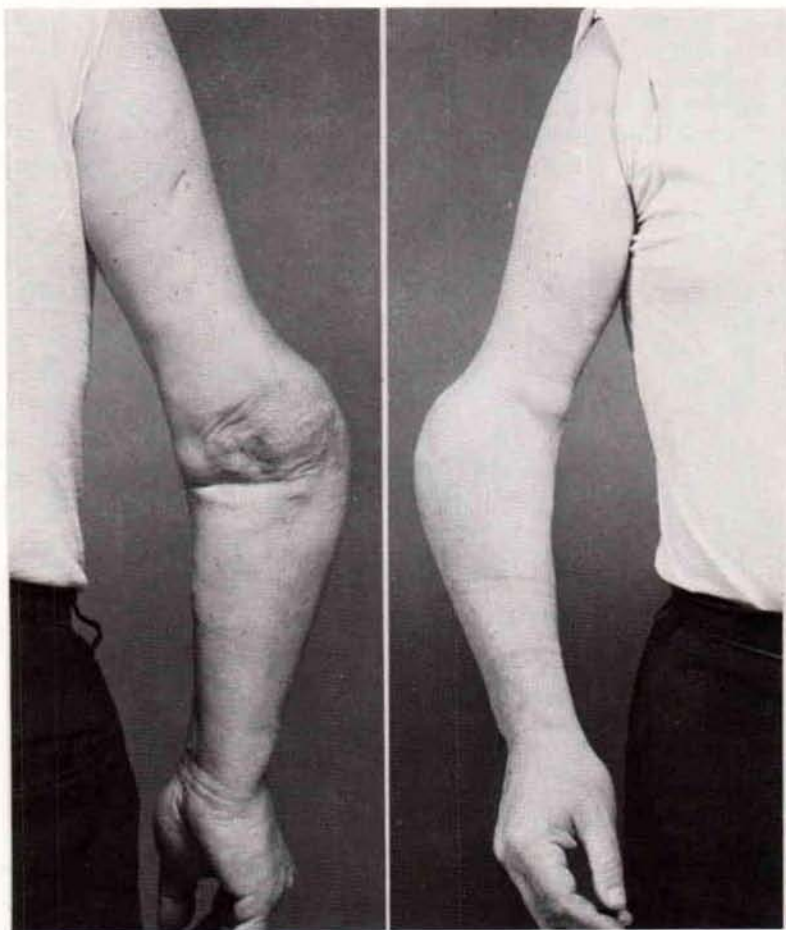
THE USE OF POLYPROPYLENE IN AN UPPER-LIMB ORTHOSIS

Gary Zamosky¹

During the past several years, sheet polypropylene has proven to be a durable plastic, useful in orthotics. Polypropylene is strong, light in weight, and relatively easy to mold. It has been used very successfully in the fabrication of ankle, foot, and spinal orthoses. Described here is a

procedure, based on research done at the Veterans Administration Prosthetics Center, New York City, using polypropylene in the fabrication of an upper-limb orthosis for control and protection of a painful elbow.

The patient (Fig. 1) has considerable lateral



Figs. 1. Anterior and posterior views of the deformed limb.

¹Orthotist, Isadore Zamosky, Inc., Monsey, New York 10952.

and anterior tilt in the plane of the humero-ulnar articulation. There is a large rectangular piece of detached bone in the cubital space. Present, also, is osteosclerosis of the dorsal fragment of the olecranon. Pain and ultrasensitivity at the elbow joint limits his ability to extend his arm fully.

The objective was to provide an orthosis with hinges at the elbow, and a cap to protect the olecranon and posterior portion of the elbow, which at the same time would serve as a stop to limit extension of the forearm. The completed orthosis is shown in Figure 2. The protective cap is part of the humeral segment of the arm.

FABRICATION TECHNIQUES

A plaster cast made with the arm fully extended proved to be the best method for use with vacuum-forming techniques.

The cast is modified with reliefs to provide for the formation of the olecranon cover and to allow for pressure sensitive areas. One sleeve of cotton

stockinet is pulled over the cast to provide the air space desirable in the vacuum-forming procedure (1). Two pieces of polypropylene, 1/4-in. thick, are fitted to the frame of the vacuum-forming machine.

The vacuum-forming procedure is carried out in two stages. The first piece of polypropylene is placed in the oven at a temperature of 375°F for approximately 10 to 15 minutes. The cast is placed on the stand with the dorsal aspect facing upward for the best results. The heated polypropylene is pulled over the cast and the vacuum applied. Excessive material is cut away to provide the forearm cuff, and the cast and cuff are placed in position for the second forming procedure.

The second sheet of polypropylene is heated and vacuum formed over the cast and forearm cuff. The upper-arm cuff is trimmed so enough material is left to cover the epicondyles and to a level 1 in. distal to the olecranon.

The elbow joint axis is located, and copper

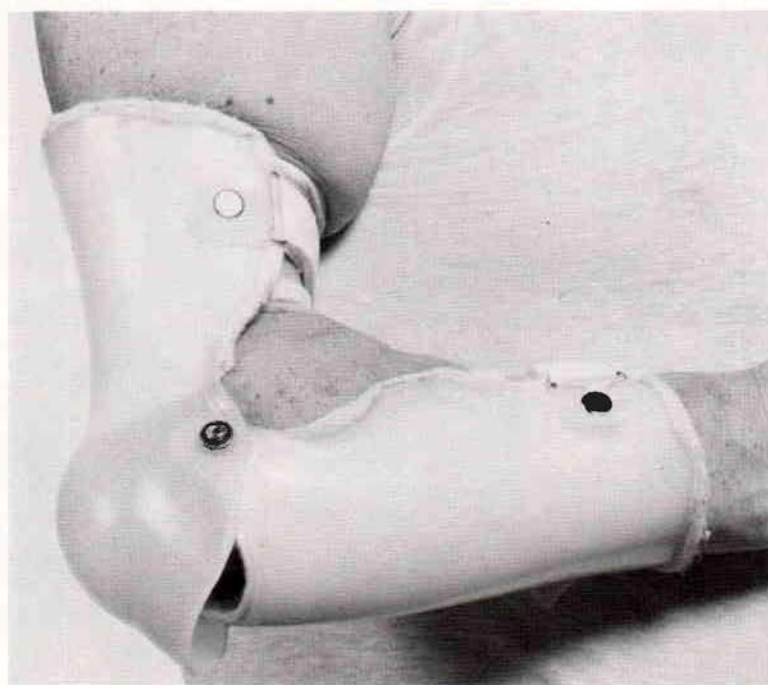


Fig. 2. Lateral view of polypropylene elbow orthosis depicting the protective cover in flexed position.

rivets and burrs are used to join the two parts. The orthosis is held to the arm and forearm by two Velcro straps with plastic loops and tongues.

The polypropylene elbow orthosis can be fabricated easily in 2 to 4 hours. It is cosmetic, light in weight, durable, and easy to keep clean.

The completed orthosis is shown in Figures 2, 3 and 4. The liner in this orthosis is moleskin, but it is not considered to be necessary.

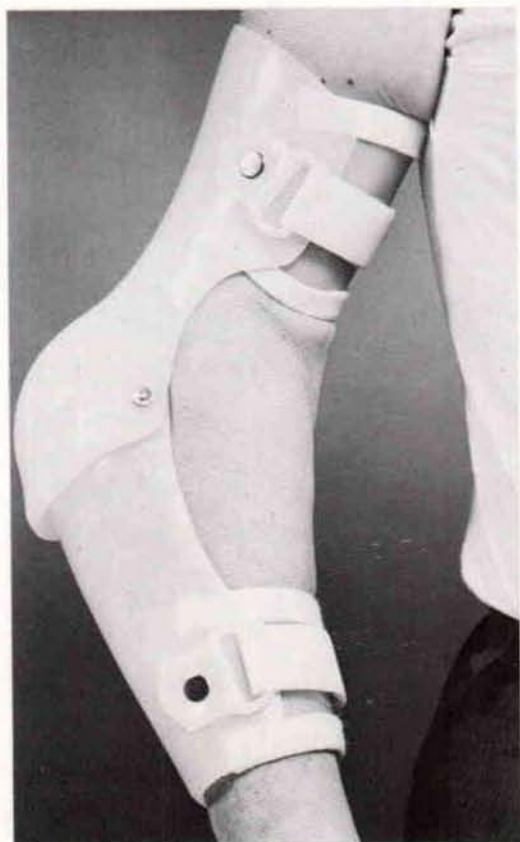
ACKNOWLEDGMENTS

I wish to give my sincerest thanks and appreciation for the time and effort contributed by Dr. Gustav Rubin of the Veterans Administration Prosthetics Center towards the development of

the Polypropylene Elbow Orthosis during my employment at the Veterans Administration Prosthetics Center.

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Figs. 3. Anterior view of elbow orthosis. Note the hyperextension stop.

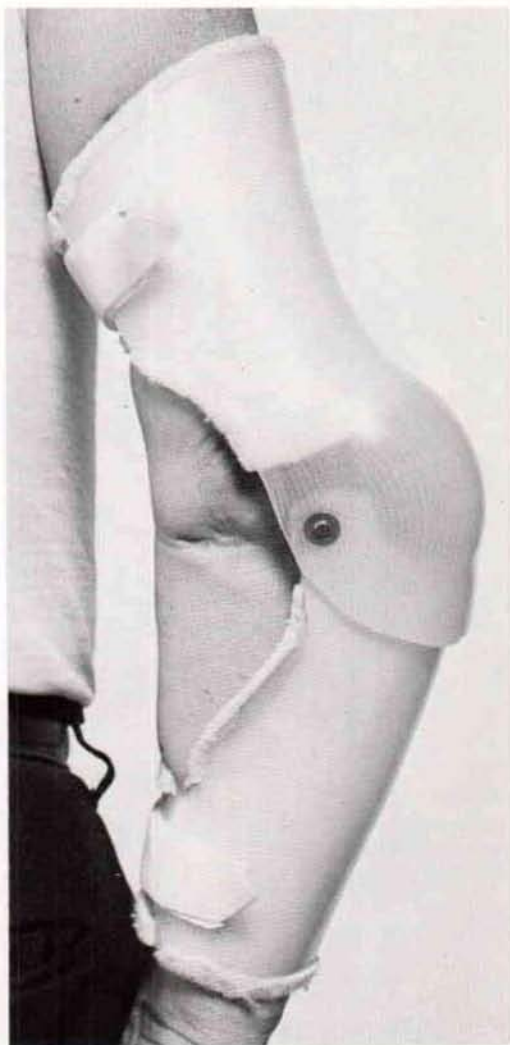
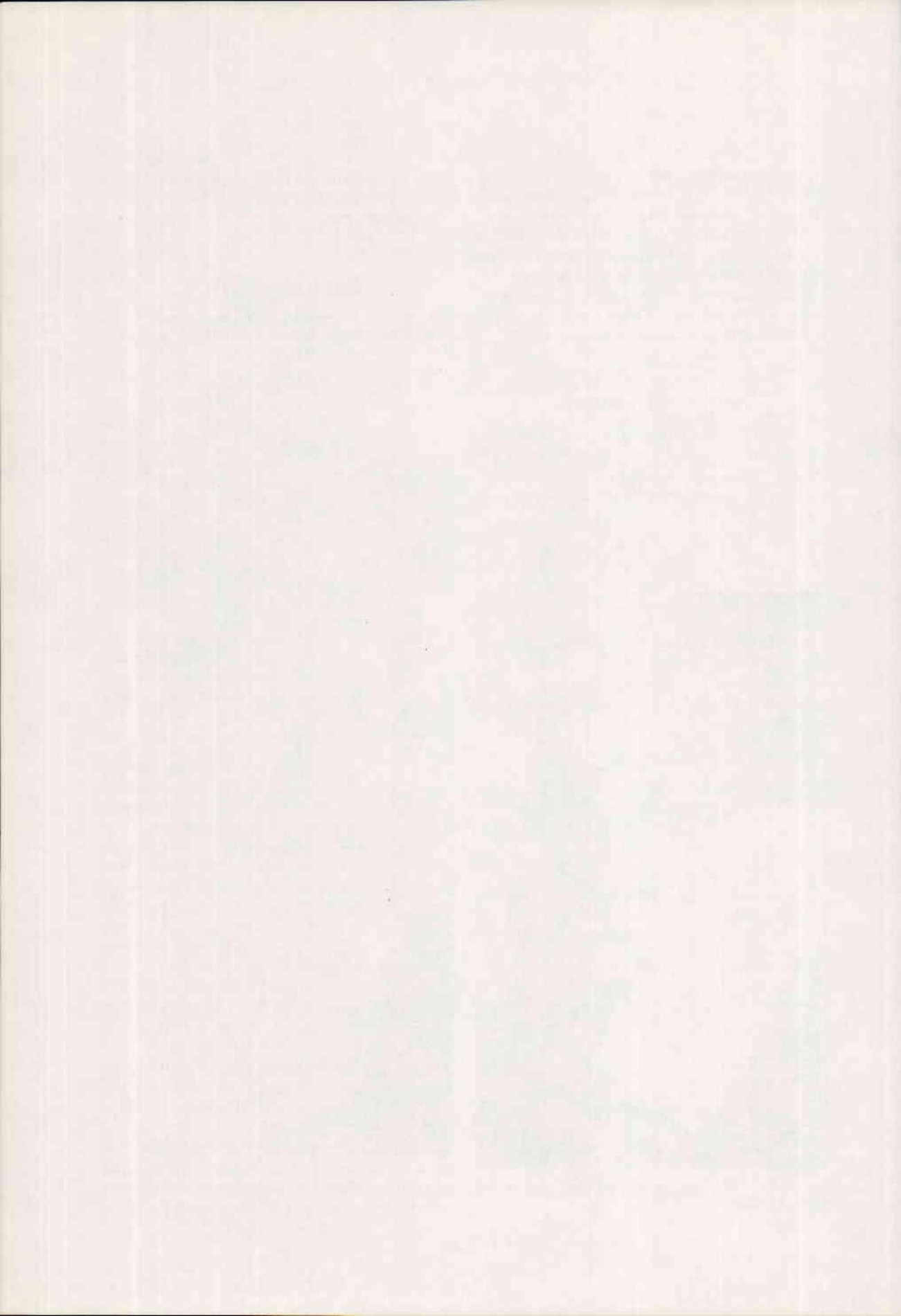


Fig. 4. Posterior view of the elbow orthosis. The liner used in this instance is moleskin, but is not considered to be necessary.



LOWER-LIMB MODULAR PROSTHESES¹ A STATUS REPORT

A. Bennett Wilson, Jr.²

The literal definition of "modular" is "*pertaining to a unit of measurement.*" The word has been used widely in this context in architectural circles for many years, but recent usage in electronics and other space-related technologies has given the word a meaning that connotes interchangeability within a system of components of the same or slightly different characteristics in order to effect a repair quickly or to change easily the characteristics of the overall system.

In recent years "modular" has crept into the language of prosthetics replacing "pylon" (which was also a poor choice of words) and has been used to describe a prosthesis made up of easily assembled and disassembled parts, sometimes interchangeable with parts providing slightly different function.

Some basic definitions concerning construction of limb prostheses developed and adopted at a conference sponsored by the Committee on Prosthetics Research and Development (CPRD) in 1971 (2) are:

- Modular:** Having accessible a number of interchangeable components which can be assembled easily and quickly into a prosthesis.
- Exoskeletal:** Used to describe a prosthesis where the supporting structure is outside of or external to the normal shape of the limb.

Crustacean: Used interchangeably with "exoskeletal"—"crustacean" connotes "shell" whereas "exoskeletal" connotes "external support."

Endoskeletal: Used to describe a prosthesis where the supporting structure is internal to the normal shape of the limb.

Endoprosthesis: A prosthesis lying inside the body.

NOTE: A prosthesis may be entirely endoskeletal, such as the new Bock system, or may be partly endoskeletal and partly exoskeletal, such as the new Blatchford system.

The obvious objectives of modular systems are a reduction in the time required to provide the patient with a functional prosthesis and the possibility of trying out various combinations of components on each patient with a minimum expenditure of time and effort. Additional objectives of some designers have been to provide for adjustability of alignment throughout the life of the prosthesis and lighter weight. In any event it was hoped that the end result would be better service to the amputee at less expense.

The common approach to modular design is to use a metal tube, or pylon, to which feet, knee joints, sockets, and other components can be attached quickly with clamps, screws, or other devices. Cosmetic appearance is provided by use of a nonstructural cover. A tube not only provides an inexpensive structure with a high strength-weight ratio, but also permits easy adjustability in planes perpendicular to the long axis. Paré (1), in the middle of the sixteenth century, offered this type of construction (Fig. 1), and Parmelee, in his suction-socket patent of

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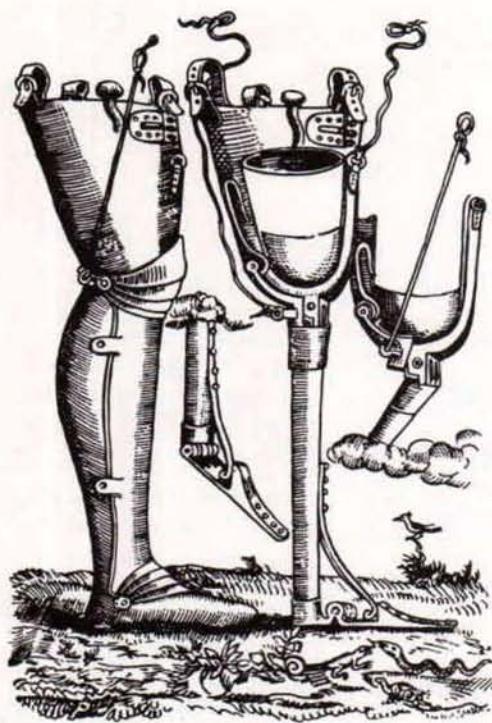


Fig. 1. Artificial leg invented by Ambroise Paré (mid-sixteenth century). From A. Paré, *Oeuvres Completes*, Paris, 1940. From the copy in the National Library of Medicine.

1863, proposed a pylon-type construction (Fig. 2). However, the crustacean type of construction has prevailed to this very day, in all probability because of the difficulty in providing a satisfactory cosmetic appearance.

In the early 1950s the University of California at Berkeley chose the pylon type of construction in designing the "adjustable legs" (Fig. 3) that have been used so successfully in arriving at a satisfactory alignment for each patient, to be transferred to and built into crustacean-type prostheses later (9). At the time of the original design, it was felt, correctly, that wide latitudes of adjustability were required and consequently the adjustable legs were too bulky for use other

D. D. Parmelee,
Artificial Leg.
N° 97,637. *Patented Feb. 10, 1863.*

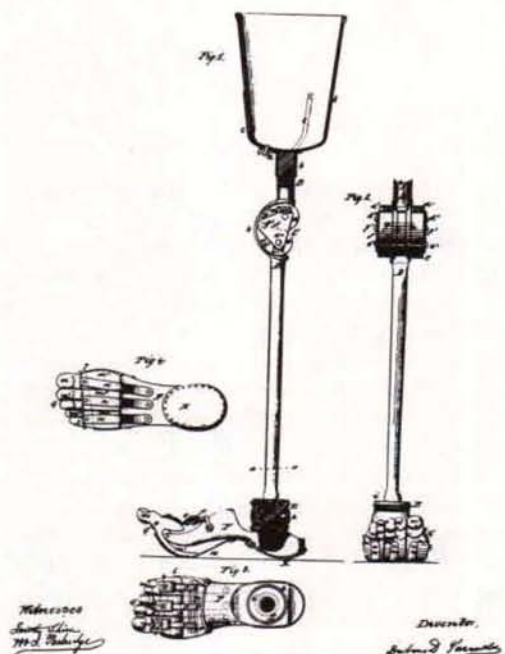


Fig. 2. The "Parmelee Leg" depicted in the 1863 patent covering the suction socket.

than for the time necessary to arrive at an adequate dynamic alignment, although an adaptation of the above-knee unit was made for use as a temporary prosthesis by the Veterans Administration Prosthetics Center.

The use of temporary prostheses was generally discouraged in the United States for many years because it was feared that ill-fitting, hastily devised sockets that might be prepared for temporary use would do more harm to the patient than good. However, in the early 1960s, after techniques had been developed for relatively quick fabrication of plastic-laminate sockets, interest in temporary limbs and fitting was revived and encouraged, and at least two devices—the so-called

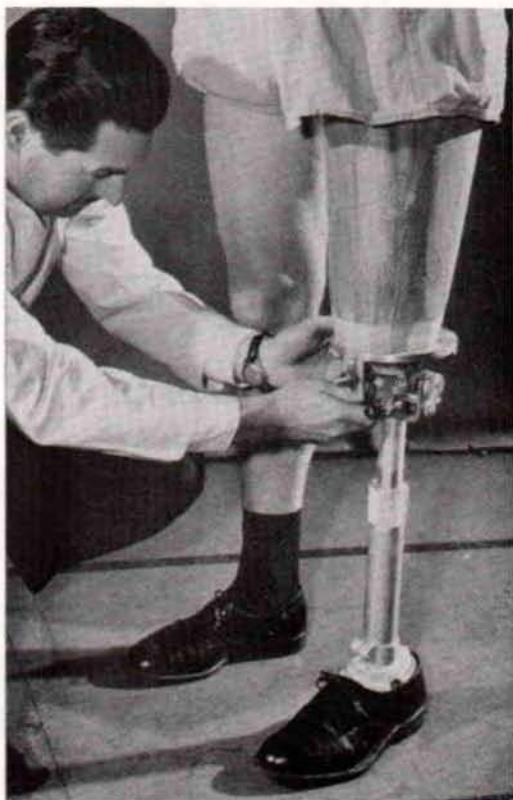


Fig. 3. The adjustable leg designed at the University of California, Berkeley, about 1950 for use in aligning above-knee legs.

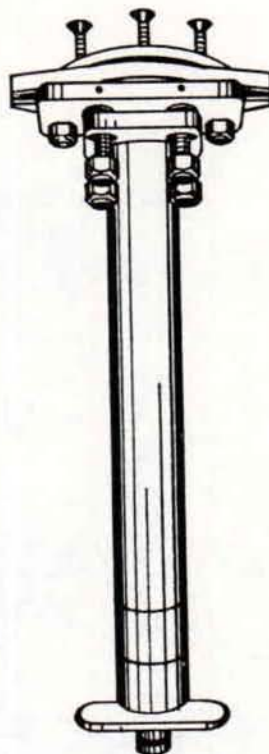


Fig. 4. The Northwestern Below-Knee Pylon, the first unit known to be designed for both temporary and definitive prostheses.

Northwestern adjustable below-knee pylon (Fig. 4)—and the Winnipeg system (Fig. 5) (1) were designed to meet this need (10).

However, it remained for the introduction of immediate postsurgical fitting of prostheses to give full impetus to the development of the modular concept (Fig. 5) (3) (7) (9) (10) (11). The minimum requirements for a lower-limb prosthesis for use immediately after surgery are:

1. Adjustability of the socket in the flexion-extension plane.
2. Adjustability of the socket in the abduction-adduction plane.
3. Adjustability of the shank in the mediolateral plane.
4. Adjustability of the shank in the anteroposterior plane.
5. Adjustability of toe-in and toe-out of the foot.
6. Adjustability of the length of the shank.
7. Provision for quick connection and disconnection of the socket to and from the rest of the prosthesis.

For maximum ease in operation, it is desirable that each adjustment be independent of other adjustments, and an inexpensive method of providing cosmesis would be a welcome feature.

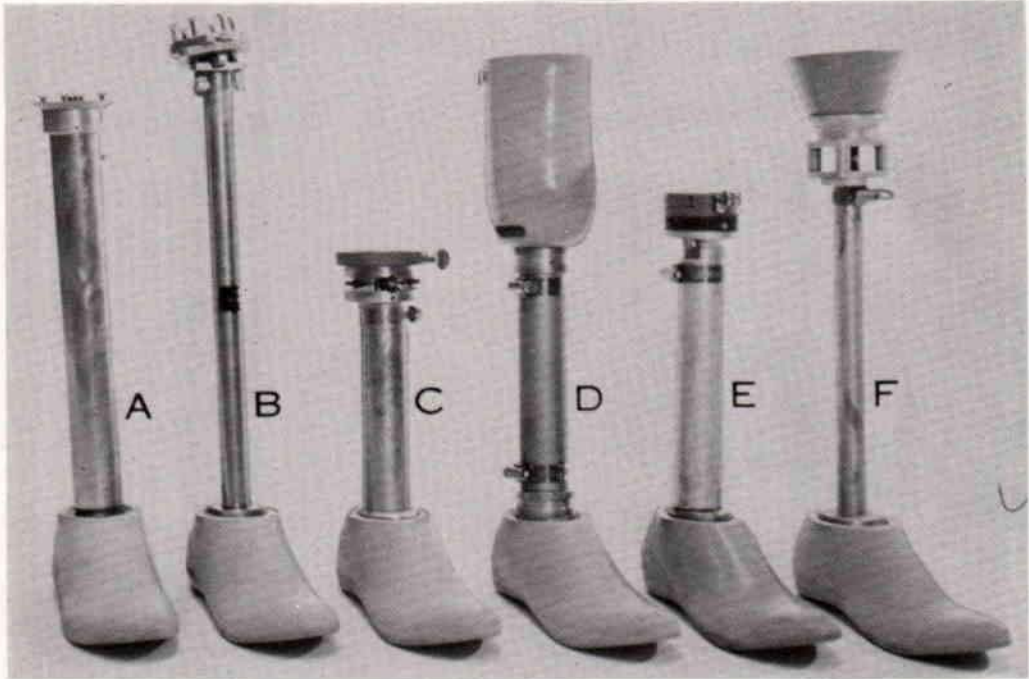


Fig. 5. Below-knee pylon-type prostheses that were available in 1968 for fitting prostheses immediately after surgery. A, Hosmer Postoperative Pylon; B, Northwestern Pylon (Hosmer); C, Veterans Administration Prosthetics Center (VAPC) "Standard" Pylon; D, Canadian "Instant" Prosthesis (Hosmer); E, United States Manufacturing Co. Pylon; F, Finnie-Jig (Arthur Finnieston Co.). Metal straps for attachment to a plaster-of-Paris socket are available, but not shown. *Courtesy of Veterans Administration Prosthetics Center.*

Better procedures for "bench alignment" permitted the reduction of the range of adjustments in comparison to the ranges provided by the original UC-B adjustable legs, and a number of very satisfactory pylon-type units were designed to meet the minimum criteria given above, and were made available commercially (Fig. 5) (10). These devices are used widely where immediate postsurgical fitting and early fitting procedures are carried out.

The minimum range of adjustments, set at a workshop on the subject sponsored by CPRD in 1971 (2), are:

Range of motion in flexion-extension plane8 deg
Range of motion in adduction-abduction plane8 deg
Horizontal movement in medio-lateral plane20 mm
Horizontal movement in antero-posterior plane20 mm

At the beginning of the immediate postsurgical fitting program, little attempt was made to provide good cosmesis, although the advantages that might accrue if it were feasible to leave the original unit in the definitive prosthesis were generally

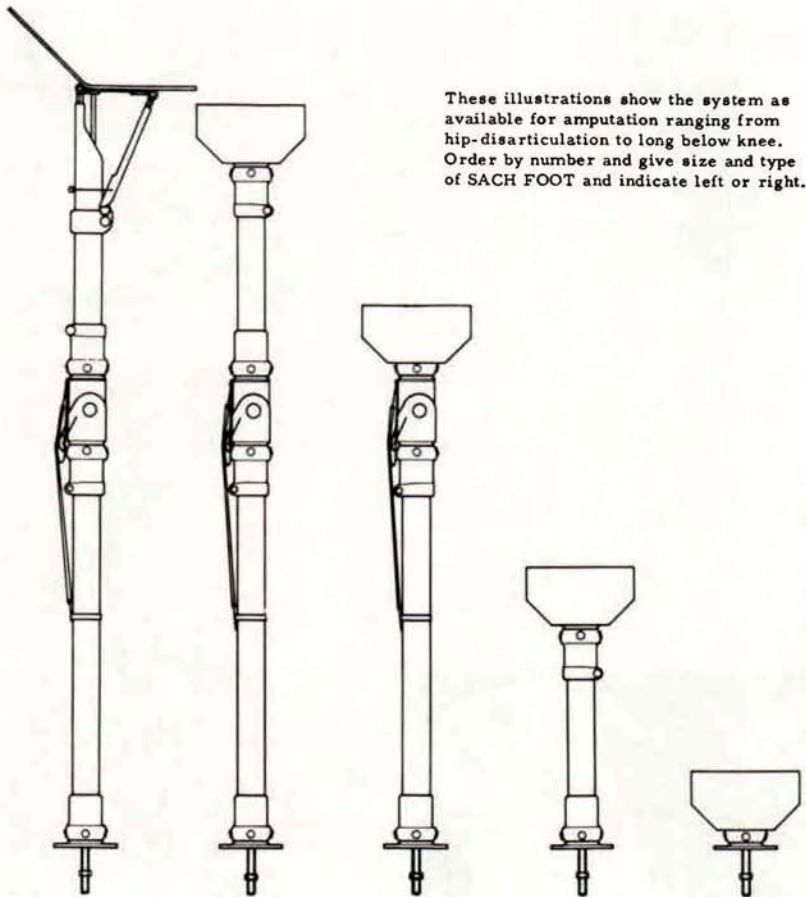


Fig. 6. Line drawings of the basic components of the Otto Bock modular system. From left to right, set-ups for hip-disarticulation, short above-knee, above-knee, below-knee, and Syme amputations.

recognized. It was felt by designers and prosthetists that the endoskeletal, modular concept offered an opportunity to provide a lighter weight prosthesis with improved cosmesis (to the eye and to the touch) at a lower cost. The obvious problem has been provision of adequate cosmesis, a problem pretty much independent of the mechanical design of the device itself.

A number of schemes to provide cosmesis have been investigated by private industry and by government-supported research groups, but the only method successful to date is the preformed resilient plastic foam that must be shaped and "fitted" by the prosthetist and for which a cosmetic cover, or skin, is needed.

All major manufacturers and suppliers in the United States offer one or more systems that include components for amputee types ranging from long below-knee to hip-disarticulation and hemipelvectomy.

The Otto Bock³ (Figs. 6, 7, 8, and 9) and IPOS⁴ systems (Figs. 10 and 11) are designed so that angular and linear adjustments can be made throughout the life of the prosthesis. The Kol-

³Manufactured by Otto Bock Orthopedic Industry, 3428 Duderstadt, W. Germany.

⁴Manufactured by IPOS Kommanditgesellschaft, 314 Luneburg, W. Germany. Available through Hosmer, Inc., 561 Division Street, Campbell, California 95008.

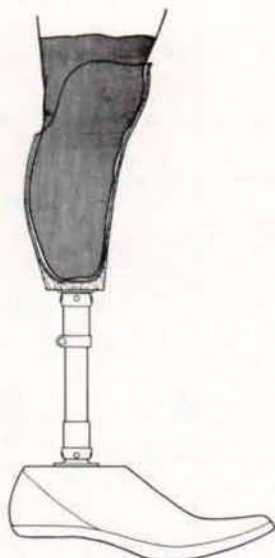


Fig. 7. Bock system used on below-knee patient. The cosmetic cover is not shown.

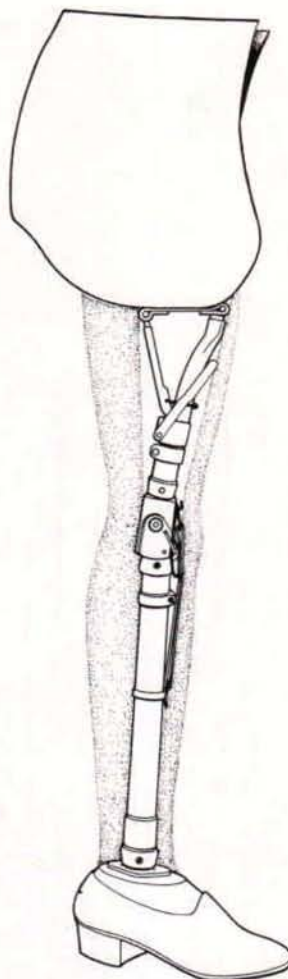


Fig. 9. The Bock system for the hip-disarticulation amputee.



Fig. 8. The Bock system for an above-knee amputee showing the foam block that will be used to provide a cosmetic cover.

man⁵ system (Fig. 12) is simpler but does not provide adjustment in as many planes as the Bock and IPOS systems.

Various types of knee joints are available for use in the Bock, IPOS, and Kolman systems, but the Universal Multiplex Unit (Fig. 13) is the only design currently offering interchangeability of various popular fluid knee-control units.

The designers and manufacturers of the VAPC system recommend that the adjustable component be removed and replaced with solid parts.

⁵Manufactured by U.S. Manufacturing Company, 623 South Central Avenue, Glendale, California 91209.

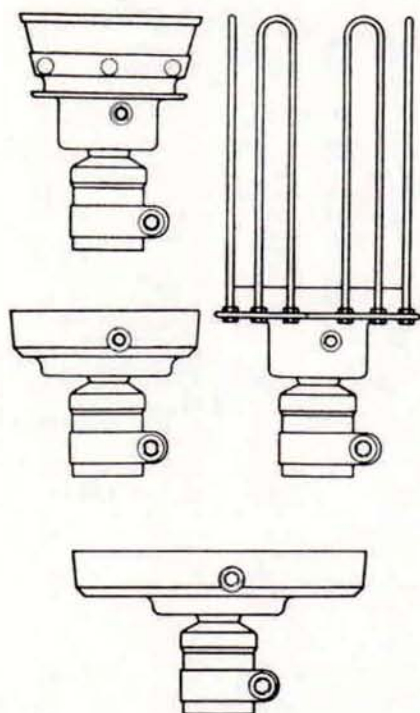


Fig. 10. The IPOS below-knee and above-knee components. Additional components for the hip-disarticulation case are also available.

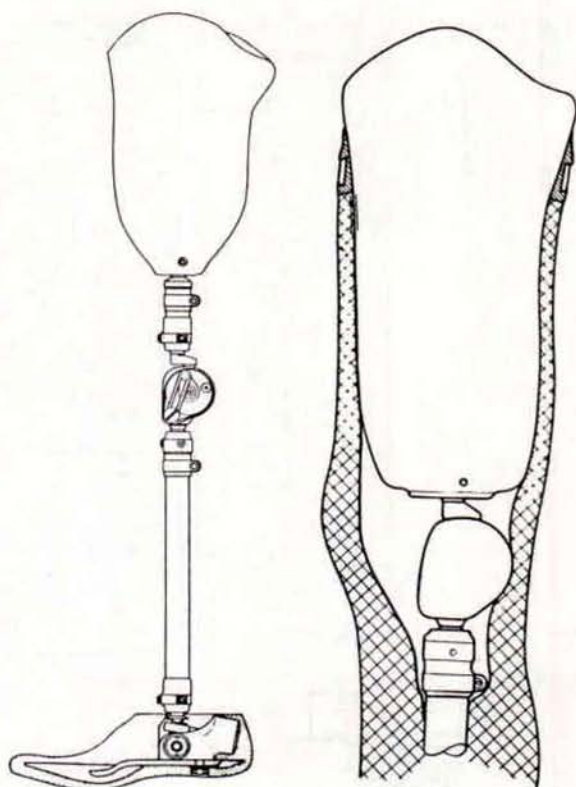


Fig. 11. Schematic view of the IPOS system for the above-knee amputee to show relationship between the cosmetic cover and the knee joint.

All systems use a plastic foam material to cover the mechanical parts (Fig. 8).

Three modular, endoskeletal systems are available for the hip-disarticulation case: the Otto Bock system (Fig. 9), the Kolman system (Fig. 12), and the IPOS system.

No formal clinical evaluation of any of these devices has been conducted in the United States because it has been felt that since no new function is provided the patient, such a program would be a waste of time and money. A research group at Strathclyde University did undertake a rather extensive evaluation project in 1971 involving 6 types of units and 23 subjects, each of whom were fitted with prostheses incorporating each unit. In all, there were 140 fittings involved. In an attempt to reduce the variables as much as possible, one prosthetist carried out all of the fittings. As might have been predicted, there seem to have been

more variations within the prosthetist's contribution than there were from unit to unit. The report of this evaluation project has never been issued.

The government of Great Britain has sponsored, in cooperation with the International Society for Prosthetics and Orthotics, a series of international meetings (4) (5) (6) (8) to develop standards and bring about a certain amount of interchangeability of components for modular prostheses. The Blatchford⁶ system has been adopted for use throughout Great Britain for an interim period until a distinctly improved model is developed.

⁶Blatchford & Sons Ltd., Artificial Limb Manufacturers, Lister Road, Basingstoke, Hants. RG 22 4AH, England.

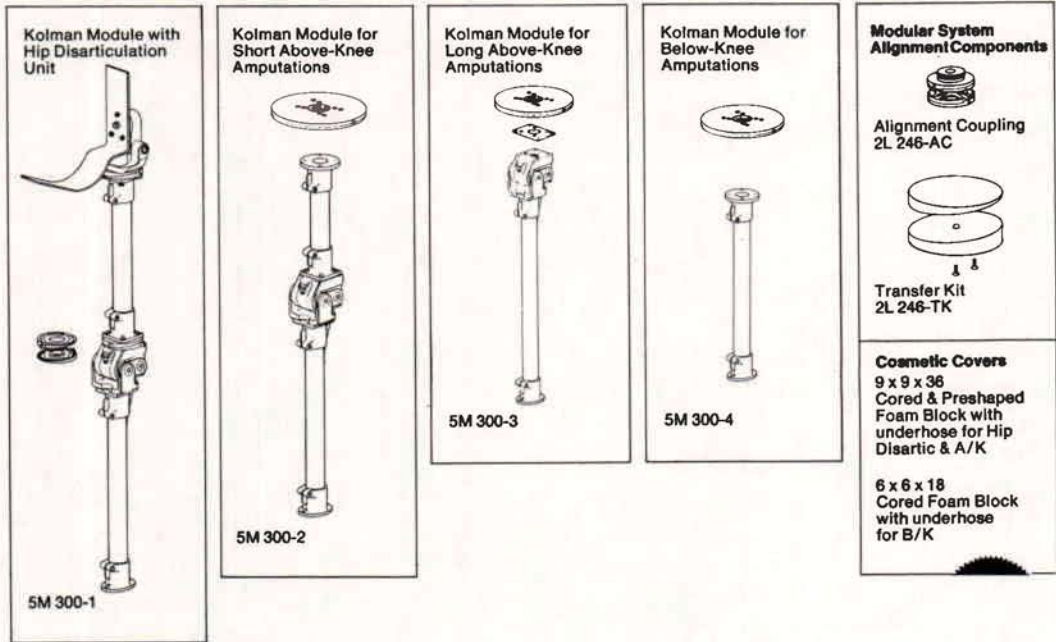


Fig. 12. Components for the Kolman system.

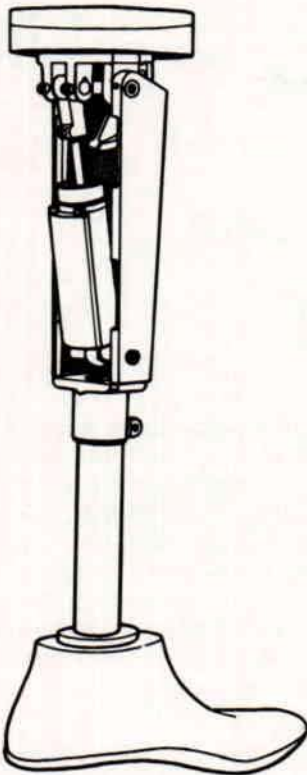


Fig. 13. Universal Multiplex Unit.

To get some idea of the use of modular prostheses in the United States, a limited survey was made by mail of 15 private limb facilities. Institutions not required to show a profit were excluded for obvious reasons. The questionnaire used is shown in Figure 14.

Very few patterns emerged from this survey. Modular prostheses are being used for immediate postoperative fitting and early fittings at all levels, but only the hip-disarticulation prostheses have found widespread acceptance for definitive prostheses.

It is clear that the differences between units designed for a given level are so small that such factors as availability and personal preference have as much to do with choice and selection as anything else.

CONCLUSIONS

Designers, manufacturers, and clinicians had hoped that the "modular" approach would yield better appearing prostheses that were lighter in weight and less expensive than the crustacean types in general use. However, modular prostheses for the below-knee and above-knee levels have been disappointing in this respect because

CPRD 6/5/74

Name of Respondent _____

MODULAR PROSTHESIS SURVEY

FREQUENCY OF USE

Type of Prosthesis and Components	Immediate Postoperative Fitting	Early Fitting	Definitive Fitting	Remarks
<u><u>BELOW-KNEE</u></u>				
1st Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
2nd Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
<u><u>ABOVE-KNEE</u></u>				
1st Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
2nd Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
<u><u>HIP-DISARTICULATION</u></u>				
1st Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
2nd Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	
<u><u>UPPER-LIMB AMPUTATIONS</u></u>				
1st Choice Name _____	Never Seldom Often Always	Never Seldom Often Always	Never Seldom Often Always	

Fig. 14. Questionnaire used in CPRD survey.

they are usually heavier than the corresponding crustacean model, and the time required to install the cosmetic filler and cover makes the modular prosthesis relatively expensive.

Nevertheless, the modular systems have proven to be valuable in the field of lower-limb prosthetics in early fitting and immediate post-surgical fitting at all levels. The hip-disarticulation systems are adopted widely for definitive prostheses, and the below-knee and above-knee systems seem to be used in definitive prostheses only when cosmesis is an unusually large factor to the patient and when hard use is not contemplated.

Work is being carried out to improve durability and to reduce the costs of achieving excellent cosmesis. In any event, the modular concept has already made a place for itself in lower-limb prosthetics.

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A MODIFIED PROSTHETIC FOOT FOR PILOTS

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Gene Helmuth¹

A number of lower-limb amputees who were airplane pilots, and who wished to continue flying, have requested a modification of the prosthetic foot so that the rudder pedal can be operated without the foot hitting the brake pedal. Also, there is concern about the foot getting "hung up" in the cockpit should bailing out be necessary.

The foot described in this article is one way in which we have dealt with this problem and still provided the patient with a functional prosthetic foot for ordinary use so that it is not necessary to change the entire foot each time he flies.

The amputee came to us wearing the foot as shown in Fig. 1. The foot, designed for the

the pilot had to sacrifice the function of the forefoot needed for normal earthbound activities.

THE NEW DESIGN

The keel (Figs. 2 and 3) for use in a molded SACH foot was constructed of stainless steel, wood, and balata belting. The stainless steel

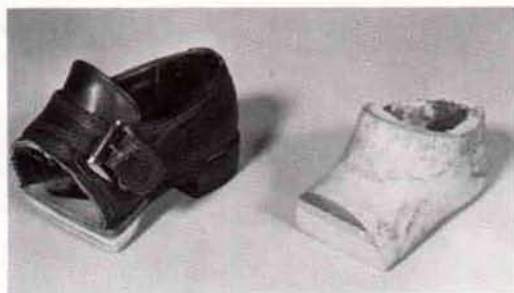


Fig. 1. Modified foot from "Hydra-Cadence" unit used by amputee in piloting an airplane. Forepart of foot has been removed to avoid undesirable operation of brake.

"Hydra-Cadence" above-knee prosthesis (2), had been used as a basis for the modified design.

The forefoot had been cut off and thus there was no "toe lever" for pushoff during walking. Because the foot was one used with the "Hydra-Cadence" unit, the cosmetic cover was glued to the top surface of the foot making it almost impossible to change the foot each time he flew. Since the shorter foot was needed in flying,



Fig. 2. Special keel assembly for use in construction of a molded SACH foot.

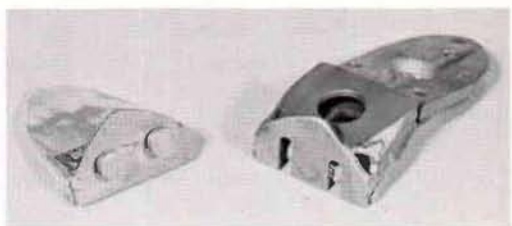


Fig. 3. Keel shown in Figure 2 showing quick connect-disconnect feature for removal and reinstallation of the forepart of the foot.

part, 304 grade, was shaped and welded to conform to the size of the foot mold to be used. The 1/4-in. thick balata belting was riveted into place with 3/16-in. copper rivets. The two locking pins were made of 3140-type steel for high strength, and welded into place. The wood block, which

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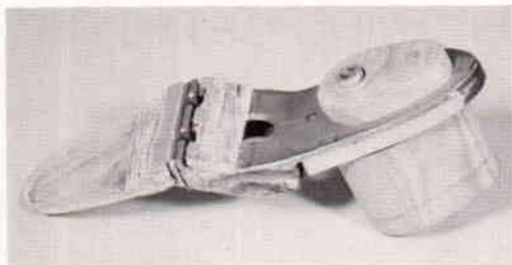


Fig. 4. Keel assembly with wooden parts installed preparatory to casting of the foam rubber.



Fig. 6. Completed two-part SACH foot for use by amputee in piloting aircraft.

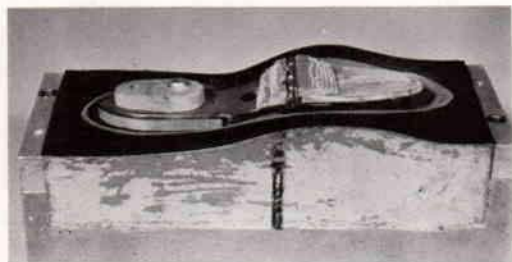


Fig. 5. Keel assembly in mold.

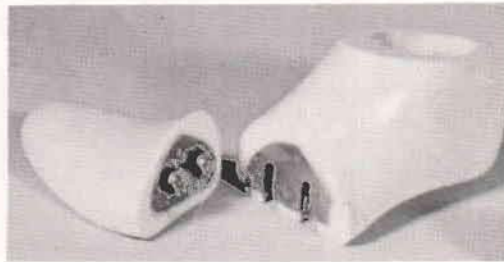


Fig. 7. Two-part SACH foot shown in Figure 6 with forepart of foot detached.

can be seen in the center of the heel section (Fig. 4), was used as a permanent filler to keep the weight down. The block was secured with three #8-32 screws. The other two pieces of wood were used as fillers for the voids needed to make attachment of the foot to the shin possible.

The foot was poured in the usual manner for making SACH feet (Fig. 5) (1). The completed foot as it appears with and without the forefoot is shown ready for attaching to the prosthesis, in Figures 6 and 7.

With this foot the amputee is able to wear a

complete artificial foot for his normal everyday activities. When he wishes to fly, the only change required is removal of the toe section of the artificial foot and a change of shoe.

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A PATELLAR-TENDON-BEARING ORTHOSIS

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"Short leg braces" with molded leather cuffs or ischial weight-bearing "long leg braces" have been used for years to unweight the leg below the knee.

In certain pathological conditions of the lower limb, the stress of weight-bearing cannot be tolerated because of pain or the possibility of tissue damage. Pathologies encountered in such situations fall into three broad categories: 1) those affecting bone, such as delayed unions or nonunions of fractures; 2) those involving the ankle or foot joints, such as traumatic arthritis or similar conditions; and 3) those involving the soft tissue, such as ulcers and traumatic loss of the heel pad or other soft tissues.

The Veterans Administration Prosthetics Center designed an orthosis in 1958 using principles of the patellar-tendon-bearing prosthesis to load the distal shin, ankle, and heel (1) (2).

Using the Veterans Administration Prosthetic Manual, "PTB Sockets for Below Knee Weight Bearing Braces," dated January 3, 1961 (1), we measured, fabricated, and fitted several braces as described therein. After experience with patients wearing the patellar-tendon-bearing socket for some time, changes in the anterior-posterior measurement of the leg seemed to be indicated. No provisions for adjustment of the hinge on the medial side of the socket had been made, but adjustment could be made on the lateral side by tightening the strap. However, the patients felt that the socket was not fitting the leg properly, as though the socket was twisted about the leg. Additional strips of Kemblo were glued to the socket, but this increased the bulk.

In an attempt to overcome these problems several two-part sockets were made. The anterior section was attached to the uprights and the posterior section was made so that it lapped over the outside of the anterior section. Straps to adjust the anterior-posterior pressure were provided. This design represented progress but the poster-

ior section tended to move up and down on the anterior section, and it was still quite bulky.

With improved techniques for plastic lamination and casting, a new design was developed, and is being used today.

FABRICATION

CASTING

A wet cast sock is pulled over the patient's foot and leg, and held up by an elastic strap around his pelvis and two Yates clamps.

In the posterior area a piece of lead or webbing or elastic strap is laid underneath the cast sock along the midline where the cast will be cut for removal.

The position of the knee joint is located and marked on the medial and lateral sides at the level of the center of the patellar tendon. A line in the horizontal plane connecting these two points through the center of the patellar tendon is drawn.

The distal, proximal, medial, and lateral aspects of the patella are outlined and connected.

A horizontal line is drawn through the center of the outline of the patella.

The lateral condyle of the tibia and the head of the fibula are located and marked.

The crest of the tibia is outlined and a line is drawn down the crest to a point approaching the ankle.

The posterior tibial flare is marked on the medial side.

Two layers of plaster-of-Paris splints are laid over the crest of the tibia. The cast is made somewhat longer than necessary, and when possible, the distal part of the splints are folded over to reinforce that part of the cast. Two more layers of splints are laid just slightly medial to the original two layers, and then two additional layers are laid over the lateral aspect. The plaster in these splints is worked with the hand so that it is well molded over the crest of the tibia and around all the bony areas of the stump. The thumbs are imprinted in the patellar-tendon area, and indentations are made the same as when a cast is

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taken for a patellar-tendon-bearing socket. After these have partly dried, the balance of the shin is wrapped with a roll of plaster bandage.

When the cast has set, it is removed by cutting it with a cast saw down the posterior aspect along the midline of the webbing that was inserted under the cast sock. After the cast has been removed from the patient it is wrapped with an additional bandage to hold it together, and the distal end is covered so that a positive model can be poured.

THE POSITIVE MODEL

The cast is coated on the inside with sodium silicate, commonly called waterglass, or any other separating agent.

The cast is then put into a box, and supported with sand. A pipe is inserted as nearly as possible in the center of the cast between the patellar tendon and the popliteal area in the AP plane, and between the medial and lateral walls in the ML plane to make alignment easier later. The cast is then filled with plaster of Paris which is allowed to cure until it is hard.

The female cast is removed and the male mold is ready for modification.

With a round Surform file, plaster in the area of the patellar tendon is removed to the point where the indentations in the plaster cast were formed by the thumbs.

With a scarpa knife, the plaster along the lateral aspect of the tibia is removed and a Surform file is used to remove a small amount of plaster along the shank of the fibula. However, the marks at the head of the fibula and the distal end of the fibula are not removed.

An additional small amount of plaster is removed from the posterior aspect of the cast coming around to the medial side and a slight amount is removed under the medial tibial plateau.

Usually the amount of plaster removed from the male mold is so slight that the marks made by the wet stump sock are still visible.

Circumference measurements should then be made of the cast. They should be the same as those taken of the stump with the exception of the one taken through the patellar-tendon area which should be $\frac{1}{4}$ -in. smaller than the measurement taken on the patient.

If any additional plaster needs to be removed, it should be removed in the posterior aspects of the cast.

The cast is then smoothed with a sand screen.

The ML dimension should be the same as that measured on the patient's leg.

The AP dimension should be the same as that measured on the patient's leg. If the AP dimension is not the same, plaster should be removed from the popliteal area.

The cast (Fig. 1) is now ready for the buildup in the posterior area so as to provide a "roll" of the socket in that area.



Fig. 1. Modified positive model for PTB orthosis.

A line is drawn on the posterior aspect of the cast in the same horizontal plane with the midline of the patella tendon.

With a "T-type" instrument and using the pipe as a reference, a horizontal line is drawn on the posterior aspect of the cast, perpendicular to the pipe.

A piece of lead about $\frac{1}{16}$ -in. thick, $\frac{1}{2}$ -in. wide, and 12-in. long is sprayed with Silicone and then shaped to conform to the back of the cast making it the width of the posterior aspect of the cast (Fig. 2). It is placed on the line just previously drawn as shown in Figure 2.

The area within the lead piece is filled with plaster of Paris. The head of the fibula is built up with plaster slightly about $\frac{3}{16}$ in. at the apex and feathered to nothing around the edge of the marks

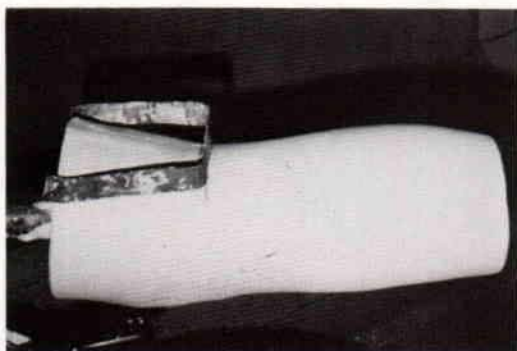


Fig. 2 Lead form in place for formation of roll in proximal posterior section of socket.

that have been on the cast outlining the head of the fibula.

The lateral tibial condyle area is built up in the same manner.

The tibia tubercle is also built up in the same manner but is extended down the entire length of the tibia while the distal third of the tibia is built up at least 3/16 in. and the sides are feathered into the cast.

The lead is then removed, and plaster is used to fill in the cracks around the edge of the buildup which is smoothed to where it is even with the rest of the cast. A little additional buildup is made on both the medial and lateral sides of the popliteal area to allow for motion of the hamstrings when the knee is flexed.

LAMINATION

After the cast is has been modified and smoothed, it is covered with a coat of Vaseline or talcum powder in order to allow the PVA bag to slide over the cast easily when the lamination is to be made over a wet cast.

A PVA bag is pulled over the entire cast, and tied off at the proximal end. Any extra material is cut off of the distal end, a small sheet of PVA is pulled over the distal end, and the excess material trimmed off so that the entire cast is covered with PVA.

A piece of nylon stockinet, cut twice the length of the cast plus an extra allowance, is sewed in the center and the first section is pulled over the cast followed by the second section.

A piece of plastic sheeting (Mylar) about 1/32-in. thick and a width approaching, but not exceeding, the buildup on the posterior aspect of

the cast in the popliteal area is sewn to a piece of stockinet about as long as the model (Figs. 3 and 4).



Fig. 3. Stockinet with Mylar sewn in distal end in place on positive model.

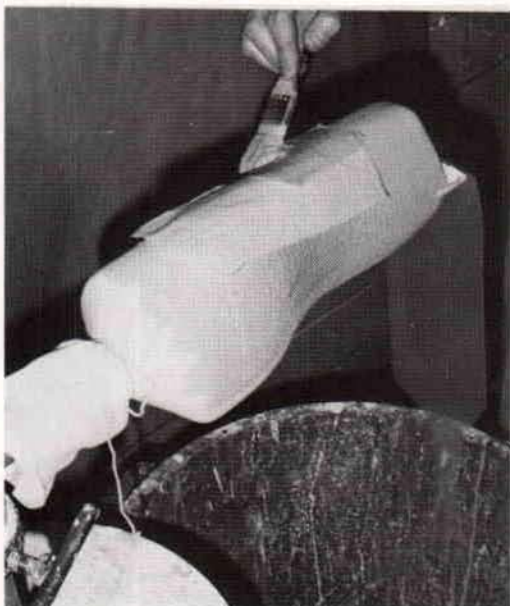


Fig. 4. The addition of Dynel for reinforcement. Note the Mylar strip that is to be used as a separator.

The stockinet is then stretched over the cast. It should be large enough so that when it is pulled over the cast bridging will not occur. The distal end of the stockinet is tied to the cast.

Then a mixture of 85 to 90 percent rigid and 10 to 15 percent flexible polyester resin is mixed together along with the catalyst, color, and promotor, and painted on the anterior area and along both sides to approximately 1 in. of the posterior line of the cast. Only a few drops of promotor are used so that the resin will cure very slowly, to give more time for carrying out the rest of the laminating procedure. Dacron felt is not used because with the opening and closing of the posterior aspect of this socket, Dacron felt seems to break down while nylon stockinet continues to hold up over several years of service. The rigid plastic should be far enough posterior on both medial and lateral walls so that the metal uprights can be attached to the rigid section.

Dynel is added to the lateral wall for reinforcement. It goes posteriorly over the head of the fibula and additional layers are added so that the Dynel gives extra support over the entire anterior, medial, and lateral aspects of the rigid part of the socket (Fig. 4).

A mixture of 85 to 90 percent flexible and 10 to 15 percent rigid polyester resin is then mixed and painted on the posterior aspect of the cast.

The Mylar strip is then pulled down over the posterior aspect of the cast, and tied at the distal end (Fig. 5).

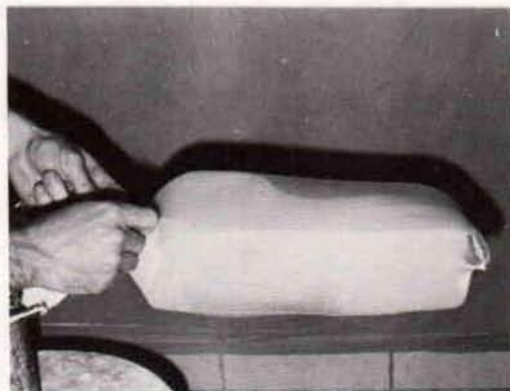


Fig. 5. The Mylar strip in place over the posterior aspect of the positive model.

Two additional layers of nylon stockinet are then applied over the entire model (Fig. 6) and the rigid mixture is painted over the same area to the point where the rigid plastic was applied previously. Two more layers of stockinet are then



Fig. 6. Application of two layers of stockinet over the positive model after the Mylar strip has been put in place.

pulled over the entire cast and tied off around the pipe. Additional rigid resin can be painted on these last two layers if desired.

A PVA bag that has been previously wrapped in a wetted towel for dampening is pulled over the entire cast at this time and the 85 to 90 percent flexible and 10 to 15 percent rigid mixture of polyester resin is poured into the bag and worked into the material over the entire model (Fig. 7).

The PVA bag is then stretched tight, and tied off by means of Yates clamps at the top.

Plastic tape is used to wrap the model in the patellar-tendon-bearing and the popliteal buildup areas. A piece of sponge rubber is placed in the popliteal area and wrapped with the plastic tape to form an even roll in the hamstring-popliteal area.

The entire model is then wrapped with an Ace bandage and allowed to cure.

After the laminate has set, the entire assembly is cured at 250°F for a minimum of 30 minutes.

TRIMMING THE SOCKET

The ends are cut off after the laminate has been cured in the oven, and while it is still hot.

The posterior wall over the area that is built up is cut away.



Fig. 7. The outer PVA bag is in place and the resin is being poured.

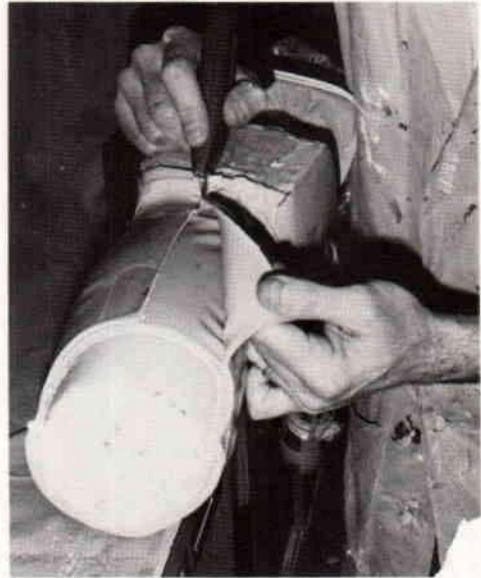


Fig. 8. The laminate is cut to the Mylar strip.

Marks are made on the posterior aspect of the cast for the centerline and the width of the Mylar that was inserted under the lamination.

The posterior aspect of the plastic is cut to, but not into, the Mylar (Fig. 8).

The inner area should be on the lateral side when it is cut so that the tongue goes from the medial to the lateral side and therefore there will be no pressure on the medial posterior condyle area (Fig. 9).

Other trimlines can be determined at this time, and the plastic can be cut while it is still warm (Fig. 10).

When all of these areas have been cut, the socket should be taped back together and allowed to cool completely on the cast.

FITTING AND ASSEMBLY

After the socket is cool, it is removed and tried on the patient, at which time additional adjust-



Fig. 9. The inner socket is cut on the lateral side.

ments can be made to the "socket."

The outline of the leg that was taken at the time of measurements is laid out in accordance with standard orthotic principles, and uprights are bent to the outline.



Fig. 10. Posterior view of the socket.

The footpiece is attached to the shoe, the socket is applied to the patient and taped together, the shoe is put on the patient, and the alignment of the uprights is checked, making sure that the socket is in the proper place and that patellar-tendon-bearing is achieved. The location of the uprights is then marked on the socket.

The socket and uprights are removed, holes are drilled in the uprights and the socket, temporary screws are inserted, and the orthosis reapplied to the patient for additional checks of alignment and weight-bearing characteristics.

Adjustments are made as required, and the socket and uprights are riveted together.

Velcro straps should be attached to the posterior aspect of the socket to keep it closed and the patellar-tendon-bearing surface is placed in the proper relationship to the lower limb.

After the patient wears the orthosis for a short time, he should remove the socket and check the pressure points. Usually he will find a red spot over the patellar-tendon area. Occasionally on people who have very thin skin, it is necessary to put a piece of stockinet under the socket when they are first using this weight-bearing orthosis.

With this orthosis, the amount of the patellar-tendon bearing can be adjusted as required.

Experience has shown that this orthosis is most effective when used with limited motion joints at the ankle. The range of motion may be increased as the patient progresses.

ACKNOWLEDGMENTS

I express sincere appreciation to Robert O. Gooch, C.P., Felton L. Elliott, C.O., and Russell E. Bland, members of the Department of Prosthetics and Orthotics, Duke University, for their assistance in developing this technique.

LITERATURE CITED

1. McIlmurray, William J., and Werner Greenbaum, *Patella Tendon Bearing Socket for Below Knee Weight Bearing Braces*. Veterans Administration Prosthetics Center, New York, January 3, 1961.
2. Kay, Hector W., Clinical applications of Veterans Administration Prosthetics Center patellar-tendon-bearing brace. *Artif. Limbs* 15:1:46-67, Spring 1971.

NEW PUBLICATIONS

PROSTHESES AND REHABILITATION AFTER ARM AMPUTATION. Leonard F. Bender, M.D., M.S., Charles C Thomas, 179 pp., \$12.50.

This volume is a textbook approach to upper-limb prosthetics rehabilitation. It is primarily of interest to physical therapists, occupational therapists, and physicians who have not been exposed on a regular basis to upper-limb amputees requiring prostheses and rehabilitation.

Pre- and postoperative care is given thorough coverage and includes many techniques and assistive devices used for activities of daily living with and without prostheses.

Complications in the use of upper-limb prostheses are discussed including neuromas, weakness, phantom pain, sensation, psychological reaction, problems of stump contour, skin problems, bony overgrowth, and burns. A detailed description of the technique for immediate post-surgical fitting of upper-limb prostheses along with a chart for determining amputation level is set forth.

Construction of definitive prostheses including wrapping the cast, using the wax check socket, master mold, harnessing and control systems are presented, and give the uninitiated insight into the prosthetists' technocracy. This section blends into a description of normally available upper-limb prosthetic components—terminal devices, wrists, elbow mechanisms, and shoulder units. Most of the components are illustrated, and anticipated application with respect to function for each is considered.

Dr. Bender provides a significant contribution with a chapter on prescription of upper-limb prostheses. He describes amputee level along with any limitations or complications, and follows this with various possible prosthesis prescriptions for each level. The range of levels described begins with transmetacarpal and extends through the forequarter. Controls training is included. Examples of check-out forms are given and their use is discussed.

A short, unique description, along with a series of illustrations, for functional partial hand prostheses completes this work.

This is an excellent text and reference source for all members of prosthetic clinic teams where upper-limb amputees are rarely seen. It provides prescription guidelines, insight into training methods, and the function that can be expected for each level of amputation through the upper limb.

C. H. Dankmeyer, Jr.

THE CONTROL OF UPPER-EXTREMITY PROSTHESES AND ORTHOSES. Edited by Peter Herberts, Roland Kadefors, Robert Magnusson, and Ingemar Petersen, Charles C Thomas, Publisher, 261 pp., \$18.75.

This book consists of 24 papers given at a small conference held in October 1971 in Göteborg, Sweden, for the purpose of interchanging experience and results of research in externally powered prostheses. Represented was work from Sweden, England, the Netherlands, Denmark, France, West Germany, Scotland, Yugoslavia, and the United States.

The Foreword from *The Control of Upper-Extremity Prostheses and Orthoses* is given below.

During the last decade, externally powered upper-extremity prostheses and orthoses have matured into important clinical assets. Among the factors promoting this development are the evolution of semiconductor technology, the control possibilities offered by myo-electric signals and the generally increased concern of society for creating improved aids for the handicapped.

Even though many patients are now successfully fitted with externally powered prostheses and orthoses, it should be stated that most of these devices provide a minor repertoire of functions only. There thus exists a demand for improved functional capability, a demand which is reflected in the research and development efforts of many

medical engineering centers. It is generally agreed that the main technical and biological obstacles encountered in these endeavors are found in the field of control. Control site availability, signal processing, and man-machine interaction including the problems of training are all factors decisive to prosthesis or orthosis acceptance by the patient.

Since a mutual exchange of experience and research results is essential to rapid progress, we decided to arrange a meeting of the foremost international experts in the field of prosthesis and orthosis control. The resulting conference, "The Control of Upper-Extremity Orthoses and Prostheses," was held in Göteborg, Sweden, in October, 1971. The papers presented covered the following four subsections of the field: *Basis of Myo-Electric Control*, *Myo-Electric Control in the Clinic*, *Biomechanical Control*, and *Systems Design*. These papers form the subject matter of the present monograph, and of the experts invited, those able to attend the conference constitute the list of Contributors.

We wish to express our deep appreciation of the substantial support given to the con-

ference by The Swedish Board for Technical Development, The Swedish Department of Social Affairs, The City of Göteborg, and Een-Holmgren Ortopediska AB.

THE EDITORS

MANUAL ON DYNAMIC HAND SPLINTING WITH THERMOPLASTIC MATERIALS—Low Temperature Materials and Techniques. Maude H. Malick, Hamarville Rehabilitation Center, Ridge Road, Pittsburgh, Pennsylvania 15238, 1974, 206 pp., \$11.00.

This is a comprehensive manual on the design, fabrication, and application of dynamic hand splints made from sheet thermoplastic materials. Orthoplast is featured, but descriptions of other suitable materials are given also.

Of especial interest is the rationale for design and application of dynamic splints. This well-illustrated book should be read and kept readily available by all clinical personnel involved in treatment of hand deformities.

RESOLUTION CONCERNING THE METRIC SYSTEM

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

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

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

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

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

METRIC SYSTEM Conversion Factors

LENGTH

Equivalencies

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

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

AREA

To convert from

square inches	square meters	0.00063616†
square feet	square meters	.092903

VOLUME

Definition

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

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

MASS

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

FORCE

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

*This double-prefix usage is not desirable. This unit is actually a nanometer (10^{-9} meter = 10^{-7} centimeter).

†For practical purposes all subsequent digits are zeros.

STRESS (OR PRESSURE)

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

TORQUE (OR MOMENT)

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

ENERGY (OR WORK)**Definition**

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

$$1 \text{ cal (gm)} = 4.1840 \text{ joules}$$

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

TEMPERATURE CONVERSION TABLE

$$\text{To convert } ^\circ\text{F to } ^\circ\text{C} \quad ^\circ\text{C} = \frac{^\circ\text{F} - 32}{1.8}$$

$^\circ\text{F}$	$^\circ\text{C}$
98.6	37
99	37.2
99.5	37.5
100	37.8
100.5	38.1
101	38.3
101.5	38.6
102	38.9
102.5	39.2
103	39.4
103.5	39.7
104	40.0

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

INFORMATION FOR AUTHORS

ORTHOTICS AND PROSTHETICS

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All submitted manuscripts should include:

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3. LEGENDS. List all illustration legends in order, and number to agree with illustrations.
4. ILLUSTRATIONS. Provide any or all of the following:
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 - b. Original drawings or charts
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4. Write out numbers less than ten.
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6. Use the word "Figure" abbreviated to indicate references to illustrations in the text (. . . as shown in Fig. 14)

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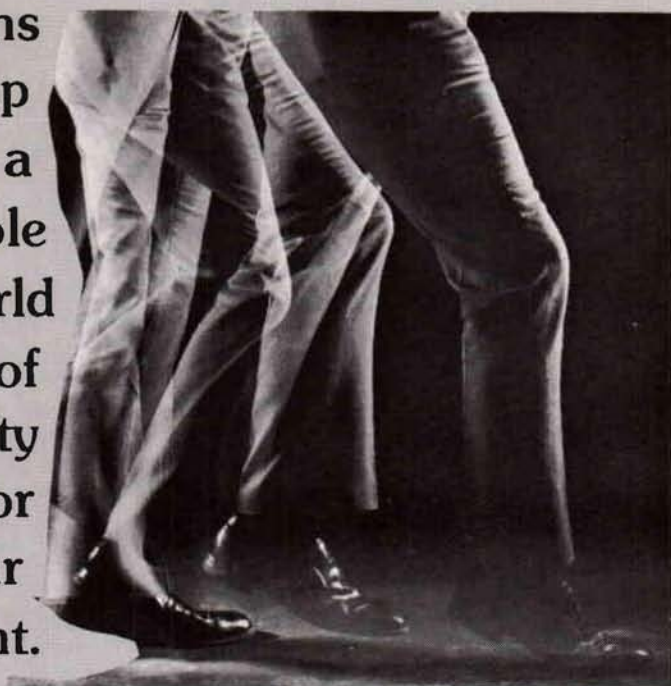
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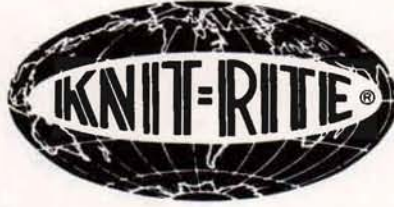
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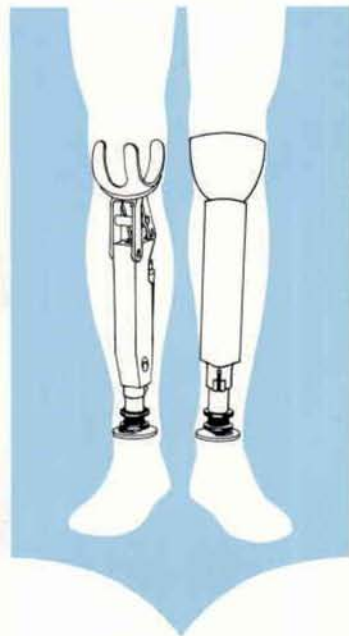
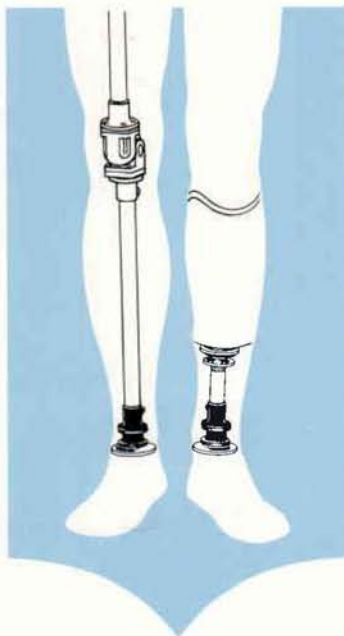
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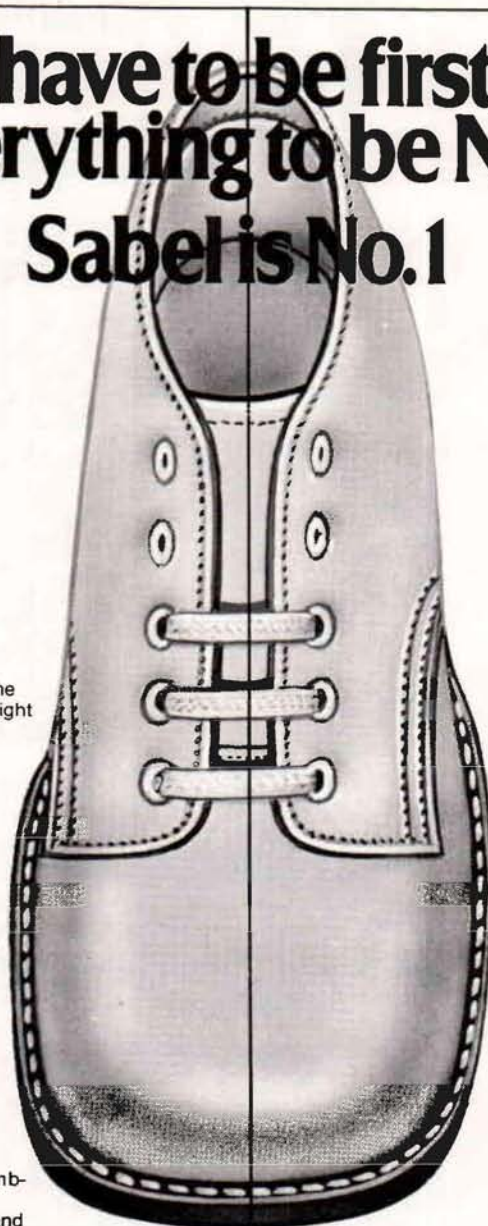
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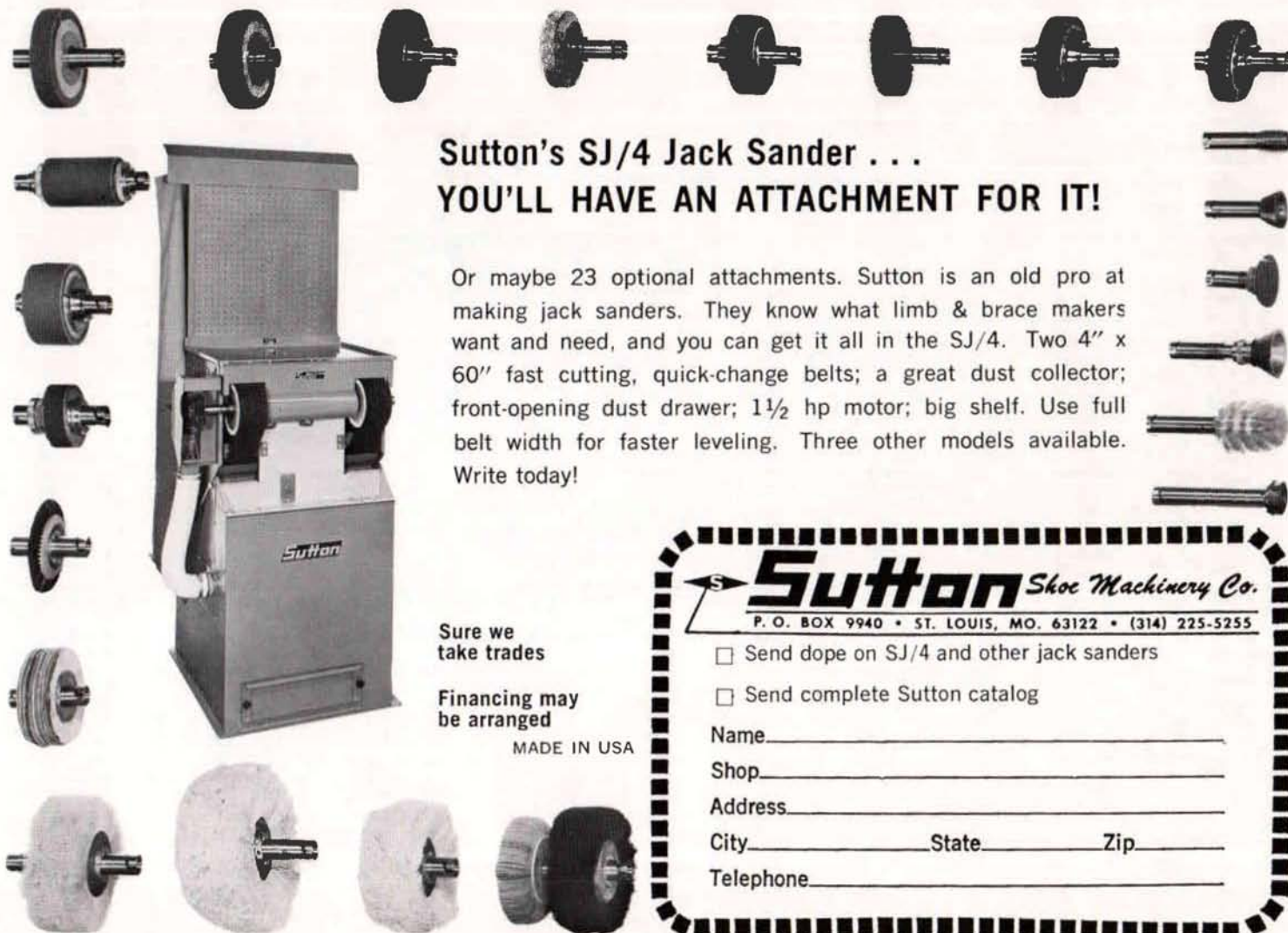
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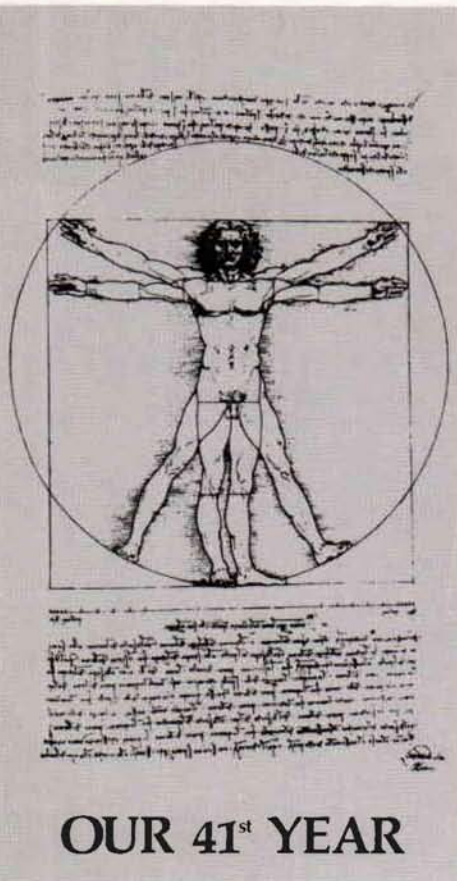
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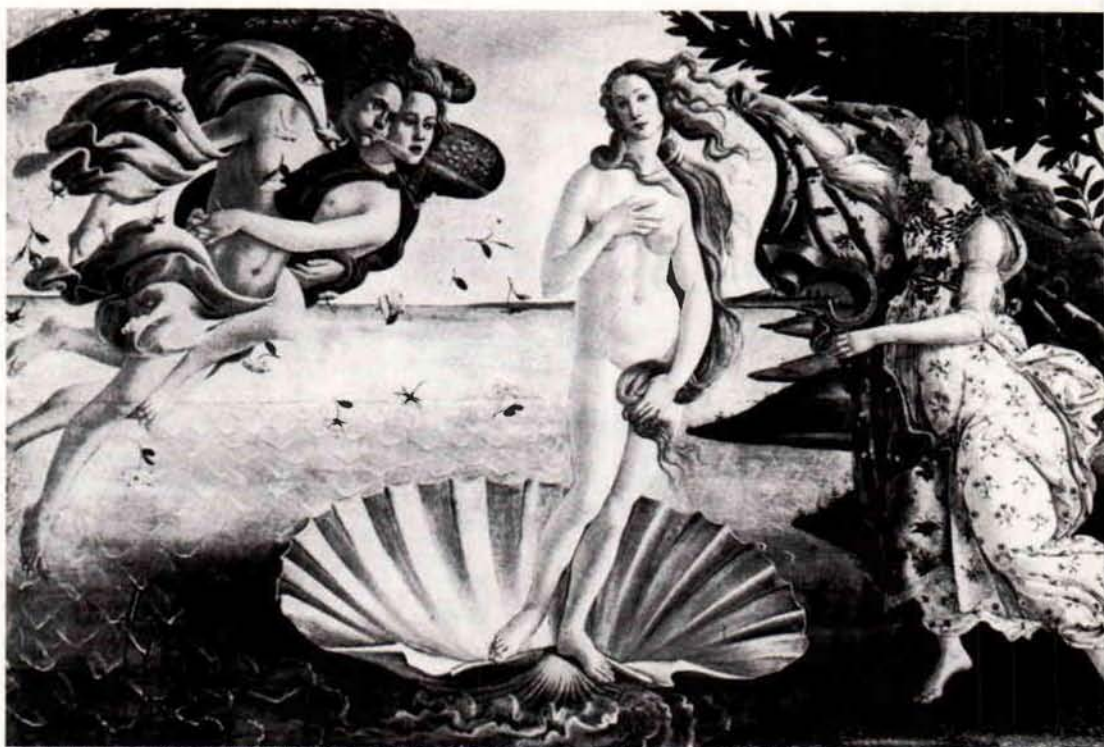
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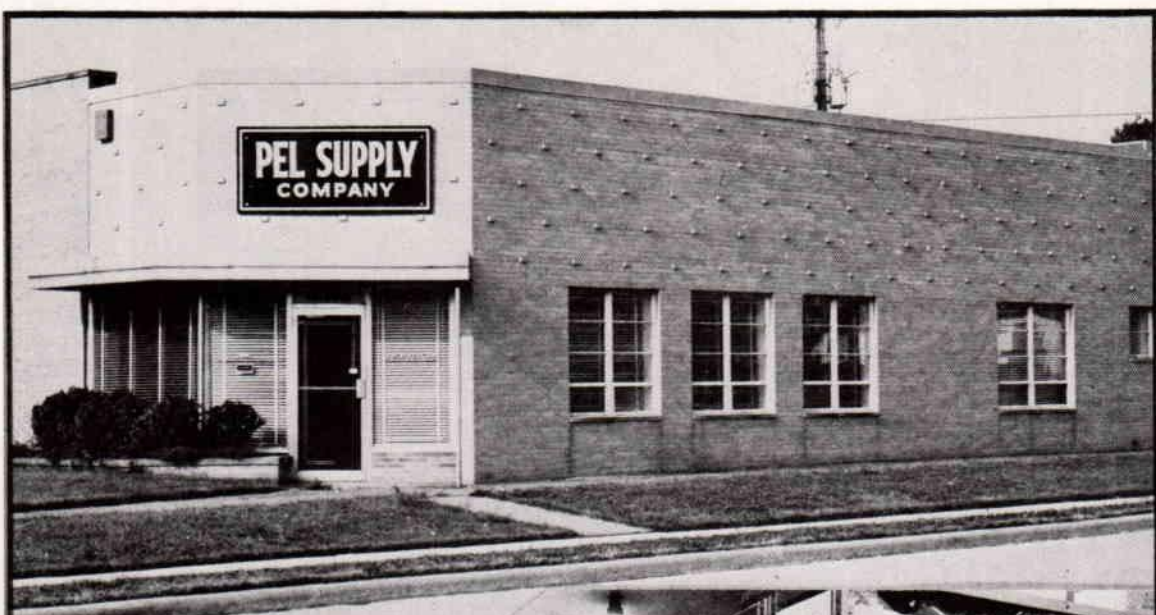
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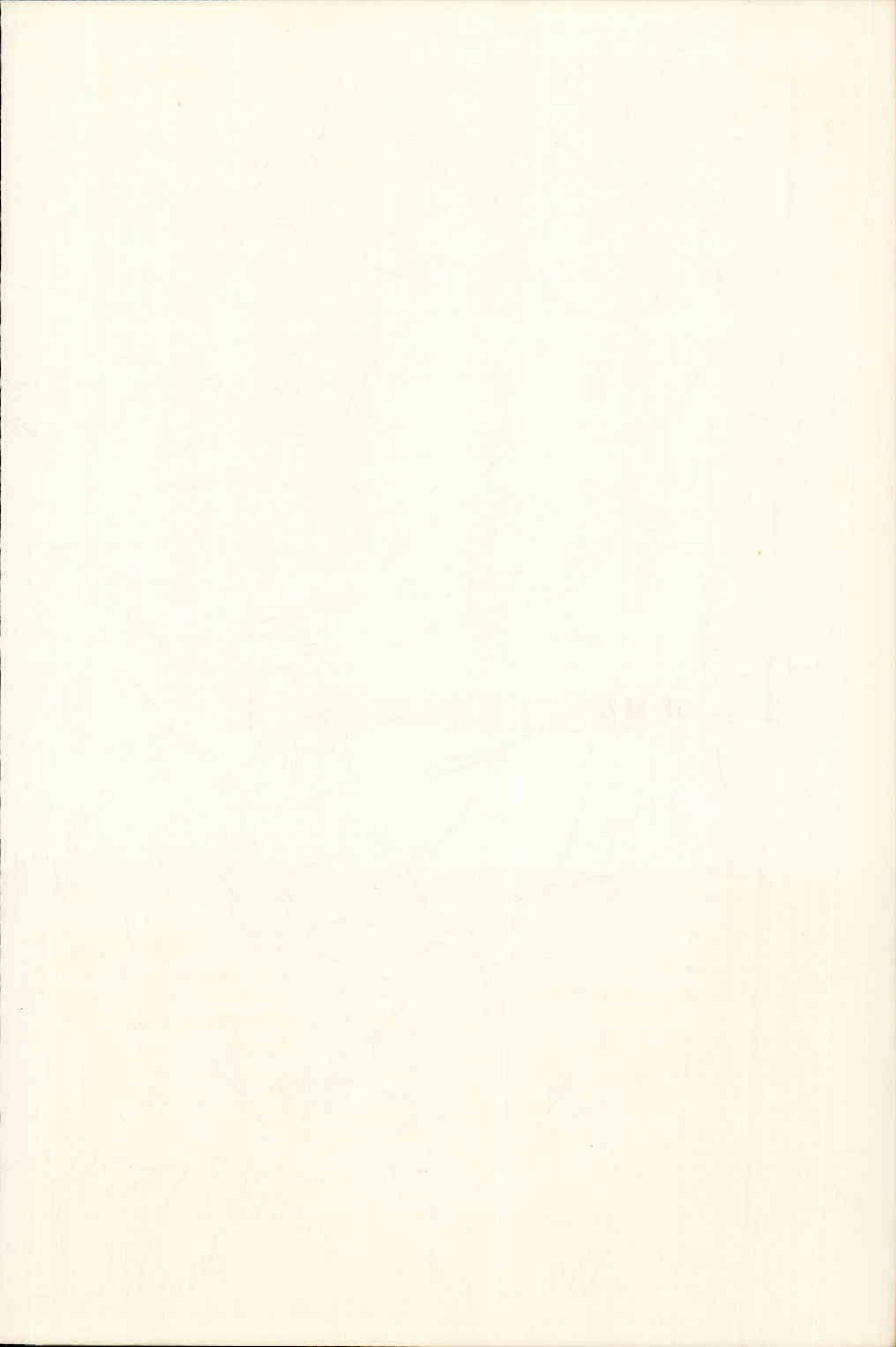
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