Clinical Prosthetics & Orthotics

Evaluative Fitting Techniques

Volume 9, Number 3 **Summer, 1985**

The Officers, Directors, and members of the Academy would like to acknowledge very gratefully the contribution of the firms listed below, whose generosity helped make possible the publication of this issue of *Clinical Prosthetics and Orthotics*.

Kingsley Manufacturing Company
1984 Placentia Avenue
P.O. Box CSN 5010
Costa Mesa, California 92628

Newington Children's Hospital
181 East Cedar Street
Newington, Connecticut 06111

Fessenden Prosthetic-Orthotic Service, Inc.
5707 Dorr Street
Toledo, Ohio 43615
Gary O. Fessenden, CP
David W. Bilby, CP

Clinical Prosthetics & Orthotics

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.

Clinical Prosthetics and Orthotics (ISSN 0279-6910) is published quarterly by the American Academy of Orthotists and Prosthetists, 717 Pendleton St., Alexandria, VA 22314. Subscriptions: $15.00 domestic, $20.00 foreign, $30.00 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.

CONTENTS

Volume 9, Number 3 **Summer, 1985**

FEATURES

PRIZES FOR PUBLISHED ARTICLES

The American Academy of Orthotists and Prosthetists, through the sponsorship of the William Hood Manufacturing Co. of Vittoria, Ontario, Canada, announces prizes for articles published in *Clinical Prosthetics and Orthotics*.

- $250.00 for the best article by an orthotics/prosthetics practitioner
- $250.00 for the best article by an orthotics/prosthetics student

Stipulations

The practitioner must be a member of the Academy. The student must be enrolled in an ABC accredited educational program. All articles will be judged by the Academy Editorial Board, and the decision of the judges is final. Prizes will be announced in *Clinical Prosthetics and Orthotics*. All published articles by authors meeting the above criteria will be considered.

Submit articles to: Charles H. Pritham, CPO, Editor, *Clinical Prosthetics and Orthotics*, c/o Durr-Fillauer Medical, Inc., Orthopedic Division, P.O. Box 5189, Chattanooga, Tennessee 37406.

The awards have been made possible through the generosity of the Wm. Hood Mfg. Co.

TO: ORTHOTISTS AND PROSTHETISTS
FROM: EDITOR, *CLINICAL PROSTHETICS AND ORTHOTICS*
RE: INVITATION TO CONTRIBUTE TO THE *C.P.O.*

As professionals, we are obligated to do what we can to advance the state-of-the-art and share new developments with our colleagues. The most efficient way to transfer information, and the way that has the greatest impact, is through the written word. But, for many professionals, writing is a task that often becomes monumental to the point that we succumb to inertia. Writing, however, is not such a monumental task if we break it down into smaller, simpler tasks which we can complete one at a time.

The initial and most difficult problem every writer faces is how to organize the material. The quickest way to organize material is through use of an outline. In its most basic form, an article is divided into three parts—introduction, body, and conclusion. The introduction states the subject and gives pertinent background information that is necessary in order to understand the topic. The main body of the article then presents the topic in as much detail as possible. At the heart of every article is the intent to inform and answer a variety of questions. The body can include subheads, such as review of literature, method, clinical materials, discussion, and results. The conclusion restates the main points presented in the article.

Clinical Prosthetics and Orthotics addresses broad, philosophical issues, and as such invites a more subjective style. Each issue of *C.P.O.* centers on a main topic. Usually, an issue will contain a lead article, an editorial, and one or more technical articles pertaining to the topic. Authors are solicited by the Academy editorial board; however, *C.P.O.* also accepts unsolicited articles. Unsolicited articles need not cover the topic at hand and may be of a more technical and objective nature. All articles are submitted to the editor, a professional in the field, who checks every article for accuracy, terminology, format, and references. The articles are then forwarded to the publications staff at the Academy National Headquarters for production and printing.

See page 48 for article guidelines.

Lead Article

An Advanced Approach Toward Improved Prosthetic Fittings

by David F.M. Cooney, R.P.T., C.P.O.
Keith E. Vinnecour, C.P.O.

The importance of amputation surgery and dedicated follow-up cannot be underestimated by those clinicians who deal with the amputee population. A prosthetist who receives a patient with a residual limb that is of the optimum configuration to receive a prosthesis and permits the lowest energy cost with maximum unilateral weight bearing comfort, is too often the exception. A concerted effort by all professionals involved—physicians, nurses, physical and occupational therapists, psychologists, social workers, and prosthetists—is required for truly successful rehabilitation.

DELINEATION OF LEVEL

Successful primary healing in patients who have experienced a trauma related amputation is not as great a concern since the average age of this group is much younger than the dysvascular amputee. For the majority of patients who require prosthetic care due to vascular insufficiency, predictions for successful healing, and therefore level of amputation, is a critical consideration and of primary address here. The following discussion and techniques employed, however, can apply to all prosthetic fittings.

In the dysvascular patient, the correct assessment of tissue viability and level of limb amputation is paramount to successful rehabilitation. Correct assessment also serves to reduce the length of the hospital stay and, therefore, costs. Patient morbidity and mortality are also reduced.[21]

A number of methods are employed to determine amputation level. Absolute determinants include ischemia and necrosis. Skin temperatures, absence of hair, sensory deficits, and peripheral pulses are also clinical tools of relative, though unreliable, demarcation. A less direct way of determining level of amputation is the condition of the underlying tissues and skin bleeding during surgery.[1,20,31]

Objectively defined methods are being used to more accurately determine surgical level. Doppler pressure measurements use systolic pressure differentials between the level of concern and brachial pressure. The literature cited offers relative values for prediction of successful healing,[1,26] but also points out the Doppler method's fallibility.[22] Two other non-invasive tests, segmental systolic pressure readings and pulse-volume recordings, can provide a reasonably valid prediction of primary wound healing, but should not be used as the sole indicators for amputation site.[9]

Thermography has been used to estimate the optimal site of amputation. Infrared emissions from the involved extremity are displayed on a screen to show temperature differentials. One study claimed a 96 percent success rate with amputation levels recommended via thermography[12].

Skin blood flow by the Xenon-133 clearance techniques to predict primary healing levels in amputation surgery have shown positive results. A 100 percent primary amputation healing is claimed by these authors for surgeries where recommendations according to their standards were followed.[20,24]

The choice of any of the above methods rests with the abilities of the institution. Though most non-invasive means are available throughout the medical community, invasive techniques using radioactive isotopes, like Xenon-133, require the availability of a nuclear medicine department. Clearly, not all facilities have this capability.

Once the level of tissue viability and surgical healing have been determined, operative procedures commence. A residual limb offering optimal function should be a "well muscled, durable stump of effective length with a pliable skin cover that has adequate sensation." The means to this end requires careful attention to the handling of the bone, nerves, and soft tissues.[7]

SURGERY

Subsequent to determining the amputation level is the actual surgical technique, which is an important adjunct to successful rehabilitation of the amputee. Handling of the bone requires close attention to the residual cortical shaping, and in standard practice it should be beveled to prevent sharp margins and potential socket problems.[19]

The reaction of the bone to surgical handling of the periosteum is not fully understood, but when dealing with tissues that are compromised initially, one cannot fault a "kid-glove" approach to dissection and ligation. Delicate handling may avoid subsequent spurring along the bony margins.[31] It has generally been considered that fibular length should be less (approximately 2.0 cm.) than the length of the tibia.[19,31] The authors feel that fibular length should be equal to or no more than 5 mm. shorter than the cut tibia. It is felt that this improves prosthetic medio-lateral stability, provides greater distal bulk, and serves to prevent mature conical shaping and increase total tissue contact and weight-bearing.

In the procedure described by Ertl,[8] the lengths of the two bones are equal. A bony bridge, or periosteal flap, is then created to afford an end bearing residual limb. This synostosis also prevents any relative motion of the two bones. The tibiofibular osteoplasty closes the open medullary canals and can recreate the normal conditions of direct weight bearing pressures and circulation in the long axis of the bone. This can help prevent degeneration in the joints proximal to the amputation.[8,17,19] It would seem that this procedure should warrant greater attention in appropriately selected patients (especially in light of the much improved fitting techniques now available).

Establishing stabilization in the distal musculature at the selected site of amputation is important to provide a more physiologically effective residual limb. Where surgically feasible, the muscles should be sutured to each other as well as to the periosteum and/or bone without excessive tension or laxity. This allows for a well contoured and generally less prosthetically troublesome limb.[3]

Nerve tissue should be handled meticulously to avoid residual problems once prosthetic wear is initiated. Each nerve should be individually dissected and have adequate traction applied. Severing of the nerve with traction maintained will cause it to retract far enough up into the soft tissue so as to be well protected and less threatened by weight bearing pressures.[19,31] Prosthetically crucial are the smaller sural and saphenous nerves, as they are sometimes neglected in lieu of the more major posterior tibial, deep and superficial peroneal nerves.[31] Redundancy of soft tissues should be avoided, but adequate coverage of the remaining structures is a must in order to provide a good limb for weight bearing. Closure of the wound should include careful suturing and handling of the already compromised tissues and care should be taken to avoid traction at the suture line so as to prevent contractures of the joint.[3,19,31]

It has been shown again and again that immediate post-surgical fitting procedures can improve residual limb viability, reduce pain and edema, and prevent contractures.[4,7,21,31] Rigid dressings are common practice in immediate post-surgical fittings, but variations on this theme include the use of pneumatic devices that can also afford the advantages of their more rigid counterparts.[2,15] More tenuous situations that may not allow for early weight bearing and ambulation, secondary to healing problems, can be approached through the use of Una boot dressings[34] and an innovative removable rigid dressing technique.[37]

Invariably, the independent and/or conjunctive use of any one of these methods can enhance the post-operative management of even the most difficult rehabilitation patient. By improving a patient's physical and mental status and by providing mobility through this approach, the clinical team can increase a patient's rehabilitation potential.[4,21]

PROSTHETIC EVALUATION

Little has changed in the physical aspects of evaluation. Standard anthropometric measures are still used to provide an objective record for modifications and fabrication, and for comparative purposes related to future changes. Accurately determining the anatomical joint range of motion (both in the involved and uninvolved limb) and strength/stability can provide criteria for prescription and serve to mediate problems during fitting.

One new tool in the evaluative process is Xeroradiography®. Xeroradiography® is a process that yields an x-ray image on an opaque background. The picture records are easier to store than their x-ray counterparts and provide a clear definition of both the bony anatomy and soft tissue. Evidence of bone spurring, vessel calcification, and presence of vascular surgery staples is readily observed. Measurements are also easy to glean. The use of this information in the treatment of the amputee is obvious and can significantly improve and objectify the prosthetist's skills and, ultimately, improve patient management.[35]

CASTING

Adopting a "hands-on" technique in the quest of obtaining an anatomical replica of the residual limb should be the goal of the prosthetist. A careful volume study of the involved limb can serve to optimize the definitive results.

The growing use of static and dynamic test sockets, and the information provided by them, has yielded a twist on the time tested practices utilized by many prosthetists. The technique of automatic build-ups over sensitive areas has been found to be less than necessary. Reversing this thought process to promote negative model modifications over areas of weight bearing can provide better total-contact, total-weight bearing sockets. Doing this in the molding process can reduce the amount of relatively educated guesswork necessary in cast modification by producing better initial cast molds. Methods which have been developed to aid in this pursuit include vacuum casting[13] or a three to four stage alginate casting technique.[32]

Another method to improve fit from the initial casting is to work toward a more dynamic casting method. As the casting is predominantly done under non-weight bearing conditions, working toward more "dynamic" casting methods which equalize the weight bearing pressures is warranted consideration. Where an Ertl procedure has been performed, distal weight bearing casting is preferred to achieve maximum results. The same intent should be attempted with the non-Ertl distal end as well. Ultimately, the better the quality of the cast and the less initial modification guesswork, the better the test socket fitting.

TEST SOCKET

Use of clear test sockets for improving fit is well documented in the literature cited. Though the technology for transparent test sockets has been available since the 1950's, the current practice of direct weight bearing modifications to the socket are relatively new.[23,25,27,32,33]

During the initial static weight bearing period, areas of the residual limb are demarcated according to weight distribution and, therefore, load. This is evidenced by varying degrees of blanching or redness. The goal of a total tissue bearing socket is then pursued to decrease areas of excessive pressure (blanching) and to increase areas of inadequate loading (redness). This goal can be met through either static or dynamic test socket volume changes, or cast model modifications.

Under weight bearing conditions, loose areas are marked by redness, and tension analysis is accomplished via "poking" the tissue through holes made in the socket. Various injectable materials (glycerine, alginate, pour-a-pad) are then added to equalize weight bearing pressures. Areas of excessive weight bearing, if not relieved by the weight borne by the newly injected materials, are either relieved in the socket or modified on the master mold.

By achieving a careful stump-socket interface tension analysis as described, greater confidence in he ultimate result and an optimum fit is possible. Difficulty of fit dictates the number of check socket fittings. Unfortunately, fittings are also affected by the reimbursement source. The fact is undeniable, however, that a transition to the use of transparent test socket fittings can increase the level of prosthetic expertise and elevate the profession to a higher plateau of fitting success.

DYNAMICS

Advancements in prosthetic componentry and gait analysis techniques, when used in conjunction with improved evaluation tools and fitting methods, provides a greater arsenal for the prosthetist seeking to optimize his patient's abilities. An exciting variety of new techniques are surfacing throughout the country which not only render prosthetics more professionally demanding to the practitioner, but also challenging to the patient. Different socket styles and theoretical bends are adding to current thought and practice.

The above-knee amputee now has a variety of alternatives in not only socket material and construction, but in functional design as well. The Swedish flexible socket offers a lighter weight, more "natural" feeling socket to the AK amputee. It also allows for greater transmission of heat via the polyethylene or Surlyn® material, and therefore a cooler feeling. The flexibility of the socket also encourages physiological muscle activity and provides sensory feedback through the thin material.[11]

Contoured Adducted Trochanteric Controlled Alignment Method (CATCAM) is an exciting new above-knee socket design. Proponents claim it increases comfort secondary to total soft tissue weight bearing, because the ischial tuberosity is no longer on the "seat" of the conventional quadrilateral design, but contained within the socket. The CATCAM also allows for more natural muscle activity by virtue of both the flexible design (a la Swedish flexible socket) and inherent socket mechanics. By improving the socket's purchase on the femur, whereby the ischium, trochanter, and adductor longus tendon are in essence "locked-in," stabilization increases, which then decreases the Trendelenberg tendencies experienced by many above-knee amputees. By obtaining a definite position of adduction of the femur, one can take advantage of the muscle stretch of the gluteus medius and therefore increase pelvic control with unilateral weight bearing.[28]

Ultralight weight components continue to be preferred in the above-knee prosthesis. The availability of titanium, carbon graphite, and higher density plastics in the manufacturing of the pylons, joints, and attachment plates allow for lighter weight limbs and, ultimately, decreased energy costs for the amputee.

The below-knee amputee has a varied repertoire of options. A greater array of suspension methods—latex rubber, neoprene sleeves, total suction prostheses—are now available. The Flex-foot prosthesis[18] utilizes a sleeve suspension and is comprised of a carbon graphite and fiberglass pylon and a heel that is very strong, light weight, waterproof, and energy cost effective. The Flex-foot design provides "stored energy" upon weight bearing that "propels" the amputee forward, mimicking "normal" muscle activity in gait. This can also be used for the above-knee amputee. The Flex-foot is proving to be a great advance toward increasing the abilities of the athletic amputee and shows great promise for the elderly and less physically challenged.

New liner materials have also provided alternatives for the below-knee amputee, with greater comfort as a result. Silicone gel and leather liners,[9] Ipocon gel,[14] and injection molded silicone gel liners[16] offer the amputee who has minimal tissue coverage and/or scarring the benefit of shock absorption and a "new skin" type feel. The active, athletic below-knee amputee also captures the benefit of the gel system and suffers less trauma as a result.

Prosthetic feet, such as the Seattle[5] foot and S.A.F.E.[6] foot, appear to offer better gait characteristics and function, and also allow for increased activity by virtue of their functional, flexible designs.

Ancillary methods of evaluating and improving gait performance are making their way into the more aggressive practices. John Sabolich, C.P.O.† in Oklahoma City has been utilizing a bio-feedback device with his above-knee patients in an attempt to re-educate the gluteus medius muscle during gait training. Utilizing the system in a dynamic fashion, i.e. patient ambulating with the electrodes over the targeted muscle, provides the patient audible feedback of muscle activity.

Use of a video tape camera also provides patients with optimum benefits during the alignment and gait training period.[32] Careful analysis of the saggital and frontal views provides the practitioner with a better opportunity to critically analyze and improve his patient's gait. Improved problem-solving subsequent to delivery is also a benefit of this technique. The film serves as a learning tool for the new amputee and the practitioner, and also serves as a record of a patient's progress and delivery status for ironing out future fitting problems relative to gait induced complaints.

The Computer Aided Design, Computer Aided Manufacturing (CADCAM) technique[29,30] is presently available for use in designing below-knee prosthetic sockets and will soon be available for design of above-knee prosthetic sockets as well. Measurements are taken from the residual limb and entered into the program. A screen display then allows for modifications to be made relative to the entered data and design scheme. Once the design is created, the information is transmitted to a computerized milling device that then carves out a model of the residual limb. From this model a socket is fabricated from polypropylene.

In the future, "shape-sensing" will allow for modifications from the sensed data rather than the standard methodology. The ability to draw from the digitalized information of Computerized Axial Tomography (CATSCAN) or x-rays is also in the offing. This system is also an excellent, accurate record keeping tool. The potential to "sense" size and shape, store the information, recall, modify, or duplicate as desired is an enticing prospect. Further research is both warranted and forthcoming.

CONCLUSION

With the advent of better technology and methods, a concomitant increase in prosthetic professionalism occurs. Improved education must also follow. Industry-wide attention to continuing the trend will help prevent our field from lapsing into the mundane.

The practice of this increased professionalism and improved techniques also commands a higher cost. Jan Stakosa, C.P.'s‡ method of using a wide variety of componentry per patient during the fitting and alignment phases in order to optimize function not only serves to improve the patient's quality of life, but carries with it an increased time commitment and cost. Due to this increased input and component variability, thorough education of the public and professionals per the costs involved is required. Ultimately, third party payers and the government will also have to be addressed. Until such time as these practices and advancements become standard, there will not be reimbursement for them.[36]

How do you value human needs in a marketplace in which the trend is toward price reduction? The reality is that all these advances will increase the cost of prosthetic care. Prosthetists, the public, third party payers, and the government will need to be willing to improve the quality of life for this sector of the population, who deserve to be rehabilitated to the maximum and be allowed to perform as well as any able-bodied individual.

It is our hope that the prosthetic industry will take up the challenge to advance the profession and invest the time in testing preferred methods and improvements. Equally important is the quest to participate in their creation. Through improved knowledge of the mechanics of amputation surgery and the variables of follow-up care, combined with mutual professional dialogue, we can better serve the amputee population.

REFERENCES

[1] Barnes, R.W.; Shanik, G.D.; and Slaymaker E.E., "An index of healing in below-knee amputation: Leg blood pressure by Doppler ultrasound," *Surgery 79*(1):13–20, 1976.

[2] Bonner, F.J. and Green, R.F., "Pneumatic airleg prosthesis: Report of 200 cases," *Archives of Physical Medicine and Rehabilitation*, 63:383–385, 1982.

[3] Burgess, E.M., "General principles of amputation surgery," *Atlas of Limb Prosthetics: Surgical and Prosthetic Principles*, St. Louis, MO, Mosby, Ch. 2, p.p. 14–18, 1981.

[4] Burgess, E.M., "Postoperative management," *Atlas of Limb Prosthetics: Surgical and Prosthetic Principles*, St. Louis, MO, Mosby, Ch. 3, p.p. 19–23, 1981.

[5] Burgess, E.; Hittenberger, D.; Forsgren, S.; and Lindh, D., "The Seattle foot," *Orthotics and Prosthetics*, 37(1):25–31, 1983.

[6] Campbell, J. and Childs, C., "The S.A.F.E. Foot," *Orthotics and Prosthetics*, 34(3):3–16, 1980.

[7] Cary, J.M. and Thompson, R.G., "Planning for optimum function in amputation surgery," *Atlas of Limb Prosthetics: Surgical and Prosthetic Principles*, St. Louis, MO, Mosby, p. 28, 1981.

[8] Ertl, J., "Uber amputationstumpfe, *Chirurg.*, 20:218, 1949.

[9] Gibbons, G.W.; Wheelock Jr, F.C.; Hoar Jr, C.S., et al, "Predicting success of forefoot amputations in diabetics by noninvasive testing," *Archives of Surgery*, 144:1034, September, 1979.

[10] Graves, J., "Selectively placed silicone gel socket liners," *Orthotics and Prosthetics*, 34(2):21–24, 1980.

[11] Hanak, R., "Fabrication procedures for the ISNY above-knee flexible socket (instruction manual)." Course at New York University, Post-Graduate Medical School, Prosthetics and Orthotics, January, 1984.

[12] Henderson, H.P. and Hackett, M.E.J., "The value of thermography in peripheral vascular disease," *Angiology*, 29:65–71, 1978.

[13] Hittenberger, D.A. and Carpenter, K.L., "A below knee vacuum casting technique," *Orthotics and Prosthetics*, 37(3):15–23, 1983.

[14] *Ipocon Silicon Liner Technical Manual*. IPOS, Luneberg, West Germany.

[15] Kerstein, M.D., "Utilization of an air splint after below-knee amputation," *American Journal of Physical Medicine and Rehabilitation*, 53(3):119–126, 1974.

[16] Koniuk, W., Personal communication. San Francisco Prosthetic-Orthotic Service, Inc., San Francisco, 1985.

[17] La Noue, A.M., "More on Ertl tibiofibular synostosis," *Newsletter . . . Amputee Clinics*, (V)4:3–4, July, 1973.

[18] Leal, J., "The Flex-foot prosthesis" (instruction manual). Presented at UCLA Prosthetics Education Program, Advanced Below Knee Prosthetics Saturation Seminar, October, 1984.

[19] Loon, H.E., "Below-knee amputation surgery," *Selected Articles from Artificial Limbs*, January 1954—Spring 1966. Huntington, NY, Krieger, p.p. 305–318, 1970.

[20] Malone, J.M.; Leal, J.M.; Moore, W.S.; et al., "The Gold Standard for amputation level selection: Xenon-133 clearance," *Journal of Surgical Research*, 30:449–455, 1981.

[21] Malone, J.M.; Moore, W.S.; Leal, J.M. and Childers,

S.J., "Rehabilitation for lower-extremity amputation," *Archives of Surgery,* 116:93–98, January, 1981.

[22] Mehta, K.; Hobson II, R.W.; Jamil, Z; et al., "Fallibility of Doppler ankle pressure in predicting healing of transmetatarsal amputation," *Journal of Surgical Research,* 28:466, 1980.

[23] Mooney, V. and Snelson, R., "Fabrication and application of transparent polycarbonate sockets," *Orthotics and Prosthetics,* 26(1):1–13, 1972.

[24] Moore, W.S.; Henry, R.E.; Malone, J.M.; et al., "Prospective use of Xenon Xe 133 clearance for amputation level selection," *Archives of Surgery,* 116:86–88, January, 1981.

[25] Pike, A.C. and Black, L.K., "The orthoglas transparent test socket—an old idea, a new technology," *Orthotics and Prosthetics,* 36(4):40–43, 1982–83.

[26] Pollack Jr, S.B. and Ernst, C.B., "Use of Doppler pressure measurements in predicting success in amputation of the leg," *American Journal of Surgery,* 139:303, 1980.

[27] Reger, S.I.; Letner, I.E.; Pritham, C.H.; et al., "Applications of transparent sockets," *Orthotics and Prosthetics,* 30(4):35–39, 1976.

[28] Sabolich, J. and Guth, T., "The C.A.T.C.A.M. above knee prosthesis pilot course" (instruction manual). Course at UCLA Prosthetic Education Program, March, 1985.

[29] Saunders, C.G., "Computer-aided socket design: A computer-aided design and manufacturing package for fitting below-knee amputees with sockets," *Medical Engineering Resource Unit,* Shaughnessy Hospital, Vancouver, BC, Canada, March, 1984.

[30] Saunders, C.G. and Fernie, G.R., "Automated prosthetic fitting." Proceedings of the 2nd International Conference on Rehabilitation Engineering, Ottawa, 1984.

[31] Schmitter, E.D., "Surgical principles and practice: Lower Extremity amputations.' Lecture—Prosthetics and Orthotics Course for Physicians and Therapists. Provided by Prosthetic-Orthotic Education Program, School of Medicine, Department of Surgery (Orthopaedics). University of California, Los Angeles, April 5–9, 1982.

[32] Staats, T.B., "Advanced prosthetic techniques for below knee amputations," *Orthopedics,* 8(2):249–258, 1985.

[33] Stakosa, J.J., "Prosthetics for lower limb amputees," *Vascular Surgery: Principles and Techniques,* Norwalk, CT, Appleton-Century-Crofts, p.p. 1143–1162, 1984.

[34] Sterescu, L.E., "Semirigid (Una) dressing of amputations," *Archives of Physical Medicine and Rehabilitation,* 55:433–434, September, 1974.

[35] Varnau, D.; Vinnecour, K.E.; Luth, M.; and Cooney, D.F., "The enhancement of prosthetics through Xeroradiography," *Orthotics and Prosthetics,* 39(1):14–18, 1985.

[36] Whipple, L. and Stakosa, J., "The not so simple ABC's of high technology," *Disabled USA,* Washington, D.C., July, 1983.

[37] Wu, Y.; Keagy, R.D.; Krick, H.J.; et al., "An innovative removable rigid dressing technique for below-the-knee amputation," *Journal of Bone and Joint Surgery,* 61-A(5):724–729, 1979.

AUTHORS

David F.M. Conney, R.P.T., C.P.O., is a senior vice-president at Beverly Hills Prosthetics and Orthotics, Inc.

Keith E. Vinnecour, C.P.O., is owner and president of Beverly Hills Prosthetics Orthotics, Inc., Beverly Hills, California.

---

† John Sabolich, C.P.O., is Vice-President of Sabolich Orthotics-Prosthetics Center, Oklahoma City, Oklahoma.

‡ Jan Stakosa, C.P. is Director of the Institute for the Advancement of Prosthetics, Lansing, Michigan.

Editorial

The New Revolution

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

The recent development and proliferation of advanced and precision fitting techniques in prosthetics have caused many prosthetists to reevaluate those principles which were held sacred for the past twenty years. In the last three years in particular, both below-knee and above-knee prosthetics have undergone tremendous changes.

Many progressive practitioners recognize that the term "Patellar Tendon Bearing (PTB)" is no longer considered descriptive of a well designed below-knee socket and use the term only in a historical sense. The term Total Surface Bearing better describes what has superseded PTB philosophy.

In above-knee prosthetics, a greater revolution is in the offing. Now the CATCAM (Contour-Adducted-Trochanteric-Controlled Alignment Method) socket is shaking the underpinnings of the Quadrilateral above-knee socket design. For those of us who are "dyed-in-blue-and-gold-UCLA-Quad-socket" prosthetists, it is both difficult and exciting to see the development and confusion a rival design causes throughout the profession. I am sure that thirty years ago the "wood-socket-plug-fit" prosthetists shared a similar feeling when the quadrilateral socket and later the introduction of plastics caused their world to turn upside down.

The point is that change and improvement are inevitable. You can fight it and it will flow over you like a river, or you can go with the flow and learn to adapt to new techniques. I have been asked repeatedly what I think about the use of multiple check socket fittings, CATCAM, alginated check sockets, and the Flex-Foot. The list goes on and on. American prosthetists in particular must understand that we are in the midst of a full blown revolution and the results of this revolution will set the path we follow for the next couple of decades. Rather than question what is right or wrong without really having proof of either, I have chosen a path as the director of a prosthetics education program of "pouring fuel on the fire." What better time or place for controversy than at UCLA, where the first school was started over thirty years ago.

Is all this extra precision and care really necessary to accurately fit an artificial limb? The answer is quite simple, and if you are an amputee the question is repulsive. If superior techniques that can improve the quality of the care provided to amputees are available but are not used, it is nothing less than criminal.

There are those who would question: how much of a good thing is enough? That is a question that the patient must answer and the prosthetist must decide based on knowledge and education. The fact that many of the newer techniques and fitting regimes demand more time and effort than methods which have been in use for twenty years is entirely a separate issue. While it may not be possible to provide these services for the reimbursements, which are now received from payment sources, this does not mean that the techniques do not work or are wrong. It only means that the third party payers are ignorant of changes which have occurred in our profession and must be introduced to the benefits of new procedures.

This same principle applies to prescribing physicians. It is totally fair to say that a physician who took his prosthetics-orthotics training over five years ago is now out of date. The same is true for practitioners who have not upgraded their practices through educational opportunities during this period.

It is always uncomfortable when you begin to wonder whether you are doing the best you can for your patient. It is even more uncomfortable when you know you are not. We should never be satisfied with our work and never doubt that a better job can be done. With such a philosophical upheaval running rampant through our profession, the time for learning is now. Are you satisfied with application of outdated techniques, or are you willing to enter a new era of prosthetic and orthotic practice? The choice is yours.

AUTHOR

Timothy Staats, M.A., C.P., is Adjunct Assistant Professor and Director of the Prosthetics & Orthotics Education Program at UCLA, Rehabilitation Center, 1000 Veteran Avenue, Rm 22-41, Los Angeles, CA 90024.

Editorial

The Role of Test Socket Procedures in Today's Prosthetic Practices

by Michael J. Quigley, C.P.O.

The proper role of a test socket procedure is a controversial topic in today's practice of prosthetics. A test socket procedure can be defined as that stage in the design of a prosthesis when a socket is fabricated solely for the purpose of determining proper socket fit. Although test sockets were originally used for upper limb prostheses, the true advent of the test socket was in 1972 when Mooney and Snelson[1] described the polycarbonate clear test socket as developed at Rancho Los Amigos Hospital. During the 13 years since that article, the proper role of the test socket procedure has still not been defined.

There are several reasons for the controversy over test sockets. First, when a test socket procedure is done, there is an implication that the mold, mold modifications, and socket design principles instilled in the prosthetist may not be correct. After all, if the prosthetist's techniques were perfect, the socket would fit perfectly and the need for a test socket would be obviated. However, any time a clear test socket is used, the prosthetist immediately notices a few things he would like to change in the definitive socket or, in some cases, the next test socket.

It is safe to say that the majority of United States prosthetists believe in the value of test sockets and use them on a regular basis. Indeed, insurance companies and most other third party reimbursers, including Medicare, pay for test sockets, thereby recognizing twin values. A test socket procedure makes good sense, and there is no question that it improves prosthetic fitting. However, it is also true that many prosthetists do not use these sockets, or use them only rarely. The group that does not use test sockets feels that they can fit nearly all prostheses well without test sockets and do not want to spend the additional effort that test sockets require, or they simply do not want to change the methods they learned many years ago. The present Veterans Administration's (VA) procedure for obtaining approval for test sockets seems to favor this latter group, since it is an intentionally cumbersome system that, in effect, discourages test socket procedures on VA patients.

Test socket users also include prosthetists who routinely use multiple test sockets on every patient, with the principle that each successive socket brings you one step closer to the perfect fit. If one test socket procedure is good, shouldn't two be better? Or three? Or more? This is a major area of controversy that could be discussed here but not resolved. Probably the best example of this use of test sockets is at the Institute for the Advancement of Prosthetics (IAP) in Lansing, Michigan (although a number of other prosthetic practices are also using multiple test sockets, or featuring them as a type of "first class" service).

An average of six test socket procedures are done on each patient in Lansing: beginning with static fittings in clear sockets with the patient now. Are you satisfied with application of outdated techniques, or are you willing to enter a new era of prosthetic and orthotic practice? The choice is yours.

AUTHOR

Timothy Staats, M.A., C.P., is Adjunct Assistant Professor and Director of the Prosthetics & Orthotics Education Program at UCLA, Rehabilitation Center, 1000 Veteran Avenue, Rm 22-41, Los Angeles, CA 90024.

Editorial

The Role of Test Socket Procedures in Today's Prosthetic Practices

by Michael J. Quigley, C.P.O.

The proper role of a test socket procedure is a controversial topic in today's practice of prosthetics. A test socket procedure can be defined as that stage in the design of a prosthesis when a socket is fabricated solely for the purpose of determining proper socket fit. Although test sockets were originally used for upper limb prostheses, the true advent of the test socket was in 1972 when Mooney and Snelson[1] described the polycarbonate clear test socket as developed at Rancho Los Amigos Hospital. During the 13 years since that article, the proper role of the test socket procedure has still not been defined.

There are several reasons for the controversy over test sockets. First, when a test socket procedure is done, there is an implication that the mold, mold modifications, and socket design principles instilled in the prosthetist may not be correct. After all, if the prosthetist's techniques were perfect, the socket would fit perfectly and the need for a test socket would be obviated. However, any time a clear test socket is used, the prosthetist immediately notices a few things he would like to change in the definitive socket or, in some cases, the next test socket.

It is safe to say that the majority of United States prosthetists believe in the value of test sockets and use them on a regular basis. Indeed, insurance companies and most other third party reimbursers, including Medicare, pay for test sockets, thereby recognizing twin values. A test socket procedure makes good sense, and there is no question that it improves prosthetic fitting. However, it is also true that many prosthetists do not use these sockets, or use them only rarely. The group that does not use test sockets feels that they can fit nearly all prostheses well without test sockets and do not want to spend the additional effort that test sockets require, or they simply do not want to change the methods they learned many years ago. The present Veterans Administration's (VA) procedure for obtaining approval for test sockets seems to favor this latter group, since it is an intentionally cumbersome system that, in effect, discourages test socket procedures on VA patients.

Test socket users also include prosthetists who routinely use multiple test sockets on every patient, with the principle that each successive socket brings you one step closer to the perfect fit. If one test socket procedure is good, shouldn't two be better? Or three? Or more? This is a major area of controversy that could be discussed here but not resolved. Probably the best example of this use of test sockets is at the Institute for the Advancement of Prosthetics (IAP) in Lansing, Michigan (although a number of other prosthetic practices are also using multiple test sockets, or featuring them as a type of "first class" service).

An average of six test socket procedures are done on each patient in Lansing: beginning with static fittings in clear sockets with the patient wearing no prosthetic socks, going on to clear socket dynamic (walking) fittings, and progressing to a definitive socket with a gel liner for below knee amputees. The patient is seen every day for two to three weeks until the socket fit is perfected, and only then is the prosthesis finished. This, of course, is an expensive undertaking, but it seems logical to assume that with so much time and energy spent, the patient would end up with a better fit. The multiple test socket users lead an utopian existence, seeking the perfect fit, and see only one or two patients every week or continue fittings for many months if they are seen only on a weekly basis. For the average prosthetist who fits 100 or more patients alone each year, the sheer logistics of using multiple test sockets on every patient is staggering.

Another area of controversy regarding test sockets is that they provide an incentive for prosthetists to delay being satisfied with the socket fit if they are paid separately for every test socket used. If they fit six test sockets, they are paid six times more than if they fit one test socket. The only response to this problem is that there are a few difficult cases where the only way a good fitting can be achieved is with multiple sockets, and the prosthetist should be reimbursed for his effort. On the other hand, there are always the few people who will abuse the system. In practice, less than five percent of all prosthetists have the time or inclination to routinely use multiple test sockets. After all, there are also very few patients or insurers who want to bear the expense, are able to make all the appointments necessary, and are willing to wait the many months for a finished prosthesis when multiple test sockets are used.

Before summarizing, one final comment is necessary. Having test socket procedures available and using test sockets properly are two different things. There are no standardized, recommended, or documented procedures for the proper use of a test socket. Some people use clear sockets, some do not. Some use "wet fit" procedures with no prosthetics socks; others use prosthetic socks. Some test sockets are used statically, others dynamically. Alginate procedures are used in some areas. Even when a clear socket is used directly against the skin, how do we interpret what we are seeing? The result of the confusion over the proper use of a test socket is that the prosthetist converts one of the few objective tools he has available (a clear socket) into a subjective one by having to use educated guesses to determine the modifications needed to improve socket fit. The whole area concerning the optimum use of test socket procedures is in great need of study and documentation.

In summary, test socket procedures are good procedures. When a prosthetist knows that the socket he is fitting is not the final product, he is more likely to make major socket modifications and, therefore, less likely to provide a poor fitting prostheses. Multiple or successive test sockets will always be required on a few difficult cases. In some areas, the patient and prosthetist will afford the luxury to use successive test sockets to try to achieve the perfect fit, but this will probably include less than one percent of the patient population.

It is obvious that these socket procedures are here to stay and that the use of test sockets will increase as new materials and techniques are introduced. Hopefully, some meaningful documentation will be developed to enable prosthetists to obtain as much information as possible from a test socket procedure. Without a true understanding of how to properly use a test socket, each prosthetist is left to practice and develop his own technique, and the art of prosthetics again overwhelms the science.

REFERENCES

[1] Mooney, V. and Snelson, R., "Fabrications and Applications of Transparent Polycarbonate Sockets," *Orthotics and Prosthetics*, 26 (1), p.p. 1–13, March 1972.

[2] Personal communication, Jan Stokosa, C.P., May 6, 1985, Lansing, Michigan.

AUTHOR

Michael J. Quigley, C.P.O. is President of Oakbrook Orthopedic Services, Ltd. 1 South 132 Summit Ave, Ste 102, Oakbrook Terrace, IL 60181.

A Below-Knee Weight-Bearing Pressure-Formed Socket Technique

by Robert F. Hayes, C.P.

I'm pleased that the Academy has requested that I update and rewrite the Below-Knee Weight-Bearing, Pressure-Formed Socket Technique article I originally wrote in 1975. It's hard to believe that ten years have passed since the original publication of this paper.

I haven't made any significant changes regarding the principles or application of this procedure, but let's go back to some of the reasons this concept was developed.

As I explained in the original article, my son was being fitted for ski boots and it occurred to me that we might make use of some of the techniques used by ski boot designers. The ski boot had an inflatable inner bladder. With the foot under weight-bearing, a conforming material similar to certain silicone compounds was injected into the bladder to give a perfect form-fitting in the attitude of weight-bearing. The incentive to apply this technique to limb prosthetics was reinforced while I was casting a below-knee patient who was a dentist. We exchanged thoughts on molds and changes when pressures are applied. Dentists take one mold for a cast which is filled with dental impression cream (similar to alginate). This is applied to the patient under pressure to give a more accurate impression, and then this is filled to form the definitive positive mold.

The standard method of fitting a below-knee amputee involves taking a negative cast in a non-weight-bearing condition, forming a positive model, modifying it in size to present dimensions by removing material to create pressure, and applying material to relieve pressure on the stump as appropriate. A socket is then molded over this model with the hope that, with small adjustments, it will fit the patient.

Wouldn't it be wiser to develop a socket under pressure that will adjust to and fit the patient, rather than fit the patient to the socket? In trying to answer this question, the procedure described here was developed.

THE PROCEDURE

Measure the patient in the usual manner. Place a sheet of plastic wrap material, such as Saran, over the patient's stump to keep it clean of indelible pencil, and to make removal of the cast easy. If a wool sock is to be used, apply it, and then apply the plastic wrap. Apply a cast sock or tube gauze over the stump. Bond 1/4″ felt over all pressure-sensitive areas: the crest of the tibia and the head of the fibula.

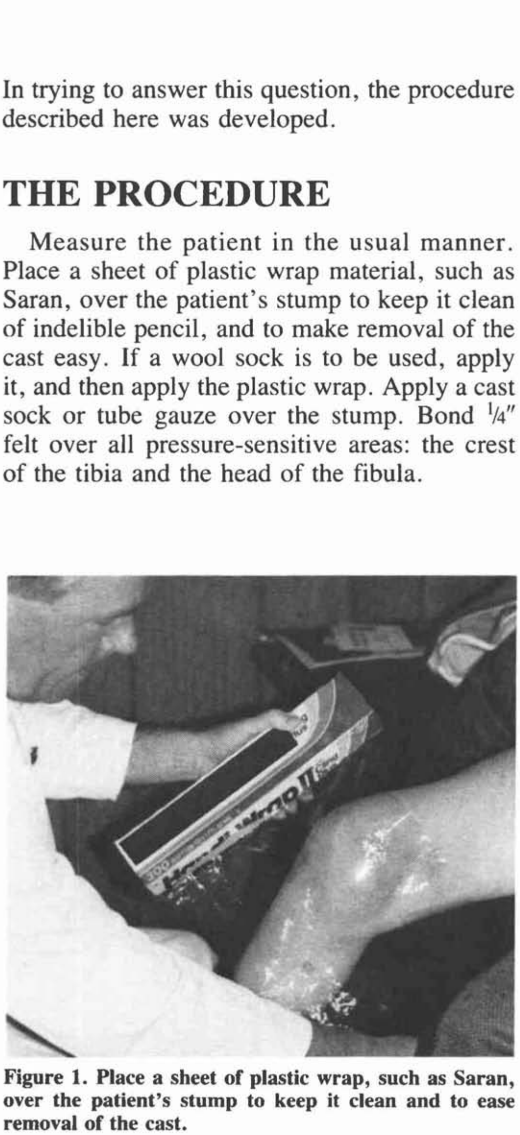


Figure 1. Place a sheet of plastic wrap, such as Saran, over the patient's stump to keep it clean and to ease removal of the cast.


Figure 2. Apply cast sock and felt relief pads.


Figure 3. Using Plaster-of-Paris, wrap the residual limb in the usual manner.

Keep in mind that all areas being built up will be filled with alginate to give contact, yet minimizing pressure. There are some prosthetists who have adopted this technique and claim they apply direct pressure over the crest of the tibia. My experiences do not agree with that. In particular, since many of our patients are diabetic with very thin skin, extra caution should be taken to reduce pressure and especially friction over all bony prominences.

Using elastic Plaster-of-Paris bandage 4″ wide, wrap the stump in the usual manner, and reinforce with 3″ or 4″ wide regular plaster bandage. Remove the cast and remove the tube gauze and felt buildup from the negative cast.

Pour the positive model, remove the negative cast, and modify in the usual manner, but do not touch areas that were covered with felt. Build up the distal end of the positive model at least 1″.


Figure 5. Pour and modify the positive model.


Figure 6. Build up the distal end of the positive model at least 1″.

Figure 4 (left). Remove the tube gauze and felt buildup from the negative cast.

Make a check socket. This is a perfect application for vacuum-forming. Plaster bandages or laminates can, of course, be used. Drill two holes 1/4" in diameter in the distal end and rough up the inside surface of the socket. For the first fitting, apply the stump sock of choice, and place plastic "wrap" over the stump sock to act as a separator. An invaginated balloon will not work because it adheres to the alginate that is to be used later.

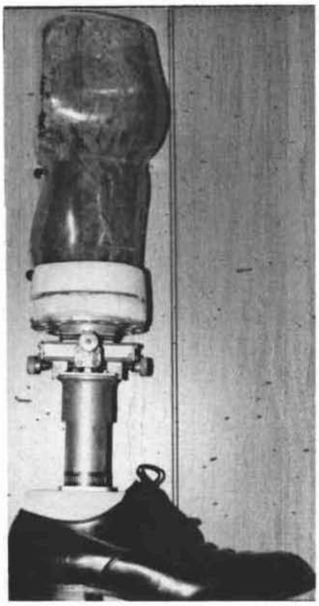


Figure 7. Set up the transparent check socket for dynamic alignment.

Some prosthetists apply the check socket to the patient's bare stump (no socks) for visual inspection. It bothers me to think what happens to the fit of this socket when the prosthesis is finished from this exact mold and the patient applies the usual stump sock of 3-ply or 5-ply. When the check socket is applied on new patients, I recommend using a thin-fitting sock in anticipation of stump atrophy. On seasoned, well-shaped stumps, I use the same sock that the patient usually wears. When using inserts that tend to compress, i.e. Pelite®, you may use a 3-ply and, after several weeks of prosthetic use, the socket should accommodate a 5-ply sock. Of course, there are many factors to be considered, and this is the area where the prosthetist's knowledge and experience will play the major role as to how well his or her patient does.

Mix about 1/2 pint of dental impression cream or alginate (which is more economical). Pour about 1/3 of the total amount in the distal part of the socket and, with a spatula, spread the rest around the remaining surface of the socket. It is necessary to work quickly at this point.

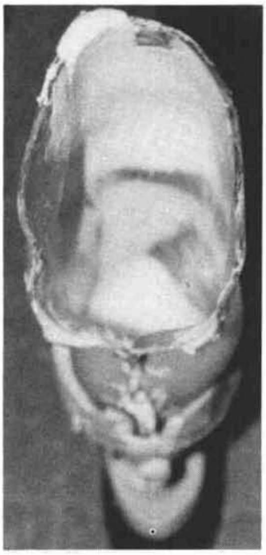
Figure 8. Pour the alginate and let it escape through the distal holes until the patient is lowered into the socket to the proper level at which time the holes are blocked. The alginate will then escape along the proximal brim of the socket.

Place the socket on a fitting stool adjusted for height. Use some sort of pad to prevent slipping and cover the drilled holes in the socket with your thumb and forefinger. Have the patient place his stump in the socket. Let the alginate escape through the distal holes until the patient is lowered into the socket to the proper level, at which time the holes are blocked. Alginate will now escape along the proximal brim of the socket.

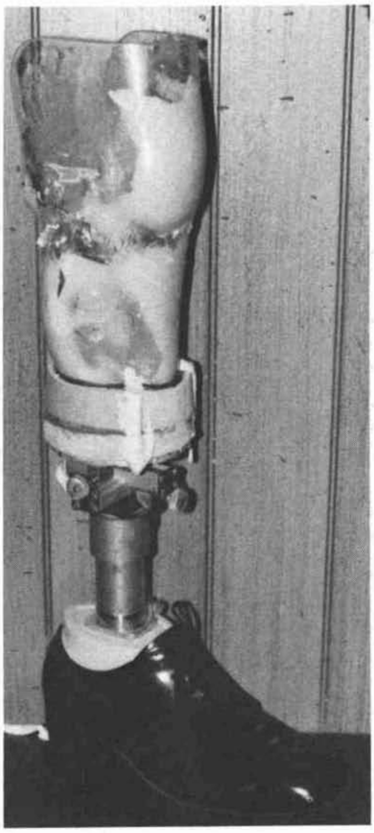
As soon as the alginate has set up, remove the stump from the socket and immediately fill the socket with plaster. The rigid socket and alginate are removed by using a cast cutter. The mold resulting is a perfectly smooth, pressure-formed, positive mold that can be used in any method of fabrication desired.

When this technique is used, patients can be fit with sockets without soft liners.

Only a minimal amount of additional time is required. I feel that the technique allows better fitting of "problem" stumps and that it may be used as a routine procedure to advantage, especially in central fabrication systems. Vacuum-forming procedures recently introduced make this approach to fitting even more attractive. We have since switched to clear plastic check sockets for the obvious advantage of visual inspection and also the ability to adjust check socket pressure areas with a heat gun on some plastics. We also now fit the check socket on the adjustable leg, rather than the fitting stool. This better simulates the pressures exerted on the stump by the definitive prosthesis, since we



Figures 9 (left) and 10 (right). The completed socket.



all agree that socket alignment greatly affects the application of pressure.

I know that this procedure has been used by many prosthetists in various parts of the country, and I have received many favorable comments about the benefits to the patient. This pleases me because this is the goal of the process. And I'm sure that in the future new devices and innovations will continue to add to and improve this concept to even greater benefit of the patient.

AUTHOR

Robert F. Hayes, C.P., is President of Hayes Prosthetics, Inc., 1309 Riverdale Street, West Springfield, Massachusetts 01089.

The following series of articles on Gait Analysis were based on a project which was supported by the Newington Children's Hospital Research Fund.

Gait Analysis

INTRODUCTION

by Ronald F. Altman, C.P.O.†

The following series of articles all have to do with using gait analysis, in orthotics as well as prosthetics, to improve function. The Gage/Hicks study traces gait analysis in prosthetics from Inman forward, and the individual articles illustrate contemporary laboratory approaches to the objective assessment of gait.

Fundamental to optimal lower-extremity prosthetic/orthotic service is an analysis of the gait of the patient. To the extent the method of analysis fails to provide adequate objective or useful information about gait, it allows for the possibility and probability that a less than optimum fit and/or alignment configuration has been or will be achieved.

While gait analysis has long been an established procedure of varying objectivity in prosthetics, in orthotics the use of gait analysis has been rather ineffectual in assisting to optimize gait, a process which for the most part fails to go beyond a most rudimentary observation. This is due in part to the rudimentary functional characteristics of most orthoses.

Advances in our profession as well as technology and materials can and do result in more functional orthoses. If we are going to provide the optimal orthotic design configuration for any given patient, it is essential that we define gait characteristics more precisely and reliably.

Though not yet universally available, the increasing number of gait analysis facilities will soon benefit us all—patients and practitioners alike—as we gain access to the resulting information flow in formats readily usable by orthotists and prosthetists.

† Director of Orthotics/Prosthetics Department at Newington Children's Hospital in Newington, Connecticut.

Gait Analysis in Prosthetics

by James R. Gage, M.D.
Ramona Hicks, R.P.T., M.A.

REVIEW

Objective measurement systems which quantify locomotion have been in use for the past century. But not until World War II, when thousands of men returned home to the United States with amputations, was technology really applied to the understanding of prosthetic gait.

Inman and colleagues[1] founded the Biomechanics Laboratory at the University of California to establish fundamental principles of human walking, particularly in relation to problems faced by lower limb amputees. Inman's measurement techniques included motion pictures of coronal and sagittal views, as well as transverse rotations from below using a glass walkway. Using interrupted light photography, the Biomechanics Laboratory team studied the motion of body segments during gait. Force plates measured the subject's ground reaction forces, and muscle activity was recorded using electromyography (EMG), which measures the electrical signals associated with contraction of a muscle. Prior to Inman's fundamental studies, *The following series of articles on Gait Analysis were based on a project which was supported by the Newington Children's Hospital Research Fund.*

Gait Analysis

INTRODUCTION

by Ronald F. Altman, C.P.O.†

The following series of articles all have to do with using gait analysis, in orthotics as well as prosthetics, to improve function. The Gage/Hicks study traces gait analysis in prosthetics from Inman forward, and the individual articles illustrate contemporary laboratory approaches to the objective assessment of gait.

Fundamental to optimal lower-extremity prosthetic/orthotic service is an analysis of the gait of the patient. To the extent the method of analysis fails to provide adequate objective or useful information about gait, it allows for the possibility and probability that a less than optimum fit and/or alignment configuration has been or will be achieved.

While gait analysis has long been an established procedure of varying objectivity in prosthetics, in orthotics the use of gait analysis has been rather ineffectual in assisting to optimize gait, a process which for the most part fails to go beyond a most rudimentary observation. This is due in part to the rudimentary functional characteristics of most orthoses.

Advances in our profession as well as technology and materials can and do result in more functional orthoses. If we are going to provide the optimal orthotic design configuration for any given patient, it is essential that we define gait characteristics more precisely and reliably.

Though not yet universally available, the increasing number of gait analysis facilities will soon benefit us all—patients and practitioners alike—as we gain access to the resulting information flow in formats readily usable by orthotists and prosthetists.

† Director of Orthotics/Prosthetics Department at Newington Children's Hospital in Newington, Connecticut.

Gait Analysis in Prosthetics

by James R. Gage, M.D.
Ramona Hicks, R.P.T., M.A.

REVIEW

Objective measurement systems which quantify locomotion have been in use for the past century. But not until World War II, when thousands of men returned home to the United States with amputations, was technology really applied to the understanding of prosthetic gait.

Inman and colleagues[1] founded the Biomechanics Laboratory at the University of California to establish fundamental principles of human walking, particularly in relation to problems faced by lower limb amputees. Inman's measurement techniques included motion pictures of coronal and sagittal views, as well as transverse rotations from below using a glass walkway. Using interrupted light photography, the Biomechanics Laboratory team studied the motion of body segments during gait. Force plates measured the subject's ground reaction forces, and muscle activity was recorded using electromyography (EMG), which measures the electrical signals associated with contraction of a muscle. Prior to Inman's fundamental studies, prostheses were customized for the individual amputee, without any particular regard to rational structural design. Inman's goal was to provide fundamental data essential for the design of prosthetic limbs. By analyzing normal human walking, he and his colleagues laid the groundwork for biomechanical analysis of amputee gait.[2] Since that time, numerous techniques have been developed to study human locomotion,[3] and numerous studies have been undertaken to evaluate prosthetic gait.

RESEARCH APPLICATIONS

Eberhart, et al.[4] described the locomotor mechanism of the above-knee amputee from kinematic and kinetic data. They compared lateral stick figures of amputees to normal subjects as a means to objectively identify gait deviations in the sagittal plane. Force plate data were used to compare the weight-bearing characteristics of the prosthetic limb and the sound limb. From these comparisons, the authors identified amputees who walked well with their prostheses and those who were less adept. Eberhart believed that ultimately "optimal" patterns of gait could be determined for amputees and used as a reference for evaluating prosthetic gait.

Zuniga, et al.[5] studied gait in 20 above-knee amputees by using electrogoniometers attached to the knee and foot switches. Their data documented asymmetry in the stance and swing phase times between the prosthetic and sound limb.

In similar investigations, James and Oberg[6] and Murray, et al.[7] studied temporal stride parameters and knee flexion-extension angles, and also examined above-knee gait at various speeds. They confirmed the stance and swing phase asymmetry between the prosthetic and sound limb. They also showed that the asymmetry was present regardless of the speed of walking.

The collection of baseline data in above-knee amputees clearly demonstrated some shortcomings in prosthetic gait. One of these, the longer swing time which is required on the prosthetic side, has led to the development of dozens of prosthetic knees. Gait analysis laboratories have been used to evaluate some of these prosthetic designs. Godfrey, et al.,[8] in a limited study that compared gait with six cadence-responsive knee units, found no significant differences among them. Murray, et al.[9] compared the gait of above-knee amputees with hydraulic knee units versus constant friction knee units. Temporal and kinematic data, which were collected at slow, free, and fast speeds, showed that the hydraulic knees improved the symmetry between the prosthetic limb and the sound limb, especially at the fast and free speeds. This finding was true for both cadence and the amount of knee-flexion at swing phase.

Hoy and colleagues,[10] in one of the few studies on gait in juvenile amputees, collected kinematic data at various speeds to compare the solid ankle cushioned heel (SACH) foot to a Child Amputee Prosthetic Project (CAPP) experimental foot. The authors found hip range of motion to be closer to normal and significantly less with the CAPP foot than the SACH foot.

Hannah and Morrison[11] studied the effect of alignment of the below-knee prosthesis on gait. Using electrogoniometers to measure hip and knee joint rotations in the coronal, sagittal, and transverse planes, they found that malalignment of the prosthetic foot was the most crucial for gait symmetry.

Grevsten and Stalberg[12] used electromyography to compare muscle activity in below-knee amputees walking with patellar tendon-bearing (PTB) and PTB-suction prostheses. Surface electrodes were placed over the tibialis anterior and gastrocnemius muscles which, in normal gait, usually fire at opposite phases. The data showed that these muscles contracted for longer periods when the PTB prosthesis was used than with the PTB-suction prosthesis, suggesting that the suction mechanism improved the adaptation to the prosthesis.

Thiele, et al.[13] investigated possible neurophysiological reasons for weakness in above-knee amputees by recording electromyographic activity of the quadriceps during gait. They did not find abnormal recordings and concluded that muscle weakness was secondary to biomechanical, rather than neurophysiological, factors.

CLINICAL APPLICATIONS

Until the present, gait analysis has been applied to prosthetics only for research purposes. Routine prosthetic fitting and checkout are still done by means of observational gait analysis. However, observational gait analysis has many disadvantages.

In the first place, even normal human walking is extremely complex. With each step, more than 30 major muscles have to contract and/or relax synchronously in each lower extremity. Also, normal human gait is rapid (approximately 105 steps per minute), and the human eye is not fast enough to separate the various components of gait at this speed. Krebs, et al.[14] have shown that data vary widely when different examiners have observed a person's gait and that observational analysis is only a moderately reliable technique. The variations between observers may be due to the preconceptions of individual observers, to limitations of human perception, or to problems in transmitting the information or data to colleagues. In light of these findings, it is not surprising that the fit and quality of the limbs fabricated by different prosthetists vary greatly.

Technology has now progressed to the point where automated gait laboratories can be built. Their capabilities vary, but most labs monitor one or more of the following parameters:

1) kinematics or movement measurements through a motion analysis system,
2) evaluation of ground reaction forces via force plates or pressure sensitive switches, and
3) dynamic electromyography (monitoring the electrical activity of contracting muscles).

The advantage of an automated motion measurement system is that automated data entry and rapid processing allow routine clinical use at a reasonable cost. Since the sampling rate of most automated motion systems is in excess of 50 Hz (50 samples/second), all movement in the lower extremities during walking can be examined in detail and with excellent reproducibility.

Thus, the analysis of walking becomes objective, rather than subjective, and a record of this objective analysis is produced by the computer in such a fashion that preconceived biases and communication errors between observers are minimized. Furthermore, some of the more modern gait analysis facilities have the ability to compare records, for example, of a patient's gait pre- and post-operatively, or of an amputee's gait with two different prosthetic devices or components. Through comparisons like these, the presence or absence of benefit can be determined objectively.

KINESIOLOGY

The field of prosthetics can make use of the new science of kinesiology, or the study of movement. Kinesiology consists of two major fields:

1) kinematics, the study of motion exclusive of the influences of mass or forces, i.e., without regard to the underlying cause of the motion; and
2) kinetics, which deals with the forces that produce motion.

Kinematic Data

Kinematic data can be gathered in a variety of ways—through interrupted light photography, cinefilm, video systems, and/or electrogoniometers—and it can be displayed in many ways. Stick figures provide a visual display of the subject walking.

Figure 1 is a stick figure representation of an 11 year old girl with a right knee disarticulation. The stick figures facilitate the identification of gait deviations, e.g., knee hyperextension on the prosthetic side at stance phase. With

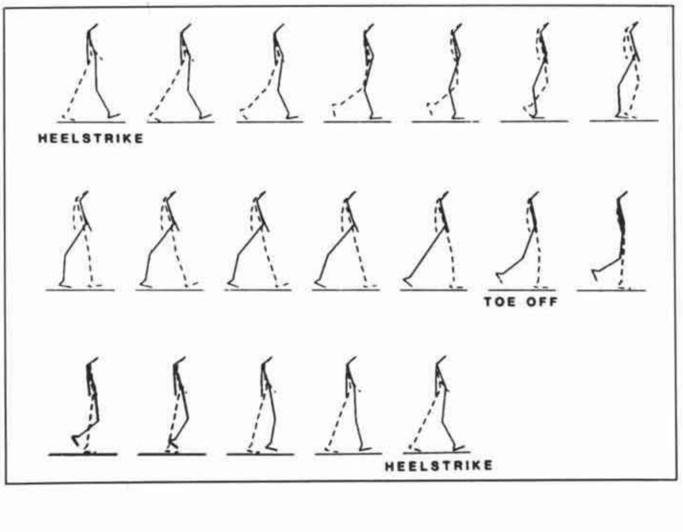


Figure 1. Lateral stick figures of the right gait cycle of an 11-year-old-girl with a right knee disarticulation.

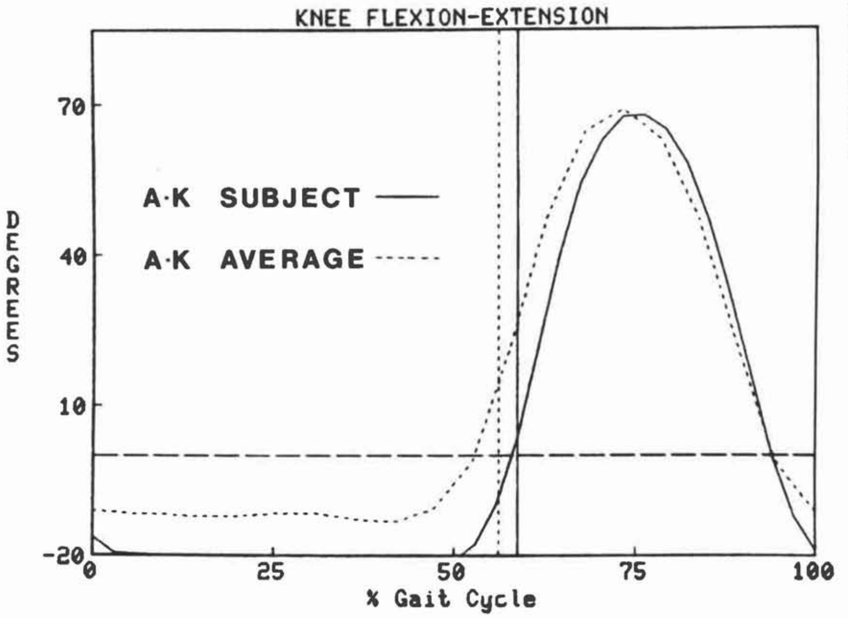


Figure 2. Comparison of knee flexion-extension motion in one above-knee amputee with an average composite of knee flexion-extension in seven above-knee amputees.

observational gait analysis, this gait deviation might be missed, or two examiners might argue about its presence. With objective gait analysis, we can prove the deviation's existence by viewing the stick figures, and we can identify the cause of the deviation by reviewing the graphs that depict motion. These graphs display motions of each joint of the lower extremities in all three planes during a representative gait cycle.

Figure 2 is a graph showing knee flexion-extension of the same child with a knee disarticulation. The child's sagittal knee motion is compared with the mean or average flexion-extension of seven other above-knee amputees. Although all above-knee amputees hyperextend their knees slightly during stance phase, this patient has 10 degrees more hyperextension than average. Following the gait analysis, it was discovered that the knee extension bumper was too soft, and it was replaced with a stiffer one.

Kinematic data can also be used to compute temporal data, such as stride length, cadence, and walking velocity. Figure 3 compares the temporal data of the child with knee disarticulation with "normal" children the same age. Notice that the stride length is normal but that the walking velocity and cadence are less than normal.

Linear Measurements

	AMPUTEE (knee disartic.)	NORMAL CHILDREN (ages 8–14)
Opposite Toe Off (%)	11.76	10.40
Opposite Heel Strike (%)	50.00	49.20
Single Stance (%)	38.24	38.86
Toe Off (%)	58.82	60.20
Step Length (cm)	53.80	59.94
Stride Length (cm)	117.50	118.66
Gait Cycle Time (sec)	1.17	1.02
Cadence (steps/min)	104.37	118.21
Walking Speed (cm/sec)	102.17	116.89

Figure 3. Linear measurements of an 11-year-old girl with a knee disarticulation compared with a composite of linear measurements of normal children.

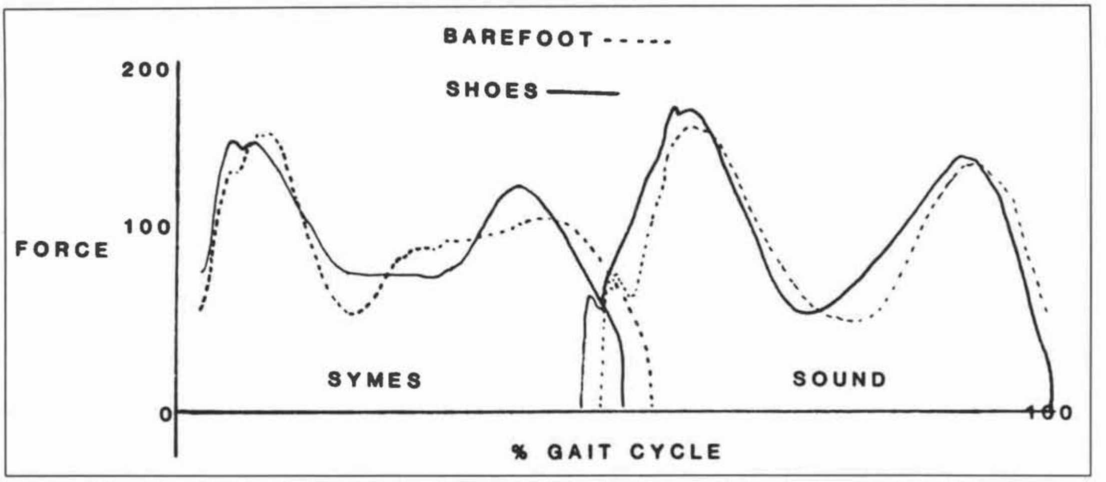


Figure 4. Graphic display of the vertical ground reaction forces in a 9-year-old boy with and without shoes.

Kinetic Data

The forces that cause movement are usually collected through pressure sensitive switches or paper, or with commercial force plates, which are designed to break down the ground reaction forces into their components (X,Y,Z force, and X,Y,Z moment). The software of a modern gait analysis laboratory is able to combine force plate data with motion analysis data to produce meaningful graphic outputs.

Figure 4 shows the vertical ground reaction force (Z force) for walking barefoot compared with walking with shoes in a 9 year old boy with a Symes prosthesis. Notice the improved symmetry at push-off between the prosthetic and sound limb when shoes are worn.

Force plates can also be used to compute the location of the center of pressure on the foot. Figure 5 compares the foot force progression pattern of a SACH foot to a multi-axis foot in a 27 year old male with a below-knee amputation. From these data, one can see that the foot force progression pattern is more lateral with the multi-axis foot than with the SACH foot. Also, notice with the SACH foot how the initial forces move from an anterior to posterior direction as the heel compresses. This pattern is not seen in the multi-axis foot.

DYNAMIC ELECTROMYOGRAPHY

Dynamic electromyography is a valuable tool for measuring the time duration of muscle activity, which is recorded through electrodes, either surface or indwelling. However, since voluntary muscle activity results in an electromyographic recording that increases in magnitude with the tension, other variables can also influence the signal, limiting the accuracy of EMG as a predictor of muscle tension.

Electromyographic data can be displayed in several ways. When used to analyze a gait cycle, the data show which muscles are active during each phase of gait. Figure 6 compares muscle activity during gait of the subject walking with the SACH foot compared with the multi-axis foot. The hamstrings and quadriceps

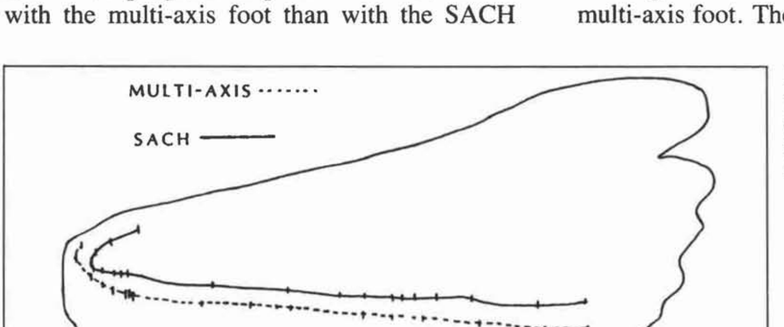


Figure 5. Path of the center of pressure on the foot in a 27-year-old below-knee amputee with a SACH foot and with a multi-axis foot.

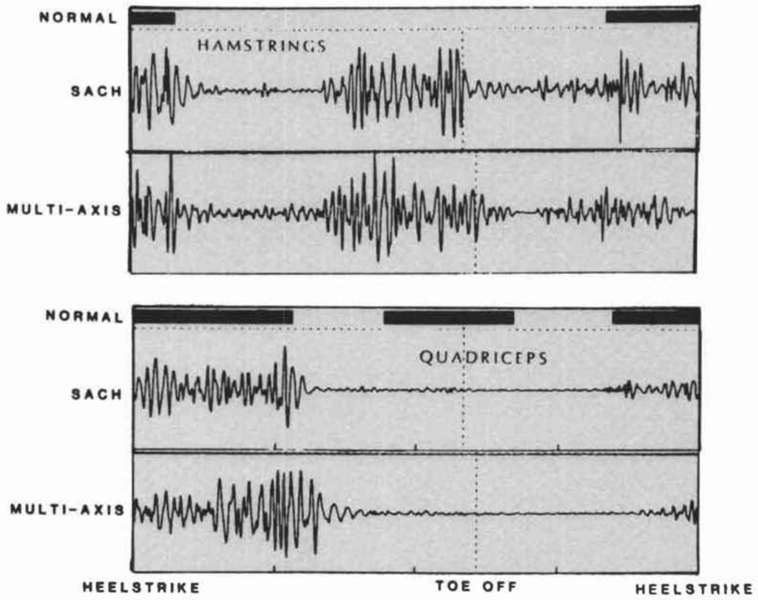


Figure 6. EMG activity of the hamstrings and quadriceps muscles during gait in a 27-year-old patient with a SACH foot and with a multi-axis foot.

muscle groups were sampled and show the same firing patterns regardless of the type of foot that is worn. What is interesting is that the hamstrings are firing just before toe-off when they are usually silent and the quadriceps are inactive at this time when normally they fire to restrain knee flexion and prevent excessive heel rise. As might be expected, this patient walks with exaggerated knee flexion at swing phase.

SUMMARY

Gait analysis is useful in evaluating an amputee's prosthesis by providing objective measurements and a permanent record of the patient's status. Kinematic, kinetic, and EMG data assist the clinician and prosthetist in identifying specific problems encountered by the amputee and in identifying the causes. Gait analysis also allows comparison of different prosthetic designs or different alignments of the same prothesis. Most importantly, however, the record provided allows examiners to objectively discuss the problems and their potential solutions.

FUTURE APPLICATIONS

The field of prosthetics will begin to change rapidly with the application of kinesiology. Soon, optimal standards of gait will be established for each prosthetic level. With the widespread availability of low-cost motion analysis, kinematic analysis will be routinely incorporated into dynamic alignment of each new prosthesis, helping to insure appropriate alignment and fit. Finally, prosthetic research, using both kinematics and kinetics, will continue as we seek to identify and rectify the problems created by loss of the body's normal limb. The ultimate outcome of this research will be the development of components that will be stronger, lighter in weight, and much more functional than those used now.

REFERENCES

[1] *Fundamental Studies of Human Locomotion and Other Information Relating to Design of Artificial Limbs.* Subcontractor's Report to the Committee on Artificial Limbs. National Research Council. Prosthetic Devices Research Project, College of Engineering, University of California, Berkeley. Serial No. CAL 5. 2 vols. The Project, Berkeley, 1947.

[2] Radcliffe, C.W., "Functional considerations in the fitting of above-knee prostheses," *Selected Articles From Artificial Limbs,* Huntington, NY, Kreiger Publishing Co, Inc, 1970, p.p. 5–30.

[3] Winter, D.A., *Biomechanics of Human Movement,* New York, John Wiley & Sons, 1979, p.p. 9–46.

[4] Eberhart, H.D.; Elftman, H.; and Inman, V.T., "The locomotor mechanism of the amputee," *Klopsteg PE, Wilson PD, et al (eds): Human Limbs and Their Substitutes,* New York, Hafner Publishing Co, 1968, p.p. 472–480.

[5] Zuniga, E.N.; Leavitt, L.A.; Calvert, J.C.; Canzoneri, J.; and Peterson, C.R., "Gait patterns in above-knee amputees," *Arch Phys Med Rehabilitation,* 53:373–382, 1972.

[6] James, U. and Oberg, K., "Prosthetic gait pattern in unilateral above-knee amputees," *Scand J Rehabil Med,* 5:35–50, 1973.

[7] Murrary, M.P.; Sepic, S.B.; Gardner, G.M.; and Mollinger, L.A., "Gait patterns of above-knee amputees using constant-friction knee components," *Bull Prosthet Res*, 17(2):35–45, 1980.

[8] Godfrey, C.M.; Jousse, A.T.; Brett, R.; and Butler, J.F., "A comparison of some gait characteristics with six knee joints," *Orthotics and Prosthetics*, 29(3):33–38, 1975.

[9] Murray, M.P.; Mollinger, L.A.; Sepic, S.B.; Gardner, G.M.; and Linder, M.T., "Gait patterns in above-knee amputee patients: Hydraulic swing control vs constant-friction knee components," *Arch Phys Med Rehabil*, 64:339–345, 1983.

[10] Hoy, M.G.; Whiting, W.C.; and Zernicke, R.F., "Stride kinematics and knee joint kinetics of child amputee gait," *Arch Phys Med Rehabilitation*, 63:74–82, 1982.

[11] Hannah, R.E. and Morrison, J.B., "Prostheses alignment: Effect on gait of persons with below-knee amputations," *Arch Phys Med Rehabil* 65:159–162, 1984.

[12] Grevsten, S. and Stalberg, E., "Electromyographic study of muscular activity in the amputation stump while walking with PTB- and PTB-suction prosthesis," *Ups J Med Sci*, 80:103–112, 1975.

[13] Thiele, B.; James, U.; and Stalberg, E., "Neurophysiological studies on muscle function in the stump of above-knee amputees," *Scand J Rehabil Med*, 5:67–70, 1973.

[14] Krebs, D.E.; Edelstein, J.E.; and Fishman, S., "Reliability of observational kinematic gait analysis," Accepted for publication in *J Phys Ther*, 1985.

AUTHORS

James R. Gage, M.D. and Ramona Hicks, R.P.T., M.A. are with the Kinesiology Laboratory at Newington Children's Hospital in Newington, Connecticut 06111.

Evaluation of a Prosthetic Shank with Variable Inertial Properties

by Scott Tashman, M. Eng.
Ramona Hicks, R.P.T., M.A.
David J. Jendrzejczyk, C.P.

Above-knee amputees walk slower than the normal population. This has been documented in adults[1,2,3] and children.[4] It has been suggested that the prolonged swing phase of the prosthesis forces a slower cadence and, therefore, a slower walking speed.[5] Since children rely on a fast cadence to obtain an adequate walking speed,[6] a prolonged swing phase can be a major obstacle to comfortable, efficient normal-speed walking.

To date, most efforts to reduce prosthetic swing phase time have been directed towards the prosthetic knee joint.[7,8] Various mechanisms have been designed to accelerate the extension of the prosthetic knee. Mechanical, hydraulic, and pneumatic systems have been developed in an effort to provide a more favorable gait.[9] Hydraulic knee units have been shown to provide a more normal cadence and walking speed for adults than simple constant friction knee units.[10]

Most of the prosthetic knee unit research has been directed towards the adult amputee population. Pediatric hydraulic knee units have been considered impractical because of size and weight limitations. Pediatric above-knee amputees are generally fitted with constant friction knee units because they are simple, light in weight, low in cost, easy to install and adjust, and require little maintenance.[11]

It has often been presumed that adjustments in the knee joint friction could be used to provide an optimum cadence for the amputee with a constant friction knee joint. A study was performed at the Newington Children's Hospital Kinesiology Laboratory to test this assumption.[12] When subjects were asked to walk at a comfortable speed, no significant changes were observed in cadence or actual prosthetic shank swing time as the knee joint friction was varied over a wide range. In all cases, the swing period of the prosthetic shank was close to the natural swing period of the shank measured off the patient. This indicates that the physical properties of the prosthetic shank play a significant role in determining the natural cadence of the above-

[7] Murrary, M.P.; Sepic, S.B.; Gardner, G.M.; and Mollinger, L.A., "Gait patterns of above-knee amputees using constant-friction knee components," *Bull Prosthet Res,* 17(2):35–45, 1980.
[8] Godfrey, C.M.; Jousse, A.T.; Brett, R.; and Butler, J.F., "A comparison of some gait characteristics with six knee joints," *Orthotics and Prosthetics,* 29(3):33–38, 1975.
[9] Murray, M.P.; Mollinger, L.A.; Sepic, S.B.; Gardner, G.M.; and Linder, M.T., "Gait patterns in above-knee amputee patients: Hydraulic swing control vs constant-friction knee components," *Arch Phys Med Rehabil,* 64:339–345, 1983.
[10] Hoy, M.G.; Whiting, W.C.; and Zernicke, R.F., "Stride kinematics and knee joint kinetics of child amputee gait," *Arch Phys Med Rehabilitation,* 63:74–82, 1982.
[11] Hannah, R.E. and Morrison, J.B., "Prostheses alignment: Effect on gait of persons with below-knee amputations," *Arch Phys Med Rehabil* 65:159–162, 1984.
[12] Grevsten, S. and Stalberg, E., "Electromyographic study of muscular activity in the amputation stump while walking with PTB- and PTB-suction prosthesis," *Ups J Med Sci,* 80:103–112, 1975.
[13] Thiele, B.; James, U.; and Stalberg, E., "Neurophysiological studies on muscle function in the stump of above-knee amputees," *Scand J Rehabil Med,* 5:67–70, 1973.
[14] Krebs, D.E.; Edelstein, J.E.; and Fishman, S., "Reliability of observational kinematic gait analysis," Accepted for publication in *J Phys Ther,* 1985.

AUTHORS

James R. Gage, M.D. and Ramona Hicks, R.P.T., M.A. are with the Kinesiology Laboratory at Newington Children's Hospital in Newington, Connecticut 06111.

Evaluation of a Prosthetic Shank with Variable Inertial Properties

by Scott Tashman, M. Eng.
Ramona Hicks, R.P.T., M.A.
David J. Jendrzejczyk, C.P.

Above-knee amputees walk slower than the normal population. This has been documented in adults[1,2,3] and children.[4] It has been suggested that the prolonged swing phase of the prosthesis forces a slower cadence and, therefore, a slower walking speed.[5] Since children rely on a fast cadence to obtain an adequate walking speed,[6] a prolonged swing phase can be a major obstacle to comfortable, efficient normal-speed walking.

To date, most efforts to reduce prosthetic swing phase time have been directed towards the prosthetic knee joint.[7,8] Various mechanisms have been designed to accelerate the extension of the prosthetic knee. Mechanical, hydraulic, and pneumatic systems have been developed in an effort to provide a more favorable gait.[9] Hydraulic knee units have been shown to provide a more normal cadence and walking speed for adults than simple constant friction knee units.[10]

Most of the prosthetic knee unit research has been directed towards the adult amputee population. Pediatric hydraulic knee units have been considered impractical because of size and weight limitations. Pediatric above-knee amputees are generally fitted with constant friction knee units because they are simple, light in weight, low in cost, easy to install and adjust, and require little maintenance.[11]

It has often been presumed that adjustments in the knee joint friction could be used to provide an optimum cadence for the amputee with a constant friction knee joint. A study was performed at the Newington Children's Hospital Kinesiology Laboratory to test this assumption.[12] When subjects were asked to walk at a comfortable speed, no significant changes were observed in cadence or actual prosthetic shank swing time as the knee joint friction was varied over a wide range. In all cases, the swing period of the prosthetic shank was close to the natural swing period of the shank measured off the patient. This indicates that the physical properties of the prosthetic shank play a significant role in determining the natural cadence of the above-knee amputee with a constant-friction knee joint. To force the shank to move at a frequency different from its natural frequency requires significant input of energy in the form of applied torque at the knee joint (from hip or pelvic muscle force). The test subjects, when asked to walk at a comfortable speed, did not supply the extra energy needed for a faster cadence; they instead aligned their cadence with the natural frequency of the shank.

PURPOSE

The above results led to the current project: the design and testing of a prosthetic shank with variable physical properties. The purpose of this study was to test the following hypotheses:

1. If the physical properties of the shank section of an above-knee prosthesis with a constant friction knee unit are changed in such a way as to alter the natural swing period, the swing period of the shank during gait will also be altered.

2. Reducing the swing period of the shank will increase natural cadence and walking speed.

METHODS

Design

The principal design goal for the shank was to reduce the natural swing period as much as possible. If the shank/foot is considered as a physical pendulum, it has a period T equal to: $\mathbf{T = 2\pi \sqrt{I/Mgd}}$, I is proportional to Md^2.

Where:

- T = natural swing period of shank as a pendulum
- I = rotational inertia of shank/foot above knee pivot
- g = acceleration due to gravity
- d = distance from knee pivot to center of mass
- M = mass of shank/foot

These equations indicate that changes in mass alone will not reduce the swing period of the shank; the center of mass must be shifted proximally (towards the knee joint) to significantly reduce the period.

With reducing distal weight as the primary goal, an experimental shank was constructed for the test subject, a 13 year old male knee disarticulation patient with a "good" amputee gait pattern. Since the limb was to be used for laboratory testing purposes only, some strength was sacrificed in order to obtain the maximum possible reduction in distal weight while still using readily available materials. The shank was thin and hollow, with layers of polyester resin and one layer of carbon filter cloth laminated over a plaster mold. Excess material was ground away wherever possible. In addition, the prosthesis was set in correct alignment using a heel build-up on an ultra-light SACH foot to eliminate shoes and further reduce distal weight. To enable changes in the natural swing period, a lead mass which attached to a metal rod could be placed proximally or distally inside the shank. The additional mass was chosen so that the experimental shank/foot would weigh the same as the patient's standard prosthesis.

The completed prosthesis is shown in Figure 1. With the moveable mass placed distally, the shank had a center of mass positioned similarly to the patient's original shank. Shifting the mass proximally caused the center of mass to move proximally by 13 centimeters. To deter-

Figure 1. Completed experimental prosthesis; shown during testing in the Kinesiology laboratory.

Comparison of Physical Properties: Standard vs. Experimental Prosthetic Shank			
		Experimental Shank	
	Standard Shank	Proximal mass	Distal mass
Mass of Shank (Kg)	1.90	1.03	1.03
Mass of Shank & Weight	—	1.86	1.86
Distance from Knee Joint to Center of Mass (cm)	31.0	18.7	31.7
Swing Period (sec)	1.30	1.12	1.32

Table 1.

mine the effect of changing the mass position, the pendulum swing period of the shank was measured by timing the swing of the shank, which was suspended by a metal rod through the knee joint axis. The light weight shank, with the mass placed distally, exhibited inertial properties very close to those of the patient's original shank. Shifting the mass to the proximal position reduced the pendulum swing period by 0.20 seconds or 15 percent (Table 1).

Evaluation

The Newington Children's Hospital Kinesiology Laboratory measured the effect on the gait of the changes made in the position of the center of mass of the experimental prosthesis. An automated video system was used to acquire three-dimensional kinematic data from 26 retroreflective markers placed at designated positions on the body.[13,14] The kinematic data were used to determine the motions of all major body segments and calculate dynamic lower extremity joint angles in three planes. Linear movement and temporal measurements, such as stride length, single stance time, swing phase time, cadence, and walking speed were also determined. Swing time was determined by measuring the time from toe-off to heel strike. The shank pendulum time was determined by measuring the time required for the prosthesis to go from full extension into flexion and back to full extension; this is equivalent to one half of the period of the shank measured as a free-swinging pendulum.

Kinematic data were acquired for two walks with the subject walking at:

a. normal speed, weight proximal
b. fast speed, weight proximal
c. normal speed, weight distal
d. fast speed, weight distal

For the normal speed walks, the subject was asked to walk at a speed that was comfortable; no further prompting was given. For the faster speed walks, the subject was instructed to walk as fast as was comfortable; again, no further instructions were given. For each mass position, the knee joint friction was set to "clinically optimal" by matching the prosthetic side heel rise to the normal side heel rise at normal speed, and the patient was allowed to walk around for a while until he seemed reasonably comfortable with the altered characteristics of the limb.

RESULTS

Stride Parameters

Stride parameters measured during the four different conditions are shown in Table 2. This data represents the first walk acquired for each condition; the variation between the first and second trials for all conditions was less than five percent. Cadence, stride length, and walking speed were all essentially the same at the "normal" walking speed with the mass placed proximally or distally. At the "fast" walking speed, the subject walked seven percent faster with the mass placed distally than with the mass placed proximally, due to both a faster cadence and a longer stride length.

Shank Swing Dynamics

At the normal walking speed, the shank pendulum time was reduced by eight percent with the weight placed proximally, resulting in an eight percent reduction in the swing phase time for the prosthetic limb (Table 3). Since the swing phase time for the normal side stayed the

Results: Stride Parameters				
	NORMAL SPEED		FAST SPEED	
	Mass Proximal	Mass Distal	Mass Proximal	Mass Distal
Cadence (steps/min)	91.6	90.0	105	109
Walking Speed (cm/sec)	94.9	94.5	126	135
Gait Cycle Time (sec)	1.33	1.33	1.15	1.10
Stride Length (cm)	126.0	125.8	144.0	149.4

Table 2.

Swing Phase Timing: Normal Speed			
	Mass Distal	Mass Proximal	% Change
Swing Time—Prosthetic Side	0.62	0.57	−8%
Swing Time—Normal Side	0.51	0.52	+2%
Swing Time Asymmetry (%)	19.5	9.1	−10%
Shank Pendulum Time	0.60	0.55	−8%

Table 3.

Swing Phase Timing: Fast Speed			
	Mass Distal	Mass Proximal	% Change
Swing Time—Prosthetic Side	0.61	0.56	−8%
Swing Time—Norm. Side	0.44	0.46	+5%
Swing Time Asymmetry (%)	32.4	19.6	−13%
Shank Pendulum Time	0.61	0.51	−16%

Table 4.

same, the swing asymmetry (prosthetic side vs. normal side) was reduced from 19.5 percent to 9.1 percent. A similar reduction in swing asymmetry was seen during the fast walk (from 32.4 percent to 19.6 percent). During fast walking with the proximal weight placement, the swing phase time was increased by five percent for the normal limb and reduced by eight percent for the prosthetic limb. The reduction in pendulum swing time was much greater (16 percent).

The dynamic knee joint motion is shown for both weight positions (Figures 2 and 3). The peak knee flexion was reduced from 64 degrees to 54 degrees at the normal walking speed and from 84 degrees to 62 degrees at the fast walking speed with the weight placed proximally. The plots also indicate delayed initiation of knee flexion and faster motion of the limb with the proximal weight placement.

DISCUSSION

As expected, proximal weight placement in the shank produced a shorter shank swing time during gait. This subsequently resulted in a shorter swing phase (toe-off to heel-strike) for the prosthetic limb. At normal speed, the decrease in swing phase was equal in time to the decrease in shank swing time (eight percent). At a faster walking speed, the same eight percent decrease in swing phase was observed, but the shank swing period was reduced by a much

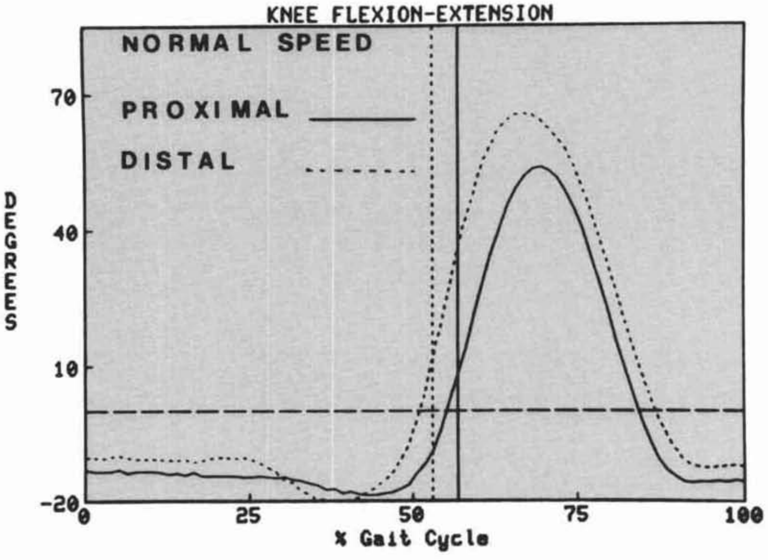


Figure 2. Knee flexion-extension angle vs. percent of gait cycle: normal walking speed, proximal and distal weight placement.

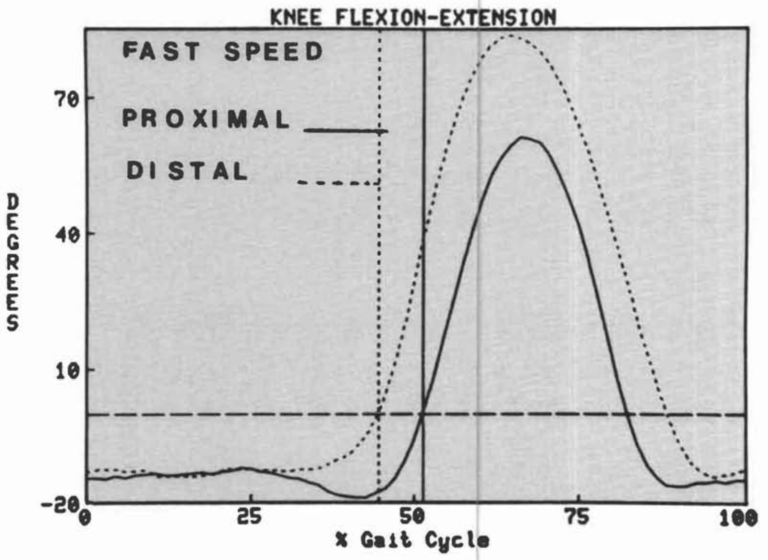


Figure 3. Knee flexion-extension angle vs. percent of gait cycle: fast walking speed, proximal and distal weight placement.

greater amount. Figure 3 illustrates the effect of this discrepancy: the limb reaches full extension well before heel strike. One explanation for this is that the subject did not have sufficient time to fully adjust to the new limb; further use should enable the subject to reduce swing phase as much as the shank swing period was reduced.

A less expected outcome was the similarity in walking speed and cadence between the two different weight placements. The reduced swing phase did not result in a reduced gait cycle time; the subject instead lengthened his stance phase to balance the decrease in swing phase. This resulted in a smoother, more symmetric gait.

CONCLUSIONS

Limited conclusions can be made based on this single-subject study. However, it appears that decreasing the natural swing period of the shank by shifting the center of mass proximally results in a faster swing phase during gait. In one subject this led to an increase in stance phase for the prosthetic side towards normal values, and considerably reduced left-right asymmetry for this subject. Improved symmetry should lead to a more energy efficient, natural appearing gait. No increase in cadence or walking speed was observed. It is possible that longer wear of the limb might have permitted the subject to naturally increase his cadence;

this could not be evaluated with the present limb design.

The outcome of this study indicates that weight distribution in the prosthetic shank/foot has a significant impact on gait. This suggests that future prostheses should be designed to minimize distal shank/foot weight.

REFERENCES

[1] James, U., and Oberg, K., "Prosthetic gait pattern in unilateral above-knee amputees," *Scand J Rehab Med,* 5:35–50, 1973.

[2] Godfry, C. M.; Jousee, A. T.; Brett, R.; and Butler, J. F., "A comparison of some gait characteristics with six knee joints," *Orthotics and Prosthetics,* 29(3):33–38, 1975.

[3] Murry, M. P., "Gait patterns of above-knee amputees using constant-friction knee components," *Bull Prosthet Res,* 17(2):35–45, 1980.

[4] Hoy, M. G.; Witing, W. C.; and Zernicke, R.F., "Stride kinematics and knee joint kinetics of child amputee gait," *Arch Phys Med Rehabil,* 63:74–81, Feb 1982.

[5] Drillis, R., "Objective recording and biomechanics of pathological gait," *Ann NY Acad Sci,* 74:86–109, Sept 1958.

[6] Sutherland, D. H.; Olshen, R.; Cooper, L.; and Woo, S., "The development of mature gait," *J Bone Joint Surgery,* 62A:336–353, April, 1980.

[7] Zarrugh, M. Y., and Radcliffe, C. W., "Simulation of swing phase dynamics in above-knee prosthesis," *J Biomech,* 9:283–292, 1976.

[8] Wallach, J., and Saibel, E., "Control mechanism performance criteria for an above-knee leg prosthesis," *J Biomech,* 3:87–97, 1970.

[9] NYU Medical Center, *Lower-Limb Prosthetics,* pp. 145–163. Prosthetics and Orthotics, New York University Post-Graduate Medical School, 1980 revision.

[10] Murray, M. P.; Mollinger, L. A.; Sepic, S. B.; Gardner, G. M., and Linder, M. T., "Gait patterns in above-knee amputee patients: Hydraulic swing control vs. constant-friction knee components," *Arch Phys Med Rehabil,* 64:339–345, 1983.

[11] New York University Medical Center, "The ISNY PTB Socket," *Lower-Limb Prosthetics,* 1980 revision, pp. 107–108.

[12] Hicks, R.; Tashman, S.; Cary, J. M.; Altman, R. F.; and Gage, J. R., "Swing Phase Control with knee friction in juvenile amputees," In press, *J Orthop Res.*

[13] Gage, J. R., "Gait Analysis for decision-making in cerebral palsy," *Bull Hosp Joint Des,* 43:147–163, 1983.

[14] Gage, J. R.; Fabian, D.; Hicks, R.; and Tashman, S., "Pre- and postoperative gait analysis in patients with spastic diplegia—a preliminary report," *J Ped Orthop,* 4:715–724, 1984.

AUTHORS

Scott Tashman, M.Eng., Ramona Hicks, R.P.T., M.A., and David L. Jendrzejczyk, C.P. are with the Kinesiology Department and Department of Orthotics and Prosthetics at Newington Children's Hospital, Newington, Connecticut, 06111.

Kinematic and Kinetic Comparison of the Conventional and ISNY Above-Knee Socket

by David E. Krebs, M.A., P.T.
Scott Tashman, M.S.

Prosthetic sockets must be comfortable, but they must also be functional. Despite advances in other prosthetic components, the materials used to construct the portion of the prosthesis that most directly contributes to amputee comfort, the socket, have remained essentially unchanged since the introduction of thermosetting resin sockets in the 1950's.[1–4]

A prosthetic socket must perform at least two functions: it must contain the stump tissues, and during stance phase, it must provide a means of transferring the amputee's weight from the pelvis and residual limb to the floor. To contain the stump tissues, the socket shape encourages optimal distribution of forces and pressure, during both swing and stance phases. Body weight is transmitted primarily from the ischial tuberosity to the proximal brim of the socket, through the vertical socket walls, and finally through the knee, shank, and foot to the floor.[5]

Conventional sockets surround the entire residual limb with rigid thermosetting resins, thus requiring this single container to perform both socket functions. The rigid proximolateral socket wall has been reputed to provide stabilization for the residual limb,[3] although no emthis could not be evaluated with the present limb design.

The outcome of this study indicates that weight distribution in the prosthetic shank/foot has a significant impact on gait. This suggests that future prostheses should be designed to minimize distal shank/foot weight.

REFERENCES

[1] James, U., and Oberg, K., "Prosthetic gait pattern in unilateral above-knee amputees," *Scand J Rehab Med*, 5:35–50, 1973.

[2] Godfry, C. M.; Jousee, A. T.; Brett, R.; and Butler, J. F., "A comparison of some gait characteristics with six knee joints," *Orthotics and Prosthetics*, 29(3):33–38, 1975.

[3] Murry, M. P., "Gait patterns of above-knee amputees using constant-friction knee components," *Bull Prosthet Res*, 17(2):35–45, 1980.

[4] Hoy, M. G.; Witing, W. C.; and Zernicke, R.F., "Stride kinematics and knee joint kinetics of child amputee gait," *Arch Phys Med Rehabil*, 63:74–81, Feb 1982.

[5] Drillis, R., "Objective recording and biomechanics of pathological gait," *Ann NY Acad Sci*, 74:86–109, Sept 1958.

[6] Sutherland, D. H.; Olshen, R.; Cooper, L.; and Woo, S., "The development of mature gait," *J Bone Joint Surgery*, 62A:336–353, April, 1980.

[7] Zarrugh, M. Y., and Radcliffe, C. W., "Simulation of swing phase dynamics in above-knee prosthesis," *J Biomech*, 9:283–292, 1976.

[8] Wallach, J., and Saibel, E., "Control mechanism performance criteria for an above-knee leg prosthesis," *J Biomech*, 3:87–97, 1970.

[9] NYU Medical Center, *Lower-Limb Prosthetics*, pp. 145–163. Prosthetics and Orthotics, New York University Post-Graduate Medical School, 1980 revision.

[10] Murray, M. P.; Mollinger, L. A.; Sepic, S. B.; Gardner, G. M., and Linder, M. T., "Gait patterns in above-knee amputee patients: Hydraulic swing control vs. constant-friction knee components," *Arch Phys Med Rehabil*, 64:339–345, 1983.

[11] New York University Medical Center, "The ISNY PTB Socket," *Lower-Limb Prosthetics*, 1980 revision, pp. 107–108.

[12] Hicks, R.; Tashman, S.; Cary, J. M.; Altman, R. F.; and Gage, J. R., "Swing Phase Control with knee friction in juvenile amputees," In press, *J Orthop Res*.

[13] Gage, J. R., "Gait Analysis for decision-making in cerebral palsy," *Bull Hosp Joint Des*, 43:147–163, 1983.

[14] Gage, J. R.; Fabian, D.; Hicks, R.; and Tashman, S., "Pre- and postoperative gait analysis in patients with spastic diplegia—a preliminary report," *J Ped Orthop*, 4:715–724, 1984.

AUTHORS

Scott Tashman, M.Eng., Ramona Hicks, R.P.T., M.A., and David L. Jendrzejczyk, C.P. are with the Kinesiology Department and Department of Orthotics and Prosthetics at Newington Children's Hospital, Newington, Connecticut, 06111.

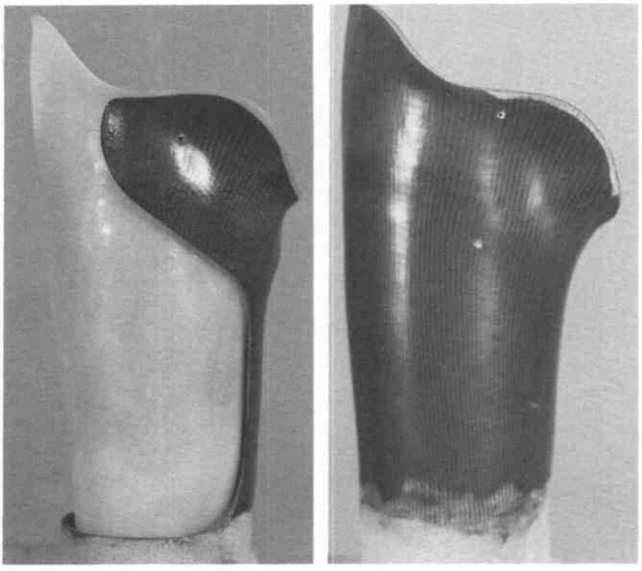
Kinematic and Kinetic Comparison of the Conventional and ISNY Above-Knee Socket

by David E. Krebs, M.A., P.T.
Scott Tashman, M.S.

Prosthetic sockets must be comfortable, but they must also be functional. Despite advances in other prosthetic components, the materials used to construct the portion of the prosthesis that most directly contributes to amputee comfort, the socket, have remained essentially unchanged since the introduction of thermosetting resin sockets in the 1950's.[1–4]

A prosthetic socket must perform at least two functions: it must contain the stump tissues, and during stance phase, it must provide a means of transferring the amputee's weight from the pelvis and residual limb to the floor. To contain the stump tissues, the socket shape encourages optimal distribution of forces and pressure, during both swing and stance phases. Body weight is transmitted primarily from the ischial tuberosity to the proximal brim of the socket, through the vertical socket walls, and finally through the knee, shank, and foot to the floor.[5]

Conventional sockets surround the entire residual limb with rigid thermosetting resins, thus requiring this single container to perform both socket functions. The rigid proximolateral socket wall has been reputed to provide stabilization for the residual limb,[3] although no em-



Figures 1A and 1B. The ISNY Socket System.

pirical evidence has been reported to substantiate this claim.

The ISNY socket system (Figure 1) has been introduced recently by facilities in Iceland, Sweden, and New York University. The ISNY system consists of two parts, each with its own function. The ISNY socket is flexible, but not elastic, and biomechanically only serves to contain the stump tissues. This containment, of course, thus provides an inelastic reaction point for the thigh and the muscles that act upon it. The ISNY frame consists of a strong thin strut located horizontally around the quadrilateral brim, and vertically along the medial portion of the prosthetic thigh section,[6,7] that performs the weight transfer function. Because the strut is extraordinarily strong, more than 75 percent of the residual limb is covered only by the soft flexible thermoplastic.

Over 1,000 amputees world-wide have been fitted with the ISNY socket. These amputees consistently report significant improvement in prosthetic comfort. No amputees have complained of instability due to the flexible lateral wall of the ISNY, nor have clinicians reported functional disadvantages stemming from the flexible ISNY socket. On the contrary, even amputees who have been converted from pelvic band suspension on a conventional hard socket to solely suction suspension with an ISNY socket, show no visually apparent gait instabilities.

The purpose of this single-subject study is to objectively investigate the kinematic and kinetic differences during walking in an ISNY and a conventional socket. In particular, we sought to answer the question: "Does the absence of a rigid lateral wall, such as in the ISNY socket, contribute to gait instability?"

METHOD

This investigation provided a single, unilateral above-knee amputee with two thigh sections, one with an ISNY and the other with a conventional socket, and then analyzed selected kinematic and kinetic gait variables during walking at a slow speed, a preferred rate, and a fast speed, to determine if biomechanical differences resulted during walking in the two different sockets.

Subject

A thirty-six year old male participated in this single-subject research design. He provided written, informed consent prior to participation. The subject, a left above-knee amputee since a motor cycle accident 12 years prior to the study, is an otherwise healthy, active non-athlete who is employed in a full-time job that demands at least eight ambulatory hours daily. The subject is 1.75 m (5 feet, 9 inches) tall and weighs 104 kg (229 lb.); his residual limb length is 80 percent of the right (normal) femur. The subject wears an above-knee exoskeletal prosthesis with SACH foot, Mauch S-N-S hydraulic knee, and quadrilateral total-contact suction socket.

Prosthesis

Two types of above-knee prosthetic sockets, identical in size, shape, and alignment, were made for the components described above. The sockets were fitted to interchangeable thigh sections, to permit easy exchange of the conventional and ISNY sockets upon a single shank section.

One socket was fabricated of conventional rigid thermosetting resins, while the other was of ISNY construction. The subject had worn a conventional socket from the time of amputation until initiation of this investigation. At that time, the conventional socket was duplicated using Plaster-of-Paris to fabricate a positive model, which in turn was utilized to construct the ISNY socket. Both types of prosthetic sockets therefore provided identical total stump contact.

Because alignment symmetry is such an important determinant of prosthetic gait, the thigh-to-shank alignments and weight-transfer lines on the prosthetic thigh sections were precisely duplicated utilizing laboratory bench-alignment techniques.[3] In addition, alignment was visually inspected while the patient stood and walked in both sockets; adjustments were made until the alignments were symmetrical. Following prosthetic fabrication, however, no further alignment alterations were possible, thus helping to insure the integrity of the data collection procedures.

Testing Procedure

Both ISNY and conventional sockets were tested during the same day, to minimize variability in gait laboratory data collection techniques, and differences within the subject that longer between-analysis trials might have entailed. The conventional socket had been worn for 12 months; the ISNY socket had been worn for the 11 months immediately prior to gait analysis. The subject wore each type of prosthesis at least two hours immediately prior to testing, to permit re-accommodation to the prosthesis to be tested.

Both sockets were tested while the subject walked at one of three self-selected speeds. The testing order for socket type and walking speed was randomly assigned; walking speed order was replicated over the sockets. One of the following commands was given to the amputee prior to each experimental run:

1. "Walk as slowly as you comfortably can"
2. "Walk at your normal pace"
3. "Walk as fast as you comfortably can"

The exact speed at which the subject chose to walk under the three different commands was self-selected: no attempt was made to interpret the commands, or to interfere with the amputee's self-selected speed during data collection on the first socket. Prior to walking trials with the second socket to be tested, the subject was instructed to attempt to duplicate the speed at which he walked with the first type of socket.

Each socket/speed combination was tested twice to assess the kinematic stability (reliability) of the test situation.

Data Collection

Kinesiological data was collected at the Gait Laboratory of Newington Children's Hospital, Newington, Connecticut, one of the most technologically sophisticated gait data collection centers in the United States. The gait lab and data collection are described in detail elsewhere.[7-9] In this study, the following data was simultaneously monitored and subsequently displayed:

Kinematics: Joint positions of the pelvis and both lower-limbs in three dimensions.

Kinetics: Floor reaction forces and moments of the prosthetic side in three dimensions.

Data collection was conducted identically for all experimental conditions. Furthermore, the amputee was "blind" to the experimental hypotheses; the data collection techniques were designed to obviate experimenter bias.

Data collection began after the amputee had taken at least three steps, to insure that steady-state walking velocity had been achieved.

Kinematic Data

The following joint motions on the amputated and the sound sides were measured:

Hip: Extension/flexion, abduction/adduction, and internal/external rotation;

Pelvis: Sagittal tilt, frontal plane obliquity and transverse rotation;

Knee: Extension/flexion; valgus/varus; and internal/external rotation;

Foot: Plantar/dorsiflexion and transverse rotation.

In addition, the following events were obtained: cycle time; toe off, heel off, and prosthetic stance duration in percent cycle time; stride and step length in cm.; cadence in steps per minute; and overall velocity (walking speed) in meters per second.

Passive, optically reflective limb position markers were placed on the prosthetic side over the posterior sacrum, anterior superior iliac spine, greater trochanter of the hip, lateral mid-thigh, lateral knee, lateral mid-shank, ankle lateral heel, and the lateral foot at the level of the fifth metatarsal head. Similar markers were placed on the subject's sound side. The markers remained in place between runs, with the exception of the markers placed on the experimental thigh sections, which were re-attached in identical positions for both types of sockets.

Prior to data collection, it was specified that output differences of five degrees or more, in any plane during stance phase, were to be accepted as significant departures from kinematic symmetry of the two socket systems. Particular attention was to be paid to residual limb hip and pelvic motions, because the prosthetic knee and foot motions cannot be affected by socket differences.

Kinetic Data

Ground reaction forces and moments generated by the amputee while walking in the two sockets were measured by two six-channel force plates. The force plates have a 12 bit resolution, and output X (antero-posterior), Y (medio-lateral), and Z (vertical) force and moment signals proportional to the ground reactions during the gait cycle. The force plates are sampled at 2 KHz, and low-pass filtered with a 300 Hz. fourth-order Bessel analog filter to dampen high frequency noise.

The force plates are mounted flush with the floor and are thus totally unobtrusive; the subject was neither made aware of their location, nor instructed to alter his gait to strike the force plates.

Prior to data collection, it was specified that differences of five percent or more were to be accepted as significant departures from kinetic symmetry. Particular attention was to be paid to residual limb medio-lateral force and shear kinetics, because alterations in the medio-lateral support of the socket would be most likely to be revealed by changes in these force-plate variables.

RESULTS

Gait analysis was performed first on the ISNY socket, followed by analysis of the conventional socket. The walking speed randomization resulted in the following trial sequence: preferred pace, then slow pace, then fast pace walking. Temporal data for each run is presented in Table 1. Kinematic and kinetic data resulting from the two sockets and three walking speeds are given in Figures 2 through 5.

No differences of greater than five degrees are found in any of the stance phase kinematics, with the single exception of late stance phase hip rotation at slow walking speed. This isolated finding is probably insignificant.

Test-retest reliability analyses revealed no more than two to three degrees difference, for any given speed, between repeated runs in the same socket.

DISCUSSION

No substantial kinematic or kinetic differences are to be found in the analysis of this subject's biomechanical data. Therefore, socket design did not differentially affect the subject's gait timing, joint motions, or forces. In particular, it would appear that the absence of a rigid lateral wall does not adversely affect dynamic stump or pelvic stability of this above-knee amputee.

It is important to note the limitations of single-subject research design. Although it is quite reasonable to infer that this subject's gait was unchanged (or more precisely, differed by less than expected error amounts) in the two types of prosthetic sockets, it would be incorrect to generalize the results of this study beyond this single subject. That is, it is possible that kinematic differences might be found in other subjects, despite their absence in this subject.

Temporal Comparisons No consistent pattern was found with regard to between-socket comparisons of walking speed, nor stance duration, nor any other important temporal variable (Table 1). For example, walking velocity in the conventional rigid socket was about eight and 13 percent faster than in the ISNY at "fast" and "slow" walking speeds, respectively, but velocity is about one percent greater at the self-selected "preferred" walking speed. Prosthetic stance phase duration was consistently, if only slightly, longer with the ISNY socket, than with the conventional socket, approaching near-normal values at the "fast" velocity. This finding agrees with the data from Murray and colleagues.[10] It is possible, therefore, that the ISNY socket permitted greater stance-phase comfort, which in turn permitted more normal

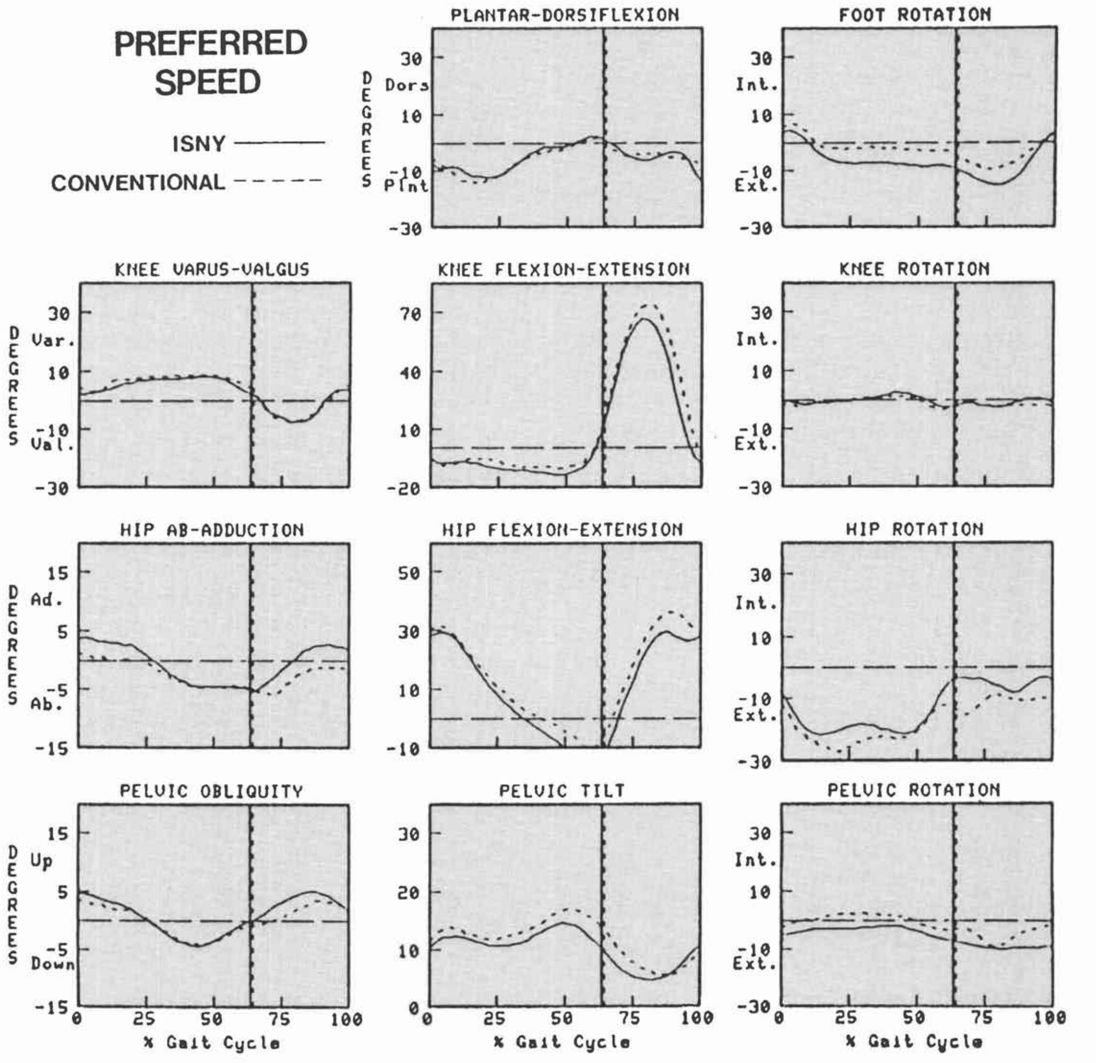


Figure 2.

weight acceptance during stance phase than with the conventional socket. That is, the ISNY socket appeared to permit more symmetrical prosthetic to sound side stance durations. Leavitt, et. al.[11] support the contention that more comfortable sockets permit longer, and more normal, stance phase durations.

Kinematic Comparisons No kinematic between-socket differences of greater than five degrees were found, except in "slow" walking hip rotation. The latter finding probably reflects normal, minor variability in walking kinematics.[12] Kinematic variability, at least in non-amputees, is greater at non-preferred walking cadences.[13] Of particular interest to this study is the fact that hip abduction in the ISNY socket was relatively invariant over the three walking velocities, and indeed, more closely approximated previously reported kinematics from normal subjects[14] than did the subject's hip kinematics while walking with the conventional socket.

Kinetic Comparisons Normal subjects walking at self-selected comfortable speeds exert vertical floor reactions that increase from 0 to approximately 120 percent of body weight from heel strike to foot flat. As the knee flexes in midstance and then extends in preparation for push-off, vertical forces decline to 60 to 80 percent, then increase to a second peak of about 120 percent of body weight, falling again to 0 after toe-off.[15-19]

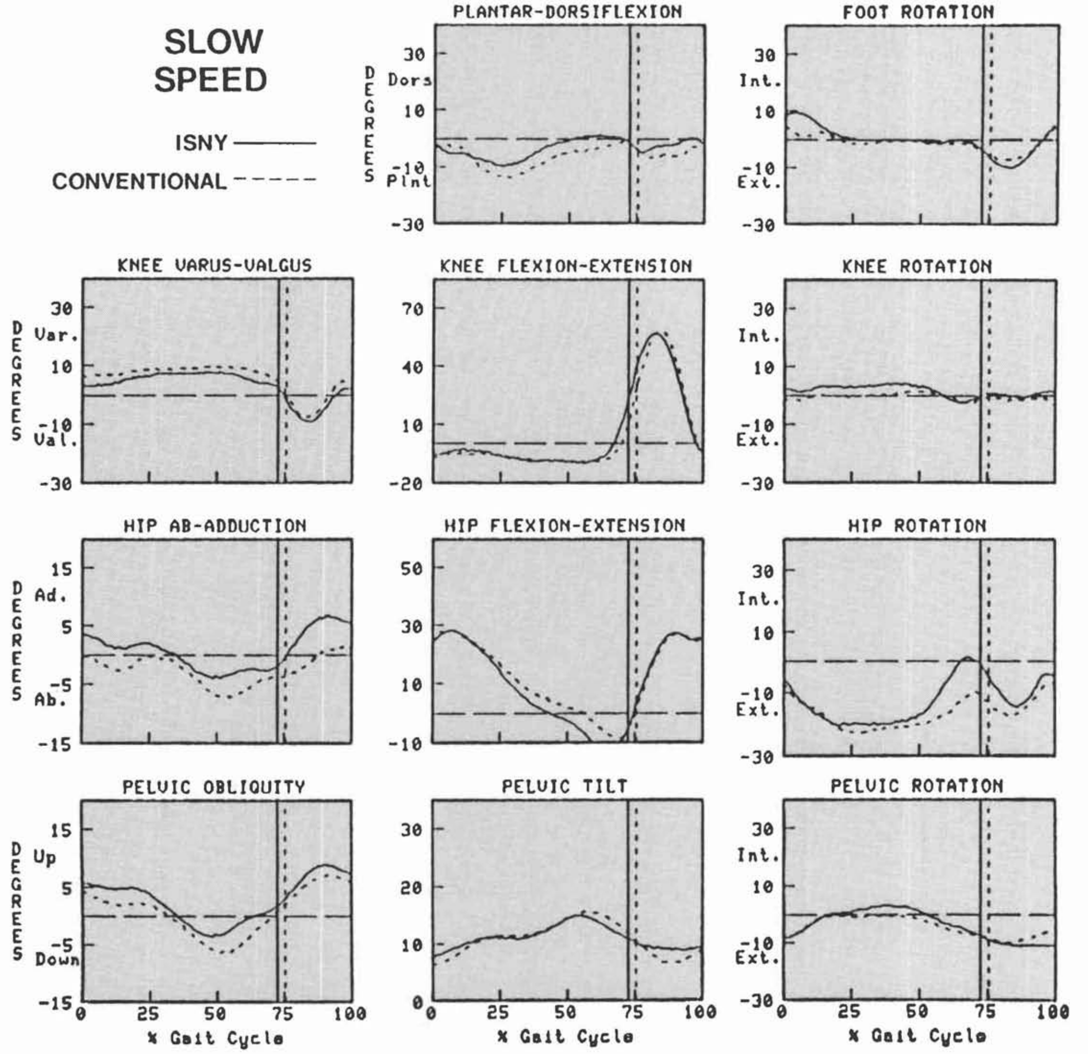


Figure 3.

The kinetic data reveal no medio-lateral differences between the ISNY and the conventional socket. With respect to vertical forces, the ISNY appears to have been loaded more quickly, and more forcefully, during "preferred" and "fast" walking than was the conventional socket. Again, this result can probably be attributed to the greater comfort of the ISNY socket, which permitted the amputee to load the limb in a more normal pattern than was the conventional socket.

General Observations Several general features of this amputee's gait are also noteworthy. Examination of the kinematic plots of knee joint motion during stance phase reveals that the knee is in hyperextension, under all walking conditions. That is, the knee bolt was set quite posterior to the hip and ankle, despite the recommendations from as long ago as 1954 that above-knee amputees with residual limbs as long as this patient's should probably not have the knee bolt offset posteriorly.[2,20] The kinetic plots indicate the unfortunate effect of this alignment stability: the floor reaction vector remains in front of the prosthetic knee even during late stance, when the amputee is attempting to roll over the keel of the SACH foot to begin knee flexion. Thus, the prosthetic limb is excessively stable, which substantially increases the work required for locomotion in comparison with an optimally aligned limb.

The fact that the prosthetic knee remained in

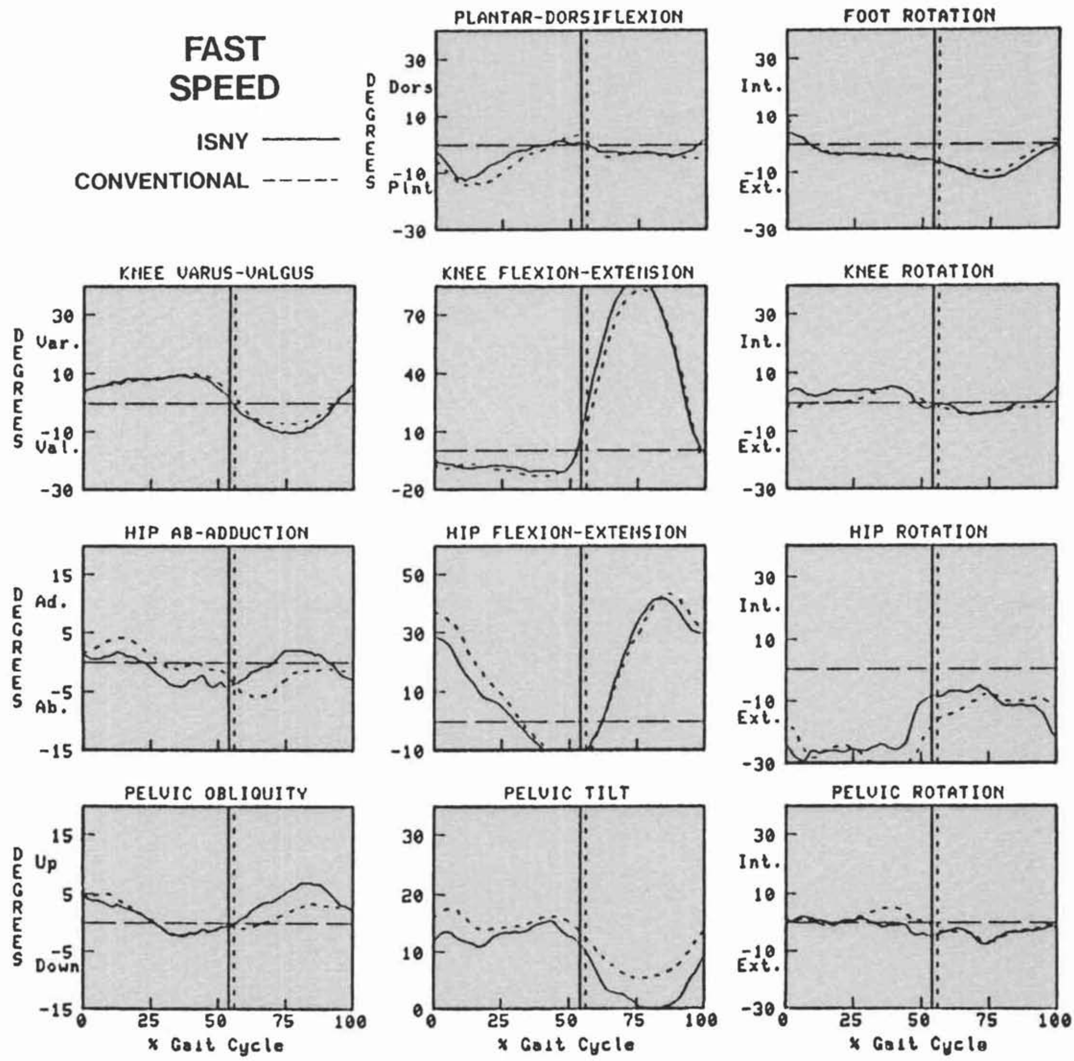


Figure 4.

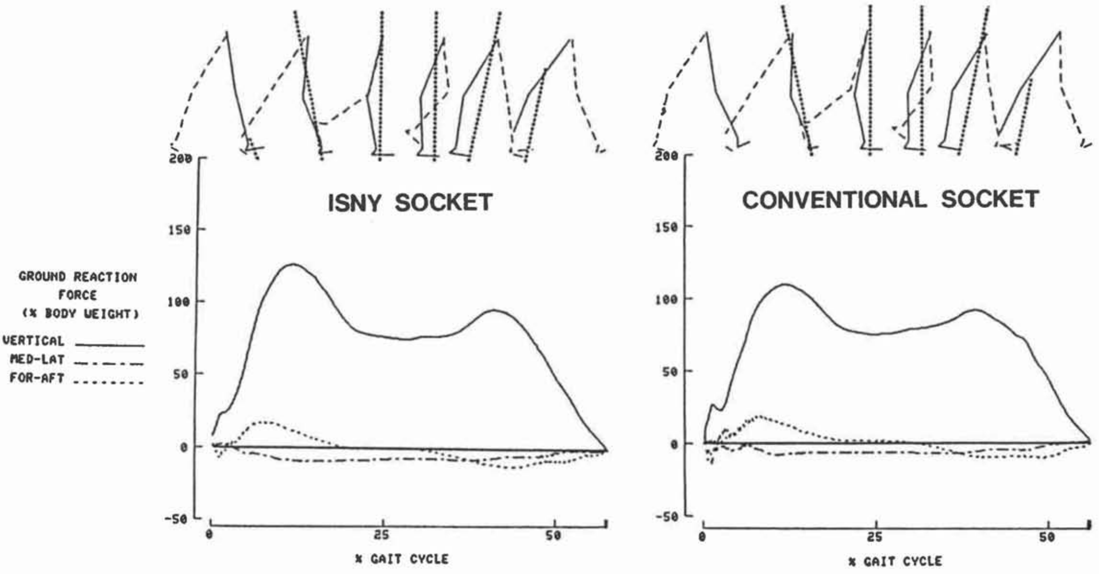
varus throughout stance phase no doubt reflects appropriate inset of the foot and shank. It is therefore not surprising that the pelvic obliquity kinematics and the medio-lateral kinetics rather closely approximate normal values.

SUMMARY

In summary, no important temporal, kinematic or kinetic differences were found during this single-subject comparison of an ISNY and a conventional rigid above-knee prosthetic socket. Indeed, we have speculated that these gait laboratory data may provide evidence of a functional advantage of the ISNY, with respect to stance phase duration and loading of the prosthesis. Therefore, given no functional disadvantage of the ISNY and given that the subject finds the ISNY to be significantly more comfortable, it may be concluded that the overall evidence favors use of the ISNY socket. Firm conclusions, however, must await further studies that compare comparably fitting ISNY and conventional sockets in larger groups of amputees.

REFERENCES

[1] Eberhart, Howard D.; Herbert Elftman; and Verne T. Inman, "The Locomotor Mechanism of the Amputee," *Human Limbs and Their Substitutes,* Ed. Paul E. Klopsteg and Philip D. Wilson. New York: McGraw-Hill, 1954b. p.p. 472–480.



Figures 5A and 5B.

	ISNY Socket			Conventional Socket		
	Slow	Preferred	Fast	Slow	Preferred	Fast
Toe-off (% of Gait Cycle)	72.2	63.4	54.0	75.7	64.6	56.1
Opposite Toe-off (%)	27.7	17.1	9.5	29.7	16.5	12.1
Opposite Heel-strike (%)	55.5	51.2	46.0	55.0	50.6	47.0
Single Stance (%)	27.7	34.2	36.5	25.2	34.2	34.9
Step Length (cm)	53.3	64.1	74.9	53.2	65.4	75.7
Stride Length (cm)	112.3	130.9	152.7	107.2	129.5	147.0
Cycle Time (sec)	1.7	1.4	1.1	1.9	1.3	1.1
Cadence (steps/min)	70.3	86.7	112.5	65.2	88.4	107.5
Walking Speed (cm/sec)	65.7	94.6	143.2	58.2	95.3	131.6

Table 1.

[2] Inman, Verne T., Henry J. Ralston and Frank Todd, *Human Walking,* Baltimore: Williams and Wilkins, 1981, p. 134.

[3] NYU Prosthetics and Orthotics Staff, *Lower-Limb Prosthetics, Including Prosthetist's Supplement,* New York: New York University Post-Graduate Medical School, 1982, p. 170, S233–263.

[4] Wilson, A. Bennett Jr., "Limb Prosthetics—1970," *Artific Limbs,* 14:1–52, 1970.

[5] Appoldt, Frank; Leon Bennett; and Renato Contini, "Stump-Socket Pressure in Lower Extremity Prostheses," *J Biomech,* 1:247–257, 1968.

[6] Kristinsson, Ossur, "Flexible Above-Knee Socket made from Low Density Polyethylene Suspended by a Weight-Bearing Frame," *Orthotics and Prosthetics,* 37(2):25–27, 1983.

[7] Gage, James R., "Gait Analysis for Decision-Making in Cerebral Palsy," *Bull Hosp Joint Dis,* 43:147–163, 1983.

[8] Gage, James R.; David Fabian; Ramona Hicks; and Scott Tashman, "Pre- and Post-Operative Gait Analysis in Patients with Spastic Diplegia: A Preliminary Report," *J Ped Orthop,* 4:715–725, 1984.

[9] Taylor, Kenneth D., Francois M. Mottier, Dan W. Simmons, et al., "An Automated Motion Measurement System for Clinical Gait Analysis," *J Biomech,* 15:505–516, 1982.

[10] Murray, M. Patricia; Susan B. Sepic; Gena M. Gardner; and Louise A. Mollinger, "Gait Patterns of Above-Knee Amputees Using Constant-Friction Knee Components," *Bull Prosthet Res,* 17(2):35–45, 1980.

[11] Leavitt, Lewis A.; Efrain N. Zuniga; Jon C. Calvert; et al., "Gait Analysis and Tissue-Socket Interface Pressures in Above-Knee Amputees," *South Med J,* 65:1197–1207, 1972.

[12] Pedotti, A., "A Study of Motor Coordination and Neuromuscular Activities in Human Locomotion," *Biol Cybern,* 26:53–62, 1977.

[13] Milner, Morris; John V. Basmajian; and Arthur O. Quanbury, "Multifactorial Analysis of Walking by Electromyography and Computer," *Am J Phys Med,* 50:235–258, 1971.

[14] Johnston, Richard C. and Gary L. Smidt, "Measurement of Hip-Joint Motion during Walking: Evaluation of an Electrogoniometric Method," *J Bone Joint Surg,* 51(A):1083–1094, 1966.

[15] Inman, Verne T., "Human Locomotion," *Can Med Ass J,* 94:1047–1054, 1966.

[16] Jacobs, N.A.; J. Skorecki; and J. Charnley, "Analysis of the Vertical Component of Force in Normal and Pathological Gait," *J Biomech,* 5:11–34, 1972.

[17] Lehman, Justus F.; Michael J. Ko; and Barbara J. deLateur, "Knee Moments: Origin in Normal Ambulation and Their Modification by Double-Stopped Ankle-Foot Orthoses," *Arch Phys Med Rehabil,* 63:345–351, 1982.

[18] Oberg, K. and H. Lanshammar, "An Investigation of Kinematic and Kinetic Variables for the Description of Prosthetic Gait using the ENOCH System," *Prosthet Orthot Intern,* 6:43–47, 1982.

[19] Chao, E.Y.; R.K. Laughman; E. Schneider; and R.N. Stauffer, "Normative Data of Knee Joint Motion and Ground Reaction Forces in Adult Level Walking," *J Biomech,* 16:219–233, 1983.

[20] Radcliffe, Charles W.; "Alignment of the Above-Knee Artificial Leg," *Human Limbs and Their Substitutes,* Ed. Paul E. Klopsteg and Philip D. Wilson. New York: McGraw-Hill, 1954. p.p. 677–692.

AUTHORS

David E. Krebs, M.A., P.T., Associate Research Scientist, Prosthetics and Orthotics New York University Post-Graduate Medical School, 317 East 34th Street, New York City, New York 10016.

Scott Tashman, M.S., Biomedical Engineer, Motion Analysis Laboratory, Newington Children's Hospital, Newington, Connecticut 06111.

ACKNOWLEDGMENT

This investigation was supported in part by grant MCJ-363-082-26-0, Maternal and Child Health, Bureau of Community Health Services, Public Health Service, Department of Health & Human Services, Rockville, Maryland.

The Application of Gait Analysis in Orthotics

by Robert S. Lin, C.P.O.

A gait analysis laboratory is an invaluable tool in the quantitative analysis of orthotic systems and their effect on human locomotion. This is particularly true in cases where the orthotic design is based on biomechanical behavior of the extremity during gait, as in the anterior floor reaction orthosis, the posterior offset knee mechanism and the Scott-Craig orthoses.

Traditionally, the success or failure of an orthosis has been based on clinical observation by the orthotist, physician or therapist, while relying on the latest medical record entry and their recollection of the patient's status. Even the most comprehensive dictations often fail to note important subtle factors.

On the other hand, a gait lab report provides a formal permanent record of the specific gait status of an individual. This detailed analysis can be reviewed any time.

Clinical application of the gait laboratory is best demonstrated in the management of an 11 year old spastic diplegic at Newington Children's Hospital. M.C. came to us with hip flexion contractures, bilateral knee flexion contractures, and equinovarus deformities of both feet. Despite these lower extremity contractures, he is ambulatory, exhibiting a markedly tenuous gait pattern and unable to stand in place.

Computerized gait analysis was performed pre-op and ten weeks post-op with and without the anterior floor reaction orthoses. In addition to the stick figures and ground reaction data, linear measurement of single stance percentage, stride length, walking velocity, and external work of walking were all obtained.

These results provided quantitative pre-op, post-op, and post-op with orthoses data which compared specific differences in gait behavior and the effects of surgery and orthotic management on these.

In addition to comparative studies of pre-op, post-op and post-op with orthoses conditions,

[14] Johnston, Richard C. and Gary L. Smidt, "Measurement of Hip-Joint Motion during Walking: Evaluation of an Electrogoniometric Method," *J Bone Joint Surg,* 51(A):1083–1094, 1966.
[15] Inman, Verne T., "Human Locomotion," *Can Med Ass J,* 94:1047–1054, 1966.
[16] Jacobs, N.A.; J. Skorecki; and J. Charnley, "Analysis of the Vertical Component of Force in Normal and Pathological Gait," *J Biomech,* 5:11–34, 1972.
[17] Lehman, Justus F.; Michael J. Ko; and Barbara J. deLateur, "Knee Moments: Origin in Normal Ambulation and Their Modification by Double-Stopped Ankle-Foot Orthoses," *Arch Phys Med Rehabil,* 63:345–351, 1982.
[18] Oberg, K. and H. Lanshammar, "An Investigation of Kinematic and Kinetic Variables for the Description of Prosthetic Gait using the ENOCH System," *Prosthet Orthot Intern,* 6:43–47, 1982.
[19] Chao, E.Y.; R.K. Laughman; E. Schneider; and R.N. Stauffer, "Normative Data of Knee Joint Motion and Ground Reaction Forces in Adult Level Walking," *J Biomech,* 16:219–233, 1983.
[20] Radcliffe, Charles W.; "Alignment of the Above-Knee Artificial Leg," *Human Limbs and Their Substitutes,* Ed. Paul E. Klopsteg and Philip D. Wilson. New York: McGraw-Hill, 1954. p.p. 677–692.

AUTHORS

David E. Krebs, M.A., P.T., Associate Research Scientist, Prosthetics and Orthotics New York University Post-Graduate Medical School, 317 East 34th Street, New York City, New York 10016.

Scott Tashman, M.S., Biomedical Engineer, Motion Analysis Laboratory, Newington Children's Hospital, Newington, Connecticut 06111.

ACKNOWLEDGMENT

This investigation was supported in part by grant MCJ-363-082-26-0, Maternal and Child Health, Bureau of Community Health Services, Public Health Service, Department of Health & Human Services, Rockville, Maryland.

The Application of Gait Analysis in Orthotics

by Robert S. Lin, C.P.O.

A gait analysis laboratory is an invaluable tool in the quantitative analysis of orthotic systems and their effect on human locomotion. This is particularly true in cases where the orthotic design is based on biomechanical behavior of the extremity during gait, as in the anterior floor reaction orthosis, the posterior offset knee mechanism and the Scott-Craig orthoses.

Traditionally, the success or failure of an orthosis has been based on clinical observation by the orthotist, physician or therapist, while relying on the latest medical record entry and their recollection of the patient's status. Even the most comprehensive dictations often fail to note important subtle factors.

On the other hand, a gait lab report provides a formal permanent record of the specific gait status of an individual. This detailed analysis can be reviewed any time.

Clinical application of the gait laboratory is best demonstrated in the management of an 11 year old spastic diplegic at Newington Children's Hospital. M.C. came to us with hip flexion contractures, bilateral knee flexion contractures, and equinovarus deformities of both feet. Despite these lower extremity contractures, he is ambulatory, exhibiting a markedly tenuous gait pattern and unable to stand in place.

Computerized gait analysis was performed pre-op and ten weeks post-op with and without the anterior floor reaction orthoses. In addition to the stick figures and ground reaction data, linear measurement of single stance percentage, stride length, walking velocity, and external work of walking were all obtained.

These results provided quantitative pre-op, post-op, and post-op with orthoses data which compared specific differences in gait behavior and the effects of surgery and orthotic management on these.

In addition to comparative studies of pre-op, post-op and post-op with orthoses conditions,

Estimated External Work of Walking (joules/kg/meter)		
Preop.	Postop. Braces	Postop. No Braces
1.118	0.485	0.479

Comparison of data from gait analysis of M.C.

gait analysis can effectively perform comparative studies between two different orthotic designs, as well as two different orthoses of the same design. A recent study performed on a 30 year old female, with poliomyelitis and unilateral lower extremity involvement, compared various orthotic configurations.

INTERPRETATION AND DISCUSSION OF DATA

History

Ms. Jones is a 29 year old female with a history of poliomyelitis at age two. The patient was left with lower extremity weakness but was an independent ambulator. Approximately five years ago, the patient sustained a right comminuted supra-condylar fracture which required closed reduction and casting. The patient had a long recovery and rehabilitation time and is still engaged actively in physical therapy. At this time she utilizes a double right metal KAFO and a cane for walking. She recently obtained a floor reaction orthosis, but is able to use this for only a brief time while walking during therapy. Her principal complaint with this floor reaction brace is that she fatigues much more quickly than with her metal KAFO. The Gait Analysis Laboratory assessed the patient's ambulation with her existing orthoses. While the patient's physical therapist and the patient relate significant improvement in both motion strength and endurance in the intervening four years, they are concerned about improving endurance.

Comments on Linear Measurements

The patient has an asymmetrical single stance time and her right step lengths are consistently longer throughout testing. The patient's stride length and velocity increased in these modes: shoe only (no orthosis), floor reaction orthosis, and metal KAFO, respectively. Her best gait in terms of linear measurements approximates only 50 percent of normal walking velocity.

RIGHT LOWER EXTREMITY

Coronal Plane

The right pelvis is down 5–10 degrees in pelvic obliquity and the right hip is held predominantly in 15–20 percent of abduction. (Knee) Varum-valgum is normal. With the floor reaction orthosis on, both pelvic obliquity and hip abduction are reduced by approximately five degrees. With the KAFO on, both pelvic obliquity and hip abduction are increased.

Transverse Plane

With the floor reaction orthosis, pelvic rotation is normal, hip rotation is slightly external, and knee rotation is significantly normalized. Foot rotation is skewed toward five degrees of internal rotation. With the KAFO on, pelvic rotation is unchanged. There is a marked change in hip rotation, being 20–30 degrees externally rotated. Knee rotation is neutral and motion, as expected, is eliminated, as is foot rotation.

Linear Measurements—M.C.						
	Right Side (11/11/82)			Left Side (11/11/82)		
	Preop.	Post-op Braces	Post-op No Braces	Preop.	Post-op Braces	Post-op No Braces
Single Stance (%)	34.48	31.25	23.08	34.48	34.47	26.83
Step Length (cm)	37.30	39.70	37.80	44.60	48.40	41.20
Walking Velocity (cms/sec)	83.5	78.41	53.89	83.55	78.41	53.89

LEFT LOWER EXTREMITY

Coronal Plane

The left hip is consistently hiked 5–10 degrees and is in 15 degrees of adduction (from her leg length discrepancy). (Knee) Varum-valgum is normal. The floor reaction orthosis does not significantly change her pelvic obliquity or hip ab-adduction. With the KAFO on, she demonstrates a mild decrease in pelvic obliquity and hip adduction.

Sagittal Plane

Pelvic tilt is off the graph 30^+ degrees. Hip flexion-extension has a fairly normal excursion and is increased approximately 10 degrees. The knee is mildly hyperextended during stance and has good excursion during swing phase. The patient demonstrates a mild drop-foot during swing phase. The patient's AFO does not change her pelvic tilt, hip flexion-extension, knee flexion-extension, nor foot plantar-dorsiflexion significantly. However, the patient's KAFO does significantly decrease her pelvic tilt and hip flexion-extension or foot plantar-dorsiflexion on that side.

Transverse Plane

Pelvic rotation is essentially normal. There tends to be slightly more external rotation at the hip. Knee and foot rotation are essentially normal. The rotational plots are not significantly altered by either the ankle-foot orthosis or the metal KAFO.

RECOMMENDATIONS

Although the patient appreciates the floor reaction orthosis as it is more cosmetically acceptable, it is obvious that she will need the use of her cane to minimize her energy requirements. It was recommended that a new floor reaction orthosis be fitted to improve the overall gait and possibly increase her endurance.

DISCUSSION

The patient returned for a repeat gait analysis so we could analyze the efficiency of the new floor reaction orthosis and shoe that she acquired after a gait analysis on October 3, 1984. In addition to the new AFO, the patient ac-

Summary of Linear Measurements—Jones

	J00;1 Left cane, shoes	J01 Left cane, right floor reaction. brace	J02 Left cane, right KAFO	Normal Adults
R. Single Stance (%)	28.26	31.71	31.82	35.07
L. Single Stance (%)	30.43	34.15	38.64	35.07
R. Step Length (cm)	42.40	38.10	57.90	65.14
L. Step Length (cm)	27.00	38.20	38.10	65.14
Stride Length (cm)	70.45	75.65	94.90	130.29
Cadence (S/min)	76.60	85.71	80.00	108.24
Walking Velocity (cms/sec.)	44.97	54.04	63.27	117.42

Summary of Linear Measurements—Jones

	J10 New AFO	J11 Old AFO	Normal Adults
R. Single Stance (%)	23.1	25.9	35.07
L. Single Stance (%)	34.0	29.8	35.07
R. Step Length (cm)	30.7	27.7	65.14
L. Step Length (cm)	33.0	34.3	65.14
Stride Length (cm)	63.5	61.6	130.29
Cadence (S/min)	69.3	61.5	108.24
Walking Velocity (cms/sec.)	36.6	31.6	117.42

quired new tennis shoes with the right one having approximately a $5/8''$ buildup on the entire heel. Subjectively, the patient stated that she is encouraged by the use of this new AFO and is both comfortable and functional; however, she still fatigues easily and she does not have total confidence in the orthosis.

SUMMARY

The purpose of this gait analysis was to evaluate the effectiveness of the new floor reaction orthosis. The orthosis still allowed too much knee flexion during stance phase, and although the patient was happy with the orthosis subjectively and it had increased her endurance, it could still be fine-tuned further. The orthosis itself is adequate but the shoe platform could be modified in one of two ways:

1) The heel could be made of a much softer material similar to a SACH heel shock absorber, thus effectively allowing her more plantarflexion and increase in the efficiency of the extension couple.
2) Alternately, the heel could be ground down, removing some of the lift.

It was decided to treat the shoe much like a SACH heel cushion of a prosthesis, and add further cushion in order to allow the orthosis to become more effective in the face of her inadequate quadriceps.

CONCLUSION

This comprehensive report is a compilation of the data generated and the physician's interpretation of this data. It is obvious that such a report gives the clinician a patient picture that is far superior to all other available documentation. It also enables the progress to be quantified and compared numerically to previous analyses run either pre-op or with different orthotic applications.

The gait analysis laboratory can be used as an adjunct to empirical clinical observation in assisting the orthotist in many of the important decision-making processes. It can reduce the level of ''artistry'' that is presently a significant component in orthotics, while introducing a level of science to the orthotic design and prescription process.

With this clinical tool, the complex orthotic problems we face daily can be better analyzed as the abnormalities of gait are monitored, documented, and interpreted by the orthotist and physician using the laboratory.

AUTHOR

Robert S. Lin, C.P.O. is with the Department of Orthotics and Prosthetics at Newington Children's Hospital, Newington, CT 06111.

Results of the Questionnaire on Upper Extremity Prostheses

There were 10 responses (one of which was considered so confusing as to be useless). Eight of the nine respondents, whose results were considered, stated that they fit less than 15 new upper limb prostheses, and six fit less than 10. One respondent stated that his facility had fit 55 new upper limb prostheses last year (this represented about 25 percent of all new prostheses delivered in that period).

Seven respondents replied that upper extremity amputees represented less than 20 percent of their patient population and two stated that they amounted to 20–40 percent.

In response to question 3, the average number of externally powered prostheses fit among the respondents was 3.2. The most common response (five) was zero. One individual reported fitting 18 (the same individual above who reported fitting a total of 55 upper extremity prostheses, 33 percent).

The responses to question 4, concerning which upper extremity prosthesis they considered most beneficial to a patient, were fairly equivocal. There were $2^1/2$ votes for body powered prostheses, $3^1/2$ in favor of externally powered prostheses, and two for hybrid prostheses.

Concerning the remarks by John Billock, C.P.O., in his article, "Upper Limb Prosthetic Management—Hybrid Design Approaches," whether they considered a hook or hand most appropriate, the respondents gave $5^1/2$ votes to externally powered hands, $2^1/2$ to hooks, and none to body powered hands.

In response to what their patients preferred, three respondents stated that their patients preferred hooks, another individual stated that half his patients preferred hooks, three others said that their patients were in favor of externally powered hands, and one said that his patients preferred body powered hands.

Eight of the respondents said that the results of LeBlanc's survey concurred with their experience and one said fifty-fifty.

In considering question 8 concerning preference for R&D, the number of number one and number two responses were totalled. The results were as follows:

Improved External Powered Prostheses, including provisions for hybrid design	8
Cosmetic Gloves and Skins	5
Sensory Feedback	3
Improved Body Powered Prostheses	2
Hooks	1
Hands	1

Questionnaire: Improved Fitting Techniques

1. Do you use Check Sockets?
   _______ Routinely
   _______ Occasionally
   _______ Never

2. Do you use alginate?
   _______ Routinely
   _______ Occasionally
   _______ Never

3. Do you use x-rays or Xeroradiography?
   _______ Routinely
   _______ Occasionally
   _______ Never

4. Do you use videotape during dynamic alignment?
   _______ Routinely
   _______ Occasionally
   _______ Never

5. Which single development would you most like to see as an aid in fitting? _______
   ____
   ____

6. Comments: _______________________
   ____
   ____
   ____
   ____
   ____

Dear Reader:

Questionnaires have been a regular feature of *Clinical Prosthetics and Orthotics* for quite some time now. Responses have never been overwhelming. Recently it has been proposed that the questionnaire be discontinued. It has also been proposed that the question be put to the readership. Therefore, are you in favor of continuing or discontinuing the questionnaire?

_______ Continue
_______ Discontinue

Send all completed questionnaires to: Charles H. Pritham, C.P.O., c/o Durr-Fillauer Medical, Inc., Orthopedic Division, 2710 Amnicola Highway, Chattanooga, Tennessee 37406.

Post Office
Will Not Deliver
Without Postage

Charles H. Pritham, CPO
Durr-Fillauer Medical, Inc.
Orthopedic Division
2710 Amnicola Highway
Chattanooga, TN 37406

FOLD UNDER AND TAPE CLOSED

"Resource for the Rehabilitation Professional"

Clinical Prosthetics & Orthotics

CLINICAL PROSTHETICS AND ORTHOTICS IS THE PROFESSIONAL PUBLICATION YOU NEED TODAY.

Originally a newsletter, *C.P.O.* has grown to a scientific publication concerned with philosophical issues which arise in the clinical atmosphere of the profession. Now, the Academy has further raised the standard of *C.P.O.* to better reflect the professional needs of its readers in content, format and style.

The new 24 page format allows room for additional articles, letters, and reports from the field. Each issue continues to publish high quality articles—lead, editorial, technical and short technical notes—which discuss a single issue/topic from several professional viewpoints.

C.P.O. also continues to run its survey—asking readers to respond to the articles in the issue. The results of each survey are published in the following issue. The survey is a valuable tool through which practitioners can voice their experiences and opinions on significant issues to aid the growth of the profession.

The prosthetic-orthotic community has found *C.P.O.* to be such an asset to the profession that O&P suppliers and facilities have decided to sponsor individual issues of *C.P.O.* This support and a 54% increase in subscribers since October, 1983 have made possible this latest expansion of *C.P.O.*

C.P.O. is what you, the health-care professional, need to keep abreast of what is happening to the fields of orthotics and prosthetics. Quality medical journals are one of the professional's greatest assets. In view of the 1st-class nature of *C.P.O.*, yearly subscription rates to the publication are very low. You therefore can obtain one of your necessary resources with little investment. And you *can* look on your purchasing of *C.P.O.* as an investment—an investment in your professional standing and abilities.

Join the over 2000 health-care professionals who have long read *C.P.O.* and who are now getting an even greater return on their investment dollar. Don't miss out. Get the new *C.P.O.* today!

SUBSCRIPTION INFORMATION AND FORM ARE PROVIDED ON THE BACK.

Clinical Prosthetics & Orthotics

RATES

Domestic Subscription (U.S., Canada and Mexico)	$15.00
Foreign subscription sent via surface mail	$20.00
Foreign subscription sent via air-mail	$30.00

Please detach, complete and mail the order form below along with your payment* to:

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

*All checks must be payable in U.S. currency to the American Academy of Orthotists and Prosthetists. Please do not send cash or UNESCO coupons.

C.P.O. Subscription Order Form

Mail to: The American Academy of Orthotists and Prosthetists
717 Pendleton Street
Alexandria, VA 22314

Upon the receipt of your order and payment we will begin your subscription with the next published quarterly issue of ***C.P.O.***

NAME: ______________________________

ADDRESS: ______________________________

CITY: ______________ STATE: __________ ZIP: __________

C.P.O. Gift Subscription Order Form

NAME OF RECIPIENT: ______________________________

ADDRESS: ______________________________

CITY: ______________ STATE: __________ ZIP: __________

Calendar

1985

September 6–7, American Academy of Orthotists and Prosthetists Continuing Education Conference 4-85, "Paraplegic Management," Boston, Massachusetts.

September 13–15, Fifth Annual *Advanced* Course in "Lower Extremity Amputation and Prosthetics," Nassau County Medical Center, East Meadow, New York. Contact: Lawrence W. Friedmann, M.D., Chairman, Dept. of Physical Medicine and Rehabilitation, Nassau County Medical Center, 2201 Hempstead Turnpike, East Meadow, NY 11554; (516) 542-0123.

September 13–14, Ohio Chapter of the Academy Meeting, Resort Inn, Kings Island, Ohio. Contact: Jon Leimkuehler, CPO, (216) 651-7788.

September 15–October 6, European Educational Tour, under the auspices of the Academy Midwest Chapter. Contact: Norbert Fliess, CP, American Limb & Orthopedic Co., 1724 West Ogden Avenue, Chicago, Illinois 60612.

September 21, Pennsylvania Chapter of the Academy meeting, Elizabethtown Hospital and Rehabilitation Center, Elizabethtown, Pennsylvania.

November 2, Midwest Chapter of the Academy Fall Continuing Education Seminar.

November 15–16, American Academy of Orthotists and Prosthetists Continuing Education Conference 5-85, "Powered Limb Prosthetics," Downtown Holiday Inn, Atlanta, Georgia.

1986

January 27–February 2, Academy Annual Meeting and Scientific Seminar, MGM Grand, Las Vegas, Nevada. Contact: Academy National Headquarters: (703) 836-7118.

February 20–25, American Academy of Orthopedic Surgeons Annual Meeting, New Orleans, Louisiana.

April 8–11, Pacific Rim Conference, Intercontinental Hotel, Maui, Hawaii.

April 12, Midwest Chapter of the Academy Spring Continuing Education Seminar/Social Event.

March 13–15, American Academy of Orthotists and Prosthetists Continuing Education Conference 1-86, "Spinal and Seating Orthotics," Birmingham, Alabama.

May 16–17, American Academy of Orthotists and Prosthetists Continuing Education Conference 2-86, "Disarticulation Prosthetics," Ann Arbor, Michigan.

June 19–22, AOPA Region VI and Academy Midwest Chapter Combined Annual Meeting, Lakelawn Lodge, Delavan, Wisconsin.

July 18–19, American Academy of Orthotists and Prosthetists Continuing Education Conference 3-86, "Spina Bifida," Cincinnati, Ohio.

September 19–20, American Academy of Orthotists and Prosthetists Continuing Education Conference 4-86, "Powered Limb Prosthetics," Newington, Connecticut.

October 24–25, American Academy of Orthotists and Prosthetists Continuing Education Conference 5-86, "Lower Limb Prosthetics," Kansas City, Missouri.

Academy Tallies Election Results

Academy members elected the following officers and directors to serve their terms in 1986:

President-Elect
Wm. C. Neumann, CPO

Vice-President
Alvin C. Pike, CP

Secretary-Treasurer
John W. Michael, CPO

Directors
Joseph M. Leal, CP
Terry J. Supan, CPO
Dennis E. Clark, CPO

Academy Representative to ABC Board of Directors
Paul V. Murka, CPO

The officers and directors named above will be seated during the Academy Annual Business Meeting in Las Vegas, January 31, 1986. Mr. Murka will be seated on the ABC Board of Directors during the AOPA National Assembly in October, 1985.

Congratulations to the newly elected officers and Board members.

LAS VEGAS

Site of Academy Annual Meeting and Scientific Seminar at the MGM Grand, January 27–February 2.

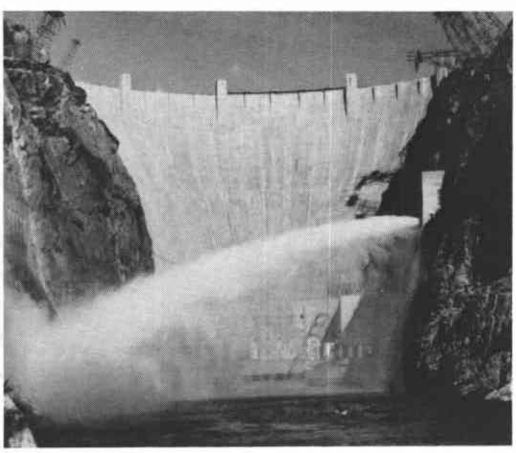
AMERICAN ACADEMY of ORTHOTISTS and PROSTHETISTS
AAOP

Fantastic Las Vegas, Nevada

On the outskirts of the growing metropolis of Las Vegas, visitors can absorb a stunning assortment of natural and man-made wonders. Lake Mead, only 30 minutes from the heart of Las Vegas, is the largest man-made reservoir in the Western Hemisphere. One hundred miles long, it boasts an even more incredible 500 miles of shoreline. Swimming, water skiing, camping, boating and fishing are among the activities to be readily enjoyed by visitors.

Equally spectacular is the world-famous Hoover Dam, also only 30 minutes from the center of town. The dam stands 726 feet high, the equivalent of 70 stories. Built 50 years ago, this spectacular engineering monument supplies power to Arizona, Southern California, Colorado, Wyoming, New Mexico, and Nevada.

Just a few miles from Las Vegas is the breathtaking Red Rock Canyon which has evolved from a 400-million-year-old sea bed into a series of magnificent geological formations, wind-sculptured sandstone out-croppings, unique desert vegetation and a new visitors center. Red Rock varies in elevation from 3,500 to 7,500 feet.



Some 20,000 years ago Indians inhabited the Valley of Fire, 50 miles north of Las Vegas. Visitors can marvel at the bright sandstone formations. Petroglyphs, etched in sandstone, may be seen at various points in the Valley of Fire State Park, where enchanting picnic sites and rustic camp grounds are readily available.

Southern Nevada's wild and turbulent history is still preserved in some of the area's ghost towns. Many are just a short jaunt from the heart of Las Vegas and offer an exciting, living testament to the American West, as portrayed for years in movies and novels.

Boat excursions on mammoth Lake Mead are a refreshing contrast to the surrounding desert. Or, for the more adventurous, day trips by plane to the fabulous Grand Canyon are a regularly-scheduled, added attraction. Camera buffs will especially enjoy the once-in-a-lifetime opportunity to witness, from the air, this 200-million-year-old natural phenomenon, where native Americans began taking up residence 300 years before Columbus discovered America.

C.P.O. MANUSCRIPT GUIDELINES

1. Manuscripts must be typewritten, double-spaced with wide margins.
2. Indicate bibliographical references by means of Arabic numerals in parentheses (6).
3. Write out numbers less than ten.
4. Bibliography should follow the correct forms as shown below.
   a. Book
      Murphy, Eugene F., Ph.D., "Lower-Extremity Component," *Orthopedic Appliances Atlas,* Vol. 2, J.W. Edwards, 1960, pp. 217–224.
   b. Journal Article
      Panton, Hugh J., B.S., C.P.O., "Considerations for Joints and Corset," *Newsletter . . . Amputee Clinics,* 8:3: June, 1975, pp. 1–3, 6–7.
   c. Lecture or Verbal Presentation
      1. Holmgren, Gunnar, "The PTB Suction Prosthesis" from the written material of a lecture delivered at the third of the "Strathclyde Bioengineering Seminars," 8–11 August, 1978.
      2. Wagner, F.W., Jr.: "Classification and treatment for diabetic foot lesions"; Instructional Course, American Academy of Orthopedic Surgeons, New Orleans, Louisiana, February, 1976.
   d. Personal Communication
      Irons, George, C.P.O., Personal communication, June 1977. Presently, Director of Research, United States Mfg., Glendale, California. Formerly, Research Prosthetist, Patient Engineering Service, Rancho Los Amigos Hospital, Downey, California.

   Arrange all references alphabetically.
5. Illustrations
   a. Write authors name on the back of each illustration.
   b. Number all illustrations to correspond to references from the text.
   c. Submit either an original or a photographic black & white reproduction of illustrations.
6. Photographs
   a. DO NOT WRITE ON THE BACK OF PRINTS OR USE PAPERCLIPS OR STAPLES ON PRINTS.
   b. Identify by number to correspond to references from the text on a piece of paper taped to the back of each print.
7. List all captions in order on a separate sheet of paper, typed.
8. Must include with the manuscript signed copies of releases from patients who are photographed and referred to in the text. For pictures of patients for which permission cannot be obtained, the eyes must be masked out.
9. Mail the original manuscript with accompanying photographs/illustrations and one xerox to:

   Charles H. Pritham, C.P.O.
   c/o Durr-Fillauer Medical, Inc.
   Orthopedic Division
   2710 Amnicola Highway
   Chattanooga, TN 37406

The American Academy of Orthotists and Prosthetists

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