Clinical Prosthetics & Orthotics

Disarticulation Amputations

Component Selection Criteria: Lower Limb Disarticulations	John Michael, M.Ed., C.P.O.
The Hip Disarticulation Prosthesis As Developed by the O.I.M. Noord Nederland	Peter Tuil
The UCLA Anatomical Hip Disarticulation Prosthesis	David H. Littig, B.A., C.P. Judd E. Lundt, B.S., A.E.
The CAT-CAM-H.D.™ A New Design for Hip Disarticulation Patients	John Sabolich, B.S., C.P.O. Thomas Guth, B.A., C.P.O.
Technical Note: Cosmesis and the Knee Disarticulation Prosthesis	Robert Gilley, C.P.
With a Spring in One's Step	D.D. Murray, M.D. W.J. Hartvikson H. Anton E. Hommonay, C.P.O.(C) N. Russell, C.P.(C)

Volume 12, Number 3 **Summer, 1988**

CONTENTS

Volume 12, Number 3 **Summer, 1988**

Disarticulation Amputations

Component Selection Criteria: Lower Limb Disarticulations 99
John Michael, C.P.O.
A brief overview is provided by the author to clarify the criteria involved in selecting the most appropriate components for disarticulations.

The Hip Disarticulation Prosthesis as Developed by the O.I.M. Noord Nederland 109
Peter Tuil
This article outlines the benefits of the O.I.M. hip disarticulation prosthesis and provides fabrication guidelines.

The UCLA Anatomical Hip Disarticulation Prosthesis 114
David H. Littig, B.A., C.P.
Judd E. Lundt, B.S., A.E.
This article discusses a new approach to hip disarticulation and hemipelvectomy fittings developed at the UCLA Prosthetics Education Program.

The CAT-CAM-H.D.™ A New Design for Hip Disarticulation Patients 119
John Sabolich, B.S., C.P.O.
Thomas Guth, M.D., C.P.O.
The authors discuss the advantages of a new CAT-CAM hip disarticulation socket developed by RGP of San Diego and Sabolich Prosthetic Research Center in Oklahoma City.

Technical Note: Cosmesis and the Knee Disarticulation Prosthesis 123
Robert Gilley, C.P.
The author recommends a fabrication method for a knee disarticulation prosthesis designed to eliminate cosmetically unacceptable gaps caused by knee flexing.

With a Spring in One's Step 128
D.D. Murray, M.D.
W.J. Hartvikson
H. Anton
E. Hommonay, C.P.O. (C)
N. Russell, C.P. (C)
An evaluation of the performance of the Seattle foot and attempt to determine, subjectively and objectively, whether or not it improves prosthetic gait.

From the Editor:

Opinions expressed in *Clinical Prosthetics and Orthotics* are solely those of the authors. No endorsement implied or explicit is made by the editorial staff, or by the American Academy of Orthotists and Prosthetists.

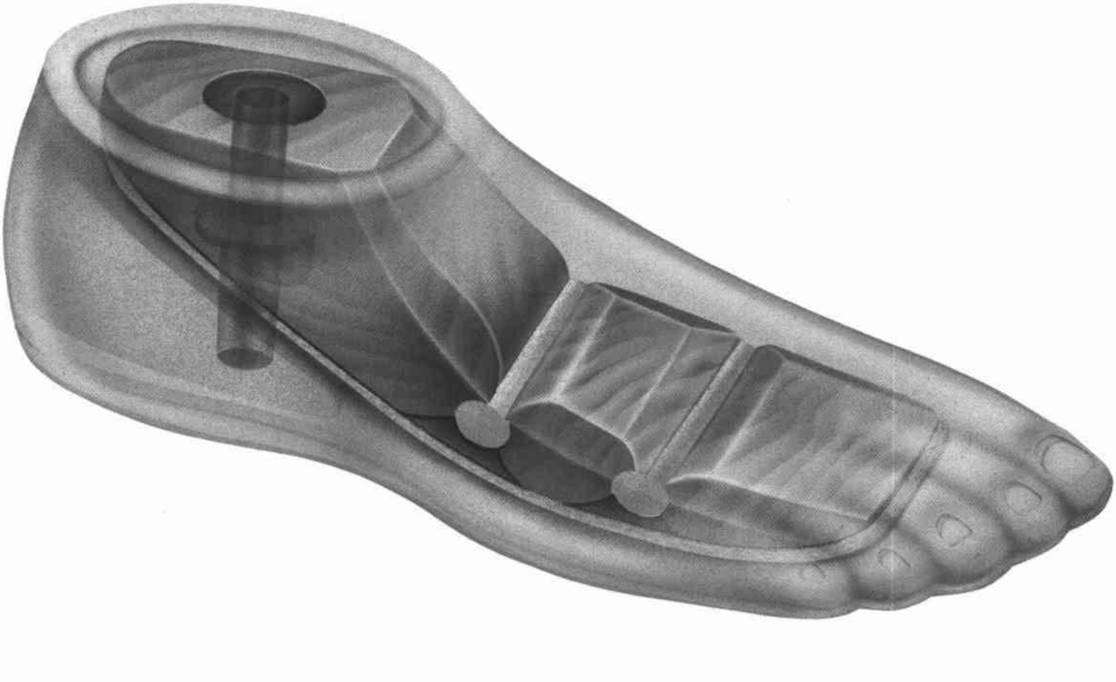
FEATURES

Letter from the Editor iv
Calendar ix

AD INDEX

Durr-Fillauer Medical, Inc. vi
Kingsley Mfg. iii
Knit-Rite, Inc. C3
Pel Supply viii
Southern Prosthetic Supply C2
United States Mfg. Co. x

Clinical Prosthetics and Orthotics (ISSN 0735-0090) is published quarterly by the American Academy of Orthotists and Prosthetists, 717 Pendleton St., Alexandria, VA 22314. Subscriptions: $25.00 domestic, $30.00 foreign, $40.00 air mail. Second-class postage paid at Alexandria, Virginia and additional mailing offices. POSTMASTER: Send address changes to *Clinical Prosthetics and Orthotics,* 717 Pendleton St., Alexandria, VA 22314.



Letter From the Editor

Dear Reader,

The American Academy of Orthotists and Prosthetists is pleased to announce that it will be publishing *Clinical Prosthetics and Orthotics* under a new title in cooperation with the American Orthotic and Prosthetic Association (AOPA). The result of this cooperative effort will be the *Journal of Prosthetics and Orthotics*. The first issue of the new quarterly journal will be published this October, 1988.

The *Journal of Prosthetics and Orthotics* will be equally supported by AOPA and the Academy and will be backed by an outstanding editor and editorial board. The new journal will be the only orthotic and prosthetic journal in the United States and will alternate between solicited and unsolicited issues. Each solicited issue will focus on a topic that the editorial board considers to be relevant and the most current concern to the orthotics and prosthetics field. Prosthetists, orthotists, and other allied health professionals who are best able to write on the chosen topic—the experts—will be solicited.

Every alternate issue will feature articles which are submitted for publication and accepted, after a full review by the editorial board. Again, articles chosen for publication will be relevant and current.

The editor of the *Journal of Prosthetics and Orthotics* will be Charles Pritham, CPO, former editor of *Clinical Prosthetics and Orthotics*. Members of the editorial board are as follows:

Larry Lange, CPO—former editor of *Orthotics and Prosthetics,* AOPA's quarterly journal (also to cease publication)
John W. Bowker, M.D.—Director of the Department of Orthopaedics and Rehabilitation at the University of Miami School of Medicine and former member of the *C.P.O.* editorial board
Wayne Daly, CPO—former member of the *Orthotics and Prosthetics* editorial board
Michael J. Quigley, CPO—chairman of *C.P.O.* and former editor for *Orthotics and Prosthetics*
Robert C. Manfredi, Jr., CPO—former member of the *Orthotics and Prosthetics* editorial board
Michael P. Greene, CPO—former member of *Orthotics and Prosthetics* editorial board
Tina-Laura Hittenberger, CO—former member of *Orthotics and Prosthetics* editorial board and chairperson of the 1988 AOPA Scientific Sessions

The editor and members of the editorial board were jointly appointed by the Presidents of AOPA and the Academy.

Thank you for your support. If you have any questions concerning the new journal, please feel free to call or write Sharada Gilkey, Managing Editor, *Journal of Prosthetics and Orthotics,* 717 Pendleton Street, Alexandria, Virginia 22314; (703) 836-7119.

Sincerely,
Charlie Pritham, CPO
Editor

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 99–108

Component Selection Criteria: Lower Limb Disarticulations

by John Michael, M.Ed., C.P.O.

Because disarticulations comprise only a small percentage of the lower limb amputations performed each year[1], questions sometimes arise regarding the most appropriate components to select. This paper will present a brief overview in an effort to clarify the criteria involved.

Hip Disarticulation/Hemipelvectomy Components

For the hip disarticulation or hemipelvectomy case, component selection is generally analogous to the more familiar above-knee patient. Endoskeletal components are preferred for the high level amputee because they offer light-weight and enhanced cosmetic appearance. A clear trend away from steel components to the much lighter titanium or carbon fiber versions is apparent. Most systems (particularly the Otto Bock "Modular") also permit subtle realignment, even in the definitive prosthesis. This can be an advantage due to the complex interplay between the mechanical hip, knee, and foot mechanisms.

Hip Joint Mechanisms

In general, a free motion hip joint is preferred, as originally proposed by McLaurin in 1954.[2] Careful attention to alignment details results in a very stable configuration by virtue of the weight line and reaction line forces. This permits very safe weight-bearing, yet allows easy hip flexion during swing phase.

Stride length is generally controlled by a spring or elastic flexion limiting apparatus, sometimes called an "extension bias." In modern practice, the joint is placed near the anterodistal quadrant of the socket, which sometimes requires a slightly shorter thigh segment for the best appearance when sitting.

Manual locking hip joints are also available but should be reserved as the component of last resort, even for bilateral amputees. In addition to disrupting swing phase, locked joints require the use of one hand on the unlocking mechanism during sitting. This often makes a difficult task more complicated, particularly for the double amputee.

More importantly, a locked hip joint may place the patient in a more dangerous position during a fall backwards. If the joint prevents flexion at the hips, the head rather than the buttocks may strike the ground first. In our last 50 consecutive fittings at Duke, both unilateral and bilateral hip/hemi patients have never required a locked joint to ambulate securely.

Two variations in hip joint design warrant mention. Peter Tuil of the Netherlands advocates the use of a reversed polycentric knee disarticulation joint (Otto Bock 3R21) as a hip joint.[3] Benefits claimed are parallel to those expected from a polycentric knee unit: in creased ground clearance during swing phase due to the inherent "shortening" of the linkage in flexion and enhanced stability at heel strike (Figure 1).

This view has been corroborated in a number of fittings over the past few years at the Royal

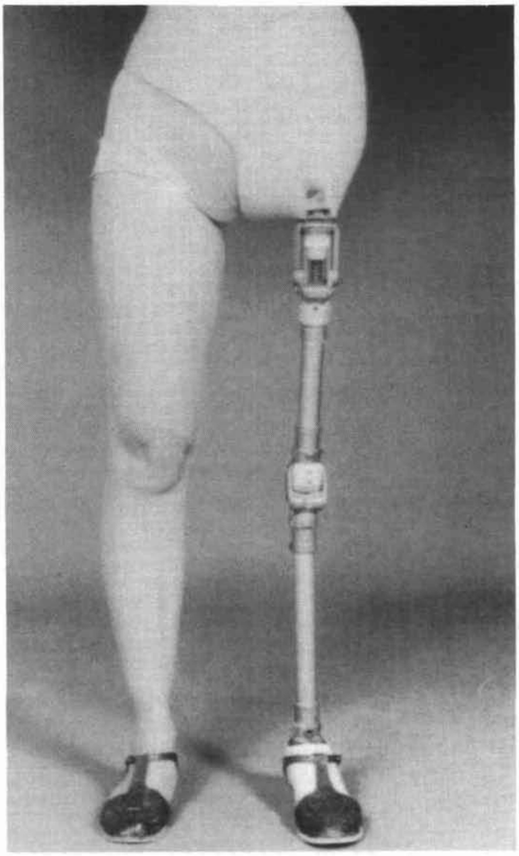


Figure 1. Prosthesis utilizing reversed polycentric knee disarticulation mechanism at the hip, as proposed by Peter Tuil of the Netherlands. (Courtesy of *Orthotics & Prosthetics,* 38/1, p. 33.)

Ottawa Regional Rehabilitation Centre in Canada.[4] Such a technique has also worked well in our hands at Duke, although we are not certain the benefits fully justify the special effort involved.

An even more intriguing concept is the "Hip Flexion Bias" modification promulgated by Haslem, et al. of Houston, Texas.[5] In this system, hip extension from heel strike to midstance compresses a specially selected spring, which encircles the endoskeletal pylon. At toe-off, this kinetic energy is released and the thigh segment is propelled briskly forward (Figure 2).

Not only does this result in a much more cosmetically "normal" gait, it also significantly improves ground clearance in swing phase.

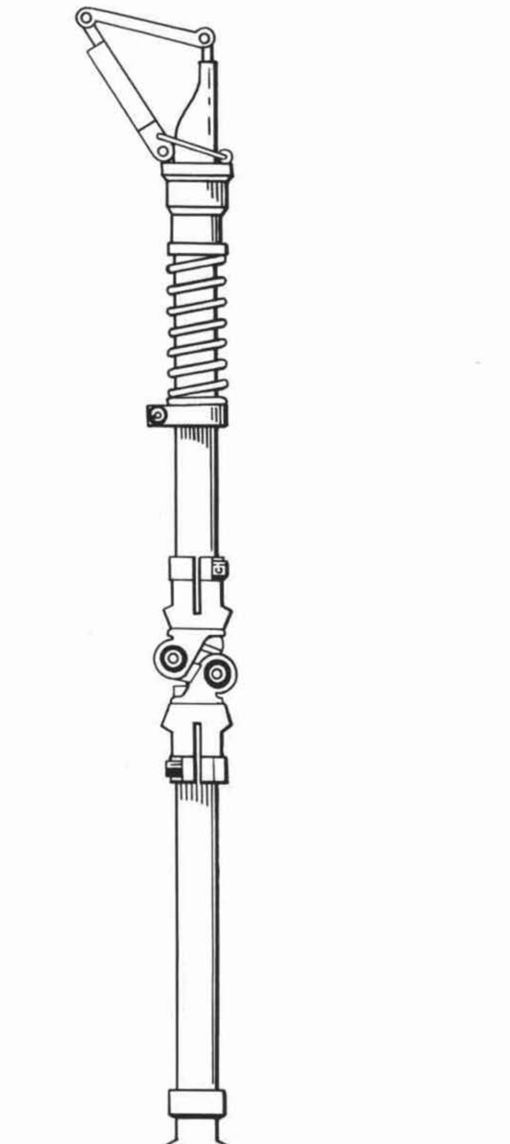


Figure 2. Hip Flexion Bias system designed by Haslam et al. of Houston, Texas. Note compression spring encircling thigh tube, which propels the limb forward during swing phase. (Redrawn from reference 5)

One of the inherent limitations of the Canadian hip disarticulation alignment system is the prosthesis must be significantly short (1cm+) to avoid forcing the amputee to vault for toe clearance.

Figures 3 and 4 illustrate the biomechanics of the Canadian design. At toe-off, the heel rises up during knee flexion and pulls the hip joint firmly against its posterior (extension) stop. The thigh segment remains vertical until the knee has reversed its direction of motion and contacted the knee stop. Only then does the thigh segment rotate anteriorly, causing the hip joint to flex. In essence, the prosthesis is at its full length during midswing. Since the patient has no voluntary control over any of the passive mechanical joints, the prosthetist is forced to shorten the limb for ground clearance.

The hip flexion bias system neatly avoids this dilemma. As a result, the prosthesis can be lengthened to a nearly level configuration in most cases. However, two potential problems have been noted with this approach. One is the development of annoying squeaks in the spring mechanism after a few months of use, which sometimes tend to recur inexorably.

A more significant concern is that as the spring compresses between heel strike and midstance, it creates a strong knee flexion moment. Unless this is resisted by a stance control knee with friction brake or a polycentric knee with inherent stability, the patient may fall. Since the friction brake mechanisms lose their effectiveness as the surface wears, the polycentric knee is the preferred component with this hip mechanism.[6]

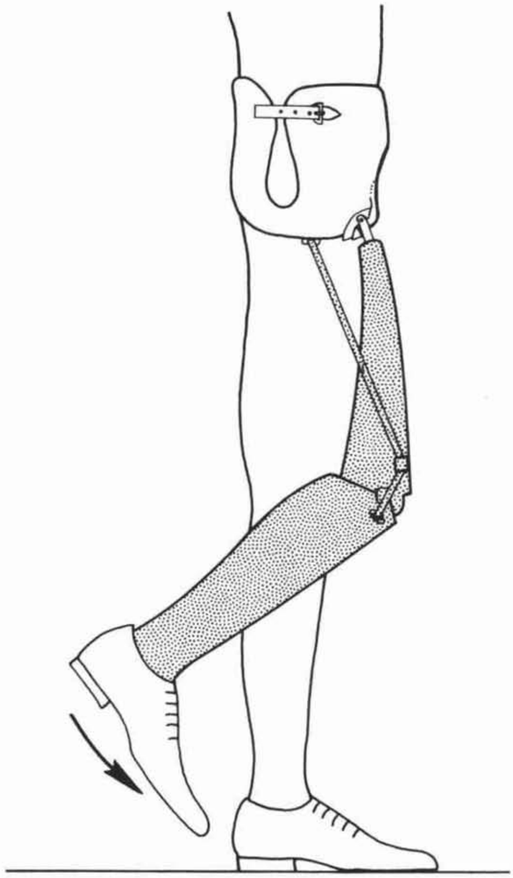


Figure 3. Canadian prosthesis in early swing phase. Hip joint remains neutral as shank swings forward. (Redrawn from reference 13)

Knee Joint Mechanisms

Other than the exception discussed above, knee mechanisms are selected by the same criteria as for above-knee amputees. The single axis/constant friction design remains the most widely utilized due to its light weight, low cost, and excellent durability. The friction resistance is often removed to ensure the knee reaches full extension as quickly as possible. A strong knee extension bias enhances this goal, offering the patient the most stable biomechanics possible with this mechanism.

Although this was proposed as the knee of choice for the Canadian hip disarticulation design, more sophisticated mechanisms have proven their value and are gradually becoming more common. The friction brake stance control knee (Otto Bock 3R15 or equivalent) is probably the second most frequently utilized component.

Because there is very little increase in cost or weight and reliability has been good, many clinicians feel the enhanced knee stability justifies this approach—particularly for the novice amputee. Mis-steps causing up to 15° knee flexion will not result in knee buckle, making gait training less difficult for the patient or therapist.

The major drawback to this knee is that the limb must be non-weight-bearing for knee flexion to occur. Although this generally

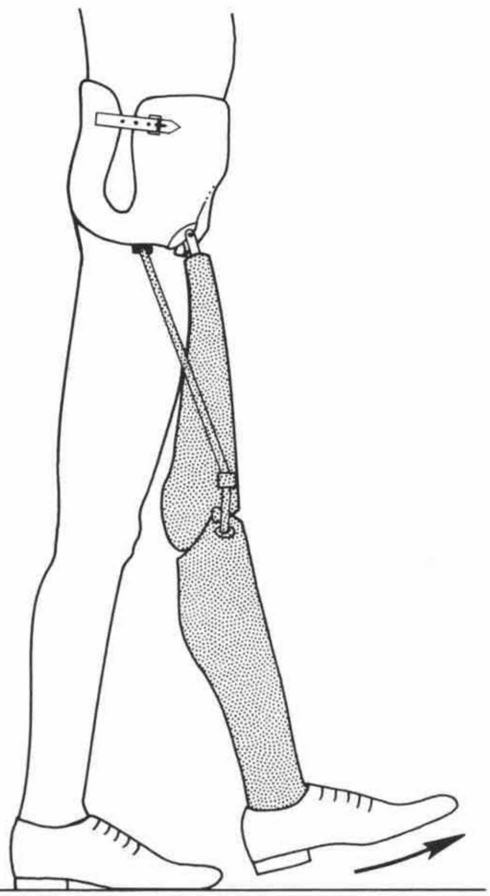


Figure 4. Canadian prosthesis just after mid-swing. Hip joint does not flex until shank motion is arrested by terminal extension stop. Prosthesis is fully extended at the instant of mid-swing. (Redrawn from reference 13)

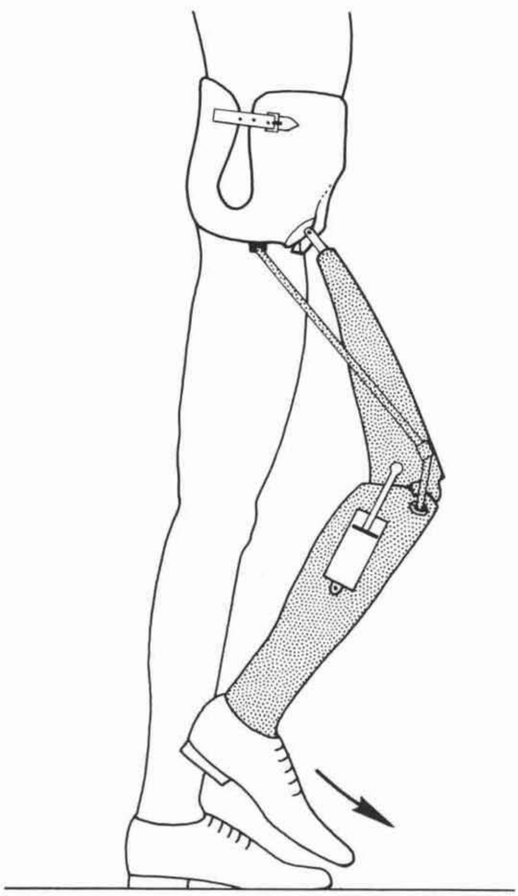


Figure 5. Canadian prosthesis with fluid controlled knee mechanism at mid-swing. Hydraulic extension resistance allows shank momentum to flex hip joint. Increased ground clearance may result. (Adapted from reference 13)

presents no problem during swing phase, some patients have difficulty mastering the weight shift necessary for sitting. It should be noted that use of such knee mechanisms bilaterally must be avoided. Since it is impossible for the amputee to simultaneously unload both artificial limbs, sitting with two stance control knees also becomes nearly impossible.

A third class of knee mechanisms which has proven advantageous for this level of amputation is the polycentric group (Otto Bock 3R20 or equivalent). Although slightly heavier than the previous two types, this component offers maximum stance phase stability. Because the stability is inherent in the multi-linkage design, it does not erode as the knee mechanism wears during use.

In addition, all polycentric mechanisms tend to "shorten" during swing phase, adding slightly to the toe clearance at that time. Many of the endoskeletal designs feature a readily adjustable knee extension stop. This permits significant changes to the biomechanical stability of the prosthesis, even in the definitive limb.

Because of the powerful stability, good durability, and realignment capabilities of the endoskeletal polycentric mechanisms, they are particularly well suited for the bilateral amputee.[8] All levels of amputation, up to and including bilateral hemipelvectomy (hemicorporectomy), have successfully ambulated with these components.

At first glance, a manual locking knee seems a logical choice. However, experience has shown this is rarely required, and should be reserved as a prescription of last resort. Only multiple medical disabilities (e.g. concomitant blindness) will require this mechanism. The complications in unlocking a joint for sitting by the unilateral have been discussed previously; expecting a bilateral amputee to cope with dual locking knees and dual locking hips can be an overwhelming task.

For many years, the use of fluid controlled knee mechanisms for high level amputees was considered unwarranted, since these individuals obviously walked at only one (slow) cadence. The development of the hip flexion bias mechanism and more propulsive foot designs have challenged this assumption. Furthermore, a more sophisticated understanding of the details of prosthetic locomotion has revealed an additional advantage for the hip/hemi amputee.

It is well accepted that any fluid control mechanism (hydraulic or pneumatic) results in a smoother gait.[9] Motion studies conducted at Northwestern University revealed that a more normal gait for the hip/hemi patient is also a by-product.[10]

The preferred mechanism has separate knee flexion and extension resistance adjustments. A relatively powerful flexion resistance limits heel rise and initiates forward motion of the shank more quickly. In essence, the limb steps forward more rapidly.

As the shank moves into extension, the fluid resistance at the knee transmits the momentum up the thigh segment, pushing the hip joint forward into flexion. In essence, the fluid controlled knee results in a hip flexion bias effect (Figure 5).

Sophisticated gait analyses have demonstrated that this results in significantly more normal range of motion at the hip joint during the walking cycle.[11] Clinical observations suggest that a more varied cadence is possible, and the prosthesis can usually be fabricated to nearly full length without swing phase difficulties.

Richard Lehneis, et al. have reported on a coordinated hip-knee hydraulic linkage using a modified hydrapneumatic unit.[12] This was designed to create a hip extension bias, and resulted in a smooth gait. We have no experience with this particular component at Duke.

Finally, a number of new components have been developed recently which combine the characteristics of some of the above classes of knee mechanisms. For example, Teh Lin manufactures a "Graphlite" knee consisting of a polycentric set-up with pneumatic swing phase control in a carbon fiber receptacle.

Foot Mechanisms

Traditionally, the Solid Ankle Cushion Heel (SACH) has been considered the foot of choice for the Canadian hip disarticulation design due to its light weight, low cost, and excellent durability.[13] Provided the heel durometer is very soft, knee stability with this foot has generally been quite acceptable.

In those cases where slightly more knee stability was desired, a single axis foot with a very soft plantar flexion bumper was preferred.[14] Added weight, maintenance, and cost, plus reduced cosmesis are the liabilities of this component.

Multi-axis designs (such as the Greissinger) have similar liabilities to the single axis versions, but add extra degrees of freedom via hindfoot inversion/eversion and transverse rotation. In addition to accommodating uneven ground, absorbing some of the torque of walking, and protecting the patient's skin from shear stresses, multi-motion feet seem to decrease the wear and tear on the prosthetic mechanisms as well.[15]

In the last five years, more sophisticated foot mechanisms have reached the market, and all have been demonstrated to function successfully for the high level amputee. The Solid Ankle Flexible Endoskeleton (SAFE) foot inaugurated a class that could be termed "Flexible Keel" designs.[16] Other members of this class include the STEN foot and the Otto Bock 1D10 Dynamic foot. All are characterized by a softer, more flexible forefoot, resulting in a smoother rollover for the patient. The SAFE version offers some transverse rotation as well.[17]

In general, a softer forefoot requires special care during dynamic alignment to ensure that knee buckle does not occur inadvertently. However, when used in concert with a polycentric knee, the reverse occurs: the prosthesis actually becomes safer during late stance phase.

The polycentric knee mechanism strongly resists a bending moment, which leads to its powerful stability at heel strike. It flexes during swing phase only if the forefoot remains firmly planted on the floor as the body "rides" the prosthesis over it.[18] This creates a shearing force which disrupts the linkage and permits easy flexion of the knee. Because the softer flexible keel delays this shearing moment, the polycentric knee is actually more stable in late stance than with a more rigid foot.

Dynamic Response feet, which provide a subjective sense of active push-off, can also be used to advantage for the hip/hemi amputee.[19] Carbon Copy II, Seattle foot, and Flex-Foot™ have all been successfully utilized for this type of patient. They seem to provide a more rapid cadence, as evidenced by one long-term hip disarticulation wearer, who stated after receiving a Seattle foot, "For the first time in my life, I can pass someone in a crowd."[20]

Once again, the interaction between the foot and knee must be carefully monitored. In general, the more responsive the foot mechanism, the more important the knee unit resistances become. Many practitioners prefer a fluid controlled knee, or at least one with powerful friction cells.[21] Otherwise, much of the forward momentum of the shank can be wasted as abrupt terminal impact of the knee. Presumed reductions in energy consumption have not yet been documented by scientific studies.

In addition to the foot mechanisms, several ankle components have recently reached the American market. These can be paired with most of the feet mentioned above, adding additional degrees of motion as desired. Examples include the SwePro ankle from Sweden, the Blatchford (Endolite) Multiflex ankle from England, and the recently announced Seattle ankle.

Torque absorbing units are often added to hip/hemi prostheses to reduce the shear forces transmitted to the patient and components.[22] Ideally, they are located just beneath the knee mechanism. This increases durability by placing the mechanism away from the sagittal stresses of the ankle, yet avoids the risk of introducing iatrogenic swing phase whips.

The major justification for such a component is that the high level amputee has lost three biological joints and, hence, has no way to compensate for the normal rotation of ambulation. Torque absorbers can be combined with virtually any foot available, if desired.

Finally, transverse rotation units originally developed for the Oriental world have become available. Installed above the knee mechanism, these devices permit the amputee to press a button and passively rotate the shank 90° or more for sitting comfort. They not only facilitate sitting cross-legged upon the floor, but also permit much easier entry into automobiles and other confined areas.

Knee Disarticulation Components

Although it is generally agreed that knee disarticulation offers the possibility of increased function over an above-knee amputation,[23] it clearly restricts patients' options in knee mechanisms and results in cosmetic compromises as well. For these reasons, its advisability remains hotly contested among knowledgeable surgeons and prosthetists.

Knee Mechanisms

The traditional knee mechanism for disarticulation has been the single pivot external hinges. Inherent disadvantages have been the lack of swing phase control (no friction adjustments) and rapid wear due to the small bearing surface compared to the typical 4″ long axle of the above-knee set-up. Even with the addition of a posterior "back check" to limit extension, rapid wear of the extension stops is common.

The major virtues of this design are its simplicity and low cost. It probably functions best for small children. Although the knee ball does not protrude when sitting, external hinges result in a slightly wider mediolateral configuration which some patients find objectionable. Heavy duty wearers can quickly destroy these relatively slender joints.

One manufacturer provides a yoke attachment permitting the use of a fluid-controlled cylinder with these hinges (Figure 6). This improves swing phase significantly, but long-term durability remains problematic.

The only other type of knee possible is a special polycentric design. By using longer linkage arms, the shank appears to fold back under the thigh when sitting, thus minimizing the apparent protrusion of the knee (Figure 7). Since no mechanism is alongside the knee, the mediolateral silhouette is more acceptable as well.

Several manufacturers offer the option of fluid controlled units along with the polycentric mechanism, and almost all have friction control options as well. For this reason, swing phase functioning is much better than the simple external hinge design (Figure 8).

All polycentrics offer powerful inherent stance phase control, and this group is no exception. However, because distal weightbearing dramatically simplifies the biomechanics of knee control, this feature is seldom of great value to the patient. One manufacturer offers a manual locking module as well, but this should be used only as a last resort.

One subtle problem with knee disarticulation polycentrics is that the relative "shortening" of the shank in sitting may lift the foot completely off the floor, particularly for husky individuals who are less than 5′ 6″ tall. The resulting sense of insecurity can be very disconcerting to the amputee and may result in rejection of the prosthesis.

Durability can sometimes be a problem, although it is generally better than for external hinges. Most knee disarticulation polycentrics work quite well for geriatric patients but can become increasingly problematic for extremely vigorous individuals.

In some cases, the only effective solution to chronic breakage problems is to switch to a conventional above-knee set-up. This results in protrusion of the knee ball by at least 2″, making sitting in tight spaces (such as bus seats) nearly impossible. Although the function and durability are excellent, the cosmetic liability of such malalignment is obvious to the casual observer as well.

Foot Mechanisms

Knee disarticulates can utilize all the feet and ankle options of the higher level amputee, as previously discussed. Knee stability is rarely a concern, but reducing stress on the relatively fragile knee mechanism is a concern. For that reason, the author favors flexible keel designs,

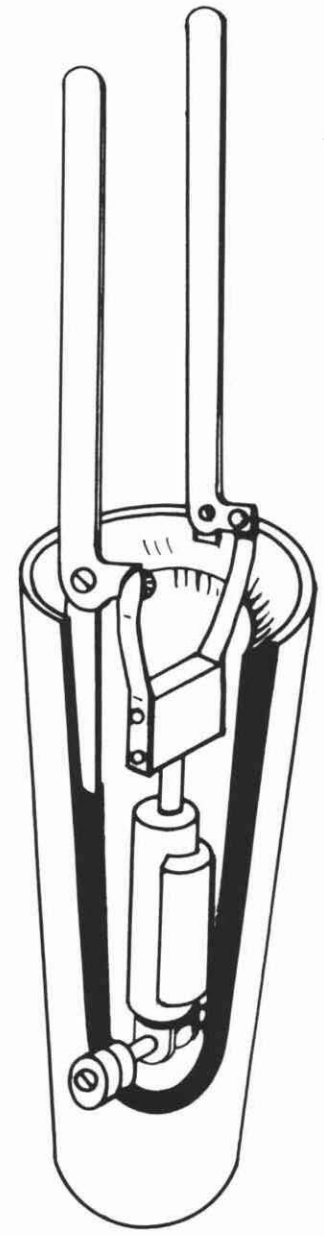


Figure 6. Cut-away drawing of special hydraulic mechanism with yoke, permitting swing phase control for knee disarticulations with single pivot external hinges. (Redrawn with permission of Hosmer-Dorrance Corporation)

with or without a torque absorbing unit, since these components reduce the forces transmitted to the limb.

Ankle Disarticulation (Symes)

Like his knee disarticulate brethren, the Symes amputee has a very limited range of choices in prosthetic componentry. In addition, a significantly poorer cosmetic result is inevitable. These disadvantages must be weighed against the functional advantages of distal weightbearing and the documented reduction in energy consumption over the below-knee amputee.[23]

Foot Mechanisms

The Symes amputation generally precludes the use of any articulated ankle mechanism, due to space limitations. The heavy metal frame of yesteryear is virtually extinct.

Most of today's Symes amputees are fitted with a SACH foot. The specially designed Symes version suffers from reduced durability due to the greater stresses the end-bearing residual limb can exert on the prosthesis. However, it can often be replaced economically if broken.

The external keel SACH design limits inversion and eversion almost completely but can be more durable and more cosmetically pleasing than the standard SACH. Since its use precludes any alteration of alignment after transfer and finishing, great care must be exercised during the fitting.

The Stationary Ankle Flexible Endoskeleton (SAFE) foot, discussed earlier, has a Symes version. This offers a flexible keel and much smoother roll-over. This reduces the forces transmitted to the prosthetic socket, increasing both patient comfort and socket durability. Reliability is adequate, and replacement is possible. The author prefers this design for Symes amputees for the reasons cited.

The Carbon Copy II has recently developed a dynamic response design suitable for many adult male Symes. Patient response has been favorable, as they sense the dynamic push-off it offers. External appearance is excellent, as is the weight reduction. Our experience at Duke

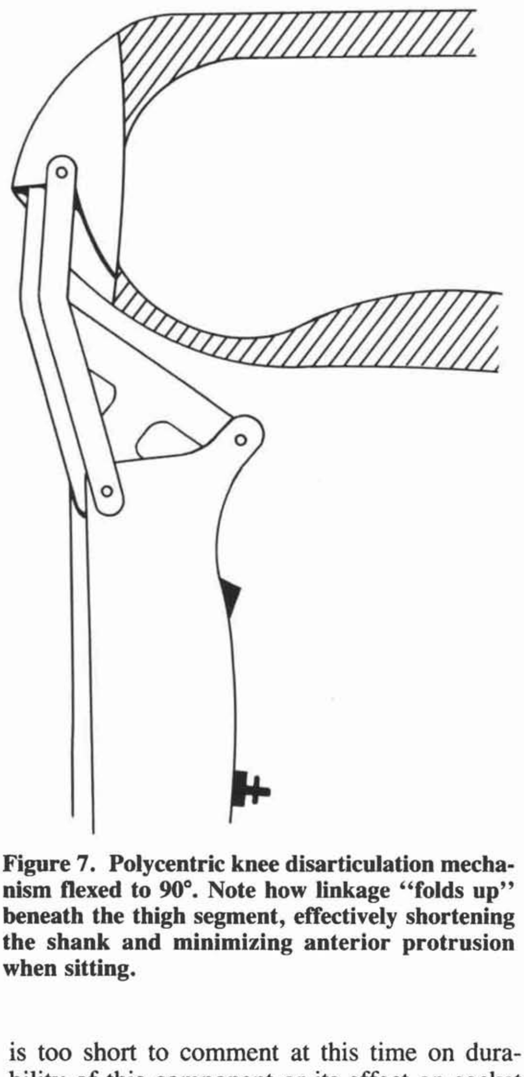


Figure 7. Polycentric knee disarticulation mechanism flexed to 90°. Note how linkage "folds up" beneath the thigh segment, effectively shortening the shank and minimizing anterior protrusion when sitting.

is too short to comment at this time on durability of this component or its effect on socket stresses.

Summary

Although disarticulations represent less than five percent of the lower limb amputees fitted annually,[25] appropriate components can be selected based on logical criteria. Both Symes and knee disarticulates, however, have limited component options, often with decreased reliability plus cosmetic limitations compared to more conventional amputation levels.

Hip disarticulates and hemipelvectomies have as broad an array of choices as the above-

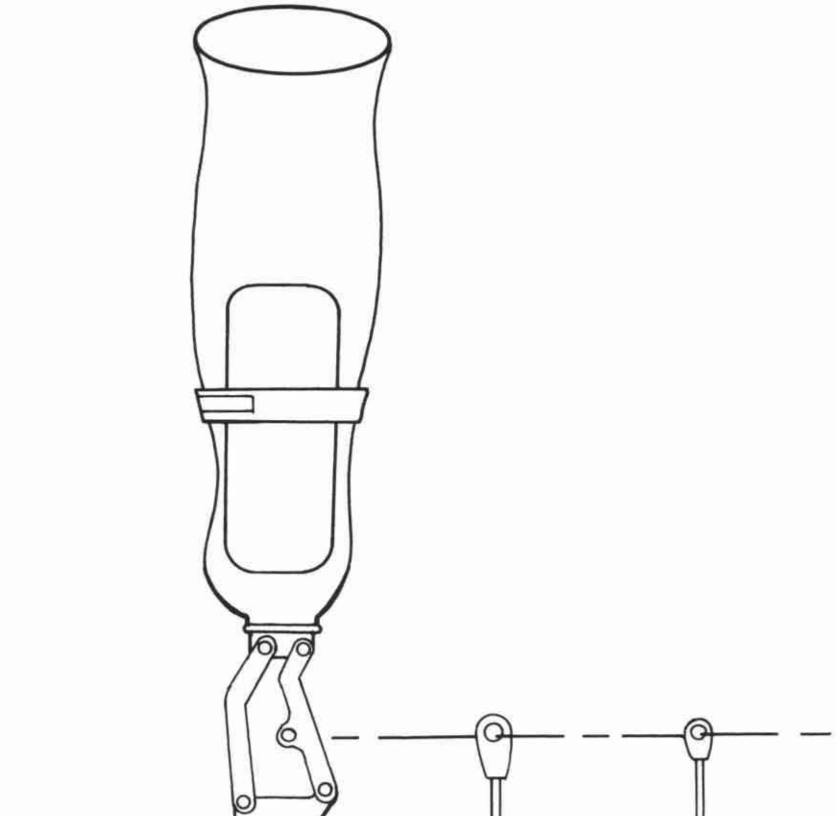


Figure 8. Example of polycentric mechanism permitting interchange of mechanical and fluid control swing phase units. (Designed by Orthopedic Hospital of Copenhagen; redrawn with permission of United States Manufacturing Company)

400 x 67

knee, prescribed for generally analogous reasons. As our understanding of biomechanics has improved, more sophisticated mechanisms have been successfully provided to this group of patients. Current state-of-the-art requires careful consideration of the subtle interactions between the foot, ankle, hip, and ancillary mechanisms to ensure the optimum result for each patient.

References

[1] Glattly, Harold W., "A Preliminary Report on the Amputee Census," *Artificial Limbs,* 7(1), 1963, pp. 5–10.

[2] McLaurin, C.A., "Hip Disarticulation Prosthesis," Report No. 15, Prosthetic Services Centre, Department of Veterans Affairs, Toronto, Canada, 1954.

[3] Webster, B. and P. Tuil, "Heupexarticulates Onder De Loep," *Infortho,* August, 1982.

[4] Van der Waarde, Tony, "Ottawa Experience with Hip Disarticulation Prostheses," *Orthotics and Prosthetics,* 38(1), 1984, pp. 29–33.

[5] Haslam, T. and M. Wilson, "Hip Flexion Bias," *Concept 80,* Medical Center Prosthetics, Houston, Texas, 1980.

[6] Ibid.

[7] McLaurin, op. cit.

[8] Greene, Michael, "Four Bar Knee Linkage Analysis," *Orthotics and Prosthetics,* 37(1), 1983, pp. 15–24.

[9] Mooney, Vert, in *Atlas of Limb Prosthetics,* C.V. Mosby Company, St. Louis, 1981, pp. 391–395.

[10] Van Vorhis, Robert, "Clinical Analysis of Hip Disarticulation Prostheses," 31st Scientific Meeting, Midwest Chapter, AAOP, Lincolnshire, Illinois, 1981.

[11] Ibid.

[12] Lehneis, H.R. et al., *Prosthetics Management for High Level Lower Limb Amputees,* Institute of Rehabilitation Medicine, New York, New York, 1980.

[13] Hampton, F., *A Hemipelvectomy Prosthesis,* Northwestern University Prosthetic Research Center, Chicago, Illinois, 1964, p. 32.

[14] Gehl, Gunter, "Proper Selection of Prosthetic Components," Northwestern University Prosthetic Orthotic Center, Certificate course, Chicago, Illinois, 1976.

[15] Ibid.

[16] Michael, John, "Analysis of Energy Storing Feet," AAOP Annual Meeting and Scientific Symposium, Newport Beach, California, 1988.

[17] Campbell, J. and C. Childs, "The S.A.F.E. Foot," *Orthotics and Prosthetics,* 34(3), 1980, pp. 3–17.

[18] Nader, Max et al., "Polycentric, Four Bar Linkage Knee Joint," *Technical Information Bulletin No. 45,* Otto Bock Industries, Duderstadt, West Germany, 1988, p. 3.

[19] Michael, op. cit.

[20] Michael, J., "Energy Storing Feet: A Clinical Comparison," *Clinical Prosthetics and Orthotics,* 11(3), 1987, pp. 154–168.

[21] Leal, J., personal communication, 1987.

[22] Nader, Max et al., "Torsion Adapter With Tube," *Technical Information Bulletin No. 2.6.1,* Otto Bock Industries, Duderstadt, West Germany, 1986, p. 3.

[23] Jensen, J.S. and T. Mandrup-Poulsen, "Success Rate of Prosthetic Fitting After Major Amputations of The Lower Limb," *Prosthetics and Orthotics International,* 7(2), 1983, pp. 119–122.

[24] Waters, R.L. et al., "Energy Costs of Walking of Amputees: The Influence of Level of Amputation," *Journal of Bone and Joint Surgery,* 58A, 1976, p. 46.

[25] Glattly, op. cit.

Author

John W. Michael, M.Ed., C.P.O., is Director and Assistant Clinical Professor at Duke University Department of Prosthetics and Orthotics, Duke University Medical Center, P.O. Box 3885, MO4 Davison, Durham, North Carolina 27710; (919) 684-2474.

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 109–113

The Hip Disarticulation Prosthesis as Developed by the O.I.M. Noord Nederland

by Peter Tuil

What characterizes the hip disarticulation prosthesis of the O.I.M. Noord Nederland is the use of a four-bar Otto Bock knee joint as a hip joint. O.I.M. Noord Nederland has used this variation with much success over the last five years. At first, it was questionable whether the joint would be strong enough, but this has proven not to be an issue. There have been some problems with the 3R21, but only when it is used as a knee joint. These complications have been due to extreme flexion, lamination sections that were too thick and caused the joint to tear apart during flexion, or too much external rotation.

There are two advantages in the use of the four-bar hip joint. First, the patient walks with a lower energy expenditure because the prosthesis shortens the swing phase. In contrast to patients who have worn older style hip disarticulation prostheses (for years patients used to be fitted with a tilting-table prosthesis or later with a wooden "Canadian hip" prosthesis), the patients with the new style prosthesis walk more and have indicated that they use less energy. Second, there is hardly any strain on the cosmetic cover, so much less damage is done.

An additional advantage of the four-bar joint is that the construction can be less critical. Besides, the whole prosthesis can be readily adjusted.

Description of the Fabrication Method

To make the plaster impression, two wooden blocks are mounted on a table or casting stand. (Editorial note: Presumably this stand is adjustable in height.)

These wooden blocks have sloped planes so that a wedge-formed gap is created between them (Figure 1). In the back, the sloped side forms a 60° angle. In the front, the sloped side is divided into two different angles (Figure 2). Both blocks can rotate around their vertical

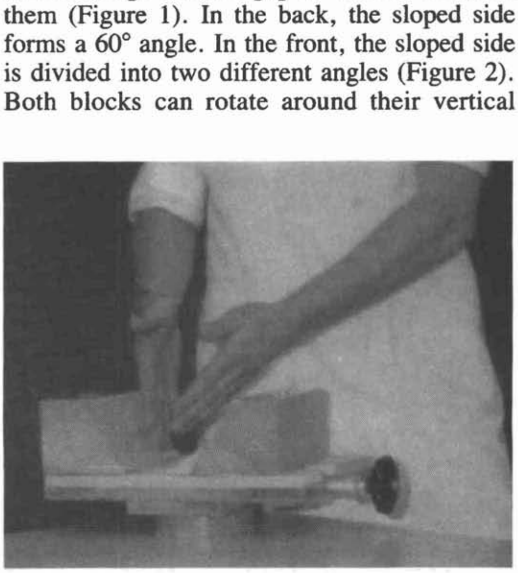


Figure 1. Apparatus used for casting.

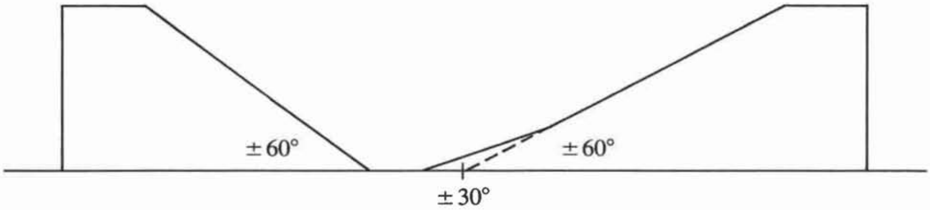


Figure 2. Side view of the wooden blocks.

Figure 3. The adaptor.

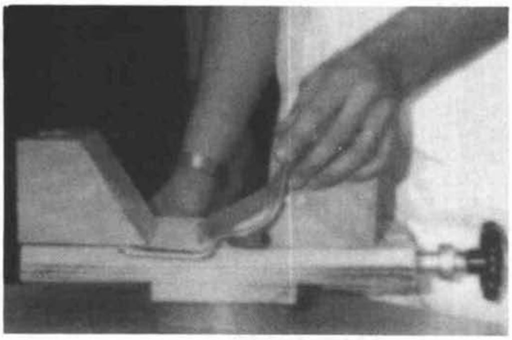


Figure 4. Shows how the adaptor, which will later be laminated into the socket, relates to the wedged form of the wooden blocks.

axles with regard to the table to which they are attached. They can also be shifted with regard to each other in the sagittal plane by means of a spindle (worm gear mechanism). The blocks are primarily meant to provide a good fitting of the residual limb and pressure relief in the places where that is necessary.

The four-bar joint is attached to the socket by means of a specially manufactured adaptor (Figures 3 and 4). The adaptor, which will later be incorporated into the socket, mimics the wedged shape of the wooden blocks (Figure 5).

Finally, the impression of the wooden table provides a good plane of reference for the plaster model (Figure 6).

This impression of the horizontal plane must remain horizontal during the construction process. During plaster modification, one should maintain unchanged the medial of the plaster model in the transverse plane, so that the impression of this edge will always indicate the line of progression of the plaster model.

The socket is laminated in three layers. First though, a layer of Pe-Lite™ is put on the plaster model, followed by a layer of stockinette, and finally a layer of P.V.A. foil. The layer of stockinette is always applied under the first layer of foil. This will provide better suction, absorb some moisture, and the plaster model need not be as smooth.

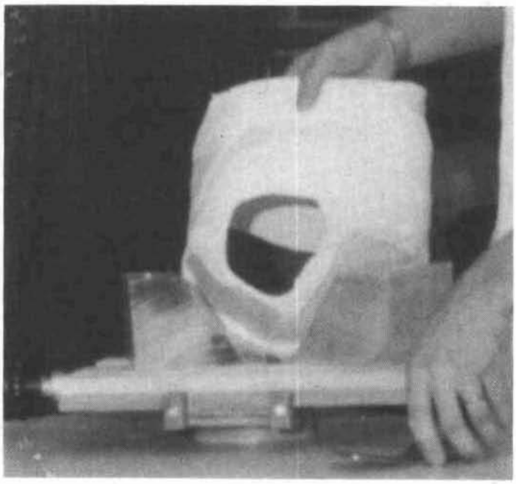


Figure 5. The apparatus forms a good plane of reference.

The first layer is laminated from flexible resin with two layers of Perlon stockinette,

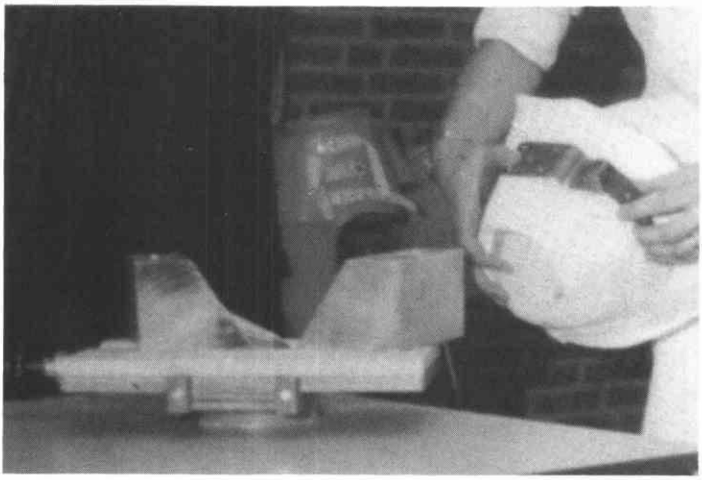


Figure 6. Position of the adaptor as related to the pelvic socket.

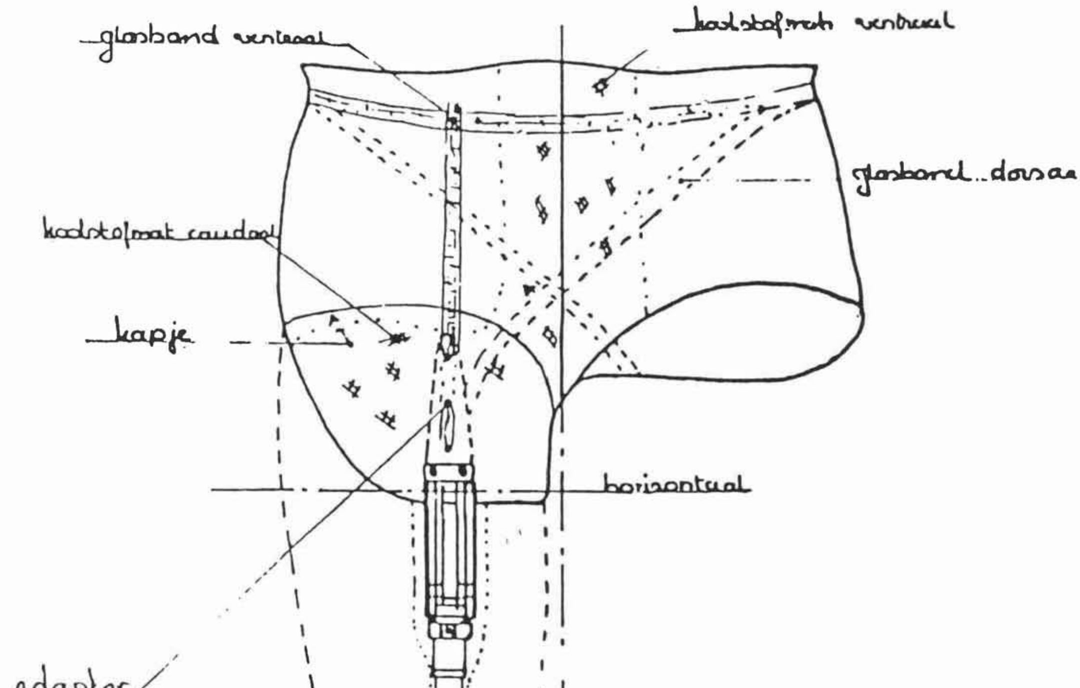


Figure 7. Reinforcement of the socket.

which is elastic in two directions. Subsequently the adaptor is located as shown in Figure 7.

The space between the adaptor and the plaster model is filled with "leichtspatel" (filler). The base of the plaster model must stand horizontally. The adaptor is placed approximately 4 to 5cm lateral of the groin. The maxim is to get the adaptor directly underneath the ischial tuberosity. However, this is influenced by the needs of the cosmetic cover.

The adaptor is then covered with two layers of stockinette and a reinforcing layer of carbon fiber matting to prevent the adaptor breaking loose from the forces generated at heel strike. A strip of carbon fiber is put in the front to prevent the pelvis socket from curling inward. A reinforcing band of glass fiber is placed diagonally as shown in Figure 8. Over this, two layers of stockinette are placed. First, rigid lamination resin is applied on those areas where the socket must be rigid. The rest is laminated with flexible resin. An adjustable "jig" is necessary in order to be able to turn the model around in the bench-vice quickly. The final layer is done with flexible resin and two layers of stockinette.

A layer of stockinette and P.V.A. foil are put on the socket. Then, the little cap needed to finish the cosmetic cover is laminated with three or four layers of stockinette and one layer of carbon fiber. The extra time needed to form this cap will later save a lot of time during the finishing of the cosmetic cover.

The prosthesis is completed with a four-bar knee joint (3R21), a single axis ankle joint foot, and a rotation adaptor.

The alignment of the prosthesis is first considered in the sitting position. One must take into account the symmetry in comparison to the healthy limb (Figures 9 and 10). The definitive alignment is settled upon during stance and walking exercises (Figure 10). The adjustment of the 3R21 knee joint is very important. Mistakes in alignment can cause malfunctions of the knee joint. Many adjustments are possible with regard to rotation in the hip joint itself. The lack of facility to adjust abduction has never been a problem.

The freedom of movement when seated is considerable (Figures 11 and 12).

The cosmetic cover is shaped in the hip area, as well as in the knee area, so that less tension will be induced in the cover during flexion and when seated. Finally, a long elastic strip is glued to the inner anterior wall of the cover. This is done to protect the foam-cover.

The construction process for a prosthesis for a hemipelvectomy is similar.

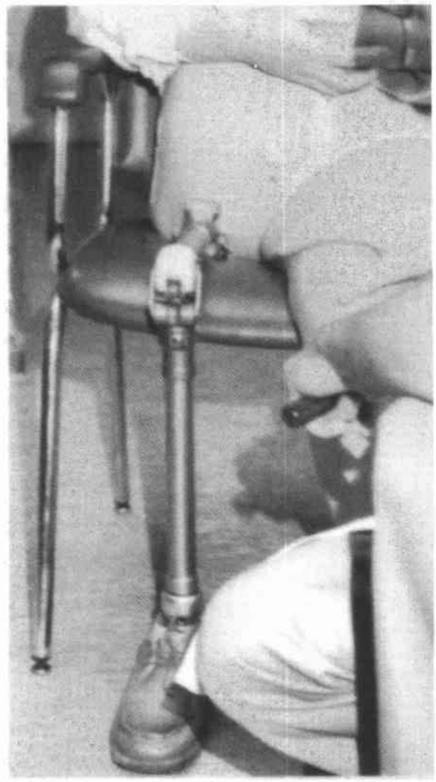
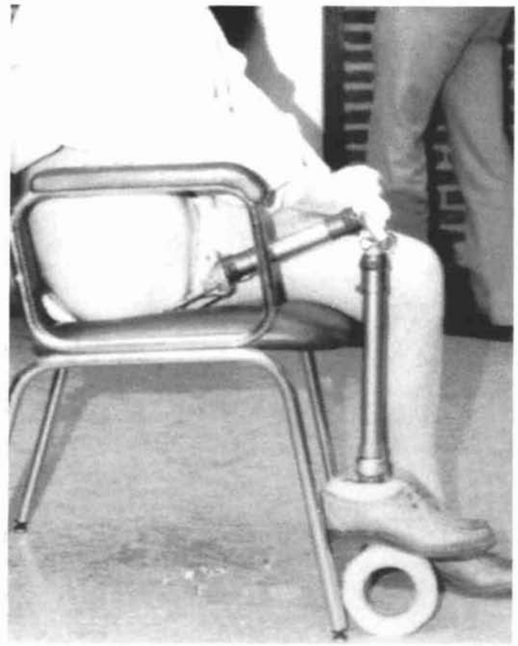


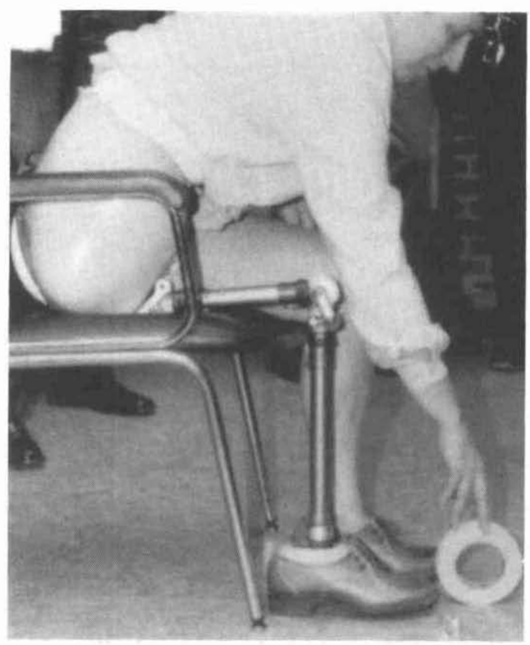
Figure 8. Alignment is first considered with the patient seated.

Figure 9. Side view of the patient sitting and wearing the prosthesis.

Figure 10. The realization of definitive alignment.

Figure 11 (right) and 12 (below). Freedom of movement when seated.





Author

Peter Tuil can be contacted at Stitchting Orthopedische Instrumentmakerij, Noord-Nederland Dilgtweg 5, 9751 ND Haren (Gn), The Netherlands.

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 114–118

The UCLA Anatomical Hip Disarticulation Prosthesis

by David H. Littig, B.A., C.P.
Judd E. Lundt, B.S., A.E.

This article discusses a new approach to hip disarticulation and hemipelvectomy fittings developed at the UCLA Prosthetics Education Program. It employs fundamental principles and methods in a new and different combination to produce a more complex and more natural biomechanical system. The results include:

- a smoother, and apparently less energy consuming gait
- improved stability
- improved suspension
- improved wearer comfort

Introduction

For several years now, the UCLA Prosthetics Education Program has been involved in an effort to understand, develop, and refine a teaching method for the CAT-CAM Socket,[1] the anatomically shaped above-knee socket. The very essence of this effort is a broader and more detailed understanding of the pelvic anatomy and its optimal containment within the socket. With the dramatic above-knee results that have been achieved through this understanding has come a compelling and obvious need to examine the application of the same principles to hip disarticulation fittings.

It was felt that a hip disarticulation socket design, which would encapsulate the ischium and ischial ramus in a more anatomical contour than previous socket designs, might produce an improved prosthetic fitting.[2] Since much of the CAT-CAM experience alluded to employed a frame supported flexible polyethylene socket, the flexibility of such a design applied to a hip socket seemed a reasonable way to provide more comfort. These factors formed the basis for this work.

To date, three hip disarticulation patients and one complete hemipelvectomy patient have been successfully fit with the design described. The hemipelvectomy application followed the hip patient fittings by a number of months and was tried only as a whimsical experiment. Based upon the initial understanding of the biomechanics, this fitting was not expected to succeed. However, the results were quite surprising and motivated another look at the biomechanical analysis. Two of the hip cases and the hemipelvectomy case will be described in this article along with the biomechanics.

Patient Experience

The first hip disarticulation patient is a 23 year old male who had an amputation on the right side at age five for tumor, and who has rejected a prosthesis since age ten because it was too limiting and cumbersome. Owing to immature muscular and skeletal development at the time of amputation, he is significantly atrophied on the amputated side. This individual is extremely active, participates and excels in athletics as an equal with the able-bodied, and is impressively agile on crutches. Consequently, his remaining limb is hyperdeveloped to the extent that the thigh musculature extends well past the midline of the body.

The second hip case, a right amputee as

Figure 1. Posterior view of suspension system showing "X" pattern strapping.

Figure 2. Anterior view of suspension system.

well, could be described as a more typical patient. He is a 40 year old professional, amputated at age 28, also due to a tumor. He has worn a prosthesis continuously since his amputation, the most recent being an Otto Bock endoskeletal design. When this project was begun, he had recently taken delivery of a new one-piece flexible socket prosthesis which combined a Flex-Foot™ with Otto Bock endoskeletal knee and hip components.

The hemipelvectomy case was a 26 year old male who had undergone complete amputation on the left side for a massive tumor in the hip joint. At the time of the work described here, which was six months post-surgery, he had not yet been fit for a definitive prosthesis, but was wearing a socket only for sitting comfort. This patient was first seen as a demonstration subject for prosthetic certificate students at UCLA. For that program, he was fit with a fairly conventional design. However, since his level of amputation is somewhat uncommon, and because the patient was willing to experiment, the design was altered to include the suspension system that had been found to be so successful with the hip disarticulation patients.

Fabrication

The hip patients were cast using a similar technique with splints, circular wrap, iliac crest definition, anterior and posterior compression, and ischial weight-bearing while the plaster hardened. Since this was an attempt at a more anatomical socket, contours detailing the ischial ramus angle and the medial inclination of the ischium were included in the cast. Unlike the above-knee socket which flexes and extends with the femur through each stride, the hip socket is expected to remain relatively fixed, relative to the pelvic anatomy. Thus, the medial brim need not extend as high or contain as much of the ischial ramus. If properly executed, a cast which includes the bony contours of the pelvis will take much of the guesswork out of cast modification and fitting and should reduce the number of check sockets needed to attain an optimum result.

The initial concept for a hip disarticulation socket was a one-piece polyethylene design with a laminated frame to which the hip joint would be attached. Accordingly, such a system was fabricated for the first fitting. The results were reasonably successful. However, sound side comfort and piston action of the resulting prosthesis were not wholly satisfying. Since this was an experiment, it was decided to push further. Through several reiterations of socket size, volume and shape, and through several experiments with suspension, the design described in this article was arrived at: a two-piece system composed of a laminated anatomically shaped socket, encompassing only the amputated side and connected to a polyethylene suspension segment for the sound side waist with Dacron® webbing (Figure 1).

Fabrication of the socket for the hip patients was relatively simple because of the two-piece design. The original model was split and only the amputated side laminated. For the contralateral side, a shell of Aliplast®-lined polyethylene was vacuum formed over that half to serve as the suspension system.

Since the initial fitting of the hemipelvectomy patient was meant to instruct the certificate students in basic prosthetics for this level of amputation, the approach was very straightforward. He was cast in a suspended attitude with a simple circular wrap. Modification involved little more than smoothing of the model. A total flexible polyethylene socket was vacuum formed over the entire model. Following this, a frame for mounting the hip joint was laminated over the amputated side only of the polyethylene socket. As this was a demonstration fitting with no intent to finish, the hip joint was only temporarily attached.

Prostheses for all patients were assembled with Otto Bock endoskeletal 7E7 hip joints and 3R20/3R36 knee units. Several feet were experimented with on the first patient until an Otto Bock single axis foot proved optimum. The Flex-Foot™ that had been included with the second patient's recently delivered prosthesis was incorporated into his set-up. The hemipelvectomy patient was also fit with an Otto Bock single axis foot.

The socket and sound side suspension segment for the hip disarticulations were joined posteriorally with Dacron® webbing. Using temporarily attached four-bar buckles and the webbing, the proximal and distal aspects of the socket and the polyethylene segment were connected with the webbing to form an "X" pattern across the posterior gap (Figure 1). At their cross point, the straps are not connected but are allowed to freely move with respect to each other. The buckles were found to be necessary for "fine tuning" adjustments of the suspension during fitting and alignment. Anteriorly, a single strap attached at the distal aspect of the laminated socket was passed through a loop on the polyethylene portion and back to a roller buckle on the anterior proximal socket (Figure 2).

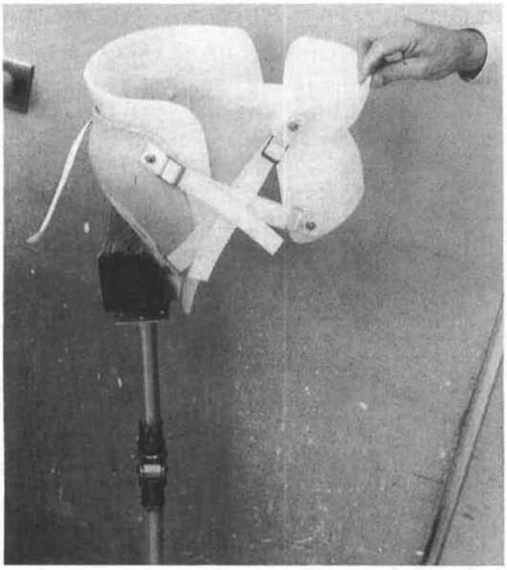


Figure 3. Posterior view of hemipelvectomy set-up showing adjustable cross strapping.

Functional Results

Results achieved with this combination of socket and suspension were dramatic. After some adjustments, the hip patients felt no discomfort from the socket, despite the obvious upward curve of the medial brim in the perineum. This edge, along with the distal portion of the socket, particularly under the ischium, were lightly padded with ⅛" Pe-Lite™, as is customary in most hip sockets. Neither patient perceived any piston action or discomfort from the proximal brim of the socket or the polyethylene waist segment. The most obvious benefit was a significant reduction in lateral trunk bending that is so common with hip disarticulation amputees. In fact, this gait anomaly was reduced beyond that usually seen with many above-knee amputees. Both patients were impressed with the comfort and secure feeling that the design afforded.

With the adjustable diagonal posterior straps, which in the finished prosthesis are replaced with buckleless double Dacron® webbing, the socket can be optimally positioned under the pelvis to more effectively encapsulate the bony pelvic anatomy. This is somewhat akin to adducting an above-knee socket of similar medial brim design (CAT-CAM). By a careful balance in the strap length adjustments, comfort and suspension in the entire system can be achieved.

Because of the success that had been achieved with the hip disarticulation patients with this suspension technique, it was decided to try it on the hemipelvectomy patient. His reasonably comfortable and functional single piece socket was modified by removing the center portion of polyethylene in the posterior and rejoining the two separate segments with Dacron® webbing in the same cross strap pattern. An anterior closure as previously described was also employed (Figure 3). The result was about ⅛″ of piston action and improved comfort over the one-piece design, probably because the prosthetic socket could more accurately follow the body contours.

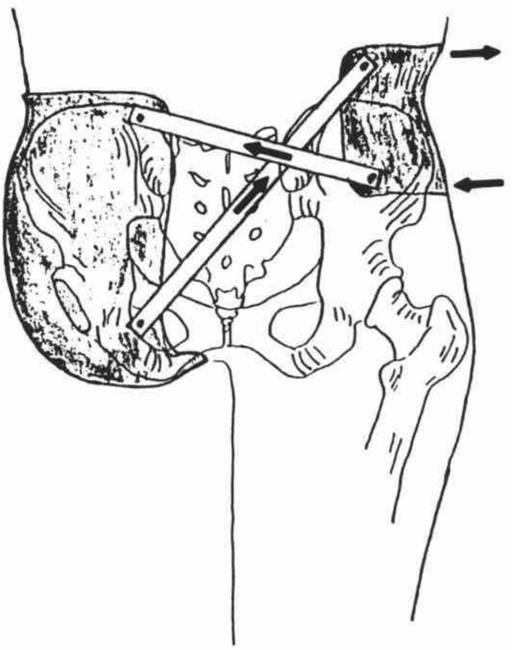


Figure 4. Suspension system forces at mid-stance.

Biomechanics

In all cases, it appears that during the gait cycle the polyethylene segment that encompasses the contralateral hip will tilt from the vertical as it follows the changing sound side body contour. The forces thus imposed on each of the posterior straps will vary alternately, and their crosspoint will shift slightly with each stride. For example, as the amputee reaches heel strike on the prosthesis, tension in the strap originating at the posterior proximal socket (the lateral support strap) will build as the body moves forward, and the center of gravity begins to shift laterally. As the patient progresses, this force reaches its maximum at mid-stance (Figure 4) and then begins to fall off. Tension in the other (suspension strap) is at its lowest at mid-stance on the prosthetic side and then begins to build toward its peak when the amputee reaches mid swing-through (Figure 5). The cycle then repeats itself with each successive stride. This alternating action in the straps, coupled with an accurately contoured socket, provides a continuously snug and secure suspension without the need for excessive tightness.

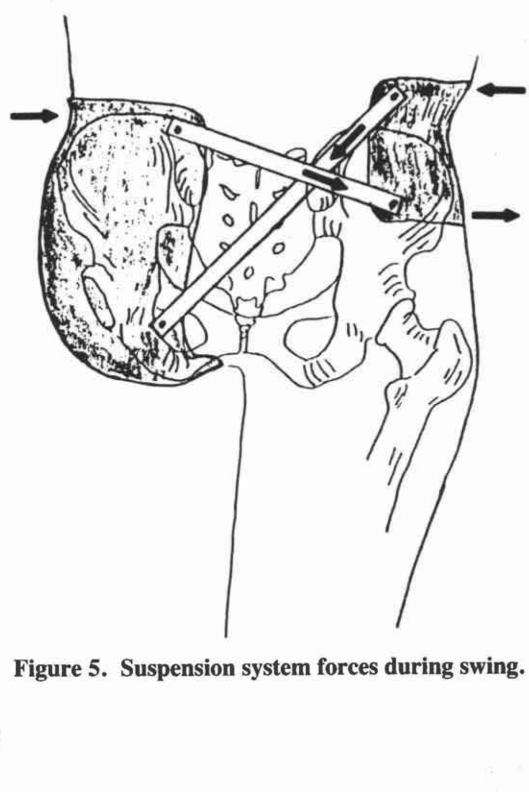


Figure 5. Suspension system forces during swing.

At the outset of these efforts, it was believed that much of the success of the suspension system depended upon a well-contoured medial brim, which accurately encapsulated the ischium and ischial ramus. The hemipelvectomy fitting quickly dispelled this consideration as a major factor. However, all hip disarticulation patients fit to date have perceived far greater comfort and control when in a socket so described. The idea behind ischial containment is to provide greater mediolateral stability in the prosthesis. It appears that the cross strap suspension is contributing the better part of this stability.

Results to date suggest that the two-part socket and posterior cross strapping provide a mechanism which more closely conforms to changing soft tissue and muscle contours through the gait cycle. With a one-piece socket, regardless of flexibility, slight and subtle motions about all three body axes are not fully accommodated by "give" in the socket, as well as they seem to be in the one described here. Thus, the body must either move inside the socket or limit its movements due to the restrictions imposed by the rigidity of the socket. In either case, the result is a less natural gait and a greater apparent expenditure of energy. With this new approach, these shortcomings of the hip and hemipelvectomy fittings seem to be significantly reduced.

References

[1] Sabolich, John, "Contoured Adducted Trocanteric Controlled Alignment Method (CAT-CAM): Introduction and Basic Principles," *Clinical Prosthetics and Orthotics*, Vol. 9, No. 4, Fall, 1985, pp. 15–26.

[2] Sabolich, John and Tom Guth, "CAT-CAM Innovations," *Ability*, Vol. 6, No. 3, Winter, 1986, p. 48.

Authors

David Littig, B.A., C.P., and Judd E. Lundt, B.S., A.E., can be contacted through the UCLA Prosthetics Education Program, Rehabilitation Center, 1000 Veteran Avenue, Room 22-41, Los Angeles, California 90024.

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 119–122

The CAT-CAM-H.D.™ A New Design for Hip Disarticulation Patients

by John Sabolich, B.S., C.P.O.
Thomas Guth, B.A., C.P.O.

The innovative features of the CAT-CAM™ above-knee socket design were outlined in the Fall, 1985 issue of *Clinical Prosthetics and Orthotics,* Volume 9, Number 4. Shortly afterwards, RGP of San Diego and the Sabolich Prosthetic Research Center in Oklahoma City combined efforts to develop a CAT-CAM™ type hip disarticulation prosthetic socket design. It was intended that this new socket would hold the ischial tuberosity and descending ramus in a special compartment of the socket. RGP worked primarily on the suspension system, and Sabolich worked on the ischial ramus containment.

The conventional hip disarticulation socket differs from the CAT-CAM™ type in that the old design has a flat inferior floor upon which the ischial tuberosity sits. Even worse, many times the tuberosity sits on the very edge of this table. As described in the original 1985 CAT-CAM™ article and in terms of the above-knee socket, this is not a desirable biomechanical situation because, first, the bone is touching a flat tangential surface rather than a contoured surface that conforms to the complex bony shape and thus distributes the load over a wider area and, second, because it does not provide medial-lateral stability. The new socket affords much more bony contact not only to the ischial tuberosity, but to the descending pubic ramus as well (Figures 1 and 2). Experience has shown that the ramus turns out to be of more importance than the ischial tuberosity when it comes to enhancing medial-lateral and rotational stability. Only the inferior pubis-ramus is allowed to exit the socket at the medial inferior dip of the medial wall (Figure 3).

In order to better understand the new hip disarticulation design, it must first be understood that the CAT-CAM™ above-knee design is not a narrow ML socket at the proximal portion. On the contrary, the proximal ML diameter of the CAT-CAM™ above-knee socket, which contains the pelvic bones, is wider than the mid and distal portions of the socket, which then narrows to conform to the medial-lateral thigh dimension in order to supply soft tissue compression. The new hip disarticulation socket follows this SCAT-CAM™ principle. Thus, it provides a better bony locking effect. Also, these bony pelvic structures are more fully encapsulated as a result of a V-shaped medial contouring of the socket and provide the hip disarticulation patient with a feeling akin to the above-knee socket, rather than that which results simply from sitting on a flat hard seat.

Some of the principles of the CAT-CAM™ total flexible brim are also utilized in this type of hip socket. The entire socket is flexible except in the area where the hip joint is attached. This can be accomplished in two ways: first, with a rigid frame and a flexible inner socket much like with the CAT-CAM™ and SCAT-CAM™ above-knee design; second, by a heterogeneous monolithic polyester socket that is rigid in the joint area and then gradually becomes flexible throughout the remainder of the socket (Figures 4 and 5).

Like the SCAT-CAM™ design, the hip socket is more bone and muscle contoured than the traditional bucket shaped hip disarticulation design (Figure 6). The new socket has a concave contour in the area of the ilium on the amputated side. On the contralateral side, there is

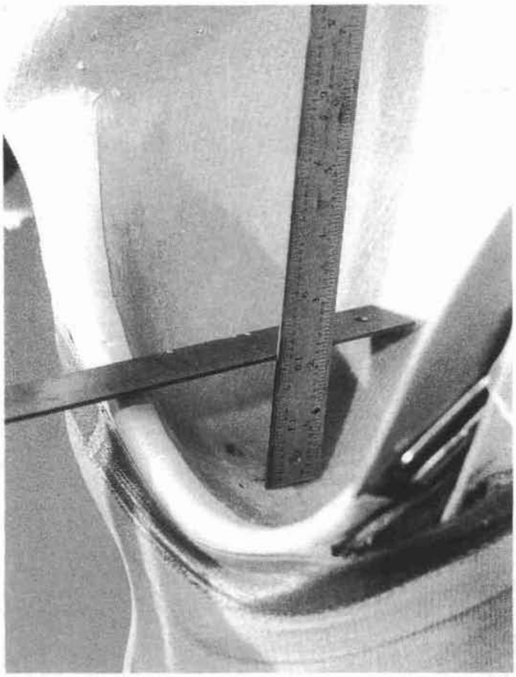


Figure 1. Demonstrates depth of ischial seat area relative to medial brim. Also shows how the ischium and ramus are in the socket.

a concave contour between the ilium and trochanter. This increases medial-lateral stability and results in improved gait when combined with the containment of the ilium, ischium, and ramus bones within the socket. This is contrasted to most conventional designs which bulge out and follow the flow of the soft tissue on both lateral sides of the socket rather than conforming to the body contours.

The "Inter Ilio Trochanteric Effect"† is one of the reasons it has been possible to suspend the socket in most cases without extending it above the iliac crests of the pelvis. Instead, the suspension is gained by conforming the socket into the notch between the ilium and trochanter and creating a counter pressure with the opposite concave shaped side of the socket. Of course, it is more difficult to suspend the socket in this manner when fitting heavy people with excessive adipose tissue.

Normally with a conventional hip disarticulation, it is easy for a prosthetist to pull the prosthesis off the patient by sliding it into abduction, away from contact with the residual

†See acknowledgments.

Figure 2. Postero-medial view of transparent diagnostic test socket on the patient with a patch of white paper delineating the ischial-ramus compartment.

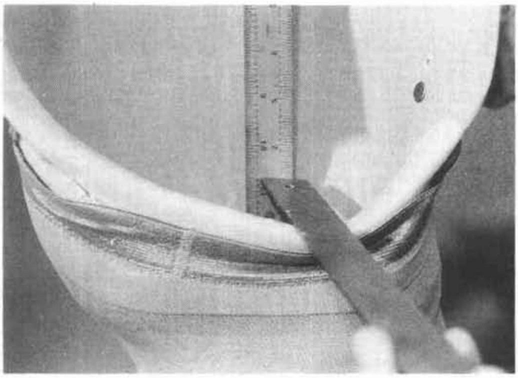


Figure 3. Medial view with rulers at the inferiormost point in the dip of the medial brim.

Figure 4. Laminated socket demonstrating flexibility of the contralateral portion of the socket. Superior portion of amputated side is flexible as well.

limb and the ischial tuberosity, when the prosthetic pylon is abducted off the floor. However, with the CAT-CAM-H.D.™ design, this maneuver is more difficult, and the socket resists this abduction tendency due to the bony lock about the ramus (Figure 7).

In the last four years, a combined number of 67 CAT-CAM hip disarticulation sockets have

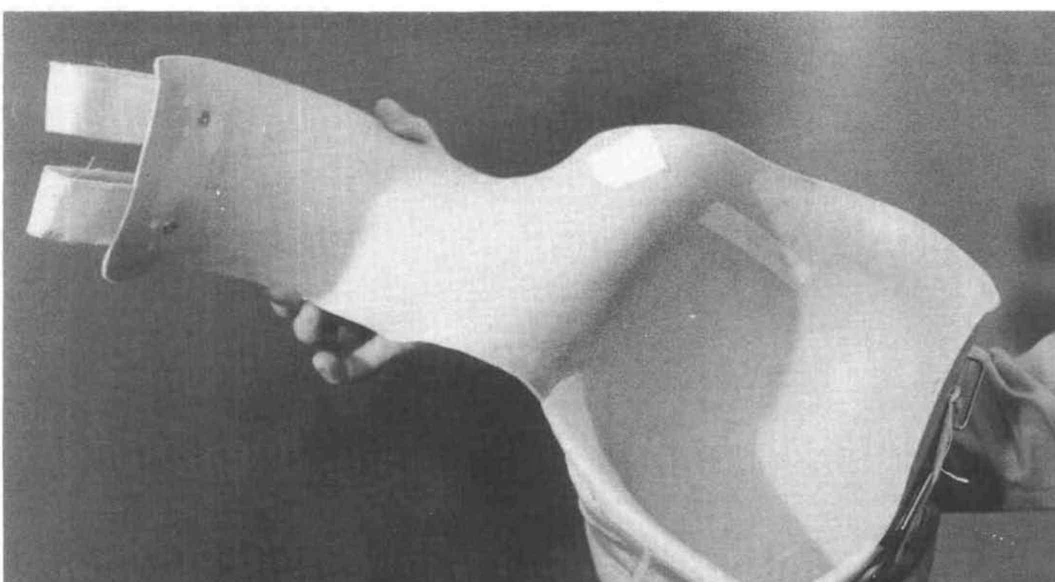


Figure 5. View similar to Figure 4 showing flexibility of socket. Also shows "V"-shaped contour of medial brim in sagittal plane.

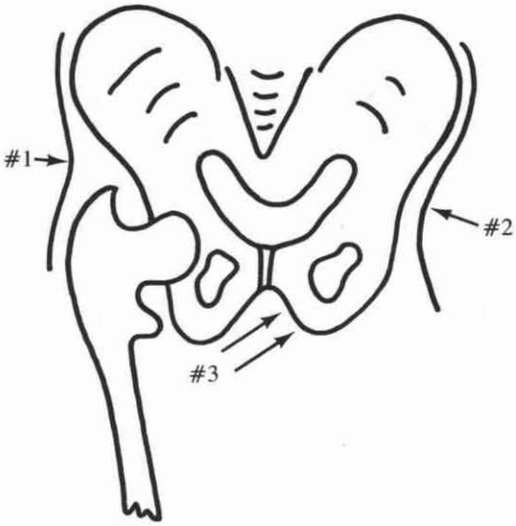


Figure 6. Schematic cross-section through the frontal plane. Vectors 1 and 2 demonstrate the suspension principle and also refer to the dark lines which represent the socket walls. Notice how the superior edges of the socket do not come above the ilium crests and the concave contouring inferior to the illiae. Vector 3 refers to the bony lock.

been fit in Oklahoma City and San Diego. These patients report that they do not feel like they are "sloshing around in a bucket" and have a "greater feeling of security and stability" (Figure 8). Three of these patients can run with their new prosthesis in a hop, skip fashion which has been recorded during video gait analysis. Two patients have been able to manage limited step over step running.

Acknowledgments

It should be noted that Mike Wilson, C.P.O., was the first person who suggested to me the principles of lateral pressure between the ilium and the trochanter on the contralateral side. He called it an "Inter Ilio Trochanter Effect."

Appreciation is given to Don Landis, B.S., R.P.T., for his editorial help in preparing this manuscript.

Appreciation is also given to Glenn Hutnick, C.P.O., and Alan Finnieston, C.P., who will be contributing to the next phase of research in hip disarticulation designs.

Authors

John Sabolich, B.S., C.P.O., is President of Sabolich Prosthetic and Research Center, 1017 N.W. 10th Street, Oklahoma City, Oklahoma, 73106.

Thomas Guth, M.D., C.P.O., is with RGP Orthopedic Appliance Co., 6147 University Avenue, San Diego, California 92115.

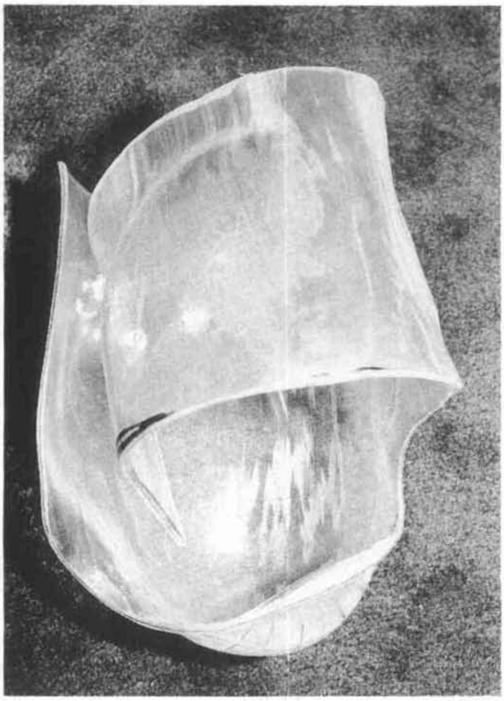


Figure 7. Medial view of the transparent diagnostic test socket showing height of medial brim relative to inferior most portion of the socket.

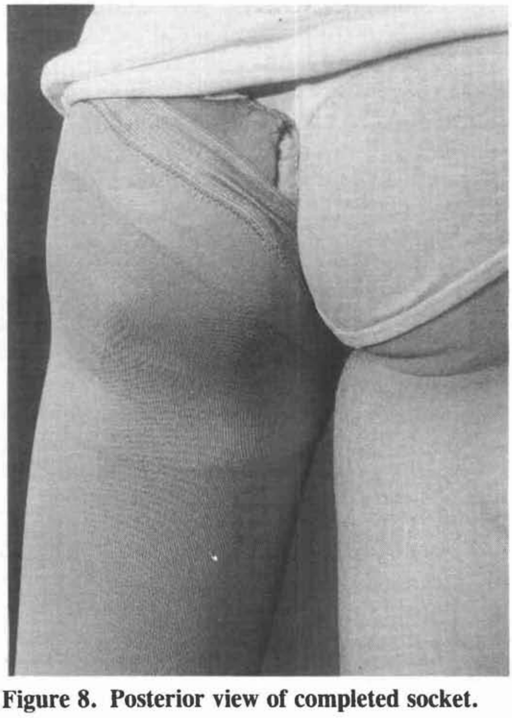


Figure 8. Posterior view of completed socket.

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 123–127

Technical Note: Cosmesis and the Knee Disarticulation Prosthesis

by Robert Gilley, C.P.

One central problem confronts the prosthetist when fabricating a knee disarticulation prosthesis with side joints, regardless of the socket style (laminated, molded, leather, or flexible frame) used. As the knee flexes, the space anteriorally between the thigh section and the inner edge of the shin increases in width (Figure 1). The result is cosmetically unacceptable. The gap develops because the radial distance from the knee joint center to the periphery of the socket gradually decreases from anterior to posterior (Figure 2). Resolution of the problem may be achieved by building up the distal end of the socket so as to maintain a constant spherical shape through the full range of motion (Figure 3). Observation over the years has led me to conclude that many younger prosthetists are not as familiar with the process as perhaps they should be. Therefore, these few notes are offered in hopes of redressing the situation.

Fabrication of the prosthesis begins by mounting the proximal portion of both joints to the thigh, and similarly, mounting the distal portions in a suitably sized block of wood. Obviously, every effort should be made to maintain the narrowest possible medial-lateral diameter at the knee joint center and to keep the joints square. Wood is removed from between the distal joint sections to permit proper mating of the shin block to the thigh and full range of motion. The interior inner edge of the shin block should fit closely to the thigh in the fully extended position, while allowing sufficient space distal to the edge for the material to be added to the thigh during finishing.

The distal end of the thigh is then built-up with rigid urethane foam. The convex shape of the anterior socket from medial to lateral, and at the level of the proximal anterior edge of the shin block, is repeated radially about the knee joint axis from anterior to posterior. This can be accomplished, rather laboriously, by first removing enough extra material from the thigh to permit it to be assembled with the shin block in the fully extended position. Then, as the knee is gradually flexed through the full range of motion, the anterior edge of the shin is used as a guide to judge how much material to remove at each successive position of flexion. Sufficient material is removed to permit full range of flexion. Care must be taken to maintain a smooth, even surface from medial to lateral and to not remove too much material. Nonetheless, it will doubtlessly be necessary to add material.

The posterior surface of the distal thigh is finished off flat from medial to lateral, so as to fill most of the posterior knee opening. It should not rise above the anterior rim of the shin in the fully flexed position, and at the same time, should not protrude too far posteriorly when in the fully extended position.

The process can be greatly expedited if the following simple apparatus is used. Two aluminum plates are modified so that a piece of stiff paper or cardboard can be clamped between them (Figures 4 and 5). The plates are cut away on one edge so as to span the largest socket. At the leading edge of the cut away side, two threaded rods with tapered points are

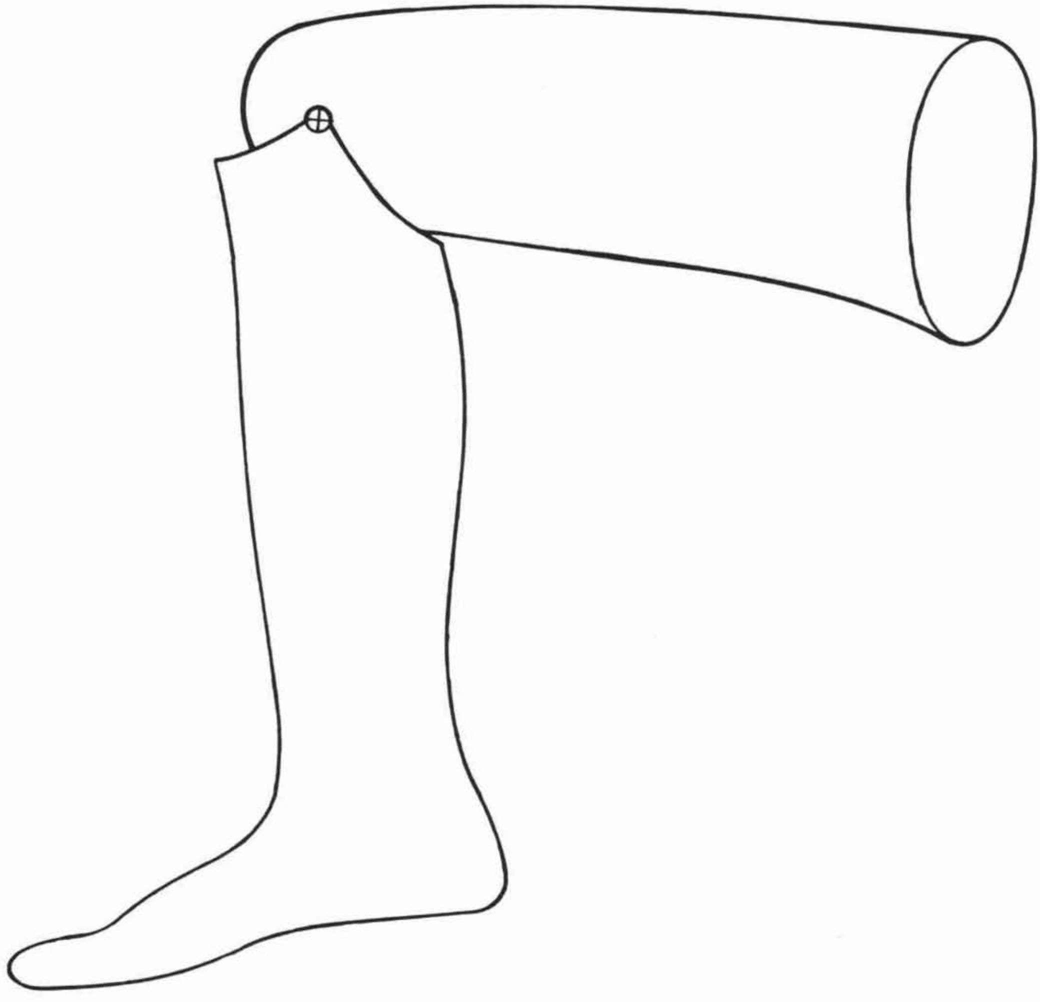


Figure 1. As the knee flexes, the gap anteriorally increases.

mounted on a common axis (Figure 6). These two rods permit the device to be mounted on the proximal knee joints and swung around the distal end of the thigh section (Figure 7). The stiff paper clamped between the two plates of the device is cut to match the shape of the anterior surface of the thigh at the requisite level. The resulting template is then used to duplicate the shape through the full range of motion. (Some prosthetists will of course identify the device as a simple adaptation of the templates that were formerly used when shaping the ball of the knee of a handmade knee-shin set-up for an above-knee prosthesis.) The device described has been in use by us now for over a year. It greatly speeds the process of finishing a knee disarticulation prosthesis.

In conclusion, it is hoped that these comments on the matter will aid a prosthetist confronted for the first time with the task of finishing a knee disarticulation prosthesis—a task that is rather infrequently confronted in the United States and not always addressed by the schools.

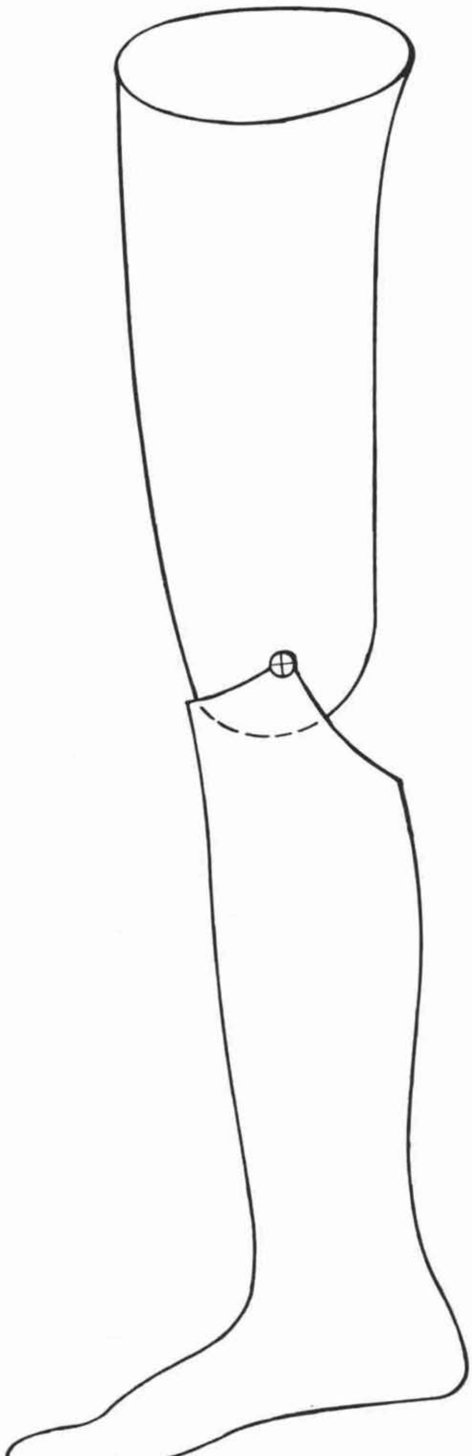


Figure 2 (left). This problem results from the fact that the radial distance from the knee joint center to the periphery of the thigh decreases from anterior to posterior.

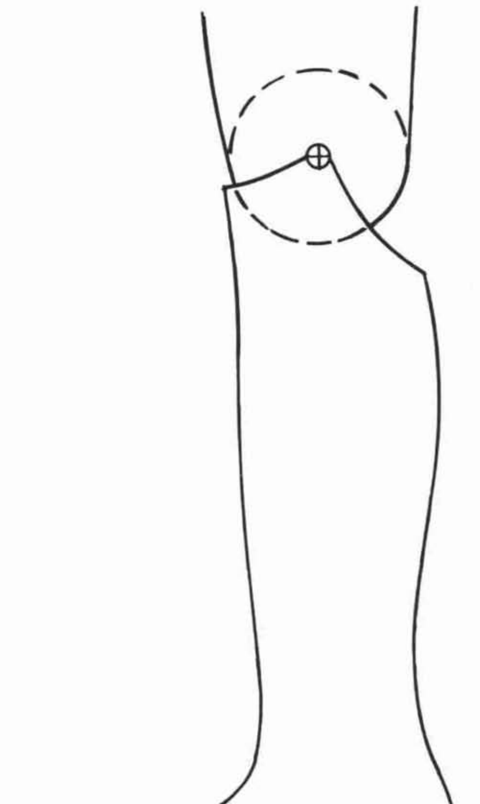


Figure 3. The distal end of the thigh has been built up so that the space between the thigh and shin is constant throughout the range of motion.

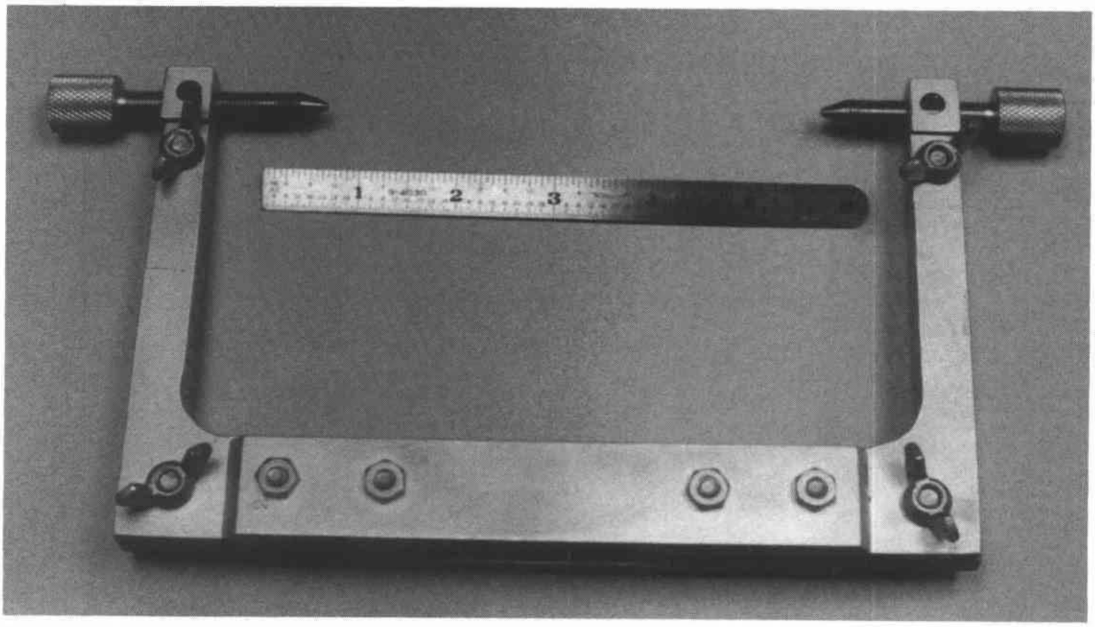


Figure 4. Knee spanning template holder. Six inch rule included in photograph to give a sense of scale.

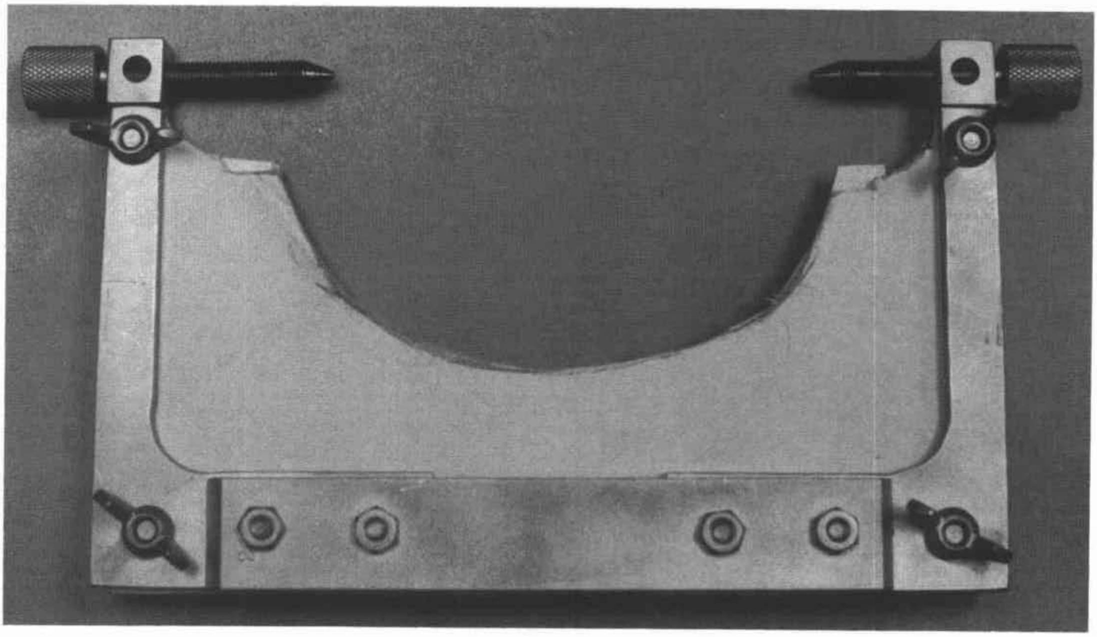


Figure 5. Template holder with stiff cardboard template clamped between the two plates of the holder.

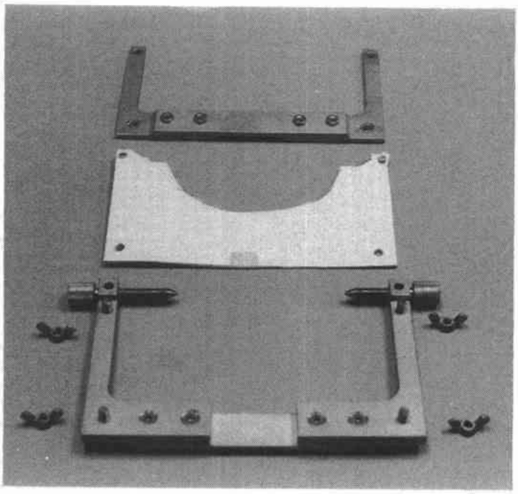


Figure 6. Exploded parts view of template holder and template.

Author

Robert Gilley, C.P., is with Durr-Fillauer Medical, Inc., 2710 Amnicola Highway, Chattanooga, Tennessee 37406.

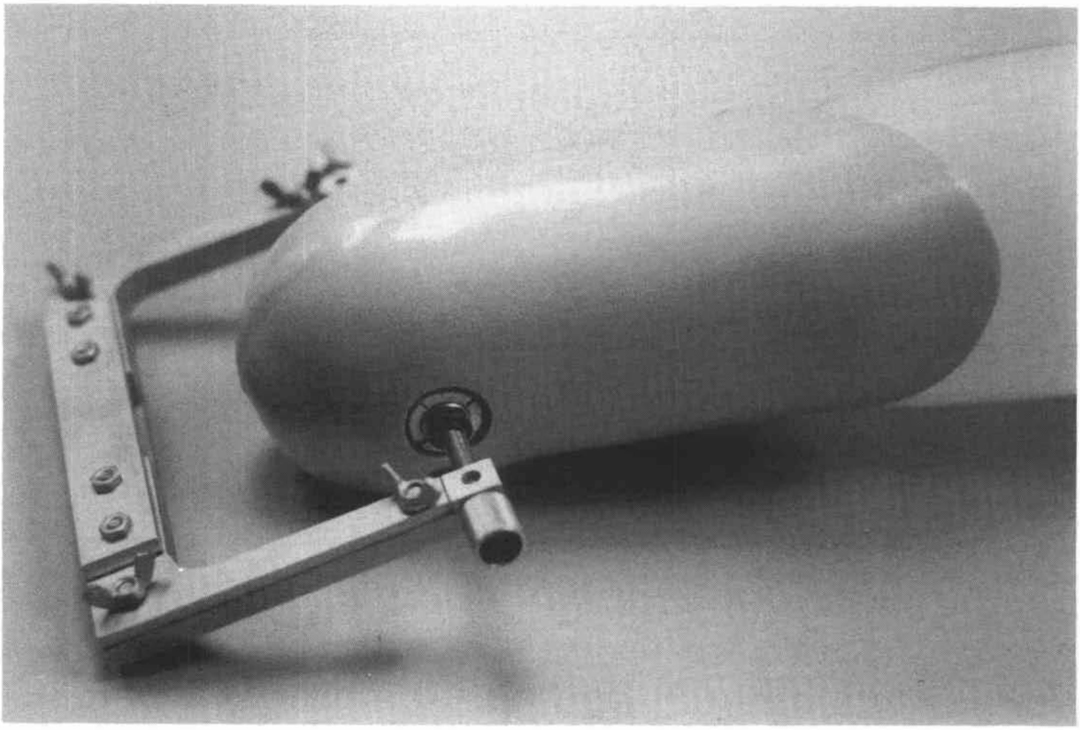


Figure 7. Template holder in place mounted on the knee bolt of a conventional above-knee prosthesis (for illustrative purposes only, a knee disarticulation prosthesis was not available at the time this article was prepared).

Clinical Prosthetics and Orthotics, Vol. 12, No. 3, pp. 128–135

With a Spring in One's Step

by D.D. Murray, M.D.
W.J. Hartvikson
H. Anton
E. Hommonay, C.P.O.(C)
N. Russell, C.P.(C)

Introduction

In recent years, there has been a significant number of new developments in prosthetics in both North America and Europe. New concepts for socket molding, knee control, dynamic foot action, and the utilization of space-age materials have expanded prosthetic development and performance.

The traditional prosthetic foot had a keel and an articulated ankle. This concept has modern derivatives with multi-axis ankles, but the principle remains the same. The S.A.C.H. foot design is that of the solid ankle and cushioned heel. By virtue of a compressible heel of a selected rubber density, the wearer achieves a simulated ankle motion at heel strike.[1] This design has been a mainstay in prosthetic fabrication for several decades. These feet are both essentially passive and accommodating. The Seattle foot, with its cushioned heel and keel spring action, stores energy through the stance phase of gait and releases it at toe-off, thus imparting a dynamic component to gait.[2] An added feature of this foot is that of cosmetic molding.

The principle of dynamic toe-off to improve the mechanical efficiency of the prosthesis is an attractive one, and it forms the basis for the design of the Seattle foot. The purpose of this study is to evaluate the performance of the Seattle foot and subjectively and objectively determine whether or not it improves prosthetic gait.

Clinical Investigation

A questionnaire was designed to gather general demographic data and review foot function in general living situations. Thirty-three patients were identified in the last two years as having been fit with a Seattle foot, and 31 (94%) responded to the questionnaire. There were 27 males and four females. The age range was from 24 years to 72 years (Figure 1).

The weight of the patients ranged from 95 pounds to 195 pounds and their height ranged from 5′1″ to 6′4″.

Amputation dates ranged from 1930 to 1986, with over half of the respondents having been injured since 1975.

On average, each patient had 3.75 surgical procedures, with a range from 1 to 24.

The length of time from amputation to prosthetic fitting was, for the most part, under one year (Figure 2).

The original foot supplied in most cases was a S.A.C.H. foot. The next most frequent, in order, was a single axis ankle with a keel foot. The remainder are unknown. A significant number of the candidates had been using their original foot an average of 14 years before having it changed to a Seattle foot. For the

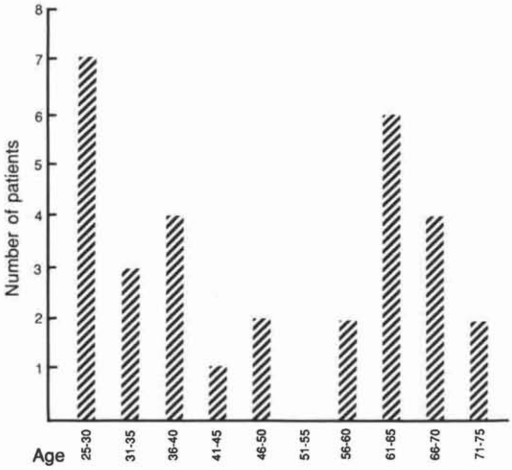


Figure 1. The age range was from 24 to 72 years.

most part, people were attracted to the Seattle foot because of a better design and newer technology. They wished for added spring, flexibility, and mobility in the foot. Some simply tried it because it was recommended by staff, or because they liked the cosmetic appearance.

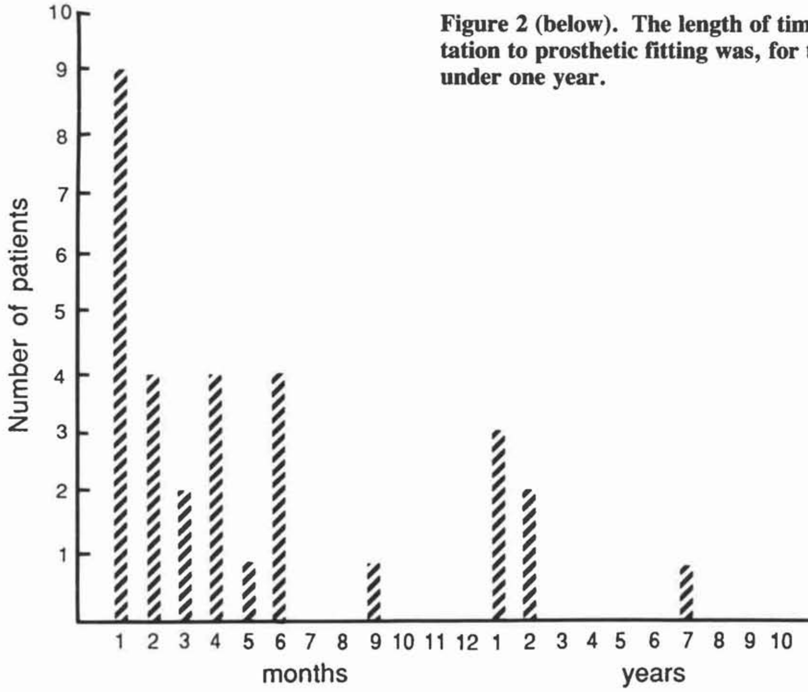
The length of time for use of the Seattle foot ranges from one month to two years with an average of 8.5 months (Figure 3).

The Seattle foot was fit on 29 below-knee amputees and two above-knee amputees.

The heel stiffness in the Seattle foot was rated as acceptable in 80% of cases. Twelve percent (12%) felt it was too stiff. Eighty-one percent (81%) of respondents felt that they had good ankle motion with the Seattle foot, and 19% felt they did not. Seventy-four percent (74%) of respondents felt that the ankle motion was greater than with the previous foot, 16% felt it was the same, and 10% felt less ankle motion.

When questioned about the shock stress at the hip or knee, 55% felt there was decreased shock stress and 39% felt that there was no change.

Figure 2 (below). The length of time from amputation to prosthetic fitting was, for the most part, under one year.



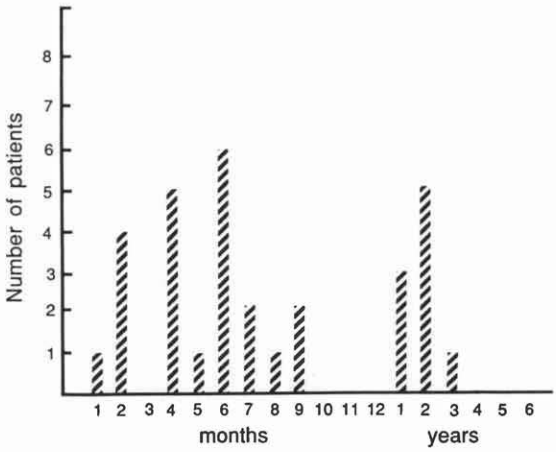


Figure 3. The length of time for use of the Seattle foot ranges from one month to two years, with an average of 8.5 months.

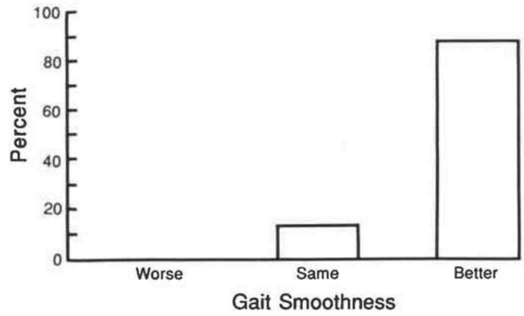


Figure 4. The greatest advantages with the Seattle foot were a more natural and smooth action.

When questioned about the effect of the Seattle foot on changing gait, 87% felt it was better and 13% felt it was the same.

Eighty-seven percent (87%) were aware of toe-off action in the Seattle foot and 13% were unaware of it. The toe-off action was most noticeable when accelerating quickly, climbing up or down, playing ball sports, and running or walking on uneven ground. Forty-eight percent (48%) of the respondents would have preferred greater toe-off action, whereas 52% were satisfied with the toe-off.

Half the respondents felt the Seattle foot had made a general difference to their recreational pursuits. When specific activities were rated, at least 50% of respondents felt that walking, going up and down stairs, hiking, dancing, and jogging were consistently easier than with the previous foot.

Balance and endurance on the prosthesis was felt to be easier by about 61% of the respondents and smoothness was better in 87%.

Uneven terrain was considered easier by 74%, but 3% said it was more difficult. In fact, the Seattle foot does not provide as much forefoot flexibility in the medial-lateral plane as with an articulated ankle joint.

Walking and running was easier for 67% of the respondents (48% of the patients jogged). Of the 61% who dance, 74% found it easier.

Of those people responding negatively to the Seattle foot, the pattern was either negative responses throughout the questionnaire (by four respondents) or negative responses for certain

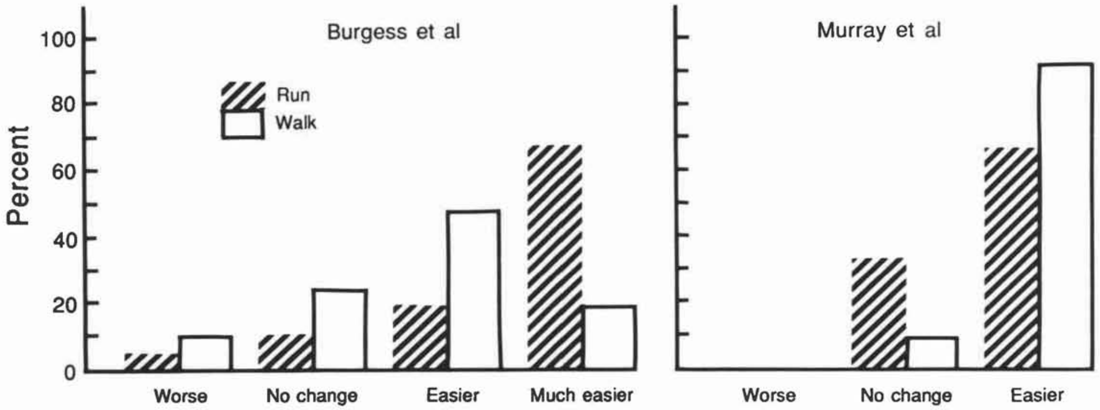


Figure 5. A comparison of two clinical surveys of the Seattle foot for running and walking.

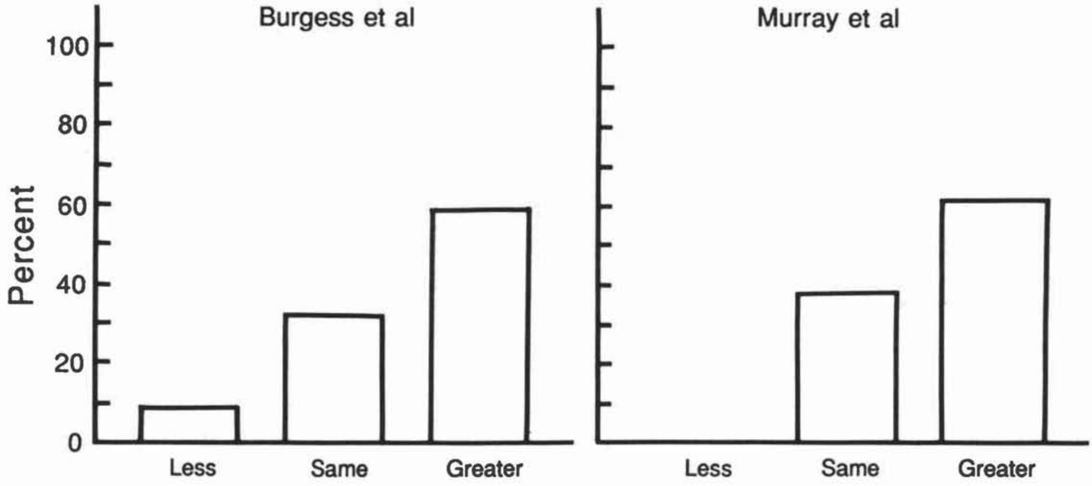


Figure 6.

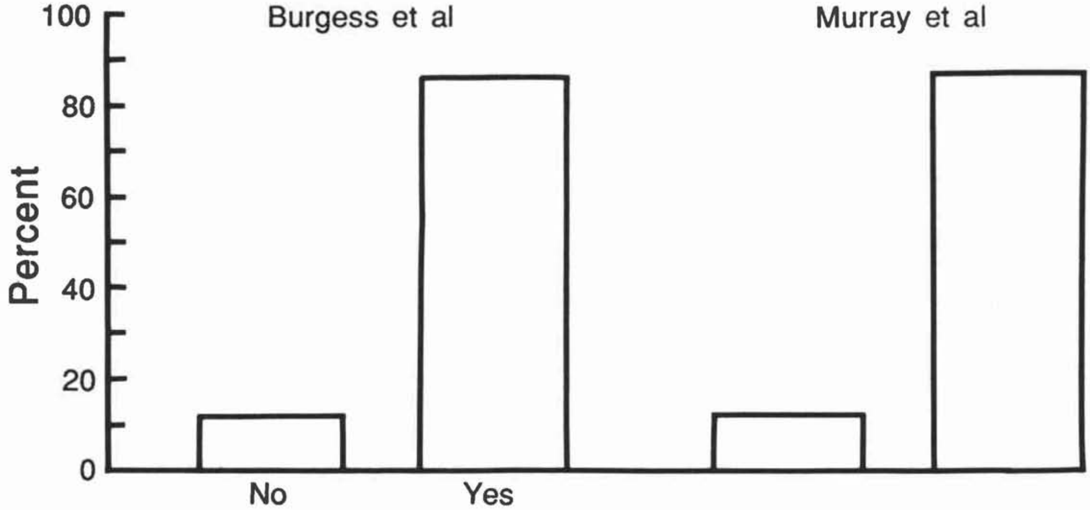


Figure 7.

functions, such as the half who felt there was no difference in the recreational pursuits. Of these negative responses, there was no pattern either in terms of age, weight, or amputation site.

The greatest advantages with the Seattle foot were reported to be a more natural and smooth action, resulting in an improved gait (Figure 4), better ability to handle stairs and uneven ground, and improved abilities in sports.

The cosmetic design and the anatomical detail were appreciated by 97% of the respondents.

Residual limb pain was felt to be decreased in 39% of respondents and unchanged in 45%. Sixteen percent (16%) did not respond to this question. The foot design had not been expected to have any effect on this problem.

Skin problems were felt to be decreased in 55% of the respondents. Thirty-five percent (35%) said there was no change. The foot design was not expected to improve this clinical problem either.

The Department of Veterans Affairs in Seattle has reported an evaluation of the Seattle foot.[3] Although a comparison of amputee

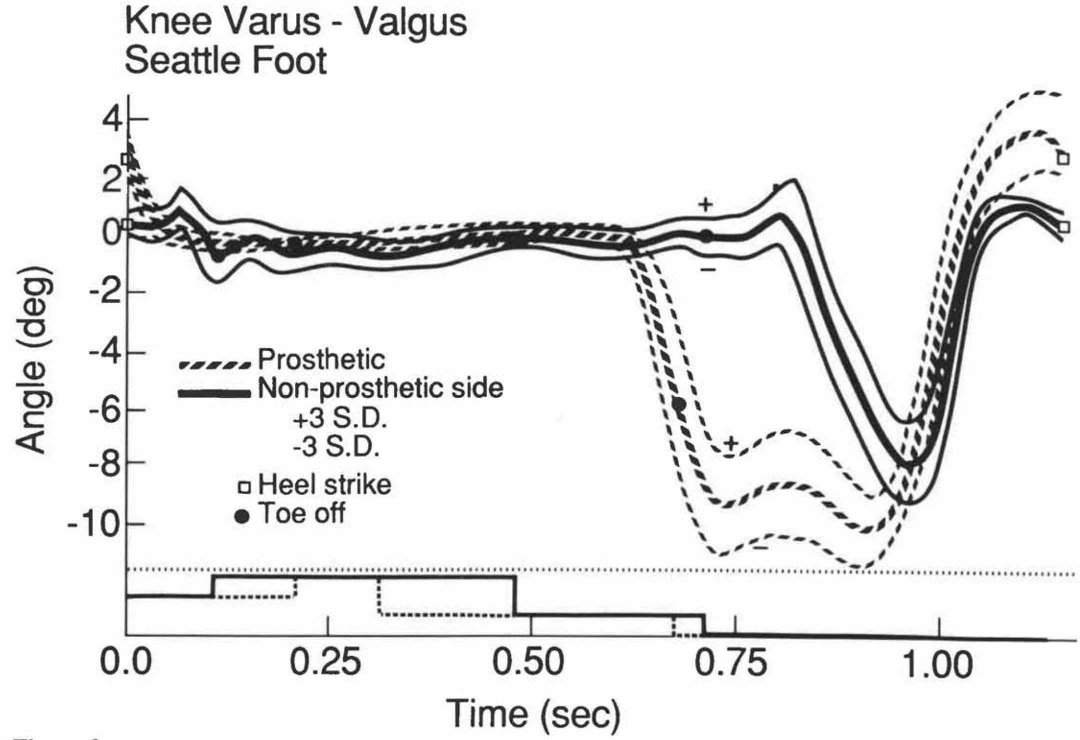


Figure 8.

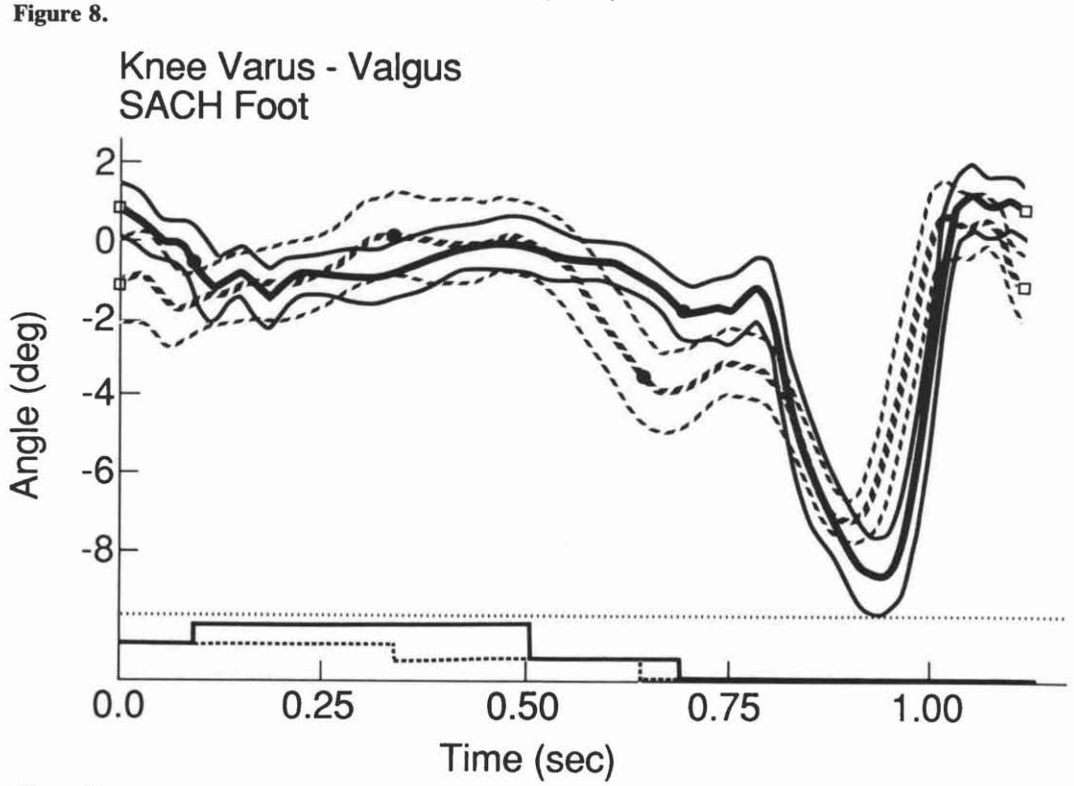


Figure 9.

groups was not possible, the results of this clinical survey compare favorably with the original study. Figures 5, 6, and 7 graphically demonstrate the comparison.

Laboratory Investigation

Electrogoniometric Evaluation

A gait study using a single amputee with many years experience with a S.A.C.H. foot and several years experience with the Seattle foot was undertaken at the G.F. Strong Gait Laboratory.

Motion in the lower extremity was analyzed using a computerized electrogoniometric system. This system accurately measures movement in three planes at the hip, knee, and ankle and stores data for subsequent analysis.[4] The S.A.C.H. foot, Seattle foot, and non-prosthetic side were compared.

Patterns of movement measured at the hip were similar for the S.A.C.H. and Seattle feet and resembled those seen on the non-prosthetic side. At the knee, the Seattle foot produced a more repeatable pattern of internal-external rotation and varus-valgus than did the S.A.C.H. foot (Figures 8 and 9).

The greatest differences between the S.A.C.H. and Seattle feet were seen at the ankle. The patterns of forefoot abduction-adduction, plantar flexion-dorsiflexion, and inversion-eversion were all more repeatable for the Seattle foot.

Also, the pattern of plantar flexion-dorsiflexion for the Seattle foot more closely resembled that of the non-prosthetic side (Figures 10 and 11).

In summary, the Seattle foot generally produced a more repeatable pattern of motion at the knee and ankle than the S.A.C.H. foot, and the pattern of plantar flexion-dorsiflexion for the Seattle foot appeared more normal.

Force Plate Evaluation

Through the facilities of Simon Fraser University Kinesiology Department, a force plate study was done on the same single subject. The vertical compression forces generated by the S.A.C.H. and Seattle feet during stance were measured. Figure 12 demonstrates typical forces measured during stance in a below-knee amputee on the non-prosthetic side. A maximum peak is seen immediately after heel strike. This is followed by a trough in mid-stance and a second, lesser peak at push-off.

Figure 13 illustrates the forces generated in the same individual during stance on his prosthetic side while using a Seattle foot. Figure 14 shows stance forces generated in the same individual on his prosthetic side using a S.A.C.H. foot.

The initial peak is greater for the S.A.C.H. than the Seattle foot. This suggests more effective shock absorption at heel strike for the Seattle foot than the S.A.C.H. foot. The second peak is less than that seen on the non-prosthetic side with both feet, but is greater for the Seattle foot than the S.A.C.H. foot. Thus, the Seattle foot more closely approximates normal push-off force than the S.A.C.H. foot. The trough at mid-stance is shorter with the S.A.C.H. foot than on the non-prosthetic side. The mid-stance trough for the Seattle foot more closely approaches that of the non-prosthetic side, suggesting a more normal pattern of foot-ankle motion than with the S.A.C.H. foot. In summary, the Seattle foot generally appears to produce a more normal pattern of vertical forces than the S.A.C.H. foot and produces a greater force at push-off.

Conclusion

The patient response to the questionnaire regarding the effectiveness of the Seattle foot was positive. Comparison with the Seattle Study revealed similar results. Gait studies undertaken tended to support the clinical impression with regard to both kinetics and kinematics. Overall, this dynamic foot design offers definite advantages to the prosthetic user. At best, prosthetic users seem to get an increased gait smoothness, with the dynamic toe action positively influencing their abilities on rough ground and inclines. At worst, their gait pattern is not negatively influenced by this spring action.

References

[1] *Orthopaedic Appliances Atlas*. Vol. 2, Artificial Limbs, Editor J.W. Edwards, Ann Arbor, Michigan, 1960, pp. 149–151.

[2] Reswick, J.B., "Evaluation of the Seattle Foot," *J. Rehab Research and Development*, Vol. 23, No. 3, pp. 77–94.

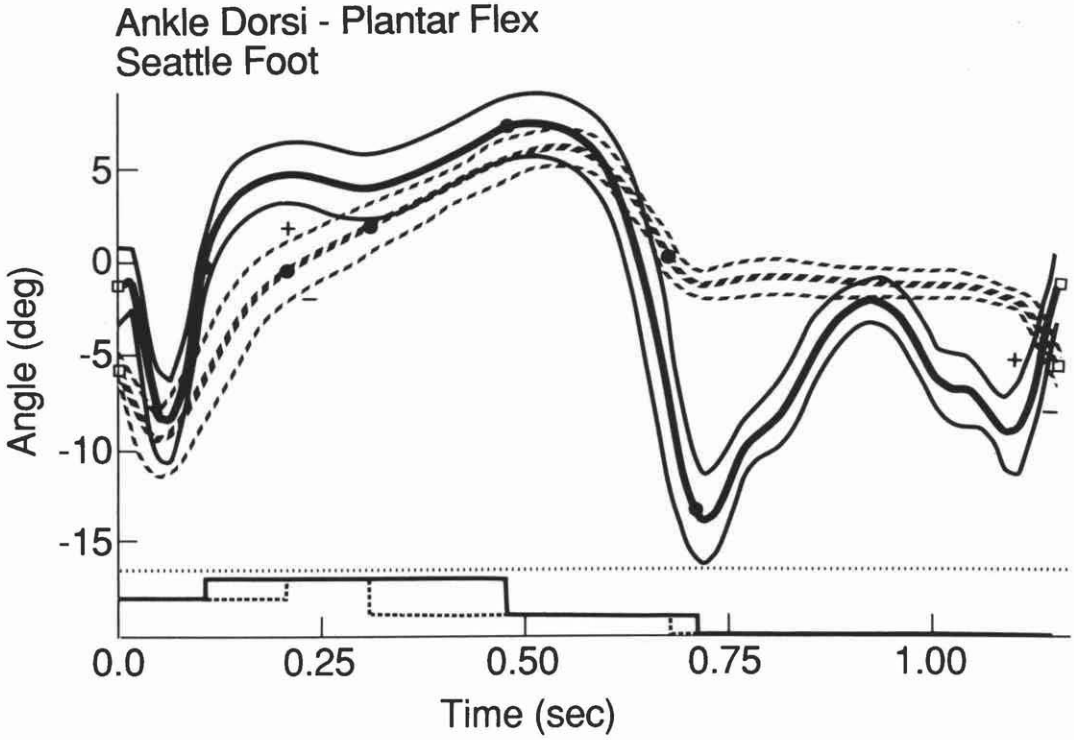


Figure 10.

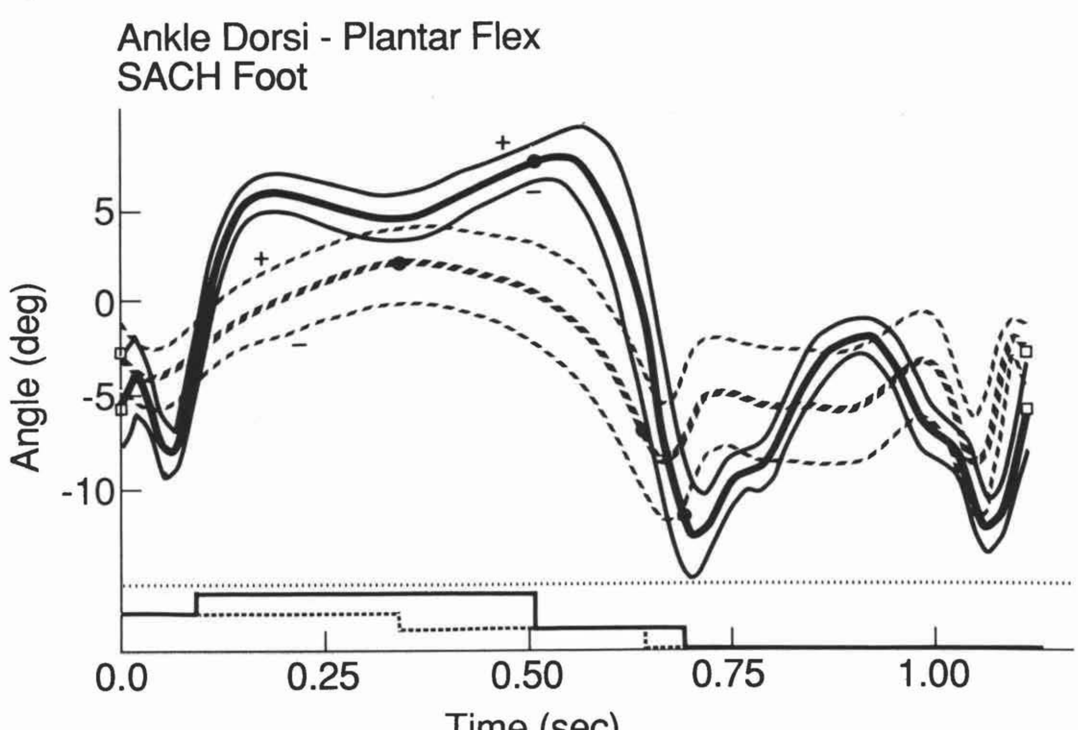


Figure 11.

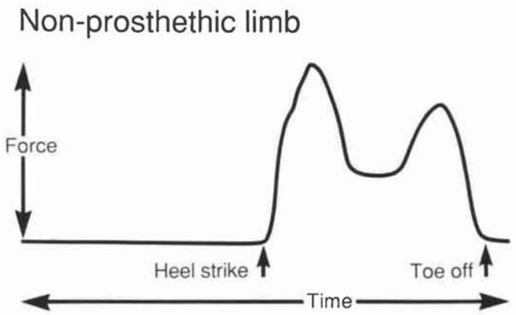


Figure 12. Typical forces measured during stance in a below-knee amputee on the non-prosthetic side.

Figure 13. The forces generated in the same individual during stance on his prosthetic side while using a Seattle foot.

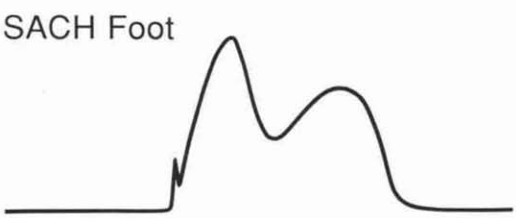


Figure 14. Stance forces generated in the same individual on his prosthetic side using a S.A.C.H. foot.

[3] Burgess, E.M. et al., "Development and Preliminary Evaluation of the V.A. Seattle Foot," *Journal of Rehabilitation Research and Development,* Vol. 22, No. 3, B.P.R. 10–42, pp. 75–84.

[4] Chao, Edmund, "Justification of Triaxial Goniometer for the Measurement of Joint Rotation," *J. Biomechanics,* Vol. 13, 1980, pp. 989–1006.

Acknowledgments

The authors wish to thank the G.F. Strong Gait Lab and Dr. Cecil Herschler, as well as the Simon Fraser Kinesiology Department and Dr. Arthur Chapman for their technical assistance in the preparation of this study.

Authors

Dr. D. Duncan Murray, F.R.C.P.(C) and co-authors can be contacted at the Department of Medicine, Shaughnessy Hospital, Vancouver, British Columbia V6H 3M1. Dr. Murray is Professor and Head of the Department of Medicine at Shaughnessy Hospital.

Calendar

Please notify the National Office immediately concerning all meeting dates. It is important to submit meeting notices as early as possible. In the case of Regional Meetings, check with the National Office prior to confirming date to avoid conflicts in scheduling.

1988

June 14–18, AOPA Regions VII, VIII, X, and XI Combined Annual Meeting, Westin Hotel, Seattle, Washington. Contact: Steve Colwell, (206) 526-7944.

June 15, 16, 17, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

June 22–25, Convention of the Canadian Association of Prosthetists and Orthotists (CAPO), Queen Elizabeth Hotel, Montreal, Quebec, Canada. Contact: C.A.P.O. Convention '88, 5713 Cote des Neiges, Montreal, Quebec H3S 1Y7, Canada; (514) 731-3378.

June 23–27, AOPA Regions V and VI and the Academy Midwest Chapter Joint Education Seminar, Pheasant Run, St. Charles, Illinois. Contact: Kathi Ensweiler, CO, (219) 836-2251.

June 25–30, International Conference of the Association for the Advancement of Rehabilitation Technology, Palais des Congres, Montreal, Quebec, Canada. Contact: International Conference, 3631 Rue St. Denis, Montreal, Quebec H2X 3L6, Canada; (514) 849-9847.

July 13, 14, 15, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

July 15–16, Academy Continuing Education Conference 3-88, ''Clinical Practice Management—Ethical and Legal Considerations,'' Vanderbilt Plaza Hotel, Nashville, Tennessee. Conact: Academy National Headquarters, (703) 836-7118.

July 16–17, ABC Board of Director's Meeting, Washington, D.C. Contact: ABC National Office, (703) 836-7114.

July 16, 17, 18, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

July 27–29, Hosmer Electric Systems Workshop, University of Washington, Seattle, Washington. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; (408) 379-5151 or (800) 538-7748.

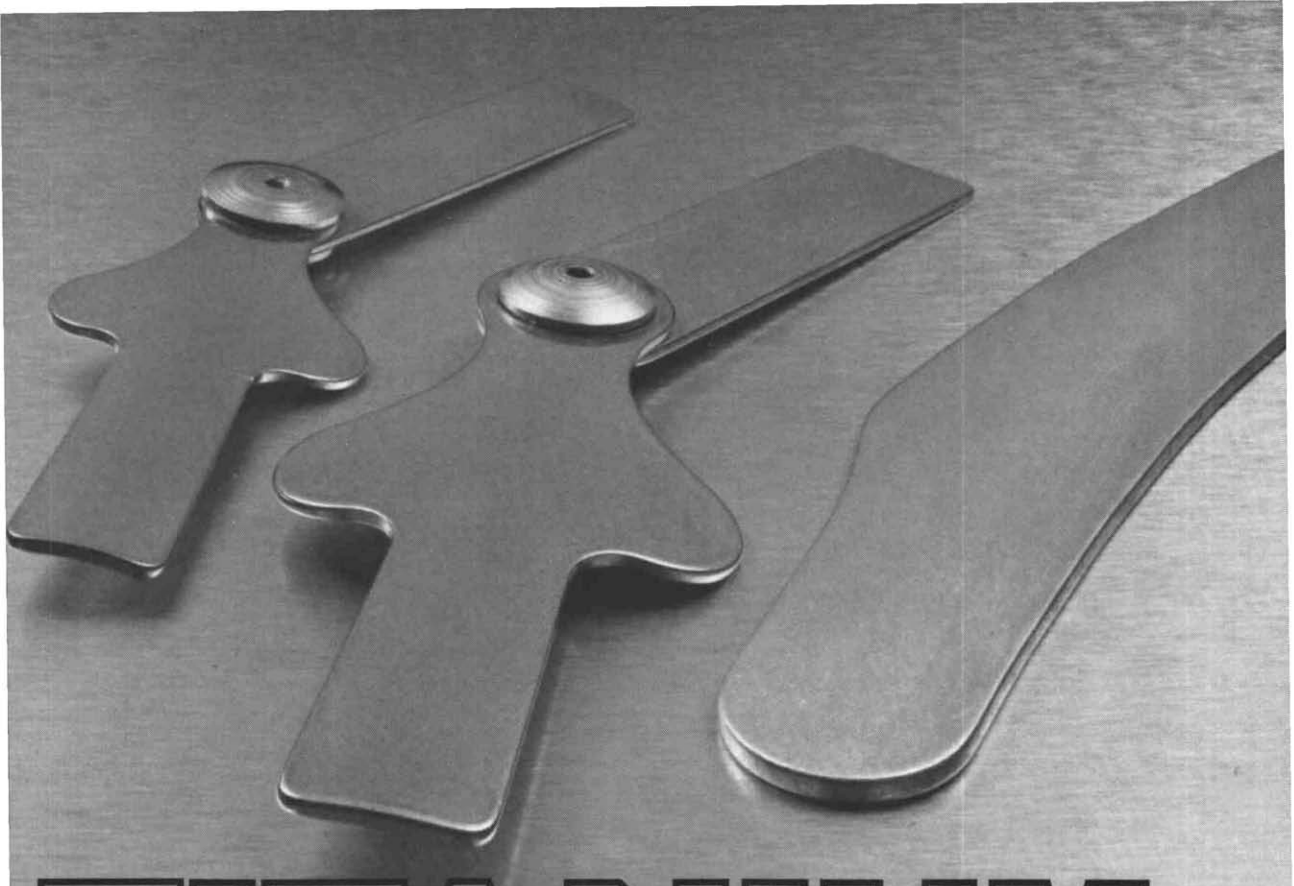
July 27, 28, 29, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 street, Hialeah, Florida 33016; (305) 823-8300.

August 8–21, ABC CPM Examination, University of Texas, Dallas, Texas. Contact: ABC National Office, (703) 836-7114.

August 10, 11, 12, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

August 13, 14, 15, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

August 15, ABC Technican Examination application deadline. Contact: ABC National Office, (703) 836-7114.



TITANIUM:

Exotic metals technology leads to ultra-light, ultra-strong hip joint design.

Reduce the weight of your patients prosthesis without compromising on strength—with either regular or small titanium hip joints from USMC.

Check the adjacent chart for a comparison and you'll find that titanium is both lighter and stronger than stainless steel—and just as easy to fabricate. Titanium cuts, drills, buffs and bends as easy as stainless steel while cutting hip joint weight by 40%.

Use the following product numbers to get more muscle out of a lighter weight hip joint.

Regular Titanium Hip Joint—P02-23H-D001
Small Titanium Hip Joint—P02-23S-M001
Pelvic Band—P02-340-0000

	STAINLESS STEEL Tensile Strength 125,000PSI	TITANIUM Tensile Strength 140,000PSI
	Weight	Weight
Regular Hip Joint	8.5 oz. (242 gm.)	5.2 oz. (147 gm.)
Small Hip Joint	4.7 oz. (132 gm.)	3.2 oz. (90 gm.)
Hip Band	9.7 oz. (274 gm.)	4.1 oz. (117 gm.)

USMC™

United States Manufacturing Company
180 N. San Gabriel Blvd., Pasadena, CA 91107-3488 U.S.A.
P.O. Box 5030, Pasadena, CA 91107-0030 U.S.A.
Telephone: (818) 796-0477, Cable: LIMBRACE, TWX: 910-588-1973
Telex: 466-302, FAX: (818) 440-9533

August 24, 25, 26, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

September 3, 4, 5, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

September 5–9, 16th World Congress of Rehabilitation International, Keio Plaza Inter-Continental Hotel, Shinjuku, Tokyo, Japan. Contact: Secretary General, 16th World Congress of Rehabilitation International, % the Japanese Society for Rehabilitation of the Disabled, 3-13-15, Higashi Ikebukuro, Toshima-Ku, Tokyo 170, Japan.

September 9–10, Annual Fall Meeting of the Ohio Orthotics and Prosthetics Association, Radisson Hotel, Columbus, Ohio. Contact: O.O.P.A. and Ohio A.A.O.P. State Office, 4355 N. High Street, #208, Columbus, Ohio 43214; (614) 267-1121.

September 14–16, Hosmer Dorrance Systems Workshop, Tulane University, New Orleans, Louisiana. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; (408) 379-5151 or (800) 538-7748.

September 14, 15, 16, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

September 15, ABC Written/Visual and CPM Examination application deadline. Contact: ABC National Office, (703) 836-7114.

September 15–16, ABC Technician Examination, Spokane Falls, Washington. Contact: ABC National Office, (703) 836-7114.

September 23–24, Academy Continuing Education Conference 4-88, "Spinal Orthotics and Seating," Holiday Inn at Kansas City Airport, Kansas City, Missouri. Contact: Academy National Headquarters, (703) 836-7118.

September 23–24, Academy Continuing Education Conference 5-88 and Northern California Chapter Combined Meeting, "Current Clinical and Technical Concepts in Lower Limb Prosthetics and Orthotics," San Francisco Airport Hilton, San Francisco International Airport. Contact: Academy National Headquarters, (703) 836-7118.

September 23–24, Freeman Orthotic Fitters Training Workshop, Seattle, Washington. For more information, write: Freeman, Drawer J, Sturgis, Michigan, or call Cameron Brown, (800) 253-2091.

September 24, Academy Northern California Chapter Seminar, San Francisco, California. Contact: Robert A. Bangham, CO, % Hittenbergers, 1117 Market Street, San Francisco, California 94103.

September 24–25, Northwest Chapter of Academy Meeting, Red Lion Hotel, Janzen Beach, Portland, Oregon. Contact: Glenn Kays, CPO, (503) 287-0459.

September 28, 29, 30, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

October 7–8, Academy New York State Chapter Scientific Seminar, Long Island, New York. Contact: Marty Mandelbaum, CPO, 5225-21 Nesconset Highway, Port Jefferson Station, New York 11776; (516) 473-8668.

October 12, 13, 14, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

October 15–16, ABC Written/Visual Examination. Contact: ABC National Office for locations, (703) 836-7114.

October 15, 16, 17, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High

Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Floirda 33016; (305) 823-8300.

October 19, 20, 21, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Floirda 33016; (305) 823-8300.

October 22–24, Prescription Footwear Association's 1988 Symposium, Mayflower Hotel, Washington, D.C. The theme of the symposium will be ''Pedorthic Management of Diabetic Foot Disorders.'' For exhibiting or registration information, contact Robert S. Schwartz, CP, Symposium Chairman, Prescription Footwear Association, 9861 Broken Land Parkway, Columbia, Maryland 21046; (301) 381-7278.

October 25–30, AOPA Annual National Assembly, Sheraton Washington Hotel, Washington, D.C. Contact: Katie Register, AOPA National Headquarters, (703) 836-7116.

October 27–28, ABC Board of Director's Meeting, Washington, D.C. Contact: ABC National Office, (703) 836-7116.

November 2–4, Fourth Annual Conference, ''Computer Technology/Special Education/Rehabilitation.'' Contact: Dr. Harry J. Murphy, California State University, Northridge, 18111 Nordhoff Street, Northridge, California 91330; (818) 885-2578.

November 2, 3, 4, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

November 12, Midwest Chapter of the American Academy of Orthotists and Prosthetists Fall Scientific Seminar, Northwestern University, Chicago, Illinois. Contact: Arnel Hope Dobrin, Northwestern University's Prosthetic-Orthotic Center, 345 E. Superior Street, 17th Floor, Chicago, Illinois 60611; (312) 908-8006.

November 12, 13, 14, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

November 16, 17, 18, AFI-ENDOLITE Prosthetic Certificate Course, ''Endolite High Technology Prosthesis,'' Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

November 21, Southern California Chapter of the Academy Fall Seminar, Marriott Hotel, Anaheim, California. Contact: Marmaduke Loke, 7910 Frost Street, San Diego, California 92123.

1989

January 31–February 5, Academy Annual Meeting and Scientific Symposium, Stouffer Orlando Resort, Orlando, Florida. Contact: Academy National Office, (703) 836-7118.

February 9–19, American Academy of Orthopaedic Surgeons Annual Meeting, Las Vegas, Nevada.

May 12–14, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting.

May 18–20, AOPA Region V Annual Meeting, Hotel Sofitel, Toledo, Ohio.

May 18–20, The Second S.M. Dinsdale International Conference in Rehabilitation, ''Visions and Controversies in Rehabilitation,'' hosted by the Royal Ottawa Regional Rehabilitation Centre, Ottawa, Ontario. Contact: Information Department, (613) 737-7350, ext. 602.

June 13–18, Regions VII, VIII, X, and XI Meeting, Embassy Suites Hotel, Downtown Denver, Colorado. Contact: Robert Schlesier, CPO, (303) 234-1756.

October 2–8, AOPA Annual National Assembly, Bally's Hotel, Reno, Nevada. Contact: AOPA National Headquarters, (703) 836-7116.

November 12–17, International Society for Prosthetics and Orthotics VI World Congress, Kobe Convention Center, Kobe, Japan. Contact: VI ISPO World Congress, Secretariat, % International Conference Organizers, Inc., 5A Calm Building, 4-7, Akasaka 8-chome, Minato-ku, Tokyo, 107 Japan.

1990

January 22–28, Academy Annual Meeting and Scientific Symposium, Hyatt Regency Hotel, Phoenix, Arizona. Contact: Academy National Office, (703) 836-7118.

February 8–13, American Academy of Orthopaedic Surgeons Annual Meeting, New Orleans, Louisiana.

May 11–13, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting.

June 27–July 1, Region V and VI Combined Meeting, Amway Grand Plaza Hotel, Grand Rapids, Michigan. Contact: Bob Leimkuehler, CPO.

September 11–16, AOPA Annual National Assembly, Sheraton Boston Hotel, Boston, Massachusetts. Contact: Katie Register, AOPA National Headquarters, (703) 836-7116.

1991

March 19–24, Academy Annual Meeting and Scientific Symposium, Town and Country Hotel, San Diego, California. Contact: Academy National Office, (703) 836-7118.

October 1–6, AOPA Annual National Assembly, Disneyland Hotel, Anaheim, California. Contact: Katie Register, AOPA National Office, (703) 836-7116.

Clinical Prosthetics and Orthotics
AUTHOR INDEX
Volume 12

Birke, James A., with Deborah A. Nawoczenski, "Orthopedic Walkers: Effect on Plantar Pressures," 12/2 (Spring, 1988), p. 74.

Bowker, John H., "Partial Foot and Syme Amputations: An Overview," 12/1 (Winter, 1987–88), p. 10.

Coleman, William C. with Arthur Plaia, "Soft Molded Sandals for Insensitive Foot Care," 12/2 (Spring, 1988), p. 67.

Collins, Jack N., "A Partial Foot Prosthesis for the Transmetatarsal Level," 12/1 (Winter, 1987–88), p. 19.

Gilley, Robert, "Technical Note: Cosmesis and the Knee Disarticulation Prosthesis," 12/3 (Summer, 1988), p. 123.

Gilley, Robert, "Technical Note: Fabrication of the Syme Prosthesis," 12/1 (Winter, 1987–88), p. 41.

Imler, Clarence D., "Imler Partial Foot Prosthesis I.P.F.P.—'Chicago Boot,' " 12/1 (Winter, 1987–88), p. 24.

Kalter, Mitchell E., with Richard L. Jacobs, "The Susceptible Insensate Foot," 12/2 (Spring, 1988), p. 61.

LaTorre, Richard, "Total Contact Partial Foot Prosthesis," 12/1 (Winter, 1987–88), p. 29.

Littig, David H. with Judd E. Lundt, "The UCLA Anatomical Hip Disarticulation Prosthesis," 12/3 (Summer, 1988), p. 114.

Michael, John, "Component Selection Criteria: Lower Limb Disarticulations," 12/3 (Summer, 1988), p. 99.

Murray, D.D., with W.J. Hartvikson, H. Anton, E. Hommonay, and N. Russell, "With a Spring in One's Step," 12/3 (Summer, 1988), p. 128.

Quigley, Michael J., with Sam E. Hamontree, and Joe Antorietto, "The Rancho Syme Prosthesis with the Regnell Foot," 12/1 (Winter, 1987–88), p. 33.

Sabolich, John with Thomas Guth, "The CAT-CAD-H.C.™ A New Design for Hip Disarticulation Patients," 12/3 (Summer, 1988), p. 119.

Schuch, C. Michael, "Orthotic Management of the Arthritic Foot," 12/2 (Spring, 1988), p. 51.

Schuch, C. Michael, "Report From: International Workshop on Above-Knee Fitting Alignment Techniques," 12/2 (Spring, 1988), p. 81.

Seale, Karen S., "Clinical Analysis of Foot Problems," 12/2 (Spring, 1988), p. 44.

Stills, Melvin, "Partial Foot Prostheses/Orthoses," 12/1 (Winter, 1987–88), p. 14.

Tuil, Peter, "The Hip Disarticulation Prosthesis as Developed by the O.I.M. Noord Nederland," 12/3 (Summer, 1988), p. 109.

Clinical Orthotics and Prosthetics TITLE INDEX Volume 12

A Partial Foot Prosthesis for the Transmetatarsal Level, *Jack N. Collins,* 12/1 (Winter, 1987–88), p. 19.

Clinical Analysis of Foot Problems, *Karen S. Seale,* 12/2 (Spring, 1988), p. 44.

Component Selection Criteria: Lower Limb Disarticulations, *John Michael,* 12/3 (Summer, 1988), p. 99.

Imler Partial Foot Prosthesis I.P.F.P.—"Chicago Boot," *Clarence D. Imler,* 12/1 (Winter, 1987–88), p. 24.

Orthopedic Walkers, *James A. Birke and Deborah A. Nawoczenski,* 12/2 (Spring, 1988), p. 74.

Orthotic Management of the Arthritic Foot, *C. Michael Schuch,* 12/2 (Spring, 1988), p. 51.

Partial Foot and Syme Amputations: An Overview, *John H. Bowker,* 12/1 (Winter, 1987–88), p. 10.

Partial Foot Prostheses/Orthoses, *Melvin Stills,* 12/1 (Winter, 1987–88), p. 14.

Report From: International Workshop on Above-Knee Fitting and Alignment Techniques, *C. Michael Schuch,* 12/2 (Spring, 1988), p. 81.

Soft Molded Sandals for Insensitive Foot Care, *William C. Coleman and Arthur Plaia,* 12/2 (Spring, 1988), p. 67.

Technical Note: Cosmesis and the Knee Disarticulation Prosthesis, *Robert Gilley,* 12/3 (Summer, 1988), p. 123.

Technical Note: Fabrication of the Syme Prosthesis, *Robert Gilley,* 12/1 (Winter, 1987–88), p. 41.

The CAT-CAM-H.D.™ A New Design for Hip Disarticulation Patients, *John Sabolich and Thomas Guth,* 12/3 (Summer, 1988), p. 119.

The Hip Disarticulation Prosthesis as Developed by the O.I.M. Noord Nederland, *Peter Tuil,* 12/3 (Summer, 1988), p. 109.

The Rancho Syme Prosthesis with the Regnell Foot, *Michael J. Quigley, Sam E. Hamontree,* and *Joe Antorietto,* 12/1 (Winter, 1987–88). p. 33.

The Susceptible Insensate Foot, *Mitchell E. Kalter* and *Richard L. Jacobs,* 12/2 (Spring, 1988), p. 61.

The UCLA Anatomical Hip Disarticulation Prosthesis, *David H. Littig* and *Judd E. Lundt,* 12/3 (Summer, 1988), p. 114.

Total Contact Partial Foot Prosthesis, *Richard LaTorre,* 12/1 (Winter, 1987–88), p. 29.

With a Spring in One's Step, *D.D. Murray, W.J. Hartvikson, H. Anton, E. Hommonay,* and *N. Russell,* 12/3 (Summer, 1988), p. 128.

The American Academy of Orthotists and Prosthetists

American Academy of Orthotists and Prosthetists
717 Pendleton Street
Alexandria, Virginia 22314