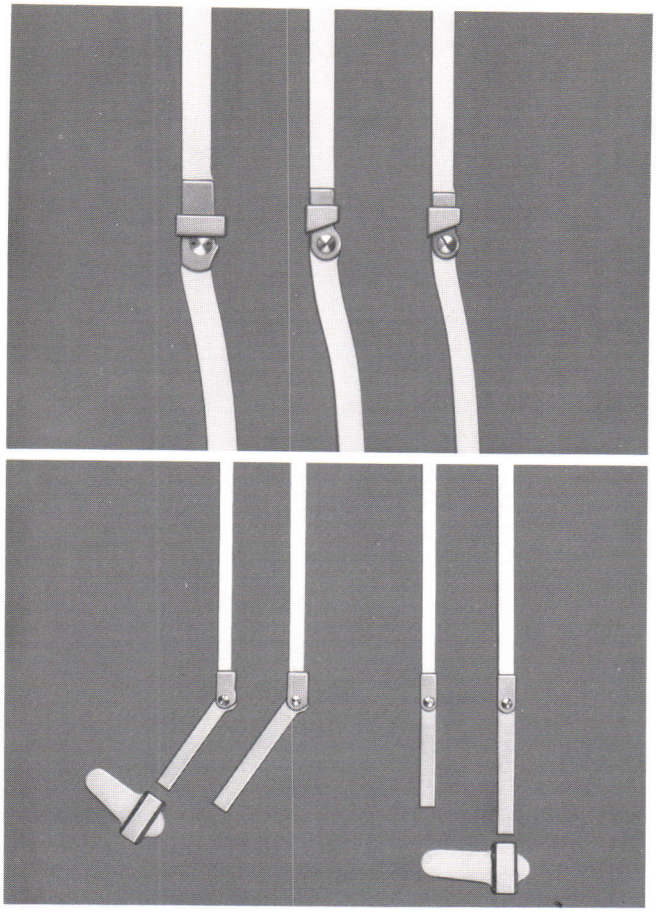
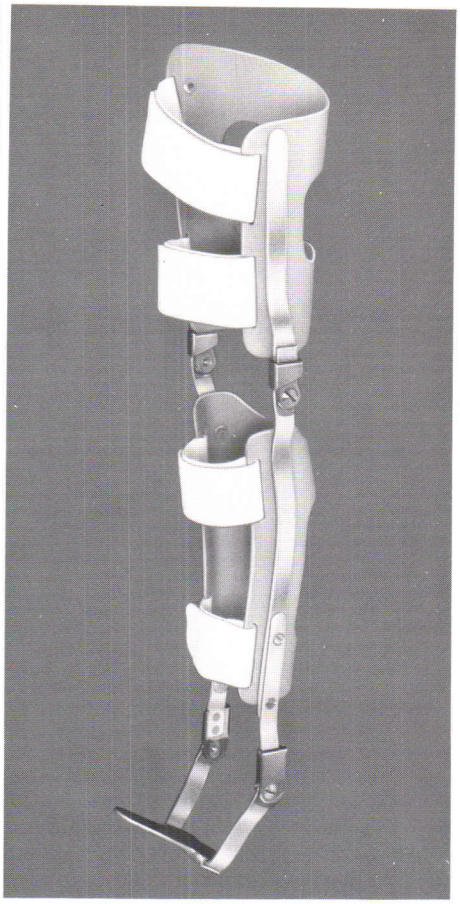
Winter 1988
Volume 41
Number 4

Orthotics and Prosthetics

Journal of the American Orthotic and Prosthetic Association





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Orthotics and Prosthetics

Winter 1988 **Volume 41, Number 4**

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Orthotics and Prosthetics (ISSN 0030-5928) is published quarterly by the American Orthotic and Prosthetic Association, 717 Pendleton Street, Alexandria, VA 22314. Subscriptions: $50 domestic (U.S., Canada, Mexico); $50 foreign; $60 foreign airmail. Membership dues of $24 for each member are set aside for *Orthotics and Prosthetics*. Second-class postage paid at Alexandria, Virginia and additional mailing offices. POSTMASTER: Send address changes to *Orthotics and Prosthetics*, 717 Pendleton Street, Alexandria, VA 22314.

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Letters to the Editor

To the Editor:

In the Fall, 1987 *Orthotics and Prosthetics,* Volume 41, Number 3, Mr. William T. Lovegreen, C.P.O., proposed using the Seattle Foot™ to provide a waterproof functional prostheses. Because many of our patients live on or near the coast, we have attempted this application on several occasions.

Unfortunately, we have not had satisfactory long-term results with the Seattle Foot™. In our experience, within a few weeks the sand and rocks of the beach abraid the polyurethane "skin" on the planter service of the foot. Once this happens, micro cracks appear and the foot gradually absorbs water until it is noticeably heavier. In essence, it becomes waterlogged.

Unfortunately, once this occurs, we have found it impossible to dry the foot completely. Over a period of time, it begins to make objectionable "squishing" noises, and sometimes supports the growth of discoloring fungus.

For that reason, we no longer recommend the Seattle Foot™ as a waterproof design. We have had somewhat more favorable results by special ordering a S.A.F.E. foot from Campbell-Childs with an all-plastic bolt block.

Perhaps Mr. Lovegreen or Model and Instrument Development could comment on their longer term experience in this regard.

Sincerely,
John W. Michael, C.P.O.
Director
Duke University Medical Center
Durham, North Carolina

Dear Editor:

After reading the critical comments made by Portnow and Boutrass in the Fall, 1987 issue of *Orthotics and Prosthetics,* concerning the New England Preparatory Prosthesis, I feel obliged to say a few words in behalf of O'Toole and DeCarteret who favor this type of prosthesis for the particular amputees they serve.

The NEPP has been used successfully in the New England area since before 1937 without producing the dire results listed by Portnow and Boutrass. For 25 years or more they have been routinely prescribed at Amputee Clinics, particularly in the Boston area. Can anyone imagine that respected orthopedic, vascular and general surgeons would allow their amputee patients to be fitted with devices which caused "problems with edema, stump shape, wound healing, pain and distal migration of the stump"? Of course not, the very thought is ridiculous.

It is my opinion that you measure success by the results produced rather than by the particular methodology. Both Dr. O'Toole and Donald DeCarteret, C.P., are experts in their respective fields and are well-acquainted with all types of immediate and early prosthtic fittings. Mr. DeCarteret, by virtue of his training, education and his many years experience in the prosthetic profession is eminently qualified to provide any prescribed prosthetic service. I am certain that neither he nor Dr. O'Toole would ever provide any amputee with a device which would be detrimental to his or her well-being.

I feel it's only fair to state that I believe that Dr. Portnow and George Boutrass,

C.P.O., have every right to provide amputee services which they feel are appropriate for their amputee patients. I'm sure that no one would deny them that right. However, I feel that they should be tolerant of the viewpoints of other prosthetists.

In closing I would say that during my 44 years in the prosthetic profession, I have heard criticisms cast at all types of prosthetic fittings. Apparently none are perfect. They all have their pros and cons. What suits one may not be good for another. But if you provide what you sincerely feel is the very best for your amputee patient you should be above reproach. You may not be right all of the time, but who is?

Very truly yours,
Howard V. Mooney, C.P.(E)
37 Walker Street
Lowell, Massachusetts 01854

Dear Editor:

The newly labelled "New England Preparatory Prosthesis" was created at least 60 some years ago. It was requested by doctors who at that time were mounting a rigid dressing on a crutch lower.

At that time, they recognized the need for early post-op fitting and ambulation.

I am proud that our area, and mainly Boston, was amongst the forerunners in this procedure.

Such old timers, as my father, Mr. Mooney, Mr. Hitchock and a few others, were directly involved in the development of the preparatory prosthesis.

This type preparatory (above and below knee) was commonly used as prescribed, by institutions of medicine as old as anesthesia.

Doctor O'Toole and Donald DeCarteret, C.P., presented in their article, the use of this type prosthesis when fashioned by artisans who were skilled in the configuration and practicality.

To the best of my knowledge (42 years) there is no one prosthesis that is the best for all.

We are, and should be, prescribing per patient. Prosthetists and doctors, other than the young, are aware that total contact (not end-bearing) variable volume plastic sockets with alignment, adjustment capabilities, have a long history, also in the Boston area.

Our clinical affiliation with many amputee clinics, always had an open mind as to the type of prosthesis that would best assist the patient's rehabilitation.

As such, many types are in use, most quite successful in patient care.

Yours very truly,
Joseph H. Martino, C.P.

REFERENCES

O'Toole, D.M. and D.R. DeCarteret, "A Commentary on the New England Preparatory Prosthesis," *Orthotics and Prosthetics,* Vol. 41, No. 2, Summer, 1987, pp. 29–31.

Portnow, D.M. and Boutross, "Letter to the Editor," *Orthotics and Prosthetics,* Vol. 41, No. 3, Fall, 1987, p. 10.

Lower Limb Prosthetics, New York University, 1980. p. 107.

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Calendar of Events

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

1988

March 18–19, Academy Continuing Education Conference 1-88 and Midwest Chapter Combined Meeting, "Current Clinical and Technical Concepts in Lower Limb Prosthetics and Orthotics," Howard Johnson O'Hare International Hotel, Chicago, Illinois. Contact: Academy National Headquarters, (703) 836-7118.

April 6, "Graph-Lite Orthotics," Daw Industries Advanced Continuing Education Seminar, Certificate CEC course. Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

April 7–8, ABC Technician Exam, Northeast Metro Technical Institute, Minneapolis, Minnesota. Contact: ABC National Office, (703) 836-7114.

April 7, 8, 9, "Anatomical Design Socket and Advanced Prosthetic Techniques," Certificate CEC course. Flexible AK/BK & Narrow ML (CAT-CAM/NSNA equivalent). Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

April 8–9, Freeman Orthotic Fitters Training Workshop, Louisville, Kentucky. For more information, write: Freeman, Drawer J, Sturgis, Michigan, or call Cameron Brown, 800-253-2091.

April 11–15, 10th Congress of the International Federation of Physical Medicine and Rehabilitation, Sheraton Hotel, Toronto, Ontario. Contact: Secretary, 545 Jarvis Street, Toronto, Ontario M4Y 2H8, Canada.

April 13, 14, 15, "Anatomical Design Socket and Advanced Prosthetic Techniques," Certificate CEC course. Flexible AK/BK & Narrow ML (CAT-CAM/NSNA equivalent). Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

April 15–17, Region I Meeting, Boston, Massachusetts. Contact: Bill Miller, CO, (617) 735-6887.

April 20–22, Hosmer Electric Systems Workshop, VoTech Institute 916, Minneapolis, Minnesota. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; (408) 379-5151 or (800) 538-7748.

April 20–23, Third Bi-Annual Pacific Rim Prosthetic/Orthotic Conference, Turtle Bay Hilton, Oahu, Hawaii. Contact: Theresa LaPorte, Pacific Rim Conference, 1621 East Fremont Street, Las Vegas, Nevada 89101; tel. (702) 382-8287.

April 23–24, ABC Written/Visual Examination. Contact: ABC National Office for locations, (703) 836-7114.

April 24–27, The National Symposium on Information Technology as a Resource to Health and Disability Professionals, The Sheraton Charleston Hotel, Charleston, South Carolina. Sponsored by the South Carolina Center for Developmental Disabilities. Contact: Ken Loud, National Symposium for Information Technology, 1244 Blossom Street, 5th Floor, University of South Carolina, Columbia, South Carolina 29208; (803) 777-4435.

April 28–30, Region IV Meeting, Hyatt Regency Hotel, Memphis, Tennessee. Contact: Ronney Snell, CP, (901) 725-0600.

April 27, 28, 29, "Anatomical Design Socket and Advanced Prosthetic Techniques," Certificate CEC course. Flexible AK/BK & Narrow ML (CAT-CAM/NSNA equivalent). Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

April 30, "Graph-Lite Orthotics," Daw Industries Advanced Continuing Education Seminar, Certificate CEC course. Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

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April 30–May 1, ABC Board of Director's Meeting, Memphis, Tennessee. Contact: ABC National Office, (703) 836-7114.

May 4, 5, 6, "Anatomical Design Socket and Advanced Prosthetic Techniques," Certificate CEC course. Flexible AK/BK & Narrow ML (CAT-CAM/NSNA equivalent). Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

May 10–13, International Trade Fair and Congress for Orthopaedics and Rehabilitation Technology. Contact: NMA Nurnberg Messe- und, Ausstellungsgesellschaft mbH, Objektleitung, Messezentrum, D-8500, Nurnberg 50, West Germany.

May 13–14, Academy Continuing Education Conference 2-88 and New York State Chapter Combined Meeting, "Current Clinical and Technical Concepts in Lower Limb Prosthetics," Albany Marriott Hotel, Albany, New York. Contact: Academy National Headquarters, (703) 836-7118.

May 13–14, Charleston Bending Brace Seminar, Park Suite Hotel. Contact: Melissa Wetherell, P.O. Box 1070, Apopka, Florida 32704-1070; (800) 327-0073.

May 13–15, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting, Mission Hills Resort, Rancho Mirage, California. Contact: Tim Jacobson, CPO, (714) 686-5325.

May 16–20, "Fitting Procedures for Utah Artificial Arm and Hand System." Contact: Harold Sears, Ph.D., 95 South Elliot, #95, Chapel Hill, North Carolina 27514; (919) 968-8492, or 1-800-621-3347.

May 18, "Graph-Lite Orthotics," Daw Industries Advanced Continuing Education Seminar, Certificate CEC course. Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

May 19, 20, 21, "Anatomical Design Socket and Advanced Prosthetic Techniques," Certificate CEC course. Flexible AK/BK & Narrow ML (CAT-CAM/NSNA equivalent). Contact: Daw Industries, 5360 A Eastgate Mall, San Diego, CA 92121; 1-800-824-7192.

May 19–21, Freeman Orthotic Fitters Training Workshop, Daytona Beach, Florida. For more information, write: Freeman, Drawer J, Sturgis, Michigan, or call Cameron Brown, 800-253-2091.

May 24–26, HIBCC '88: The Health Industry Electronic Communications Conference, Hyatt Regency Chicago, 151 East Wacker Drive, Chicago, Illinois 60611. Contact: Health Industry Business Communications Council, 70 West Hubbard, Suite 202, Chicago, Illinois 60610; (312) 644-2623.

June 8–11, AOPA Regions II and III Combined Annual Meeting, Trump Plaza Hotel and Casino on the Boardwalk, Atlantic City, New Jersey.

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

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

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

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

June 27–July 1, "Fitting Procedures for Utah Artificial Arm and Hand System," College Montmorency, Dept. of P and O, Montreal, Quebec. Contact: Harold Sears, Ph.D., 95 South Elliot, #105, Chapel Hill, North Carolina 27514; (919) 968-8492, or 1-800-621-3347.

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

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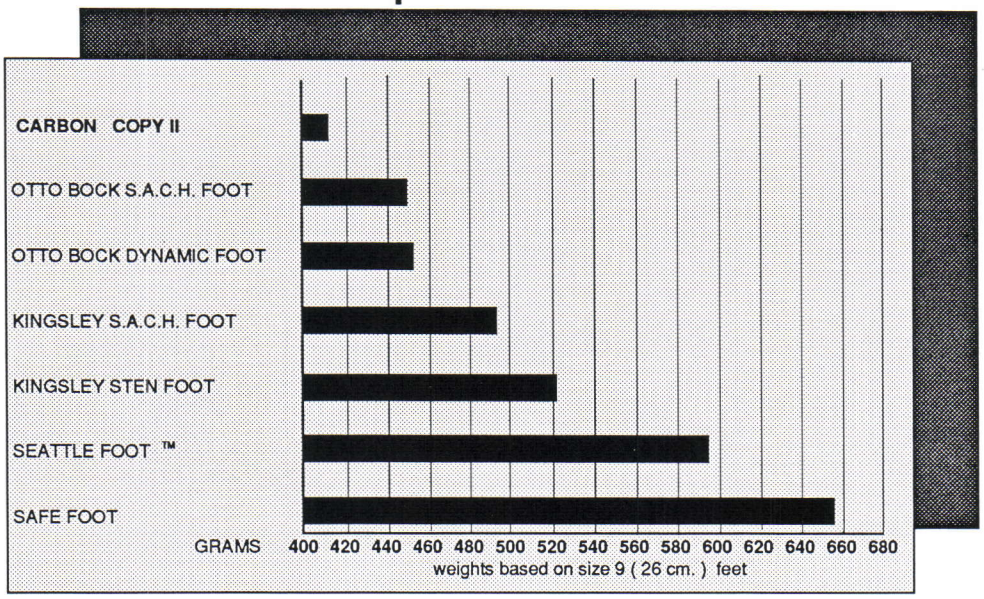

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July 16–18, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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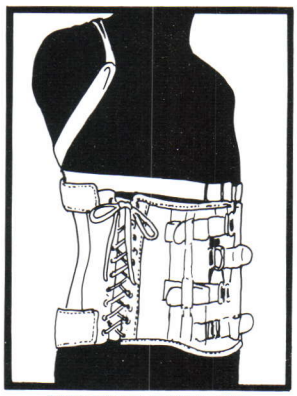

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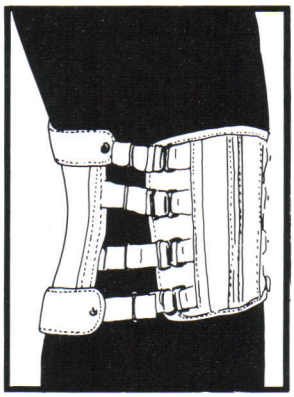

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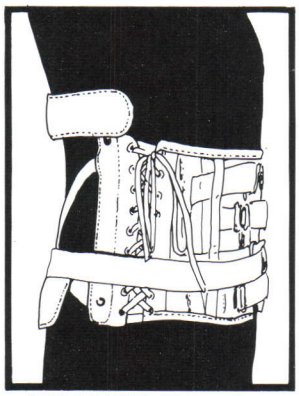

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October 15–17, AFI-ENDOLITE Prosthetic Certificate Course, "Endolite High Technology Prosthesis," Miami Lakes Inn & Country Club, Miami, Florida. Contact: Karen Hewitt, Registrar, AFI-ENDOLITE, 2480 West 82 Street, Hialeah, Florida 33016; (305) 823-8300.

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

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

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

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

1989

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

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

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

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

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

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

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

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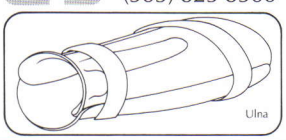

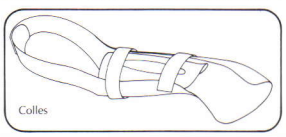

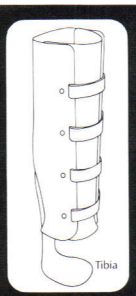

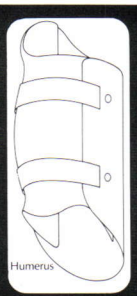

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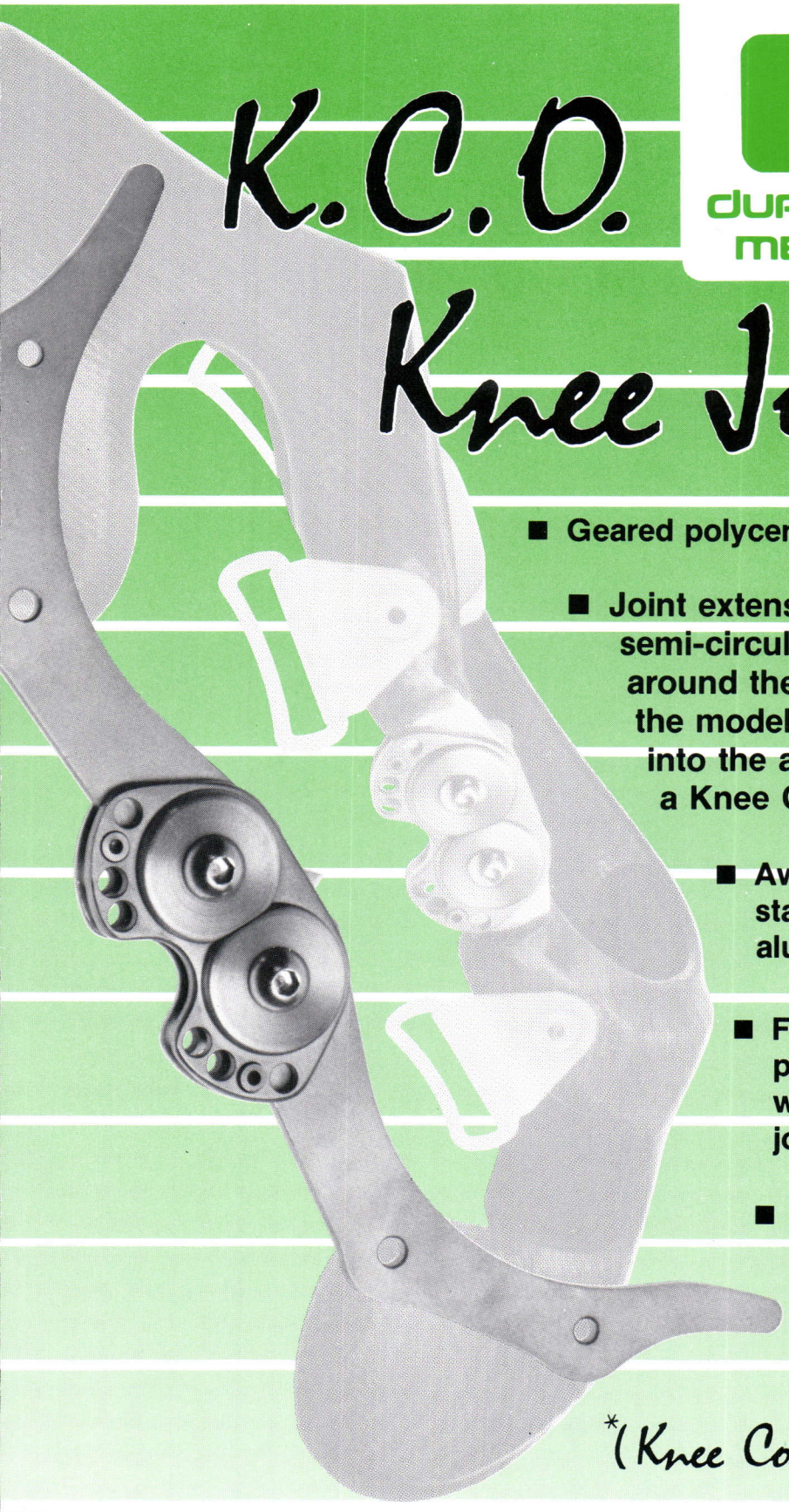

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 18–28.

Stabilizing Anterior Cruciate Ligament Injuries: Biomechanical Requirements of Orthotic Design

Stuart H. Marquette, C.O.

THE ACL IN ITS NORMAL AND INJURED STATES

The anterior cruciate ligament (ACL) is located in the intra-articular area of the tibio-femoral joint. It is attached on one end to a fossa on the posterior aspect of the medial surface of the lateral femoral condyle. This attachment is in the form of a segment of a circle, with its anterior border straight and its posterior border convex. Its long axis is angled somewhat forward from the vertical, and the posterior convexity is parallel to the posterior articular margin of the lateral femoral condyle. The other end of the ACL is attached to a fossa anterior, and lateral to the anterior tibial spine[2] (Figure 1).

The ACL itself consists of collagen and elastic fibers conformed in a band-like fashion. The band can be divided into two sections: an Anterior Medial Band (AMB) which is taut during flexion and lax near full extension, and a Posterior Lateral Band (PLB) which is taut during extension and lax in higher degrees of flexion.[2, 6, 39] The ACL's loosest position occurs between 40° and 50° of flexion, where neither the PLB nor the AMB have significant tension.[18]

The stability of a knee at any single moment is a function of four factors: ligament tightness, congruency of the articular joint surfaces, the effect of internal forces from the menisci, and the effect of muscle action.[22] For example, joint loading has been shown to significantly reduce anterior/posterior laxity—probably due to increased joint congruency and meniscal compression.[32] Cabaud describes the ACL as an essential structure on which mammalian knee joint stability depends, and as the "Keystone" of control in the complex fluid flexion and rotation of the normal knee.[6] The ACL has been shown extensively to be the primary structure controlling anterior drawer of the tibia in relation to the femur.[4, 6, 7, 11, 18, 19] In fact, values as high as 88.8% of the total resisting force to anterior tibial motion have been attributed to the ACL.[16] Internal rotation of the tibia increases tension on the ACL in all degrees of flexion.[6, 11, 18] The ACL serves to transfer force from the anterior superior tibia to the posterior distal femur. And, finally, it assists in limiting internal rotation and contributes to the rotation centers of the normal knee.

ACL Injuries

ACL injuries are quite prevalent and, as society becomes more oriented to fitness and sports activity, are becoming increasingly more common. Torg presents statistics showing that 69% of all knee injuries presented for arthroscopy for "internal derangement" of the knee involved damage

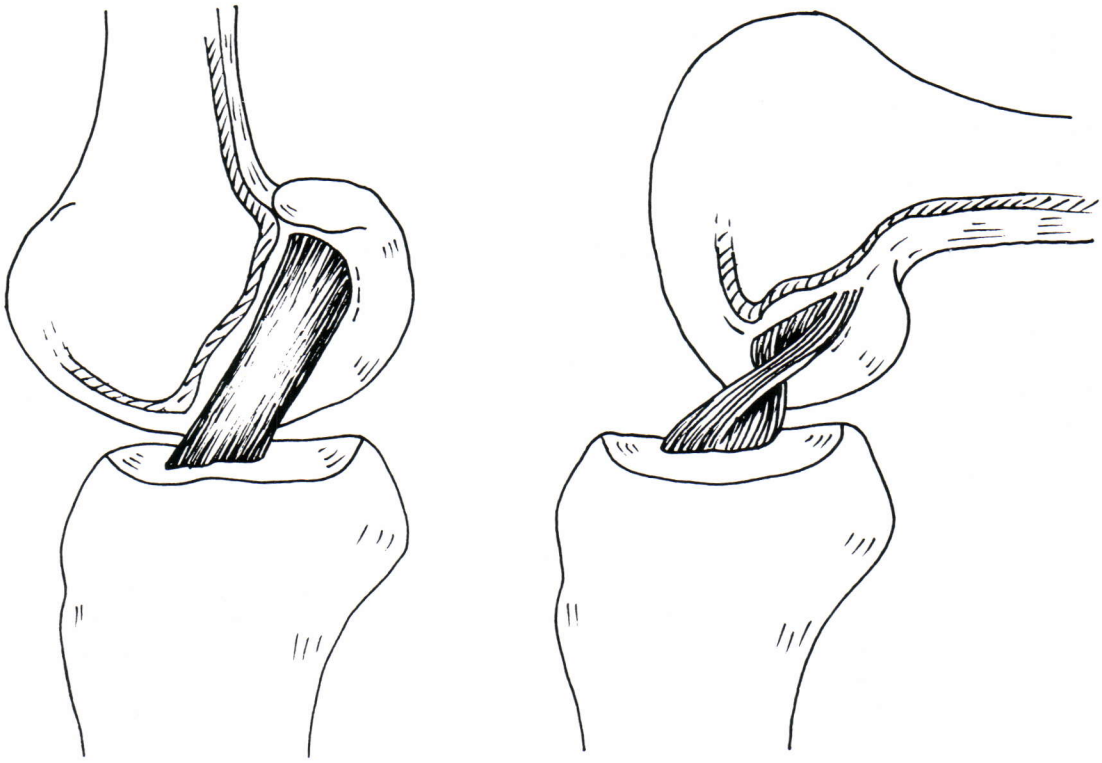

Figure 1. The anterior cruciate ligament.

to the ACL.[36] The literature describes four mechanisms capable of producing forces on the ACL larger than it can transfer, thereby severing it. The most common of the four is external rotation and abduction, the force mechanism present in most football and skiing injuries. An external rotatory moment is created by a cleated shoe in combination with an internal hip rotation. The same type of force is applied by a ski acting as a lever arm. When a large abductive force is applied to the tibia by another player or the skier continuing downhill and then firing his femoral adductors, the ACL tears.[8,18] A second common method of injury is extreme internal rotation of the tibia, such as crossing of ski tips.[6,11,22] The third injury mechanism is hyperextension to the degree where the posterior capsule and posterior cruciate have already been compromised. The last injury mechanism is a large anteriorly-directed posterior force, such as clipping in football.[18]

Anterior Drawer

Many studies have documented the difference in anterior drawer, both in vivo and in vitro, when the ACL is torn. Kochan, et al. found that at 20° of flexion with 200 N (Newtons) or about 45 lbs of anterior tibial force applied, the anterior drawer increased from 5.9 ± 2.1 mm for a normal knee to 12.2 ± 4.2 mm for an ACL-deficient knee.[19] These numbers may be affected by the limitation of internal tibial rotation upon anterior drawer caused by a locked foot plate used in these studies. Daniels, et al., using a less accurate KT-2000 and allowing for tibial rotation, obtained values for cadaver and in vivo knees at 89 N (20 lbs). Cadaver studies showed an increase from 5.8 ± 2.3 mm for the intact knee to 12.1 ± 2.9 mm when the ACL is severed (mean increase was 6.3 ± 2.0 mm, with a range of 2 to 10 mm). In vivo, this same study obtained values of

7.4 ± 1.7 mm for the normal knee and 13.0 ± 3.5 mm with the ACL absent.[7]

In a study by Fukubayashi, et al. anterior drawer increased 30% when free tibial rotation was allowed. The study explained that this rotation was internal and that the largest anterior drawer was obtained at 30° of flexion (7.0 mm at 100 N—22.5 lbs). They also demonstrated that anterior drawer of an ACL-deficient knee increased 2.5 times that of the intact knee betwen 15° and 45° of flexion at this same force level.[11] While actual anterior drawer values need to be determined using larger sample sizes and consistent force levels, the fact remains that anterior drawer of the tibia is primarily controlled by the ACL, and that these motions are extremely small in magnitude. To control anterior drawer to within normal values means to restrict movement to less than 1/5". At this point, the stiffness of the force versus displacement curve is high (83 ± 17 N/mm for normal knees and 52 ± 18 N/mm for ACL-deficient knees at 200 N of force), and increased force yields a less proportional increase in anterior drawer.[25] Allowing even 1/2" of anterior tibial motion is severely detrimental to secondary structures and normal knee motion.

Rotatory Instabilities and Pivot Shift

The most problematic and destructive occurrence to the ACL-deficient patient is not necessarily pure anterior drawer, but the resulting rotatory instabilities that are allowed to occur. These instabilities have been classified by Hughston, et al. and have received considerable recent attention in the literature.[14,15] The reasons for this attention is the realization that rotatory instabilities cause severe degenerative changes and jeopardize secondary structures, which can be further disabling to the patient. The long term degenerative effects of rotatory instabilities are not yet known, but early evidence suggests the acceleration of degenerative changes.

Rotatory instabilities are classified as four major types. Here, we will focus on the two types that are related to anterior tibial movement: Anterior Lateral Rotatory Instability (ALRI), in which the lateral aspect of the tibia subluxes anteriorly and Anterior Medial Rotatory Instability (AMRI), where the medial aspect of the tibia subluxes abnormally in the anterior direction.[14,15]

ALRI is the more debilitating of the two anterior rotational instabilities, primarily because the lateral compartment experiences greater stress during extension than does the medial compartment. The greatest stress occurs on the posterior lateral horn of the lateral meniscus.[15] Hughston, et al. continues, stating that this can be demonstrated by the fact that the posterolateral ligaments are bigger and stronger than the medial structures, possibly due to evolutionary reaction to the increased stress.[15] ALRI is brought about by two events: increased anterior drawer of the tibia and a shift of the transverse axis from its normal position to a point further medial. The literature identifies this latter event as the Pivot Shift Mechanism.

The normal location of the transverse axis (the center of tibial rotation) between 0° and 30° of flexion is slightly posterior and slightly medial to the bi-section of the tibial plateau.[37,38,39] As higher degrees of flexion occur, its location migrates slightly medially and posteriorly.[3] (Figure 2).

When the ACL is absent, the instant center of the knee moves from this position.[10] The center of rotation moves medially and posteriorly from its normal point to a point well inside the medial compartment, as shown in Figure 3.[9,22] The effect of this pivot shift is an increase in the length of the lever arm to the lateral tibial condyle. This allows production of a larger angular radius.[35] The combination of the pivot shift (increased lever arm) and increased anterior drawer allows the lateral tibial condyle to significantly sublux anteriorly. A number of studies have noted the action of the femur sliding off the posterior lateral condyle of the tibia.[12,23] This sliding causes impingement of the posterior horn of the lateral meniscus and will most likely cause damage to this structure over time.[5] It is important to understand that ALRI is a two-event situation and that

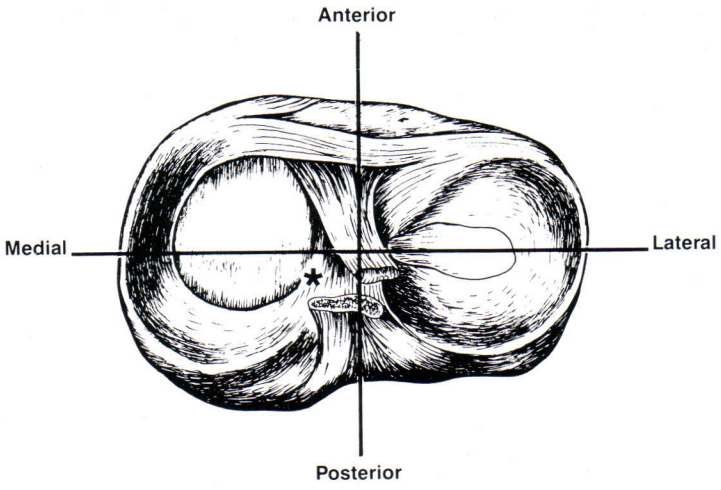

Figure 2. The tibial plateau in a normal knee, showing location of normal rotation center at 20° flexion.

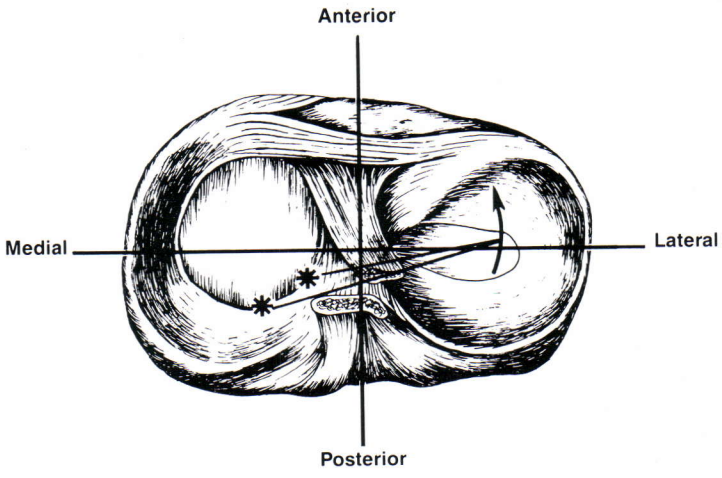

Figure 3. ALRI. Tibial plateau in an ACL injured knee, showing pivot shift. Lever arm to lateral tibial plateau increases.

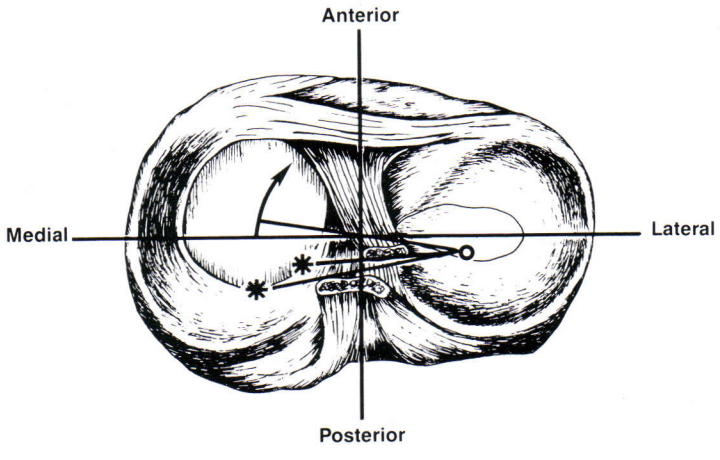

Figure 4. AMRI. The tibial plateau, showing both ACL and MCL pivot shift. Lever arm to lateral tibial plateau becomes significant.

rotatory instability is an anterior/posterior motion, and is not controlled primarily by medial/lateral stabilizers. It is reasonable to expect, then, that by limiting anterior drawer, ALRI will be reduced.

In injuries tested within two weeks of arthroscopy, the pivot shift was more sensitive in accurate diagnosis (88.8%) when compared to the Lachman and Anterior Drawer Sign Tests. In cases presented after two weeks, the Lachman and Pivot Shift tests were not found to be significantly different (84.6%).[16] The primary influencing factor here was the change in the center of rotation.

AMRI occurs primarily in the patient who has a combination of injuries which compromise the medial collateral ligament (MCL) as well as the ACL. Isolated sectioning of the MCL does not produce significant increases in anterior drawer, or in AMRI while the ACL is intact.[32,34] When the ACL and MCL have both been fully incapacitated, the pivot center of the knee shifts far laterally to the lateral compartment. The long lever arm to the medial aspect of the tibia, in combination with increased anterior drawer, allows anterior medial tibial subluxation.[33] (Figure 4).

This discussion of rotatory instabilities indicates that these harmful motions are anterior/posterior in direction and, therefore, are primarily controlled by the ACL. The ACL is a large contributor to the location of the transverse axis of rotation. Not until it is absent do the medial/lateral structures largely influence the location of this rotation center significantly.

ACL Tears

The area of the ACL that tears has been correlated in the literature to the strain rate at which the force is applied. The strength and function of the ACL under loading is determined by the structural arrangement of the fibers and the proportion of elastic to collagen fibers.[13,28] When the strain rate is slow, the ligament/bone complex is compromised first, and complete separation usually occurs in this area. The elastic modulus of canine preparation under slow strain rate was shown to be 200 megapascals.[13] At a fast strain rate, tears occur in the mid-ligament section.[26,27] Even though an ACL may appear intact arthroscopically, electron microscopy shows the micro fibrils of the collagen fibers to have been disrupted, decreasing the effectiveness of the ACL in controlling anterior tibial forces.[17,26,27]

The force required to disrupt the ACL varies widely due to differences in the size and shape of individuals. Older individuals have less ACL strength than younger adults.[27] Noyes, et al. demonstrated this, presenting maximum ACL forces of 1730 ± 660 N (388 ± 148 lbs) in a group of 16 to 26 year olds and 734± 266 N (165 ± 60 lbs) for a 48 to 86 year old group. Further, their study concluded that the ratio between ligament separation and body mass was 33 N/kg for the 16 to 26 year old group and 10 N/kg for 48 to 86 year olds.[27] Trent, et al. discussed the force per unit strain on the ACL (resting length times average stiffness of force versus deflection curves) as being 310 kg (141 lbs).[37] Resultant forces in knees range from 480 ± 35 N (107 ± 7.8 lbs) during walking, to 1020 ± 48 N (229 ± 10.7 lbs) during running, and 3280 ± 103 N (737 ± 23 lbs) in world-class long-jump athletes.[24] Forces generated during running have been presented as being 250% of the body weight.[1]

Since ligament tears do occur, it follows that the forces to which knees are sometimes subjected exceed these normal values. Complex ligamental tears imply even higher force levels. It is important to recognize that the forces to which knees are subjected are in the range of hundreds of pounds. The energy that the knee's ligaments attempt to transfer is extremely large, occasionally more than the ligaments can absorb. Even more important is the realization that the ligaments must contain motion of less than 7mm ($^3/_{10}$") while sustaining these forces. It is this problem that today's surgeons, physical therapists, and orthotists face when attempting to rehabilitate the knees that have sustained ACL tears.

Secondary Ligamental Structures

The secondary structures that assist in anterior tibial resistance should also be considered. Once the ACL has been torn, there is minimal secondary ligamental back up. When an ACL-deficient knee is required to stabilize against anterior tibial subluxation, the force must be absorbed and transferred by the secondary structures. Secondary restraints may often block clinical laxity tests, but commonly stretch out in reaction to increased stress, and do not stabilize the higher forces experienced during activity. In the ACL-deficient knee, the anterior restraining structures absorb force in the following sequence: iliotibial tract (24.8%), mid-section of the medial area of the capsule (23.3%), mid-section of the lateral area of the capsule (20.8%), medial collateral ligament (16.3%), and the lateral collateral ligament (12.4%).

The hamstrings are also positioned to help resist anteriorly-directed forces on the tibia. They are seldom flexed during the weight-bearing phase and, therefore, are unable to restrict abnormal motion.[30] External rotation of the knee increases tension in the medial structures of the knee, whereas internal rotation tightens the iliotibial tract, lateral structures, and possibly the posterior cruciate.[20] Shoemaker and Markolf showed that normal knees could produce a maximum torque of 20 ± 6.7 Newton-meters (N-m) externally and 30.9 ± 9.6 N-m internally against a fixed force plate (20° of flexion, hips extended, neutral foot position). In the same study, they showed that in vitro ligament failure occurred at 41.3 ± 10.6 N-m in external rotation.[31] This implies that even when fully flexed, the internal rotators of the knee cannot resist the forces that are applied to the knee. In summary, clearly there is no structure in the normal knee evolutionarily designed for or capable of transferring force with the efficiency of the ACL.

Secondary anterior force producing muscle activity is a very important point, since the quadriceps are very powerful and must be contracted during any bent-knee activity. Isometric quadriceps activity significantly increases the strain within the ACL relative to normal passive strain at flexion angles of 10° to 45°. The hamstrings can reduce some strain during simultaneous quadriceps activity between 0° and 30° of flexion, but not a significant amount.[30] Perry, et al. studied quadriceps stabilizing forces for flexed knees in relation to femoral head force. They reported that the quadriceps had to produce 75% femoral head force at 15° of flexion, 210% at 30°, and 410% at 60°. They explained that a 225 N (50 lb) force on the femoral head, in combination with the quadriceps stabilizing activity, produced over 600 N of tibio-femoral joint force.[29] It is important to realize that although only a portion of quadriceps force is translated into anterior tibial force, it must be considered in all treatment plans of ACL deficiencies.

The ACL in Summary

The ACL is the primary knee structure that provides significant stabilization of anterior and internal rotation movements of the tibia. The small magnitude of motion and the large force levels that must be controlled in an ACL-deficient knee present a difficult treatment problem to the rehabilitation team. ALRI results from a two-part composite event: a medial and posterior shift of the transverse axis of rotation combined with an increase in anterior drawer. Such events allow a significant subluxation of the lateral tibia and impingement of the posterior lateral menisci. ALRI is the most detrimental abnormal motion that can occur when the integrity of the ACL has been compromised. And, finally, secondary ligamental structures or muscle activity are of little benefit in resisting abnormal motion.

BIOMECHANICAL CONSIDERATIONS IN ORTHOTIC DESIGN

The ultimate orthotic design for treatment of the ACL-deficient knee would be one in which normal knee motion occurred when the orthosis was in use. It would be

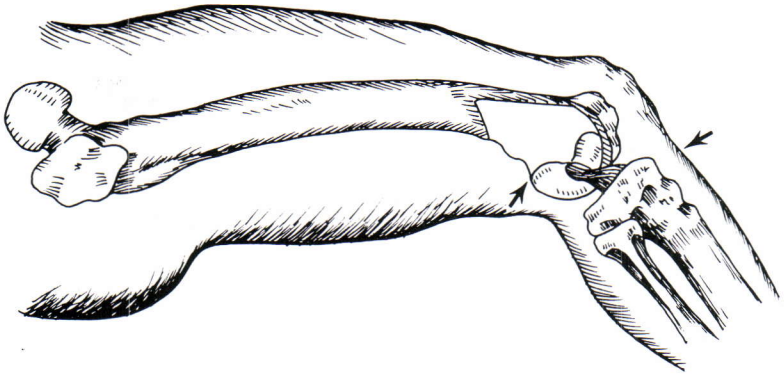

Figure 5. Ideal stabilizing forces.

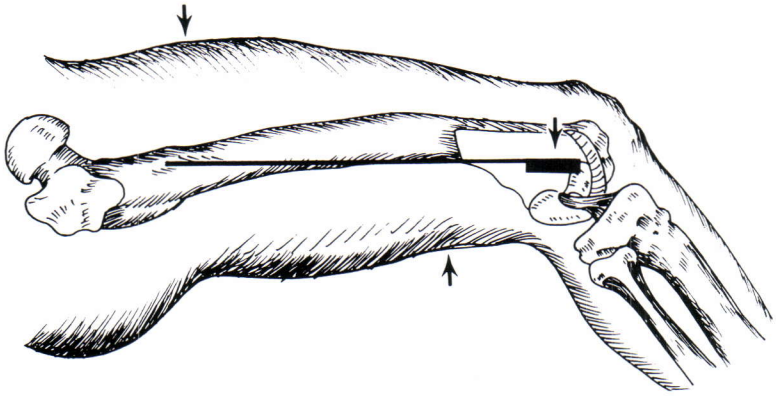

Figure 6. Two-point pressure system creating stable lever arm. Creates posteriorly directed force at joint center.

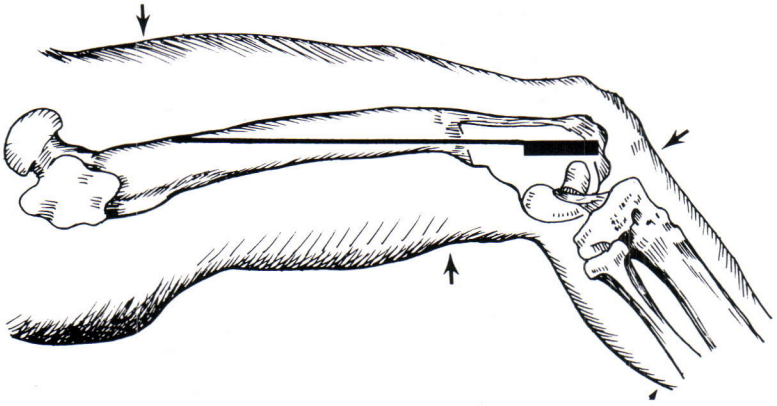

Figure 7. Four-point stabilization system required to orthoticly stabilize ACL deficient knees.

comfortable, light weight, and remain in perfect alignment at all times.

Perhaps the paramount test of an orthosis is how effectively it prevents re-injury when the patient experiences a force and situation similar to the one that originally tore the ACL. The orthosis must transfer enough force to prevent injury to the secondary structures, which become stressed in the absence of the ACL. The orthosis should generate relative motion between the tibia and femur that closely resembles normal motion. To accomplish this, the orthosis must transfer force from the anterior tibia to the distal posterior femur, replicating the primary function of the ACL. The orthosis should also resist internal rotation of the tibia on the femur, a secondary function of the ACL.

An Anatomical Joint

The ideal orthosis should incorporate in its design a joint that replicates the exact anatomical motion of the knee, that is, it rotates internally upon flexion. The foremost and most critical function of the joint is to transfer force in the anterior posterior plane, without allowing anterior motion of the tibia. If the joints are anatomical and are held in rigid relationship to the femur, the tibia will trace anatomically.

At this time, a joint of pure anatomical design and force transference does not exist. There are a number of joint designs on the market that are advertised as being "anatomical." However, most of these incorporate a slide pivot feature, which allows almost no resistance to forward motion. These designs do not restrict anterior drawer of the tibia.

An Effective Force Transfer Mechanism

Next, an ideal orthosis must stabilize the joints in relation to the femur so that their relationship does not change during activity. The main objective is to transfer force from the anterior superior tibia to the posterior distal femur. If equal and opposite forces are directed posteriorly on the anterior superior tibia, and anteriorly on the posterior distal femur, the relationship of the tibia and femur remains constant (Figure 5).

The application of pressure directly to the anterior superior tibia presents no problem, since it lies directly under the skin. However, applying force to the posterior distal femur is another matter. Pressure may not be placed directly on this area due to the large concentration of circulatory and nervous system anatomy beneath. Therefore, it becomes necessary to employ an alternate method of force transfer.

To create this stabilizing situation, a two-point pressure system must be implemented. The first pressure point is located on the femur, as far distal as comfortable knee flexion will allow. The second point is on the anterior superior thigh. These two points create a lever-arm system of stabilization with a mechanical advantage, as shown in Figure 6.

Poor anterior/posterior test results have characterized many of the orthoses presently available on the market. When the tibia moves anteriorly, the force is transferred to the joints. If the joints transfer this force effectively, they must be stabilized by the orthosis at the posterior femur. In conventional orthoses, this stabilization is effected by either an elastic or non-elastic strap. Elastic straps cannot limit motion to a few millimeters under hundreds of pounds of force. Non-elastic strapping, on the other hand, does not stretch, but instead imbeds itself into the soft tissue with a compression force equal to the original anterior tibial force. The distance the strap imbeds into the soft tissue under hundreds of pounds of force is more than the few millimeters of normal anterior motion that can be allowed. Static tests have shown that both elastic strap models, such as Lenox Hill, and non-elastic strap designs, such as C.Ti., control anterior drawer effectively only at force levels of 15 and 20 pounds, respectively. Neither, then, can effectively control anterior subluxation under dynamic situations.[3]

Other Considerations

The ideal orthosis must incorporate three other features in its design. First, it must produce a resistant force equivalent to the anterior force generated by quadriceps contraction. Second, the tibial section of the orthosis must not be allowed to leave the tibia when the leg reaches terminal swing. The orthosis has anterior inertia and does not have hamstrings to decelerate it as the human leg does.

Third, if the orthosis cannot restrict abnormal anterior tibial motion as precisely as the ACL does, it must limit extension to –10° of the full position. Limiting extension is necessary due to the increase in ligament stress and articular pressure, classically termed the "screw-home" mechanism. If the knee is abnormally rotated when approaching full extension, secondary structures such as the menisci can be pinched and damaged. Since an ideal joint design does not exist, an extension-limiting feature must be incorporated in the design of the orthosis. Once again, compression of the soft tissue on the femur using elastic and non-elastic straps creates problems. They cannot exert proper force on the posterior femur and gastrocnemius to control extension and hyperextension forces.

Knee Orthoses In Summary

The primary concern in the design of an ideal orthosis must be to limit anterior drawer to less than 10 mm of motion under hundreds of pounds of force. To accomplish this, the orthosis must incorporate joints that can effectively transfer force in all degrees of flexion and at least four pressure points (Figure 7): a two-point stabilization system to resist anterior joint motion, a point on the anterior superior tibia to transfer force to the joints, and a fourth point in the distal posterior gastrocnemius are to help decrease the orthosis' own inertial energy at terminal swing. It must also counteract the anterior tibial force generated by quadriceps contraction. Finally, the ideal orthosis must limit full extension to avoid the strong pivot shift which occurs during the "screw-home" mechanism.

IN REVIEW

The anterior cruciate ligament is the main stabilizing structure to control anterior drawer of the tibia. It also resists internal rotation of the tibia at any single degree of flexion. Its primary function by anatomical attachment is to transfer force from the anterior superior tibia to the posterior distal femur.

The difference in the amount of motion that occurs in normal and ACL-deficient knees is very small (approximately 7 mm, or $^{3}/_{10}$"). The knee can potentially experience forces of several hundred pounds.

When the ACL is compromised, rotational instabilities occur. The severity of these rotational instabilities is determined by the total amount of damage sustained by the secondary structures. The combination of the medial and posterior shift of the transverse center of rotation and increased anterior drawer produces a large angular moment of the lateral tibial condyle. This subluxation allows the femur to slide off the posterior lateral corner of the tibia and often causes impingement of the posterior horn of the lateral menisci.

The primary function of an orthosis designed for the ACL deficient patient is to replicate a normal ACL, reducing abnormal anterior motion of the tibia by transferring the force to the femur.

A minimum of four pressure points are required to stabilize the large forces involved in the ACL-deficient knee. The femoral section must be contoured to apply forces accurately in minimizing soft tissue compression and internal tibial rotation at single degrees of flexion. The femoral section must utilize a two-pressure-point design to resist anterior joint motion.

Further research must be conducted to produce the ideal orthosis—one which prevents degenerative changes and injury to secondary structures, and allows patient activity with comfort and security.

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 29–36.

The Northwestern University Knee Orthotic System Part I: The N.u.K.O. Knee Joint

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INTRODUCTION

When an orthosis is applied to the knee, it should, hypothetically allow a full, unrestricted range of motion to occur. Orthotic constraints may be introduced to provide the extra stability required to compensate for soft tissue insufficiency, thereby limiting full range of motion. For example, an orthosis applied to the knee to correct recurvatum should in no way restrict normal flexion, but rather should introduce a constraint force only near extension where the extra stability is required to stop hyperextension.

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

The N.u.K.O. knee joint offers some significant advantages over existing orthotic joints, closely mimics the motion of the natural knee, and allows design of more effective knee orthoses.[4] This report describes the proposed orthotic joint, the rationale behind its design, and its advantages.

DESCRIPTION OF THE JOINT

The joint consists of a metal, multicurvature femoral component in the shape of the sagittal profile of the distal femur, and a slotted plastic tibial component with a

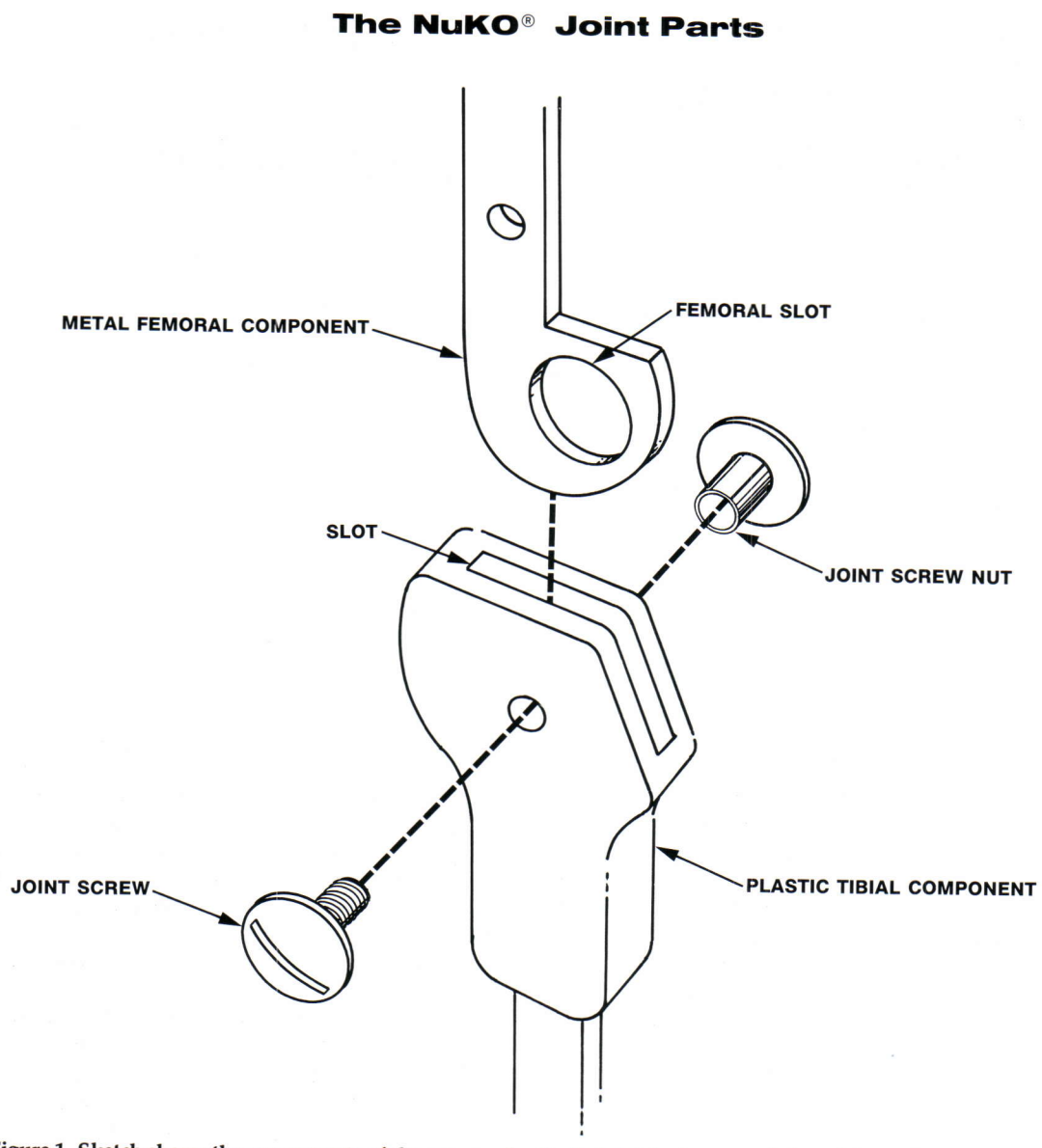

Figure 1. Sketch shows the components of the proposed orthotic joint design: metal slotted femoral component, slotted plastic tibial component with metal sidebar, and joint screw.

larger, flatter articulating surface approximating the profile of the proximal tibia (Figure 1). These incongruencies promote a rolling and sliding of the femoral component within the tibial component cup. The femoral component articulates within the tibial slot so that the surfaces become highly conforming or engaged in extension, preventing anterior-posterior motion, as in the natural joint. However, in flexion, the smaller posterior femoral curvature provides for a low degree of conformity or capture by the tibial curvature, allowing the femoral articulating surface to roll and slide anterior-posteriorly over the tibial component, thereby imitating the natural knee (Figure 2).

Stability is added to the orthotic joint through a specialized "femoral slot and joint screw" mechanism (Figure 1). The slot is formed in the metal femoral head component and captured between the walls of the tibial cup (Figure 3). The specialized mechanism simulates the action and function of the knee ligaments. The "femoral slot and joint screw" mechanism also allows the NU knee joint to tighten and become lax at different times during range of motion activity, yet allow the anterior-posterior rolling and sliding of the femoral component over the tibial component to occur (Figure 2). A computerized mathematical model was used to define "femoral slot and joint screw" placements. The data generated by the model defined the location for proper placement of this mechanism and allows the orthotic knee joint to follow the natural kinematics or motion pathways of the human knee.[4]

BIOMECHANICAL DESIGN RATIONALE

Because the surfaces of the human knee articulate without a great deal of inherent stability, the muscles and ligaments (their attachment locations and orientations) must precisely interact with the geometry of the articular surfaces to produce controlled flexion and extension motions. For example, it has been hypothesized by Lewis, et al.[3] that knee ligaments have a dual function. The "high-level" function occurs when ligaments provide stability in a traumatic situation. In this situation, the

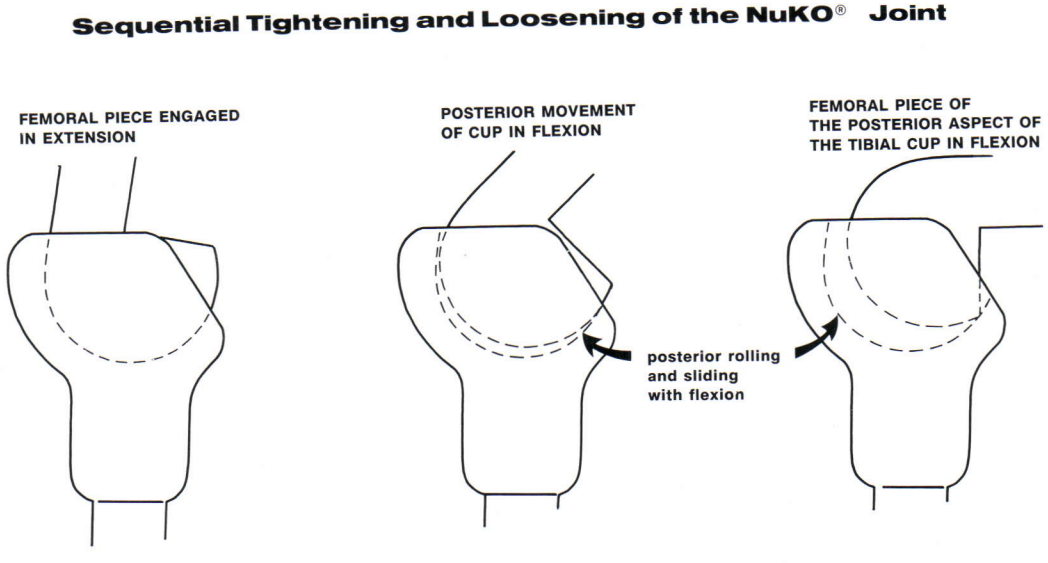

Figure 2. Sketch shows the sequential tightening and loosening of the knee joint at extension (left), 45° (middle), and 90° flexion (right).

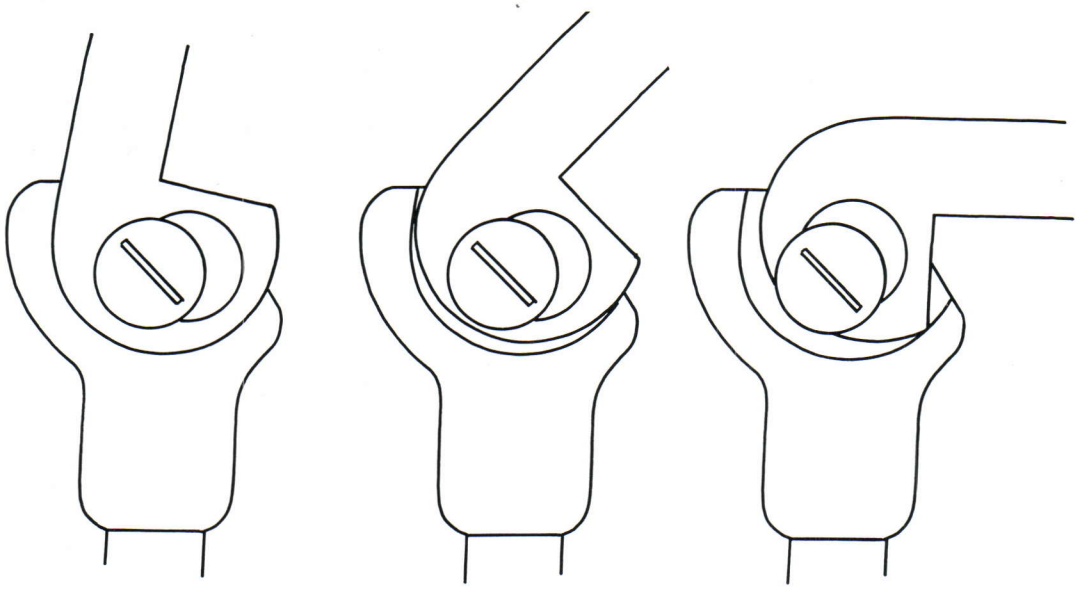

Figure 3-A. Joint at full extension.

Figure 3-B. Joint at 45°.

Figure 3-C. Joint at 90°.

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

To examine this hypothesis, Lew and Lewis[1] performed a study in which cruciate ligament forces were measured during the flexion of specimens containing a low conforming, anatomically shaped knee implant design, as well as high-conforming, non-anatomical implant design (Figure 4). In the non-anatomical implant, which did not allow rolling and sliding to accompany flexion as in the natural joint, the full range of motion was restricted to 60° of flexion, and an abnormally large constraint force was also measured in the posterior cruciate ligament.

The anatomical implant, on the other hand, allowed the rolling and sliding of the natural joint so a full range of flexion was attainable, and cruciate ligament forces approached that of a normal knee. The above findings could be extrapolated to the design of orthotic joint components. The orthotic articular surfaces should have the freedom to reorient themselves as dictated by ligaments and muscles for the full range of the natural joint motions. In this way, unwanted constraints will be minimized, and an unrestricted range of motion can be obtained.

The design of the proposed orthotic joints closely follows this biomechanical principle. The NU orthotic knee joint has articular surfaces that allow five of the six possible components of knee motion, the exception being medial-lateral displacement (Figure 5). Anterior-posterior rolling and sliding of components during flexion-extension are possible, as described earlier. Distraction of component articular surfaces is allowed, which in turn permits

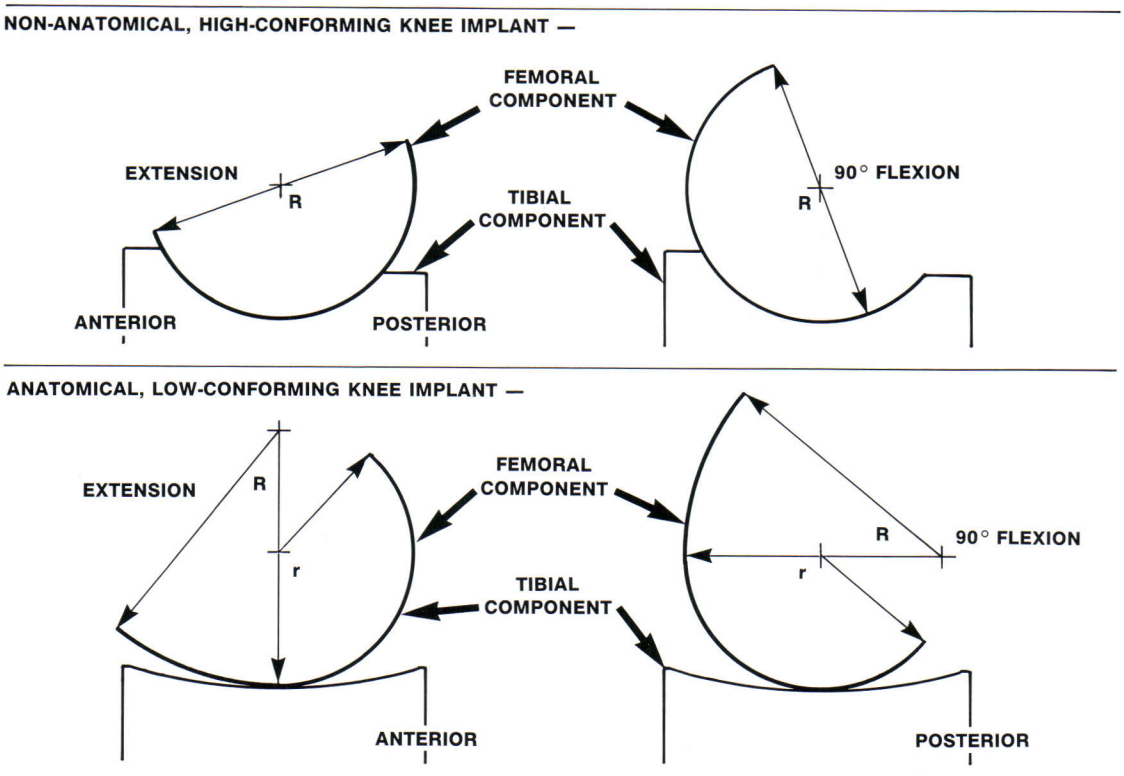

Figure 4. The proposed orthotic joint design is based upon earlier research regarding the interaction of knee ligament mechanics with internal knee prostheses. A sagittal view of the cross sections (radii of curvatures—R,r) of the tibial and femoral components of these implants are shown.

varus/valgus angulations to occur at any flexion angle. Transverse shifts are possible through the anterior/posterior and distractive displacements of the joint components. Thus, the orthotic joint articular surfaces reorient themselves as dictated by the internal knee structures to a greater degree than other commonly used orthotic joints.

Another biomechanical principle considered relates the sequential loading of knee ligaments and various bands of a specific ligament to the geometry and loading conditions of the knee through a wide variety of activities. Lewis, et al.[3] measured ligaments forces in a series of seven specimens, to correlate external joint action and ligament reaction loads. Near full extension, the anterior cruciate was found to be highly loaded during anterior-directed force or anterior drawer, varus, and internal rotation conditions. The posterior cruciate ligament was highly loaded near 90° flexion for posterior-directed forces or a posterior drawer, varus-valgus motions, and internal-external rotation. The medial collateral ligament was highly loaded during internal/external rotation and valgus force, throughout the flexion range. The lateral collateral ligament was highly loaded during varus and internal rotation and throughout the flexion range. During hyperextension, the anterior cruciate and both collateral ligaments were highly loaded. Given the previous argument for anatomically shaped orthotic joint components, the above information is important when designing constraints into

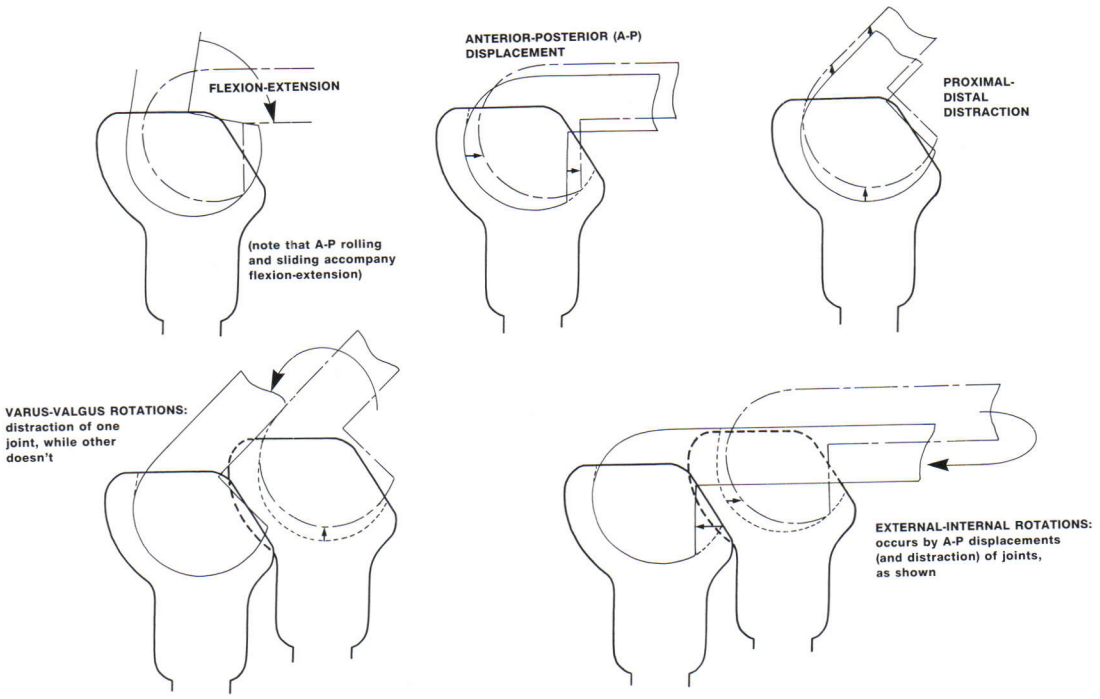

Figure 5. Sketch shows how the various components of knee joint motion can occur with the proposed orthotic joint design. Medial-lateral displacement is the only motion not allowed.

the orthotic joints or the complete orthosis to provide stability for specific ligament insufficiencies.

The orthotic "femoral slot and joint screw" mechanism is oriented and located in relation to the orthotic articular surfaces so as to function similar to the natural knee ligaments. The mechanism allows sequential tightening and loosening of the orthotic knee joint. When the orthotic joint is in extension, no anterior-posterior motion is allowed. As the knee joint begins to flex, anterior-posterior displacement is permitted along with rolling and sliding. Human ligaments also sequentially tighten and loosen about the knee joint and allow both anterior-posterior motion with rolling and sliding. The mechanism provides stability to the anatomically shaped orthotic joint surfaces, and allow the NU knee joint to work on a non-fixed asis of rotation.

MECHANICAL VERIFICATION OF THE DESIGN

The authors have previously reported a procedure for comparing the efficacy of orthotic knee joints, based upon their tendency to produce pistoning.[2] Pistoning transducers were designed, which were incorporated into the sidebars of various orthotic joint designs. As a subject wearing an evaluation orthosis performed functional activities, the transucers on the medial and lateral orthotic joint sidebars would directly measure the resulting pistoning constraint forces generated between the simplified orthotic joint motion and the complex natural joint motion. This procedure was used to compare the pistoning tendency of the proposed anatomically shaped orthotic joints with three

commonly used, commercially available orthotic joint designs: single axis hinges, posterior offset hinges, and polycentric hinges. Resultant pistoning constraint forces were measured during loaded and unloaded flexion, level walking, rising from a chair, and stair climbing activities.

The combined results over all the activities are presented in Figure 6 for each joint design. The mean and standard deviation of the combined resultant pistoning forces are given below the bar graphs, and the normalized mean forces are plotted. The data suggests that the proposed orthotic joints, because of their semi-constrained, anatomically shaped design, generated an average of 76% less pistoning constraint than the commercially available joint designs.[4] Also note that there is no statistical significance in the differences among the pistoning forces of the other three commercially available joint designs.

SUMMARY

An improved orthotic knee joint system has been designed, based upon biomechanical principles associated with knee motion and ligament mechanics. The orthotic joint articular surfaces are anatomically shaped and semiconstrained to approximate natural knee motion, particularly the anterior-posterior rolling and sliding which accompanies knee flexion and extension. Stability is added to the orthotic joint system through a "femoral slot and joint screw" configuration. The pre-

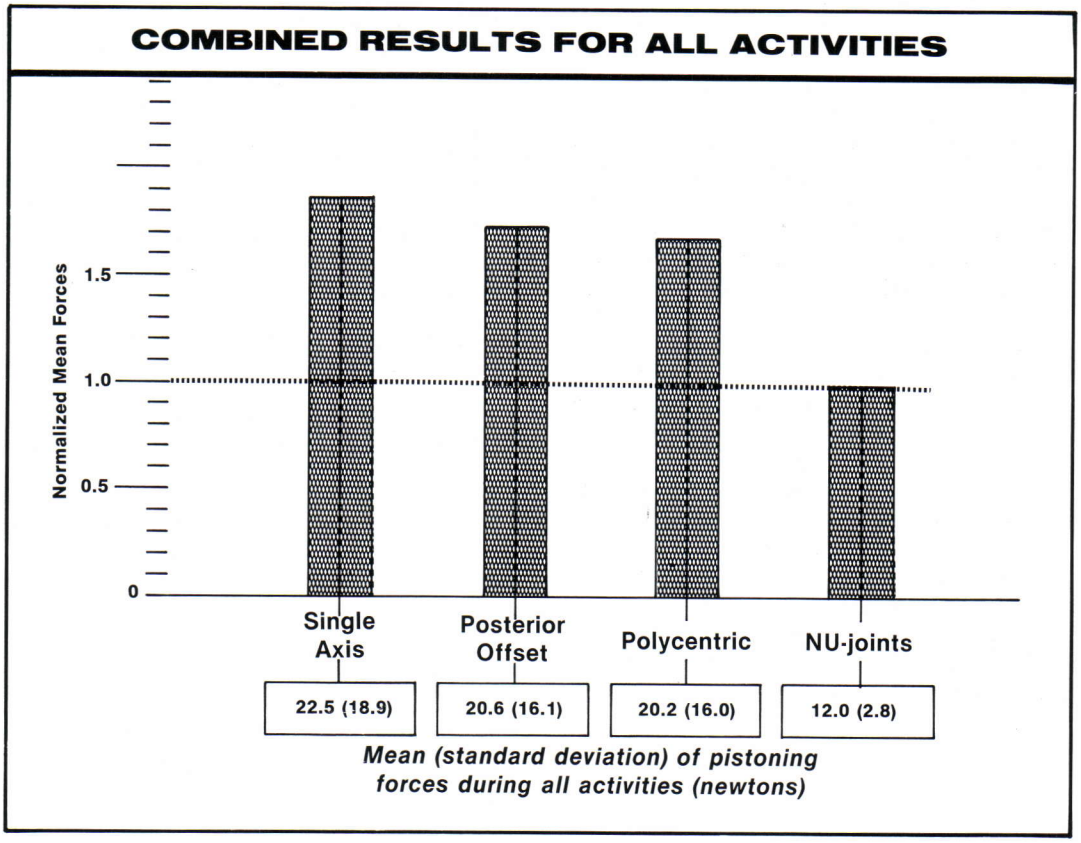

Figure 6. Sketch is a summary of the resultant pistoning constraint forces for a combination of all the test activities for each joint design. The means and standard deviations of the resultant pistoning forces are presented, as well as the normalized mean resultant pistoning forces over the four joint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

cise location and orientation of this mechanism was determined by a mathematical model.

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

The degree of suspension or fixation of a knee orthosis effects, and in most instances limits, the motion pathways allowed by the associated orthotic joints. Since the motion of most currently available orthotic joints does not match natural knee kinematics, a tightly fitted interface will magnify the pistoning constraint. This situation would be particularly harmful, for example, if an orthosis was intended to protect surgically reconstructed knee ligaments. In this case, the pistoning constraint may cause stretching of the healing tissue. On the other hand, if the interface components do not intimately secure the orthosis to the lower limb, the device would also not provide the necessary stability to the joint. Thus, improvements to the interface suspension are limited by orthotic joint kinematics. In the case of the proposed orthotic joints, it was demonstrated that the motion mismatch and resultant pistoning were reduced, thereby setting the stage so that improvements to the orthotic interface can be realized.

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ACKNOWLEDGMENTS

The authors wish to acknowledge the contributions of George T. Shybut, M.D., William Lew, and Jack L. Lewis, Ph.D. to this report.

This work was supported by Grant No. G00820024 from the National Institute of Handicapped Research, Department of Education, Washington, D.C. 20202, and from private industry.

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

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 37–50.

The Northwestern University Knee Orthotic System Part II: The Complete Orthosis

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INTRODUCTION

When an orthosis is applied to the knee, it should, hypothetically, allow a full, unrestricted range of motion, except at limits where orthotic constraints are intentionally introduced. This ideal situation is limited by the type of orthotic joints incorporated into knee orthoses. When using orthotic knee joints which are unable to follow the motion pathways of the natural joint, a tighter fitting interface will magnify the pistoning constraint. This is due to the motion mismatch between the orthotic and natural joint and will cause patient discomfort, motion restriction, and misalignment of the orthosis.[2]

A new orthotic knee joint system was described (as Part I of this report), which decreased the pistoning effect by allowing the orthotic joints to more closely imitate the natural knee kinematics.[2] These semi-constrained, anatomically-shaped joints allow improvements in orthotic suspension, since a tighter fitting orthosis with these joints will not increase the pistoning constraint.

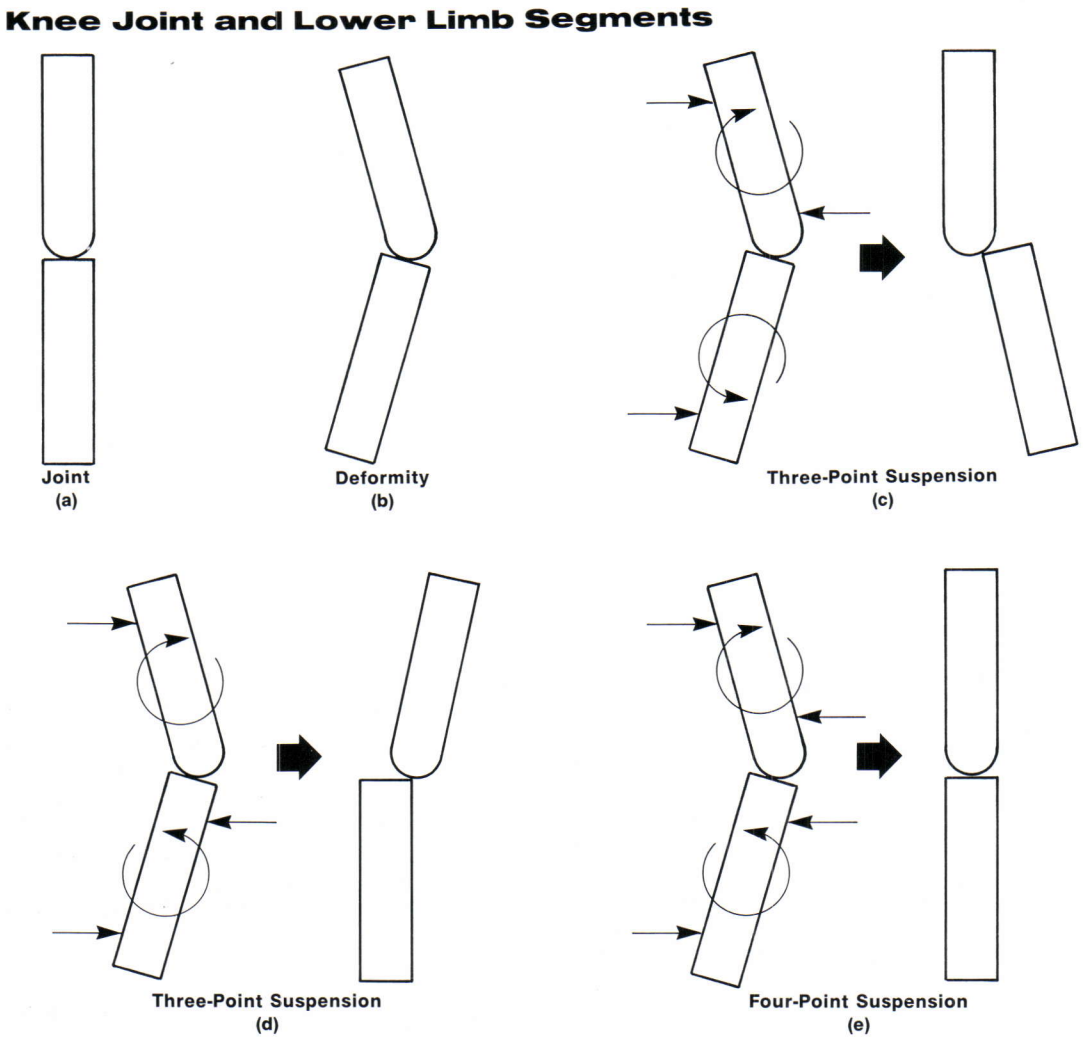
A second requirement for a knee orthosis is that the orthotic interface should be designed to complement the function of the orthotic joints. This interface should be capable of being modified to handle particular knee instability problems.

This report contains a description of a knee orthosis incorporating an improved orthotic joint design, its biomechanical rationale, significant features of the fabrication and fitting process, and a description of several case studies.

BIOMECHANICS OF KNEE ORTHOSIS SUSPENSION

A basic feature of the proposed orthosis is the use of a "four-point" suspension principle rather than "three-point" fixation, as is commonly practiced in orthotics. Three-point support is suitable for stabilizing a joint when ligamentous integrity exists as a constraint across the joint. Three-point suspension is inadequate with a ligamentous deficit, which is frequently the indication requiring the application of a knee orthosis.

The limitation of the three-point fixation system in the unstable knee is demonstrated in Figures 1-A through 1-D. When



Figures 1-A–1-E. Sketches representing the knee joint and lower limb segments: (A) with a deformity; (B) showing the effect of a "three-point"; (C, D) a "four-point"; (E) orthotic suspension.

suspension forces are applied to a joint (Figure 1-A) with a deformity (Figure 1-B), the functional forces, and the moments they create, will tend to bend and shear the joint. With three-point support (Figures 1-C and 1-D), one segment of the limb can be held in place at any one instant; however, only one force remains to support the second bony segment. Even though the point of application of this force may remain fixed, the limb segment can rotate about the single support point, causing a shearing or displacement motion at the joint, which is the very motion that is to be prevented. By contrast, as shown in Figure 1-E, a four-point fixation system will allow two suspension points on each limb segment, thus controlling and preventing motion of both segments.

The orthotic interface components and

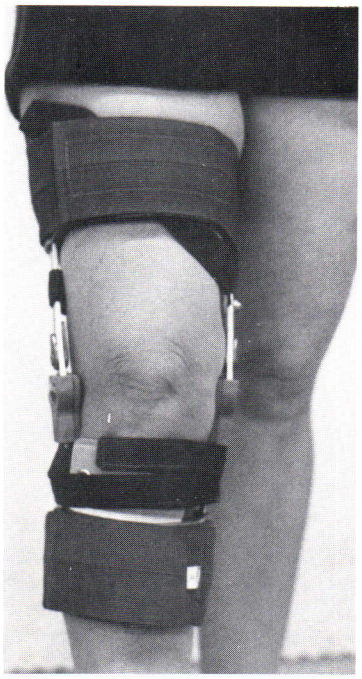


Figure 2-A. Anterior view of a representative NuKO derotational knee orthosis.

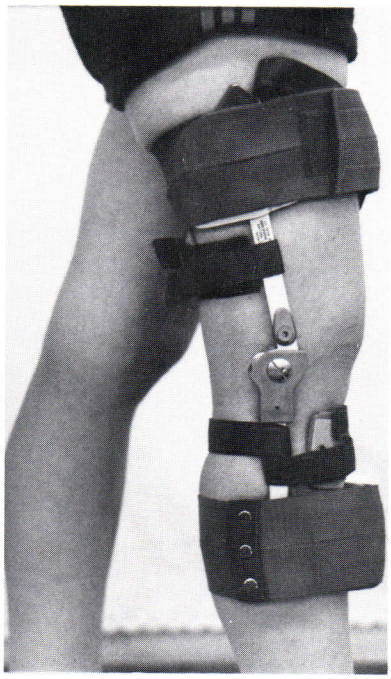


Figure 2-B. Lateral view.

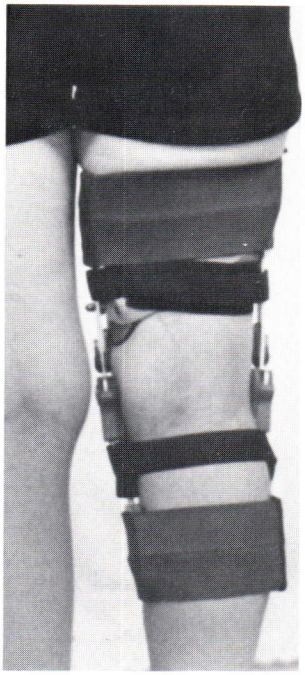


Figure 2-C. Posterior view.

strapping arrangement should be such that, given an instability direction (such as varus, valgus, anterior or posterior drawer, or rotation), they are capable of being altered to apply the four forces and resulting moments necessary to provide stability or correct a deformity, while still controlling motion at the joint.

SIGNIFICANT FABRICATION FEATURES

If a standard plaster negative impression is taken on the patient's lower limb, then as the plaster sets, apply constant pressure to the medial femoral supracondylar region. This is done so that the impression retains an accurate description of the individual's anatomy in this area. An accurate impression of the medial tibial flair region should also be obtained. A positive plaster impression is then made and modified. Emphasis is given to the parallel buildups on both sides of the knee, to ensure that

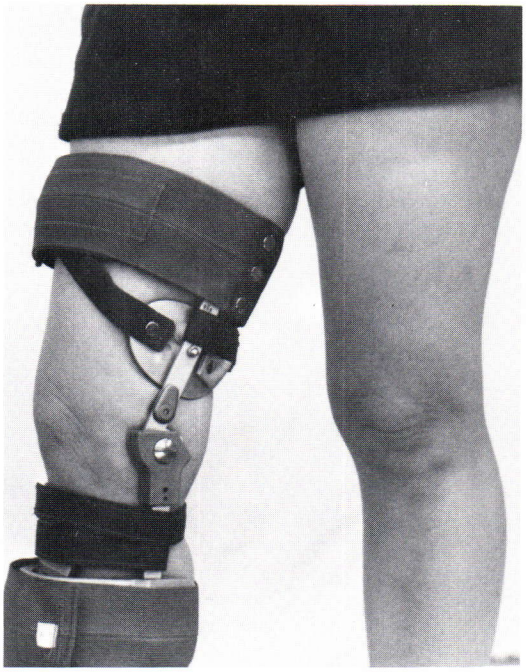


Figure 2-D. Medial view.

orthotic joints are parallel to each other and perpendicular to the joint space. Sidebars containing the orthotic joint designs are contoured to the positive plaster impression, so that the joint space of each orthotic joint is located at the level of the natural joint space, midway in the anterior-posterior plane of the knee. In the completed orthosis, the orthotic joints are positioned as closely as possible to the natural knee joint. Proximal and distal interface components are fabricated by vacuum-forming a thermoplastic material over the positive plaster impression. In this process, the orthotic joint sidebars are mechanically thermobonded to the interface, which is itself composed of two layers of thermoplastic thermobond. Using the medial femoral supracondylar depression on the positive plaster impression, an adjustable swivel medial femoral suspension pad is fashioned and is securely attached to the proximal medial joint sidebar.

Figures 2-A through 2-D present the four views of a completed derotational orthosis with an NU orthotic joint. The joint sidebars are attached to the proximal and distal interface components, which are in turn circumferentially suspended in the thigh and calf regions by broad straps. To insure adequate fixation of the orthosis, the interface components are accompanied by a swivel medial femoral suspension pad and a proximal tibial suspension pad, each with its own associated strapping arrangement.

COMMON FITTING MODIFICATIONS

Several modifications of the previously described interface components can be easily made at the time of fitting. The pressure from the medial femoral suspension pad can be increased or decreased depending upon the individual's musculature in this region. The copolymer thermoplastic of the proximal and distal interface components can be easily heated and flaired away from problem pressure areas. The plastic can also be conveniently ground away for comfort considerations. However, the fit of the knee orthosis should remain extremely intimate, as the system was designed to be worn directly against the skin. Given the fact that the improved orthotic joints minimize the pistoning constraint, tightly fitting interface components insure a functionally efficient and reliable knee orthosis system. At the same time, they provide for patient comfort and a cosmetically acceptable device.

The NuKO rehabilitative knee orthosis is intended to provide stability during the period between the plaster fracture orthosis and a definitive knee orthosis. As shown in Figure 10, the NuKO rehabilitative knee orthosis is fabricated with flexible linear low density polyethylene, in a number of sizes, and is more flexible than the NuKO derotational orthosis. This type of orthosis is used in situations (post-surgical or post-injury) in which the lower limb musculature has atrophied, and when the patient will subsequently undergo physical therapy. During this time, the lower limb musculature will increase in volume with therapy, thus making it impractical to fabricate several different "definitive" orthoses during this relatively brief period. Instead, the NuKO rehabilitative orthosis is easily, inexpensively, and reliably applied throughout the period of muscle volume changes. The patient is finally fit with the definitive orthosis only when the lower limb has stabilized in volume and shape.

INTERFACE SUSPENSION IMPROVEMENTS IN THE NU ORTHOTIC SYSTEM

A common problem which occurs with knee orthoses, is distal slippage during function. To provide additional resistance against slipping, the adjustable suspension pad in the medial femoral supracondylar region is incorporated (Figure 3-A). This particular pad is made from a copolymer plastic, an adjustable screw, and foam. The amount of pressure from the medial femoral pad can be varied by turning the adjustment screw clockwise for more pressure or counterclockwise for less

Figure 3-A. Inside view of the medial femoral suspension pad. Note the thin layer of padding covering the floating pad.

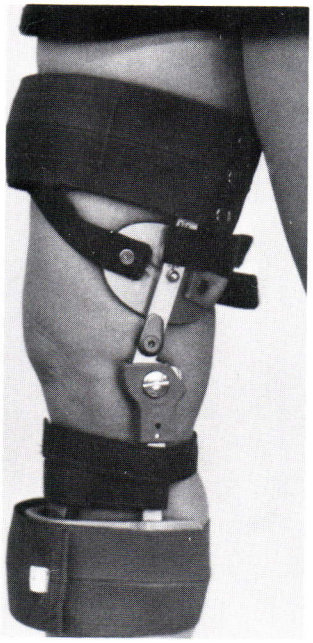


Figure 3-B. Outside view of the medial femoral suspension pad and associated straps. Note the floating pad attaches to the medial sidebar and the straps to the lateral thigh interface.

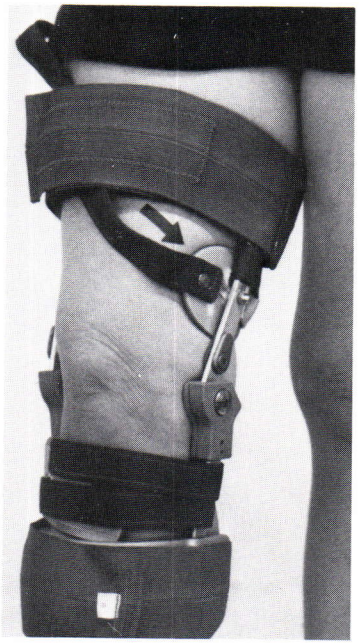


Figure 3-C. Medial femoral pad engages the medial femoral supracondylar region.

pressure depending upon the clinical situation. For example, given a patient with associated muscle atrophy, weight or volume loss, or muscle hypertrophy, the depth of the pad can be adjusted accordingly.

Figures 3-B and 3-C show the medial femoral suspension pad securely attached to the inner surface of the medial orthotic joint sidebar. To insure that the medial femoral pad is securely placed against the femur, two straps, of which the origin is on the outer surface of the medial femoral pad, encircle the thigh anteriorly and posteriorly reattach laterally over the broad plastic of the proximal interface (Figures 3-D and 3-E). Tightening this strap pulls the thigh (medial femoral supracondylar region) against the medial femoral suspension pad (Figure 3-E). The forces generated by this pad and strap assure fixation of the orthosis to the patient leg.

Proximal and Distal Interface Components

Proximal and distal interface components have been designed based on the following criteria. They are:

- rigid and strong enough to withstand repeated functional loads, or correct and hold a deformity;
- lightweight;
- unobtrusive and cosmetically acceptable;
- comfortable; and
- can be modified to generate different combinations of four-point suspension forces.

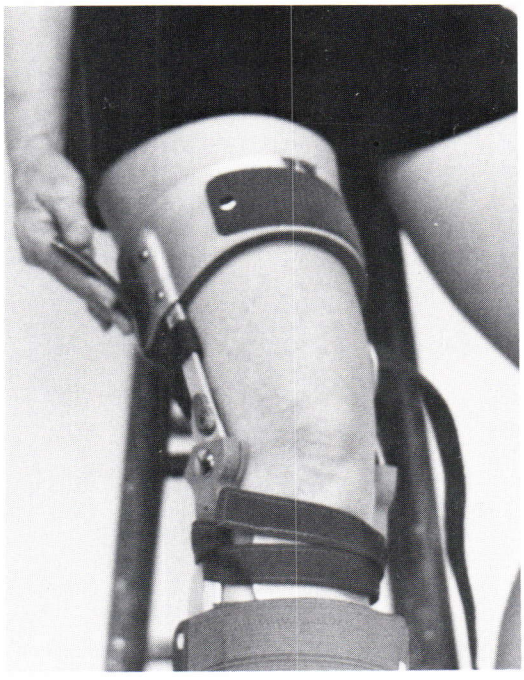


Figure 3-D. Strapping for the medial femoral pad begins with the posterior gray strap attaching to the lateral thigh band, pulling the pad against the medial femoral condyle.

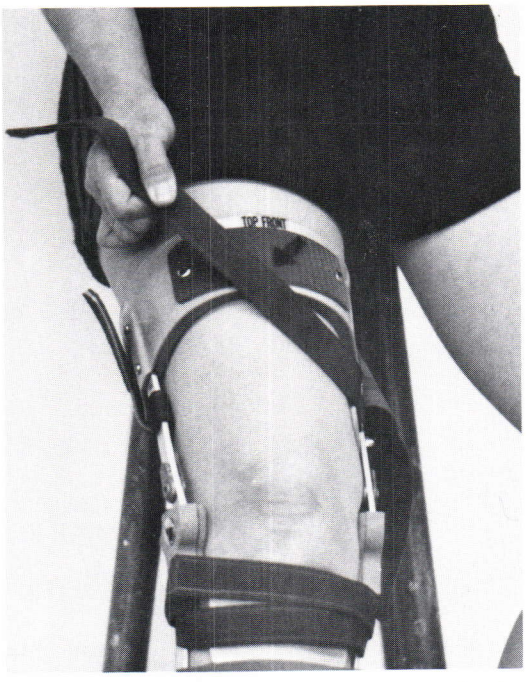


Figure 3-E. Completed strapping arrangement for the medial femoral suspension pad.

The proximal and distal interface components of the NuKO derotational knee orthosis are manufactured from a copolymer, with a composition of 10% polyethylene and 90% polypropylene. This material is also used in the automotive industry for such applications as fenders and rocker panels. For the orthosis, this material is lightweight and has high impact resistance. A completed orthosis using this copolymer weighs approximately one pound.

The NuKO derotational knee orthosis is a posterior opening design (Figures 2-A through 2-D). The proximal and distal interface components are constructed by vacuum-forming two layers of thermoplastic over the positive plaster impression, providing a rigid interface along with a method of mechanically attaching the orthotic sidebars (Figure 2-D). The interface components are suspended circumferentially in the thigh and calf regions by broad straps composed of Micro-splint™. Micro-splint™ is a semi-elastic breatheable material that offers maximal fixation of the orthosis to the patient's leg. The straps originate just posterior to the thermobonded sidebars, encircle the limb, and attach again via Velcro® strips on the anterior surfaces of the proximal and distal components. The plastic and strapping arrangement provides a secure rigid interface. Yet, because of the strapping, the components can accommodate the volume changes of the lower limb musculature during activities.

Anterior Tibial Suspension Pad

The anterior tibial suspension pad is fabricated from a linear low density polyethylene to obtain rigid fixation over the crest of the tibia. This feature provides the

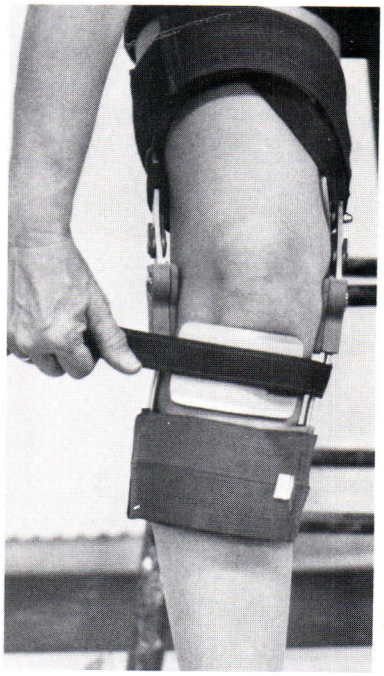


Figure 4-A. The anterior tibial plate is centered and wrapped around the medial sidebar and back onto its contact closure.

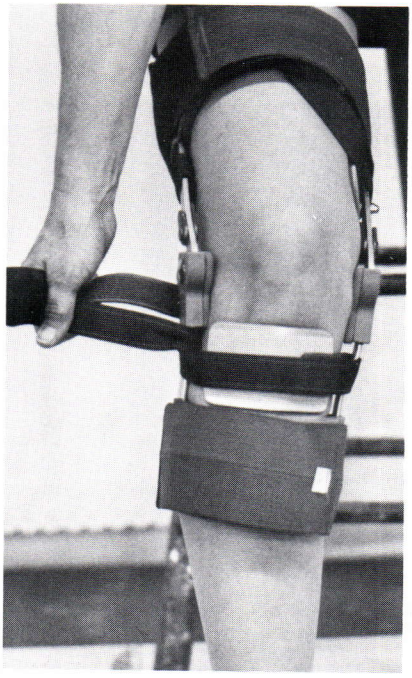


Figure 4-B. The tibial strap passes laterally around the lateral sidebar and continues posteriorly around the calf.

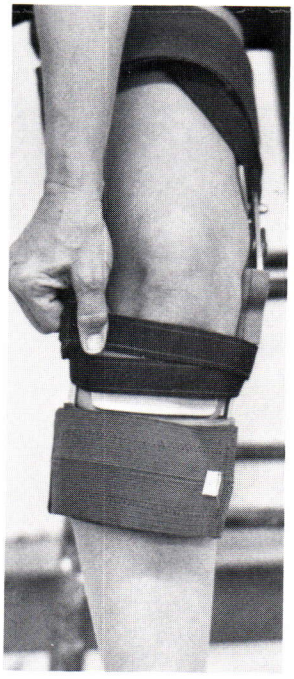


Figure 4-wc. The tibial strap continues anteriorly under the medial sidebar and onto the tibial plate's contact closure.

derotational component of the NuKO knee orthosis. With this anterior tibial suspension pad, strapping arrangements for anterior-lateral, anterior-medial, and posterior rotary instabilities can be protected.

Strapping sequence for anterior-lateral rotary instability is as follows:

1. Wrap the tibial plate strap around the medial sidebar with the tibial plate centered on the tibia (the tibial plate may be cold formed by hand to fit the tibia) (Figure 4-A).
2. Bring the strap across the top of the tibial plate securing its position on the Velcro® contact area (Figure 4-A).
3. Wrap the strap around the lateral sidebar (coming off distally) away from the popitial area and continue posterior around the calf (Figure 4-B).
4. Pass the strap under the medial sidebar and continue anteriorly across the lower portion of the tibial plate Velcro® contact area. Tightness is crucial here for derotational control. This strapping mechanism should be as tight as possible based on patient comfort and pressure tolerance (Figure 4-C).

The proposed force system generated by the proximal tibial pad and strap arrangement is shown on the cross sectional sketches in 5-A and 5-B.

In the strapping arrangement just described, tightening the strap would force the tibia posteriorly and promote external rotation (Figure 5-A). This arrangement would be used to restrain a knee with an antero-lateral rotary instability (anterior cruciate plus lateral collateral and/or lateral capsular insufficiency). Note that in the case of antero-medial rotary instabilities, the anterior tibial pad strap is applied in the same way as described above, the only difference being the pad is initially secured

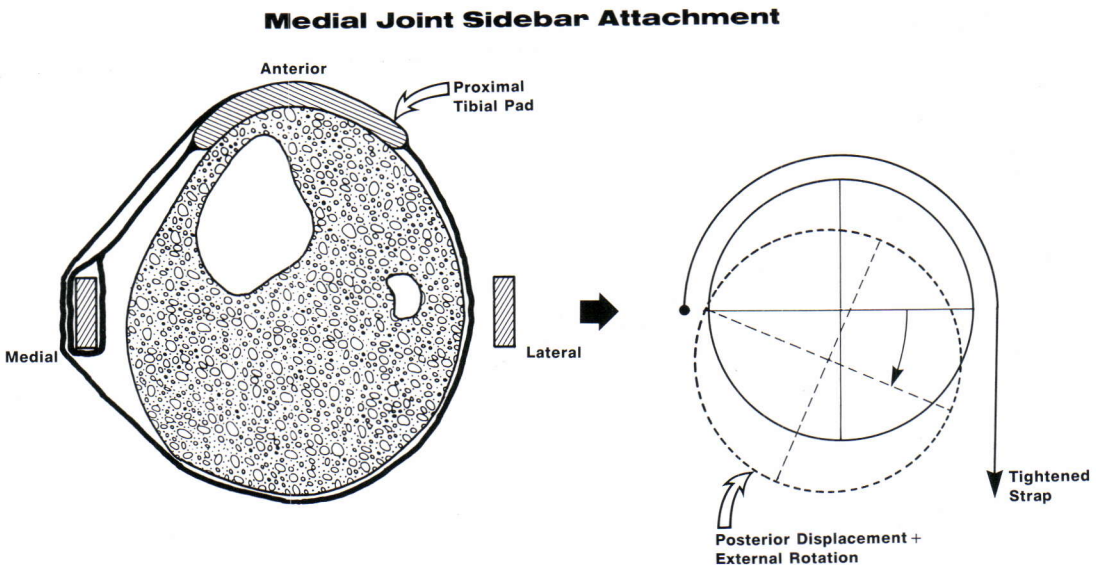


Figure 5-A. If the proximal tibial pad is attached to the medial orthotic joint sidebar, tightening the upper strap displaces the tibia posteriorly and rotates it externally.

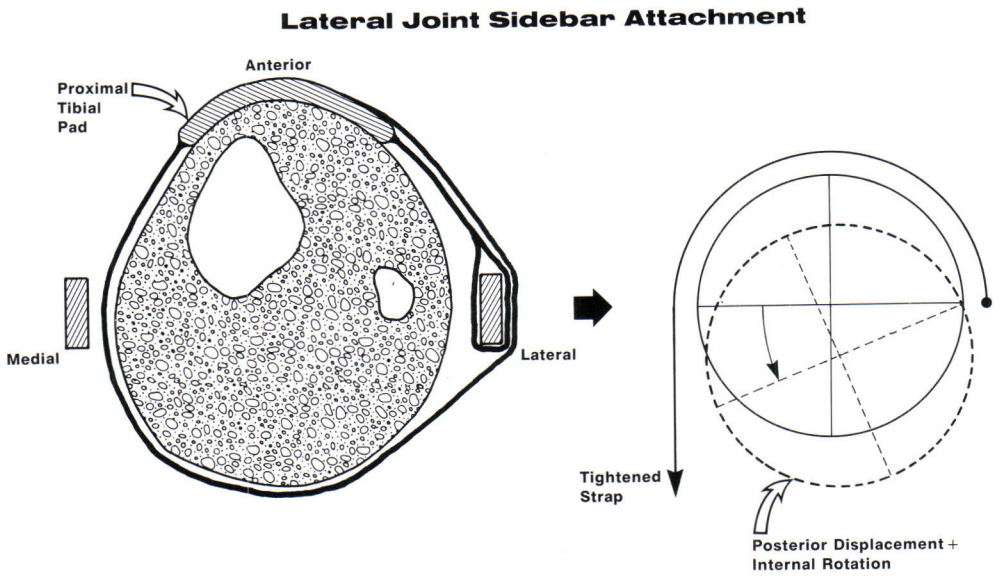


Figure 5-B. If the proximal tibial pad is attached to the lateral joint sidebar, tightening the upper strap which encircles the calf itself displaces the tibia posteriorly and rotates it internally.

to the lateral rather than the medial joint sidebar (Figure 5-B).

The proximal tibial pad also interacts with the other suspension components to provide knee stability. When an orthosis is placed on a knee with the objective of reducing anterior displacement of the tibia, the proximal tibia pad is placed anteriorly (Figures 4-A through 4-E). The four-point suspension forces generated by the orthosis are shown by the sketch in Figure 6-A. When combined with the anteriorly-directed force of the distal interface component, the posteriorly-directed force of the proximal tibial pad creates a moment which forces the tibia posteriorly, as well as straightens the tibia. This prevents it from pivoting about the distal interface component. An anteriorly-directed force of the proximal interface components combines with the posteriorly-directed force of the medial femoral suspension pad (with its strap encircling the thigh anteriorly, thereby forcing the thigh posteriorly) to create a moment controlling motion proximal to the joint. Thus, the above four forces limit the anterior displacement of the tibia and help control the motion at the joint.

If the orthotic objective was to control posterior subluxation at the knee, the proximal tibial pad can be placed posteriorly as demonstrated in Figure 6-B. The sketch shows that the anteriorly-directed force of the proximal tibial pad combines with the posteriorly-directed force of the distal interface component to create a moment which straightens the tibia and forces it anteriorly. The posteriorly-directed force of the proximal interface component combines with the medial femoral pad's anteriorly-directed force (obtained by this strap encircling the thigh posteriorly), controlling the motion proximal to the joint. Thus, the above four forces limit the posterior displacement of the tibia and help control the motion at the joint.

CASE STUDIES

Case #1

A 22 year old female collegiate basketball player sustained an acute injury to the anterior cruciate and medial collateral ligaments of her left knee (Figure 7). She underwent arthrotomy which included a medial collateral ligament repair, a pes anserinus transfer, and a medial menisectomy. She was evaluated 9 months post-surgery for knee instability symptoms. Her affected knee exhibited anterior laxity, an antero-medial rotatory instability, and a valgus deformity, but had a negative pivot shift test. The orthotic goal in this case was to stabilize the chronically unstable knee resulting from her injury.

The patient was fit with an NuKO derotational orthosis. In the sagittal plane, the interface components and strapping resisted the anterior displacement of the tibia by generating the four-point suspension forces (Figure 6-A). Since an antero-medial rotatory instability was present, the anterior tibial suspension pad was attached to the lateral orthotic sidebar, similar to that shown in Figure 5-A. Tightening the anterior tibial pad strap pre-positioned the tibia in internal rotation, while limiting the antero-medial instability. After being fit with the orthosis, the patient was able to resume vigorous athletic activity, including basketball while wearing the orthosis.

Case #2

A 20 year old male, who played basketball for a local university, injured his right knee when he went up for a rebound, and came down off-balance while simultaneously being hit by another player. Examination revealed an antero-lateral rotatory instability, with a possible anterior cruciate injury. The patient was initially treated with two weeks in a knee immobilizer and then sent back to full pre-injury activity.

Subsequently the patient reinjured the same knee, this time sustaining a partial tear of the medial collateral and anterior cruciate ligaments. He was put in a long leg cast for two weeks, then was fit with a NuKO rehabilitative knee orthosis (Figure 10). Following rehabilitation, the patient was fit with the NuKO derotational knee orthosis shown in the sequence of pictures in Figures 8-A–C. The construction of this orthosis is identical to the first case, except for anterior-lateral instability strapping.

Four-Point Suspension Forces with Anterior Tibial Pad Placed Posteriorly

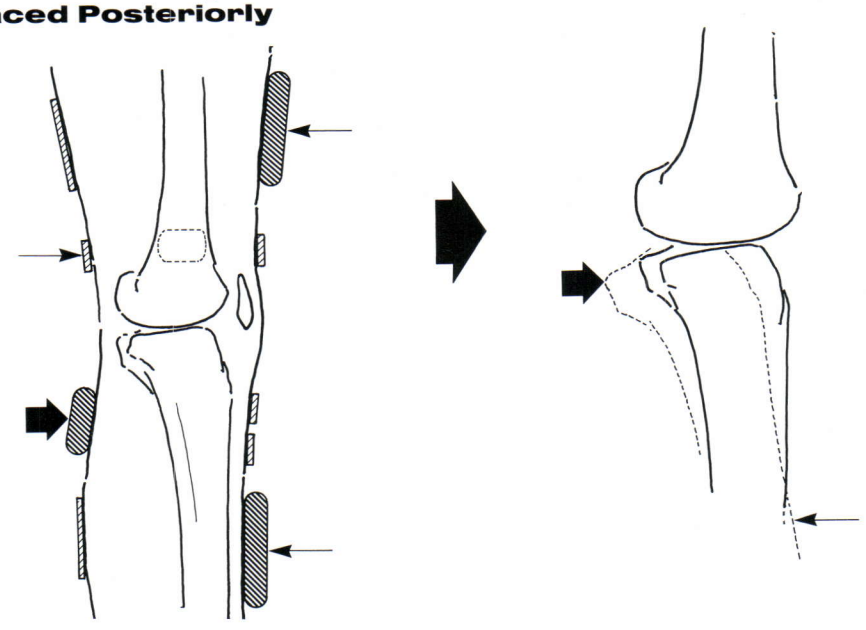


Figure 6-A. Four-point suspension forces generated by the orthosis to control anterior subluxation of the tibia. Note the anterior position and subsequent posteriorly-directed force of the proximal tibial pad.

Four-Point Suspension Forces with Anterior Tibial Pad Placed Anteriorly

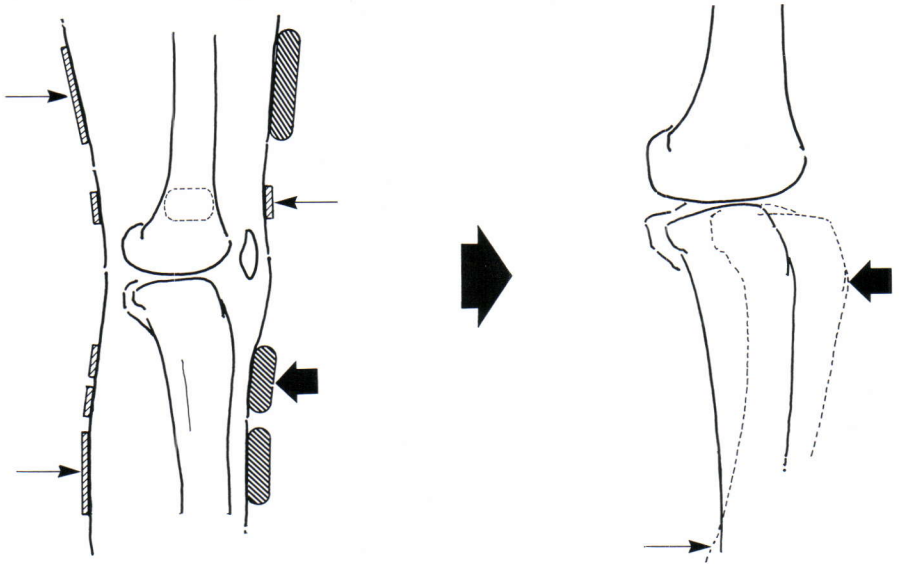


Figure 6-B. Four-point suspension forces generated by the orthosis to control posterior subluxation of the tibia. Note the posterior position and anteriorly-directed force of the proximal tibial pad.

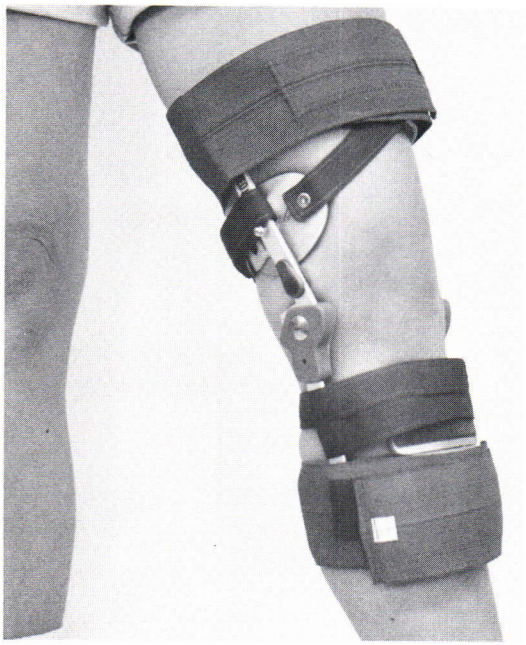


Figure 7. An orthosis for the patient in Case Study #1, providing correction for her antero-medial rotatory instability.

Case #3

A 15 year old female sustained a torn left posterior cruciate ligament from a "dashboard injury" during an automobile accident. The torn posterior cruciate was surgically repaired by a modified Jones procedure.

The patient also developed a large pressure sore on her mid-posterior calf from a postoperative cast. The orthotic objective in this case was to prevent knee motions and loads which would disrupt the posterior cruciate repair, while still allowing the patient to undergo a physical therapy program.

The patient was initially placed in an NuKO rehabilitative knee orthosis with the posterior cruciate instability strapping set up (Figure 9). The NuKO rehabilitative knee orthosis allowed the patient to complete a physical therapy program, rebuilding her lower limb musculature, and provided stability during normal walking. The orthosis was initially fit so as not to impinge upon the pressure sore region, allowing it to heal rapidly.

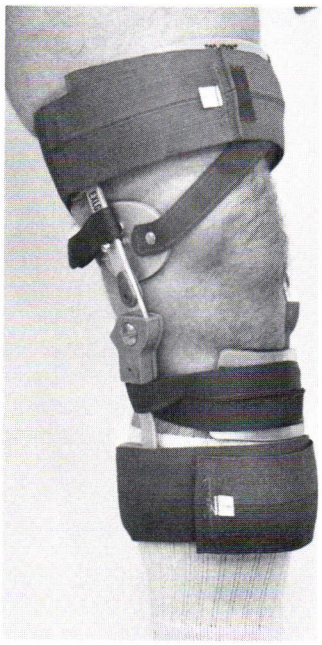


Figure 8-A. Anterior view for anterior and medial instability.

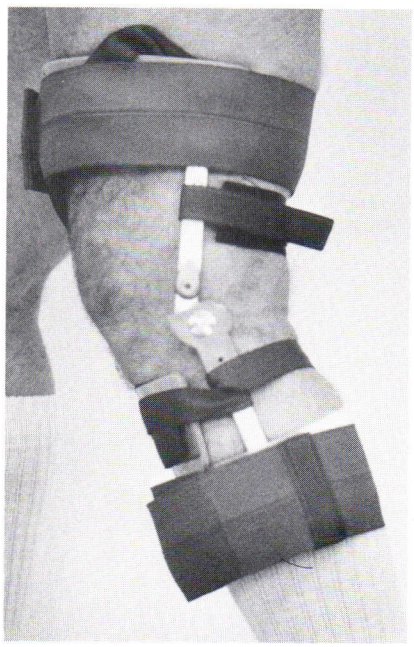


Figure 8-B. Lateral view.

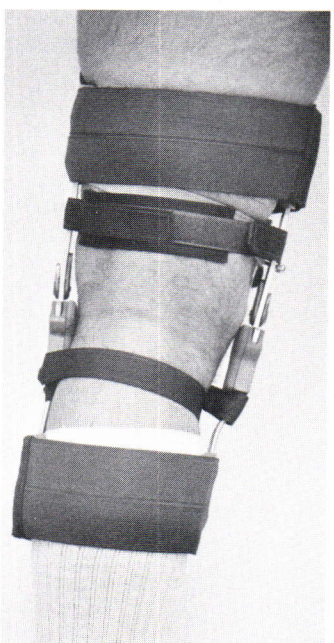


Figure 8-C. Posterior view.

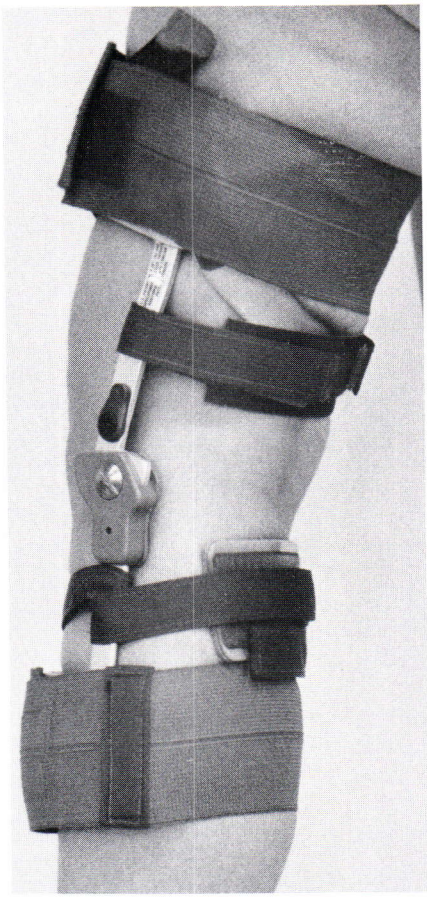


Figure 9. Posterior cruciate design for the patient in Case Study #3. Note the posterior position of the proximal tibial suspension pad. The four-point suspension in this case is the same as in the sketch of Figure 6-B.

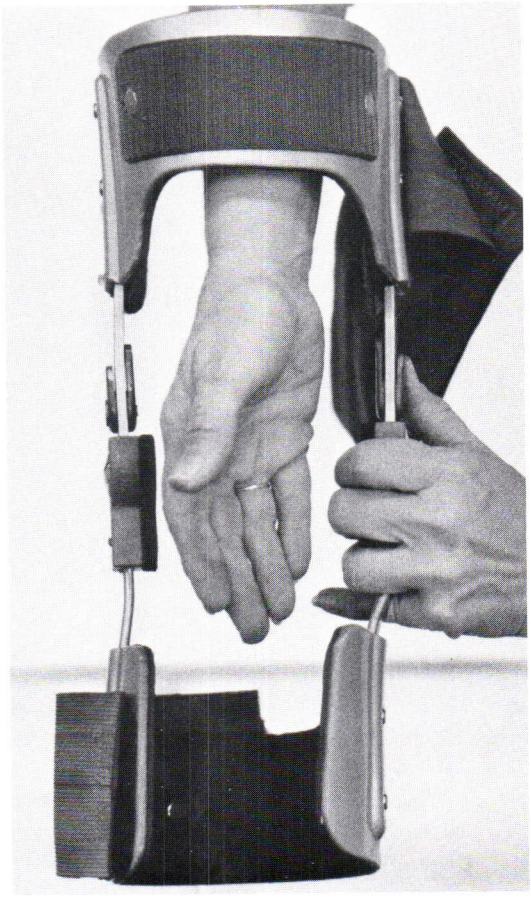


Figure 10. A typical rehabilitation orthosis design using linear low density polyethylene incorporating NuKO knee joints and an anteriorly placed tibial pad.

Four months postoperatively, as the patient's muscle volume increased to near normal, she was fit with an NuKO derotational posterior cruciate control knee orthosis. The NuKO derotational knee orthosis was used in conjunction with posterior placement of the proximal tibial pad and normal posterior-opening of the proximal and distal interface components. The patient was soon able to resume daily activities while the ligament repair continued to heal.

Case #4

A 21 year old male sustained a hyperextension injury to both knees. Examination of his left knee revealed a fracture of the medial tibial plateau, and an antero-lateral rotatory instability. His right knee exhibited both antero-lateral and postero-lateral rotatory instabilities. Since the patient also presented a marked posterior instability, a special posterior tibial pad was used in conjunction with an anteriorly placed anterior tibial pad (Figures 11-A and 11-B).

The patient underwent surgery, having a partial medial menisectomy of his right knee, as well as a repair to the anterior cru-

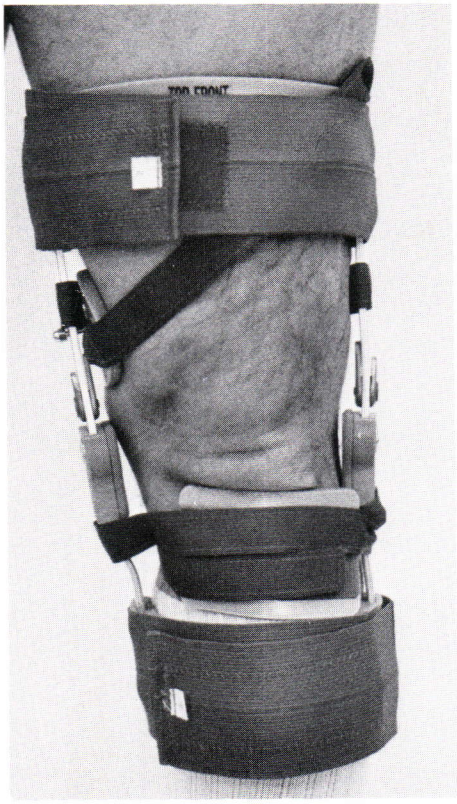


Figure 11-A. Anterior view NuKO derotational.

Figure 11-B. Medial view of NuKO derotational showing construction for both anterior and posterior instability.

ciate ligament and postero-lateral capsule of the same knee. The orthotic objective for his right knee was to prevent the repaired structures from becoming overloaded. The anterior and posterior tibial pads kept the knee in its neutral anterior-posterior position during flexion/extension and allowed the repaired tissue to remain unloaded. The anterior tibial pad was attached to the medial orthotic sidebar, restraining the anterior lateral instability as shown in Figure 5-A. The patient also eventually received an NuKO derotational knee orthosis to provide restraint for chronic antero-lateral rotatory instability in his left knee.

DISCUSSION

We have applied the NuKO orthotic knee system to a wide range of patient problems, including those with chronic ligamentous laxity, post-traumatic instability, and postoperative ligamentous reconstructions, and patients with total knee replacements, post-polio applications, and others. These applications probably represent the spectrum of potential users of knee orthoses. The results to date have been excellent. There have been complaints common to all knee orthoses, such as cosmesis and inconvenience, but generally, the clinical results have fulfilled our design expectations of a tighter fitting, more functional orthosis with minimal pistoning by virtue of the improved anatomically shaped orthotic joints. Early clinical follow-up has demonstrated improved results compared to our previous experience with other commercially available orthoses.

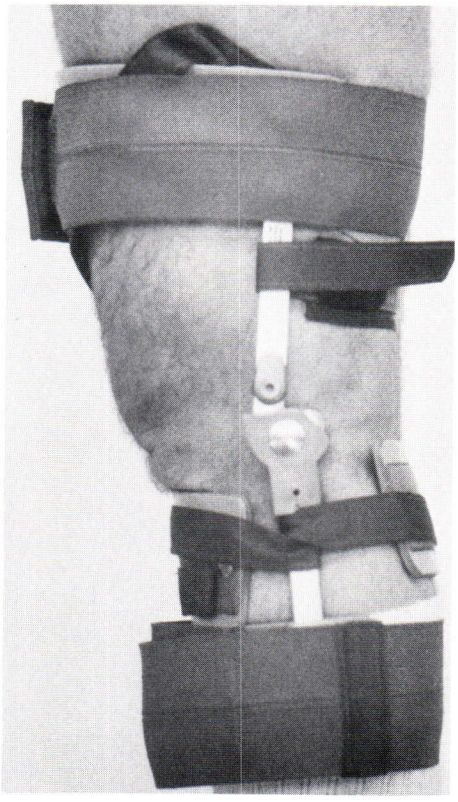


Figure 11-C. Lateral view of NuKO derotational.

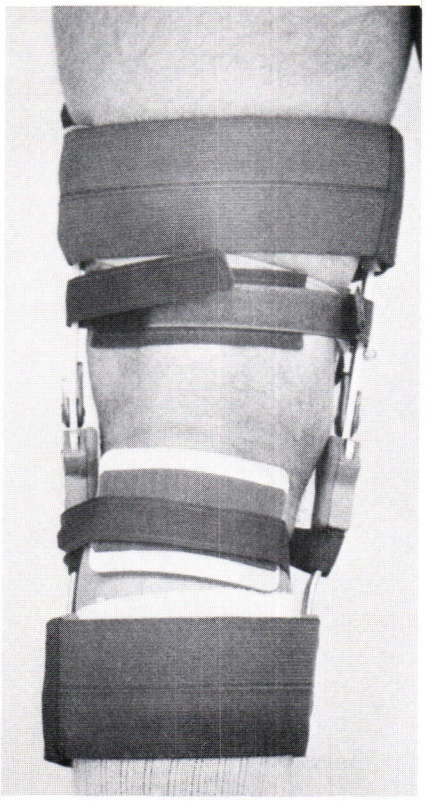


Figure 11-D. Posterior view of NuKO derotational.

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ACKNOWLEDGMENTS

The authors wish to acknowledge the contributions of George T. Shybut, M.D., William Lew, and Jack L. Lewis, Ph.D. to this report.

This work was supported by Grant No. G00820024 from the National Institute of Handicapped Research, Department of Education, Washington, D.C. 20202, and from private industry.

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

Photography by Frank Cazzola and Janice Russ.

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 51–56.

Two Way Stretch Knee Orthoses

Peter Brummer

INTRODUCTION

As is the case with all orthotic devices, a correct fitting is mandatory in order to achieve the maximum benefit and therapeutic value. To provide a service to the patient, the orthotist must know the biomechanics of the knee joint as well as the causes and effects of various impairments to that joint. Once this is known, the correct orthosis may be selected based on its function.

THE KNEE JOINT

The knee joint is the largest, and in its construction, the most complicated joint of the human body. The joint consists of the femur articulating with tibia and patella (Figure 1). Due to the extensive flexion allowed by these joints, the articulating facets do not fit as well on each other as with other less complex joints. Therefore, there exist two menisci, medial and lateral, which act to allow the femur to track on the tibia more efficiently and, because of their elasticity, also act as shock absorbers. The menisci and other non-bony joint structures offer their most elasticity during flexion and extension in the ranges between 40° and 180° (Figure 2). It is the purpose of the knee joint to support our movements during walking, jumping, or sitting in a chair. In order for the joint to remain sturdy and stable during these motions, it relies heavily on the ligaments connecting the bony structures about the knee. These ligaments prevent the shifting

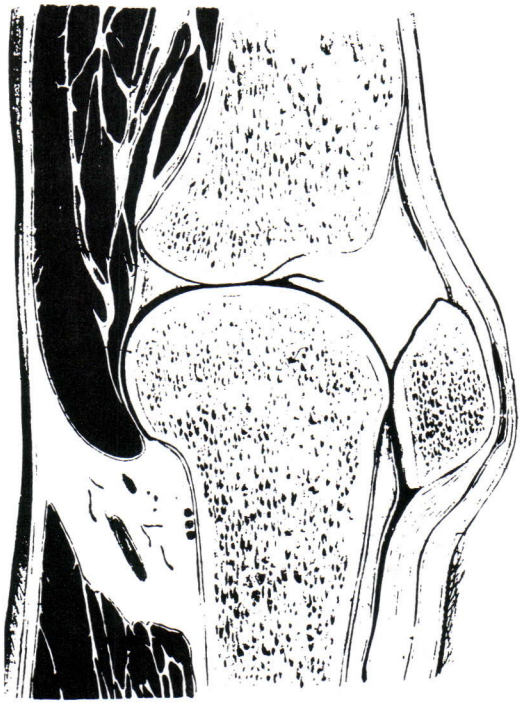

Figure 1. Cross section of knee joints showing femur, tibia, patella, and their articulating arrangement.

of the tibia on the femur. However, they do allow mild internal and external rotation of the joint during flexion (Figure 5).

KNEE ORTHOSES

It is the responsibility of the physician to recommend the correct therapeutic aid. The responsibilty of the orthotist is to apply the precribed knee orthosis such

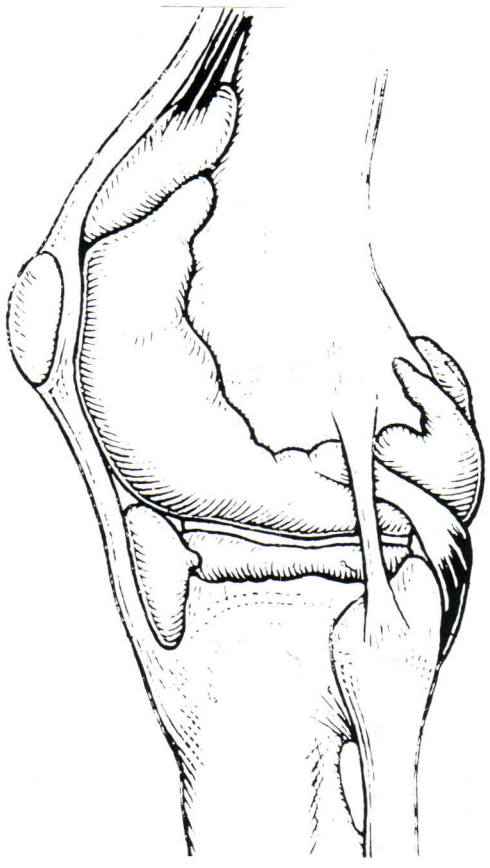

Figure 2. Lateral view of knee joint showing articular cartilage, bursas, patello-femoral ligament and lateral collateral ligament.

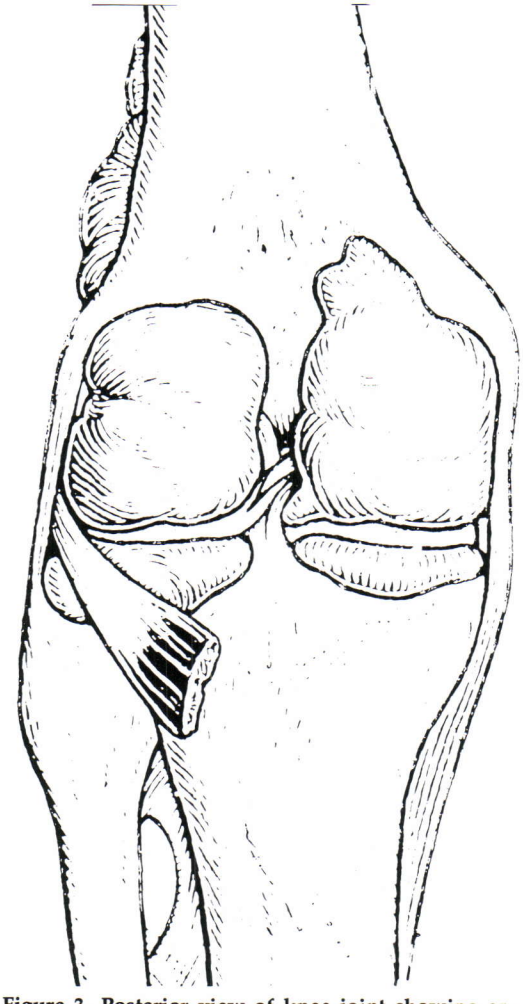

Figure 3. Posterior view of knee joint showing articular cartilage, posterior cruciate ligament, medial and lateral collateral ligaments.

that function and wearing comfort are met optimally.

The purpose of the knee orthosis is not to relieve the muscles of their function, but rather to support them during their function. If the compression rating of the elastic knee orthosis is too low; it is worthless. If the compression rating is too high, you may inadvertently cause thrombophlebitis or muscle atrophy.

Two way stretch knee orthoses, those whose elastic is bidirectional in both length and circumference, offer optimal support to the patient's impaired knee. Ideally, you would expect less stretch to occur circumferentially than in length for an optimal fit and smooth function. This is especially true for the two-way-stretch knee orthoses equipped with additional longitudinal elastic around the knee joint. Improper fit will cause the orthosis to be displaced during flexion of the joint (Figure 4). These reinforced knee orthoses are indicated for strains and sprains at the joint, following minor surgery, and as prophylactic treatment for weak ligaments.

The "Greifswald Knee Brace" was designed by the orthopaedic clinics of Greifswald and consist of two-way stretch

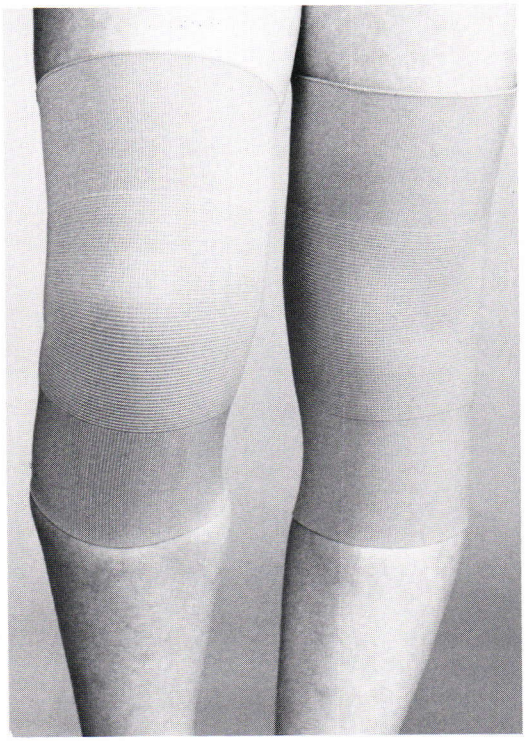

Figure 4. Elastic knee orthoses with reinforced elastic at the knee joint.

Figure 5. The Greifswald knee brace.

material with coil springs sewn into elastic knitted pockets (Figure 5). This orthosis is indicated in cases where there is damage to the meniscus or lateral collateral ligament injury. The coils may be varied in width and/or length in order to affect the mobility of the knee joint (Figure 6).

The Patellaligner knee orthosis is equipped with two pads and an elastic strap. It is recommended in cases of chronic subluxation of the patella, infra and suprapatellar tendonitis, as well as chondromalacia caused by chronic subluxation of the patella.

The L-shaped lateral pad locks the patella during its entire range. The elastic strap sets the lateral pad tightly against the patella. The medial epicondylar pad maintains the orthosis on the leg (Figure 7).

The Marshall PAC orthosis was developed in New York. It features a horseshoe-shaped pad around the patella, and is indicated for dysplasia, subluxation, and dislocation of the patella as a prophylaxis for high performance sports or after traumatic injury to the knee.

The horseshoe pad prevents a shift of the patella off of its correct path, while the strap about the thigh decreases the tension from the quadriceps muscle onto the patella (Figure 8).

The Hamburg Patella Brace, with lateral coils and infrapatellar strap, is indicated in cases of non-surgical treatment of chondromalacia and patellofemoral osteoarthritis. Along with the aforementioned orthoses, it is available with an open patella to reduce discomfort caused by pressure on the patella. The strap alters the mechanical action of the patellofemoral knee articulation during walking. The patella is slightly lifted by the pressure to the patellar tendon. This relieves stress to the patella by causing a change in the tension of the extensor muscles of the knee joint (Figure 9).

The Articula orthosis is equipped with

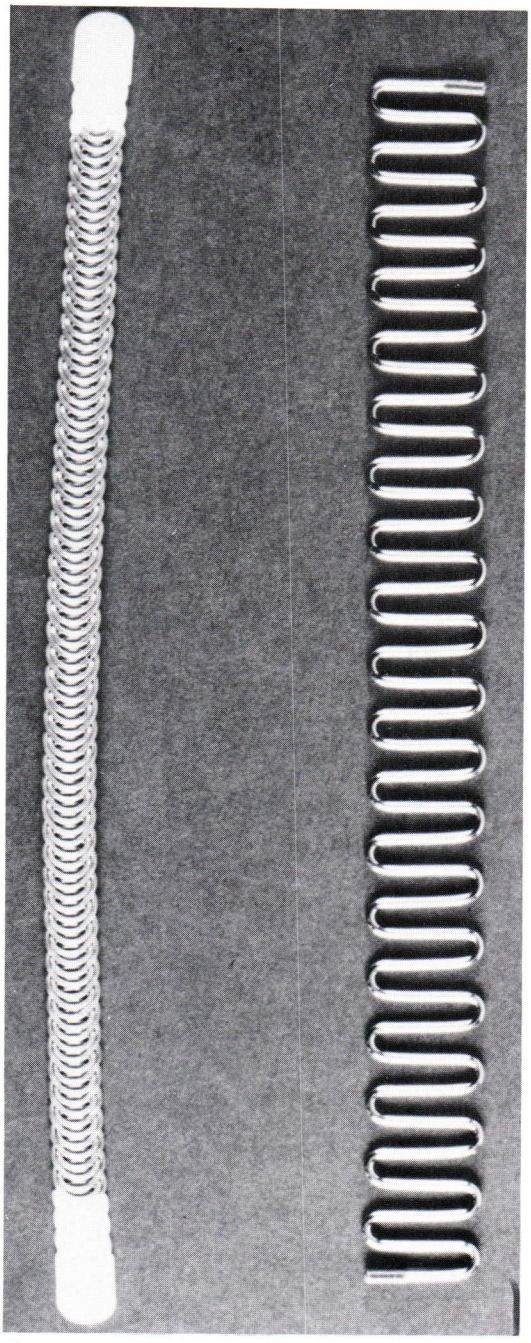

Figure 6. The coils may be varied in length and width to affect mobility of the knee joint.

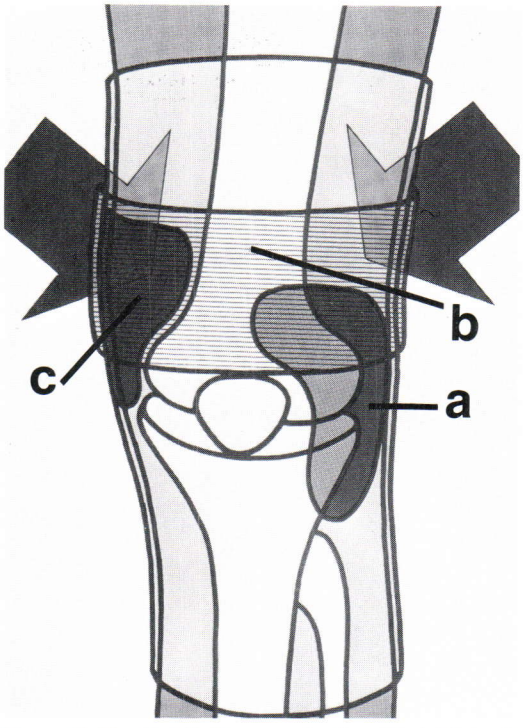

Figure 7. The Patellalinger orthosis with (a) lateral patella pad, (b) suprapatellar strap, and (c) epicondylar pad.

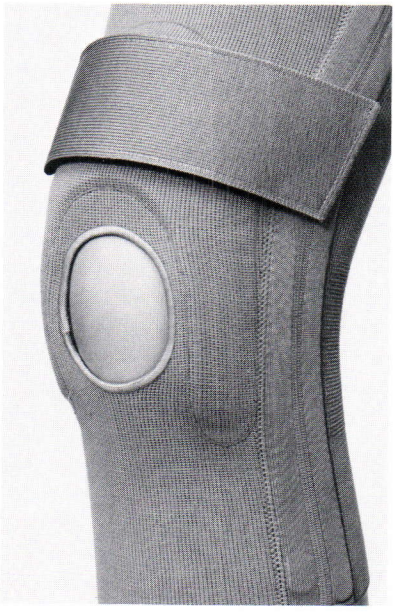

Figure 8. The Marschall Pac orthosis.

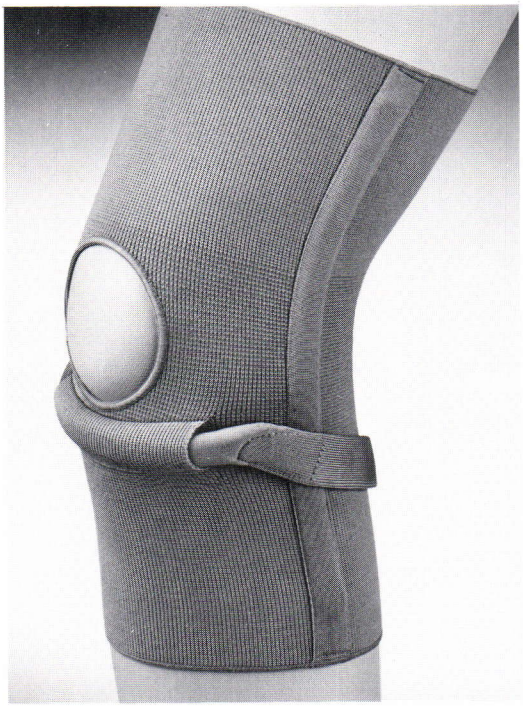

Figure 9. The Hamburg Patella Brace.

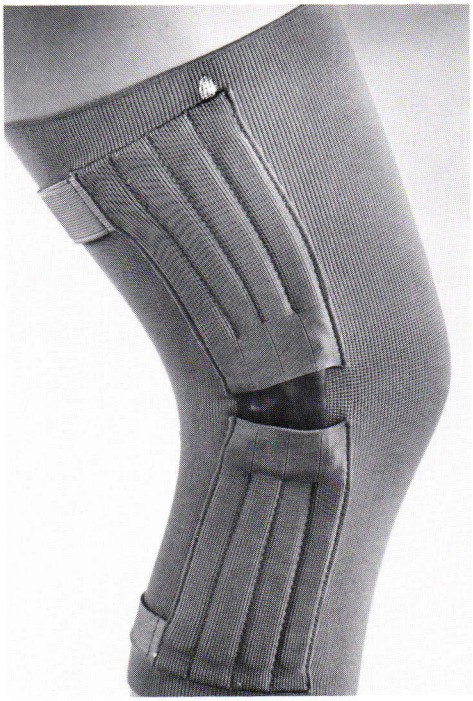

Figure 10. The Articula Orthosis.

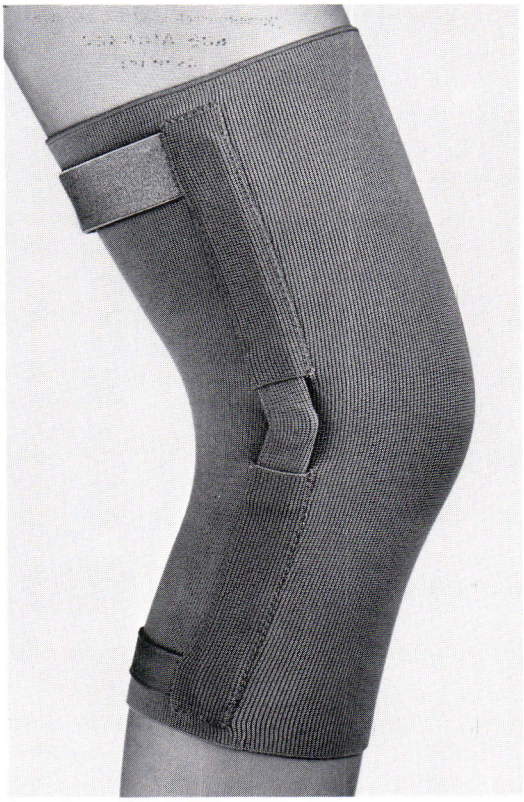

Figure 11. The Lastic Steel Hinged Brace.

malleable side bars and centered joints, and provides support to the knee joint without restricting motion (Figure 10).

The Lastic Steel Hinged Brace is supplied with steel hinges. These may come with or without flexion stops, and provide for controlled motion of the knee joint, especially with damaged lateral collateral ligaments. The steel hinges are inside knitted elastic pockets while maintaining the two-way stretch properties of the over-all orthosis (Figure 11). Both of the above articulating orthoses are indicated for arthritis of the knee, damaged cruciate or collateral ligaments, as well as for postoperative management of the knee.

A wide variety of adaptations are available to alter these orthoses to the customized needs of the patient. Various closure systems, i.e. Velcro® or laces, are available, as are assorted sizes, dimen-

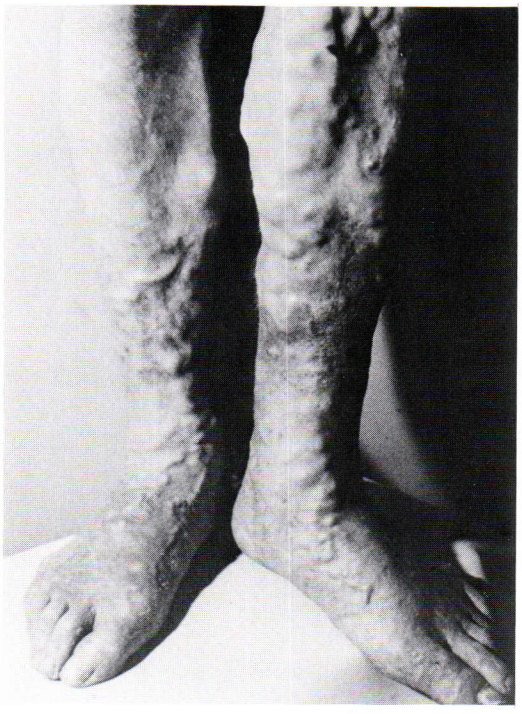

Figure 12. Knee instability in the presence of varicose veins should not be treated by knee orthoses alone.

sions, and durometers of corrective padding.

In cases where to treat the knee alone would cause damage elsewhere, as with varicosities of the calf and ankle with concomittant risk of phlebitis, elastic knee orthoses can be incorporated into a compression stocking (Figures 12 and 13).

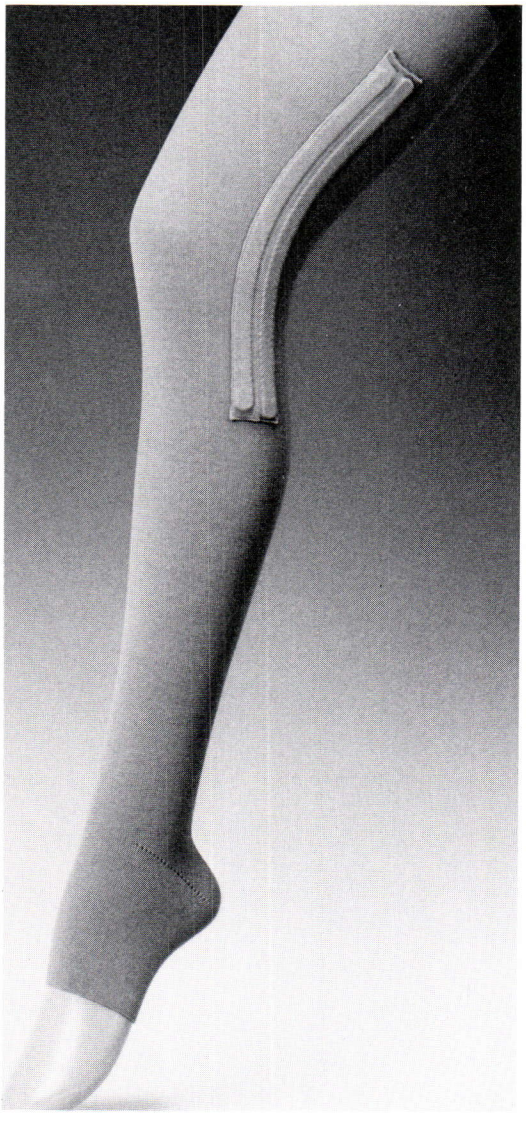

Figure 13. A knee orthosis may be integrated into a compression stocking.

CONCLUSION

To properly treat the patient with knee instability or insecurity, it is important to know the function of the various supports that are available to treat the associated disorders.

This article has described several elastic knee orthoses which may be used to treat mild to moderate disorders at or about the knee joint, and was developed to serve as a reference source for orthotic practitioners who must fit these orthoses.

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 57–65.

A Computerized Ultrasound Shape Sensing Mechanism

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Norman G. Gall, M.D.

INTRODUCTION

The most important aspect of a lower extremity prosthesis is socket design. The design determines the prosthetic fit which affects cost, suspension of the prosthesis, comfort, energy expenditure, and ultimately the utility of the device in patient ambulation. To quote T. Walley Williams, "A socket is unique to each patient and socket fabrication and fit are the essence of the prosthetic art."[10] There are several disadvantages to the present method of prosthesis socket fabrication and fitting. First, the prosthesis is primarily the result of hand craftmanship. Because of this, there is no scientific way to quantify and accurately record details of modifications made to produce the finished socket. Second, the present method of fabrication involves making a positive plaster model, which is a modified replica of the topographical shape of the residual limb. During the fabrication process, this model is often destroyed thus eliminating any topographical record of the limb for future reference. It is well known that the average amputee will require several new sockets during the residual limb maturation process following the amputation. Many amputees will require periodic replacement of the prosthesis as long as they live.[5,9] Each time the process must start anew without the benefit of information previously gained. The lack of topographical data also makes it impossible to describe "typical" changes that occur in the residual limb, thus deterring development of a socket which would have a longer optimal fit.

Limited attempts have been made to image the residual limb for research purposes. Fernie, et al.[4] in Toronto have used a video camera and laser sensing system to provide topographical information for student prosthetists. Investigations by Oshimna and Saito[8] and Aguallo, et al.[1] using a three-dimensional digitizer for shape sensing have provided early data in the use of computer assisted design for prosthetic devices. Researchers at Baylor University are in the early investigation stages of evaluating ultrasound as a tool for predicting load shape of residual limbs.[6]

The Rehabilitation Engineering Laboratory (REL) at the University of Texas Health Science Center in San Antonio has explored three-dimensional reconstruction of the residual limb from computerized tomography image data using the CEMAX-1000 computer system with excellent data acquisition, but at a prohibitive cost.[3]

The Veteran's Administration Research and Development Section under the direction of Dr. Margaret Giannini held a four

day seminar/workshop to evaluate state-of-the-art techniques of CAD/CAM in prosthetics and orthopedic footwear in the fall of 1985. Participants from the United States, Canada, and European rehabilitation communities concluded that there is a need for development and evaluation of automated shape sensing devices and CAD/CAM of artificial limbs.[11]

The greatest barrier to the implementation of CAD/CAM in prosthetics is the inability of the prosthetist to capture the exact topographical shape of the amputee's residual limb. Once objective shape information is stored in the computer, it is relatively easy to then devise a computer program that will allow the prosthetist to manipulate the shape for socket design. When this shape is perfected, it can be transmitted to a computer controlled milling machine for socket or positive model manufacture. If the previous socket design information was available, it would be possible to maximize the fit of the prosthesis each time a new one was needed, based on the computer generated data.

METHODS

To shape sense a residual limb, the Rehabilitation Engineering Laboratory at the University of Texas Health Science Center at San Antonio (REL) used an ultrasonic transducer mounted on a servomotor controlled carriage. The assembly was controlled by a personal computer and was mounted in a tank of water to provide a medium in which ultrasonic pulses/echos were transmitted. The carriage allowed the transducer to be rotated horizontally 360° in 0.5° steps and to move vertically 7″ in 0.01″ steps around the surface of a lower extremity limb placed in a stationary position in the tank (Figure 1).

An A-mode control module sends pulses at regular intervals (100HZ-10KHz) to the transducer. A pulse causes the transducer to send an ultrasonic wave front through the water. When the wave front encounters tissues of varying density, a return echo is produced, which is detected by the transducer. The pulse from

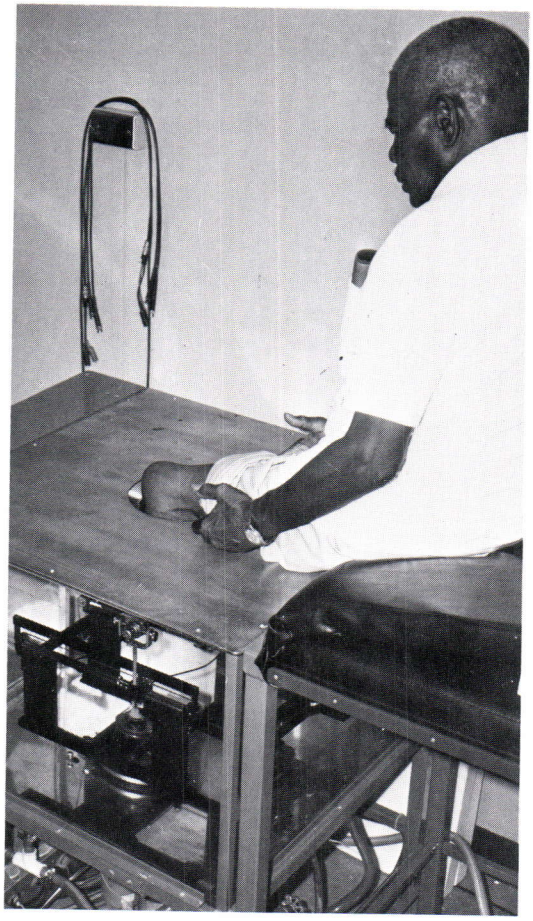


Figure 1-A. A patient's residual limb is scanned by the ultrasound shape sensing device.

the control module is also used to trigger a frequency counter configured for interval timing. The return echo from the transducer is then amplified and filtered to provide a gate signal for the frequency counter. The counter measures the elapsed time between the trigger pulse and the first return echo (Figure 2). The counter is then disarmed until the next trigger pulse. This measures only the surface of the residual limb; Tissue under the skin, bone, and the back surface of the limb is not detected. The counter reading is directly proportional to the distance between the transducer head and the surface of the limb, and the distance is accurate to within 0.002″ on a surface perpendicular to the beam axis (Figure 3). A small error occurrs when the surface being scanned is not

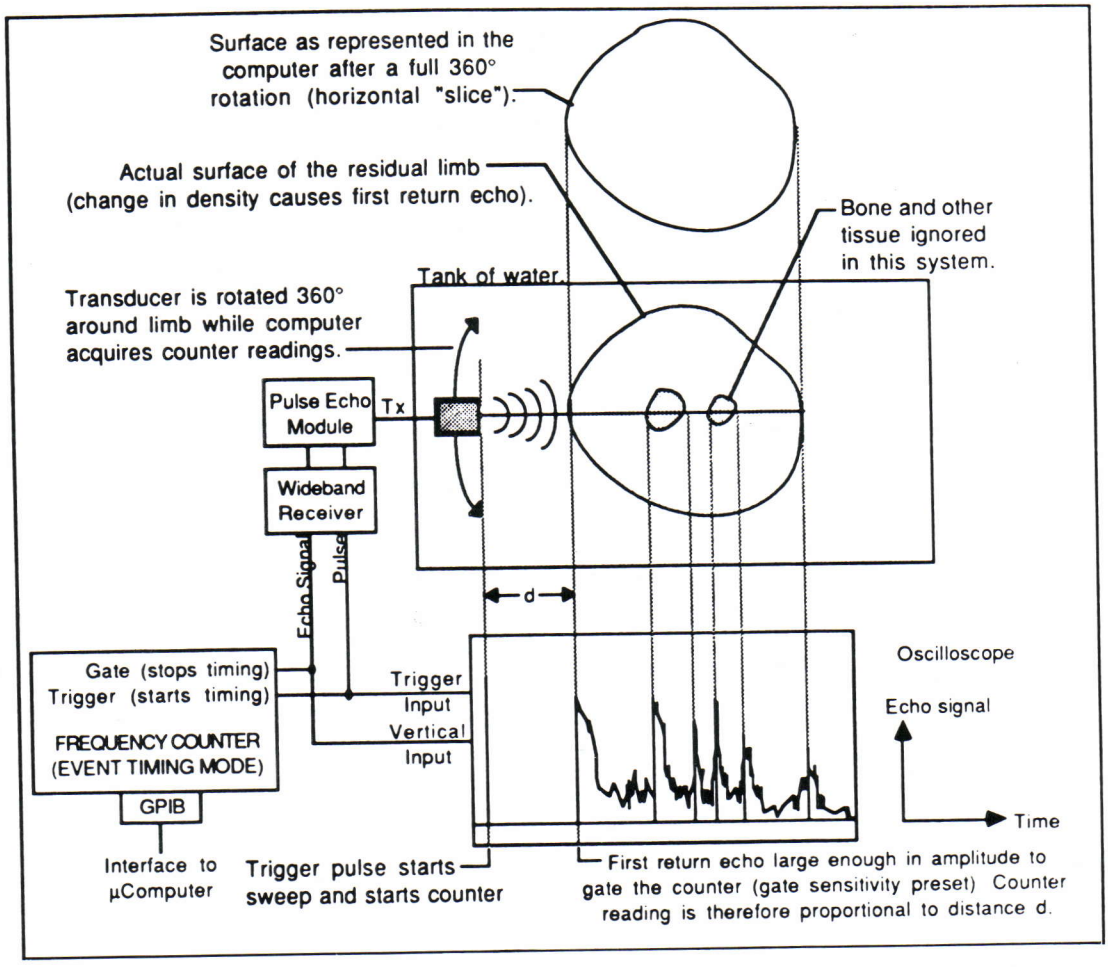


Figure 2. Operation of the A-mode Ultrasonic transducer and analog electrics subsystem.

perpendicular to the beam axis (Figure 4). This error can be minimized by selecting a transducer with the appropriate focal length and beam pattern.

The computer controls the position of the transducer by sending a pulse train through a parallel interface circuit to the servomotor controllers (Figure 5). Data representing the distance from the transducer to the residual limb surface is acquired from the frequency counter through a GPIB interface board (Figure 6). As the computer rotates the transducer around the limb, readings are taken from the frequency counter and converted to absolute x and y coordinates. These coordinates represent a horizontal "slice" of the residual limb surface. After a full 360° rotation, the transducer is moved vertically a known distance and another slice is acquired. Each slice is located on a known z coordinate and is combined with the x and y data to reconstruct the topological surface of the residual limb.

The program which runs the data acquisition procedure may be configured by the operator by varying the number of data points per slice (280 max), the number of slices acquired (99 max), the distance between slices (0.01" to 7"), etc. This program is also used to convert the data into a format compatible with the commercially available "Advanced Space Graphics" CAD program. This software product is used to display, rotate, translate, scale, and modify the topological surface (Figure

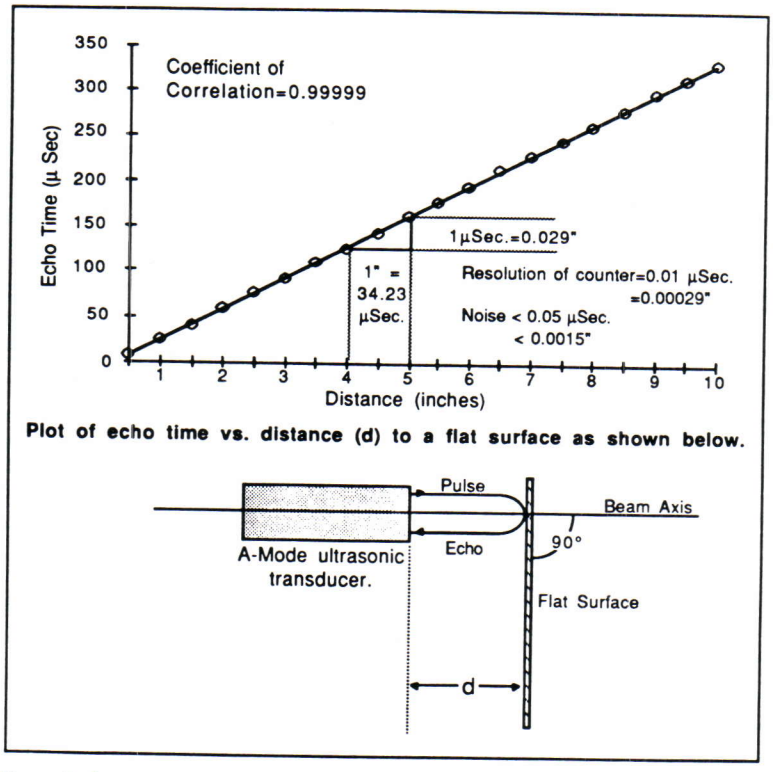


Figure 3. Arrangement used to measure accuracy of the ultrasonic transducer.

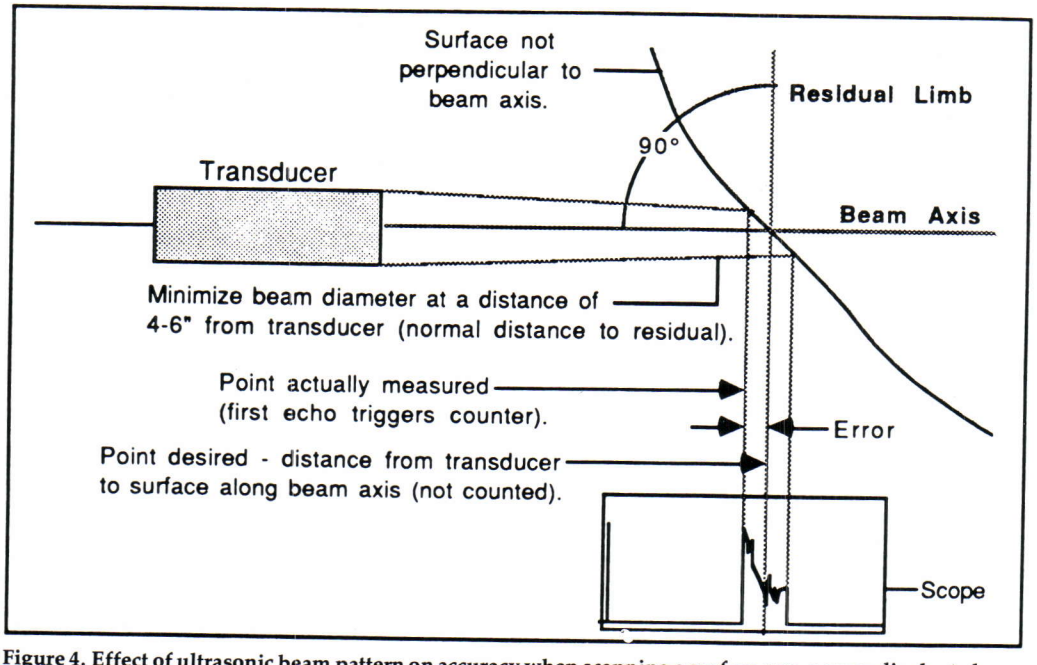


Figure 4. Effect of ultrasonic beam pattern on accuracy when scanning a surface non-perpendicular to beam axis.

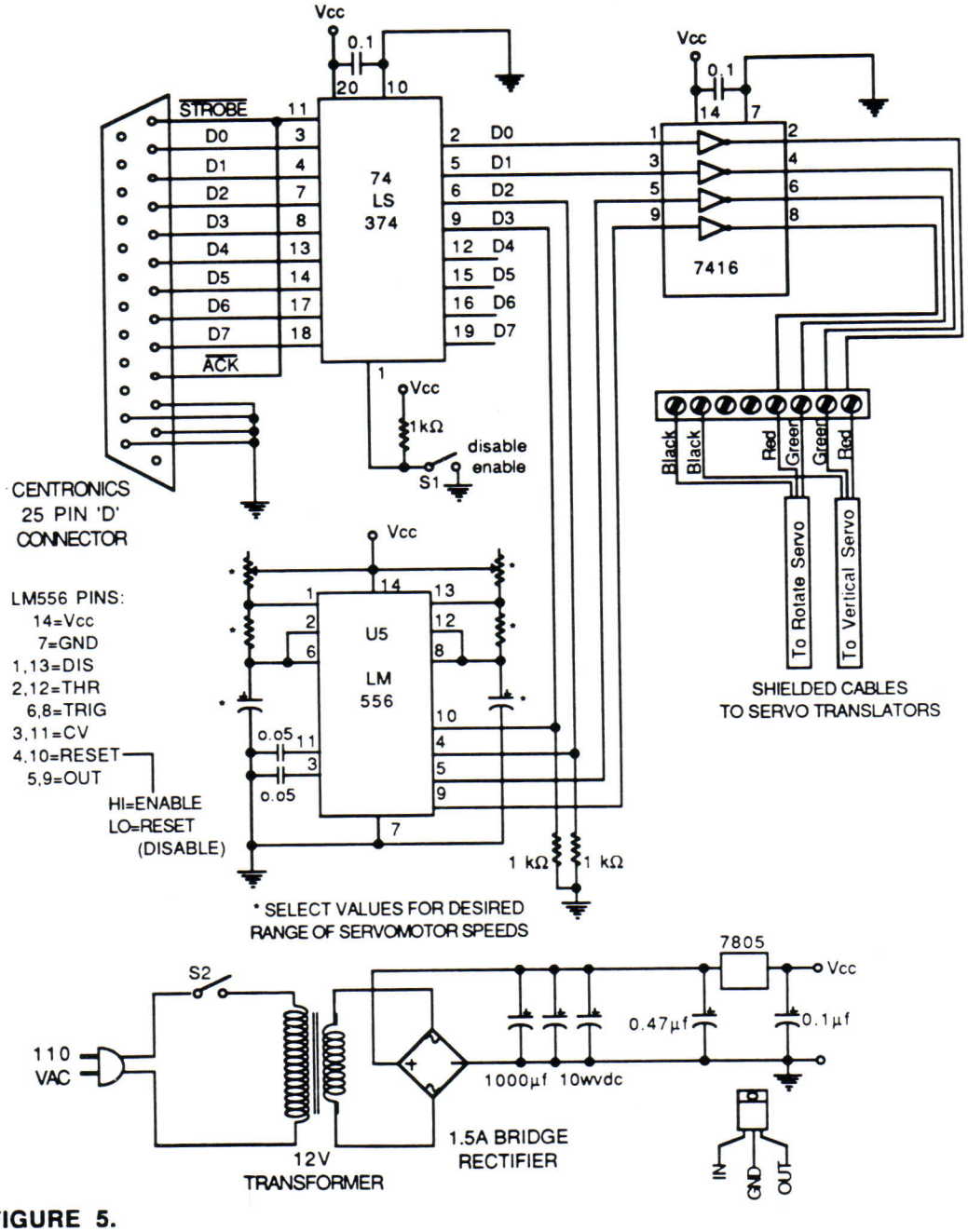
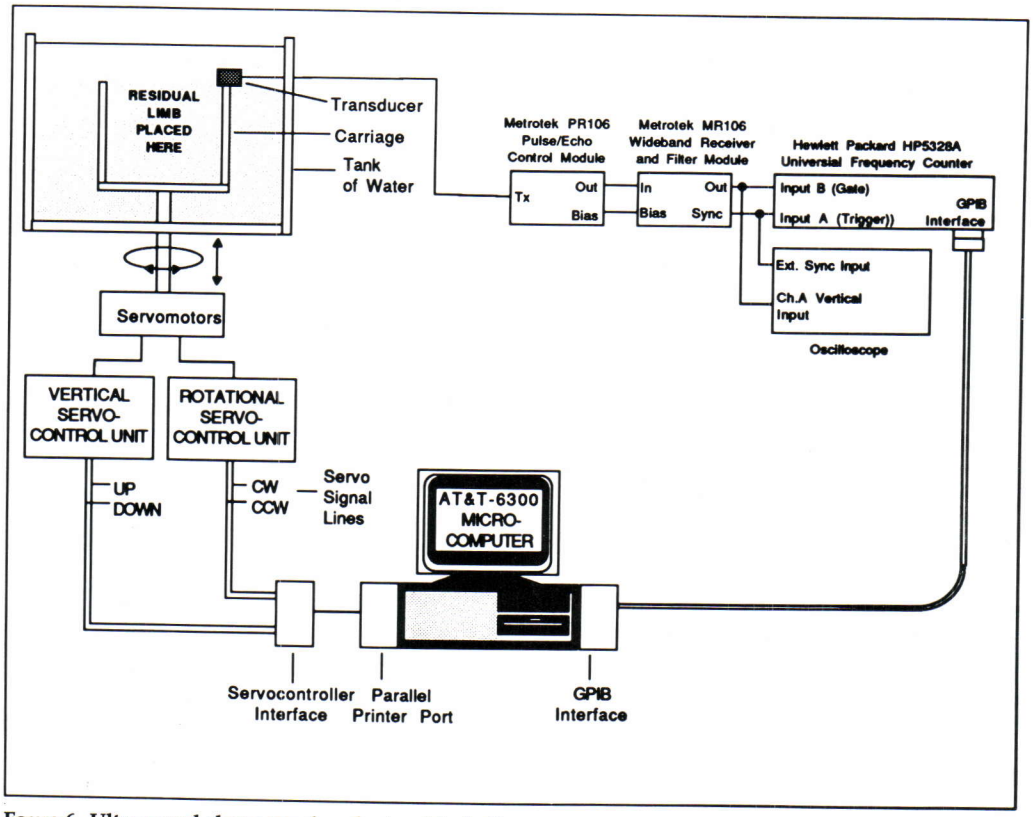


FIGURE 5.

Figure 5. Servo translator interface to the centering parallel port.



Fgure 6. Ultrasound shape sensing device, block diagram.

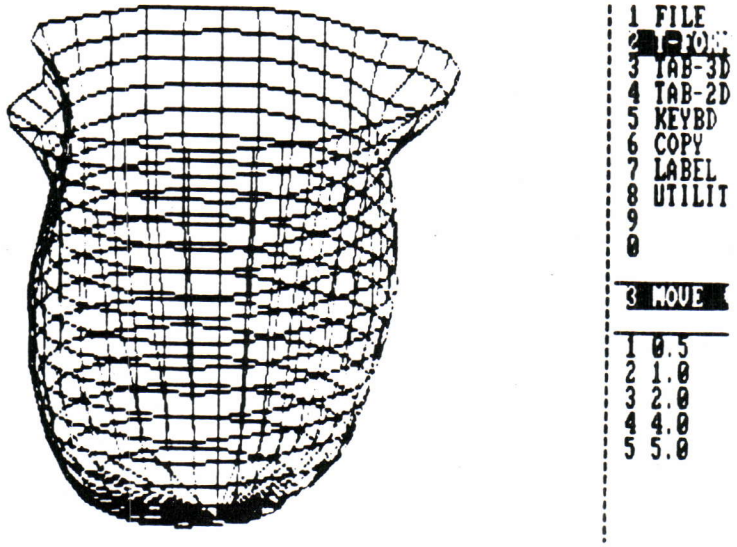


Figure 7. Example of a residual limb surface acquired by the ultrasound shape sensing device and displayed with the advanced space graphics program.

7). The REL has developed software to reformat the data for compatibility with a Computer Controlled Milling Machine.

RESULTS

The ultrasound shape sensing device currently in use at the San Antonio REL is capable of producing an accurate computer generated topographical image of a residual limb. The system is composed of two basic subsystems; the ultrasound and analog electronics subsystem, and the computer based data acquisition and control subsystem.

An image of a patient's residual limb is automatically generated in about ten minutes; the time is determined by the residual limb's length. The data representing the topological shape of the limb is stored on the computer's hard disk unit as a set of x, y, and z coordinates. This data is then available for modification by a number of custom and commercially available application programs. The ideal program would allow a prosthetist to easily modify the data, reshaping the topological surface according to normal prosthetic principles. The final version of the data may be sent via modem to a fabrication facility for socket or positive model manufacture.

DISCUSSION

The ultrasound shape sensing device (USSD) was developed as a proof of principle project. It has demonstrated high resolution for the perpendicular areas of the residual limb, but is less accurate when describing the distal end by utilizing the present scanner pathway. The USSD is limited to seven inches of vertical movement with the present carriage assembly. This is an inadequate range for patients with longer residual limbs. Therefore, these design limitations are being corrected.

Future hardware enhancements of the system will allow it to quickly and accurately capture the residual limb's surface and bone topography. Software enhancements will include a smoothing routine for the topological surface, utilizing user-friendly real-time interaction.

An algorithm, offered by Fujio Yamaguchi[12] may be used to accomplish this curve smoothing. This method is extremely fast and simple, requiring no multiplication or division by the computer. This routine will add more points to those already existing in the data set and smooth the topological surface. Because points are added, fewer points need to be acquired in

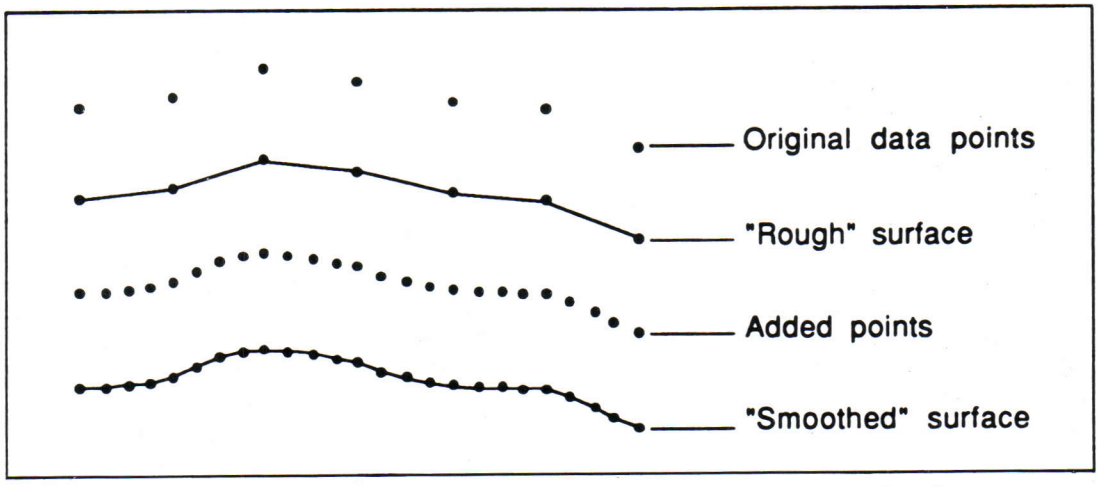


Figure 8. Example of how the curve smoothing routing would add points to a topological surface.

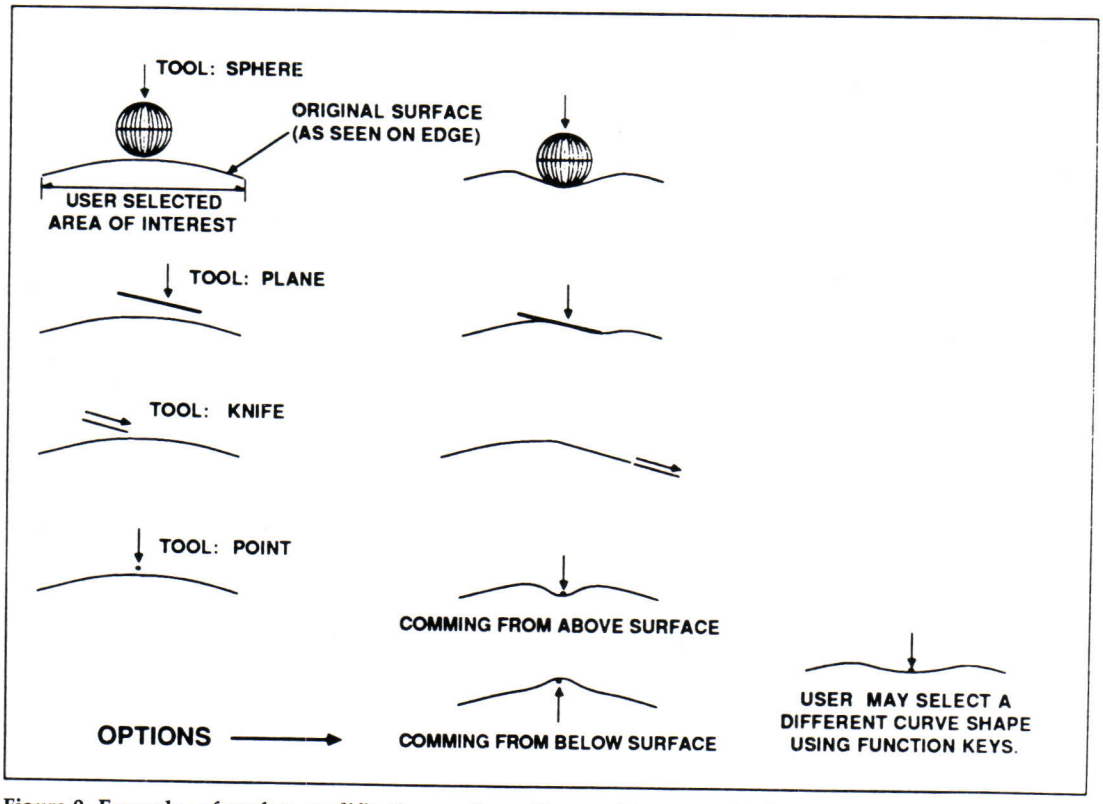


Figure 9. Examples of surface modification routines. The graphics cursor is shown as a "tool," and is used to modify the surface in real-time.

the scanning process, resulting in a decreased amount of time necessary to scan a patient (Figure 8).

A method described by Brewer and Anderson[2] for designing a CAD program allowing free form modification of a surface will be used for modification (Figure 9). This program provides the display of selected areas of interest including points to be modified, so other points will not be affected. The user may then select a "tool" to modify the surface. The tool is displayed on the cathode ray tube and may be moved about with the cursor control keys. As the tool is moved to the surface, all points in the area of interest are modified. How the points are modified depends on the tool selected and its characteristics. This offers a vast improvement over the Advanced Space Graphics Program, which only allows modifications on a point by point basis.

CONCLUSION

The current system for prosthesis design and fabrication is a very expensive and time consuming endeavor, which is beyond the reach of many third world countries[7]; CAD/CAM may make prosthetics and orthotics available to amuptees in these regions.

The greatest barrier to the implementation of a computerized system for making these apliances is the inability of the prosthetist to capture the exact topographical shape of the affected limb. A proof of concept ultrasound shape sensing device, that provides a computer controlled method to obtain descriptive topographic data of an amputee's residual limb, has been developed. While design limitations of the prototype device require modification, the application of ultrasound technology for topographical quantification of residual limbs appears promising.

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Orthotics and Prosthetics, Volume 41, Number 4, pp. 66–70.

An Alternative Casting Technique For Narrow ML Sockets

Justin Horowitz, C.P.O.
Michael Lefton, C.P.O.

INTRODUCTION

Many articles have been published with regard to new casting techniques for narrow ML socket design. All involve new methods or new equipment. The authors would like to present a new casting technique which will greatly cut down on cast modification time as well as the number of check sockets required.

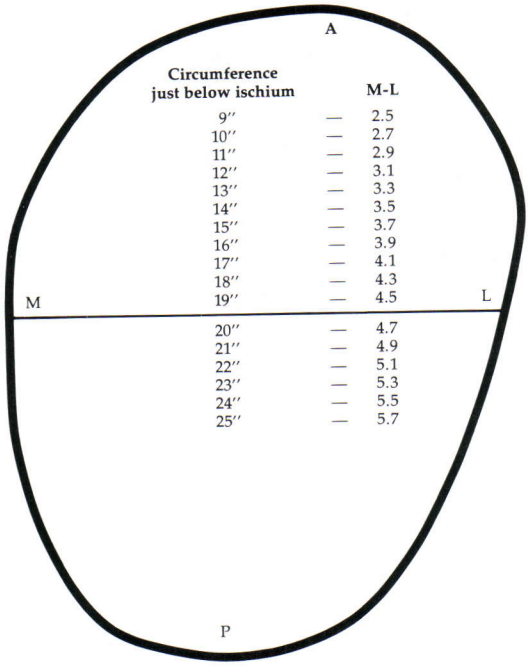
We began a study on the possible application of the Berkley Brim for casting of a narrow ML socket. Taking a brim for the contralateral side, rotating it 90°, the original posterior seat becomes the medial wall and the medial wall becomes the posterior seat. The Scarpus bulge contours to the posterior lateral region. With this brim, we were able to adjust the ML dimension.

Initially, we found that by utilizing the graph established by Mr. Long (Figure 1-A) and the measurement data sheets designed by Mr. Sabolich and Mr. Shamp (Figure 1-B), we could select the appropriate brim. As we became more comfortable with the system, we found that we could select the appropriate brim size as if we were casting for a conventional quadrilateral socket, but using the contralateral brim.

CASTING

Following the brim selection, a circumferential gauge should be employed to establish your reduced ischial level measurement. Tube gauze is placed on the patient's residual limb and appropriately suspended. The brim is then placed around the limb and moved proximally to contact the ischial tuberosity. The gentle drawing downward of tissues may be required (Figure 2). At this time, check the tissue distal to the brim for even continuity. Adjustments may be required by loosening or tightening the adjustments screws, but do not be concerned with any gaps that may occur between the limb and brim in the posterior region. In this region the residual limb, not the brim, will give you the appropriate contour (Figure 3).

The brim is then removed from the limb, and the trochanter and distal femur are marked. At this time the brim is widened by 2.5 rotations of the adjustment screws. This allows for the thickness of the plaster. In addition, the brim is coated with silicone spray.



Circumference just below ischium		M-L
9″	—	2.5
10″	—	2.7
11″	—	2.9
12″	—	3.1
13″	—	3.3
14″	—	3.5
15″	—	3.7
16″	—	3.9
17″	—	4.1
18″	—	4.3
19″	—	4.5
20″	—	4.7
21″	—	4.9
22″	—	5.1
23″	—	5.3
24″	—	5.5
25″	—	5.7

Figure 1-A. Socket pattern for thigh measuring 19″. Actual measurement of pattern circumference is 18″.

In order to support the femur, it is necessary to narrow the M-L dimension of the socket. The resulting greater A-P allows muscular function not possible with the crowded effect of a narrow A-P. This chart is to be used as a guide in establishing the width of the finished positive model. The figures were taken from approximately 500 sockets made in this facility and many of these sockets have now been worn eight years. Very few, if any, sockets have been replaced because of shrinkage. Many sockets have been replaced as muscles return to normal and the thigh takes on its original shape and size increases. Most of the increase in size will take place in the A-P dimension, with very little change in M-L. Increasing the M-L dimension by anything more than ¼″ will result in a lateral gap at the top of the socket.

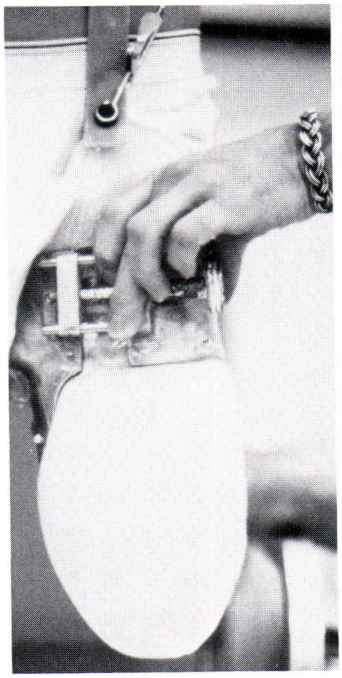


Figure 2. In donning the selected brim on the patient, a gentle drawing downward on the tissue may be necessary.

Figure 3. Any gapping in the posterior region should not be a concern, as the residual limb will give you the appropriate contour.

Contoured Adducted Trochanteric
Controlled Alignment Method

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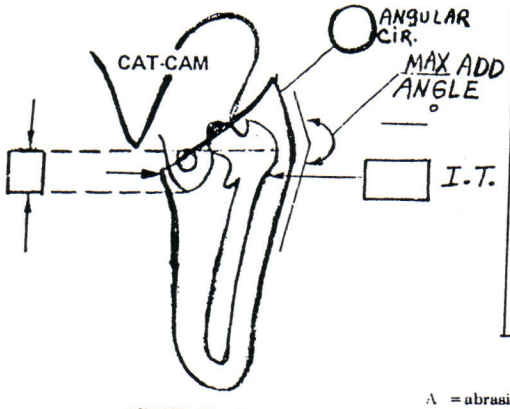
Amp. Date_____ Reason_____ Street Address_____

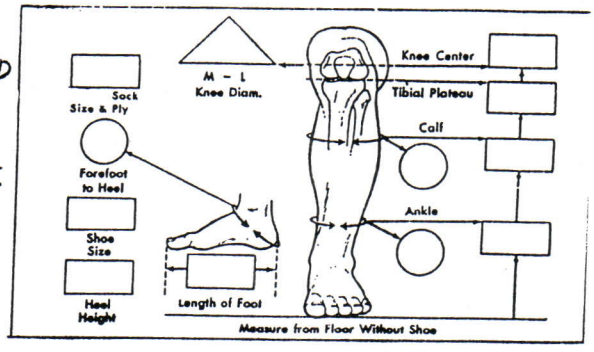
Site of Amputation_____ City_____ State_____ Zip_____

Height_____ Weight_____ Age_____ Clinic or Hospital_____

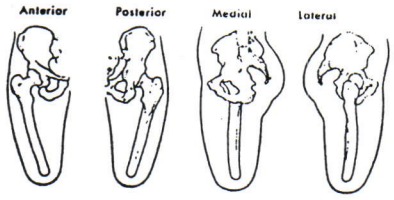
MALE [ ] FEMALE [ ] LEFT [ ] RIGHT [ ] Prescribing Physician_____

Prosthetist's_____





ABOVE KNEE



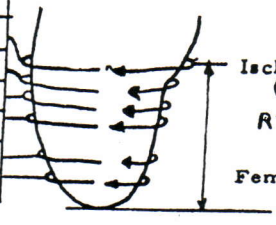
A = abrasion
B = boil or skin infection
Bu = bursa
Bs = bone spur
D = discoloration
E = edema
I = irritation
M = muscle bunching
P = pressure point
R = redundant tissue
S = scar
T = trigger point

SPECIAL NOTES:
Trochanter
Ischial Tub.
Sub-Tissue
Flex- Cont.
ABD-Cont.

Suction reduct.	ML	Distance Below Ischium	Circum-ference	
		0		
		2		
		4		
		6		
		8		
		10		

Mod. Unmod.

CAT-CAM



Trochanter to Anterior Midline

Ischial Tuberosity (Standing)

RL. Length

Femur Length

To Floor

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Figure 1-B. Measurement data sheet.

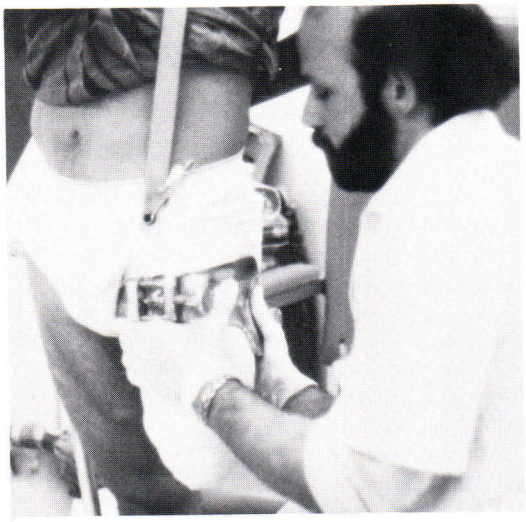

Figure 4. The brim is hand held during the taking of the negative impression with the residual limb in a relaxed position.

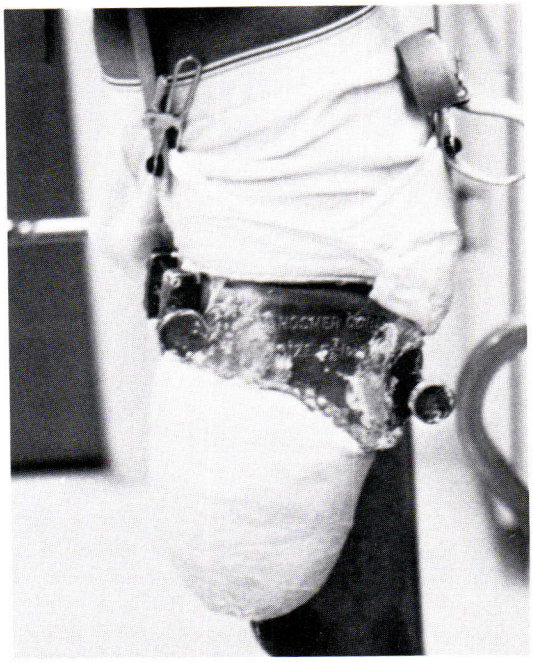

Figure 5. Viewed laterally, the top of the brim is maintained in the same flexion attitude as the residual limb.

Elastic bandage is applied to the limb, covering at least a third of the gluteal musculature. The bandage is then brought into the perineum and drawn towards the inguinal region to finish at least 2″ proximal to the trochanter. The proximal wrapping should be four layers thick. Continue wrapping the limb. Once the wrapping is completed, the brim is then applied. The brim will be hand held; The affected limb is held in a relaxed position (Figure 4). With the brim being held firmly against the ischial tuberosity, view the fixture laterally (Figure 5). The top of the brim is held in the same attitude as the limb. When viewing the brim anteriorly, just prior to the plaster setting, have the patient actively adduct, while keeping the brim horizontal (Figure 6). This position is maintained until the plaster has hardened. At this point, a top coat of rigid plaster can be applied to the casting fixture and plaster wrap (Figure 7). In addition, three layers of rigid plaster should be applied to the gluteal region from the brim proximally. Once these layers are hard the negative impression is removed and the length is verified. Positive model modifications are performed according to the previously mentioned authors.

Figure 6. Just prior to the plaster hardening, have the patient actively adduct his residual limb.

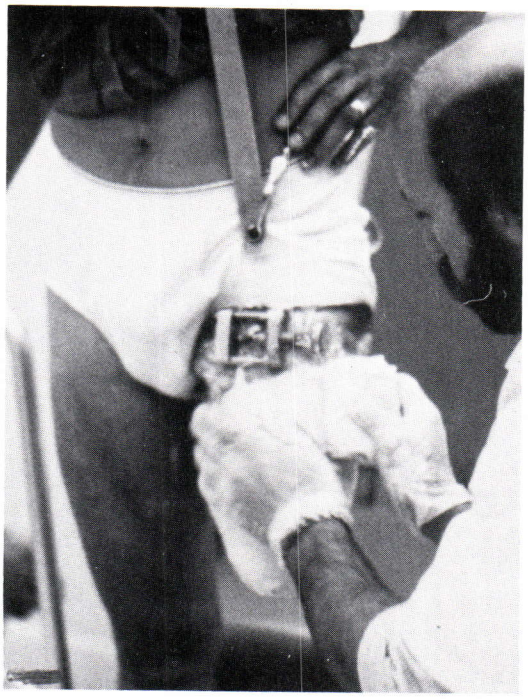

Figure 7. Once the elastic plaster bandage has hardened, it may be reinforced with rigid plaster bandage.

CONCLUSION

This technique has been tried successfully on three patients. Two patients required only one check socket. Our third patient required two sockets due to weight loss and suction not being maintained. Because of a preshaped proximal portion of the positive model, modification time is reduced by half.

AUTHORS

Justin Horowitz, C.P.O., and Michael Lefton, C.P.O., are with M&M Prosthetic Associates, Inc., RD 2, Rt 28W, Box 194P, Kingston, New York, 12401.

Orthotics and Prosthetics, Volume 41, Number 4, pp. 71–72.

Orthotic and Prosthetic Management of the AIDS or Infectious Patient

Carey A. Glass, C.P.

INTRODUCTION

Over the past two years the problem of Acquired Immune Deficiency Syndrome (AIDS) has become an ever increasing concern to the health care professional. Until now, articles and reports written by health care professionals have dealt with how to protect yourself from direct contact by controlling the work environment. As health care professionals and specialists, we are called upon to supply prosthetic and orthotic services to these patients, allowing them to lead as normal a life as possible. Considering the confusion over how to handle AIDS patients, let's first review some basic facts about the disease.

A REVIEW

What is AIDS?

Acquired Immune Deficiency Syndrome is a disease that effects the immune system of the body and its ability to fight infection.

What Causes AIDS?

The cause has been identified as a virus referred to as HIV Positive (HIV+), or Human Immunodeficiency Virus. HIV+ attacks the white blood cells that protect the body from infection. By destroying the white blood cells, the virus reduces the body's resistence to infection. Eventually a simple flu virus or other infections cannot be controlled by the body's immune system, causing the patient to become seriously ill and eventually die.

What are the Symptoms of AIDS?

The HIV+ virus can be latent and, while remaining infectious, some people do not have any symptoms at all. In some cases, it could be months or even years before any true symptoms are noticed. Some of these are white spots on the mouth and tongue, chronic diarrhea, swollen glands, excessive weight loss, fever with night sweating, Karposi's Sarcoma, or purplish sores on the skin.

How Can AIDS Be Spread?

The AIDS virus has been found in blood, breast milk, saliva, urine, semen, and tears. It can be passed on by intravenous drug use, direct contact with contaminated blood or blood products, as well as sexually.

What Procedures Should be Taken to Protect the Prosthetist and Orthotist?

Over the last eight months, with cooperation of doctors and staff of both hospital and health care facilities, we have developed a procedure to follow when handling infectious patients in the hospital. First, vinyl or latex gloves should be worn when

examining, casting, or measuring a patient as well as when working with garments or devices in intimate contact with the patient. It is suggested that when dealing with open wounds during cast removals, double gloves should be worn. Second, respiratory masks and goggles should be worn when removing casts, for protection from airborne debris. All open wounds should be covered, and all instruments that are used in measuring the patient should either be sterilized or disposable. It is very important to wash your hands with an approved soap before and after seeing the patient. The use of a balloon or plastic bag as an interface during casting is quite protective, and following useage it should be discarded with infectious waste. The HIV+ virus is very sensitive to basic sterilizing procedures such as Cidex baths, germicide, alcohol washes, or gas sterilization.

CONCLUSION

Acquired Immune Deficiency Syndrome is a health problem that must be dealt with in a common sense manner. When seeing patients as we do in our offices, as well as in hospitals, we must not forget to protect ourselves adequately. The estimates of the number of HIV+ carriers are high enough that certain procedures should be the rule, and not the exception. The health hazards to the prosthetic and orthotic professional are such that these precautions should be taken in handling all patients with open wounds.

AUTHOR

Carey A. Glass, C.P., is President of C.G. Medical Supply, Inc., 216 Somerset Street, New Brunswick, New Jersey 08901.

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Book Reviews

Useful Orthopaedic Eponyms

Richard B. Raney, M.D., Vantage Press, Inc., 516 West 34th Street, New York, New York 10001. 1987, 186 pages.

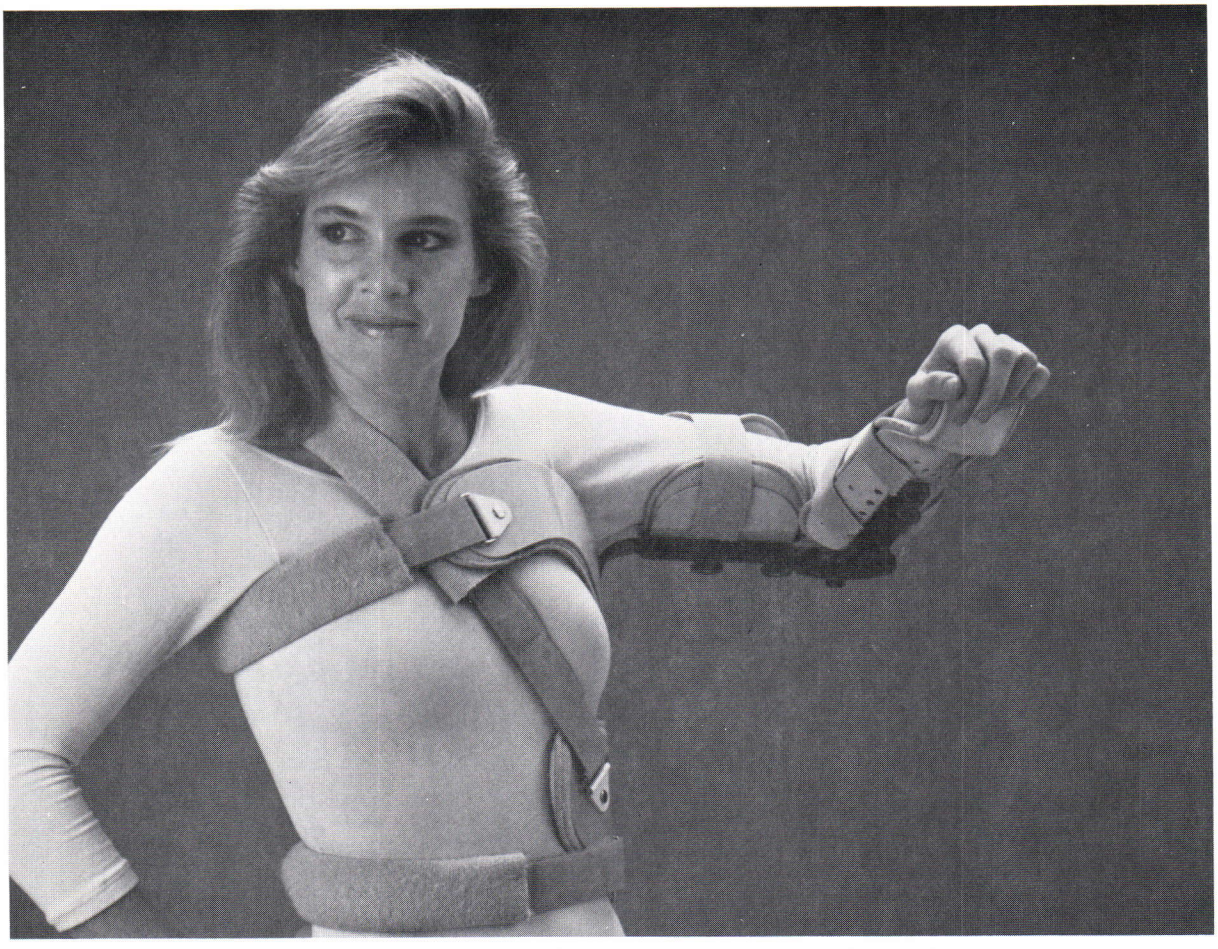
This is a very interesting book. It lists commonly used eponyms, gives a description of the word and a brief biographical sketch of the individual involved. As such it reads like a who's who of medicine. The information it imparts is useful and the biographical information give an image of the people involved and the times in which they lived.

Spinal Cord Injury, A Guide for Patient and Family

Lynn Phillips, Mark N. Ozer, Peter Axelson, and Howard Chizech. Raven Press, 1185 Avenue of the Americas, New York, New York 10036. 1987, 303 pages and index.

This book was written in cooperation with the Paralyzed Veterans of America and contains a good deal of useful information for those confronted with the catastrophic changes that paralysis can bring about. It goes beyond matters of medicine and health and covers such topics as transportation and recreation. While no book this size can cover such a broad and diverse topic area exhaustively, it is quite thorough and each chapter concludes with listings of suggested readings. The book also includes directories of rehabilitation and treatment centers and various organizations dedicated to self help. All in all, it contains a good deal of useful information for laymen and professionals alike.

Charles H. Pritham, C.P.O.



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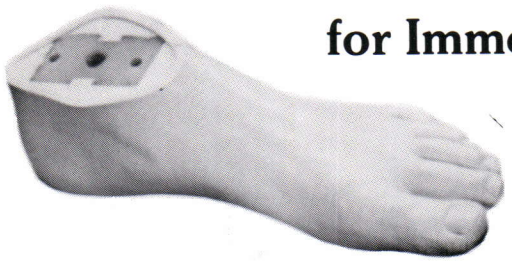

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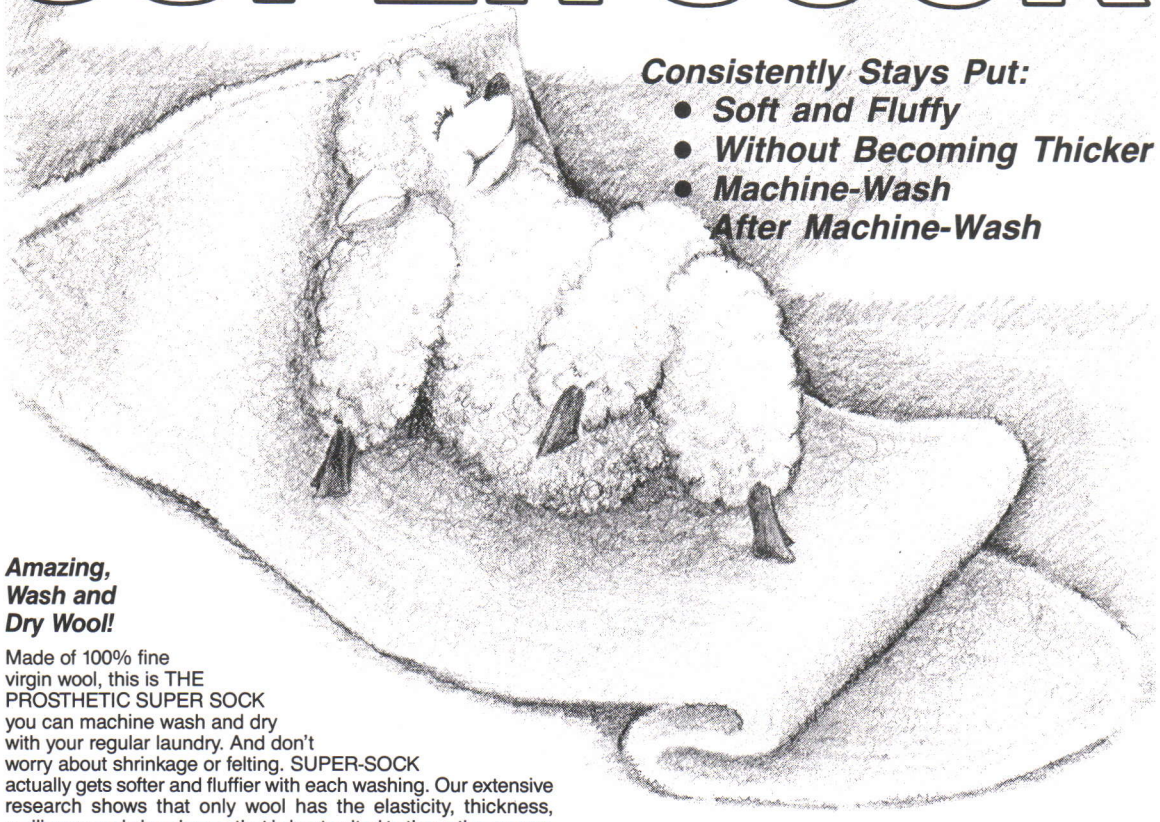

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