SEPTEMBER 1975

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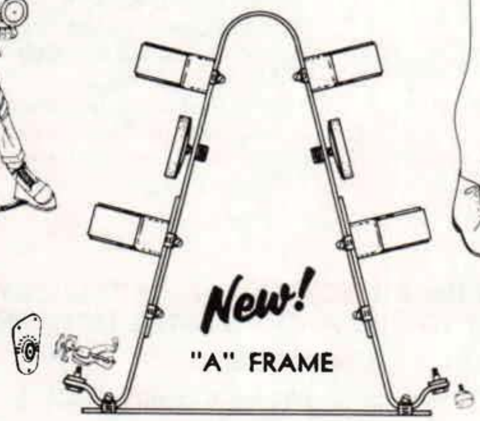

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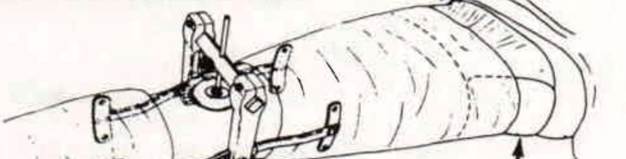

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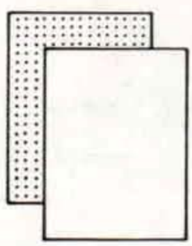

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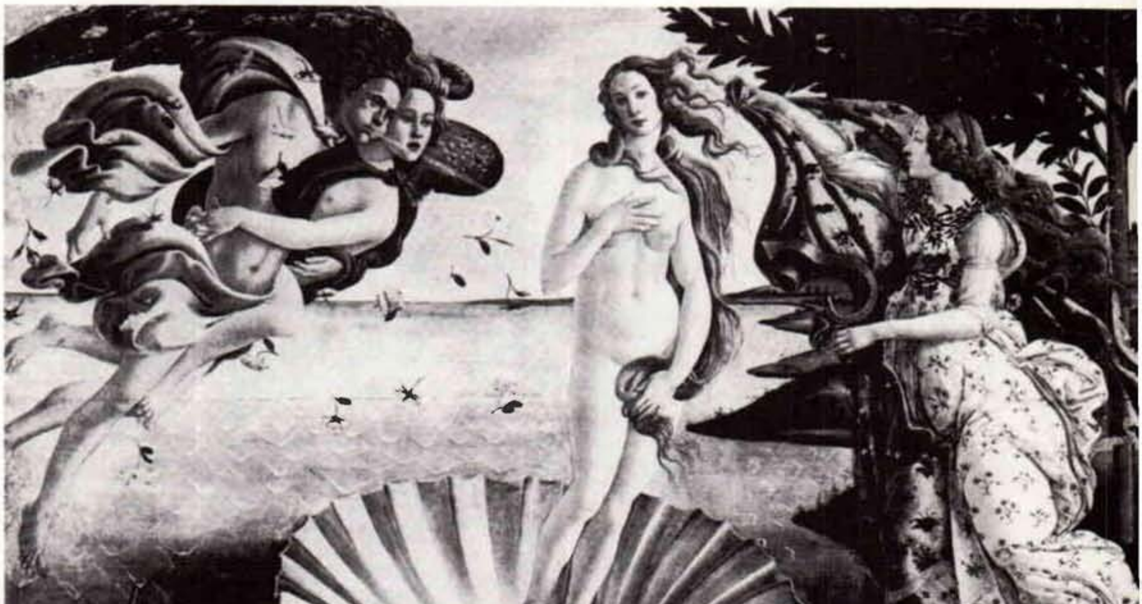

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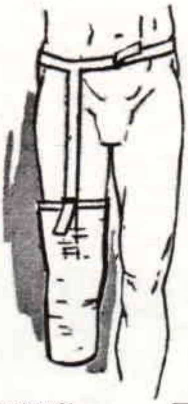

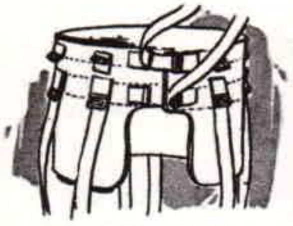

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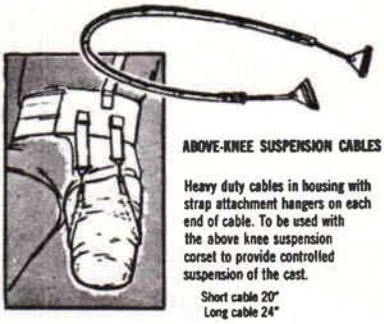

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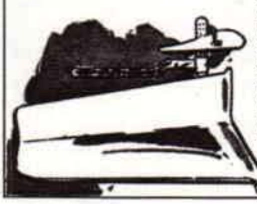

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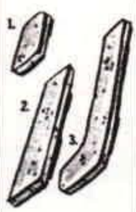

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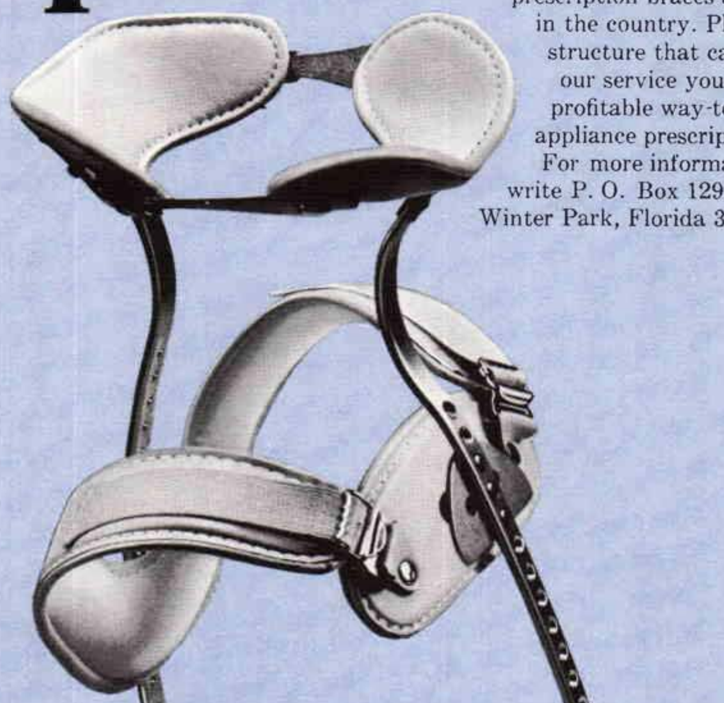

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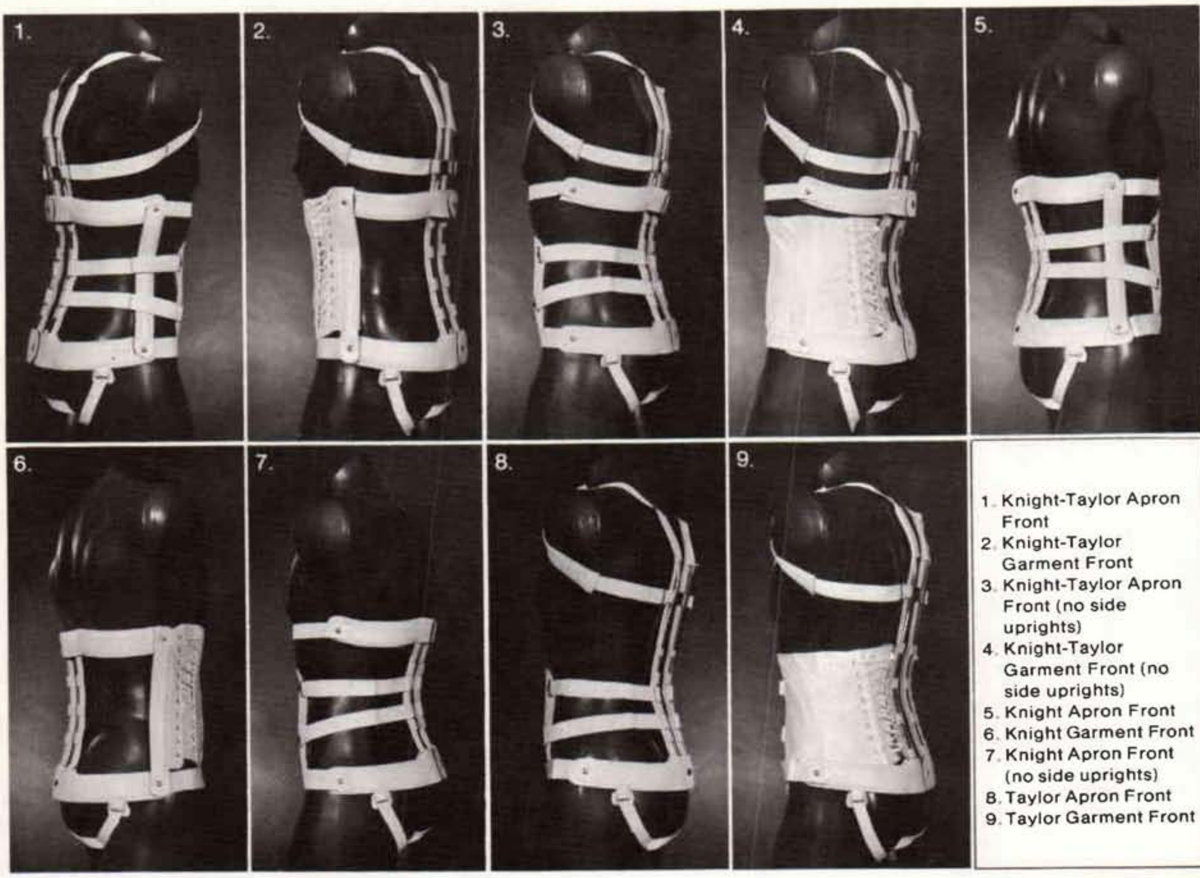

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THE CHALLENGE OF A NEW AGE

In many ways, the orthotic and prosthetic professions in the U.S. have come to a crossroads in their march towards true professionalism. With the right kind of perception and guidance, they can move ahead into the mainstream of well recognized health care providers. With anything less, they can become too easily diverted from their main mission. Hence, perhaps more than at any prior time in history, careful planning, a clear charting of future courses, and a well defined set of goals now become imperative.

Prompting this observation is the fact that a major era in the escalating objectives of the American Board for Certification in Orthotics and Prosthetics has just come to an end. No more will it be possible for aspiring orthotists and prosthetists to become certified with the base credentials of a high school diploma and three short term university level courses in their field of specialty. ABC is now embarked upon a four year transitory phase preparatory to inception of its next major educational objective in 1980, namely the attainment of a bachelor's degree in these two disciplines.

This interval will be a difficult one for the Board as it seeks to develop some commonality of education and experience which will assure that successful certification examination candidates from 1976 through 1979 demonstrate a level of proficiency at least equal to and desirably superior to that registered by their predecessors through 1975. That this will be the case seems likely in view of the Board's interim requirements for a minimum of two years of college training, desirably leading to an Associate of Arts degree in orthotics or prosthetics.

Time will tell how realistic this educational requirement is to administer and also whether it meets increasing demands for more certified practitioners. Happily, the influx of new certifees to date appears to have compensated for some earlier attrition in the professions. The 1975 examinations for example, are expected to add in excess of 250 new Certified Orthotists, Certified Prosthetists, or Certified Prosthetists-Orthotists to the professions. This record total no doubt is due to the realization by many applicants that this was the last year in which they could hope to be admitted under the now expired high school plus three short course requirement.

How manpower needs are met in the immediately foreseeable future as more stringent educational criteria are imposed is difficult to predict, too. But it will be highly important, as indicated above, for ABC and its sister organizations, AOPA and AAOP, to assure that the public need for adequate numbers of qualified personnel can be met. How well this challenge is understood and responded to could be critical.

David A. H. Roethel
Executive Director

Orthotics and Prosthetics, Vol. 29, No. 3, pp. 3–10, September 1975

AN EXTERNAL-CRUCIATE-LIGAMENT ORTHOSIS[1]

Thomas A. Martin, C.O.[2]

Orthopaedic surgeons assigned to our regional health care center observed that many patients with internally deranged knees were unable to participate in strenuous athletic activities. We, as orthotists, recognized the validity of their observation and furthermore recognized that bracing knees in the traditional manner is not satisfactory for patients who wish to engage in contact sports. Both the leather, polycentric-knee orthosis and the rotational-control orthosis (Fig. 1) are inadequate to control fore-and-aft displacement of the joint components because the axis or axes of the mechanical knee joints are confined to a single plane. We were challenged to design and construct an orthosis that would guide the knee through a helical arc. We believe we have been successful in fabricating with plastic materials an effective orthosis for the knee which enables patients to engage in athletics. In this paper, the external-cruciate-ligament orthosis (Fig. 2) is described.

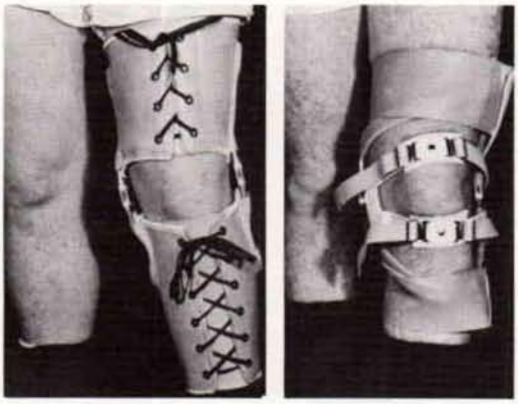

Fig. 1. Contemporary approaches to providing anteroposterior stability to the knee joint. *Left*, knee orthosis with leather cuffs and polycentric joints. *Right*, the "derotation" orthosis.

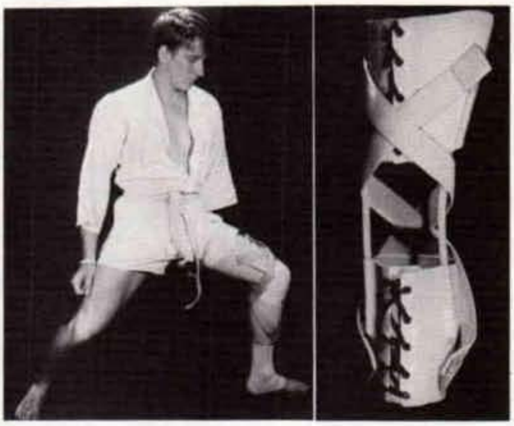

Fig. 2. The external cruciate orthosis demonstrated by a patient who is a student of karate.

ANATOMICAL CONSIDERATIONS

The femoral condyles slide and roll on the tibial condyles as the knee joint is flexed. The amount of sliding and rolling determines the axis of rotation at any given instant. Excessive sliding results in displacement or subluxation fore and aft (Fig. 3). The quadriceps, hamstrings, and the gastrocnemius assist in controlling fore-and-aft displacement, but sound ligaments are essential for absolute control. Fore-and-aft displacement between the condyles imposes tremendous strains upon sound ligaments, tendons, and cartilage. Normally, subluxation is prevented by the anterior and posterior cruciate ligaments and the posterior capsules.

Orthopaedic surgeons at Letterman Army Medical Center (LAMC, 1972–74) are convinced that many severely internally deranged knees are really the result of a two-stage insult. The initial

[1]Presented at the American Orthotic and Prosthetic Association (AOPA), National Assembly, Atlanta, Ga., Oct. 23–26, 1974. The opinions or assertions contained herein are the private views of the author and are not to be construed as official or as reflecting the views of the Department of the Army or the Department of Defense.

[2]Orthopaedic Brace and Limb Section, Orthopaedic Service, Department of Surgery, Letterman Army Medical Center, Presidio of San Francisco, Calif., 94129.

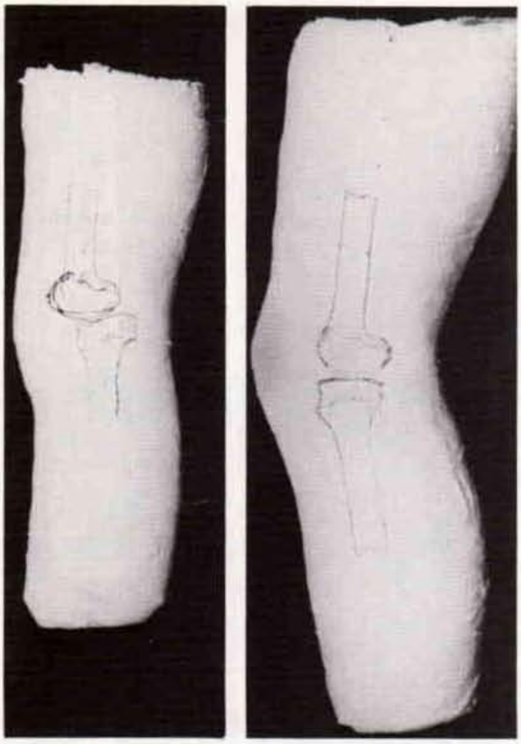

Fig. 3. The cast on the left reflects subluxation of the knee joint, probably resulting from a severed or otherwise damaged posterior cruciate ligament. The cast on the right reflects correct alignment of the tibia and femur.

injury may be a tear of one or both cruciates, followed by a progressive laxity of stabilizing ligaments on the posterior aspects of the knee joint. Continuous or strenuous activities after injury to a cruciate ligament contribute further to laxity of the ligamentous structures surrounding the joint, a condition which contributes to a change in the glidepath of the joint, and compromises the meniscii.

SELECTION OF MATERIALS FOR ORTHOSIS

We proposed that an all-plastic knee orthosis would be desirable and possible. Polypropylene can assume high compressive and tensile loads, and we selected polypropylene rod for use as "knee joints." Polypropylene has been used for fracture bracing, and has been used effectively for a below-knee orthosis; therefore we had good reason to believe that it could lend, in the form of an orthosis, excellent stability to the knee.

On a knee orthosis, the cuffs must be strong enough to carry the loads through to the polypropylene rod and elastic straps. The fit must be snug to ensure that no piston action occurs between the orthosis and the limb. A polyester-resin laminate was an obvious choice of material because it is impervious to water, perspiration, and radical temperature changes. The polyester resin can be mixed to provide a laminate as rigid or flexible as desired. An additional advantage of polyester resins is that they are incompatible with polypropylene, and should one of the rods break it can be replaced easily. Nylon stockinet and fine mesh fiberglass were chosen as the filler materials. Elastic webbing strap was used because it was cheap and available.

EVALUATION OF THE KNEE

Evaluation of the knee before making the plaster wrap is a prerequisite for a well-fitting useful orthosis. The evaluation must reveal: a) the cruciate ligament that is injured and needs support; b) the amount of pressure that is acceptable over the femoral condyles; and c) the activity in which the person expects to participate. Cast modification, the amount of flexibility of the cuffs, and placement of the rods and straps are determined using the results of this evaluation.

FABRICATION

A cotton stockinet is applied over the length of the limb, and a plaster wrap is taken with the patient sitting with the knee in 10 to 15 deg. of flexion. The outline of the patella, the apex of the fibular head, the prominence of the hamstring tendons, and the mediotibial plateau (MTP) are marked. Wrapping is begun from the lower third of the tibia, just superior to the malleolus, to the upper quarter of the thigh (Fig. 4). The first layer of elastic plaster-of-Paris bandage is molded closely to the limb. Two additional layers of standard plaster bandage provide the rigidity necessary. Certain precautions must be taken during wrapping to prevent tibial rotation and subluxation of the knee. When the patient keeps visual contact with the medial border of his foot while the wrap is being taken, excessive tibial rotation is avoided. To minimize subluxation, the knee must be supported during the entire wrapping and plaster-setting procedures. We contend that an accurate wrap, correctly molded, will limit the amount of positive mold modification needed later.

The positive mold is obtained in the traditional manner. We find that the lamination is easier when the pipe and long axis of the wrap are perpendicular to the floor. Preparation of the positive model consists only of removing and smoothing the plaster. Usually, plaster buildups are not

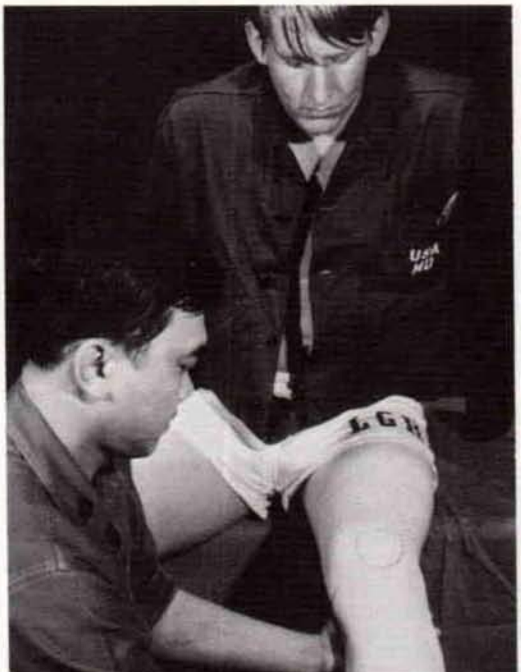

Fig. 4. In wrapping the cast the foot and ankle must be kept in alignment and subluxation must not be permitted to take place. The areas about the tibial flare, the shaft of the fibula, the distal femoral condyles, and the hamstring tendons must be molded closely.

necessary. Radical modifications include the removal of a predetermined amount of plaster above the femoral condyles. The tibial flare is blended to sustain load. The shaft of the fibula is loaded in a manner similar to the lateral stabilization provided in the below-knee prosthesis. The goal of this is not to provide lateral stability, but to provide rotary control of the orthosis against the leg. Posteriorly, the popliteal cavity is exaggerated and blended into the area of hamstring and gastrocnemius muscles. Care must be taken not to modify severely the impressions of the hamstring tendons. The MTP is measured and recorded from either end so it may be found later for definition of the trim line.

After radical modifications have taken place, the stockinet marks are removed from the entire surface of the mold with a Surform file, with the exception of the sensitive areas, i.e., crest of the tibia, tibial tubercle, fibular head, and patellar surface. After the mold has been prepared, its entire surface is then wet-sanded and all traces of stockinet marks are removed.

The materials needed for fabrication of the orthosis are:

1 layer 0.5 oz Dacron felt
5 layers nylon stockinet
2 layers tricot tube
2 strips (3 in.) fine mesh fiberglass
4 polypropylene rods 3/8 in. in diam.

A layer of Dacron felt is applied first, followed by two layers of nylon stockinet. Fiberglass strips are placed at the same location as metal bands would be on a conventional knee orthosis and serve the same purpose. The ends of these strips are rolled around to form "holsters" for the ends of the rods which are placed vertically.

Using the rationale offered by Radcliffe and Foort (*2*) for determination of the correct placement of the side joints on the conventional below-knee prosthesis, we concluded that the rods should be applied slightly posterior to the midaxial line of the knee if the anterior cruciate is to be supported; conversely, the rod is to be placed anterior to the midaxial line if the posterior cruciate ligament is to be supported. The rods and the fiberglass (holsters) are covered with one layer of stockinet (Fig. 5) and one layer of tricot. A separator is applied anteriorly to form a tongue. A second layer of tricot is pulled over the model, and the final two layers of stockinet are applied.

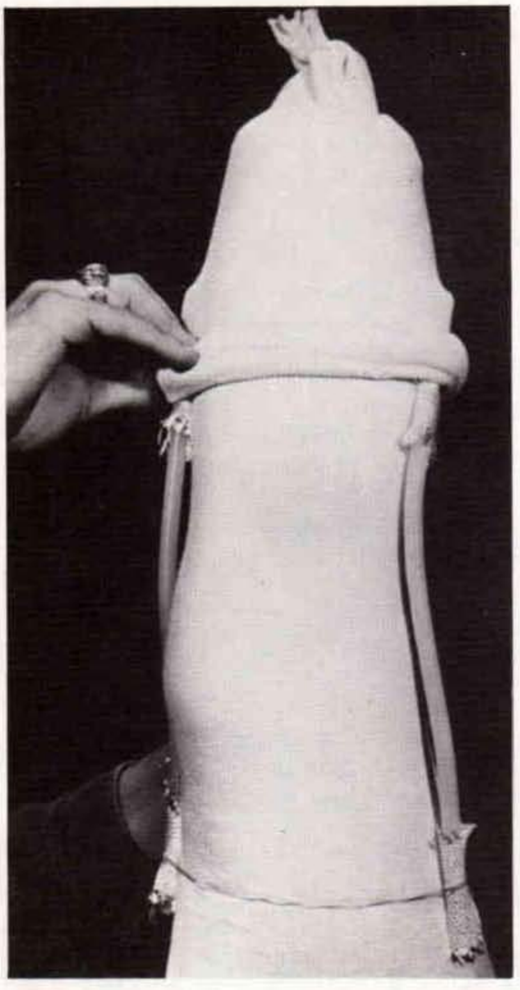

Fig. 5. The polypropylene rods and fiberglass "holsters" are covered with a layer of stockinet, followed by a layer of tricot.

The amount of flexibility selected for the cuff is based upon the activity and objectives of the individual patient. We use a combination of laminating resins, under 10 to 15 lb pressure, to acquire the appropriate rigidity in the orthosis for each patient. We have most frequently used ratios of 75:25 and 80:20 of polyester resin 4134 to polyester resin 4110. We realize that there are differences in the requirements of one who is primarily a motorcyclist and one who is a student of karate. The motorcycle rider's orthosis is flexed and exposed to constant vibration; it should be more rigid than the orthosis worn by the karate student.

When the resin has set up, the laminate is removed from the model, and vertical cuts are made before the horizontal (distal and proximal) circumferential cuts. The initial cut with a sharp knife is made precisely down the middle to the separator. After the entire layup has been cut along its entire length, the required flap is folded back, and another vertical cut is made underneath the flap and through the model (Fig. 6). After the horizontal cuts are made the entire layup is wrapped with an Ace bandage, and the final curing is allowed to take place.

The straps are usually made while the orthosis is curing. Their paths, or routes of travel, are determined by the cruciate ligament that needs to be supported. The "origins" of the straps are

Fig. 6. Vertical cuts are made in the anterior portion of plastic laminate to provide torque for distribution of loads between the cuffs and the soft tissues of the leg and thigh. The first layer is cut directly through the middle. The cut in the second layer, of course, is offset.

distal to the rod ends. One originates on the medial side and one on the lateral side of the leg. The straps spiral around the leg and "insert" on the same side of the limb above the knee. If the anterior cruciate ligament is severed, the straps wind anteriorly from their orgins: conversely, if the posterior ligament is torn, the straps spiral posteriorly. Both ends of the straps should be reinforced with a material such as Naugahyde to prevent fraying. After the resin has cured, the site of the mediotibial plateau is located, transferred circumferentially, and the trim lines drawn on the lamination (Fig. 7). The mediolateral distal trim line of the thigh cuff should be equidistant from the MTP as the mediolateral proximal trim line of the calf cuff. The posterior distal thigh cuff should be equidistant from the posterior proximal calf cuff. The anterior trim lines can be determined proximally at a point 3/4 in. superior to the patella and distally at the apex of the tibial tubercle. The respective points are joined, and trim lines are cut through the plastic, including the polypropylene rods. Attempts to salvage the rods have been unrewarding. The polypropylene must always present a smooth surface because the slightest nick in the material greatly magnifies its fracture potential. An accurate fit is determined and new rods are inserted at the initial fitting.

At the time of initial fitting we ensure that there is no piston action and that there is ample room for flexion; also, we determine the proximal attachment point for the elastic straps (Fig. 8). We put the orthosis on the leg over a stockinet with the knee in the same degree of flexion as it was during the casting procedure. The orthosis is held in position by applying nylon reinforced tape. The individual is asked to stand and then to sit again. Any indication of piston action must be

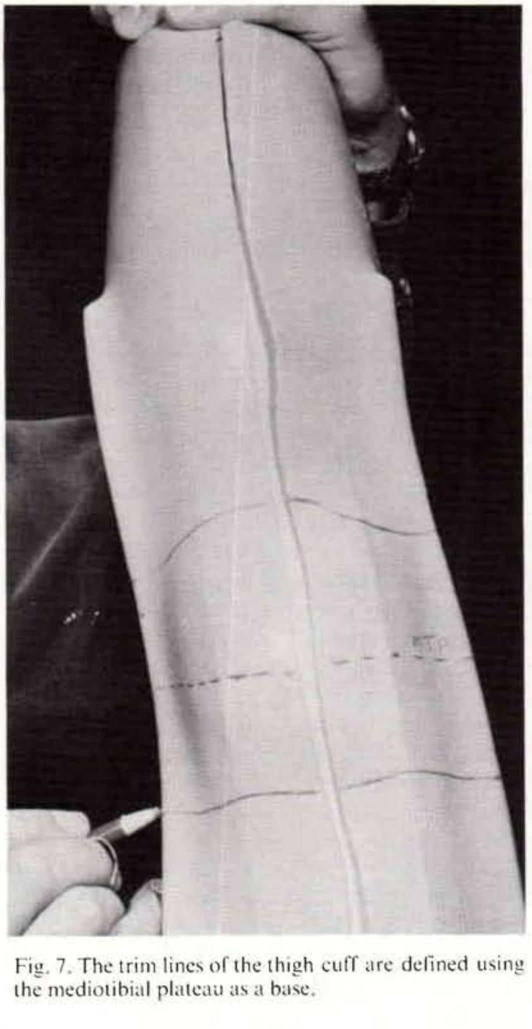

Fig. 7. The trim lines of the thigh cuff are defined using the mediotibial plateau as a base.

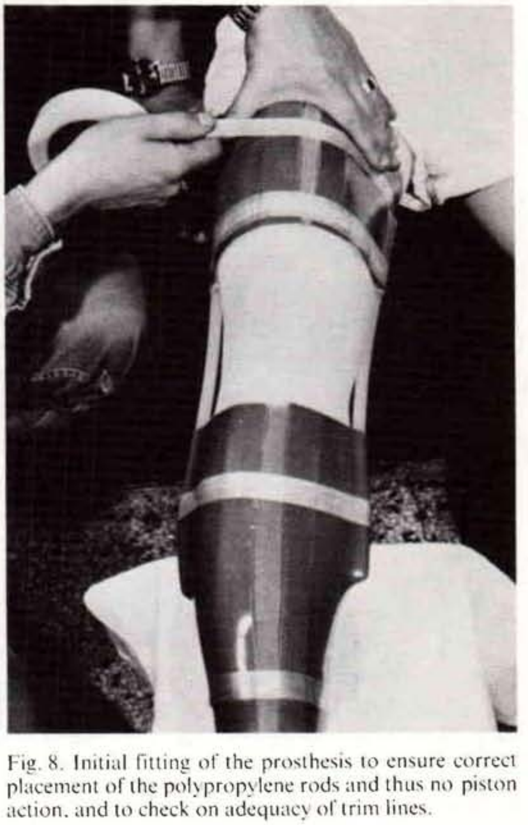

Fig. 8. Initial fitting of the prosthesis to ensure correct placement of the polypropylene rods and thus no piston action, and to check on adequacy of trim lines.

eliminated either by tightening the closure with tape or retrimming the periphery of the orthosis at the knee joint. If the patient can flex his leg to 90 deg. while sitting, it has been our experience that acceptable trim lines have been made and enough rod is exposed. *This is a dynamic orthosis, and to impede other motion or to require an excessive amount of energy to flex the knee is contrary to its design and principle.* The proximal attachment points are determined by the amount of tension needed for each individual. When the amount of elasticity has been determined, patellar-tendon-bearing (PTB) studs are inserted into the cuffs and the initial fitting is concluded (Fig. 9).

Fig. 9. Determination of the location of the elastic straps. The "origins" are distal to the calf end of the rods and the "insertions" are proximal to the thigh end of the rods. The path is determined by the nature of the injury.

EXPERIENCE

During the two-year period that we have fabricated the external-cruciate orthosis, we have fitted 32 patients, the age range being 19–82 yr., the weight range, 120–220 lb., and the height range, 5 ft. 2 in.–6 ft. 4 in. One patient was unsuccessful, probably because he was a victim of multiple combat-inflicted injuries which included brain damage, and he has not been able to cooperate in a therapeutic regimen. Ten patients were lost to follow-up evaluations. Table 1 shows the length of time 15 patients wore the orthosis and their ages. There are two women in the group, and the 68-year-old woman is the only patient on whom we have fitted the orthosis bilaterally. The six patients not listed in the table paralleled the picture of the typical candidate—a man in his middle 20s, who has an old knee injury from repeated trauma sustained in intramural or varsity sports; he wishes to continue participation in "backyard" activities, individual sports, and is on active duty with the Army.

CONCLUSIONS

Our orthosis is designed for active people—persons who want to continue to participate in a sport even though stability of the knee has been compromised by a previous injury. Failures have not occurred because of cold, wetness, rotational shearing, or sudden temperature changes while wearing this orthosis.

The experience in designing and fabricating the external-cruciate-ligament orthosis emphasizes the importance of fore-and-aft subluxation of the knee. It also has shown that orthotists can provide a supportive appliance made from plastic materials which will *stabilize* effectively a deranged knee so that an athlete may continue to participate in active sports.

We want to reassert our total agreement with Thorkild Engen (*1*) who suggested at an Academy seminar in 1972 that old concepts, traditional rationale, and historical conclusions must be reevaluated by orthotists when using plastics in fabrication of appliances. The external-cruciate-ligament orthosis made from polypropylene meets the standards of our profession and the needs of those with deranged knees who wish to participate in sports.

TABLE 1. SUMMARY OF PATIENTS WHO ARE WEARING THE EXTERNAL CRUCIATE ORTHOSIS

Case No.	Age	Cause of Injury	Extent of Injury	Comment
1	30	Trauma	Posterior cruciate	First patient to use external cruciate orthosis (ECO); active-duty Army doctor who wears ECO for jogging and softball. Has used ECO over two years.
2	30	Trauma; surgery	Posterior cruciate and other internal weakness	Active-duty military policeman who uses ECO primarily while participating in company athletics (practice and competition).
3	22	Trauma from participation in sports; Surgery	Posterior cruciate and other internal damage	Active-duty soldier, 6 ft. 4 in. 220 lb., who uses ECO primarily for skiing, tennis, and team sports.
4	24	Trauma	Posterior cruciate	Wears ECO primarily while playing his favorite sport—handball.
5	45	Old injury; repeated trauma	Posterior and anterior cruciate ligaments	Active-duty physician; sports participant, primarily wears ECO while playing tennis.
6	23	Trauma	Medical meniscus tear with posterior cruciate injury	Active-duty Army nurse (woman) who wears ECO primarily while skiing.
7	68	Arthritis	No patellae; joint deterioration; spurs; generalized muscle and ligamentous weakness	Woman, bilateral ECOs; one ECO is primarily for support; wears ECOs every day for all activities. Has Velcro enclosures because of arthritic hands.
8	28	Trauma from varsity football	Posterior cruciate, mediolateral instability	Lawyer, 6 ft. 3 in. who continues to participate in sports—basketball, tennis, skiing.
9	19	Trauma; surgery	Posterior cruciate	Karate participant and wears ECO during this activity.
10	27	Trauma	Anterior cruciate; excessive hyperextension	Uses ECO for sports participation, particularly basketball. Because of excessive hyperextension, an extra strap has been added posteriorly.
11	35	Trauma; surgery	Anterior and posterior cruciate ligaments; patellectomy with patellar prosthesis; hyperextension	By wearing ECO, he is able to stay on active duty as a air-traffic controller.
12	28	Trauma	Posterior cruciate; mediolateral instability	Uses ECO to ski and play handball; active-duty physician stationed in another country.
13	25	Trauma from collegiate football participation; surgery	Posterior and anterior cruciate ligaments; (medial compartment surgically reconstructed)	Uses ECO while skiing and playing softball; alternates with derotation orthosis to see which is more effective.
14	23	Trauma	Posterior, anterior and collateral ligaments	Used ECO originally for karate, but has strengthened quadriceps muscles sufficiently so that he only uses ECO while skiing.
15	82	Arthritis and trauma	Fracture distal femur; mediolateral instability; hyperextension	Uses ECO in order to stand; wears it over a fracture-brace stockinette; hyperextension strap works well.

ACKNOWLEDGMENTS

I wish to thank Lt. Col. John A. Feagin, Jr., Medical Corps, United States Army, and Col. Sterling Mutz, Medical Corps, United States Army, Chief, Orthopaedic Service, Letterman Army Medical Center, for their inspiration, motivation, and instruction during the period that we were evaluating the knee-joint functions and designing the external-cruciate-ligament orthosis. SFC Daniel E. Anderson, C.P.O., was in charge of the Orthopaedic Brace and Limb Section during the period the external-cruciate-ligament orthosis was fabricated originally.

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A KNEE-STABILIZING ANKLE-FOOT ORTHOSIS

Gustav Rubin, M.D., F.A.C.S.[1],
and Michael Danisi, C.O.[2]

The knee-locking ankle-foot orthosis (KAFO) designed by Saltiel (*2*) appeared to be an ideal replacement for the double-bar KAFO customarily used for patients with paralysis of the quadriceps. However, initially we were unable to apply the concept successfully.

The problems encountered came about because of Saltiel's description of the basic fabrication technique, i.e., "it is indispensable to maintain the foot in a fixed equinus position because the force acting at the knee acts only as long as *the heel does not touch the floor*. (Italics ours.) The moment (*instant that*) the heel touches the floor the brace ceases to be functional." (*2*)

When the orthosis is fabricated and applied in the prescribed fashion, we found that the force exerted at the proximal-anterior segment of the orthosis produced pain. When this dominant complaint was eliminated, another cause of extreme discomfort came to the fore: the stresses produced in the popliteal area elicited a response to pain and then to patient rejection of the device because of the excessive hyperextension produced.

The AFO developed at VAPC eliminated both of these problems when certain conditions were met:

1) a knee-flexion contracture does not exist.

2) An adjustable ankle joint is included in the AFO to correlate the degree of equinus with the degree of knee hyperextension. This arrangement permits the heel to make contact with the ground before popliteal stresses can be introduced.

3) The orthosis should be prescribed for patients with impaired quadriceps on one side only. The orthosis should not be prescribed for patients with bilateral involvement.

The VAPC orthosis is illustrated in Figure 1. Proximally, the posterior force is applied at the level of the patellar tendon, and a patellar-tendon shelf is included in the orthosis. This area tolerates pressure well.

Distally, Pope[3] ankle joints are used, incorpo-

Fig. 1. Three views of the VAPC knee-stabilizing AFO.

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[2]Supervisor, Orthotics Laboratory, VAPC.

[3]Pope Foundation, Kankakee, Ill. 60901.

rating a spring in the posterior section and a solid rod in the anterior section to stop dorsiflexion. The dorsiflexion stop allows for equinus just sufficient to stabilize the knee. The amount of dorsiflexion can be controlled and can be "fine-tuned" during dynamic fitting. The spring loading permits adjustment for accommodation for stride variations. When the patient takes a long stride and strikes on the heel, the foot goes into the foot-flat position. With a rigid ankle, a "de-stabilizing" moment is introduced that flexes the knee.

We have found this to be a useful approach within the limits of the previously outlined criteria. However, it is our impression that it may be worthwhile to present not just the end result but all of the experimental devices tried, with a view to preventing others from unnecessarily following unrewarding courses we pursued.

After fabricating the Saltiel AFO as the developer described it (*1*), we decided to use polypropylene rather than a plastic laminate and to apply the anterior force to the patellar tendon (Fig. 2). Even this area could not tolerate the force that developed when the heel was not permitted to make contact with the ground.

In an attempt to overcome this problem, the device was designed according to Lehneis (*1*) (Figs. 3 and 4) to distribute the anterior force over a broad area. The immediate complaint by the patient was related to cosmesis (Fig. 5).

Fig. 2. Polypropylene version of the "Saltiel" orthosis. Included here are efforts to take rather large loads over the patellar-tendon area.

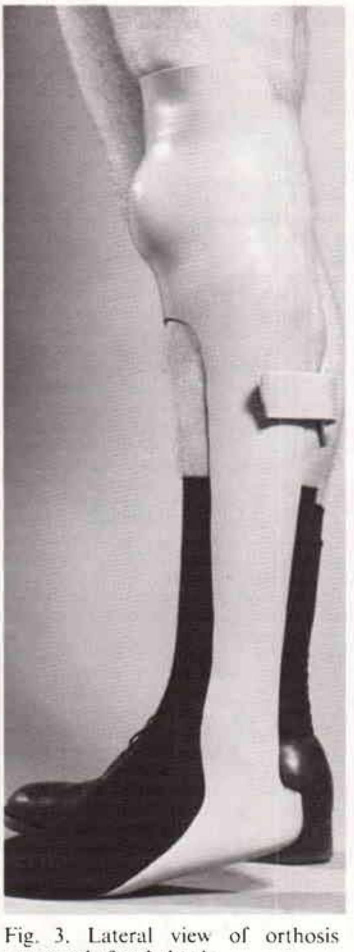

Fig. 3. Lateral view of orthosis patterned after Lehneis.

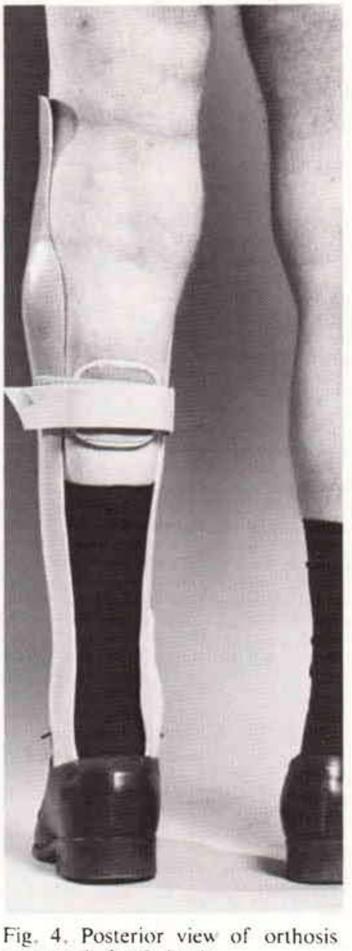

Fig. 4. Posterior view of orthosis patterned after Lehneis.

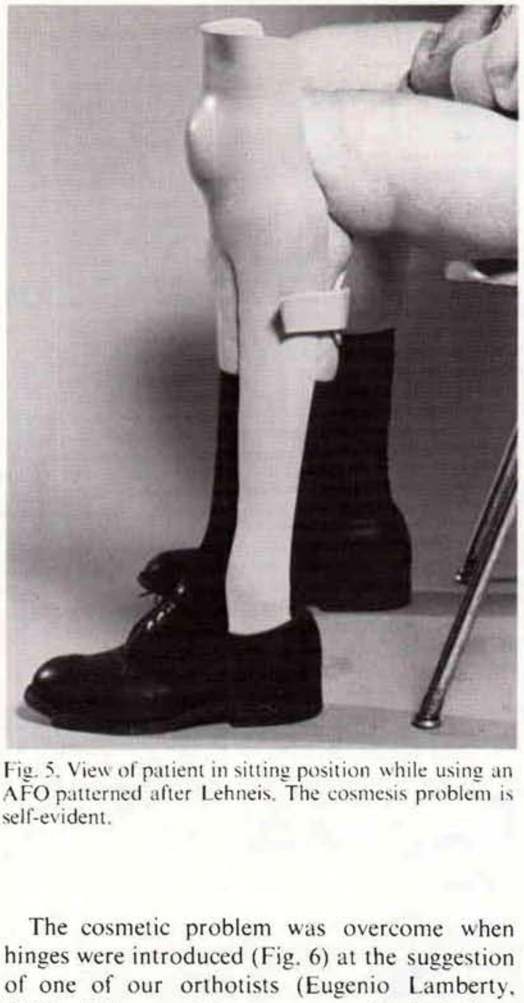

Fig. 5. View of patient in sitting position while using an AFO patterned after Lehneis. The cosmesis problem is self-evident.

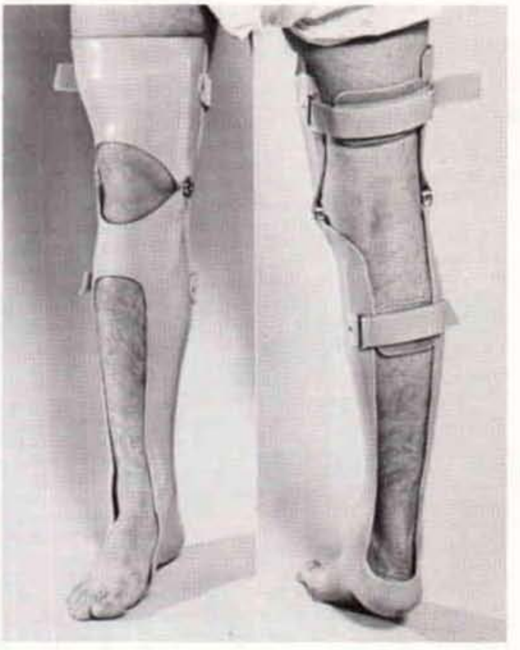

Fig. 6. Application of knee hinges to the Lehneis design to improve cosmesis. Cosmesis was achieved at the expense of excessive pressure in the popliteal area.

The cosmetic problem was overcome when hinges were introduced (Fig. 6) at the suggestion of one of our orthotists (Eugenio Lamberty, C.O.). When satisfactory cosmesis had been achieved the patient became more conscious of the stress in the popliteal area, and it became the principal source of pain and rejection.

This led to the use of a ratchet joint at the ankle (Fig. 7) with the view that the equinus would be adjusted to permit heel contact before excessive hyperextension stresses were introduced at the knee. In this fashion popliteal stresses could be avoided. In fact, the anterior force would be reduced to a minimum and an AFO could be used rather than a KAFO. The adjustments permitted by the ratchet joints were too coarse, but the principle seemed to be worth pursuing.

The orthosis reported at the beginning of this presentation was the final development and one that the authors believe to be useful, within the limits set forth.

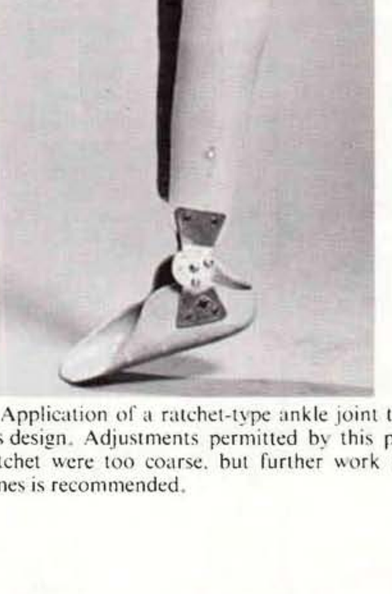

Fig. 7. Application of a ratchet-type ankle joint to the Lehneis design. Adjustments permitted by this particular ratchet were too coarse, but further work along these lines is recommended.

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A SURVEY OF PROSTHETICS PRACTICE—1973–74

Sidney Fishman, Ph.D., Norman Berger, M.A., and Daniel Watkins, B.A.[1]

Although members of prosthetics clinics are in contact with their colleagues at regional and national meetings and through professional journals, it is difficult for them to obtain precise information about the nature of the amputee population as well as the prescriptions, components, and techniques which prove successful and are in common usage in other clinics. Over the years several studies have been conducted to supply such data, notably: the Litt and Nattress report on prosthetic fabrication procedures and component choices, published in 1961 (*3*); the Davies, Friz, and Clippinger study of both amputee population and component prescription during the years 1965–67 (*1*); and Glattly's study, which investigated the composition of the amputee population in the years 1961–63 (*2*).

Because all of these studies are dated, the Prosthetics and Orthotics faculty at the NYU Post-Graduate Medical School initiated a survey to develop as accurate a picture of present practice as practical.

SURVEY METHOD

During the period January 1973 to June 1974 each student enrolled in short-term courses for prosthetists at NYU received a questionnaire[2] concerning the amputees treated at his facility, and the components, procedures, and materials used in fitting these patients. For each item on the 94 questionnaires completed, an average was calculated, and it is this figure that is cited throughout this report as an expression of the experience at the "average prosthetics facility." It should be understood that the raw data consist of estimates made by the prosthetists without the guidance of exact statistics gathered at their facilities. The students were requested to leave unanswered any questions for which they lacked sufficient information. The homogeneity of the responses speaks well for the ability of this group to make accurate estimates.

Geographically, the sample is somewhat biased by the not-surprising fact that most of the prosthetists attending short-term courses at NYU are from states relatively close to New York City. An analysis of enrollment data shows that 50 percent of the students in the courses covered by the survey came from the Northeast, 30.5 percent were from the Southeast, and 19.5 percent from the West (Fig. 1). Although it would be presumptuous to interpret the findings of this survey as accurately depicting national practice, the fairly wide geographical distribution should be kept in mind.

RESULTS

Davies, Friz, and Clippinger (*1*), as well as Glattly (*2*), reported that the amputee population that receives artificial limbs consists of 14 percent upper-limb and 86 percent lower-limb patients, which is identical with our results. In the present survey more than half of the amputations (53%) of the upper-limb group, are below-elbow, 29 percent are above-elbow, with partial-hand and shoulder amputations each accounting for 9 percent (Table 1). It is interesting to note, however, that one-quarter of the prosthetists surveyed see no partial-hand patients and one-third see no shoulder amputees.

Just as below-elbow amputees comprise a majority of the upper-limb patients, so are below-knee amputees a majority of lower-limb patients (56%). Another third are above-knee amputees, with these two levels making up nearly 90 percent of the patients seen. The percentages of Syme's (7%) and hip-disarticulation or hemipelvectomy amputees (4%) reported in this survey are somewhat higher than in previous studies, which placed the size of these groups at about 3 and 2 percent, respectively. The larger percentages indicated by the prosthetists surveyed here may

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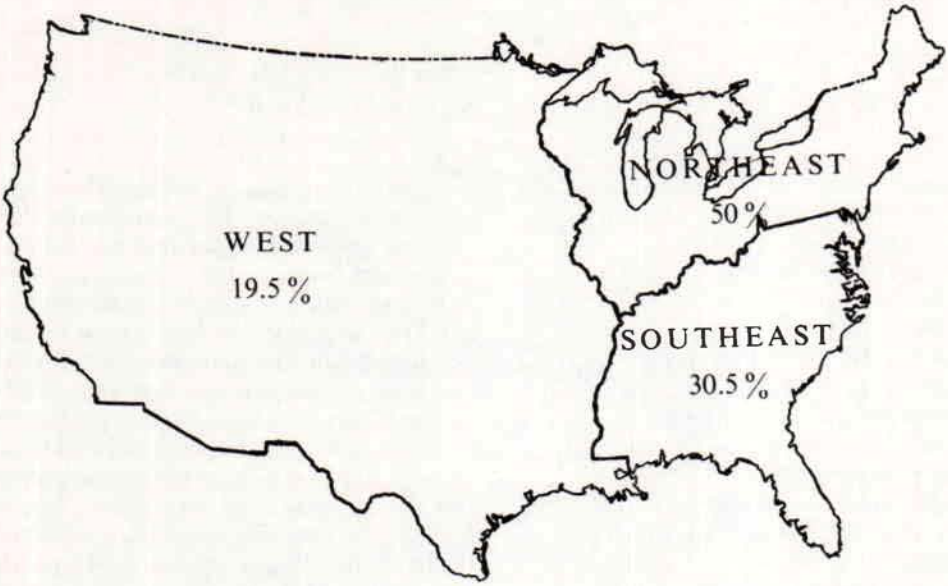

Fig. 1.

reflect some increase in the number of such amputations, or may be simply an inaccurate estimate.

The distribution of patients by age varies in one important and expected way between upper- and lower-limb groups. The proportion of patients under 18 years is quite similar, but a greater number of individuals over 60 are found in the lower-limb group while more of the upper-limb patients are between 18 and 60. The causes of amputation

TABLE 1. PATIENT POPULATION BY AMPUTATION LEVEL AND AGE

UPPER-LIMB AMPUTEES (14%)	%	LOWER-LIMB AMPUTEES (86%)	%
Partial hand	9	Syme's	7
Below elbow	53	Below knee	56
Above elbow	29	Above knee	33
Shoulder	9	Hip disarticulation and hemipelvectomy	4
TOTAL	100	TOTAL	100
Bilateral	3	Bilateral	5
Under 18 years	21	Under 18 years	18
19–60 years	71	19–60 years	60
Over 60 years	8	Over 60 years	22
TOTAL	100	TOTAL	100

explain this difference: most upper-limb amputations result from trauma to relatively young, active persons, while lower-limb amputations are more often caused by peripheral vascular disease in the elderly.

The patellar-tendon-bearing design of below-knee sockets is clearly the overwhelming prescription choice, accounting for 91 percent of all sockets (Table 2). The great majority of these utilize a soft insert, though a number of prosthetists regularly fit hard-socket PTB limbs: 16 of the respondents make them for 80 percent or more of their patients. The traditional carved wood socket is used in only 6 percent of the below-knee prostheses made today compared to 38 percent reported in the Litt and Nattress survey (*3*).

Cuff suspension is prescribed most frequently (58%) while supracondylar-suprapatellar (SC-SP) suspension is utilized in 16 percent of below-knee limbs. It is interesting to note, however, that one-quarter of the respondents feel that the SC-SP socket is preferable for the bulk of the amputee population rather than special cases and use this method for 50 percent or more of their patients. The supracondylar wedge suspension is used somewhat less extensively (9%), though several of the individuals who contributed to the "other" figure of 3 percent stated that they manufactured a soft insert which included a supracondylar wedge, and two prosthetists regularly use the removable medial-wall variation of this suspension system. To complete the suspension picture, side joints and thigh corsets are utilized for 14 percent of the patients. The SACH foot is prescribed for 81 percent of the below-knee amputees, which reflects greater use than the 73 percent reported by Davies, Friz, and Clippinger (*1*).

TABLE 2. BELOW-KNEE PROSTHESES

SOCKET	%	SUSPENSION	%
PTB with insert	53	Cuff	58
PTB hard socket	37	SC-SP (PTS)	16
PTB air cushion	1	Corset	14
Wood	6	Supracondylar wedge	9
Other	3	Other	3
TOTAL	100	TOTAL	100

FOOT-ANKLE COMPONENT	%
SACH	81
Single axis	15
Other	4
TOTAL	100

The plastic, total-contact, quadrilateral, above-knee socket was introduced approximately 12 years ago. It is clear that this socket design is now predominant and, in fact, used almost exclusively. As indicated in Table 3, 86 percent of all sockets are made of plastic, 80 percent are designed for total contact, and 97 percent are quadrilaterally shaped. The remaining small percentage of sockets that do not display these characteristics is generally found in the prostheses of long-time amputees who prefer not to change.

Apparently, many clinics recognize the advantages of suction techniques since this suspension method, used alone or in combination with a Silesian bandage or pelvic belt, accounts for 61 percent of the suspension prescriptions. Despite this excellent acceptance of the pressure-differential principle, the single, most widely used suspension method is still the pelvic belt (34%). This is not surprising in view of the significant numbers of elderly people in the lower-limb amputee population.

The simple single-axis knee with friction is clearly the most extensively used knee component (42%). Used less frequently are the friction-locking knees (23%) and hydraulic or pneumatic units (22%). However, each of these is used selectively by certain facilities, and the large number of patients fitted with one or the other speaks for the frequency with which specific clinics feel the need to provide patients with additional security and/or improved swing-phase characteristics, despite increases in weight and design complexity. The manual-locking knee, representing concern for stability even at the expense of a normal gait pattern, is used by an average of 8 percent of the limb-wearing population. Polycentric units account for only 2 percent of the knees prescribed, and no prosthetist reported that his facility used them for more than 20 percent of his above-knee patients. It is not clear if this sparse use of polycentric components is the result of dissatisfaction with their function, or lack of information, availability, and experience with these units.

The conventional and most common prosthesis for the hip-disarticulation or hemipelvectomy patient incorporates a socket and suspension of

TABLE 3. ABOVE-KNEE PROSTHESES

KNEE UNITS	%
Single axis, no lock	42
Friction lock	23
Hydraulic or pneumatic	22
Manual lock	8
Polycentric	2
Other	3
TOTAL	100

SUSPENSION	
Pelvic band	34
Suction only	32
Suction and Silesian bandage	20
Suction and pelvic band	9
Other	5
TOTAL	100

PROXIMAL SOCKET DESIGN	%
Quadrilateral	97
Peripheral bearing	3
TOTAL	100

DISTAL SOCKET DESIGN	
Total contact	80
Open end	18
Other	2
TOTAL	100

FOOT-ANKLE COMPONENT	
SACH	74
Single axis	23
Other	3
TOTAL	100

MATERIALS	%
Plastic	86
Wood	14
TOTAL	100

the Canadian design (84%), a SACH foot (70%), and a single-axis knee unit (59%) (Table 4). In contrast with the above-knee prosthesis, a lesser proportion of friction-locking knees is used (23% A/K as compared with 9% of hip amputations) and a greater proportion of manual-locking knees are prescribed (8% A/K as compared with 12% hip). Though fewer hydraulic or pneumatic units are reported (22% A/K vs. 18% hip), it is worthy of note that five prosthetists used these

TABLE 4. HIP-DISARTICULATION AND HEMIPELVECTOMY PROSTHESES

KNEE UNITS	%
Single axis, no lock	59
Hydraulic or pneumatic	18
Manual lock	12
Friction lock	9
Polycentric	1
Other	1
TOTAL	100

SOCKET DESIGN	%
Canadian	84
Tilt table	9
Other	7
TOTAL	100

FOOT-ANKLE COMPONENT	%
SACH	70
Single axis	29
Other	1
TOTAL	100

components for virtually every hip-disarticulation or hemipelvectomy patient.

While the SACH foot is used extensively for all amputees, it is clear that the frequency of prescription decreases as the level of amputation moves proximally (Table 5). Correspondingly, the frequency of use of the single-axis ankle increases. Apparently, the ability of the single-axis ankle to provide faster plantar flexion after heel-strike, and thus enhance stability, leads to its more frequent prescription at higher amputation levels.

TABLE 5. FOOT-ANKLE COMPONENT BY LEVEL OF AMPUTATION

	SACH	Single Axis	Other	Total
	%	%	%	%
Below Knee	81	15	4	100
Above Knee	74	23	3	100
Hip Disarticulation and Hemipelvectomy	70	29	1	100

Modular prostheses are prescribed most frequently for hip-disarticulation and hemipelvectomy patients for whom the combination of light weight and superior cosmesis is of great advantage, with three prosthetists reporting use of modular components for 100 percent of their hip amputees (Table 6). The modular systems for above-knee patients were used least (7%) and only slightly more often for below-knee patients (9%). One may expect that these components will become increasingly popular as clinical experience brings further design improvements and wider availability with lower unit cost.

TABLE 6. MODULAR (ENDOSKELETAL) PROSTHESES

	%
Below Knee	9
Above Knee	7
Hip Disarticulation and Hemipelvectomy	14

Though the principles and procedures of immediate and early postoperative management have been widely disseminated in the past five years, it is only after below-knee surgery that the technique is practiced with any frequency (19%) (Table 7). Only one prosthetist in ten reported that rigid dressings had been applied to 75 percent or more of their below-knee patients, and the procedure is used much less frequently for higher-level amputations. This is disappointing since, in the view of many experts, amputees at all amputation levels are greatly benefited by an aggressive rehabilitation program which includes rigid dressings and early physical and psychological mobilization of the patient.

TABLE 7. THE IMMEDIATE POST-OPERATIVE RIGID DRESSING

	%
After Below-Knee Surgery	19
After Above-Knee Surgery	9
After Hip Disarticulation or Hemipelvectomy	3

SUMMARY AND CONCLUSIONS

Reviewing the data from this survey, one is immediately struck by the fact that for each of the three amputation levels (below-knee, above-knee, and hip-disarticulation), a relatively *small* number of components make up the prescriptions for a relatively *large* number of the patients in each category. As examples, the usual below-knee prosthesis consists of a PTB socket with a soft insert (53%), cuff suspension (58%), and a SACH foot (81%): the usual above-knee prosthesis incorporates a quadrilateral socket (97%) designed for total contact (80%), suction suspension either alone (32%) or in combination with other suspension means (29%), a single-axis knee with adjustable friction (42%), and a SACH foot (74%): for the hip disarticulation, the prosthesis includes a socket of Canadian design (84%), a single-axis knee with adjustable friction (59%), and a SACH foot (70%). The concept, then, of a "standard" prescription for each amputation level may be proposed and the prescription problem stated as: "When and under what circumstances should the standard prescription be modified?" While this formulation is neither original nor startling, it may serve to clarify and expedite the decision-making process required for successful prescription.

Finally, in the absence of a more exhaustive, nationwide study, it is hoped that these data will provide clinic members with a useful indication of current lower-limb prosthetics practice, will encourage the exchange of information concerning various prosthetic components and techniques, and will be of value to the faculties of the prosthetics teaching centers as a guide to curriculum planning and emphasis.

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Orthotics and Prosthetics, Vol. 29, No. 3, pp. 21-23, September 1975

A MODIFICATION TO LOCK MECHANISM OF THE HYDRA NU-MATIC[1] KNEE CONTROL

Donald R. Pemberton[2]

In the fabrication of an above-knee prosthesis, the goal is a prosthesis that provides the amputee with a comfortable, natural gait, and is light in weight.

Because of its durability, lightness, and smoothness of fluid control, the Hydra Nu-Matic knee control has been the choice of this author in many cases.

In some instances, for the severely handicapped —such as bilateral above-knee amputees, above-knee amputees with a very short residual limb, or the elderly—a locking knee control is desirable and often is prescribed. The Hydra Nu-Matic unit is provided with a mechanism that locks the knee by restricting the flow of fluid through the unit. The lock system furnished by the manufacturer consists of a string, a pulley, and a control ring.

To operate the lock, the string must be pulled in one direction to unlock, and the opposite direction to lock. In my experience, this method leaves much to be desired, because the string stretches and becomes out of adjustment and the control ring is difficult to locate and operate through the clothing.

A bilateral above-knee amputee with locked knees is at a great disadvantage because both upper limbs are occupied with a supportive device such as crutches, walker, etc., and he is not able to use his hands to manipulate bilateral locks when he wants to sit down.

To overcome these problems of a bilateral above-knee amputee, we have devised a system that adapts easily to the Hydra Nu-Matic unit

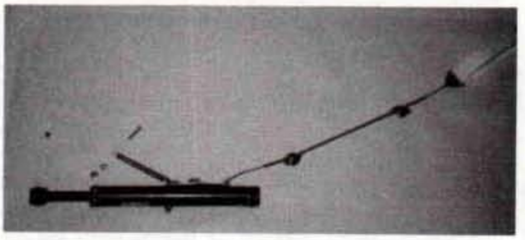

Fig. 1. Modified Hydra Nu-Matic unit. Note the coil spring that is attached distally to the slide bar.

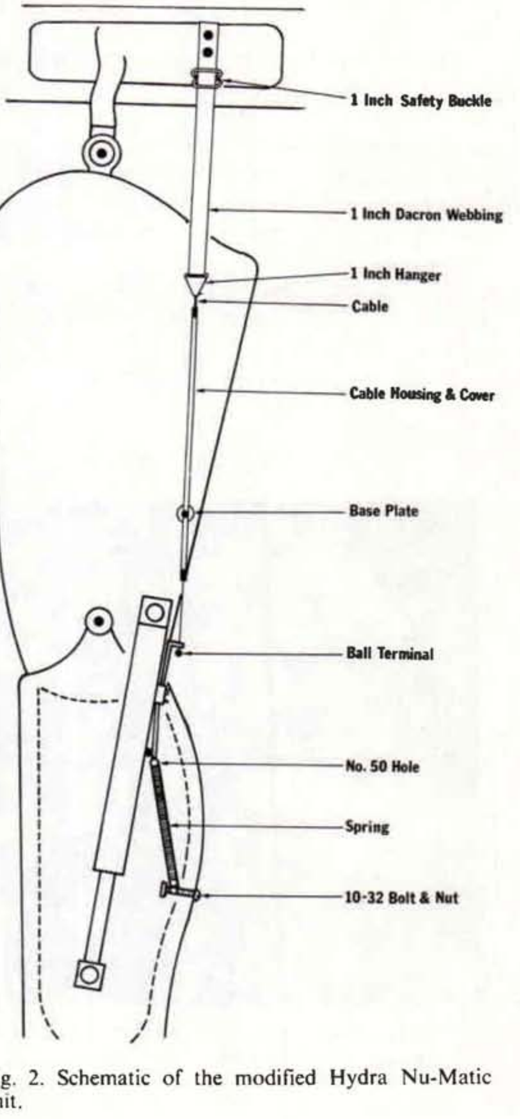

Fig. 2. Schematic of the modified Hydra Nu-Matic unit.

[1]Kingsley Manufacturing Company, Costa Mesa, CA 92627.

[2]Prosthetist, Eastside Orthopedic Company, Kirkland, WA 98033.

with knee lock (Figs. 1 and 2), and allows the knees to be unlocked by use of hip flexion, automatically locked upon extension.

In a normal standing position the knees and hips are fully extended in balance with each other. When sitting position is begun, the hips start to flex first, and then the knees, and finally the body is lowered to the sitting position. In rising, the knees begin to extend and continue to complete extension, the body is almost in complete standing position before the hips are fully extended. At full hip extension the body is then erect and in complete standing balance. Utilizing these basic biomechanical facts, the lock control is harnessed and adjusted intimately to hip flexion and extension.

The case described here involves an 84-year-old male Caucasian with bilateral above-knee amputation with no hip-flexion contractures. Gangrene as a result of vascular insufficiency was the cause for amputation.

At the start of the cycle, the patient is in standing balance with trunk erect, hips extended, and knees locked.

Hip flexion begins and, at 15 deg.[3] of hip flexion, the knees unlock (Fig. 3) lowering the patient to the chair.

To rise, the patient pulls himself from the chair utilizing a walker and his upper limbs. The knees are extending. At full knee extension, the hips are in 15 deg. of flexion, and the automatic knee locks are engaged (Fig. 4).

The patient continues to full hip extension and standing balance (Fig. 5).

Fig. 4. As the hips extend to 15 deg., the knee locks are engaged.

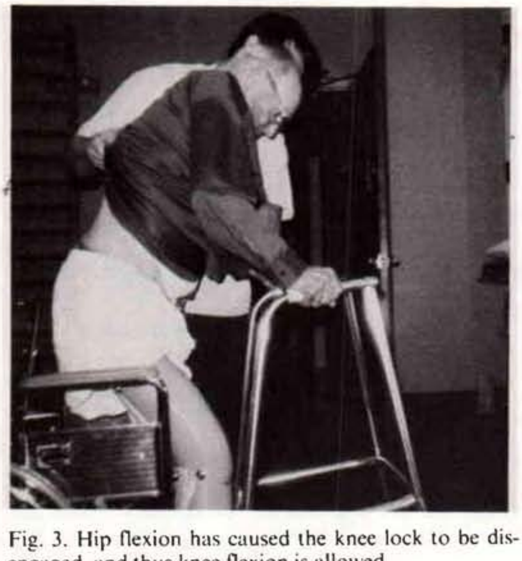

Fig. 3. Hip flexion has caused the knee lock to be disengaged, and thus knee flexion is allowed.

Fig. 5. The hip joints are extended and the knees are locked in full standing.

[3]Degree of hip flexion may vary with the individual.

PROCEDURE TO ADAPT "HIP-FLEXION CABLE-CONTROL SYSTEM" TO HYDRA NU-MATIC UNIT

Drill a No. 50 hole in the extreme distal end of the slide bar for spring attachment. With unit installed, measure posterior and at midline 5-3/4 in. distal to end of slide bar and mark. Remove unit and drill a No. 11 hole through shin. A coil spring measuring 5/16 in. in diameter and 3-1/2 in. in length is attached inside the shin section with a 10/32 machine screw and nut. Install the Hydra Nu-Matic unit in the shin section, and hook the proximal end of spring through the No. 50 hole in lock slide. Complete assembly of prosthesis, thigh to shin section, with knee bolt in place.

For the cable assembly, standard component parts for an upper-limb prosthesis are used. Swage the ball terminal to distal end of cable, clip off the ball, and round the edges. Thread the cable through the posterior hole of the proximal end of lock-slide bar. Attach the cable housing to thigh section of prosthesis with base plates. Thread cable through housing, distal to proximal. Swage 1-in. hanger on the cable[4].

The control adjustment strap is riveted posterior-lateral to pelvic belt with a copper rivet to complete assembly.

ACKNOWLEDGMENTS

I wish to express sincere thanks and appreciation for the time and effort contributed by the Physical Therapy Department at Skagit General Hospital, Mt. Vernon, Washington, and special thanks to Mr. Gary Smith, R.P.T.

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[4]Cable should be of sufficient length to allow travel of hanger during hip flexion, but not to extend over the proximal trim line of socket.

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MATERIALS INVESTIGATION OF FAILED PLASTIC ANKLE-FOOT ORTHOSES

E. P. Lautenschlager[1], S. C. Bayne[1], R. Wildes[1], J. C. Russ[2], and M. J. Yanke[2]

In April 1974 the Department of Biological Materials received for analysis the Ortholen ankle-foot orthosis (AFO) that had failed as shown in Figure 1. This device was one of two worn bilaterally by an active man who weighed 175 lb., and was 5 ft. 11 in. tall. The failure began as a small crack at the base of the right angle, and then propagated slowly for several months until the integrity of the device became questionable.

The purpose of the investigation reported here was to determine the reason for failure and to make suggestions for the improvement of the orthosis.

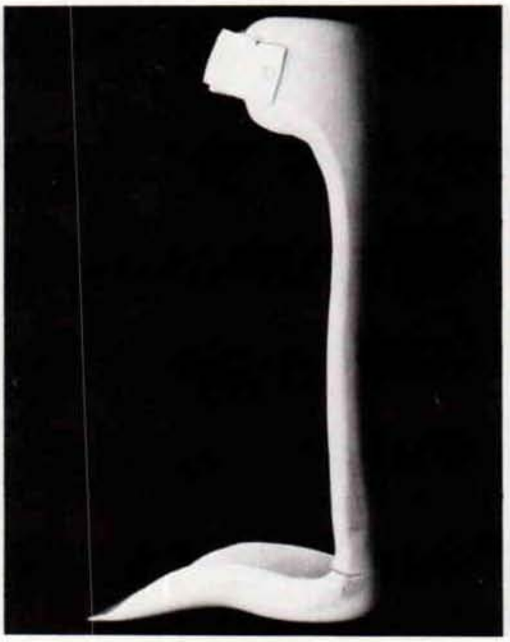

Fig. 1. Failed ankle-foot orthosis.

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[2]Prosthetic-Orthotic Center, Northwestern University, 345 East Superior Street, Chicago, Ill. 60611.

METHODS OF INVESTIGATION

After gross observations as to the influence of loading the device were made (Fig. 2), the orthosis was sectioned in the area of the failure, and scanning-electron micrographs (SEM) were taken of the fractured surface (Figs. 3 and 4).

In addition, four specimens for mechanical testing were cut from the device and tested in three-point bending in an Instron machine at a strain rate of approximately 0.4 per minute.

RESULTS

Figure 2 shows that downward loading near the ball of the foot places the right-angled area in a state of tension and opens up the crack.

Because tension opened the crack and the propagation was reported to be slow, fatigue was suspected as the mode of failure. Fatigue can occur with cyclic loads which are well below the ultimate or maximum load-bearing characteristics of the material. Because fatigue progresses in a stepwise fashion with a crack that is opened up a small additional amount during every cycle, it can be detected from a fracture surface characterized by small steps or striations. This is the case as shown in Figure 4 for the SEM of the fracture surface of the AFO supplied.

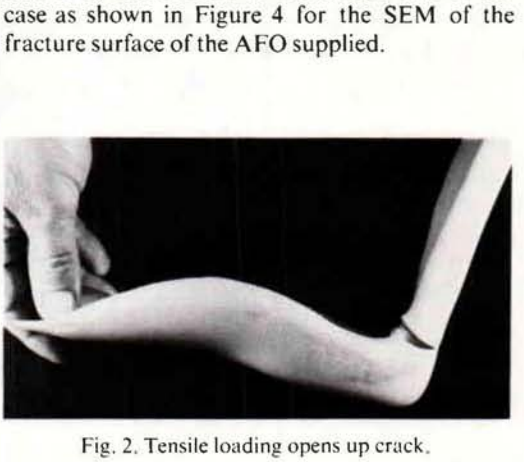

Fig. 2. Tensile loading opens up crack.

Fig. 3. Low magnification SEM of fracture surface. Area in white square presented in Figure 4.

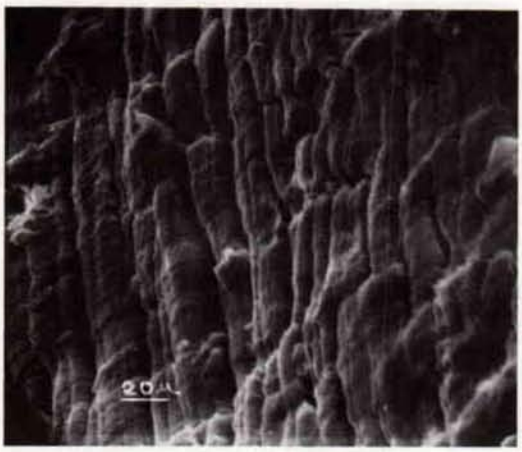

Fig. 4. High magnification SEM of fracture surface. Note the fatigue striations.

The bending strength, elastic modulus, and testing strain rate are listed in Table 1 as the average $\pm$ one standard deviation, as measured for four specimens. The formulae used in the calculations are as follows:

$$\text{Bending strength} = \frac{3}{2}\frac{PL}{BH^2}$$

$$\text{Modulus} = \frac{\triangle P L^3}{4BH^3\triangle y}$$

$$\text{Strain rate} = \frac{6H}{L^2}\cdot CS$$

where:

L = length between base support = 0.61 in.

P = yielding load applied at $L/2$,

B = breadth or width of specimen,

H = height or thickness of specimen,

$\triangle P/\triangle y$ = change of load per change of deflection in straight line elastic region of curves as shown in Figure 5,

CS = crosshead speed = 0.1 in./min.

When compared to other orthotic materials listed in Table 2, the strength of the material is among the weaker normally employed.

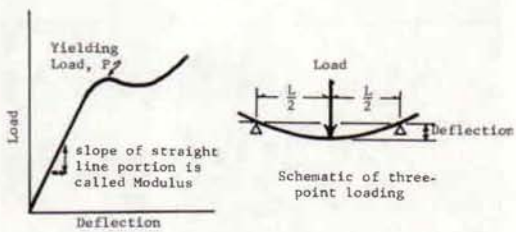

Fig. 5. Typical deformation curve obtained in three-point bending.

TABLE 1. MECHANICAL PROPERTIES OF SPECIMENS SECTIONED FROM AFO SUPPLIED

Specimen	Thickness	Width	Bending Strength (psi)	Elastic Modulus (psi)	Strain Rate min^{-1}
1	.255	.499	7760	31970	0.41
2	.254	.490	7670	28640	.41
3	.257	.488	7390	24320	.41
4	.263	.485	6660	25920	.42
Avg $\pm$s			7370$\pm$500	27710$\pm$3350	

TABLE 2. TYPICAL MECHANICAL PROPERTIES OF ORTHOTIC LAMINATES

Code Number	4110/Polylite Ratio**	Laminates*	Bending Strength (psi)	Modulus (psi)	Strain Rate min^{-1}
1A	50:50	GT/N/D	13,826±544	202,247±11086	.20±.01
1B	50:50	GL/N/D	14,223±628	243,280±33124	.20±.01
1C	70:30	GL/N/D	12,181±581	122,031±25298	.27±.01
1D	70:30	N	6,539±495	154,085±25122	.11±.00
1E	70:30	C	10,845±920	222,231±71441	.13±.01
1F	80:20	C	8,117±381	254,575±14854	.06±.00
2A	50:50	GL/N	11,753±413	249,539±37303	.05±.01
2B	50:50	GT	15,155±331	181,195±13637	.16±.01
2C	70:30	GT/D/N	15,709±1104	151,007±13007	.17±.00
2D	70:30	GL/N	8,911±697	227,540±40460	.13±.01
2E	80:20	C/GL	17,951±1059	330,839±71139	.13±.01
X1	80:20	N	6,835±956	126,199±12240	.13±.01
X2	80:20	GL/N	10,668±1584	164,817±25477	.14±.01
X3	80:20	N/D	8,962±607	186,000±15568	.17±.01
X4	80:20	GL/N	13,815±1680	202,462±53734	.17±.02

*N = nylon; D = dacron; C = cotton; GL = glass, loose weave; GT = glass, tight weave.
**Weight ratio of 4110 resin (American Cyanamid Co.) to Polylite resin (Reichhold Chemicals, Inc.) employed.

SUGGESTIONS FOR IMPROVEMENT

Although one might consider utilizing a stronger material, the basic change should result in a reduction of the high stress concentration produced by the sharp right-angle bend (i.e., the fracture site).

Reduction in stress at the radial edges could be provided by a larger radius, as shown in Figure 6, *left*, but this would change the resistance to both dorsiflexion and plantar flexion. However, failure occurs by fatigue, which is primarily a tensile phenomenon (plantar flexion). Therefore, when an orthosis is required to resist only plantar flexion (high tensile forces), and need not necessarily rigidly counteract compression forces, an orthosis of the type shown in Figure 6, *right*, would suffice. Here a webbing strap provides the resistance to tension during plantar flexion but folds or collapses during compression, thus preventing fatigue failure.

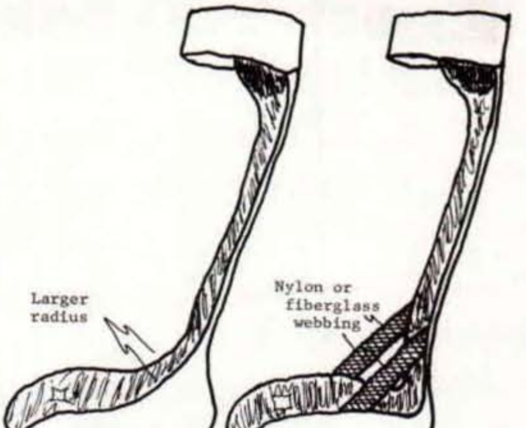

Fig. 6. Two methods of reducing the high tensile stress concentration at the AFO right-angled bend.

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POLYPROPYLENE KNEE ORTHOSIS WITH SUPRAPATELLAR LATEX STRAP

Malcolm Dixon, M.A., R.P.T.[1],
and Robert Palumbo[2]

With the advent of plastics and their utilization in orthotics, continued advances have been made in the development of new orthopedic devices. Plastic orthoses are lighter in weight and usually are more acceptable cosmetically than conventional orthoses. In recent years all-plastic orthoses have been fabricated for the ankle and foot as, for example, the Texas Institute of Research and Rehabilitation polypropylene ankle-foot-orthosis (*1*), and, for the entire lower limb, the VAPC polypropylene knee-ankle-foot orthosis.

A major problem with the fabrication of satisfactory knee orthoses has been the provision of adequate suspension. This has been approached in the past in several ways:

1) Use of a waist belt and strap.

2) Use of a circumferential elastic suspension device such as an elastic knee support or the Lenox Hill derotation orthosis[3] with elastic suspension cuffs.

3) Use of a bar, or bars, extending to the shoe (i.e., Jones knee cage; double-bar knee-ankle-foot orthosis).

The polypropylene knee orthosis described here is a plastic knee support which uses the principle of the strap suspension used for the original patellar-tendon-bearing below-knee prosthesis (*2*). The VAPC plastic knee orthosis with suprapatellar latex strap provides support for the knee without involving another joint, and avoids the use of a constricting circular elastic band.

POLYPROPYLENE KNEE ORTHOSIS

The VAPC polypropylene knee orthosis is fabricated over a positive mold of the patient's knee, which provides for a very accurate and intimate fit. The basic components are:

1) Polypropylene full cuffs: 4 1/2 in.–5 in. in length at the calf and 5 in.–6 in. in length at the thigh. Closure is obtained by use of Velcro straps (Fig. 1).

2) A box joint using ring or slip locks with bars that are 3/4 in. x 1/4 in. The box joint is made of polyethylene and the pivots are made of Delrin. The orthosis provides both mediolateral and anteroposterior stability (Table 1). The placement of a latex strap across the superior pole of the patella in the manner of a PTB suspension strap prevents the orthosis from slipping distally. This was shown in clinical tests on patients.

To prevent buckling of the knee, an additional strap of one-inch wide webbing is placed across the anterior aspect of the proximal tibia as illus-

Fig. 1. Lateral and anterior views of an early model of the VAPC polypropylene knee orthosis with a suprapatellar latex strap.

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[3]Lenox Hill Hospital, New York, NY.

TABLE 1. FUNCTIONAL DESCRIPTION OF POLYPROPYLENE KNEE ORTHOSIS WITH LATEX SUPRAPATELLAR STRAP SUSPENSION

Lower Limb	Horizontal Displacement	Flex.	Ext.	Abd.	Add.
Hip					
Thigh	H				
Knee	Locked	H	H	H	H
	Unlocked	F	F	H	H

trated in Figure 1. Also, to provide additional security from slippage, the closing flaps on the thigh and calf cuffs are lined with a layer of 1/16-in.-thick latex. Six polypropylene knee orthoses have been prescribed and fitted by the VAPC Special Clinic Team in the past year.

Fig. 2. X-ray of knee of the subject described in the text. Note that the patella has been removed.

CASE HISTORY

The patient sustained wounds to his right knee as a result of shell fragments in 1969. He underwent several operations which included removal of the medial and lateral menisci, and eventually a patellectomy (Fig. 2).

The patient walks with a marked limp on his right side. Knee-flexion range is from 10 to 80 deg. There is marked atrophy. Beyond the 70 deg. range of motion the patient has pain. Stress testing of the medial and lateral collateral ligaments and cruciates elicits pain. Instability is not elicited.

At the time of examination the patient was using a hinged elastic knee support that was inadequate. Patient feels less pain and more security when knee is held stiff in extension. It has been suggested to him that knee fusion may be necessary. The "Lenox Hill derotation brace" was prescribed with a slip lock at the knee. It was learned later that the Lenox Hill device cannot be fabricated with a slip lock. The next step considered was a Jones knee cage. The authors, members of the VAPC Clinic Team, suggested an experimental polypropylene knee orthosis with solid polypropylene joints, slip lock and suprapatellar latex strap.

The new orthosis proved to be very satisfactory. It is all plastic except for the latex suprapatellar strap for suspension and a proximal pre-tibial strap to prevent knee flexion and to replace the knee-cap pad. Although the patella had been removed, the fibrous tissue prominence of the reconstructed quadriceps tendon produced a sufficient shelf to provide an area of suspension support for the "suprapatellar" strap. The orthosis is very cosmetic and holds its position well. The patient is extremely pleased with the results.

With the assistance of M. Danisi, C.O., and E. Lamberty, C.O., a significant improvement was made in the suspension system, and it is the one now used routinely. The new suspension device resembles the strap of the PTB prosthesis more closely as can be seen in Figures 3 and 4.

SUMMARY AND CONCLUSION

A well-fitting lightweight polypropylene knee orthosis providing mediolateral and anteroposterior stability can be fabricated without involving the hip or ankle joint. The success of this orthosis depends on its intimate fit, with thigh and calf cuff lined anteriorly with a thin layer of latex and with a suprapatellar latex strap in the correct position.

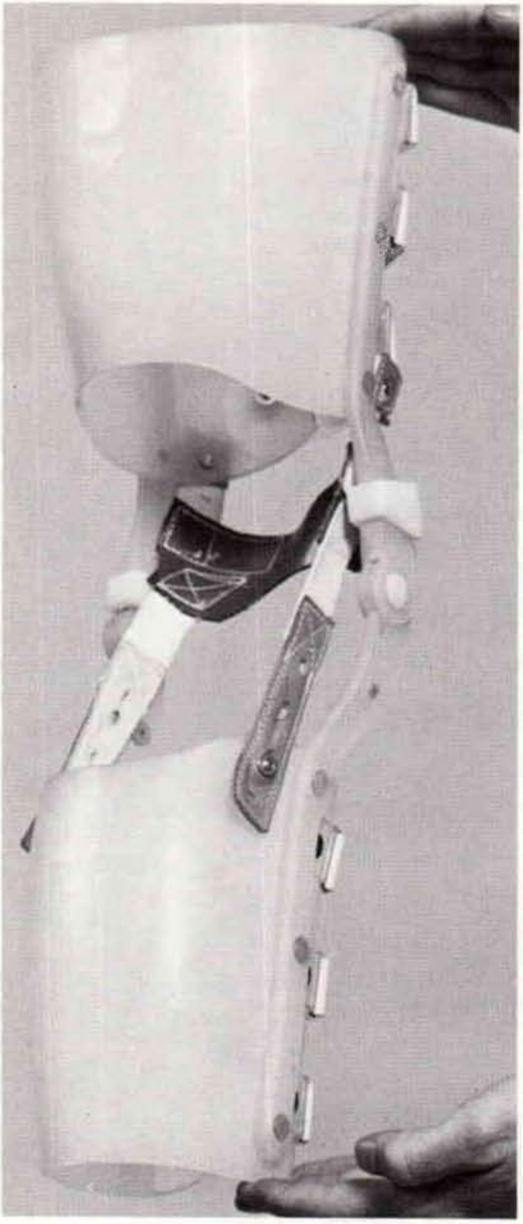

Fig. 3. Posterolateral view of current model of the VAPC polypropylene knee orthosis. Note the suprapatellar suspension strap modeled closely after the suspension strap originally designed for the first PTB prostheses.

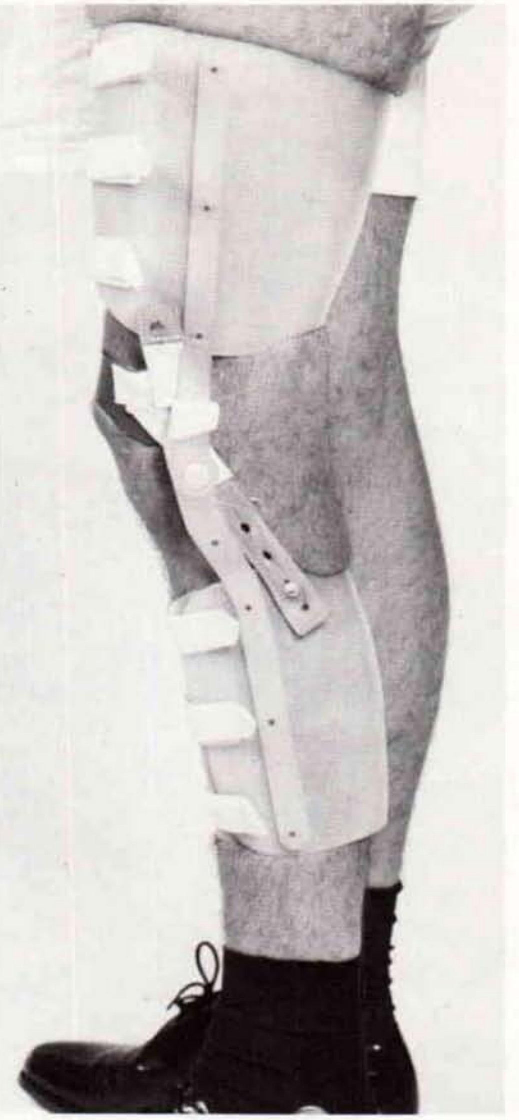

Fig. 4. Lateral view of latest version of the VAPC polypropylene knee orthosis.

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Orthotics and Prosthetics, Vol. 29, No. 3, pp. 33-38, September 1975

A COMPARISON OF SOME GAIT CHARACTERISTICS WITH SIX KNEE JOINTS[1]

C. M. Godfrey, M.D., F.R.C.P.[2], A. T. Jousse, M.D., F.R.C.P.[3], R. Brett, C.P.O.[4], and J. F. Butler, M.Sc.[5]

Patients with above-knee amputations are provided empirically with various types of knee joints. Factors which may be considered are the stability of the knee in stance phase, maintenance problems, and cost of the prosthesis. Little consideration has been given to the effect on gait of different knee joints. Various claims are advanced by manufacturers for different brands of knee joints but little evidence is available from dynamic studies to support such claims.

The present investigation involved seven male above-knee amputees who were fitted serially with six types of knee joints. Gait characteristics of the subjects were analyzed with each device to ascertain any differences that resulted as the joints were used. The fittings were carried out by a well-qualified prosthetist who fabricated a lightweight plastic socket which permitted the substitution of various knees. This, and suitable attention paid to alignment, allowed reproducible study of the gait variables of each knee.

METHODS

The knees used in the study were:

Dyna-Flex Hydraulic Knee	[38]
Hosmer Pneumatic Knee	[39]
Bock 3P23 Knee	[45]
Kolman Safety Knee	[49]
Mauch S-N-S Knee	[59]
Blatchford Stabilized Knee with Pneumatic Swing Control	[61]

as catalogued in the United States Veterans Administration publication entitled *Selection and Application of Knee Mechanisms* (5). The numbers in the brackets indicate the page number in the VA publication.

Each subject was fitted with the knee joint and given an opportunity to practice during morning hours. In the afternoon of the same day he was observed in our laboratory and, after satisfying himself of his best gait, recording of gait variables was made. The next day the same procedure was carried out with a different knee joint. Gait characteristics of the intact leg and prosthesis were determined by direct measurement of knee angle, heel strike to toe-off interval, and average stride length.

A goniometer was mounted concentrically on the knee joint. A 10kΩ Ohmite potentiometer was attached to the arms of the goniometer (Figs. 1 and 2). Suitable resistances were placed in series with the potentiometer to give an output of approximately 0–5 volts within 0–100 degrees of knee flexion. Repeatability and accuracy of the goniometer angle measurement were about 2 percent.

A miniature spring-loaded push-button foot switch was attached to the heel and another to the toe of the shoe to indicate the time of heel strike and toe-off. The output from these devices was fed by lightweight twisted-pair wire to a miniature chassis box mounted on the subject's belt. A heavier gauge cable fed the signal away to a tape recorder. An assistant walked beside the patient to prevent the cable from interfering with freedom of movement.

A Crown 700 Series, MOD IM 7 tape transport allowed the standard tape recording of DC signals. Two channels of the FM section were used to record the knee-angle and foot-switch contact times. Another channel recorded the observers' comments from start to finish of the walk.

The data from the tape recorder were displayed

[1]This study was funded by a Province of Ontario, Ministry of Health, Grant PR206.

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[3]Professor, Department of Rehabilitation Medicine, University of Toronto, Toronto, Ont.

[4]Prosthetist, Kingston, Ontario.

[5]Research Assistant, Sunnybrook Hospital.

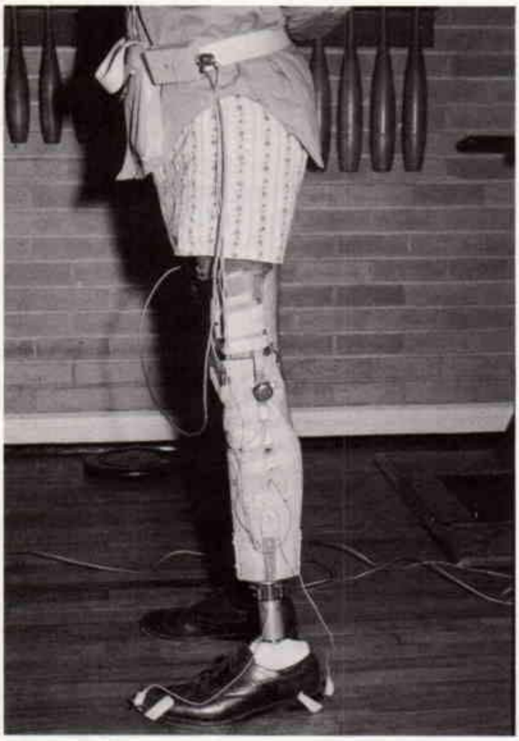

Fig. 1. Instrumented prosthesis for measuring characteristics of gait of above-knee amputees.

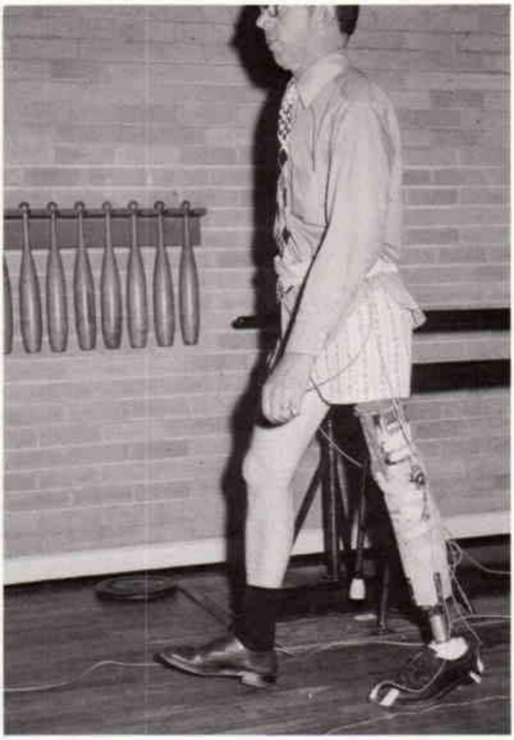

Fig. 2. Above-knee amputee subject walking while using instrumented prosthesis to record gait characteristics.

on a Tectronic Oscilloscope, Model 564, which allowed monitoring and rapid analysis of gait pattern. The oscilloscope has a split-screen storage capability which permits the signal to be viewed at various speeds and amplitudes. Each flexion cycle was superimposed onto previous ones to determine average times during a walk.

The measurements were made on a level walkway 15 meters long. The first and last three meters were used to stabilize gait and permit deceleration. Three walks were performed with each prosthetic knee and the results were examined for repeatability before being retained. In the statistical analysis the arithmetical mean value and standard deviation were calculated.

The variation around the mean values for each subject for the cycles during which measurements were made was very small for all variables. This was considered to mean that during the recordings, the patient walked naturally and attained steady state as regards his gait. None of the subjects used a cane or other support while the recordings were being made.

Mean age of the subjects was 41 years (range 22–56), body height 179 cm (range 165–188), and body weight 71 kg (range 64–77). The mean stump length was 31 cm (range 25–39) corresponding to about 65 percent of the length of the intact femur.

A prosthesis-evaluation questionnaire was given to each subject and to a trained observer to provide additional information on the amputee's walking pattern. The factors investigated included pain, fatigue, cosmesis, gait rhythms, and specific complaints.

As a comparison point for this group of amputees, studies reported by James *et al* (*1*) and Murray *et al* (*4*) were chosen. In those studies, gait characteristics at a normal walking speed were studied in 34 male unilateral above-knee amputees wearing various knee joints and 30 normal male subjects, respectively.

RESULTS AND DISCUSSION

Table 1 gives the mean value, standard deviation, and range for normal walking speed and the corresponding values for walking cycle duration, step frequency, and stride length in the group of amputees. Comparative data reported by James *et al* (AK amputees) and Murray *et al* (normal subjects) are also given.

Table 2 presents the mean values and standard deviations for duration of stance and swing phase, the step-length difference between prosthesis and intact leg, and heel rise. Comparative data, as in Table 1, are also given.

Table 3 and Table 4 summarize patients' and observers' comments, respectively. McAndrew was selected as being representative of patients whose experience was of free-swinging knees. Reid had used the Hosmer pneumatic prosthesis for four years prior to the study.

The characteristics that were measured directly are knee angle, heel strike and toe-off time, stride length, and walking cycle. From these data one can calculate accurately walking speed,

TABLE 1. VALUES FOR GAIT VARIABLES OF SIX PROSTHESES AT NORMAL WALKING SPEED
(Standard deviation and range are given in parentheses.)

	Kolman Bock Blatchford	Mauch S-N-S	Dyna-Flex Hosmer	James (*1*)	Murray (*4*)
Walking Speed (cm/sec)	85 (18) (55–115)	83 (16) (59–104)	92 (15) (56–113)	94 (15) (63–123)	151 (20)
Stride Length (cm)	119 (20) (75–150)	120 (20) (89–147)	125 (20) (86–135)	129 (17) (95–162)	156 (13)
Step Frequency (steps/min)	82(10) (63–94)	84 (7) (74–93)	87 (4) (79–94)	85 (72–105)	113
Walking Cycle (sec)	1.50(0.24) (1.26–2.01)	1.44(0.19) (1.26–1.82)	1.47(0.15) (1.26–1.87)	1.41(0.12) (1.14–1.67)	1.06 (0.09)

TABLE 2. VALUES FOR WALKING CYCLE COMPONENTS AT NORMAL WALKING SPEED
(Standard deviations are given in parentheses.)

	Kolman Bock Blatchford	Mauch S-N-S	Dyna-Flex Hosmer	James (*1*)	Murray (*4*)
Stance Phase (sec) Percent of Cycle	0.087 (0.03) 58	0.92 (0.03) 59	0.93 (0.02) 59	0.80 (0.08) 57	0.65 (0.07) 61
Swing Phase (Sec) Percent of Cycle	0.63 (0.031) 42	0.52 (0.03) 41	0.54 (0.02) 41	0.61 (0.05) 43	0.41 (0.04) 39
Step Length Difference Prosthesis—Intact leg (cm)	1.0 (0.10)	0.0 (0.12)	1.0 (0.09)	5.8 (0.10)	0.0 (0.09)
Heel Rise (cm)	20 (10)	16 (10)	12 (6)	------	------

TABLE 3. SUMMARY—SIX KNEES—PATIENTS' COMMENTS

Knee	McAndrew	Reid
Own	—Thinks it's OK, although he knows it's not very good and that better units are available (free swing—no resistance).	See Hosmer (suction pneumatic)
Bock	—Much better than his own, likes security. Terminal impact, medial whip and vaulting present.	—Felt flexion was less than it actually was. Likes it better than Kolman.
Kolman	—As Otto Bock, but more security.	Good stability—not as smooth as his own (i.e., terminal impact objectionable).
Dyna-Flex	Stable upon heel strike—thought flexion was less than it actually was—Kolman better.	Didn't like it at all—worked opposite to his own (pneumatic).
Mauch S-N-S	—Felt heavy. Likes ease of adjustment. Feels there is almost no flexion even though it may be 50 percent.	Very smooth walk—the only one we tested that he would like to wear.
Hosmer	—Very smooth walk; he said it's a "rolling" type of walk (after proper adjustment).	He is happy with it—except slight abduction makes him "pull in" the prosthesis to get it "under him." (Fitting problem.)
Blatchford	—As Hosmer—more "safety" upon heel strike with this one.	Not as good as his own—likes Mauch S-N-S better—stance not smooth and even.

step frequency, stance- and swing-phase duration, heel rise, and the difference in stride length between prosthesis and intact leg.

The knee-flexion angle was studied throughout each angle to calculate the time interval between the start of flexion and toe-off and the degree of heel rise. The interval between the start of flexion and toe-off was constant to within 0.03 second for each subject and did not vary with the type of prosthesis. However, subjects with longer time intervals, *e.g.*, Reid, exhibited a better quality gait in the opinion of the observer.

Heel rise was assumed to be a function of knee angle only, and was calculated as the distance between heel and floor at maximum knee angle. Hip rotation and vertical hip motion [4 cm. for normals according to Lamoreux (*3*)] were assumed to be constant at a normal walking speed. This was observed to be especially valid for the cadence responsive knees. Table 2 indicates the large variation in heel rise for all subjects wearing the same prosthesis. The pendulum effect was pronounced since the time interval from toe-off to maximum flexion tended towards a constant value of 0.25 second.

The most significant conclusion from the data of Table 1 is that cadence-responsive knees allow significantly greater walking speed with a minimum of effort. Further confirmation is found in the comments of Tables 3 and 4. As expected, the stance- and swing-phase intervals decrease in proportion to walking speed. The patients with the greatest body height walked with a longer step, and had a somewhat shorter walking-cycle

TABLE 4. BASIC SUMMARY—SIX KNEES—OBSERVERS' COMMENTS OF PATIENTS' GAIT

Knee	McAndrew	Reid
Own	Very bad—vaulting, terminal impact excessive, medial whip, excessive heel rise.	See Hosmer
Bock	Some improvement only, over his own.	As Kolman, but somewhat better.
Kolman	Walks better than with his own but still with same gait problems (terminal impact).	Very irregular, jerky walk. Terminal impact.
Dyna-Flex	Kicked his stump to overcome knee's resistance with success, resulting in a bad walk (excessive terminal impact).	A poor walk—pneumatic resistance of his own worked opposite to this one. (Prosthesis poorly built—terminal impact and resistance in swing phase high.)
Mauch S-N-S	Too dramatic change from his own. Even at minimum resistance is too much. Ext. resistance stops stump kicking—good stability.	Very smooth walk—no problems.
Hosmer	Very good—smooth and easy.	This is his own—a very good walker prosthesis, has slight abduction.
Blatchford	Same as Hosmer—it seems the hydraulic knees are too "stiff" for him. The lower resistance of the pneumatics (Hosmer and Blatchford) seems ideal for him.	Walks better with his own—the stance control was jerky (seemed to take too much effort to get knee extended).

duration than those with the lowest body height. They had, furthermore, higher maximal and normal walking speeds.

With few exceptions, the material of James *et al* indicates that the subjects took longer steps with the prosthesis than with the intact leg. They believe that the full extension of the hip on the prosthesis side was prevented by pressure of the prosthetic socket against the ischial tuberosity. This limitation can also contribute to making the stance phase of the prosthesis shorter than that of the intact leg. The data from our Table 2 show that the stride length of the intact leg follows closely that set by the prosthetic limb as first reported by Karpovitch (*2*). This is most likely a result of the close attention paid to alignment and construction of a comfortable socket for this study. None of the subjects experienced any great discomfort or pain with any of the interchangeable knee joints.

The values of gait variables for each subject were usually clustered within a narrow range regardless of the prosthesis. This may indicate the influence of a stable, well-fitted socket with the prosthetic knee joints. It appears that the subject walks with a characteristic gait pattern of his own on any stable prosthesis.

All units tested were stable on heel strike. As expected, the Kolman and Bock units had to be readjusted constantly as a result of wear on the braking surfaces.

The knees with mechanical friction exhibit excessive heel rise and terminal impact of the prosthesis as it goes into full extension.

MAIN CONCLUSION

In spite of the claims of manufactuers, it appears that on flat level surfaces it is the patient who walks the prosthesis and controls gait variables. The more complicated designs seem to have no great advantages over the simpler ones for walking on flat level surfaces.

Stability at the beginning of stance (heel strike) and mid-stance is common to all these knee joints when properly fitted and aligned. Complex units do not necessarily give a better gait.

This study was confined to straight walking on a flat level surface and the number of subjects was limited because of a lack of funds. It is suggested that the study be expanded to include more subjects on varied terrains.

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Orthotics and Prosthetics, Vol. 29, No. 3, pp. 39-41, September 1975

TECHNICAL NOTES

A MODIFIED ORTHOSIS FOR PREVENTION OF RECURRENT DISLOCATION OF HIP ARTHROPLASTIES

A dislocated hip orthosis can be an extremely difficult management problem. Traditionally, the hip has been reduced and the patient placed at bed rest in Wilke Boots for four to six weeks. This complication usually occurs in the elderly patient, and the period of forced bed rest is undesirable since it may, in turn, cause additional complications such as pneumonia, decubitus ulcers, and pulmonary embolism. Dislocation can also be recurrent, particularly in the total-hip-replacement prosthesis.

The advantage of holding the hip and limb in a stable position, yet allowing the patient to be ambulatory, is obvious. With this in mind, we decided to modify an orthosis that consists of a lumbosacral support (chair back or Knight spinal) with a hip joint and an extension around the thigh. The modified orthosis (Fig. 1) holds

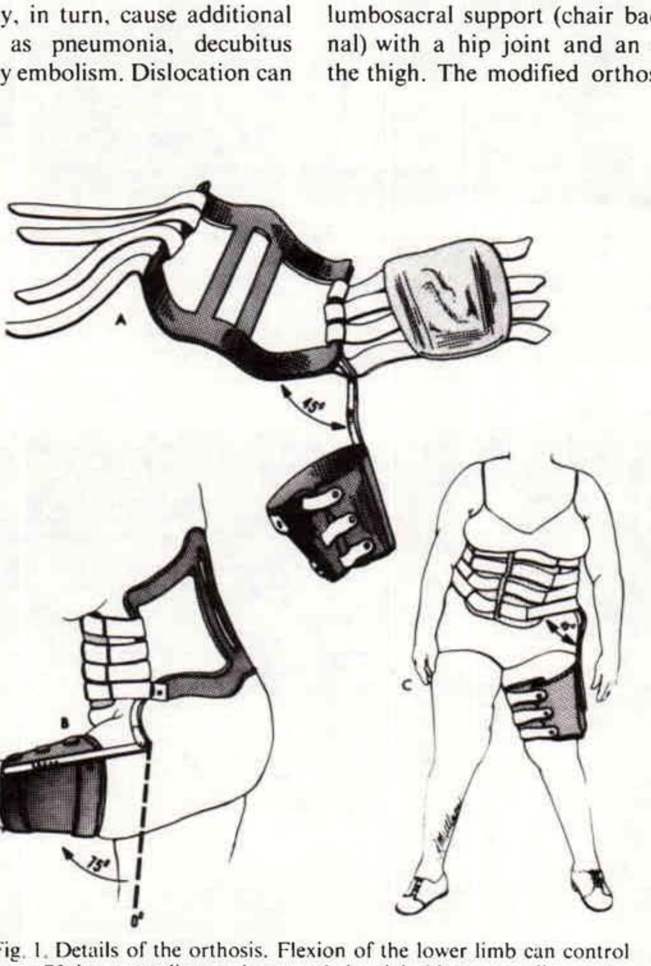

Fig. 1. Details of the orthosis. Flexion of the lower limb can control up to 75 deg. according to the case. It is advisable not to allow more than 60 deg. of flexion following a Charnley type of total hip arthroplasty.

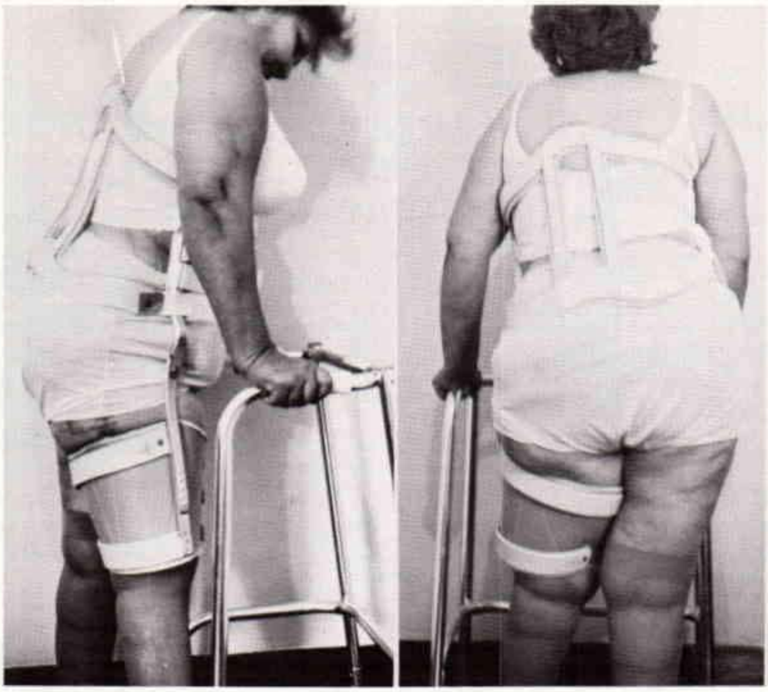

Fig. 2. A 57-year-old white female, who underwent a total-hip-replacement-type McKee-Farrar on April 15, 1970, following osteoarthritis of the hip. A revision of the hip was done due to loosening of the components and a low friction arthroplasty was performed on July 6, 1974. Due to tissue laxity, the prosthesis dislocated on two occasions. The orthosis was ordered and fitted, and the patient was able to walk immediately. Patient was discharged one week later.

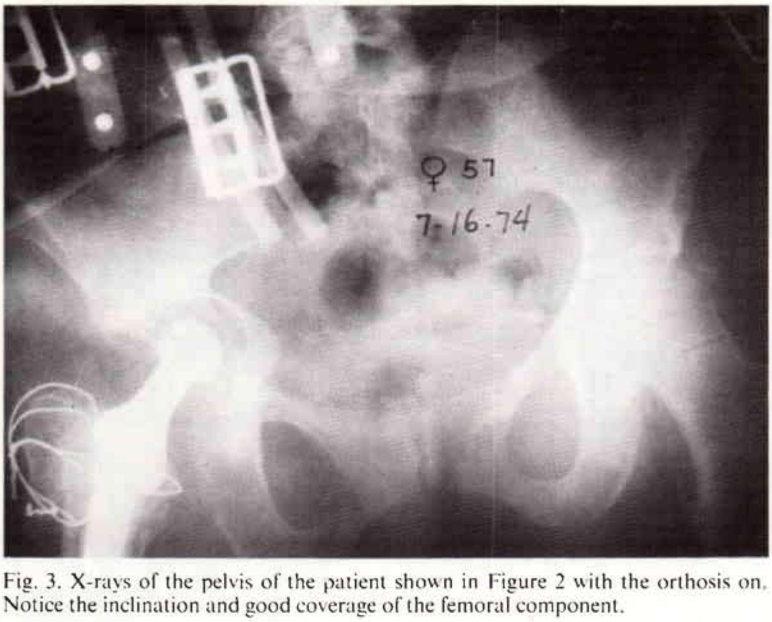

Fig. 3. X-rays of the pelvis of the patient shown in Figure 2 with the orthosis on. Notice the inclination and good coverage of the femoral component.

the limb in 45 deg. of abduction, does not allow hip extension, and allows no more than 50 to 70 deg. of flexion. For accurate control of hip flexion one may use a heavy-duty joint of the Klenzak type used for ankle dorsiflexion. If this joint is used it must be reversed with the upright attaching to the thigh section of the brace. If one wishes to control rotation, a twister cable can be added from the thigh to the shoe.

The abduction of the thigh forces the patient to tilt his pelvis down on the operated side and gives better coverage to the prosthesis. With the limitation of flexion and extension, and rotation if necessary, dislocation is prevented.

We have treated five patients (Figs. 2 and 3) at the University of Miami School of Medicine with this orthosis. There have been no dislocations after institution of the orthosis. All patients were allowed to ambulate as soon as they were fitted. Hospital stay was, therefore, markedly reduced with this method. The brace was utilized from four to six weeks which allowed soft-tissue healing in the stable position. The brace can be used for endoprostheses or total-hip replacements.

SUMMARY

A modification of the lumbosacral support with a thigh extension to prevent redislocation of hip arthroplasties in the early postoperative period has been presented. The specifications are, namely, a lumbosacral support with a hip joint and a thigh extension that does not allow more than 75 deg. of flexion, and a twister cable if rotation has to be controlled. Five patients have been treated with this method at Jackson Memorial Hospital of the University of Miami School of Medicine with satisfactory results.

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MODIFICATION OF OUTRIGGER ON MILWAUKEE ORTHOSIS

Sometimes, when fitting the Milwaukee orthosis on female patients, the level of the thoracic pad is such that the front outrigger is at the level of the fullness of the breast, and, with proper clearance, passes directly over it (Fig. 1).

This placement imposes an unnecessary cosmetic and thus psychological burden on the typical adolescent female patient, because the softness and contour of one breast is obscured by the protrusion of the overlying hard metal.

This problem can be eliminated easily by substituting a simple L-bracket for the horizontal outrigger (Fig. 2). The L-bracket is fastened to the front upright at a level just below the bottom of the breast. The horizontal leg is contoured to the front of the chest wall just below the breast (allowing clearance for chest expansion), and is extended laterally beyond the lateral fullness of the breast. From this point, the vertical leg begins, and rises to the level at which the strap for the thoracic pad must be fastened.

The two breasts now present equal, normal fullness and softness—an aid in fulfilling the cosmetic and psychological requirements of the wearer.

This method of substituting an L-bracket for the usual front outrigger to anchor the strap for the thoracic pad has been used on 12 patients, and has proved to be successful biomechanically and psychologically in each case.

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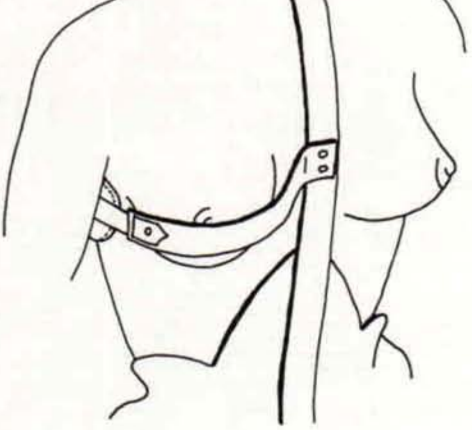

Fig. 1. Conventional, horizontal outrigger on the Milwaukee-orthosis.

Fig. 2. New L-bracket for the Milwaukee orthosis.

NEW PUBLICATIONS

HUMAN LOCOMOTOR ENGINEERING. A review of developments in the field including advances in prosthetics and the design of aids and controls. The Institution of Mechanical Engineers, 1 Birdcage Walk, London S.W. 1, England. 224 pp., 170 illus., $10.00 net.

This nicely presented volume contains 29 papers that were given at a conference sponsored by The Institution of Mechanical Engineers and held at the University of Sussex, September 7–10, 1971.

Actually the subtitle is much more descriptive of the contents than the principal title since more than a third of the articles deals almost exclusively with problems of the upper limbs, and four or five articles deal with subjects common to all human limbs.

Although this report is being made available more than three years after the conference, the majority of the material is timely and all of it is well presented. Everyone involved in research and development in prosthetics and orthotics should add a copy of "Human Locomotor Engineering" to their reference collection.

The contents are:

EXPLANATORY NOTE

HOW ELECTROMYOGRAPHIC PATTERNS AND LIMB MOVEMENTS ARE RELATED TO TO THE SPEED OF WALKING—*D. W. Grieve and P. R. Cavanagh*

COMPARISON OF EMG SIGNALS FROM LEG MUSCLES WITH THE CORRESPONDING FORCE ACTIONS CALCULATED FROM WALKPATH MEASUREMENTS—*John P. Paul*

THE STIMULATION OF INCAPACITATED LIMB MUSCLES BY MAGNETO-INDUCTIVE ENERGY—*D. P. Photiades and S. C. Ayivorh*

SINGLE MOTOR UNITS AS SOURCES OF CONTROL SIGNALS—*Z. Török and P. H. Hammond*

FUNCTIONAL ELECTRICAL STIMULATION FOR THE TREATMENT OF THE DISABLED—*S. A. G. Chandler and E. M. Sedgwick*

IMPROVEMENT OF LOCOMOTION IN HEMIPLEGIC PATIENTS WITH MULTICHANNEL ELECTRICAL STIMULATION—*A. Kralj, A. Trnkoczy, and R. Ačimović*

COMPARISON OF SEGMENTAL ANALYSIS AND EXPERIMENTAL OBSERVATIONS IN ATHLETIC ACTIVITIES—*A. J. Smith, D. Dowson, G. T. Adamson, and V. Wright*

ENGINEERING AND PHYSIOLOGICAL CONSIDERATIONS IN THE DESIGN OF A TOTAL KNEE PROSTHESIS—*B. B. Seedhom, E. B. Longton, D. Dowson, and V. Wright*

THE DEVELOPMENT OF A TOTAL SHOULDER ENDO-PROSTHESIS—*B. Reeves, B. Jobbins, D. Dowson, and V. Wright*

THE FRICTION OF INTERNAL ARTIFICIAL JOINTS—*Peter S. Walker*

AN AUTOMATICALLY STABILIZED POWERED WALKING DEVICE—*J. I. Hall and D. C. Witt*

AN EXPERIMENTAL UNPOWERED WALKING AID—*G. K. Nichols and D. C. Witt*

STANCE PHASE CONTROL OF AN ABOVE-KNEE PROSTHESIS—*Philip J. Lowe*

AN OPTIMAL PROGRAMMING STUDY OF HUMAN GAIT—*C. K. Chow and D. H. Jacobson*

THE DEVELOPMENT OF A COMPREHENSIVE EXTERNAL PROSTHETIC KNEE CONTROL—*D. R. W. May*

THE OBJECTIVE ASSESSMENT OF AMPUTEE GAIT—*J. Hughes and M. B. McGechan*

HYDRAULIC POWER FROM THE HEEL—*R. D. McLeish and J. F. D. Marsh*

PHOCOMELIC HANDS—ARE THEY SUITABLE FOR CONTROLLING ARTIFICIAL ARMS?—*N. D. Ring and J. M. Rudd*

RESPONSES OF A PNEUMATICALLY POWERED ELBOW JOINT—*C. R. Burrows, D. J. Martin, and N. D. Ring*

ADAPTIVE CONTROL OF A MULTI-DEGREE-OF-FREEDOM HAND PROSTHESIS—*J. M. Nightingale and R. W. Todd*

PLANTAR POWER FOR ARM PROSTHESES USING BODY WEIGHT TRANSFER—*J. P. Marsden and S. R. Montgomery*

A MINIATURE ELECTRICALLY OPERATED PROPORTIONAL VALVE—*J. C. Cool and P. V. Pištecký*

A PORTABLE HYDRAULIC POWER SUPPLY FOR PROSTHESES—*B. L. Davies*

A COMPUTER-BASED SYSTEM FOR THE ANALYSIS OF SOME ASPECTS OF HUMAN LOCOMOTION—*T. Kasvand, M. Milner and L. F. Rapley*

STABILIZATION OF PNEUMATIC PROSTHETIC SYSTEMS—*M. Lord and A. Chitty*

A GRAVITY ASSISTED WALKING AID—*P. A. Isherwood*

SELF-ADAPTIVE AID IN ARTIFICIAL ARM CONTROL—*Amos Freedy, Frederick Hull, Luigi F. Lucaccini and John Lyman*

HUMAN LOCOMOTION—*Edward Peizer and Donald W. Wright*

KNEE-JOINT KINEMATICS AND CONTROL-SIGNAL ECONOMY—*T. D. M. Roberts*

GLOSSARY

INDEX

THE PHYSIOLOGY OF THE JOINTS. Vol. 3, The Trunk and the Vertebral Column, 2nd Ed., I. A. Kapandji, Churchill Livingston, Edinburgh and London, 251 pp., 397 illus.; $14.00 (For U. S. orders, contact Longman Inc., 72 Fifth Avenue, New York, New York 10011).

This beautifully illustrated book, in English, is the third of a series published originally in France.

The preface to the French edition is reprinted here in full:

> To understand the diseases of the musculo-skeletal system a thorough knowledge of its physiology is essential. However, its mechanical aspects, intimately concerned with the anatomy of the skeleton, have received only scant attention. The classic work of Duchenne de Boulogne concentrates only on the functions of individual muscles without specific reference to the related joints. As a result, this part of physiology, largely neglected during medical training, was little understood by surgeons, even those practising orthopaedics.
>
> This gap has been filled now by the dedicated work of Dr. Kapandji, the artist and teacher, who has not only imagination and a flair for mechanics but also the ability to communicate his ideas in a precise and simple fashion.
>
> His performance, already remarkable in the two previous volumes, is even more striking in the present volume which I have the honour of prefacing, since the complex movements of the vertebral column are far more difficult to understand and explain.
>
> I feel that he has been completely successful in attaining his goal and I envy the young surgeons who have this work at their disposal. I am confident that this volume, which goes a long way towards explaining the basic mechanics of the vertebral column, will help greatly in improving the treatment of lesions of the vertebral column.
>
> Professor R. Merle d'Aubigné.

INTERNATIONAL CONFERENCE ON PROSTHETICS AND ORTHOTICS (Cairo and Alexandria, Egypt, May 1-11, 1972). Report of International Society for Prosthetics and Orthotics. 552 pp.; $5.00 to members, $15.00 to non-members. Send orders to ISPO, P.O. Box 42, DK 2900, Hellerup, Denmark.

This is an extremely comprehensive volume, firstly by virtue of the sheer volume of practical information it contains, and secondly in respect of the regard given to all aspects of prosthetics/orthotics activity.

The report is arranged in two sections. The first section, styled *Observations and Recommendations*, consists of the reports of the Working Groups; while the second section, the *Appendixes*, contains the detailed individual reports of the contributors. Each section deals sequentially with the three topics—Research, Evaluation, and Education.

The Research section presents a brief summary of present work in the principal subject categories while concentrating upon detailing the requirements for improving the organisation for research and, in particular, upon the projected role of ISPO in this sphere.

The individual research reports contain practical details representative of almost every aspect of current research and development effort. This section will require patient study if one intends to extract the information relevant to a particular subject.

The Evaluation section presents for the first time a comprehensive treatment of this most important topic. Once again the authors have placed the emphasis on the detail of the requirements for this activity and the vital role of ISPO in enabling international cooperation. It is refreshing to find specific statements regarding the protocol for evaluation projects and practical advice regarding administrative requirements. The individual evaluation reports contain several excellent examples which illustrate the principles embodied in the recommendations.

The Education section summarises the current world scene and examines the role played by a number of international authorities over the previous ten years.

Analysing the basis of present educational schemes, it stresses the very large degree of agreement on principle with respect to course content and duration. The apparent disparity in practise is largely due to the differing objectives of the educational authorities regarding the role of the students in relation to the prevailing local situation.

The proposals for closer international cooperation included in the recommendations are already being realised with the support of an increasing number of institutions and nations.

The individual education reports provide detailed information on the status and organisation of educational programmes ranging from the well-established North American degree courses through the short-term course provided through the auspices of the U.N. in the developing nations to the traditional European craft-based approach.

The reports should be of considerable value to those persons currently contemplating the creation of educational programmes by providing essentially a catalogue of current experience on which to base preliminary planning.

In general it must be stated that this is a most positive document. The definitive statements, endorsed by the representatives of 24 nations, are a tribute to the spirit of international cooperation which enabled this conference to take place and which is embodied in the recommendations for an even more active role for our International Society.

It is most appropriate that ISPO should have selected this truly international volume to be issued as their "Year Book" in conjunction with the 1st World Congress in Montreux.

LOW BACK AND NECK PAIN—Causes and Conservative Treatment, by Paul C. Williams. Charles C Thomas, Springfield, Ill. 82 pp., 32 illus., $5.95.

The preface follows.

> The primary purpose of this publication is to assist the patient in understanding the cause of low back and/or neck pain and to give him reasons for specific methods of conservative treatment. It is also for the physician who, with the many demands on his time, cannot go into detailed explanation of the anatomy and principles of the problem, but can prescribe the publication to the patient for this information.
>
> It is my hope this material will be made available to all those who have been taught the erroneous idea by Western Civilization that the "strut attitude" is the ideal human posture.
>
> The evolution of the conservative treatment as well as the exercises for the most part are products of my own research. A more extensive scientific explanation is available in a book I wrote for the medical profession which was published in 1965 by Blakiston Division, McGraw-Hill Book Company entitled *The Lumbosacral Spine: Emphasizing Conservative Management*. It includes references to the many who have greatly contributed to this problem. It also includes some of the figures which are being used in this publication. This is also true of a chapter I wrote on low back pain for Harrison's *Principles of Internal Medicine*, original and subsequent editions published by McGraw-Hill Book Company.
>
> The Low Back and Neck Postural Exercise Instruction Sheets in the back of this book can be reproduced by physicians for use in their practice without further permission from the author or the publisher.
>
> Paul C. Williams, M.D.

DEVICES AND SYSTEMS FOR THE DISABLED, Temple University Health Sciences Center, Krusen Center for Research and Engineering at Moss Rehabilitation Hospital, 12th Street and Tabor Road, Philadelphia, Pa. 19141, USA. Eugene Kwatny and Ronnie Zuckerman, editors; 207 pp., $12.

This publication is the pre-conference digest of papers provided to participants at the April 29–30, 1975, conference of the same name, held in Philadelphia, Pennsylvania.

The opening paper by Margaret Pfrommer of Chicago very accurately sets the tone for the whole conference. Ms. Pfrommer, a severely disabled quadriplegic, described her own experience with developmental systems designed to assist her mobility, communications, and comfort. Although the individual systems accomplish the purposes for which they were developed, when used together they provide greater independence and improvement in morale than would be expected—a phenomenon called synergism. Ms. Pfrommer stresses the importance of immediate feedback from disabled persons themselves to the developers of devices, in order that the developers become more sensitive to the varying human needs of the handicapped.

Many of the remaining 36 conference papers carry on this theme of close communication between developers and disabled persons. The conference was divided into eight wide-ranging sessions: Sensory Substitution/Biofeedback, Aids for the Deaf and Hearing Impaired, Verbal and Nonverbal Communication, Prosthetics/Orthotics, Aids for the Blind, Investigatory Studies, Mobility Aids, and General Systems and Vocational Aids.

Being a collection of individually contributed papers, the quality is somewhat variable and papers do not always fit comfortably under the session headings. For example, there are no papers on limb prosthetics in the Prosthetics/Orthotics session. In fact, the only paper related to artificial limbs is under Sensory Substitution/Biofeedback. Nevertheless, the papers are generally well written and illustrated with diagrams and very clearly reproduced photographs. The depth of the papers range from basic research studies, such as the neural mechanisms of sensory substitution, to simple descriptions of aids. The range of disabilities discussed is equally broad—from mental retardation through blindness, deafness, spinal-cord injury and scoliosis, to decubital ulcers. Many well-known authors contributed papers, such as John Basmajian, Paul Bach-y-Rita, Dudley Childress, and Richard Herman from the United States; Michael Albisser from Canada; and John Armstrong and A. D. Heyes from England.

The publication has a few mechanical shortcomings in that there is no index or table of contents. A copy of the conference program is provided, however, and the papers from each session are separated by colored inserts. The printing is on one side only, making a bulky volume. These minor details should not detract, however, from the value of these proceedings in providing current awareness of a broad range of research, development, and clinical applications for the handicapped. Taken as a whole, the book catches some of Margaret Pfrommer's synergism—"a gentle technology" is enthusiastically improving the lives of the disabled. As such, the book is of great value to all those involved in work for the handicapped.—*Peter J. Nelson, Staff Engineer, CPRD*

LITERATURE RETRIEVAL CATALOGUE NOW AVAILABLE

Previously produced in Winnipeg, Canada, the *Prosthetics and Orthotics Research Reference Catalogue* is available now in Volume Two. This new edition, containing over 560 references compiled during 1973 and 1974, is published under the sponsorship of the Committee on Prosthetics Research and Development (CPRD), National Academy of Sciences—National Research Council. Use of CPRD's resources enables this computerized retrieval system to be offered to subscribers at a lower cost, thereby ensuring a wider distribution of this much-needed service.

Endorsed by ISPO, the *Catalogue* provides ready access to literature in the specialized fields of prosthetics/orthotics/assistive devices for the physically handicapped. A particular feature is the expanded Subject Listing, permitting retrieval of references by very specific and detailed subject categories.

Volume Two is priced at $27.00 (U.S.). Orders, with payment enclosed, should be addressed to: Printing and Publishing Office, National Academy of Sciences, 2101 Constitution Avenue, N.W., Washington, D.C. 20418 USA—*Peter J. Nelson, Staff Engineer, CPRD.*

OF INTEREST

Symposium on the Care and Rehabilitation of the Spinal-Cord-Injured Patient.

The Research Center for Prosthetics and the Spinal Cord Injury Service of the Veterans Administration will conduct a symposium on the care and rehabilitation of the SCI patient, October 6–10, 1975. The VA Hospital in Long Beach, California, is serving as host station.

The agenda will include presentations on the following topics: interdisciplinary contributions in the care of the Spinal Cord Injured Patient, facilitating the transition to the home, vocational and educational opportunities and obstacles, current practices in SCI orthotics, bioengineering contributions to SCI patient care and rehabilitation, with exhibits and demonstrations.

The program is open to all physicians, therapists, nurses, orthotists, prosthetics representatives, bioengineers, rehabilitation counselors, educators, psychologists, social workers, and other professionals with special interest in the individual with a spinal-cord injury, both in the Veterans Administration and in the medical and rehabilitation community generally. The VA does not offer traineeships or other forms of financial support to non-VA participants. Registration will be limited and advance registration is necessary as there is no fee or tuition charge for this program.

Enquiries should be addressed to: Mr. Earl A. Lewis, Assistant Director, Research Center for Prosthetics, Veterans Administration, 252 Seventh Avenue, New York, New York 10001.

INFORMATION FOR AUTHORS

ORTHOTICS AND PROSTHETICS
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All submitted manuscripts should include:

1. THE ORIGINAL MANUSCRIPT AND TWO COPIES. If possible, the duplicate manuscripts should be complete with illustrations to facilitate review and approval.
2. BIBLIOGRAPHY. This should be arranged alphabetically and cover only references made in the body of the text.
3. LEGENDS. List all illustration legends in order, and number to agree with illustrations.
4. ILLUSTRATIONS. Provide any or all of the following:
   a. Black and white glossy prints
   b. Original drawings or charts

   Do *not* submit:
   a. Slides (colored or black & white)
   b. Photocopies

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1. Manuscripts must be TYPEWRITTEN, DOUBLE-SPACED and have WIDE MARGINS.
2. Indicate FOOTNOTES by means of standard symbols (*).
3. Indicate BIBLIOGRAPHICAL REFERENCES by means of Arabic numerals in parentheses (6).
4. Write out numbers less than ten.
5. Do not number subheadings.
6. Use the word "Figure" abbreviated to indicate references to illustrations in the text (. . . as shown in Fig. 14)

PREPARATION OF ILLUSTRATIONS

1. Number all illustrations.
2. On the back indicate the top of each photo or chart.
3. Write the author's name on the back of each illustration.
4. Do not mount prints except with rubber cement.
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—Rejected manuscripts will be returned within 60 days.
—Publication of articles does not constitute endorsement of opinions and techniques.
—All materials published are copyrighted by the American Orthotic and Prosthetic Association.
—Permission to reprint is usually granted provided that appropriate credits are given.
—Authors will be supplied with 25 reprints.

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

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

AREA

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

VOLUME

Definition

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

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

MASS

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

FORCE

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

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

STRESS (OR PRESSURE)

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

TORQUE (OR MOMENT)

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

ENERGY (OR WORK)

Definition

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

1 cal (gm) = 4.1840 joules

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

TEMPERATURE CONVERSION TABLE

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

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

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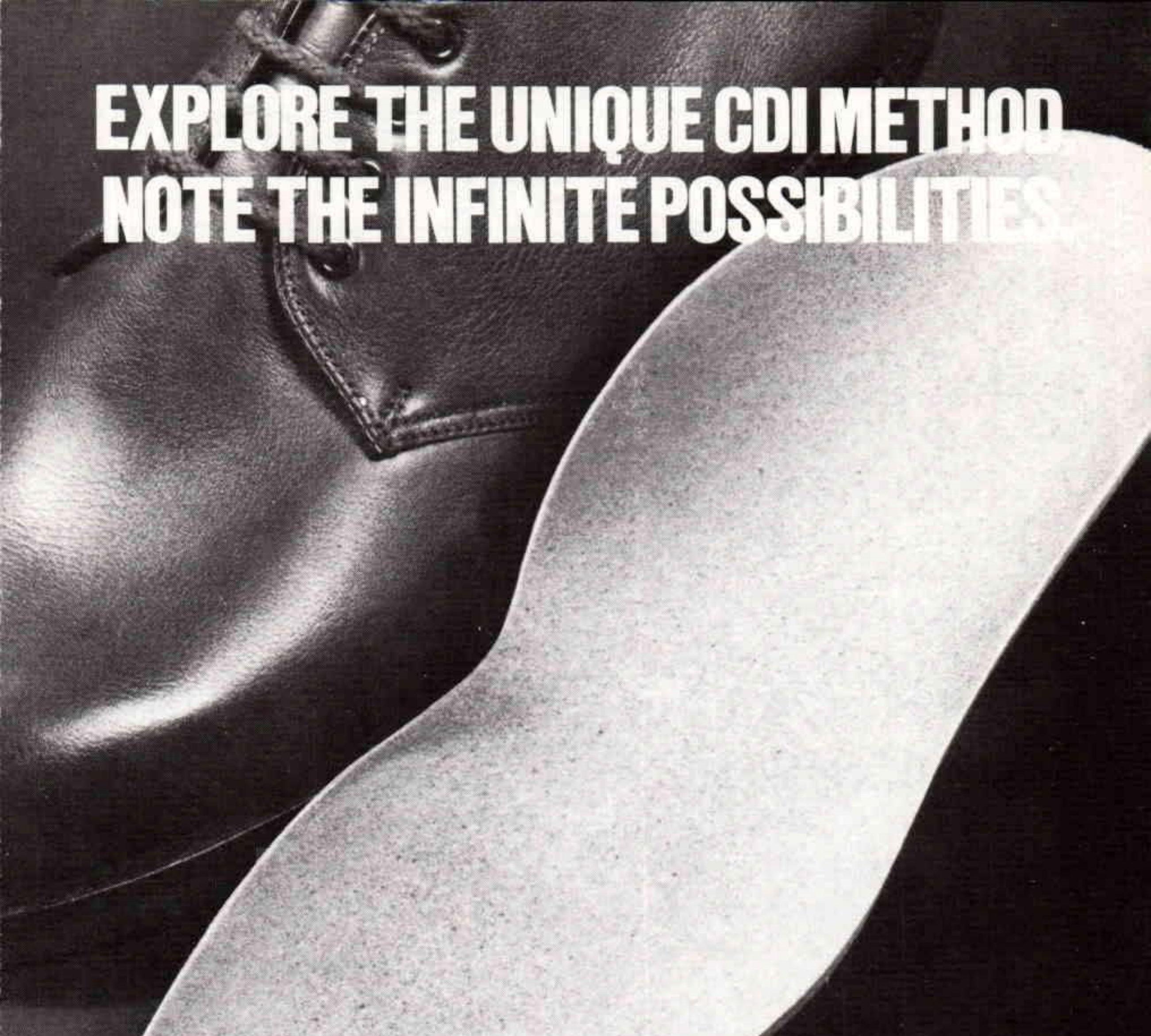

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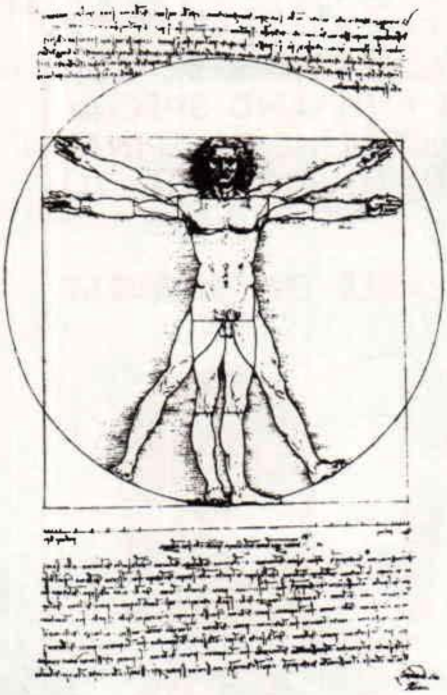

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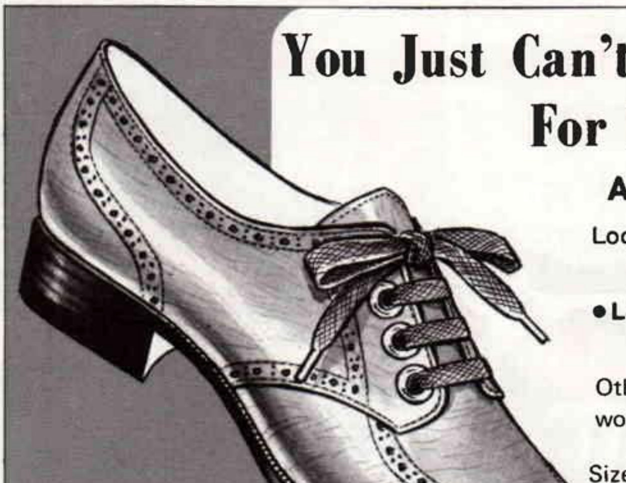

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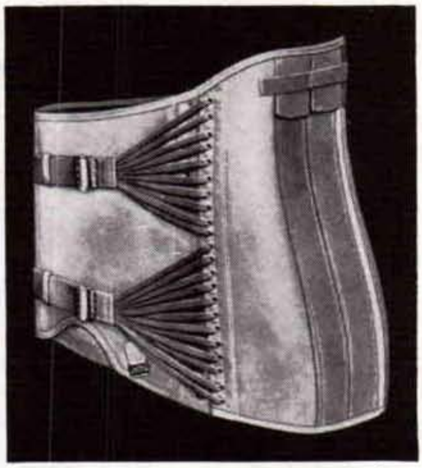

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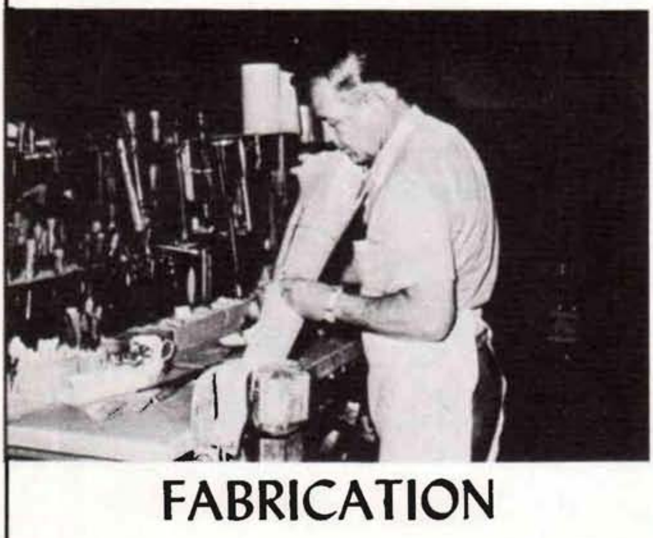

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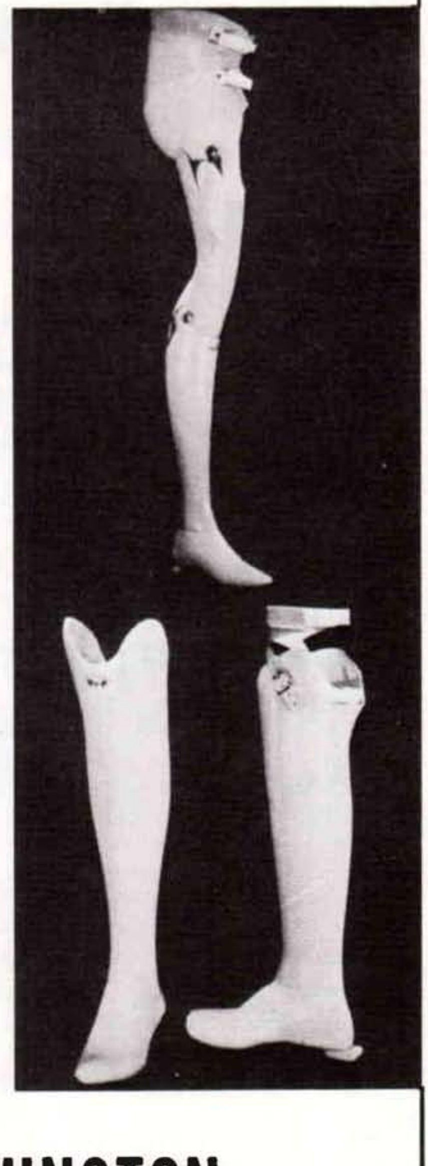

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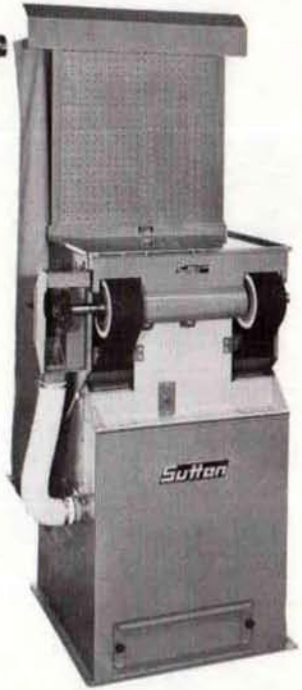

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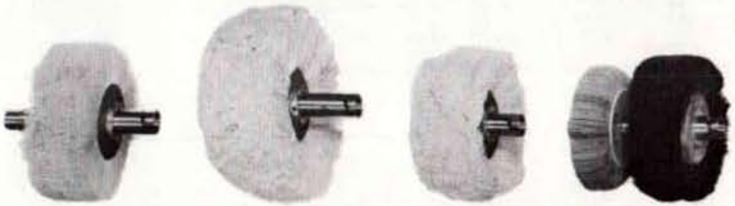

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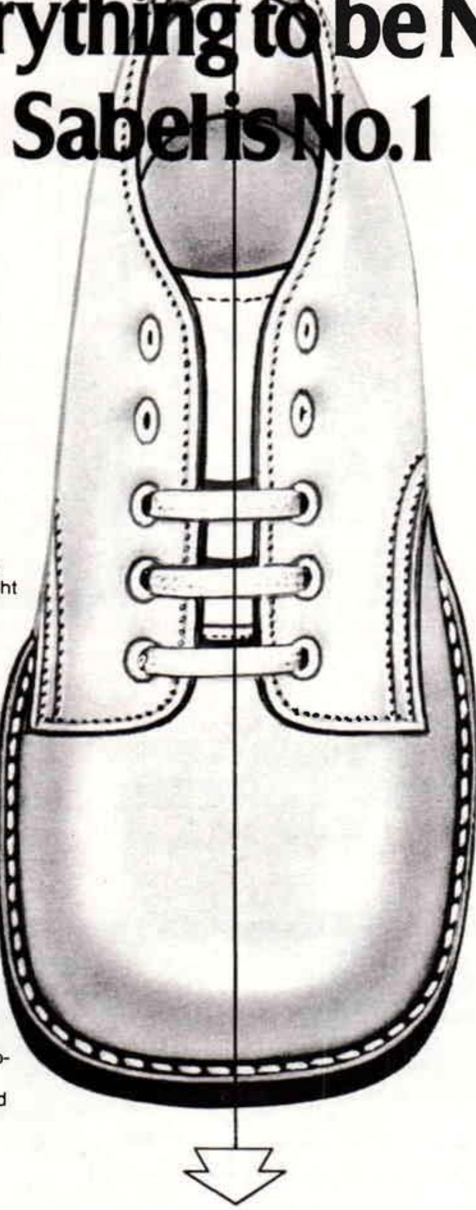

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