Volume 37
Number 4

Orthotics and Prosthetics

Journal of the American Orthotic and Prosthetic Association

The Backbone

Catalog K01

The Low Profile Sach Foot

Orthotics and Prosthetics

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Winter 1983–84 **Volume 37, No. 4**

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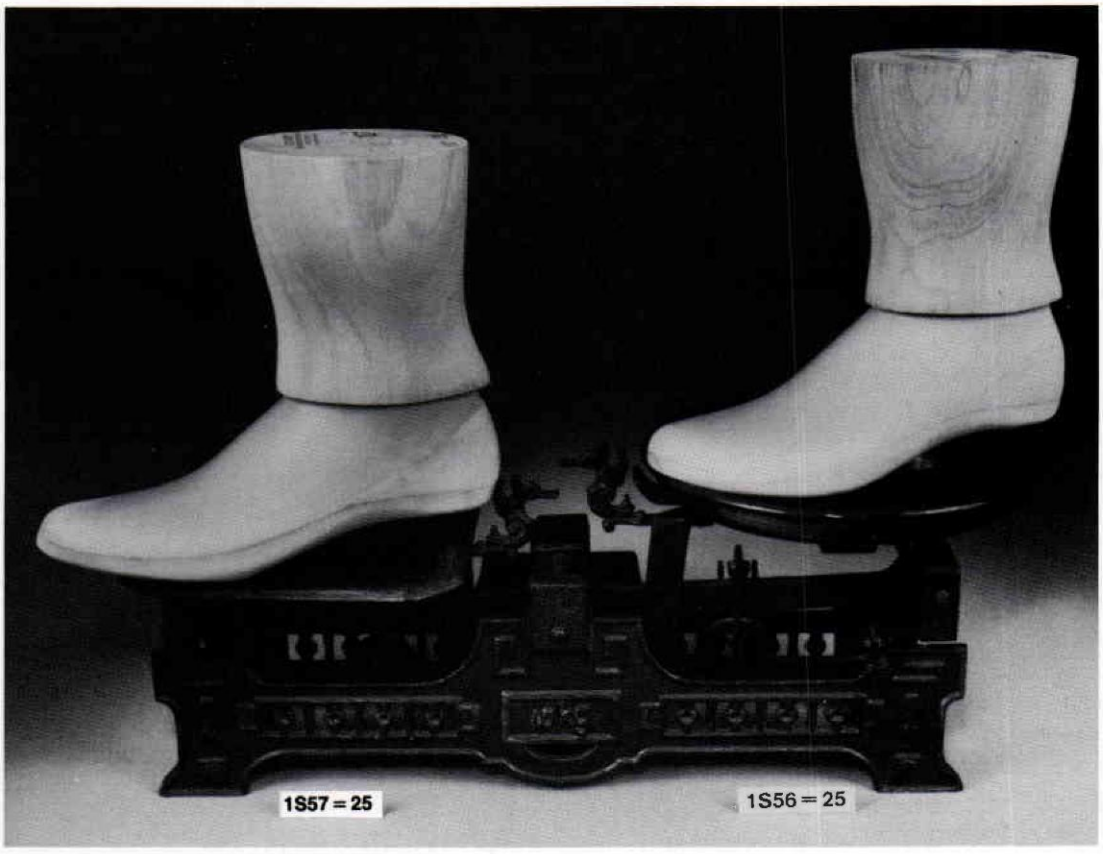

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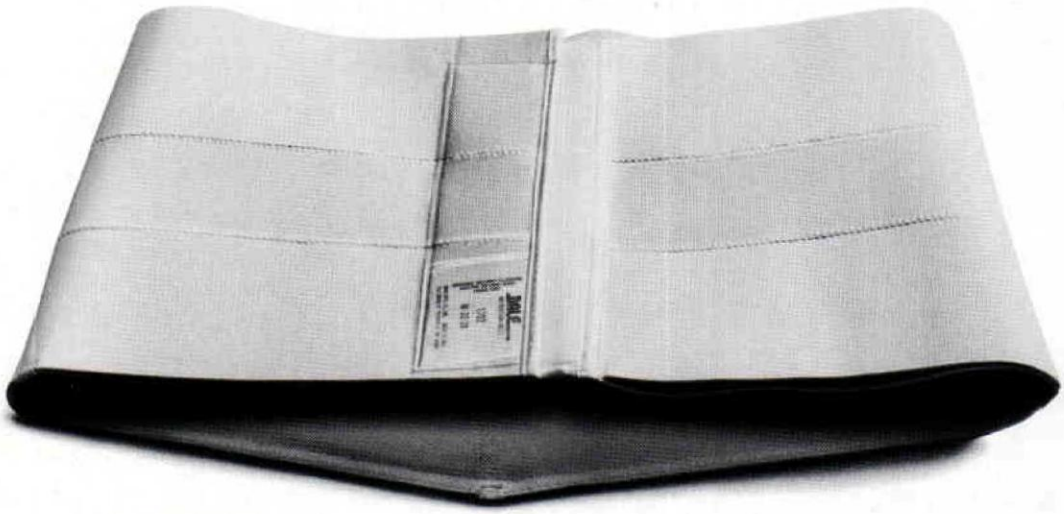

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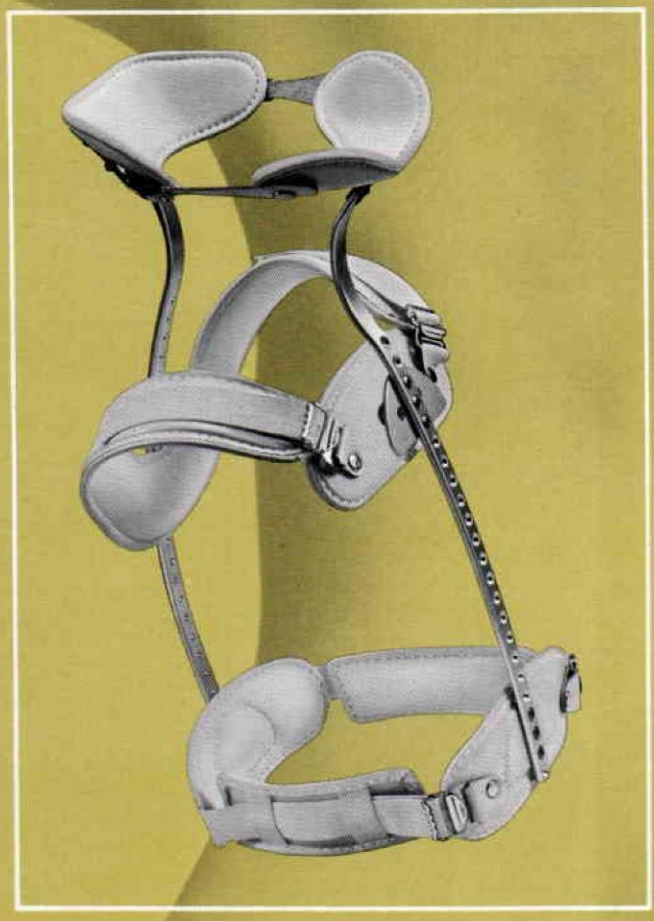

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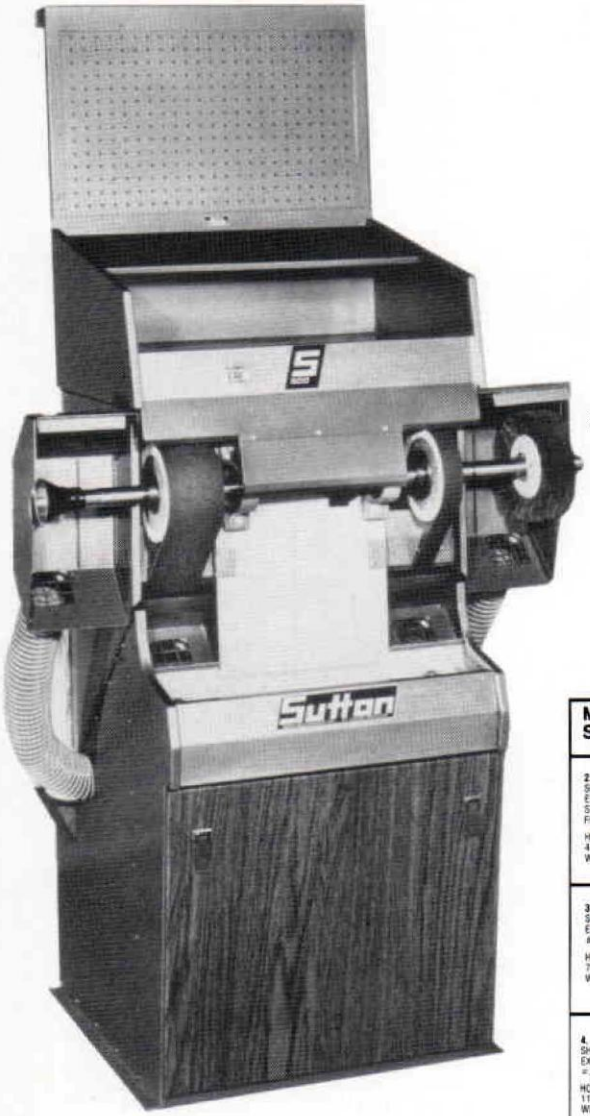

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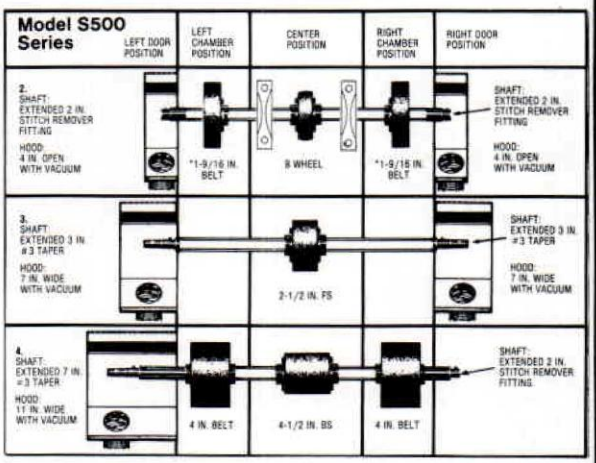

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April 2, Canadian National Society for Prosthetics and Orthotics, Biennial Meeting, Westin Bayshore Hotel, Vancouver, British Columbia, Canada. Contact: Canadian National Society—ISPO, Prosthetics-Orthotics Department, Chedoke-McMaster Hospitals, P.O. Box 2000, Station "A," Hamilton, Ontario L8N 3Z5 Canada.

April 3–5, Canadian Association of Prosthetists and Orthotists Biennial National Convention, Westin Bayshore Hotel, British Columbia, Canada. Contact: Box 320-810 West Broadway, Vancouver, B.C. V5Z 4O9 Canada.

April 6–7, New England Academy Chapter Annual Meeting, Worcester Marriott, Worcester, Massachusetts.

April 7–8, ABC Spring Written Examinations, Ramada Inn Old Town, Alexandria, Virginia; Holiday Inn O'Hare Kennedy, Chicago, Illinois; Airport Executive Inn, San Francisco, California. Contact: ABC National Headquarters, 703-836-7114.

April 11–15, The Pacific Rim Orthotics and Prosthetics Conference, Hotel International, Maui, Hawaii. Endorsed by the Academy and INTERBOR.

April 12–15, AOPA Region IV Annual Meeting, Waverly Hotel at the Galleria, Atlanta, Georgia.

April 19–24, 1st International Meeting on Leisure, Recreation, and Sports, organized by the Rehabilitation International Commission and sponsored by R.I., the Japanese Society for Rehabilitation of the Disabled and the governments of Gamagori City and Aichi Prefecture, Gamagori, Japan. Contact: Japan Sun Industries, Kamegawa, Beppu 874-01 Japan.

April 25–27, "Spinal Cord Injury: Rx for Management," a three day conference sponsored by Dallas Rehabilitation Institute and Dallas Rehabilitation Foundation will be held at the Holiday Inn Brookhollow, Dallas, Texas. Pre-registration fee (before April 16, 1984) is $150. Special one day rates and group discounts (ten or more) are available. Over 26 experts in the field of spinal cord injury rehabilitation will be speaking. For more information or to receive registration materials, contact the public relations department, Dallas Rehabilitation Institute, 7850 Brookhollow Rd., Dallas, TX 75235, (214) 920-8040.

May 3–5, AOPA Regions I, II, and III Combined Annual Meeting, Concord Hotel, Kiamesha Lake, New York.

May 13–19, Ninth International Congress of Physical Medicine and Rehabilitation, Jerusalem, Israel. Contact: Kenes, 29 Mamred Street, P.O.B. 29784, 61297 Tel-Aviv, Israel.

May 24–26, AOPA Region V Annual Meeting, Amway Grand Plaza Hotel, Grand Rapids, Michigan.

June 1–3, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting, Lake Arrowhead, California.

June 4–8, 15th World Congress of Rehabilitation International on theme, "Information, Awareness, and Understanding for Integration of Disabled Persons and Society," Lisbon, Portugal. Contact: (Program) Rehabilitation International, 432 Park Avenue South, New York, New York 10016.

June 12–July 4, 7th World Wheelchair Games (formerly Paralympics), University of Illinois, Champaign, Illinois. Contact: Prof. Timothy Nugent, Rehabilitation Education Center, 1207 South Oak Street, Champaign, Illinois 61820.

June 16–28, 1984 International Games for the Disabled, sponsored by the International Sports Organization for the Disabled, Nassau County, Long Island, New York. Contact: Mr. Michael Mushett, Director, 1984 International Games for the Disabled, c/o Special Populations Unit, Eisenhower Park, East Meadow, New York 11554.

June 17–22, "1984—The Bright Side," The Second International Conference on Rehabilitation Engineering, combined with the 7th Annual Conference on Rehabilitation Engineering, Congress Centre, Ottawa, Ontario, Canada. Sponsored by the National Research Council of Canada, the Rehabilitation Engineering Society of North America, and the Canadian Medical and Biological Engineering Society. Contact: Conference Services, National Research Council of Canada, Ottawa, Ontario, Canada K1A 0R6.

June 21–24, AOPA Region VI and the Academy Midwest Chapter Annual Combined Meeting, Holiday Inn, Merrillville, Indiana.

June 28–30, AOPA Regions VII, VIII, X, and XI Combined Meeting, North Shore Convention Center, Coeur d'Alene, Idaho.

September 30–October 5, 16th Congress of the International Society for Orthopedic Surgery and Traumatology (SICOT), London, England. Contact: Conference Services, Ltd., 3 Bute Street, London, SW7 3EY, United Kingdom.

October 15–21, AOPA General Assembly and International Congress, Fontainebleau Hotel, Miami Beach, Florida. Contact: AOPA National Headquarters, 703-836-7116.

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January 30–February 3, Academy Annual Meeting and Seminar, Cathedral Hill Hotel, San Francisco, California. Contact: Academy National Headquarters, 703-836-7118.

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A New Concept in Orthotics Joint Design—The Northwestern University Knee Orthosis System

Jack L. Lewis, Ph.D.
William D. Lew, M.S.
Carl M. Patrnchak, R.P.T., C.O.
George T. Shybut, M.D.

INTRODUCTION

When an orthosis is applied to the knee, it should, hypothetically, allow a full, unrestricted range of motion to occur, except at appropriate limits of motion where orthotic constraints are intentionally introduced to provide the extra stability required to compensate for soft tissue insufficiency. For example, an orthosis applied to the knee to correct recurvatum should in no way restrict normal flexion, and should introduce a constraint force only near extension where the extra stability is required.

In reality, however, commercially available orthoses fall short of this ideal. One of the major problem areas is that orthotic knee joints used currently follow kinematic or motion pathways which are considerably simpler than those of the natural knee joint, whose motion is three dimensional in nature. Single axis hinges are most common, although other designs, such as the polycentric, have evolved in an attempt to more closely simulate the complex rolling and sliding which accompanies flexion-extension of the natural joint. The mismatch between the orthotic and natural knee joint motions can cause an unwanted constraint force or binding to occur, with the subsequent pistoning of the orthotic components over the lower limb, producing restriction of the normal range of motion, distal migration and misalignment of the orthosis, and skin pressure discomfort.

Our laboratory has developed an orthotic knee joint which more closely mimics the motion of the natural knee. We believe it offers some significant advantages over existing orthotic joints, and allows design of more effective knee orthoses. This report describes the proposed orthotic joint, the rationale behind its design, its advantages, and evidence of its value. A second report will present a description of a complete knee orthosis using the improved joints.

DESCRIPTION OF THE JOINT

The joint consists of a metal, multicurvature femoral component in the shape of the sagittal profile of the distal femur, and a slotted plastic tibial component with a larger, flatter articulating surface approximating the profile of the proximal tibia (Figure 1). The femoral component articulates within the tibial slot so that the surfaces become highly conforming or en-

gaged in extension, preventing anterior-posterior motion, as in the natural joint. In flexion, however, the smaller posterior femoral curvature provides for a low degree of conformity or capture by the tibial curvature, allowing the femoral articulating surface to roll and slide anterior-posteriorly over the tibial component, thereby imitating the natural knee. The component curvatures and amount of capture are variable in design, and can be chosen depending upon the patient application. We have chosen a set of curvatures which limits the size of the orthotic joints, yet allows a possible eight millimeters of anterior-posterior displacement to accompany flexion.

Stability is added to the orthotic joint through the presence of inextensible dacron straps which cross the joint (Figure 1). These straps are attached distally to the plastic tibial component, and proximally to a metal ligament attachment plate, which is in turn fastened to the sidebar of the femoral component. The straps simulate the location, orientation, and function of knee ligaments. These "ligament straps" tighten and become lax at different times during the range of motion, yet allow the anterior-posterior rolling and sliding of the femoral component over the tibial component to occur. The number and location of these ligament straps can be varied and is dependent upon the particular mode of stability required. At present, we routinely use three basic ligament strap configurations: an anterior cruciate design, a collateral design, and a posterior cruciate design (Figures 2a, b, c). Note that each joint design has three straps which are located and oriented so as to sequentially tighten at 0°, 45°, and 90° of flexion, respectively.

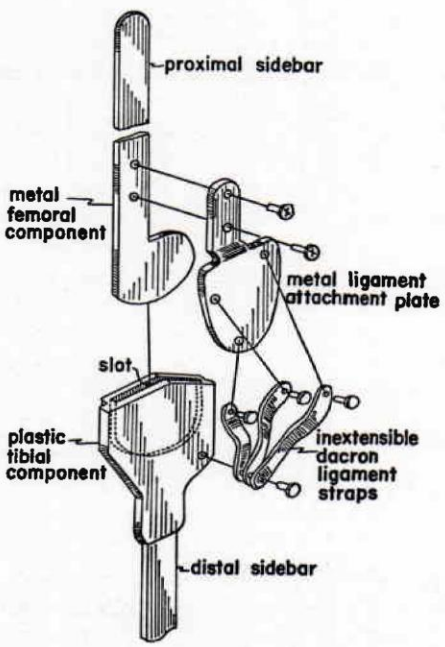

Figure 1: The above sketch shows the components of the proposed orthotic joint design: metal femoral component, slotted plastic tibial component with metal sidebar, metal ligament attachment plate, and three dacron ligament straps (the collateral design in this example).

For example, Figure 3 shows the sequential tightening of the three straps of a posterior cruciate joint. Each one of the straps tightens at a particular flexion angle, while the other two remain lax. Note that when a ligament strap is tight, it has approximately the same line of action or orientation as the natural posterior cruciate ligament. Figure 3 also shows the femoral component rolling and sliding posteriorly with flexion. A computerized mathematical model was used to define potential strap locations. The data generated by the model provides length patterns for all possible strap attachments, predicting where in the flexion range the strap would get tight, and in what direction the subsequent strap force would act.

BIOMECHANICAL DESIGN RATIONALE

Because the surfaces of the anatomical knee articulate without a great deal of capture, the muscles and ligaments (their attachment locations and orientations) must precisely interact with the geometry of the articular surfaces to generate lower limb function yet provide stability. For example, it has been hypothesized by Lewis, et. al. (1983)[3] that knee ligaments have a dual function. The "high-level" function is when ligaments provide stability in a traumatic situation, where the external

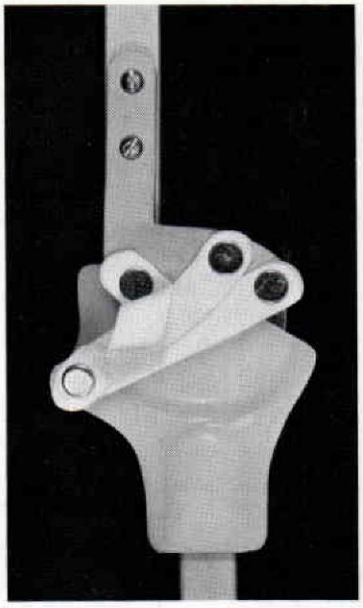

Figure 2A: Anterior cruciate joint design.

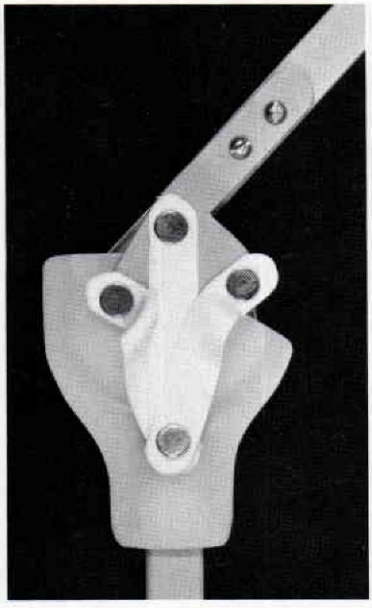

Figure 2B: Collateral joint design.

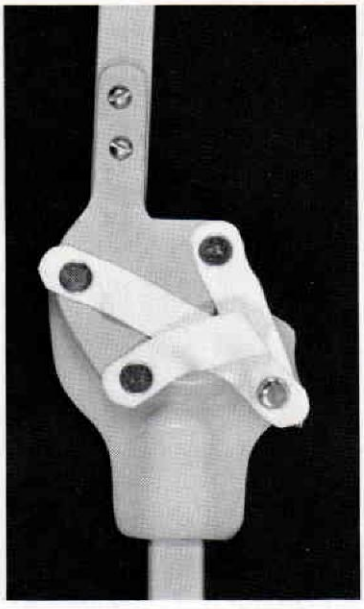

Figure 2C: Posterior cruciate joint design.

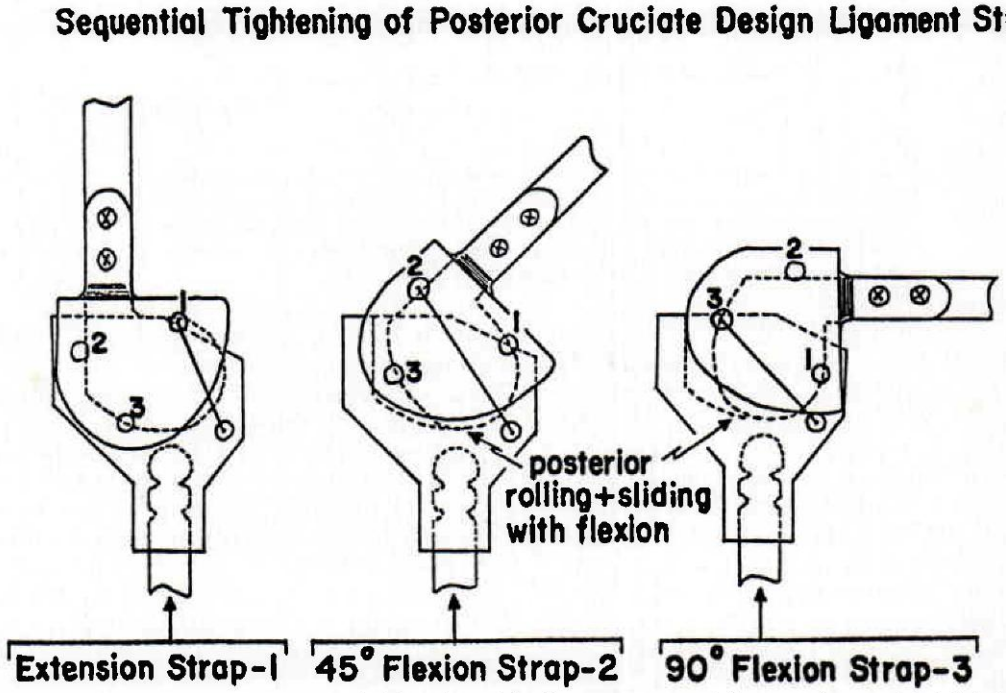

Figure 3: Sketch shows the sequential tightening of the ligament straps of the posterior cruciate joint design, at extension (left), 45 degrees flexion (middle), and 90 degrees flexion (right).

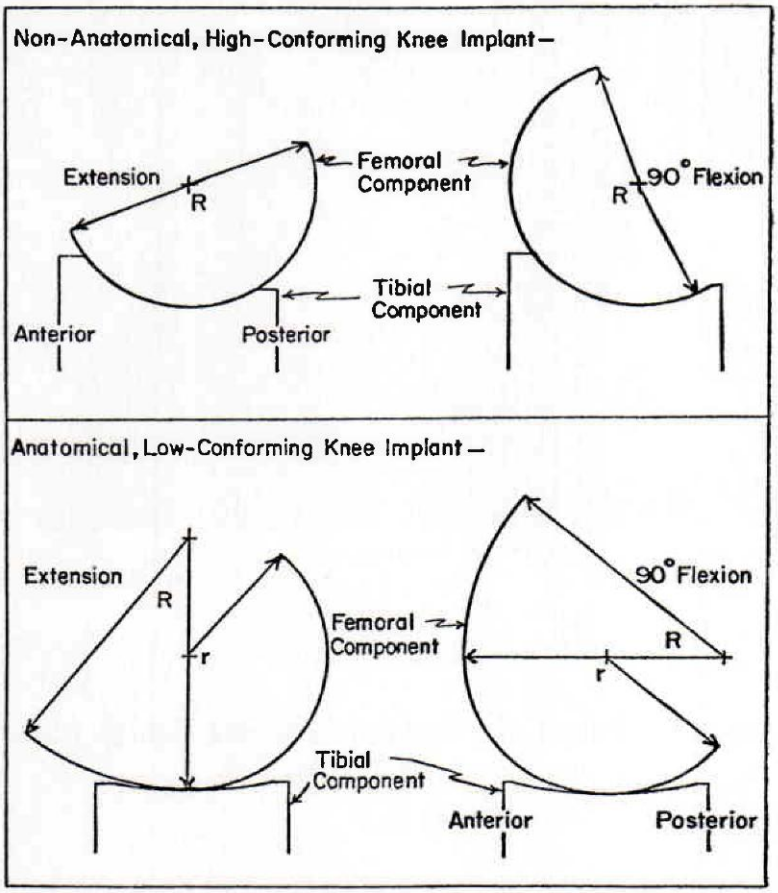

Figure 4: The proposed orthotic joint design is based upon earlier research regarding the interaction of knee ligament mechanics with internal knee prostheses. A sagittal view of the cross sections (radii of curvatures—R,r) of the tibial and femoral components of these implants are shown above.

load occurs too rapidly for the muscles to equilibrate. The "low-load" function is when ligaments keep the correct apposition of the articular surfaces during muscle-generated function, providing for proper joint lubrication and normal contact forces. This low level function is particularly dependent upon the relationship of the geometry of the articular surfaces and ligaments. As previously mentioned in the Introduction, when simplified artificial joints are placed in (total joint replacements) or around (orthoses) the knee, constraints are generated in the natural joint structures as they oppose the motions imposed by the artificial joints. This constraint is recognized externally as pistoning, and internally as, among other things, ligament incompatability.

As an example of the above, Lew and Lewis (1982)[1] performed a study in which cruciate ligament forces were measured during the flexion of specimens containing a low conforming, anatomically shaped knee implant design, as well as a high-conforming, non-anatomical implant design (Figure 4). In the non-anatomical implant, which did not allow rolling and sliding to accompany flexion as in the natural joint, the full range of motion was

restricted to about 60 degrees of flexion, and abnormally large constraint forces appeared in the posterior cruciate ligament. The anatomical implant, on the other hand, allowed the rolling and sliding of the natural joint, so that a full range of flexion was attainable, and cruciate ligament forces approached that of a normal knee. The above findings can be extended to the design of orthotic joint components. The orthotic articular surfaces should have the freedom to reorient themselves as dictated by the internal ligaments (or muscles, for that matter) for as many of the components of natural joint motion as possible. In this way, unwanted constraints will be minimized, and an unrestricted range of motion can be obtained.

The design of the proposed orthotic joints closely follows this biomechanical principle. Depending upon the constraints introduced to the orthotic joints and complete knee orthosis (because of the patient's particular condition), the proposed orthotic, articular surfaces can potentially allow five of the six possible components of knee motion, the exception being medial-lateral displacement (refer to Figure 5). Anterior-posterior rolling and sliding of components during flexion-extension are possible, as described earlier. Distraction of component articular surfaces is allowed, which in turn permits varus-valgus rotations to occur at any flexion angle. Transverse rotations are possible through the anterior-posterior and distractive displacements of the joint components. Thus, the orthotic joint articular surfaces reorient themselves as dictated by the internal knee structures, to a greater degree than other commonly used orthotic joints. Given any particular soft tissue insufficiency, specific ligament strap constraints can be added to restrain any of the above affected motion components, so that the joints allow a full range of knee motion to occur. The only exception is for points in the motion where extra stability is required to compensate for the soft tissue instability present. The validity of this design concept will be demonstrated by a mechanical evaluation, a description of which follows in the next section of this report.

Another biomechanical principle to be considered is that the geometry of the natural knee ligaments is such that they become loaded during a wide range of external joint load conditions. Lewis, et. al. (1983)[3] measured ligament forces in a series of seven specimens, with the purpose of cataloging external load conditions (applied to the tibia) which loaded particular ligaments.

The anterior cruciate was found to be highly loaded during several load conditions near extension (anterior-directed force or anterior drawer, varus, and internal rotation). The posterior cruciate ligament was highly loaded for various external load directions near 90 degrees flexion (posterior-directed force or posterior drawer, varus-valgus, and internal-external rotation). The medial collateral ligament was highly loaded during internal-external rotation and valgus, throughout the flexion range. The lateral collateral ligament was highly loaded during varus and internal rotation, throughout the flexion range. The anterior cruciate and both collateral ligaments were highly loaded during hyperextension. Given the previous argument for anatomically shaped orthotic joint components, the above information is important when designing constraints into the orthotic joints (such as "ligament straps") or the complete orthosis, to provide stability for specific ligament insufficiencies.

The orthotic ligament straps are oriented and located in relation to the orthotic articular surfaces so as to function similar to the natural knee ligaments. The straps sequentially tighten within the flexion range, such that their lines of action (when taut) can assist in the restraint of the array of external load conditions described above. The ligament straps, in conjunction with changes to the orthotic interface (to be described in a later report), provide stability to the anatomically shaped orthotic joint surfaces, and can be tailored to handle specific ligament insufficiencies.

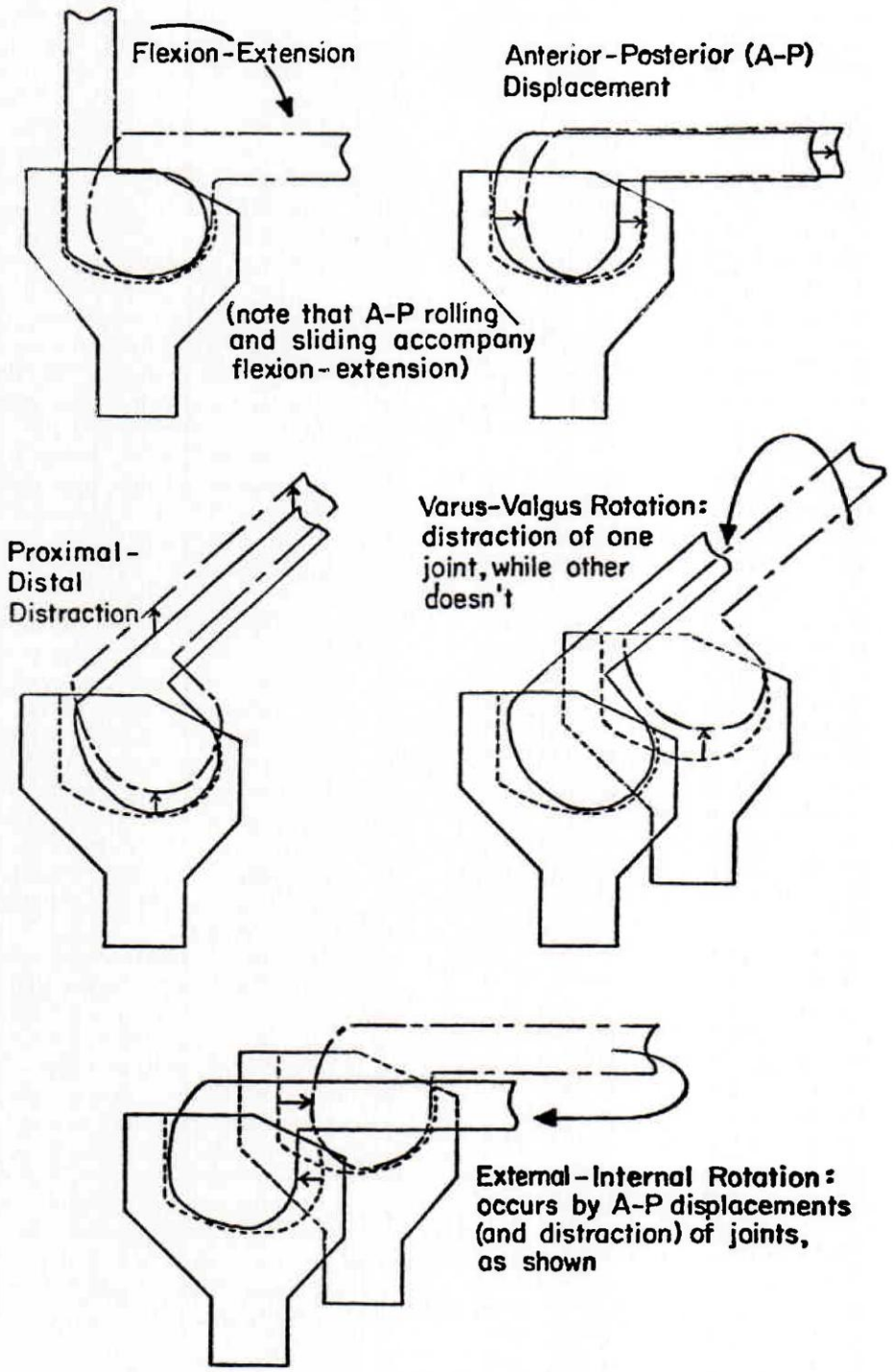

Figure 5: Sketch shows how the various components of knee joint motion can occur with the proposed orthotic joint design. Medial-lateral displacement is the only motion not allowed.

MECHANICAL VERIFICATION OF THE DESIGN

The authors have previously reported a procedure for comparing the efficacy of orthotic knee joints, based upon their tendency to produce pistoning (refer to Lew, et. al. 1982[2] for details). Pistoning transducers were designed, which were capable of being interchangeably incorporated into the sidebars of various orthotic joint designs. As a subject wearing an evaluation orthosis performed functional activities, the transducers on the medial and lateral orthotic joint sidebars would directly measure the resulting pistoning constraint forces generated by the conflict between the simplified orthotic joint motion and the complex natural joint motion. This procedure was used to compare the pistoning tendency of the proposed anatomically shaped orthotic joints (the collateral design) with three commonly used, commercially available orthotic joint designs: single axis hinges, posterior offset hinges, and polycentric hinges. The resultant pistoning constraint forces were measured as a subject performed loaded and unloaded flexion, level walking, getting in/out of a chair, and climbing up/down a step.

Figure 6 presents the combined results over all the activities, for each joint design, the mean and standard deviation of the combined resultant pistoning forces, and the normalized mean forces. The data suggests that the proposed orthotic joints, because of their semi-constrained, anatomically shaped design, generated an average of 76 percent less pistoning constraint than the commercially available joint designs ($p \leq .01$). Also note that there is no statistical significance in the differences among the pistoning forces of the three commercially available joint designs.

DISCUSSION

In summary, an improved orthotic knee joint system has been designed, based

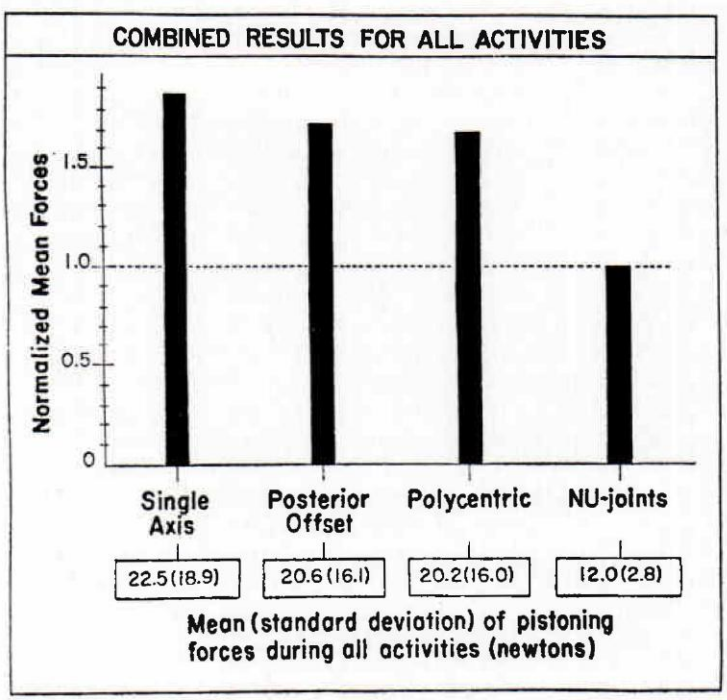

Figure 6: At left is a summary of the resultant pistoning constraint forces for a combination of all the test activities for each joint design. The means and standard deviations of the resultant pistoning forces are presented, as well as the normalized mean resultant pistoning forces over the four joint designs.

upon biomechanical principles associated with knee motion and ligament mechanics. The orthotic joint articular surfaces are anatomically shaped and semi-constrained, so as to allow a close approximation of the components of natural knee motion, particularly the anterior-posterior rolling and sliding which accompanies knee flexion and extension. Stability is added to the orthotic joint system through various configurations of inextensible dacron straps crossing the joints. The precise location and orientation of these ligament straps is determined by a mathematical model.

The functional result of this design concept is that the orthotic joint motion can closely match the motion pathways of the natural knee, except at particular points in the range of motion where extra stability is added to the orthotic joints and interface to compensate for soft tissue insufficiency. This improvement was demonstrated by the fact that in a mechanical evaluation, the proposed orthotic joints generated an average of 76 percent less pistoning constraint force than other currently popular joint designs. Thus, the improved design more closely matches natural knee motion, decreasing the effects of the binding, motion restriction, and discomfort, often associated with pistoning.

Another advantage of the anatomical orthotic joints is their high degree of variability; that is, various joint parameters can be tailored to address specific soft tissue problems. Many variations of soft tissue instability exist, due to the involvement of individual or combinations of knee ligaments and/or capsular structures, each with its own instability direction. Many of the current, commercially available orthotic joints cannot be modified to handle specific ligament problems, as one design is used regardless of the type of soft tissue insufficiency present.

The following are some of the potential variations of the basic proposed orthotic joint designs. The mathematical model can be easily adapted to compute locations of ligament straps which tighten at other flexion angles of interest (besides the 0°, 45°, and 90° straps of the basic design).

It is also possible to design various hybrid strap configurations by combining one or more ligament straps from the cruciate and/or collateral joint designs. For example, an individual presenting with an antero-medial rotatory instability resulting from an anterior cruciate and medial collateral ligament deficit may require a hybrid combination of the anterior cruciate and collateral strap configurations.

Other variables are the curvatures of the articulating surfaces of the orthotic joints. For example, given a person with a total knee replacement requiring orthotic treatment, the orthotic joint component curvatures and strapping configuration can be tailored to match the geometry of the internal implant. Application of the orthotic joint designs to specific clinical situations will be described by several case studies presented in a later report.

The degree of suspension or fixation of a knee orthosis to the lower limb is intimately related to the motion pathways allowed by the associated orthotic joints. Since the motion of most currently available orthotic joints does not match natural knee kinematics closely enough, a tightly fitting interface will just magnify the pistoning constraint. This situation would be particularly harmful, for example, if an orthosis was intended to protect surgically reconstructed knee ligaments, since the pistoning constraint may cause stretching of the healing tissue. On the other hand, if the interface components were not tightly fitting to relieve the effects of the pistoning constraint, the orthosis would not provide the necessary stability to the joint. Thus, improvements to the interface suspension are limited by orthotic joint kinematics.

However, in the case of the proposed orthotic joints, it was demonstrated that the motion mismatch and resultant pistoning were reduced, thereby setting the stage so that improvements to the orthotic interface can be realized. This also will be dealt with in great detail in a later report.

In a subsequent report, the inclusion of the improved orthotic joints in a completed knee orthosis will be described. Some of the topics included will be the biomech-

anics of knee orthotic suspension, suspension improvements contained in the Northwestern orthosis system, fabrication details, and a group of case studies showing the adaption of the orthosis system to specific knee ligament problems.

NOTES

[1]Lew, W.D. and Lewis, J.L., "The Effect of Knee Prosthesis Geometry on Cruciate Ligament Mechanics During Flexion," *Journal of Bone and Joint Surgery*, Vol. 64-A, 1982, pp. 734–739.

[2]Lew, W.D., Patrnchak, C.M., Lewis, J.L., and Schmidt, J., "A Comparison of Pistoning Forces in Orthotic Knee Joints," *Orthotics and Prosthetics*, 36, 1982, pp. 85–95.

[3]Lewis, J.L., Lew, W.D., Shybut, G.T., Jasty, M., and Hill, J.A., "Biomechanical Function of Knee Ligaments," in press, to appear as a chapter in a book entitled *Sports Medicine of the Knee*, published by the C.V. Mosby Company.

ACKNOWLEDGMENTS

This work was supported by Grant No. G008200024 from the National Institute of Handicapped Research, Department of Education, Washington, D.C. 20202.

United States Letters Patent Number 4,361,142—November 30, 1982.

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Fabrication and Fitting Instructions: Trilateral Perthes Orthosis

(Developed by M.A. Tachdjian, M.D. and Loren D. Jouett, C.P.O.)

Carlton Fillauer, C.P.O.
Charles H. Pritham, C.P.O.

Legg-Perthes' disease, osteochondrosis of the femoral head, is a condition that affects the femoral head of pre-adolescents, primarily boys of three to ten years of age. For some unknown reason, the femoral head suffers an episode of avascular necrosis followed by spontaneous regeneration of the bone. During the process, the head is malleable, and, if subjected to deforming forces, will adopt a flattened aspect that can eventually cause pain and arthritis.

To preclude this possibility, a number of treatment schemes have been developed over the years. Many of them rely on the so-called containment theory. This holds that, if the femoral head is held inside the acetabulum as the bone regenerates, it will adopt a normal congruent aspect. To achieve the position of containment, the femur is abducted and internally rotated. In addition to these criteria, the Trilateral Perthes Orthosis[1] endeavors to unweight the femur with ischial-gluteal weight-bearing. In contrast, the Toronto Orthosis[2] relies on bilateral abduction and internal rotation and dispenses with the weight-bearing component, and the Scottish Rite Orthosis[3] relies solely on bilateral abduction.

The orthosis described in this article is essentially the same as was originally described by Mihran A. Tachdjian, M.D. and Loren D. Jouett, C.P.O. in their article of 1968. That orthosis (Figure 1) included a medial upright with knee joint and drop lock and a patten bottom. An ischial-gluteal brim unweighted the affected hip. The medial edge of the brim was purposely fit

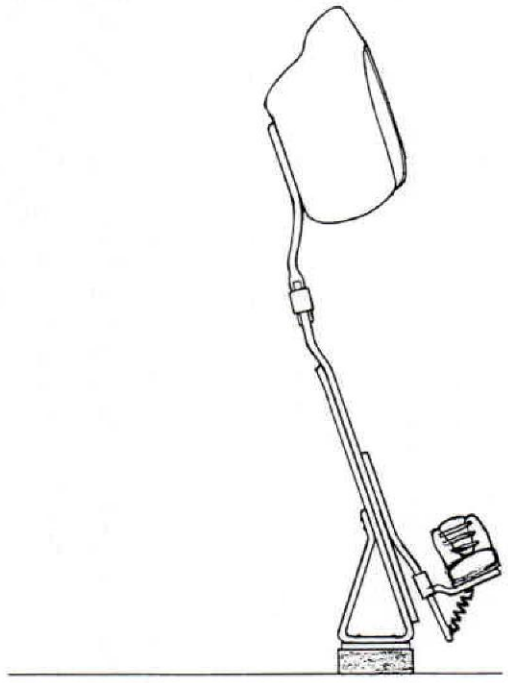

Figure 1: Trilaterial orthosis as originally described by Tachdjian and Jouett.

high enough to cause discomfort if the patient adducted to 20° or less. This, in combination with the lengthening of the distal components, encouraged abduction. The posterior portion of the socket, proximal of the ischial seat, was fit high to encompass the inferior portion of the buttock and to block external rotation. This, in combination with the strap on the shoe, encouraged internal rotation.

The combination of the three factors—unweighting of the limb, abduction, and internal rotation—strove to accomplish the goals of protecting the diseased hip and maintaining it in the proper position for healing. The stirrup was rivetted to the shoe and free to move up and down and rotate on the round slide guide extension bar. A spring running from the end of the slide guide extension to the stirrup provided suspension, and a 2"–2½" compensatory lift on the contralateral shoe completed the orthotic design.

There have been, of course, other designs similar to the design of Tachdjian and Jouett. Writing in 1970, Glimcher, Radin, and Amrich[4] described an orthosis they referred to as the Pogo Stick Brace (Figure 2). Like the Trilateral Orthosis, this device incorporated provisions for ischial-gluteal weight-bearing, abduction, and internal rotation. Unlike the Trilateral Orthosis, it attempted to reduce forces on the hip to the absolute minimum by reducing the activity of the hip abductors. The authors correctly argued that the muscular forces acting on the hip were, in some phases of gait, many times the body weight and cited in support of this fact the work of Inman, Paul, and Rydell. Their decision to eliminate the abduction forces on the hip is, of course, a logical extension of the decision to eliminate weight-bearing.

This objective of eliminating abduction forces was to be accomplished not just by putting the hip abductors in a relaxed position, but also by denying the lever arm (the leg in this case) any purchase. To this end, there was no rigid attachment to the patient's leg distally. An "ankle band retainer" consisting of spring steel, about 12 inches long, and bent in a "U" shape, was fastened to the weight-bearing pylon tube.

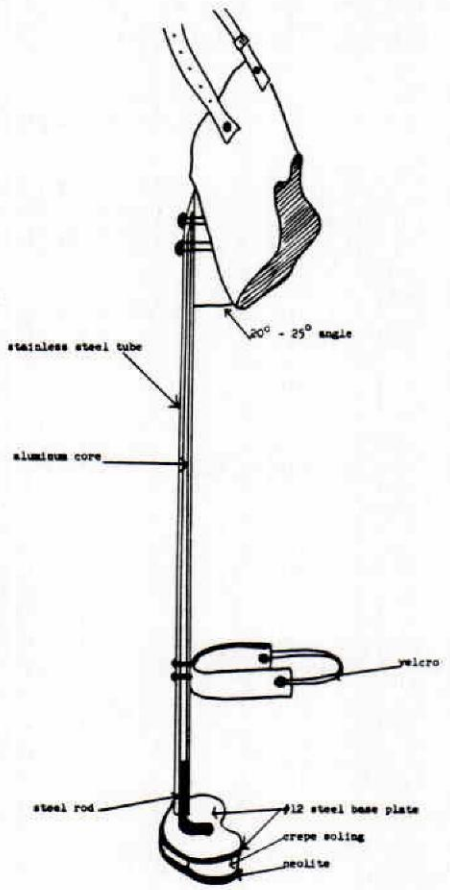

Figure 2: Pogo stick brace as described by Glimcher, Radin, and Amrich. Illustration reprinted from the original article (4) and used with the permission of the editor and authors.

A loose Velcro® strap laterally completed the retainer. This retainer and its strap were loose enough that no mechanical purchase could be obtained by the leg. Indeed, the parents of the child being fitted were to modify his pants legs to very loose oversize bell bottoms so that no stabilization from that source could be obtained. A shoulder strap to suspend the orthosis was to be worn, as well as a spring wire dorsiflexion assist orthosis on the involved side. A 3½"–4" buildup on the contralateral shoe completed the prescription. The authors stated that the first child had been fitted in March of 1964 and that at the time of writing, 24 children had been fitted.

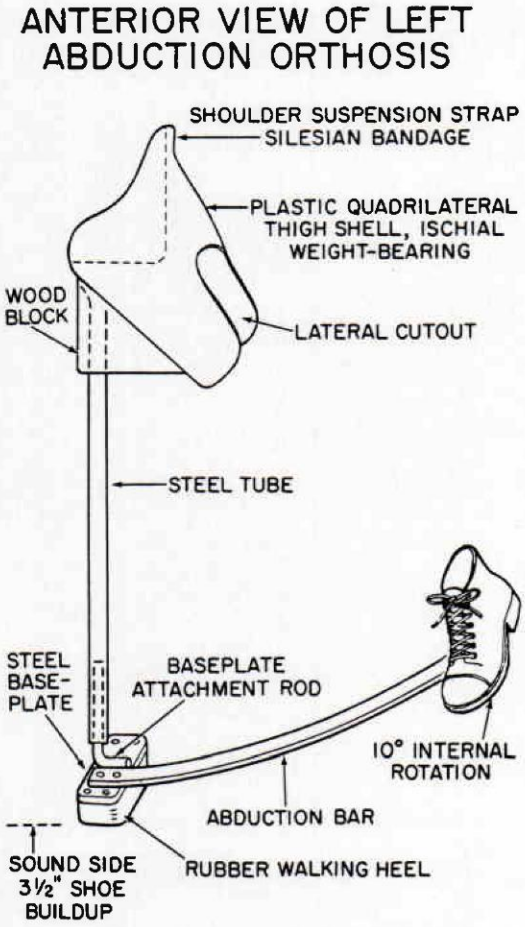

Figure 3: Abduction orthosis described by Birkeland and Zettl. Illustration reprinted from the original article (5) with the permission of the editor and authors.

In 1974, Birkeland and Zettl[5] described another variation on the same theme. This (Figure 3) consisted of the now familiar plastic trilateral socket, a medial weight-bearing pylon tube, and an abduction bar attached to the pattern bottom and to the patient's shoe to hold the leg in abduction and internal rotation. A shoulder strap and sometimes a Silesian band were used to suspend the orthosis, and, like the pogo design, no provision was made for knee flexion.

Birkeland and Zettl deliberately eliminated all moving parts from the orthosis and cited as the advantages of such a move ruggedness of design and minimal maintenance. Apparently, little restriction of patient's activities were experienced, for they described instances of children climbing trees, riding bicycles, and playing baseball with the orthosis on. The authors stated that they had had six years experience with this design and that it was their prescription of choice for unilateral involvement. For instances of bilateral involvement they relied on the Newington and Toronto orthoses.

Birkeland and Zettl, as did the other authors cited, described the use of a shoe lift (approximately 2½"–3½") on the opposite side. They felt that use of such a lift was necessary to prevent the child from bearing weight on the involved limb. Without the lift it would be possible to flex the knee of the sound side, adduct the hip of the involved leg, and bear weight through the abduction bar. The use of a lift on the sound side meant that so much flexion of the involved hip would be necessary that it would quickly become uncomfortable.

In contrast, the trilateral orthosis depends on the length of the medial support bar and patten bottom as well as the discomfort of the medial brim to hold the leg in abduction. This, of course, obviates the expense of obtaining and maintaining a lift on the opposite shoe.

It has also been argued that a shoe lift is necessary to protect the diseased hip should the patient adduct it. The theory is that if the child is not wearing a shoe lift, and if he adducts the involved limb, then a state of pelvic obliquity will result and an extreme amount of the femoral head will be exposed laterally. If he is wearing a shoe lift on the opposite side, and if he should adduct to the neutral position, at the very least, he will not be exposing an extreme proportion of the femoral head even if the position is undesirable.

The problem with this argument is that if the patient is fitted with a shoe lift, and if the pelvis is to be level, then the orthosis must be adjusted to establish a level pelvis. The result is the same relationship as without a shoe lift, only the child is 2"–3" taller. In contrast to this theory, the authors con-

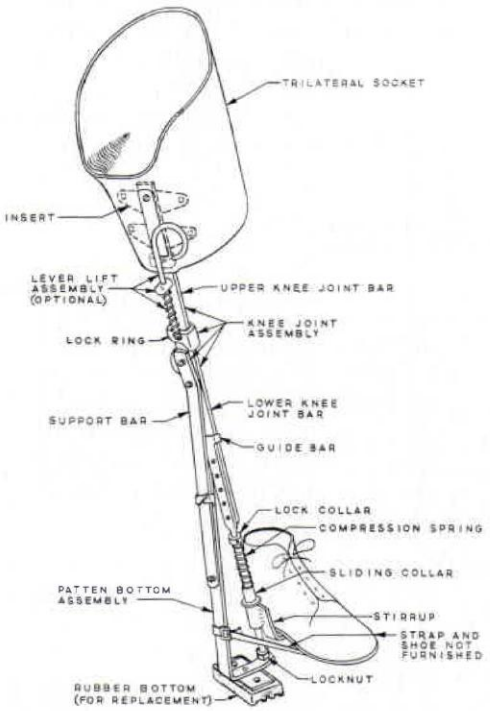

Figure 4: View of orthosis showing relationship of parts and proper nomenclature.

tend that the medial brim should be relied on to prevent adduction past a position of 20° abduction and should be deliberately modified so as to be uncomfortable in this position.

As originally described by Tachdjian and Jouett, the trilateral orthosis (Figure 1) was fabricated with a slide guide extension bar mounted on the lateral side of the distal knee joint upright and unsupported distally. Clinical experience in Chattanooga, Tennessee led to the conclusion that this arrangement was too weak, and Durr-Fillauer adopted the present design (Figure 4) in the version it now markets. Dates on the blueprints of parts for the prefabricated kits reveal that work began in the early part of 1969. The present day design has remained substantially unchanged since then.

Originally, when orthoses were centrally fabricated, it was specified that the orthotist must furnish full measurements and a mold of the patient for the weight-bearing brim. However, as was often the case, orthotists neglected to secure an impression of the patient's leg before dismissing him from the office. As a convenience to orthotists in this predicament, we would endeavor whenever possible to furnish a laminated brim from one of the models on hand, selected on the basis of the measurements furnished by the orthotist. The resulting brims proved to be so suitable that some orthotists began ordering on the basis of measurements alone. Over a period of four years, from 1971–1975, a complete range of brim sizes accumulated. Today, prefabricated brims are available in a range of fourteen sizes, from 10½"–17", in one-half inch increments.

Indications for use of the Trilateral Perthes Orthosis include treatment of unilateral Legg-Perthes disease in pre-adolescents. While the orthosis may be used bilaterally, consideration should be given to orthoses intended for bilateral involvement, such as the Scottish Rite Orthosis.

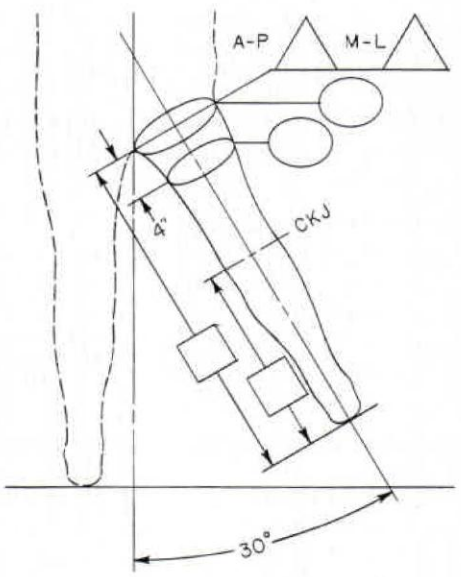

Figure 5: Tracing and measurements required. Solid line indicates the portion to be traced. Dotted line indicates patient's position.

Experience has shown that use of the trilateral orthosis bilaterally can be very uncomfortable in the perineum as well as difficult to walk on.

OBJECTIVES OF THE TRILATERAL PERTHES ORTHOSIS

1. To hold the hip in a position of
   a. 30 degrees abduction
   b. 10–15 degrees internal rotation
2. The application of ischial-gluteal weight-bearing to unweight the hip
3. To permit ambulation

FABRICATION

Measurements & Tracing: (Figure 5)

1. Patient position: A mid-sagittal line, longer than the patient, is drawn on a sheet of paper, and the patient is centered on the line:
   a. supine and with spine aligned in a straight manner
   b. sound limb parallel with mid-sagittal line
   c. affected limb—30 degrees hip abduction, neutral rotation, knee extended, ankle neutral.
2. Trace the affected limb from perineum medially to above the iliac crest laterally.
3. Indicate on tracing:
   a. plantar surface of the foot
   b. knee joint center
   c. perineum
   d. proximal edge of greater trochanter of affected limb (in 30 degrees of abduction).
4. Measure and record on measurement sheet:
   a. length from plantar surface of the affected foot to the perineum
   b. length from plantar surface to center of knee
   c. circumference at perineum
   d. circumference 4" (10.2cm) distal of the perineum
   e. anatomical anterior-posterior diameter from the ischial tuberosity to the adductor longus tendon
   f. medial-lateral diameter at the perineum and perpendicular to the long axis of the leg.

CASTING

1. Drape patient with appropriate sized cotton stockinette from the waist to the distal thigh on the affected side. Stockinette should be slit medially to fit about perineum and secured with an elastic strap and clamps. If the cast is not extended beyond mid-thigh, it can always be removed without cutting.
2. Position patient standing on the sound leg with pelvis level. The hand on the sound side is used to hold on to a table edge, chair back or some similar object for balance. A box of an appropriate size should be used to rest the foot, and hold the leg in proper position.
3. Use an indelible pencil to mark on the stockinette the following landmarks:
   a. ischial tuberosity
   b. adducter longus tendon
   c. greater trochanter
   d. superior edge of the iliac crest
   e. perineum
   f. mid-thigh
   g. using points a, b, d, and e as a guide, draw a smooth flowing line to indicate proximal trim line of socket. Include the gluteus maximus posteriorly inside the trim line (Figure 7).
4. Wrap the leg from above the proximal trim line to the mid-thigh with the plaster of paris bandage. Before the bandage sets, use the hands to indent it and form the medial brim, Scarpa's triangle, and posterior seat. The ischial seat should be perpendicular to the floor and the medial brim should be parallel with the line of progression.

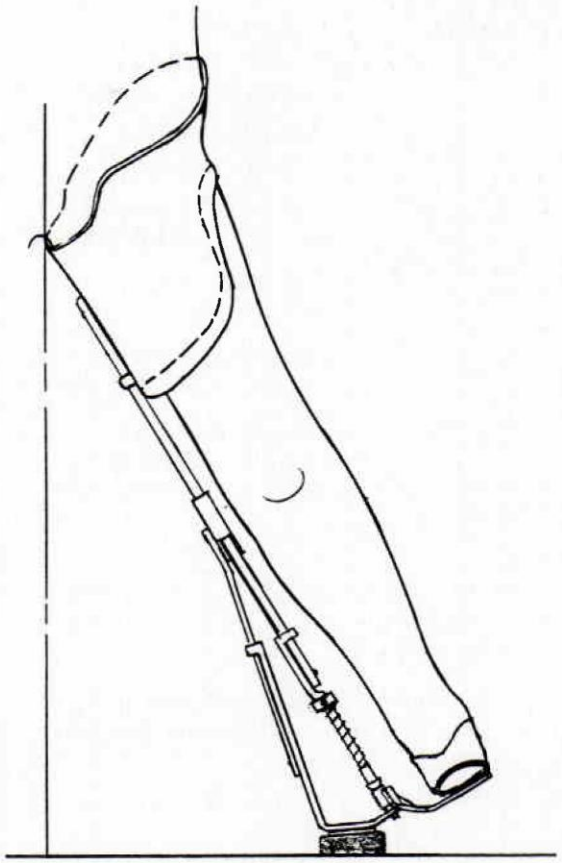

Figure 6: Assembled orthosis showing relationship to patient's leg and vertical and horizontal reference lines.

5. Remove the cast when it has set. The cast is sealed shut distally, the brims are extended proximally to the height of the lateral edge, and the negative model is poured with plaster of paris.

MODIFICATION OF THE POSITIVE MODEL

The positive model is modified in a fashion similar to a quadrilateral socket:

1. The A-P diameter of the brim medially is ¼" (7mm)–½" (13mm) less than the recorded anatomical diameter depending on size of the patient and musculature.
2. The ischial shelf is horizontal with the model in the desired position of 30 degrees abduction.
3. The M-L diameter of the socket is the same as the recorded anatomical diameter, or no greater than the proximal circumference divided by three.
4. The medial brim is parallel to the line of progression and flared. It should be high enough to impinge upon the pubic ramus in less than 20°–25° of hip abduction and ½" (13mm)–¾" (20mm) wide.
5. The proximal circumference is reduced ½" (13mm)–1" (25mm) less than the recorded circumference, depending on patient size and tissue firmness.
6. The distal circumference is reduced to anatomical measurement.
7. The posterior brim is ¾" (20mm) wide at the ischial tuberosity and tapers to nothing laterally.
8. Laterally the height of the brim is at the iliac crest and posteriorly it curves in a smooth flowing line to enclose the inferior portion of the gluteus maximus. Anteriorly over the Scarpas' triangle and laterally of it, the brim flares outward for comfort and is trimmed low enough to prevent impingement of the ASIS. Lateral of this point it rapidly turns proximal to a point on the lateral side just inferior of the iliac crest. The brim in this region exhibits an outwardly radiused edge to prevent pressure from a sharp edge when the leg is abducted.

LAMINATION

1. The appropriate sized trilateral kit is selected from among those available (see Table 1).
2. The modified positive model is prepared for vacuum laminating in the usual fashion.

Catalog Number	Description	Knee Joint Bar Size	Guidelines
3176-XLK-LR 3176-XLK-LL	X-LARGE, LONG, Right X-LARGE, LONG, LEFT	3/16" × 3/4"	Perineum to floor greater than 30 inches
3174-XLK-SR 3174-XLK-SL	X-LARGE, SHORT X-LARGE, SHORT		Perineum to floor 26"–30", Obese
3175-LK-LR 3175-LK-LL	LARGE, LONG, Right LARGE, LONG, Left	5/32" × 3/4"	Perineum to floor 26"–30", Normal size
3173-LK-SR 3173-LK-SL	LARGE, SHORT, Right LARGE, SHORT, Left		Perineum to floor 24"–26", Obese
3172-MK-R 3172-MK-L	MEDIUM, Right MEDIUM, Left	5/32" × 5/8"	Perineum to floor 20"–24", Normal
3171-SK-R 3171-SK-L	SMALL, Right SMALL, Left	1/8" × 5/8"	Perineum to floor less than 20 inches

Table 1.

3. One layer of one-half ounce dacron felt is applied to the model with additional strips about the brim. Two layers of nylon stockinette are applied (three for large sizes).
4. The insert furnished with the trilateral kit (either #3137 or #3138 depending on size) is contoured to fit about the model so that the longitudinal portion of the insert lies along the mid-line of the medial wall of the model and parallel to the long axis. A piece of scrap stainless steel appropriately sized (as long as the insert and the same size as the knee joint upright) is selected, secured in place to the insert with a cap head screw proximally, and the adjustment clips of the insert are formed around it distally. This piece of stainless steel is a laminating dummy and will form a recessed channel in the lamination for the proximal upright.
5. The screw holes of the insert are greased with vaseline and it is wrapped in fiberglass. It is sandwiched between four layers of fiberglass, and gauze is used to securely wrap it in place on the model. The fiberglass should extend just short of the trimlines.
6. An additional two layers of nylon stockinette are applied (three for large sizes).
7. The model is laminated with an 80/20 mix of rigid and flexible polyester resin.
8. Once the lamination has reached its peak exotherm and while it is still pliable, a knife is used to trim the proximal and distal trim lines exposing the distal end of the insert and the laminating dummy. The knife is used to remove the lamination from over the laminating dummy while leaving the adjustment clips of the insert enclosed in plastic. The cap head screw is removed and the laminating dummy is driven out distally using a hammer and pin punch. The socket is removed from the model while still warm.
9. The socket is trimmed as follows:
   a. proximal—described under Modification of Positive Model.
   b. distal—perpendicular to long axis and at level of distal end of insert.
   c. lateral keyhole—proximally, the cutout is as high as the proximal edge of the greater trochanter and as wide in the anterior and post-

erior dimension, distally the cut-out meets the distal edge in smooth flowing curves.

d. The screw holes (8-32 for small 3138, or 10-32 for large 3137) are drilled clear and retapped.

e. All edges are finished and buffed.

LAYOUT

(Figure 6)

1. Draw a line parallel with the longitudinal (parasagittal) axis of the leg and ¼" (7mm) medial of the medial condyle of the knee from the perineum to a point well distal of the foot. This becomes the medial longitudinal reference line and the orthosis will be laid out along it.

2. Transfer the heights of the knee joint center and of the perineum to the drawing measuring proximal from the sole of the foot.

3. Draw a line parallel and ¾" (20mm) distal of the anatomical knee joint center. This will be the orthotic knee joint center. The orthotic knee joint is fit distal to the anatomical so that when the patient kneels, weight will be transferred through the orthosis and not through the leg.

4. Draw a line parallel and 1¾" (45mm) distal of the sole of the foot. This is clearance for the patten bottom.

5. Draw a line perpendicular with the midsagittal line such that it intersects the point defined by the patten bottom line (step 4 above) and the medial longitudinal reference line (step 1). This is the floor line and it, in conjunction with the two lines mentioned in steps 4 and 5, defines the position of the patten bottom.

ASSEMBLY OF THE ORTHOSIS

(Figures 4, 7)

1. The knee joint assembly is laid out along the medial reference line and

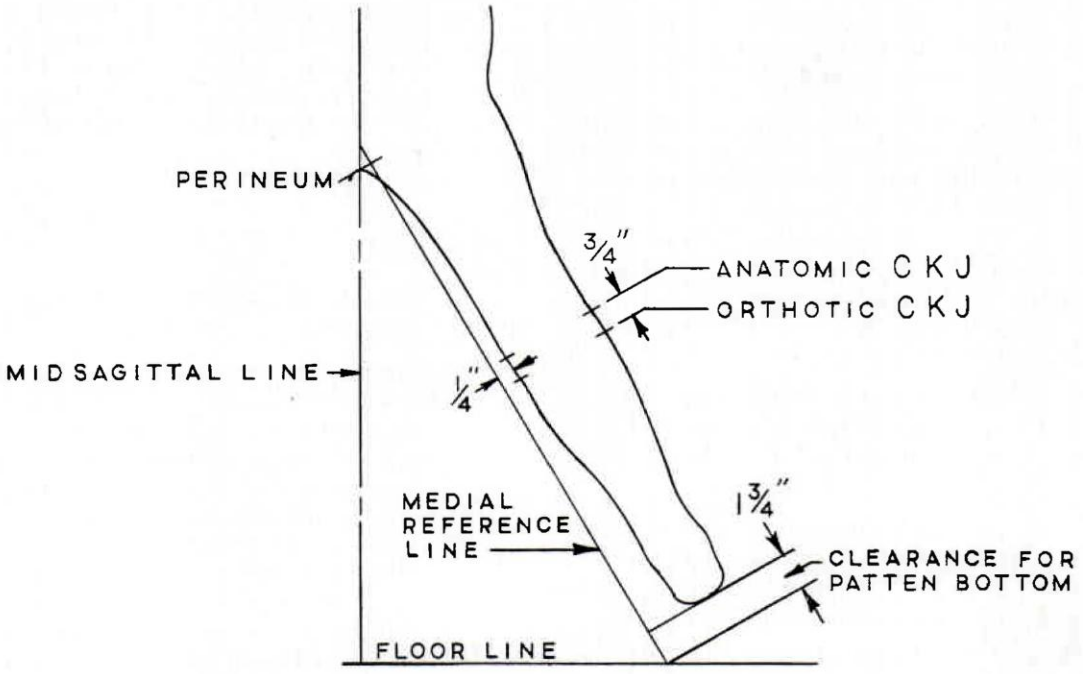

Figure 7: Layout of reference lines on the tracing for construction of the orthosis.

the knee support bar (#3111 or #3112) is contoured proximally so that the distal surface of the rubber bottom (#3118) of the patten bottom (#3114, #3115, #94 3116) lies parallel to, and on, the floor line.

2. Contour the adjustment clip of the patten bottom upright about the support bar and adjust the upright and support bar to the proper length.
3. To remove excess material from the support bar, cut it off ½" (13mm) distal to the distal screw hole of the patten bottom plate upright.
4. Drill and countersink a hole (8-32 body drill) for the attachment screw at the distal screw hole of the patten bottom plate.
5. Assemble the upright guide bar (#3121, #3123, #3125, #3126, #3127, #3129, or #3131) and the patten bottom plate. In a fashion identical to steps 2, 3, & 4 above, the distal upright of the knee joint (#365, #366, #361, #362, #371, #372, #375, or #376) is cut to length and it and the upright guide bar are assembled together.
6. Excess material is removed from the proximal knee joint upright (#400, #402, #403, or #404) so that the medial brim coincides with the perineum. A screw hole (8-32 body drill #19) is drilled and countersunk in the upright for attachment to the proximal hole of the insert in the trilateral socket.
7. The orthosis is disassembled and all cut edges are smoothed and rounded and it is finished in the usual fashion.
8. The heel of the shoe is removed and a "D" ring on a leather chafe is sewn between the welt and the sole in the vicinity of the first metatarsal head.
9. The stirrup (#2001, #2002, or #20040) is riveted to the shoe in the usual fashion. If desired, a layer of leather can be used to cover the area of the stirrup.
10. The lock collar (#3105LG or #3105SM), compression spring (#3107 or #3109), sliding collar (#3108 or #3106) with broad head down and stirrup are assembled on the upright guide bar in that order.
11. The guide bar is assembled on the patten bottom. The hex nut (#885145 or #885160) is tightened to lock the two together so as to prevent motion and wear.
12. The distal knee joint upright, support bar, guide bar, and upright of the patten bottom are assembled with the appropriate screws, lockwashers, and backing nuts.
13. The drop lock is put in place and the trilateral socket is assembled on the proximal upright with the appropriate screw.
14. A leather strap with the buckle on one end is passed through the "D" ring on the shoe, about the patten bottom upright, and is fastened to itself. This strap will internally rotate the leg while the pressure of the socket against the gluteus medius will prevent external rotation of the orthosis.

FITTING

1. The orthosis is donned with a layer of stockinette on the thigh.
2. The patient is checked standing, in 30°–40° of abduction, for a level pelvis and proper ischial weight-bearing. If the patient complains of discomfort in the region of the pubic ramus, and if he is standing in abduction, check to be sure that the ischial tuberosity is on the ischial seat. If the ischial tuberosity is slipping into the socket a loose fit should be suspected and additional socks added. Only as a last resort should the medial brim be modi-

fied—and only then minimally—bearing in mind that the orthosis is intended to be uncomfortable in a position of adduction. The stirrup should be in about the middle portion of the upright guide bar. The toe strap should be adjusted to maintain the leg in internal rotation.

3. With the patient kneeling, check to be sure that weight is borne through the orthosis and not through the leg. If necessary, adjust the upright above and below the knee to provide kneeling weight-bearing through the orthosis and yet maintain the proper overall height.

4. With the patient walking, check for abduction and internal rotation. The stirrup should be checked to make sure it does not "bottom out" (contact the metal at the ends of the guide bar) proximally or distally. The lock collar should be adjusted so that compression of the spring provides suspension of the orthosis without undue pressure in the perineum which will cause discomfort. If necessary, the spring and/or the slide collar may be cut to reduce the pressure.

5. Recheck and tighten all screws. Locktite® may be used on the knee joint screw.

The patient and parents should be instructed in the principles of the orthosis and to expect discomfort—medially, if the orthosis is adducted, and over the ischial tuberosity for the first few days. The patients should be encouraged to attempt internal rotation and abduction during ambulation. They should be instructed in care of the orthosis. The parents should be advised to supervise the patient's walking initially and to prevent such gait deviations as hopping or vaulting from becoming habits that cannot be undone.

Initial shrinkage may be expected during the first few weeks and may well necessitate additional socks to maintain proper fit. Eventually growth and increase in the patient's size should offset this decrease in circumference.

NOTES

[1]Tachdjian, M.A. and Jouett, L.D., "Trilateral Socket Hip Abduction Orthosis For the Treatment of 'Legg-Perthes' Disease," *Orthotics and Prosthetics*, Vol. 22, No. 2, June, 1968, pp. 49–62.

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[4]Glimcher, M.J., Radin E.L., and Amrich, M.M., "The Design of a New Style Ischial Weight-Bearing Brace for Use in the Treatment of 'Legg-Perthes' Disease," *Orthotics and Prosthetics*, Vol. 24, No. 3, pp. 11–20, Sept., 1970.

[5]Birkeland, I.W., Zettl, J.H., "A Hip Abduction Orthosis for 'Legg-Perthes' Disease," *Orthotics and Prosthetics*, Vol. 28, No. 3, pp. 49–55, Sept., 1974.

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Use of the Anterior Floor Reaction Orthosis in Patients with Cerebral Palsy

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INTRODUCTION

The knee joint that is unstable in the sagittal plane can be controlled by using one of three types of orthoses: knee, knee-ankle-foot, or anterior floor reaction orthosis. If the ankle mortise is unaffected, a knee orthosis alone, with a mechanical locking mechanism, can effectively guard against inadvertent knee flexion. There are many designs of knee orthoses that will accomplish this as long as suspension can be achieved and tolerated. This form of ambulation with locked knees lends for a rigid, energy-inefficient gait pattern.

This same gait pattern is demonstrated when the patient with an unstable extremity uses the knee-ankle-foot orthosis and drop lock mechanism. Although the KAFO solves the suspension problem and controls an unstable ankle mortise, it still offers an energy-inefficient gait pattern, interferes with balance mechanisms, and is heavier than the knee orthosis.

In patients with cerebral palsy, labored gait associated with excessive knee flexion in stance phase has presented the orthopaedist and orthotist with significant difficulties in treatment. Crouch gait commences with overactivity of the hamstrings, which increases knee flexion, and thus requires large increases in quadriceps force to stabilize the knee[2] (Figure 1). Calcaneus deformity, a known complication of overlengthened heelcords following tendo-Achilles lengthenings, also significantly adds to further progression of crouch gait. Reasons cited for this complication include overcorrection, failure to protect the heelcords postoperatively, and lengthening of the tendo-Achilles when significant hip and knee-flexion deformities remain uncorrected (Figure 2).

THE SALTIEL ANKLE-FOOT ORTHOSIS

In June 1969, Jimmy Saltiel described an ankle foot orthosis which was designed to stabilize the paralyzed limb without limiting knee movement. The brace extended only as high as the knee joint and was constructed from a reinforced laminate polyester plastic. He stated, "Even a totally paralyzed knee is usually stable in the AP or sagittal direction when in full extension."[1] This principle involves harnessing the floor reaction and directing it to the anterior aspect of the distal shank, thus stabilizing the joint in extension. The brace acts as a first-class lever and the plantarflexed attitude at the ankle utilizes the fulcrum established by the distal trimline of

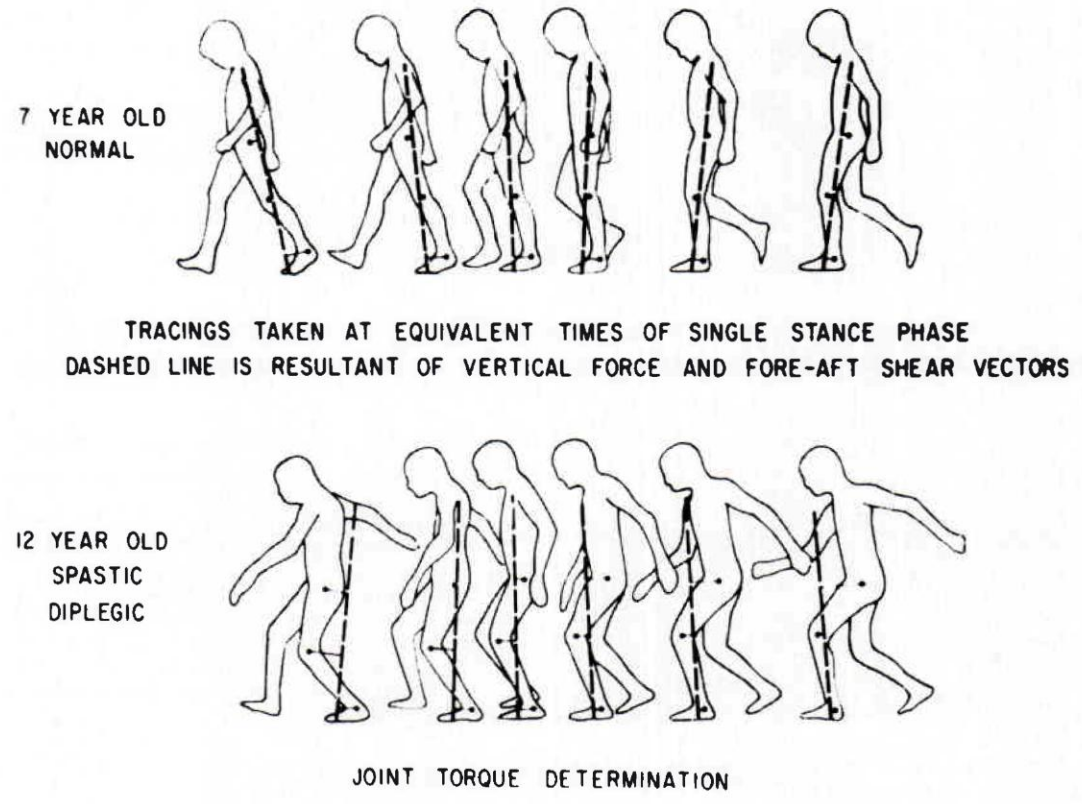


Figure 1: Line of application of floor reaction force. Contrast the line of application of the floor reaction force in the two children. Excessive arm swing and trunk movements of the patient with spastic diplegia cause early aft shear and anterior alignment of the line of application of floor reaction force. Flexion torque of the hip is excessively high in early stance, necessitating straining action of the gluteus maximus. Knee flexion torque is normal initially but progresses to pathologically high levels in late stance, necessitating straining action of the quadriceps femoris. (From Sutherland, D.H. and Cooper L., "The Pathomechanics of Progressive Crouch Gait in Spastic Diplegia," *Orthop Clin. N. Am.*, 9(1):150, 1978. Reproduced with permission.)

the footplate at the metatarsal heads (Figure 3).

The Saltiel ankle foot orthosis employs the plantarflexion-knee extension couple seen in normal gait. At heel strike in the gait cycle, the quadriceps is eccentrically contracting to control knee flexion, thereby allowing for smooth deceleration. This continues until the center of mass passes in front of the knee, at which time there is cessation of firing of the quadriceps. From that point on in the gait cycle, knee extension is caused by the plantarflexion-knee extension couple under the control of the eccentrically contracting triceps surae. This, in addition to the rigid immobilization of the ankle mortise, enables the transference of the floor reaction to the front of the knee resulting in a biomechanic assistance of knee extension.

With increased plantarflexion at the ankle, the moment is increased (Figure 4). With residual knee flexion contractures of ten degrees or more, the floor reaction passes at or behind the knee, thus greatly reducing the knee extension moment to a point where an actual flexion moment is generated once hamstring contractures exceed about fifteen degrees.

Figure 2: A typical crouch gait pattern in cerebral palsy. Note the effect of the overlengthened heel cord resulting in the calcaneous deformity.

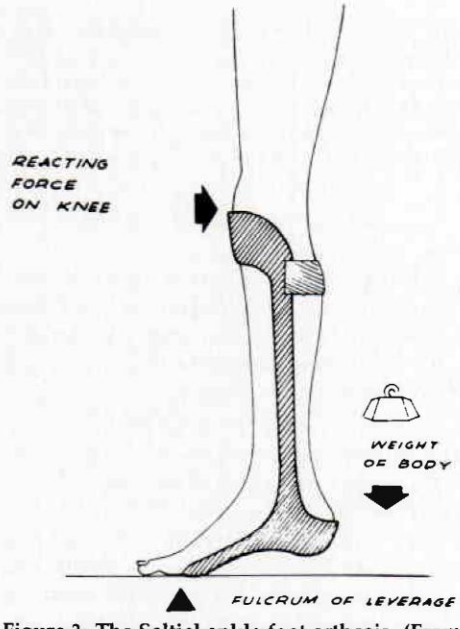


Figure 3: The Saltiel ankle foot orthosis. (From, Saltiel, J., "A one-piece laminated knee locking short leg brace," *Orthot. Pros.*, 23:7, June 1969.)

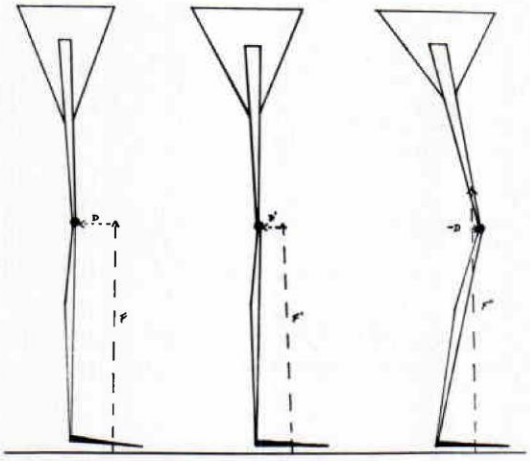


Figure 4: Biomechanics of the anterior floor reaction orthosis.
A. Shows ten degrees of plantarflexion. The more plantarflexion the ankle is set in, the greater the knee extension moment.
B. In neutral position note the reduction of D^1 resulting in the reduction of the extension moment.
C. Shows 15 degrees of dorsiflexion. At 15 degrees of dorsiflexion or greater, the ground reaction force passes behind the knee resulting in the generation of a flexion moment.

The Saltiel type ankle foot orthosis is very effective with the unilaterally involved paralytic limb that readily achieves full knee extension and is stable mediolaterally. This orthosis was originally developed for use with the post-polio lower limb and did, in many cases, preclude the need for a knee ankle foot orthosis. It should be noted, however, that this particular design cannot be applied bilaterally without significant dependence on auxiliary walking aids to improve balance. This is because with bilateral fixed equinus of the orthoses, the center of gravity is forced behind the base of support with resultant loss of balance.[3]

Because of these limitations and because many cerebral palsy patients require bilateral application, we modified the Saltiel design to use with these children. The principal differences in design include a shorter proximal trimline and an angulation at the ankle which we determine for each patient.

The purpose of this paper is to review the use of the anterior floor reaction orthosis in patients with cerebral palsy and, by doing so, to outline the prerequisites and specifications for prescription.

CLINICAL APPLICATION

The charts of 11 patients were reviewed. Eight patients had had clinical examinations only, and three had also been evaluated by computerized gait analysis in the Gait Analysis Laboratory at Newington Children's Hospital. Nine patients had bilateral anterior floor reaction orthoses and two used the orthosis unilaterally. The average age at delivery of the orthosis was 10.5 years (range 3.9 years to 16.2 years). Prior to using the anterior floor reaction orthosis, five of the 11 patients had used bilateral polypropylene knee-ankle-foot orthoses with drop lock knee mechanisms; one had used a double upright conventional KAFO; two had used standard polypropylene solid-ankle AFOs; and one patient had used a standard double-upright conventional AFO. In two patients who had not used orthoses preoperatively, floor reaction orthoses were prescribed immediately following bilateral release of flexion contractures at the hips, knees, and ankles.

Length of use of the anterior floor reaction orthoses ranged from five to 52 months (mean: 17 months). In patients in which the orthoses were discontinued, the main reasons were: outgrowing the orthosis, improved function, and additional surgical procedures.

The degree at which the ankle was set varied. Fourteen of the orthoses were set at five degrees dorsiflexion, five orthoses to five degrees plantarflexion, and one to ten degrees plantarflexion.

Eight of the 11 patients required hamstring releases at or before the start of treatment with the anterior floor reaction orthosis (two of these releases were repeat procedures). The other three patients were cast for knee flexion contractures before they used the orthosis, one of whom required surgery after the serial casting. It should be noted that before any of these 11 cerebral palsy patients could use the orthosis, they required elongation of hamstrings either by surgical release or by serial casting.

Ambulation Status with the Orthosis

Eight of the 11 patients are still using the orthosis effectively and three no longer need orthotic management. Eight patients are community ambulators and three are household ambulators. Four of the community ambulators use Lofstrand crutches; one, a quad-cane; one, Bobath poles; and one requires no walking aids. The community ambulators use the auxiliary walking aids mainly for balance, whereas the household ambulators use the aids to assist with gait progression.

Selection of Foot-Shank Angle

The degree of dorsiflexion or plantarflexion at the ankle of the orthosis is a compromise between the size of the extension moment generated at the knee and the smoothness of forward progression. Theoretically, when the orthosis is set in plantarflexion, the extension moment is maximal but the forward progression is at least partially blocked. If the knee extension moment is too great before the center of mass passes in front of the knee axis, overall forward progression halts. When the orthosis is set in slight dorsiflexion, more time is allowed for the center of mass to proceed in front of the knee axis before the maximal knee-extension moment is exerted by the floor reaction orthosis. While this is desirable, there is a reduction of the extension force generated as well as a delay in the harnessing of the floor reaction during the course of the gait cycle and, hence, more quadriceps strength is required. Because of these factors, it is imperative to set the ankle mortise at the most appropriate angle within the orthosis in order to achieve optimum function. This must be individualized from patient to patient since it depends on a number of factors such as whether the orthosis is bilateral, the quadriceps strength, and the maximum knee extension possible.

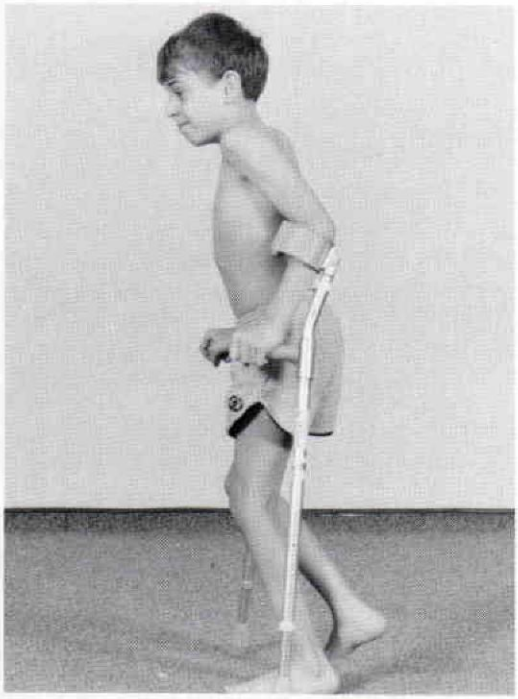


Figure 5: M.C., six weeks postoperatively. Note the dependence on forearm crutches.

Case Report

The effects of the orthosis on gait parameters can best be demonstrated by showing the results of computerized gait analysis in the following patient.

M.C., an 11 year-old male with spastic diplegia, was presented with hip flexion contractures, bilateral knee flexion contractures, and equinus deformities of both feet. At stance he demonstrated severe contractures of hip flexors, hamstrings, and triceps surae. He underwent bilateral psoas and adductor releases, bilateral varus derotation osteotomies, bilateral hamstring lengthenings, and bilateral Baker tendo-Achilles lengthenings.

Six weeks postoperatively, after casting and physical therapy, his gait revealed weakened quadriceps and beginning crouch gait (Figure 5). He was fitted with bilateral floor reaction orthoses which immediately improved his gait.

Computerized gait analysis was performed preoperatively and ten weeks postoperatively with and without the orthoses. Postoperatively, the lateral stick figures

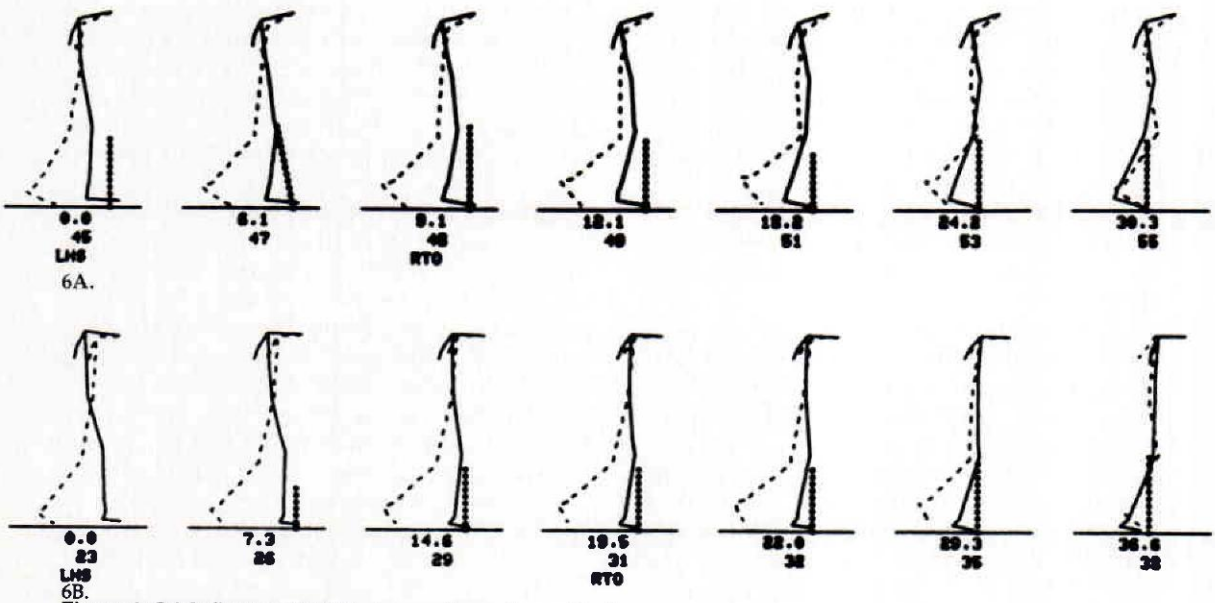


Figure 6: Stick figures of M.C., ten weeks postoperatively.

A. Lateral stick figures from a gait analysis run without anterior floor reaction orthoses. Note where the floor reaction falls respective to the knee axis.

B. Lateral stick figures from a gait analysis run with anterior floor reaction orthoses. Note the improved extension moment generated at the knee as a result of floor reaction passing more anterior to the anatomical knee.

Linear Measurements

	Right Side			Left Side		
		11/11/82			11/11/82	
	Preop.	Postop. Braces	Postop. No Braces	Preop.	Postop. Braces	Postop. No Braces
Single Stance (%)	34.48	31.25	23.08	34.48	34.47	26.83
Step Length (cm)	37.30	39.70	37.80	44.60	48.40	41.20
Walking Velocity (cms/sec)	83.55	78.41	53.89	83.55	78.41	53.89

Estimated External Work of Walking
(joules/kg/meter)

	11/11/82	
Preop.	Postop. Braces	Postop. No Braces
1.118	0.485	0.479

Table 1. Comparison of data from gait analyses of M.C.

demonstrated that the floor reaction force remained anterior to the orthosis and the knee joint axis throughout stance (Figures 6A and 6B).

As shown in Table 1, his linear measurements improved significantly when he used the anterior floor reaction orthoses. Since the single stance time increased when he used the orthoses, he had more limb support and stability (Figure 7). Step lengths also increased significantly with the orthoses, as did his walking velocity. The estimated external work of walking remained approximately the same with and without the orthoses but because of the increased walking velocity when the braces were worn, his gait appeared to be more efficient (Table 1).

In summary, for this patient:

1. Single support time improved with the orthoses.
2. Stride length and walking velocity improved.
3. Energy consumption remained unchanged.
4. Knee flexion was still slightly more than desired and it was postulated at

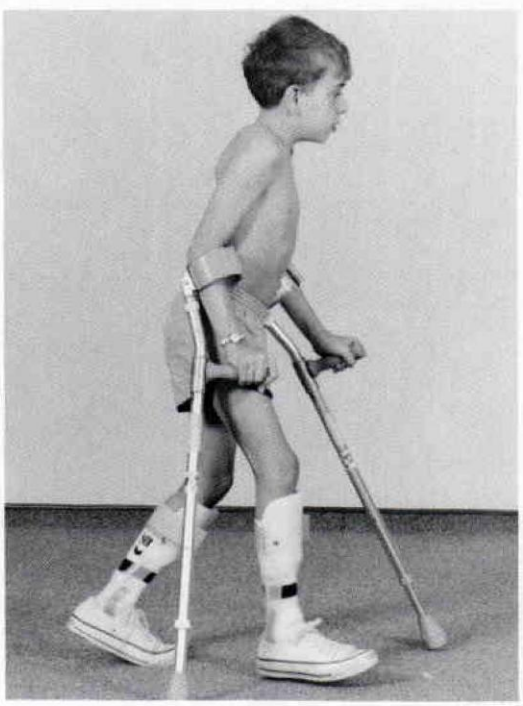


Figure 7: M.C., ten weeks postoperatively, exhibits much less dependence on forearm crutches. With the orthoses, crutches are used only to assist in balance.

this point that this patient might perform better if the orthoses were set in slightly more plantarflexion.

TECHNICAL CONSIDERATIONS

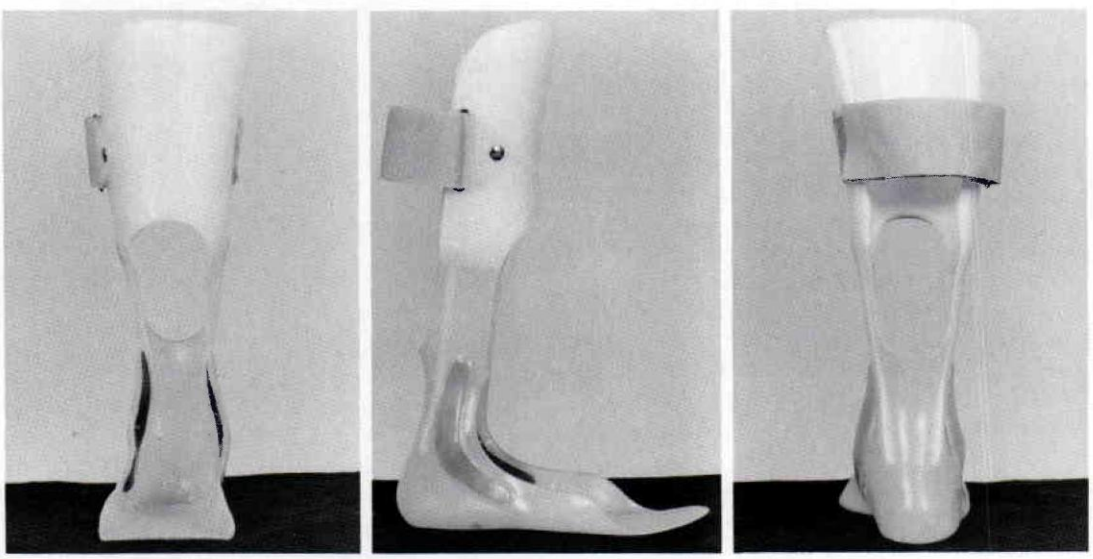
While it is virtually impossible to eliminate the elastic component of the standard orthotic materials, it should be noted that this elasticity must be reduced to a minimum. In order to harness the floor reaction and transfer it to the anterior aspect of the distal shank, a very rigid ankle is imperative. The reinforced lamination of the Saltiel orthosis can be adequately duplicated with the use of $^{3}/_{16}$-inch polypropylene, carbon composite reinforcements, and corrugations strategically placed. These materials offer a less time consuming fabrication procedure without compromising structural integrity or function (Figure 8A-8C).

In pediatric application, special consideration must be given to patellar derangement as a result of absorbing the floor reaction directly on the patella. Because of this, at Newington Children's Hospital we have modified the anterior proximal trimline to end at the distal tubercle, thus not traversing the anatomical knee joint or encompassing the patella. While this trimline does reduce the effective length of the lever arm in the application of the posteriorly directed forces, this compromise is quite acceptable as there is sufficient transference of the floor reaction to achieve knee extension.

Fabrication

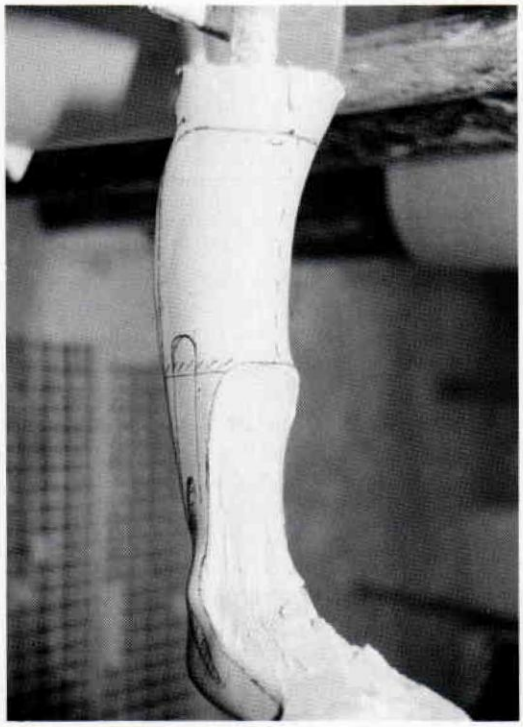
Cast modification of the positive mold follows standard procedures with buildups on all boney prominences, including the crest of the tibia and the tibial tubercle. In addition, the periphery of the trimlines is built up following the line of progression (Figures 9A & 9B).

Once plasterwork is completed, the positive mold is prepared for vacuum

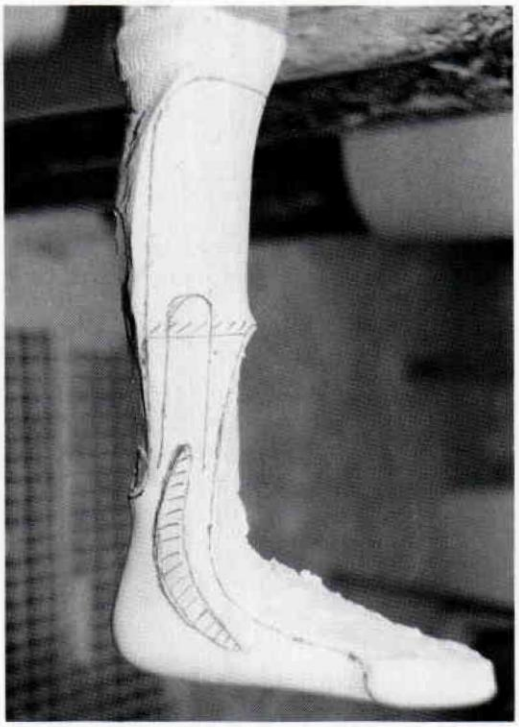


A. Anterior view. B. Lateral view. C. Posterior view.

Figure 8: Finished orthosis.



9A.



9B.

Figure 9: A modified positive model of an anterior floor reaction orthosis.
A. Model is ready for lay-up of carbon composite reinforcements and rope for corrugation.
B. The horizontal line with the diagonal lines immediately above denotes the distal border of the plastazote padding.

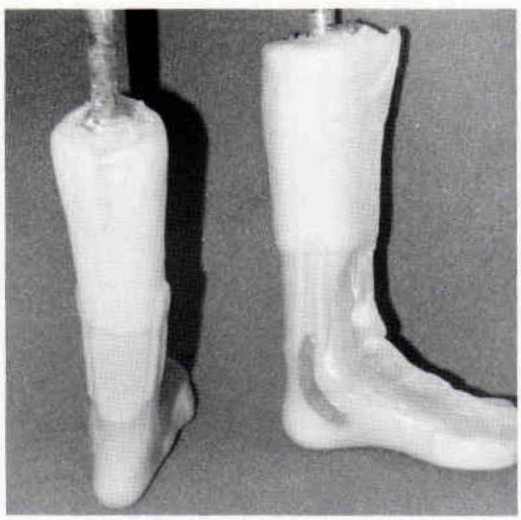


Figure 10: The orthosis molded with the seam kept as straight as possible anteriorly.

molding of polypropylene by securing the carbon composite reinforcements, rope, and Plastazote™ on the cast. Vacuum molding of the polypropylene is done with the heel of the cast facing up, with careful consideration given to obtaining a straight seam anteriorly (Figure 10). This seam enables removal of the orthosis from the cast and must be reinforced afterwards by welding polypropylene rods to the anterior aspect of the orthosis, covering the seam and extending peripherally.

Distal trimline of the footplate extends to the end of the toes. This increases the floor lever arm and enables maintenance of fit for a longer period of time. This modified design has enabled us to achieve optimum clinical results while still taking into account those factors that are unique to treating children.

SUMMARY

It was observed that by using anterior floor reaction orthoses in patients with cerebral palsy, crouch gait was greatly reduced or eliminated. Computerized gait analyses revealed improvement in endurance, ease of ambulation, and increased linear measurements in the three patients who had pre- and postoperative gait analyses. Since the start of treatment with this orthosis, the number of patients with cerebral palsy using the orthosis has increased and results have been similar to those described in the case report. This success is contingent on the prerequisites that need to be met when considering an individual for brace treatment. These prerequisites include:

1. Absence of knee or hip flexion contractures exceeding ten degrees.
2. Careful determination of the angle to which the ankle mortise is set within the orthosis in order to optimize forward progression of the weight line at mid-stance while still maintaining an adequate knee extension moment.
3. Presence of some trunk balance and/or the ability to use auxiliary walking aids in the event of diminished trunk balance.
4. Presence of a minimum of Grade three quadriceps strength.
5. Whether or not the orthosis is unilateral or bilateral. If the weight line is placed behind the base of support, bilateral application makes balance virtually impossible. However, the same posterior placement of the weight line is tolerated if only one extremity is involved.

When these prerequisites and specifications were met, the anterior floor reaction orthosis proved to be more successful than its predecessor, the KAFO, in the treatment of unstable lower extremities in patients with cerebral palsy. The anterior floor reaction orthosis has also been used successfully in patients with other diagnoses such as myelomeningocele, muscle disease, and poliomyelitis. Thus, the patient's clinical picture is more important than the actual diagnosis in determining whether or not to prescribe this particular orthotic design. Following the careful determination of candidacy in the patient with crouch gait, the anterior floor reaction orthosis can provide improved stability and minimize componentry, weight, and bulk.

NOTES

[1]Saltiel, Jimmy, "A One-Piece Laminated Knee Locking Short Leg Brace," *Orthotics and Prosthetics*, 23:68–75, June 1969.

[2]Sutherland, D.H., and Cooper, L., "The Pathomechanics of progressive Crouch Gait in Spastic Diplegia," *Orthop. Clin. N. Am.*, 9(1):143–154, 1978.

[3]Perry, Jacquelin, "Kinesiology of Lower Extremity Bracing," *Clin. Orthop.*, 102:18–31, July–Aug., 1974.

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A Medical-Social Study of Upper Limb Amputees in Hong Kong— A Preliminary Report

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INTRODUCTION

Occupational hand injuries are the most common occupational injuries in Hong Kong.[3] Some of the severe injuries will lead to loss of the hand, frequently the dominant one. The orthopaedic care of a patient with an amputated upper limb does not stop after surgery. The essence of rehabilitation is prevention and integration. The importance of preventive measures in occupational hand injuries has been highlighted in a previous study.[2] The aim of this study was to assess the medical, social and psychological impact on patients with an amputated upper limb. The data collected will hopefully provide useful guidelines on the planning and development of services in the rehabilitation of upper limb amputees and on future research in upper limb prostheses.

MATERIALS AND METHODS

One hundred patients were referred to the South Kwai Chung Prosthetic Clinic for assessment and fitting of prostheses from 1977 to 1982. Twenty of them were selected at random for this pilot study. There were 18 males and two females. The ages varied from 15 to 57, and the marital status included seven singles, eleven married, and two widowed. The educational levels were below primary in eight, primary in ten, and secondary in two.

Eighty-five percent of the injuries were occupational. All of the victims were right-handed and 70 percent of the injuries involved the dominant hand. The levels of amputation were broken down into 75 percent below-elbow, 15 percent wrist disarticulation, and ten percent above-elbow. The ages for the first prosthetic fitting ranged from 3½ to 11 years. Clinical and social assessments were made at a joint interview by a surgeon and a prosthetist. The following information was obtained:

- The type of prosthesis with respect to its fitting, cost, training, condition under use, and maintenance
- Function of the prosthesis
- Cosmesis of the prosthesis
- The amputee's Activities of Daily Living (A.D.L.) and employment
- Changes in their family conditions
- Social welfare assistance given to the amputees
- Psychological assessment of the amputees
- Special comments from those interviewed on the different members of

the rehabilitation team, i.e., the surgeon, therapists, prosthetists, and medical social workers.

RESULTS

The Prosthesis

All patients were fitted wth the body-powered prosthesis, voluntary opening hook, and an optional cosmetic hand with a built-in prehensile grip mechanism (Figure 1).

More than 90 percent of the fittings were done within three months of the amputation and the patients usually received a two to three month period of training supervised by the prosthetists. The follow-up had been generally regular in the first two years. After that, patients only had appointments for repairs or other specific problems.

The frequencies of breakdowns were related to the duration of usage. The frequent users had an average of two to three minor repairs a year. Usually the repair work could be accomplished within one week. Some of the more innovative patients invented their own minor modifications on the prostheses and handled minor repairs by themselves.

The cost of the prostheses varied from HK$1,000 to HK$2,000. The patient usually paid the fee out of the lump sum of workman's compensation granted, but 20 percent of them received financial assistance from the Social Welfare Department.

Function of the Prosthesis

Sixty percent of the patients used the prosthesis for over ten hours a day, and the majority of them (80 percent) used it at home (Figure 2). Only 60 percent of them regularly used the prosthesis at work, and these amputees were primarily heavy metal workers who utilized the hook in pinning down an object. Nearly all patients used the cosmetic hand on social occasions. Relatively few (15 percent) regularly engaged their prostheses in sports or recreational activities.

The overall comment given by the amputees on the functional achievement of the prosthesis was good in 70 percent, fair in 20 percent, and poor in ten percent.

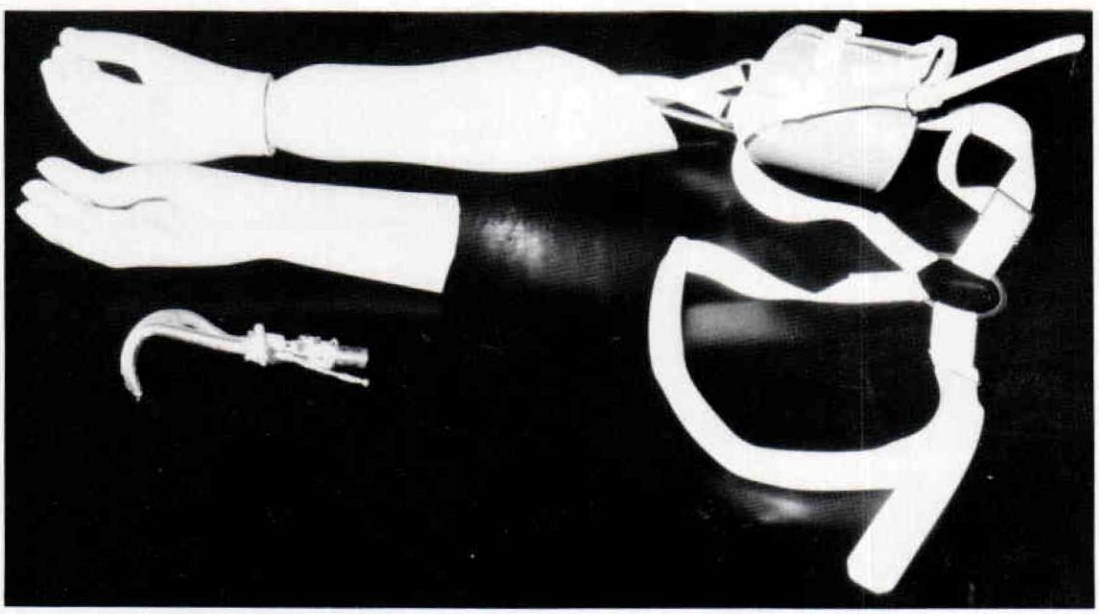


Figure 1: All upper limb amputees were fitted with a body powered prosthesis with a hook terminal device and functional cosmetic hand.

2A.

2B.

2C.

Figure 2: Functional achievement
A 45 year old seamstress had a below-elbow amputation of her dominant right hand. She was fitted with a prosthesis two months later. She demonstrated a high degree of dexterity and skill in the use of the hook terminal device (A), (B), (C). She continued in her gainful employment, and managed most of the housework.

A.D.L. and Employment

Table 1 illustrates the pattern of A.D.L. shared by amputees while using the prosthesis. Sixty to 70 percent of them were satisfied with the different modalities of activities, while the unsatisfactory group included the non-users and infrequent users.

It was interesting to note that the two non-users were both right hand dominants and their left hands had been amputated. They seemed to have developed a single-handed pattern of activities and rejected the alternative of using the prosthesis.

The majority of the right hand dominants with right hand amputations successfully shifted the dominance to the normal left hand and used the prosthesis on the right hand as an assistive device.

Only 15 percent of the patients retained their previous job after injury. Eighty percent found new jobs which adapted much better to the capability of their now much impaired upper limb function. Five percent remained unemployed.

As a result of the injuries, all the patients suffered a significant financial loss. The amount granted from the Workman's Compensation Board was usually just adequate for subsistence during the period in the hospital and during rehabilitation. The personal income per month dropped by 25 percent.

Responses Regarding Use of the Prosthesis During Activities of Daily Living

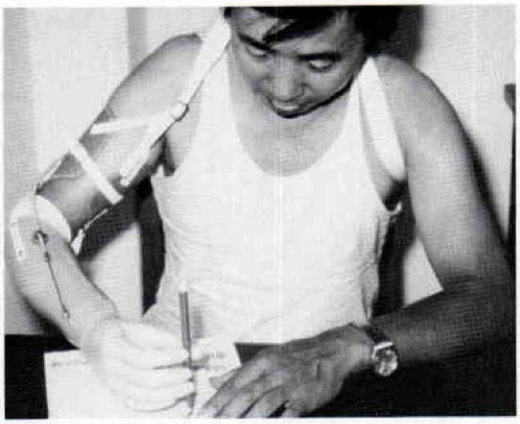
Activities in Daily Living	Satis-factory	Unsatis-factory
Feeding/ cooking	14/20 (70%)	6/20 (30%)
Bathing/ cleaning	15/20 (75%)	5/20 (25%)
Dressing	15/20 (75%)	5/20 (25%)
Toileting	12/20 (60%)	8/20 (40%)

Table 1.

Changes in Family Conditions

Among the 20 patients, 12 were the chief provider for the family. The injury and

3A.



3B.

Figure 3: Activities of daily living
A 40 year old clerk had a below-elbow amputation of his dominant right hand. He used a functional cosmetic hand well in most of the activities of daily living (A). He also attempts to write with the prosthesis (B).

subsequent loss of previous earning capacity led to serious financial problems for six of them. The wives and older children were forced to seek part-time jobs in order to support the family. However, no disruptions of marital ties were detected amongst the married patients.

Social Welfare Assistance

Table 2 illustrates the varieties of social welfare assistance available to the amputees.

Varieties of Assistance Available to the Subject Patients

	No. of Patients
Counselling	20/20 (100%)
Disability allowance	10/20 (50%)
Public assistance	11/20 (55%)
Employment	3/20 (15%)
Housing accommodation	1/20 (5%)

Table 2.

Cosmesis of the Prosthesis

The functional cosmetic hand was generally well accepted by the patients, their families, and friends. The hooks were regarded as "strange-looking" by over 60 percent of the patients. All of them preferred to hide the prosthesis in their trouser pockets.

Psychological Assessment

This was the most difficult part of the study. Most patients were reluctant to disclose their genuine feelings regarding the injury they received and the ensuing consequences. The majority of them did reveal a period of grief, frustration, and uncertainty that lasted about six to 12 months following the injury. The feeling of helplessness gradually left when they found that the prosthesis was actually functional and succeeded in assisting them through different activities. No patient received a proper referral to see a clinical psychologist. Any encouragement and counselling were given mainly by the surgeon, the prosthetist, and the medical social worker.

Amputees' Comments on the Members of the Rehabilitation Team

Doctors

Twenty-five percent of those interviewed complained that they were not adequately informed of the consequences of the amputation and the prospects of using the prosthesis. Sixty-five percent of them suggested that they would like the follow-up to be conducted by the doctors

and paramedical staff together, because some of the administrative procedures required the authorization of the medical staff and sometimes they had problems with the residual limb. Ten percent complained that the doctors were not helpful in the course of rehabilitation.

Prosthetists

Eighty-five percent said that the maintenance service had been satisfactory. Twenty-five percent complained that their prosthesis took too long to be fitted (longer than three months). But the majority of the patients were satisfied with the technical supervision given at the initial period of prosthesis training.

Therapists

Only ten percent of the patients were referred for training to occupational therapists and physiotherapists. They were not sure whether the assistance given was of much help.

Medical Social Workers

The majority of patients relied heavily on the assistance of the medical social workers. Thirty percent of them would have preferred more frequent counsellings, but were aware of the limited time available for these visits.

DISCUSSION

From this preliminary study, it can be seen that an upper limb amputee patient is confronted with various problems, including medical, psychological, social, and financial concerns. The upper limb amputee group is relatively neglected in the overall consideration of rehabilitation for amputees.[1] If the ultimate aim of rehabilitation, the true integration of the disabled into the community, is to be realized, there should be a well-coordinated system of rehabilitation care. The function of the team approach should be based on a special amputee clinic and registry headed by the surgical staff. The expertise of the participating members is then called upon under appropriate circumstances.

Of outstanding significance in this preliminary report is the need all patients have of increased financial and psychological support in the early phases of rehabilitation.

The help of the medical social worker, and if possible, a clinical psychologist, will be most appreciated in future rehabilitation efforts. It should be emphasized that only 15 percent of the patients retained their old jobs, and none of them acquired a better personal income following rehabilitation.

The training in use of the prosthesis should be a joint effort between the prosthetist, the physiotherapist, and occupational therapist. This retrospective review does demonstrate the lack of cooperation and coordination among the paramedical staff concerning this area. It must be recognized that if the patient fails to acquire the habit and skill of using the prosthesis in the first six to 12 months, the chance of a regular usage will be very remote. The initial training period is doubtlessly the most important phase of the entire program, and the medical staff should supervise very closely during this period.

Age does not seem to have a significant influence on the usage of the upper limb prosthesis. However, none of the patients are over 60 years of age. The majority of the patients are within the active income earning years, and are eager to obtain the maximum benefits from the prostheses.

The body powered hooks and functional cosmetic hands are quite well accepted by the majority of patients, although the prostheses are mainly used as assistive tools to the normal hand. The rehabilitation of a non-dominant hand to take over the job of its amputated counterpart is understandably faster than the training for the utilization of a prosthesis. However, after about one year's training, most patients adapt comfortably to a revised two handed pattern of activity.

The possibility of fitting a myoelectric prosthesis had been inquired by some patients who were aware of the recent developments in prosthesis manufacturing.

None of them however, was aware of the cost of such prostheses, nor were they informed about the complexity related to the maintenance requirements. The body powered prosthesis is reasonably priced, and easily maintained at the prosthetic workshop. It seems obvious that the expensive program of myoelectric prostheses will take many years to come to maturity. The optimism of some early workers in myoelectrics[4] does not seem to be present in Hong Kong.

Although the functional cosmetic hand is far from meeting all the aesthetic requirements, all patients preferred having it fit for special social occasions. Since the interchange of the hook terminal devices and cosmetic hands is simple, such combination does encourage an almost continuous use of the prosthesis throughout the day.

One of the non-users in our series lost his hand at the age of five, but the prosthesis was fitted 11 years later at the age of 16. Obviously, he totally missed the chance of childhood rehabilitation and reeducation. Timely fitting of prostheses for children below the age of 12 should certainly be considered and special prostheses should be developed accordingly.

Having completed this preliminary study, we are looking forward to a large scale continuation study, which is expected to cover a target population of 400 amputees, in the near future.

NOTES

[1]Chadderton, H.C., "Consumer Concerns in Prosthetics," *Prosthetics and Orthotics International*, 1983, 7, pp. 15–16.

[2]Chan, K.M., Cheng, Y.H., Chan, W.T., Leung, P.C., "The Rehabilitation of Occupational Hand Injuries in Hong Kong," presented at the 1980 World Congress of Rehabilitation International, Winnipeg, Canada, June, 1980.

[3]Leung, P.C., Chan, W.T., Occupational Hand Injuries—A Medicosocial Research, 1979–1980, O.H.I. Research Report No. I—Profile Study, March, 1981.

[4]Waring, W. & Antonelli, D.J., "Myoelectric Control Systems," *Orthopedic & Prosthetic Appliance Journal*, March, 1967, pp. 27–30.

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A Case Study: Use of a Terminal Device to Augment a Paralyzed Hand

Judd A. Katz
David Perkins

ABSTRACT

A split-hook terminal device was used to allow writing skills for a C-6 level quadriplegic. In this case, the subject has his right upper limb, but could not use his natural hand for printing. Any use of the hand, even with the aid of a support device, would cause spasms of the hand, which would not allow him to write. The voluntary opening style hook is attached to an arm support, parallel to the right arm, with the hook being placed next to the palmar side of the hand. A prehensile support for holding is in the hook, and not in the spastic hand. The orthosis is cable driven, and utilizes all traditional hardware except for the arm socket, which is replaced with a pivoting arm support. The pivoting arm support will allow the hook to turn out of the way when the subject desires to operate his wheelchair with his natural hand.

INTRODUCTION

The subject of this project is a 40-year old male who is classified as a partial quadriplegic. In 1965, the subject suffered through an automobile accident that crushed the sixth cervical vertebrae and caused his quadriplegic condition. Presently, he is an undergraduate student at Auburn University at Montgomery, Alabama, and the skill of independent writing, especially class notes and examinations, would greatly facilitate learning. Prior to the orthosis, the subject would attempt to print through the use of a pencil support device attached to his hand, but printing in this manner elicited spasms in the hand. After three to five minutes of writing, his right hand would become fatigued, and begin an uncontrolled spasm until he stopped the task. Typing would generate a milder spasm than that elicited when printing, but it proved to be extremely difficult to take a typewriter to class for the purpose of notetaking.

In addition, the use of a typewriter for Algebra, Statistics, and Physical Sciences, courses that required mathematical and scientific notation, could not be handled by a standard typewriter. The problem facing the subject's academic advisor, Dr. Katz, was to develop an orthosis that would allow the subject to print alphabetical letters, as well as scientific notation at home or in the classroom without inducing debilitating hand spasms. The orthotic device described in this article was constructed and fitted by Mr. Perkins. All materials used were readily available from a prosthetic/orthotic facility.

DESIGN DESCRIPTION

The orthosis can be viewed as having four main parts: 1) the hook and wrist assembly; 2) the bracket and rotating rod assembly; 3) the arm attachment cuff assembly; and 4) the cable harness assembly (Figure 1).

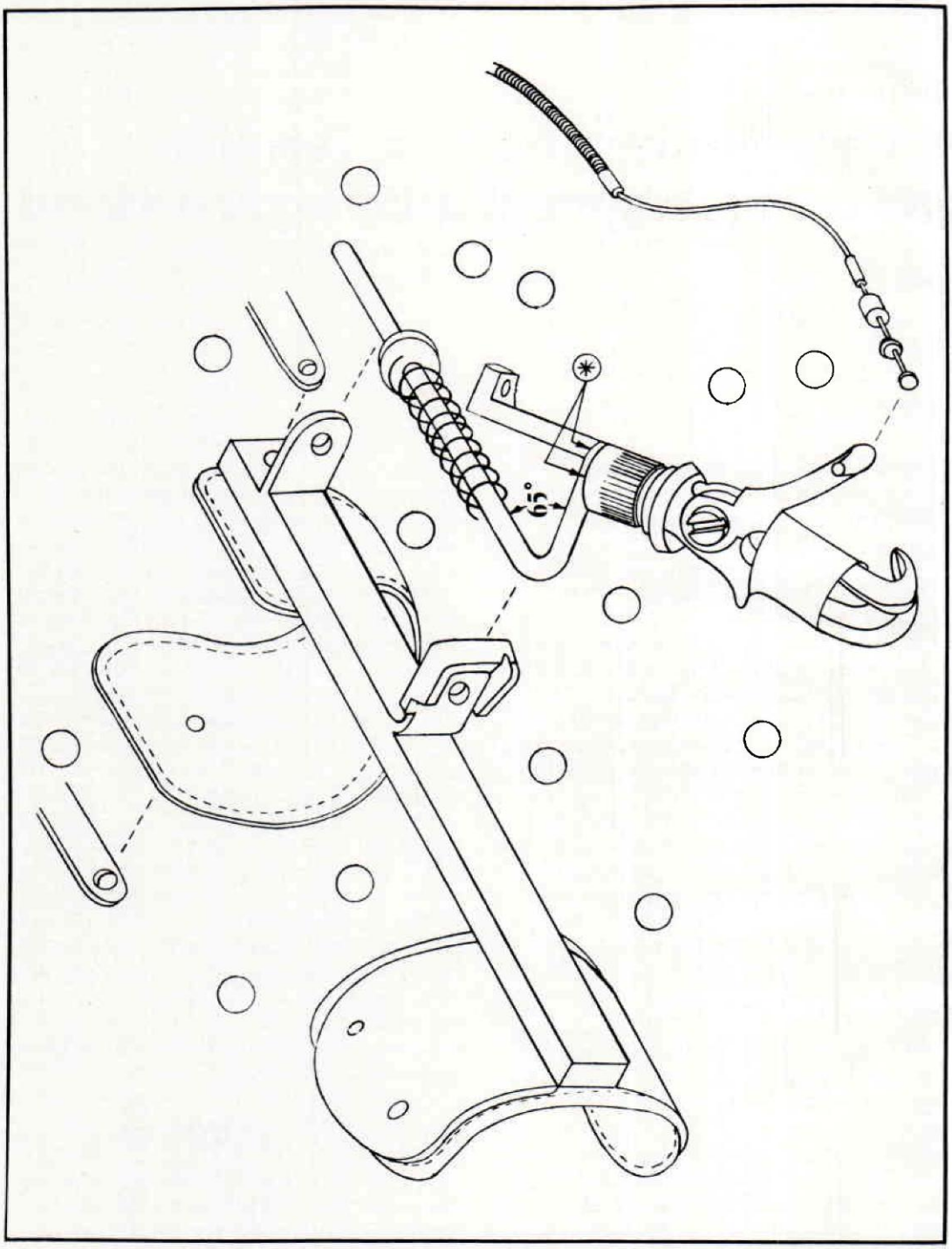

Figure 1: Exploded isometric drawing of the orthosis emphasizing the hand, rotating rod, and arm attachment assembly.

The Hook and Wrist Assembly

- A Hosmer Dorrance #99X hook was used. The smaller sized appliance was more appropriate since space was crucial when the hook was folded out of the way for propelling the chair. Three hook-tension bands (approximately 7.5 pounds prehension) were used to hold a pencil and comfortably open the appliance via the shoulder harness.
- An FM 100 wrist unit attached the hook to the rotating support. The wrist unit was welded to an angled end tab of the support rod (See Figure 1 for the site of the weld).
- A one-inch rubber lock washer was inserted between the terminal appliance and wrist to suppress hook rotation while writing.
- The cable was attached to the thumb of the hook in the traditional fashion.

The Rotating Support Rod

- A 6" × ¼" stainless steel rod was used to support the hook and give it its swivel action. The forward two inch section of the rod was bent back 65 degrees and angled down from the center line of the arm; this position aimed the tip of the hook away from the natural hand so that the hand would not interfere with the activity of the hook.
- The rod was housed inside a 3⅜" × ⅝" × ⅛" steel bracket and sheathed with a 3½" spring.
- The distal end of the bracket terminated in a two position gate allowing the rod to lock in a writing position (Figures 2 and 3), or rotate and lock behind the natural hand in a clearance position (Figure 4).
- A 3½" spring provided pressure for locking in both positions. The spring was held in place by a pin and washer

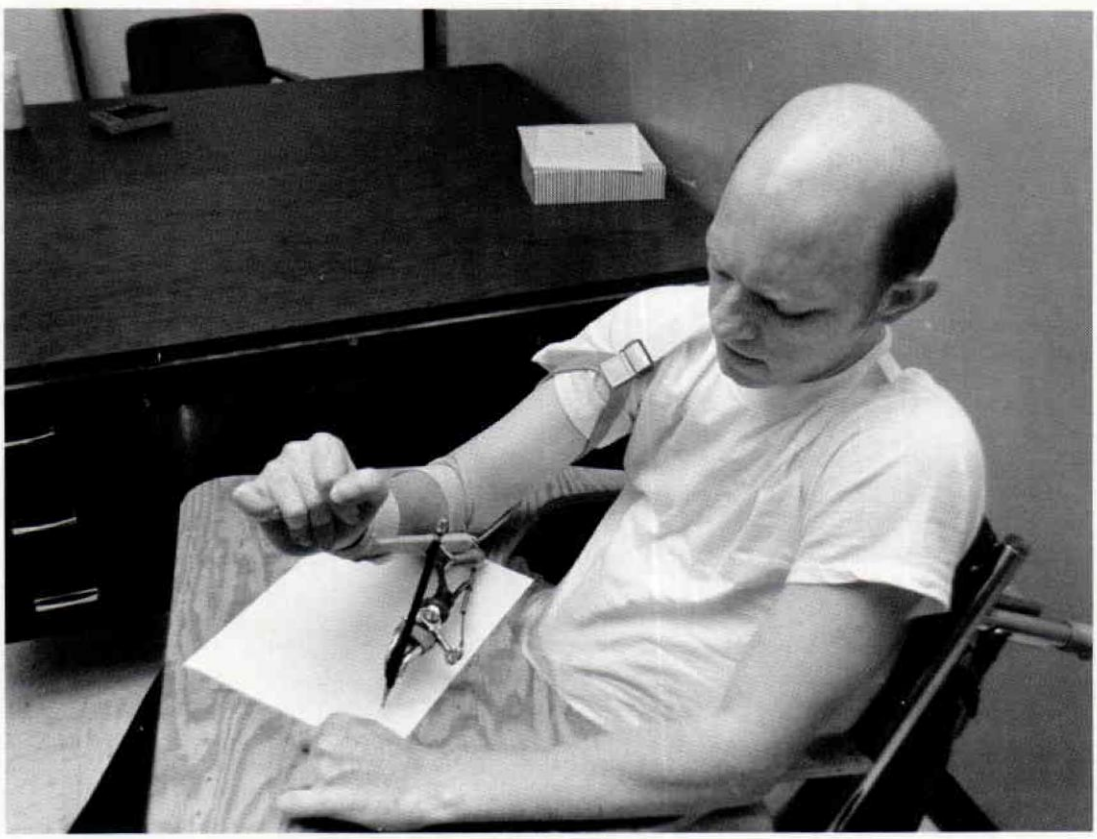

FIgure 2: The subject shown using the hook in the writing position.

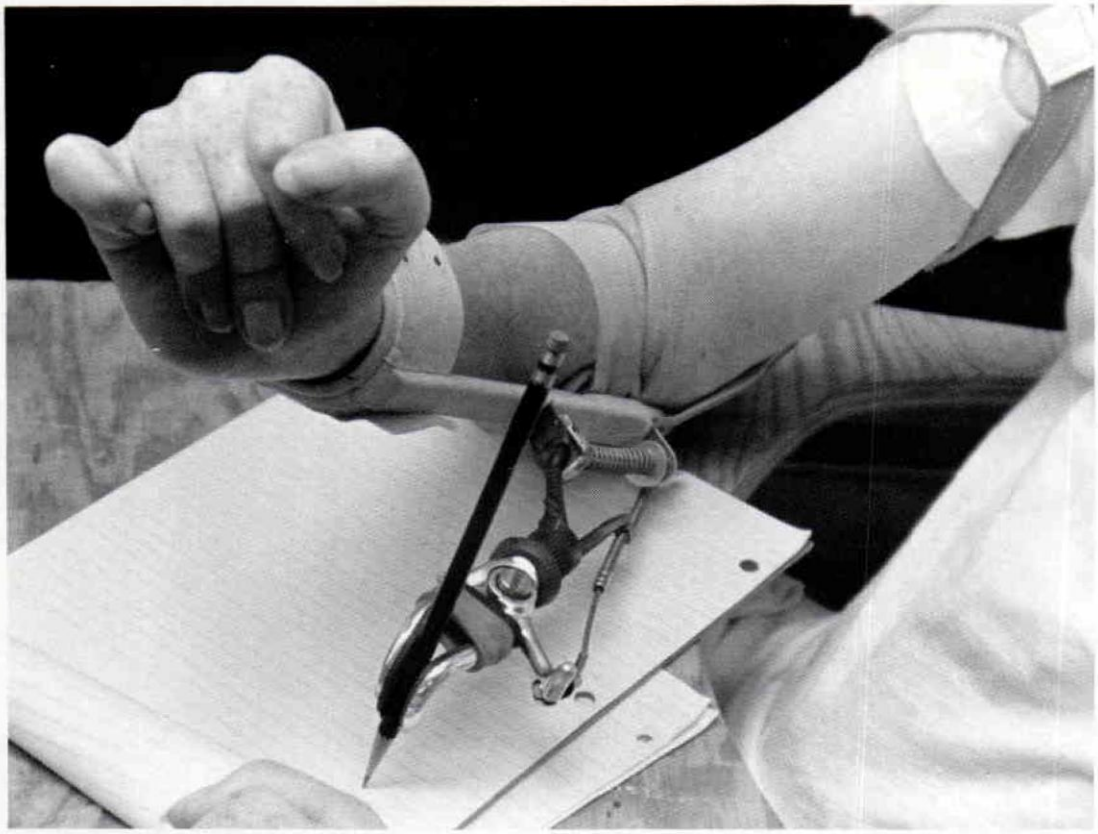

Figure 3: Closeup of hook in writing position. Bracket and rotating rod assembly can be seen.

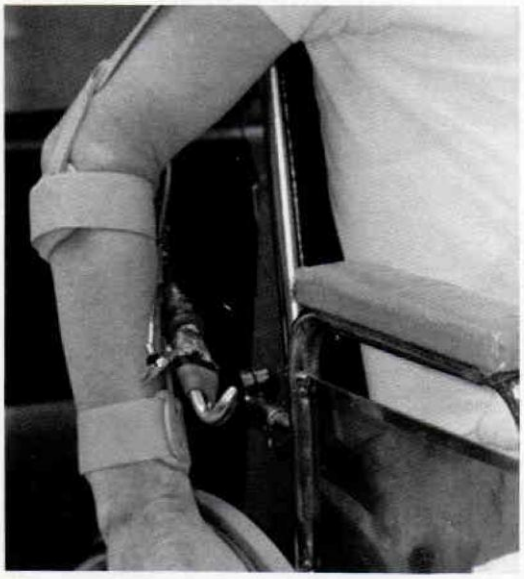

Figure 4: The hook in the clearance position, so that the subject can operate his wheelchair.

mounted at the proximal end of the rod.
- The cable guide was 1¾″ × ¾″ stainless steel rod welded to the back of the primary support rod.

The Arm Attachment Cuff Assembly

- An 8¾″ × ½″ × ¼″ bar was used as the main frame to support the rotating rod and bracket assembly.
- Two cuffs were riveted to the side opposite the hook and bracket assembly. The wrist cuff was 5″ × 2″ and the upper forearm cuff was 6½″ × 2″. Velcro® band material was used to secure the cuffs to the client's arm. The bar and cuffs were covered with gray leather which gives the prosthesis a comfortable fit and aesthetic appearance.

The Cable Harness Assembly

- A humeral cuff was attached to the upper forearm cuff by two straps. The cable was connected at the thumb of the hook with the cable housing run under the arm and connected to the outer side of the half cuff. A shoulder harness was connected to the upper arm half cuff by an inverted "Y" strap. The cable was attached to the left arm loop of the harness with a hanger connector. This is the same system used by amputees to operate the opening of a terminal device.

USING THE ORTHOSIS

With assistance from an attendant, the orthosis can be affixed to the subject prior to class. He has two wheelchairs—a motorized chair with a hand-operated four-way switch, and a standard nonmotorized chair that he can propel.

By grasping the rotating knob projection hand rims that extend from the wheels, he can exercise his arms and move his wheelchair about on flat surfaces. When the hook is not used for academic tasks, he can move it out of the writing position and into the clearance position by pulling forward on the hook with his left hand, which releases tension on the gate, and rolling the hook over the back of his right hand and locking it out of the way. He can propel his chair from class to class without removing the hook.

When class begins, he can grab the hook with his left hand, pull forward and roll the hook out of the clearance position and back into the writing position. When he flexes his back, the hook opens and he inserts a pencil into the appliance. Relaxing his shoulders, the hook grasps the pencil and he is ready to print letters or mathematical symbols. The downward angle of the hook allows him two points of contact on his writing board: the hook-held pencil and his right elbow. The arm is supported in a comfortable position which does not elicit spasms in the arm or hand.

SUMMARY

A voluntary opening hook terminal appliance was used in a nontraditional fashion to permit a partial quadriplegic patient to print. The subject has both of his natural arms and hands, yet has only minimal use of his hands for prehensile tasks. Hand spasms greatly limit the use of support devices that could augment the use of his hand. The hook was mounted on a rotating and locking bracket so that it could be moved out of the way without removing the appliance when he operated his wheelchair. This feature allowed the device to be placed on and off only once per day.

The subject is presently involved in training, which is sharpening his printing skills of both numbers and alphabet letters. Soon, he will be able to take lecture notes in class, write mathematical formulas, and take multiple-choice examinations by himself.

ACKNOWLEDGMENTS

The authors are very grateful to the Alabama Rehabilitation Center for their approval and funding of the orthosis, and to Mr. Joe Albree, Assistant Professor of Mathematics, Auburn University at Montgomery, for his technical drawing.

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The Use of Prophylactic Knee Orthoses at Iowa State University

Frank Randall, A.T.C., M.A.
Harold Miller, C.P.O.
Donald Shurr, L.P.T., M.A.

INTRODUCTION

Major knee injuries are a comon concern to all people involved with the sport of football. At the college and interscholastic level, minimizing the numbers of knee injuries would allow the game to be played at its optimal level of skill and safety. This would save everyone involved with the sport much time, effort, and money. A successful program incorporating prophylactic knee orthoses could allow such injuries to occur less frequently.

PURPOSE

During the 1982 fall football season, Iowa State University initiated a protective knee orthosis program. This program was initiated with the cooperation of the coaches, orthopaedic surgeons, and trainers, together with the developers of the Iowa knee orthosis, which was previously used for athletes with collateral ligament instability. The target population consisted of offensive and defensive linemen, linebackers, and tight ends, because of their high incidence of knee injury (See Table 1). Some of the athletes had a documented history of knee instability. The purpose of the program was to see what effect prophylactic knee orthoses had on the prevention of serious knee injuries in a major college football program, and to evaluate the problems associated with the implementation of such a device.

METHOD

Several commercially available knee orthoses had been used previously with players suffering from knee injuries, but none had been used in a prophylactic sense. That is, orthoses used previously were for athletes who already had knee injuries, in the hope of preventing further injury. This evaluation was also aimed at the normal knees which had no history of injury.

The device chosen for evaluation was an all plastic knee orthosis with polycentric hinges bilaterally and proximal, and distal cuffs made of a combination of polypropylene and polyethelene, prefabricated to models in four sizes (See Figure 1). Spray adhesive and tape were used to suspend the orthosis. The advantages of this prophylactic device were:

1. Bilateral support—medial and lateral uprights;
2. Polycentric joint construction providing a changing center of rotation;
3. Maximum length for maximum support (length of lever over thigh and leg);

Major Knee Injuries (Grades 2 and 3)*
by Football Position
1979–1983

	Quarter-Back	Running Back	Offensive Lineman	Defensive Lineman	Defensive Back	Line-Backer	Tight End	Wide Receiver	Total
1979 Spring			1	1				1	3
1979 Fall		1	1	1	1			1	5
1980 Spring						1			1
1980 Fall			1	5	1	1	1		9
1981 Spring			1	1	1	1			4
1981 Fall		3	4	1	2	1		2	13
1982 Spring			3	4	1		1		9
1982 Fall			2	3	3	1			9
1983 Spring						1			1
1983 Fall									

*Grade 2: Requires splinting or other immobilization.
Grade 3: Requires surgery.

Table 1.

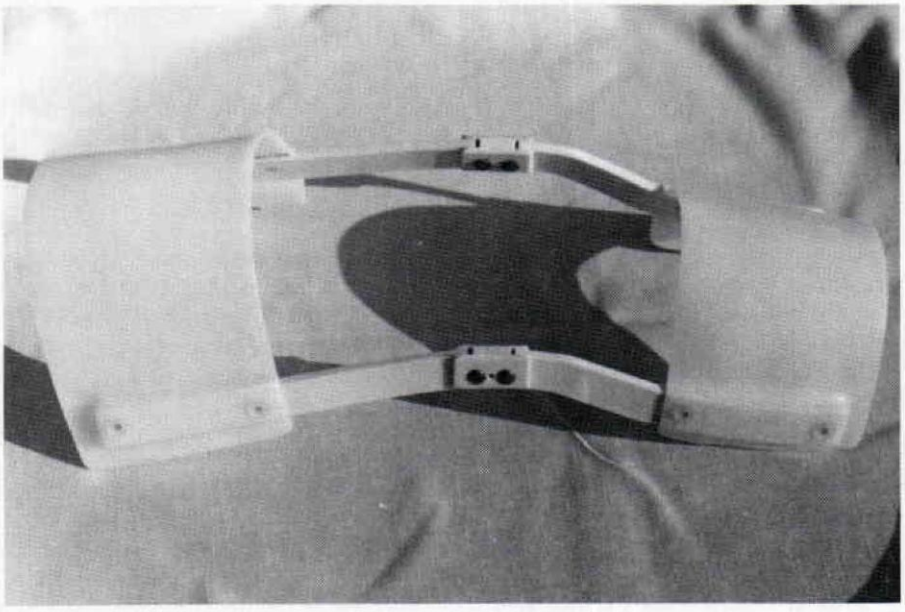

Figure 1: The prophylactic knee orthosis.

4. Semi-flexible (copolymer) cuffs on thigh and calf to give snug fit and comfort;
5. Lightweight joints (7 oz. per pair; overall 13.1 oz.);
6. Adjustable hinges to control range of motion (hyperextension to partial flexion).

The original device was fabricated in a manner consistent with that of the Iowa knee orthosis. After pouring a positive model, the orthosis was fabricated using the nylon polycentric joint. The proximal and distal cuffs were made of a fiberglass tape wrap. The orthoses were originally secured by using an elastic wrap around the thigh and calf cuff with no adhesive. Problems associated with this method and design were:

1. Irritation of skin by fiberglass;
2. Suspension;
3. The cuffs were undersized and added to the irritation of the skin, since the athletes were in a leaning position when casted, allowing musculature to sag.

Following the evaluation of the initial program, orthosis and materials, changes were made in response to problems. These changes included:

1. During casting the athlete stood upright with legs slightly flexed to get true muscle conformation of the thigh and calf;
2. All cuffs were cut to a minimum size and the material was changed from fiberglass wrap to copolymer;
3. Medical adhesive was utilized on the underside of the cuff for a tacky effect;
4. A tape anchor (tape rolled with adhesive mass out) was applied directly to the leg;
5. A closed spiral wrap was applied directly to the leg for suspension;
6. Three or four inch elastic tape was applied over the cuffs in a manner that would overlap the cuff and the underlying wrap to finally hold the orthosis in place.

RESULTS

Thirty-one athletes, taken from the first two teams, wore 62 orthoses. The 31 players, divided by position, included two quarterbacks, one linebacker, 11 offensive linemen, seven defensive linemen, two defensive backs, and two tight ends. Ten of 31, or 33 percent, had histories of previous knee injuries. During the spring of 1983, our records indicated only one significant knee injury occurred during 20 days of spring football. This may be compared to the nine injuries suffered the spring of the year before.

When considering any knee orthosis for use during game conditions, players and coaches question its effects on speed and agility. To help answer this question, some athletes were tested in a timed agility test around three cones with and without the orthosis. The results are listed in Table 2. Of those athletes tested, times revealed no change to .2 second slower with the orthoses on. The athletes who had previously injured knees had faster times with the brace on. This finding may indicate more confidence when cutting with the brace on than without. No data can be gathered from the fall 1982 trial, since the coaches were pessimistic toward the project, and less than six athletes finished the season wearing the devices. Only one of these six sustained a serious injury in the fall of 1982, during game conditions when he chose not to wear his orthosis.

DISCUSSION

The current coaching staff is very positive concerning protective devices for knees. In spring practice of 1983, only one major knee injury occurred. Several athletes who wore the brace had previously injured their knees. Their attitudes were positive toward the orthosis, and there were no new knee injuries in this group.

The double upright, plastic, prophylactic knee orthosis appears to be well tolerated by football players. With education of the athlete being a major concern, a positive attitude on the part of the coaching

Comparison of Speed and Agility With and Without Brace by Football Position

Position	Number Tested	Time in seconds With	Change	Without
Quarterback	1	8.6	No change	8.6
Defensive Lineman	2	8.8	No change	8.8
Offensive Lineman	6	9.3	.1 Slower	9.2
Linebacker	2	8.7	.2 Slower	8.5
Defensive Back	2	8.8	No change	8.8

Table 2.

staff is very important. The greatest deterrent to the success of this program was the coaches' attitudes towards the orthosis, and peer pressure from the nonwearers. These negative comments directed toward other athletes wearing the prototypes in fall, 1982, resulted in a minimal number of consistent wearers. Coaches' attitudes toward the brace must be positive if the program is to be successful.

The lateral hinge appears to be the most stressed during useage. It might be that the medial ligaments are spared if the force comes from an oblique angle to the knee. Closer study of the field position of the athlete may clear up this interesting finding.

It is interesting to note that, as of this date, the players believe that this device is worth using, and is not detrimental to their abilities. Also of interest is that they tolerate the small amount of mechanical breakage and the occasional need to change joints and uprights or to alter the limits of the orthotic hinge by using stops.

Weight and bulkiness of the device has not been a problem. A pair of the nylon joints without the plastic cuffs weighs only seven ounces. The device with the copolymer cuffs and the nylon joints weights 13 ounces.

CONCLUSION

Based on a retrospective analysis of 31 football players at Iowa State University wearing 62 plastic, prophylactic knee orthoses, it appears that the wearing of such a device is tolerated by both coaches, trainers, and players. Additionally, the evidence of only one knee injury in this group during 20 days of spring practice demands that more study be given to this concept in order to determine statistically how valid the relationship is between injury and use of the orthosis.

AUTHORS

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Technical Note:

ORTHOSIL Silicone Gel for Pads and Soft Insert Liners

Lynette K. Black, R.T. (P)

INTRODUCTION

Before becoming concerned with the functional components of prostheses and their alignment, we must first address the connection between body and prosthesis. "Considerable forces are involved in standing and walking"[1] which must be absorbed by the prosthesis, and most importantly, by the interface between the soft tissues of the residual limb and the prosthetic socket. "These soft tissues will be deformed and displaced by the external forces,"[1] affecting the prosthesis. During casting, and through socket design, these soft tissues are purposely manipulated and shaped. These external forces and the forces resulting from movement of the human body are taken into consideration in the design of the prosthetic socket.

In the majority of cases, amputees can tolerate these forces if they have a well fitted socket, with or without a soft insert liner. However, there are always exceptions. Even with conventional inserts of leather, kemblo, or polyurethane foam materials, patients with extensive scar tissue, skin grafts, or minimal tissue covering of the skeleton may require something special to prevent tissue breakdown.

SILICONE GEL

When silicone gel was introduced, its unique properties seemed to be ideal to meet the special needs of these amputees. Silicone gel is a semi-fluid material that will maintain a constant volume. Pure silicone gel padding provides an excellent weight-bearing support.[2] When used to fabricate soft insert liners, it can help absorb external forces by allowing movement within the gel, rather than between the skin and the socket or liner.

Although silicone gel has many excellent characteristics, there have been some disadvantages with using it. Previously, it required a complicated fabrication technique which took a great deal of time. Silicone gel was often bonded or injected between layers of leather. There were problems of durability. Delamination would occur between the leather and the silicone. Sometimes, if the leather were not sealed properly, the silicone gel leaked through the liner. After repeated weight-bearing, there also was a tendency for the silicone gel to migrate, thus no longer providing protection for those areas where it may be needed most.

A NEW APPROACH

This article will describe a new product and a simplified fabrication technique for silicone gel padding and soft insert liners. The product is ORTHOSIL Silicone Gel, available from OTTO BOCK Orthopedic Industry.

PROPERTIES OF ORTHOSIL

ORTHOSIL Silicone Gel consists of two liquid components which are combined to fabricate a socket insert liner or padding. ORTHOSIL was developed so that its physical and chemical properties would meet the special requirements of the prosthetics field. Advantages include:

1. It does not irritate the skin and is physiologically non toxic.
2. Due to its special molecular structure, its mechanical properties are similar to the physical characteristics of subcutaneous tissue.
3. There should not be any problems with migration of the ORTHOSIL Silicone Gel Liner. The Liner can be washed with mild soap and water and temperatures up to 356°F will not harm it.
4. ORTHOSIL Silicone Gel is available in two different types. One produces a flexible product and is used for the fabrication of soft insert liners. The other produces a more firm product, which may be used for end pads and distal end-bearing cushions for Symes and knee-bearing sockets. The flexible silicone and the more firm silicone can be mixed together in varying amounts before adding the catalyst, to achieve varying degrees of density and elasticity.

FABRICATION TECHNIQUES

To fabricate an ORTHOSIL silicone gel below knee socket liner, prepare the plaster positive model the same way as you would for lamination. Use PVA sheeting as a parting agent for dry plaster models. To seal wet plaster models, use a parting lacquer and then PVA Sheeting. Do not get talcum powder on the outside of the PVA. If talcum powder is absorbed into the silicone, it will reduce the sheer strength of the silicone gel.

To make a distal end-bearing cushion in a socket liner, a plastic mold in the shape of a cup is available. Spray the mold with silicone spray and then pull it over the distal end of the plaster positive model (Figure 1). 617H44 Silicone Gel and 617H45 ORTHOSIL Catalyst are carefully measured by weight. The mixing proportion is nine parts of silicone gel to one part of ORTHOSIL catalyst. The liquid components must be thoroughly mixed with care, yet not whipped.

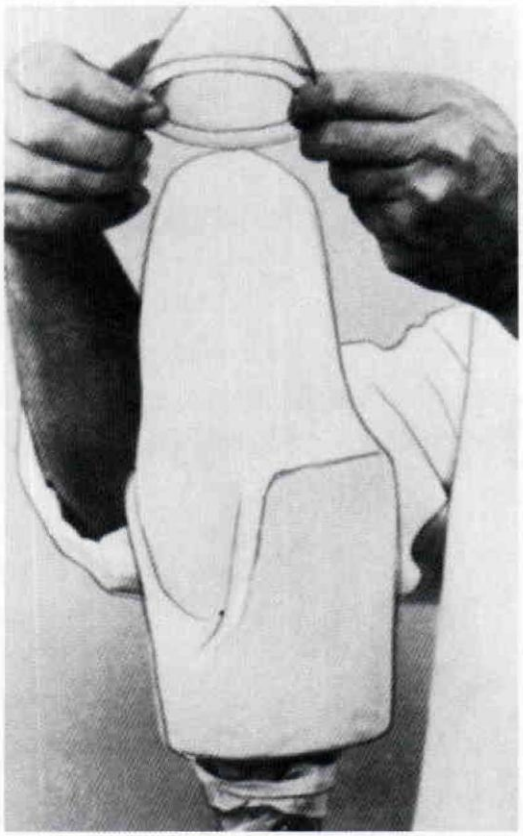


Figure 1: A plastic mold in the shape of a cup is pulled over the distal end of the positive model.

The mixture is then poured into the opening in the top of the distal end pad mold (Figure 2). To make a pull sock hole in the socket liner, sand a 4 mm thick, ¾ inch wide piece of polyethylene to the contour of the distal end of the plaster model. Insert this polyethylene piece vertically into the opening in the distal end pad mold (Figure 3). A bigger slot can be made for patients who may have trouble removing the pull sock. After about 45 minutes, the silicone mixture will vulcanize so that the mold can be removed.

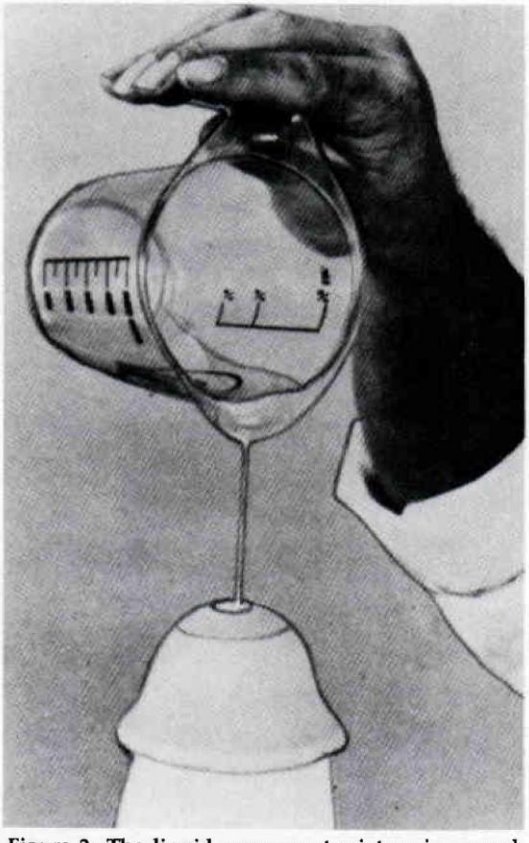


Figure 2: The liquid component mixture is poured into the opening at the top of the distal pad mold.

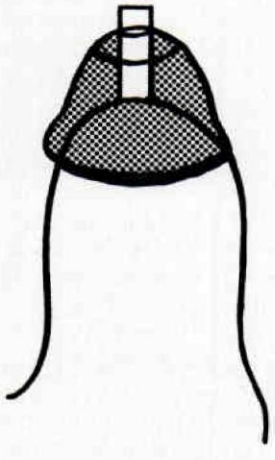


Figure 3: To create a pull sockhole, a polyethylene tube is placed through the opening in the distal end pad mold.

If the silicone gel liner is fabricated immediately after the end-bearing cushion, the ORTHOSIL will chemically bond to itself. ORTHOSIL will not adhere to itself after about three hours, when the vulcanization process has been completed. If you want to bond another layer of silicone to an existing one, a special bonding agent 617H46 must be painted on the surface.

In preparation of the gel insert, six layers of elastic stockinette are used in the layup and then an outer PVA bag is applied. Use only 623T13 Elastic Stockinette for fabrication of the silicone gel liner. Only this elastic stockinette has been specially woven and treated to be compatible with ORTHOSIL silicone gel. The elasticity of the stockinette allows stretch circumferentially, but not lengthwise.

Weigh out and mix thoroughly nine parts to one respectively of 617H43 ORTHOSIL Silicone Gel and ORTHOSIL Catalyst. A special caucasian pigment paste is available for ORTHOSIL. Only this pigment can be used with ORTHOSIL. Immediately pour the silicone mixture into the opening of the PVA bag. Do not evacuate the air from the layup before pouring in the silicone gel. Squeeze the bulk of the ORTHOSIL into the layup at the distal end of the model. Tie off the PVA bag around the polyethylene piece, still found in the distal end pad, and turn on a light vacuum. Invert the plaster positive model to a position of about 130° to allow the air to be evacuated ahead of the gel (Figure 4). Distribute the mixture slowly, and return the cast to a vertical position.

When the silicone gel begins to vulcanize, distribute the mixture to the boney prominences or any other areas that require extra padding. Once vulcanization begins, the silicone gel mixture does not shift easily and it will remain where it is distributed when fully vulcanized.

At a room temperature of 70°F the vulcanization process should be completed in about two hours. The silicone gel liner can be removed and sprinkled with talcum powder on the inside and outside. The proximal brim should be trimmed using a pair of sharp scissors, as sanding will expose fibers of the stockinette (Figure 5).

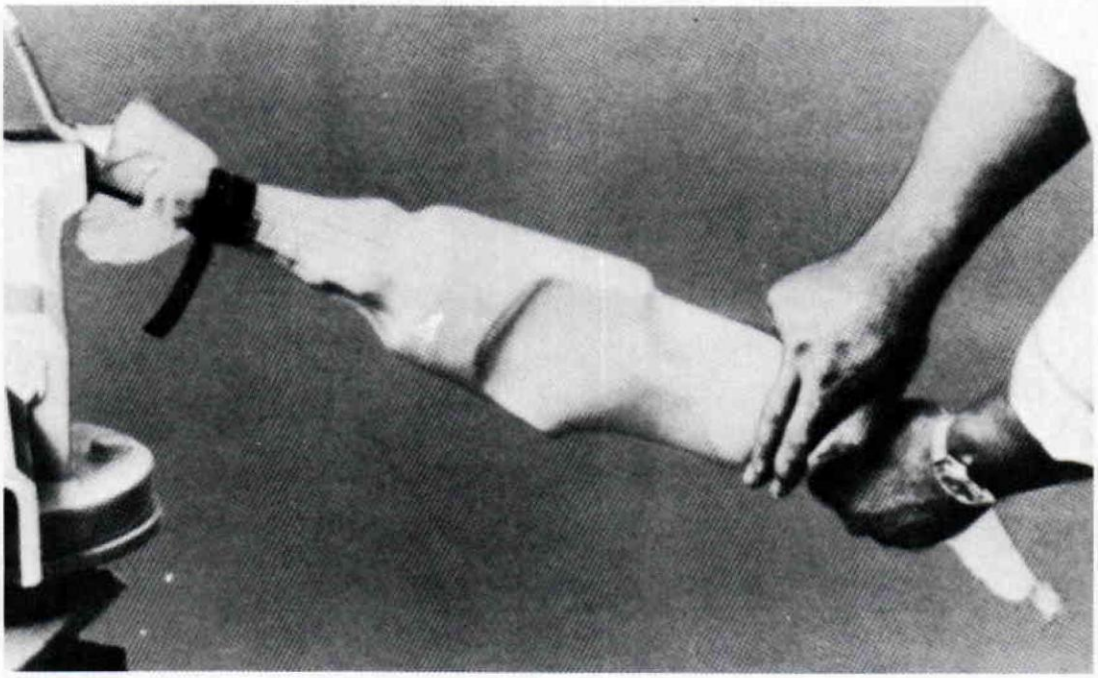


Figure 4: Invert the positive model to allow air to evacuate ahead of the gel.

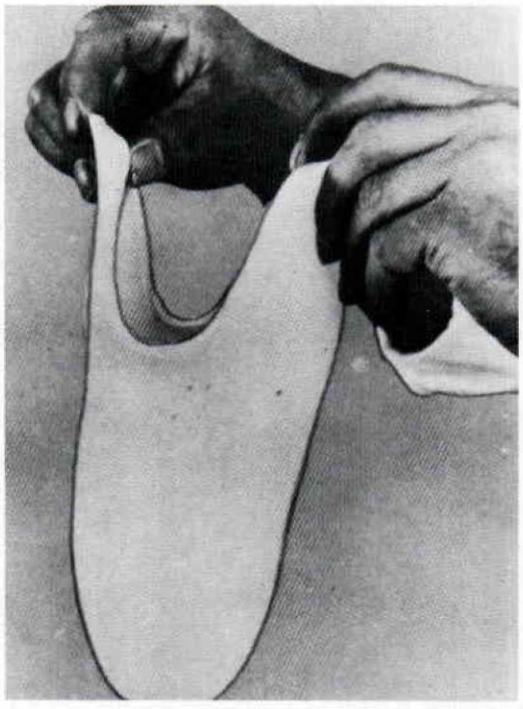


Figure 5: The completed gel insert should be trimmed with a pair of scissors to avoid a rough finish.

ORTHOSIL silicone gel can be used for custom padding too. It could be especially useful as a toe filler for amputations of the forefoot. It can be used for an endbearing cushion in a Symes socket, and for ischial seats and anterior above knee socket brims, virtually anywhere where pressure points could cause problems.[3] When fabricating pads, prepare the plaster positive model in the same manner as stated previously, and use silicone spray as a parting agent.

SUMMARY

The many unique properties of silicone gel have been found to be of great benefit to amputees whose residual limbs have a large amount of scar tissue, skin grafts, or minimal tissue covering the skeleton. Silicone gel has also been used in sports prostheses, where greater stress is placed on the residual limb than in normal daily activities (sports such as hiking, skiing, racquetball, and tennis).

With the development of ORTHOSIL silicone gel and the accompanying 623T13 elastic stockinette, the fabrication process for padding and soft insert liners is relatively quick and easy. The lamination technique does not require the use of leather, and the problems of migration, sealing, and bonding are eliminated.

ORTHOSIL silicone gel should prove to be beneficial in fitting special problem cases.

NOTES

[1]Holmgren, G., "The Interface Between the Body and the Above-Knee Prosthesis," *Prosthetics and Orthotics International*, 3(1):31–36, April, 1979.

[2]Graves, C.P., Jack M., "The Selectively Placed Silicone Gel Liner System for PTB Prostheses," *Orthotics and Prosthetics*, Vol. 34, No. 2, pp. 21–24, June, 1980.

[3]Potter, C.O., John W., and Sockwell, Jack E., "Custom-Foamed Toe Filler for Amputation of the Forefoot," *Orthotics and Prosthetics*, Vol. 28, pp. 57–60, September, 1974.

AUTHOR

Ms. Black is with Otto Bock Orthopedic Industry, Inc., 4130 Highway 55, Minneapolis, Minnesota 55422.

Technical Note:

Water Safe Prosthetics

Edmond E. Koester, C.P.

INTRODUCTION

The following article is a presentation on the fabrication of a waterproof prosthesis for the below knee amputee. It is an economical process because it is basically a modification of an existing prosthesis.

The prosthesis is made using a 4mm subortholene socket, Otto Bock pedilen foam, and a Kingsley Beachcomber foot set up (Figure 1). The adhesive for attaching the foot is a Dow Corning general purpose adhesive. A silicone sealant is used around the edge of the section where the ankle and foot are bonded together (Figure 2).

FABRICATION

The following is a description of the molding of the subortholen socket and the set-up of the prosthesis.

First, the lay up of the PTB or PTS positive model is shown with the use of one cast sock, one nylon, and clear sealant (Figure 3). The next step is to insert the PTB or PTS positive model in a vacuum forming system, pulling the cast sock and nylon over the positive model. Now the subortholen is ready to be vacuum molded over the cast.

Figure 1: The components of the waterproof below knee prosthesis include a subortholene socket, "Beachcomber" foot and ankle, and Pedilen foam.

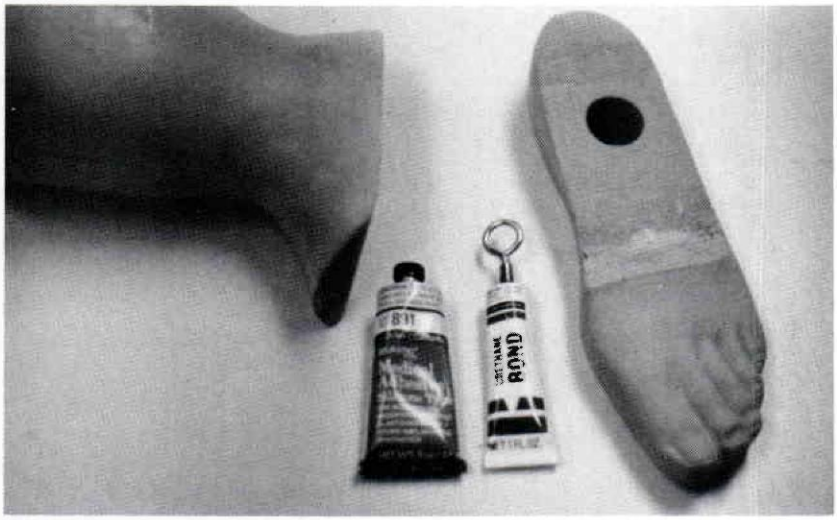


Figure 2: Attachment of the foot is accomplished by the use of Dow Corning general purpose adhesive and a silicone sealant.

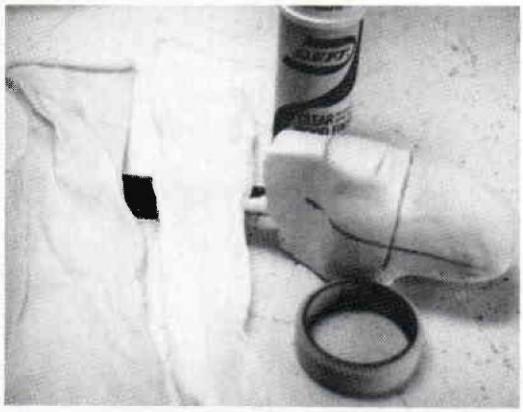


Figure 3: The layup for the subortholene socket includes clear sealant, one nylon, and a cast sock.

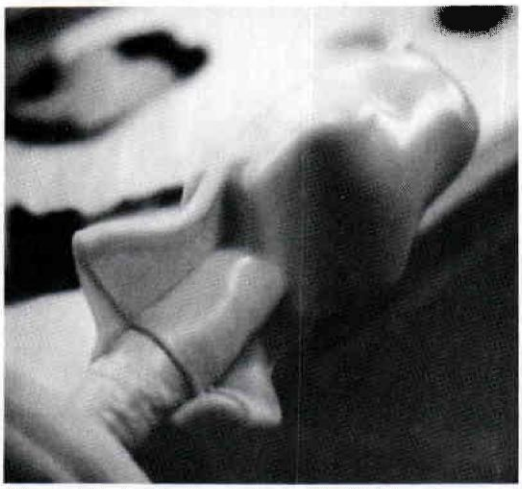


Figure 4: Subortholene is heated to 400°F and vacuum molded over the cast.

After the subortholen has been heated to 400°F, it is then molded, sealed off, and a vacuum is allowed to form, causing a weld. The excess subortholen is trimmed off while it is hot, with vacuum still on (Figure 4). The subortholen socket is then cooled with an air gun, and allowed to become completely cool. This is very important in order to allow the subortholen to bond well. Trim lines are then determined on the socket, and the cast is broken out. At this point, the socket is completely finished as a standard PTB or PTS socket.

The basic components of the waterproof prosthesis may now be assembled. These consist of the subortholen socket, Kingsley Beachcomber foot, pedilen Otto Bock foam system, and standard three layer nylon laminate. In our work, the urethane bonding agent for the Kingsley foot is virtually unbreakable as compared to any other agent which has been used in the fabrication of waterproof prostheses.

The first step is to assemble these components temporarily on an alignment jig, preferably a Staros Gardner coupling. The patient is then allowed to walk barefoot with the prosthesis through the normal alignment procedure. Emphasis is placed on the fact that the action will not be that of

a SACH foot or any kind of multi-axis ankle joint. It is therefore primarily necessary to align the prosthesis more for standing comfort than functional walking comfort. Since the prosthesis is constructed only for light duty use, the prosthesis weight is kept to a two to three pound maximum.

Once the prosthesis has been dynamically aligned, it is then finished as a standard prosthesis. The subortholen socket is completely roughed up with the large rough sanding cone. It is then shaped, and the bonding is done with the pedilen foam. The ankle block from Kingsley is cut to the lowest possible point. The prosthesis is then painted and sealed (Figure 5). Three layers of nylon are applied over the prosthesis and a normal lamination procedure is carried out. The only specialized lamination requirement is the extra time taken around the foot-ankle area and around the under cuts to eliminate as much weight from excess resin as possible. The base of the prosthesis is allowed to cure, and the distal end of the ankle section is ground to allow for a smooth flat bonding surface to the foot plate.

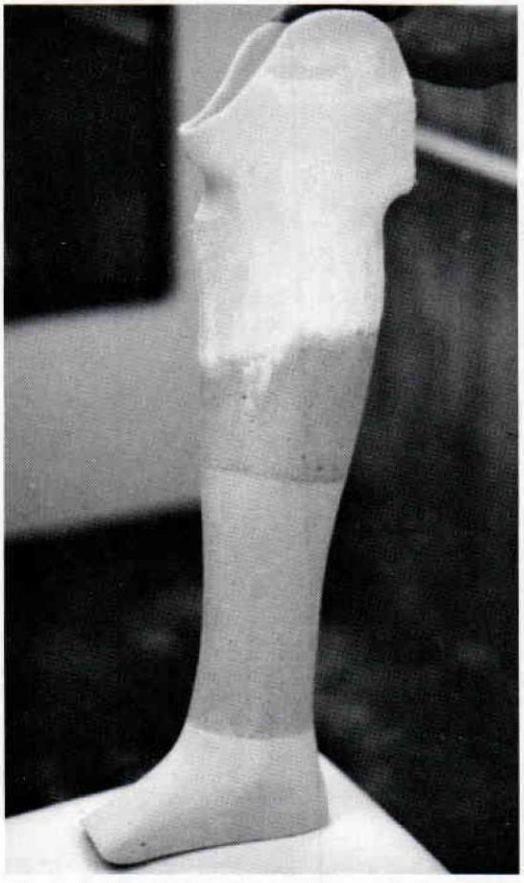


Figure 5: The limb, after bending and shaping, is ready to have the finish lamination applied.

The urethane bond is then applied to the foot, and the prosthesis (Figure 6), and requires twenty-four hours to cure. It is suggested that the adhesive be made smooth and even to make for a uniform bond. It should dry over night before removing tape (Figure 7).

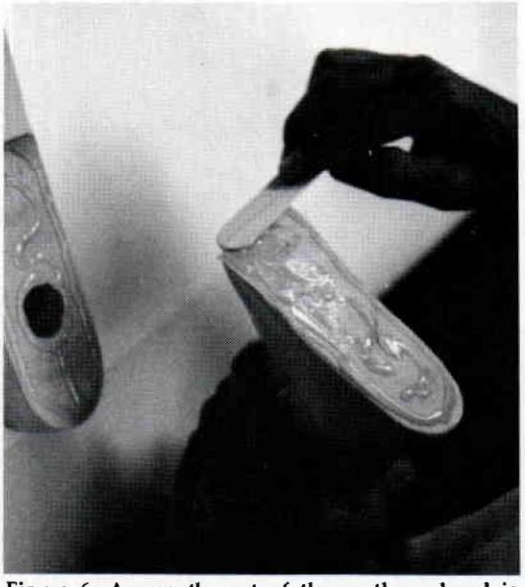
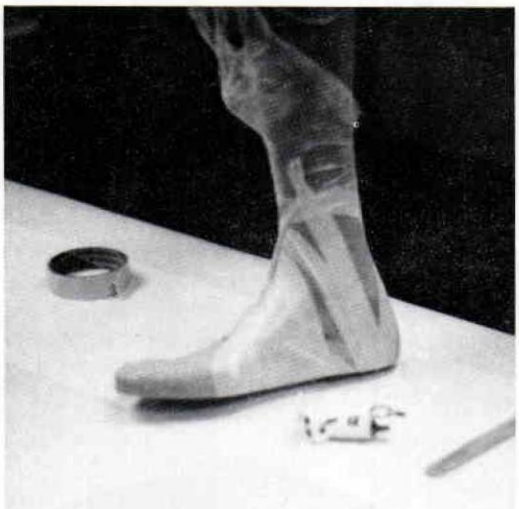


Figure 6: A smooth coat of the urethane bond is applied to the foot and ankle block.

Since urethane adhesive is a foaming agent, it does bubble and expand. Any excess should be trimmed off with a sharp knife. It should be smoothed with a felt cone or a very fine sanding cone. A medical adhesive should then be used to seal any areas which have been opened or exposed. A tongue blade may be used for this to give a uniform cosmetic appearance.

The upper section of the prosthesis is smoothed and completed as with a normal polyester laminate prosthesis. The advantage of subortholen in this system is its ease of workability, and ease of smoothing. The subortholen socket should only be made over a modified plaster positive



Fgure 7: After the foot is bonded, the foot and ankle block are taped and allowed to cure for 24 hours.

model that has proven to be a successful fit. Since the subortholen is only 4mm thick, it is impossible to relieve the socket much. The positive model used for the original prosthesis is saved. After the patient has worn his prosthesis for an extensive period of time, and proven it to fit well, then any small changes may be made to the positive mold, and the subortholen socket made from the corrected positive model.

CONCLUSION

I feel this system provides for an economical means to provide the patient with two prostheses: a standard prosthesis which may have a soft insert, and a multi-axis ankle joint or a SACH foot which cannot be allowed to get wet. The waterproof prosthesis enables the amputee to shower and enjoy water sports which he may have missed.

This prosthesis is very easy to construct, and offers reliability to the amputee, for its limited use. It can also be used at night as a temporary prosthesis, as it incorporates a flat heel SACH foot.

From the time this presentation was given, at the AOPA National Assembly in 1981, to now, more than two years have passed. My facility has supplied approximately twenty amputees with this prosthesis, and this system has proven itself to be successful. Work is presently being done on a similar system for above knee amputees, and within a few years a more complete study on waterproof prostheses should be available.

ACKNOWLEDGMENTS

The author wishes to thank his two assistants, John MacGregor and Holly Rames, for their help in performing this task. Also thanks to Kingsley Company for their Beachcomber foot, and Otto Bock for their waterproof components.

AUTHOR

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INDEX TO ORTHOTICS AND PROSTHETICS

Volume 37

By Title

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Number 4

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Volume 37

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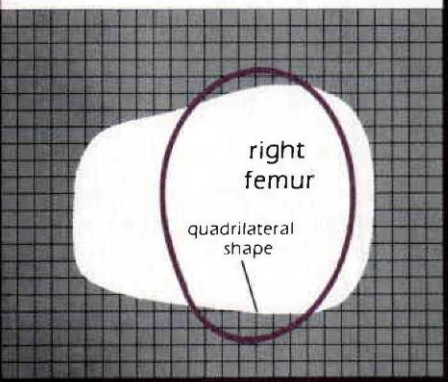

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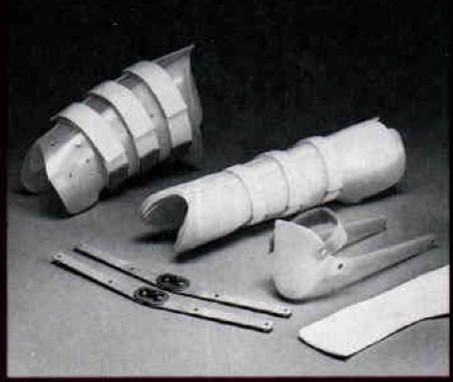

*U.S. Patent 4,320,748

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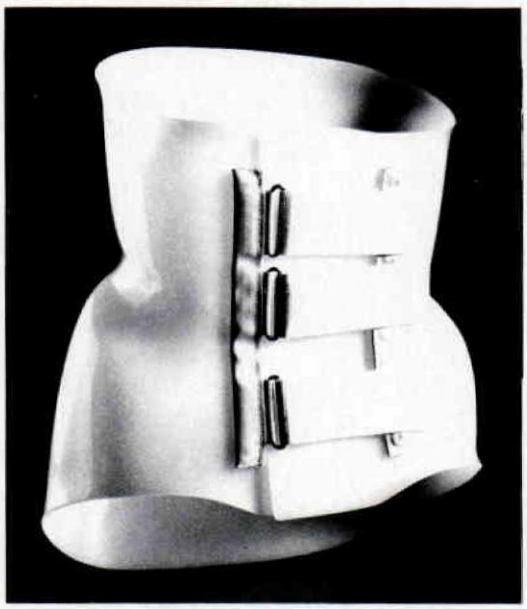

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REVIEWS

by Charles H. Pritham, C.P.O.

Low Back Pain Clinical Diagnosis and Management, Leonard P. Seimon, M.D., Appleton Century Crafts, 25 Van Zant St., East Norwalk, Connecticut 06855, 313 pages.

This book is written as a practical low key guide to clinical treatment of a complex problem. It stresses basics, covering in detail: anatomy, pathology, and sources of pain, as well as how to record a history and conduct a physical examination. Subsequent chapters consider specific problems and methods of treatment. Of particular interest in this latter vein is the discussion of techniques for spinal manipulation. The use of orthoses and spinal supports is similarly treated from a practical point of view, while underlying theoretical biomechanical principles are slighted. Approximately half the book is devoted to an extensive series of case histories.

Any orthotist interested in the treatment of low back pain might well find this book of interest.

Congenital Deformities of the Spine, Robert B. Winter, M.D., with contributions by John E. Lonstein, M.D., and Arnold S. Leonard, M.D., Ph.D., Thieme-Statton, Inc., 381 Park Avenue South, New York, N.Y. 10016, 343 pages and index, 1983, $59.00.

Dr. Winter in his preface describes this book as one of the results of a project that began 20 years ago at Gillette Children's Hospital, and that includes the results of some 1,250 cases seen in the St. Paul-Minneapolis area. He further describes the book as "a non-statistical" state-of-the-art review of the topic. The statistical research data is in the process of being presented in a series of scientific papers, each concerned with a separate part of the total topic. This seems to be an excellent approach, for it enables the clinician concerned with the treatment of patients to concentrate on the heart of the matter. All too often charts, graphs, and statistics—while important and essential to those engaged in research—impede a more general reader in his attempt to concentrate on the basics.

This book should be of interest to all orthotists involved in the treatment of scoliosis and other spinal deformities. It is perhaps best considered as a sequel to Blount and Moe's *The Milwaukee Brace*. It includes chapters on embryology, genetics, natural history, and classification and terminology. Separate chapters are devoted to various surgical techniques, surgical complications, and post-operative immobilization. Similarly, the implications and treatment of various congenital deformities and conditions are considered in separate chapters. Perhaps of greatest interest to orthotists is the chapter on non-operative treatment. In it, Dr. Winter makes firm recommendations for the use of therapy, body casts, CTLSO's, and TLSO's. In addition, the use of orthoses and external corrective devices are mentioned where appropriate in the various other chapters.

The various appropriate and inappropriate uses of orthoses and the attendant complications are discussed. Throughout, the book is well illustrated with photographs, x-rays, and line drawings.

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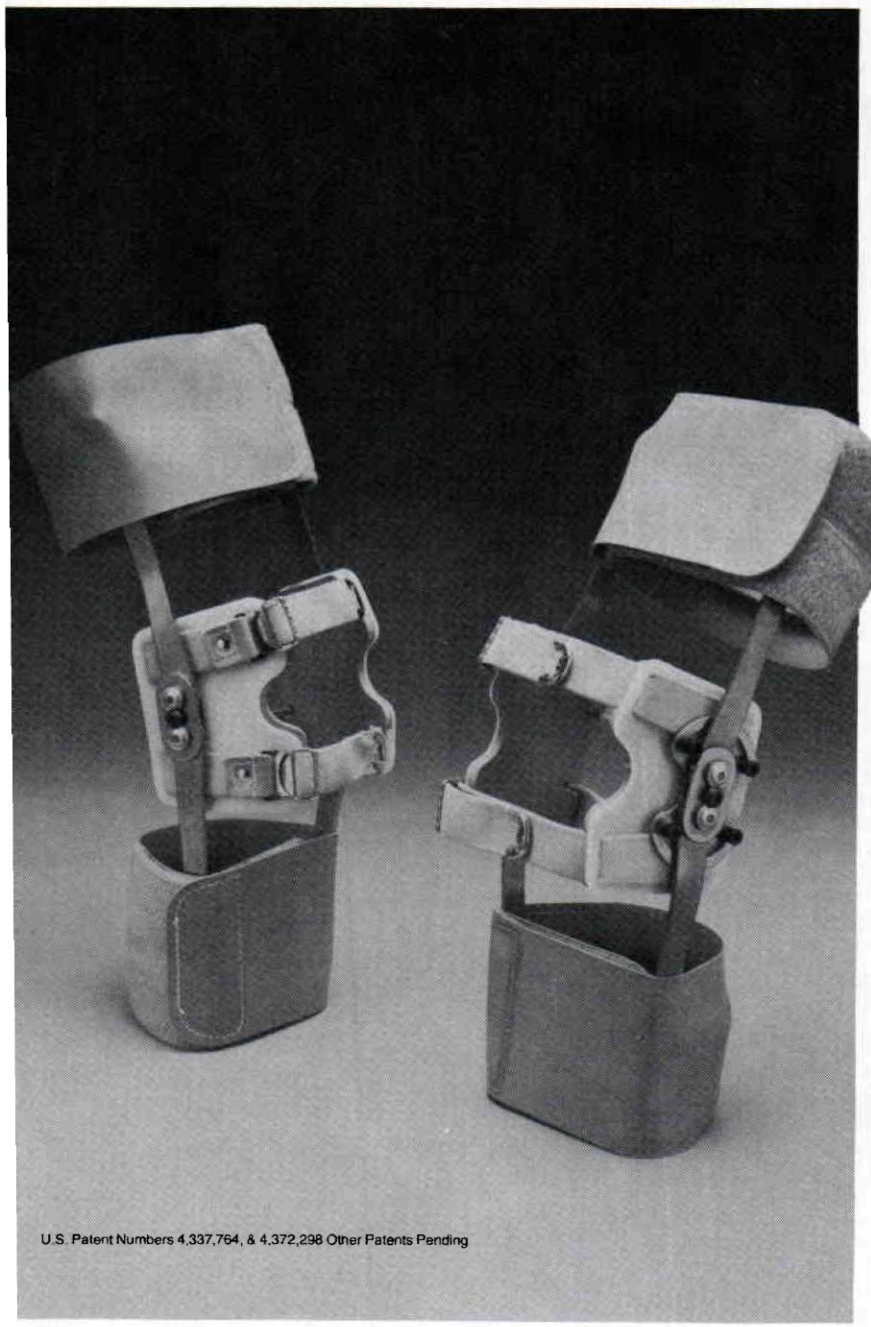
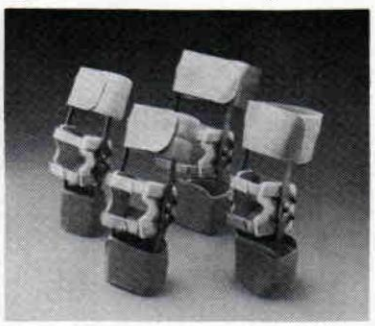


U.S. Patent Numbers 4,337,764, & 4,372,298 Other Patents Pending

™ Two completely pre-fabricated models are now available. The adjustable model may be used to control knee flexion and extension to any desired degree while the free knee model is used when knee joint motion need not be restricted. Available in four sizes, left and right, the Lerman Multi-Ligamentus™ Knee Control Orthosis may be applied to most patients right off the shelf. It's easy to apply and remove, may be fit without precise measuring devices or techniques and is easily contoured to achieve an excellent fit without disturbing knee joint alignment.



Adjustable Model Product Numbers:
Small - Right — A16-6RS-M000
Medium - Right — A16-6RM-D000
Large - Right — A16-6RL-G000
X-Large - Right — A16-6RX-L000
Small - Left — A16-6LS-M000
Medium - Left — A16-6LM-D000
Large - Left — A16-6LL-G000
X-Large - Left — A16-6LX-L000

Free Knee Model Product Numbers:
Small - Right — A16-6RS-M001
Medium - Right — A16-6RM-D001
Large - Right — A16-6RL-G001
X-Large - Right — A16-6RX-L001
Small - Left — A16-6LS-M001
Medium - Left — A16-6LM-D001
Large - Left — A16-6LL-G001
X-Large - Left — A16-6LX-L001

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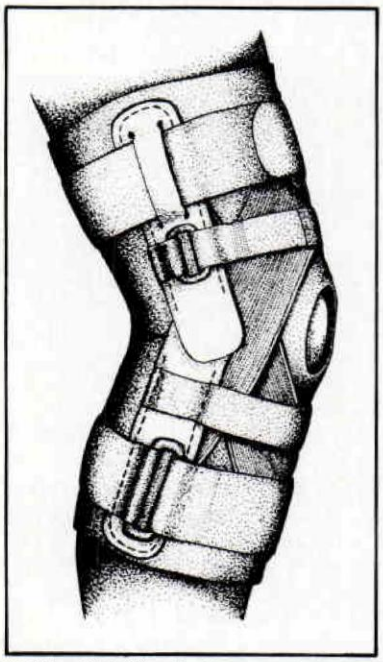
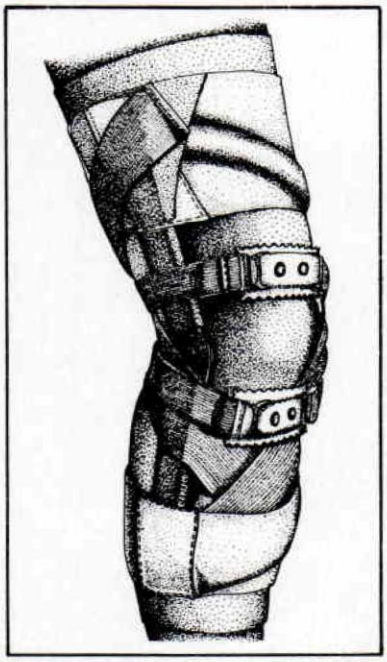
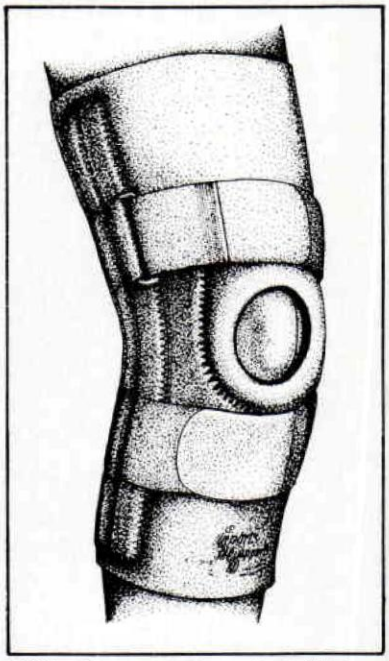


1111-F - HINGED FAN KNEE SUPPORT "NEW PRODUCT"



1000-LH - SLEEVE TO FIT UNDER DEROTATION BRACE



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We at Sports Supports realize that there are certain times when a patient can not be fitted with a stock product or size or that modifications must be made for a patient's special needs. If you will supply us with the proper measurements, we will be happy to fabricate a custom orthosis.

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TO: PERSONS WORKING IN REHABILITATION

FROM: SIEGFRIED PAUL, CPO (E), INTERNATIONAL SCIENTIFIC PROGRAM CO-CHAIRMAN
ROBERT E. FANNIN, CO, DOMESTIC SCIENTIFIC PROGRAM CO-CHAIRMAN

RE: CALL FOR CONTRIBUTED PAPERS FOR THE 1984 ASSEMBLY SCIENTIFIC PROGRAM

The American Orthotic and Prosthetic Association is an organization whose 800-plus membership consists of firms involved in the design, manufacture, and fitting of orthoses and prostheses. The primary objective of AOPA is to promote high levels of orthotic/prosthetic patient care services to the orthopedically handicapped. To aid in achieving this goal, each year the Association provides a forum, via its annual National Assembly, for orthotics and prosthetics professionals to share information on the many new ideas and/or concepts of or relating to orthotics/prosthetics. Nearly everyone working in orthotics and prosthetics in the United States attends the Assembly, along with many professionals from abroad. The 1984 Assembly will be held at the Fontainebleau Hotel, Miami Beach, Florida on October 15-21, 1984.

AOPA invites all interested persons to submit an abstract(s) for presentation during the Assembly's Scientific Program. The subject(s) for the abstract(s) should be new ideas, techniques, devices, and/or research that have a practical application in orthotics and prosthetics or a related field. Interested persons are invited to submit more than one abstract. Most presenters will be given 15 minutes for their presentation.

If you are interested in participating in the 1984 Assembly, please fill out the enclosd abstract form and return it to the AOPA National Headquarters no later than April 30, 1984.

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Thank you.

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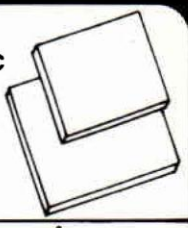

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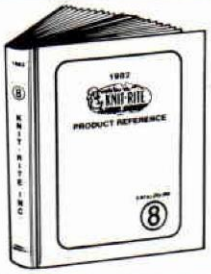

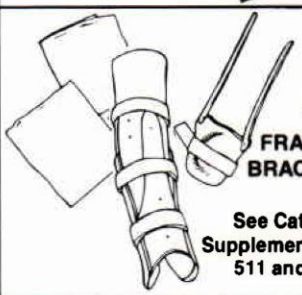

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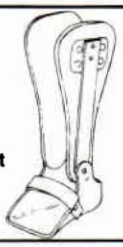

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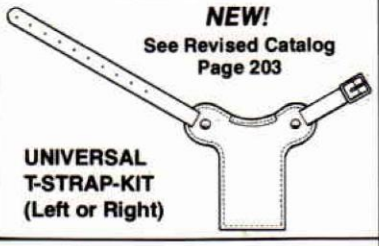

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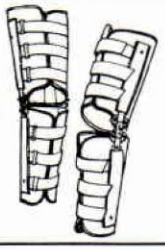

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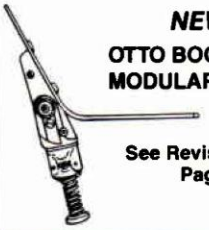

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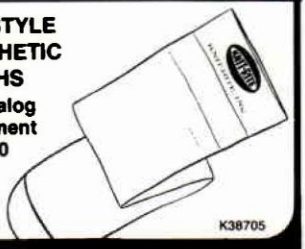

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