Winter 1987
Volume 40
Number 4

Orthotics and Prosthetics

Journal of the American Orthotic and Prosthetic Association

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Orthotics and Prosthetics

Winter 1987 **Volume 40, Number 4**

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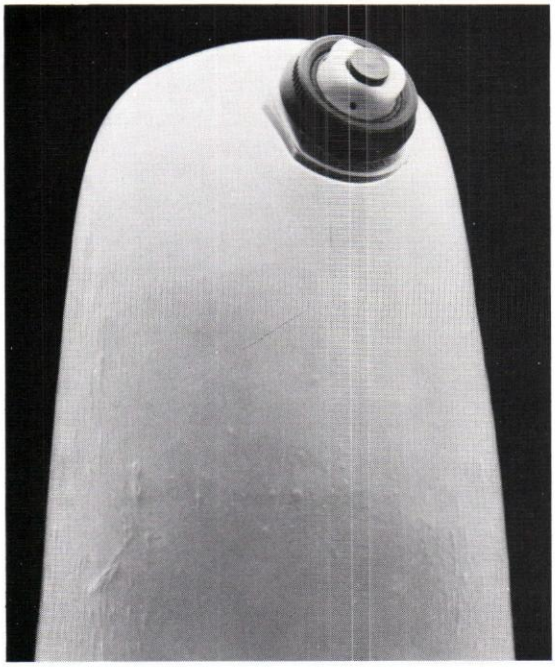

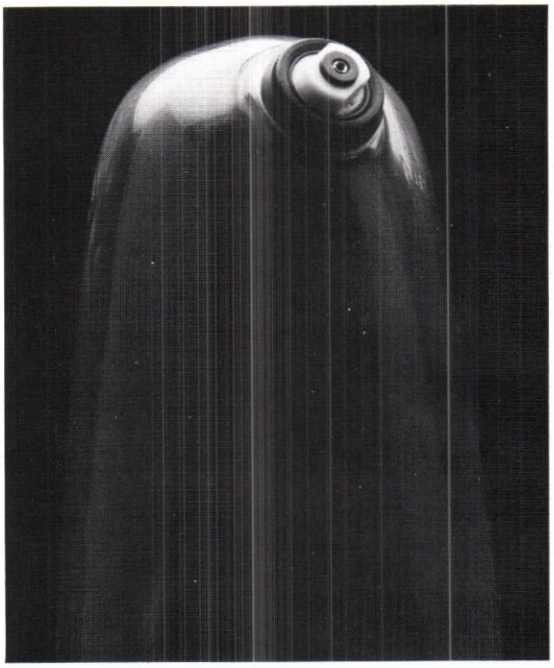

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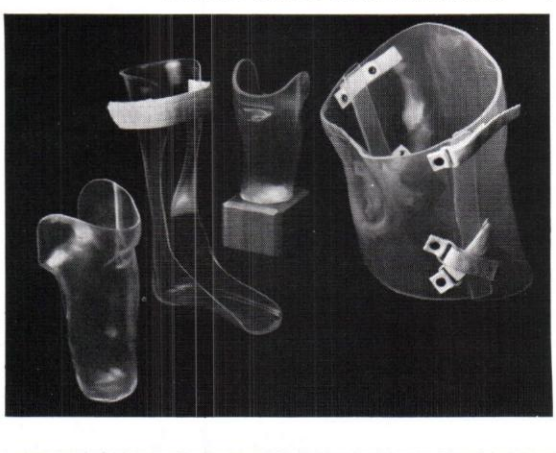

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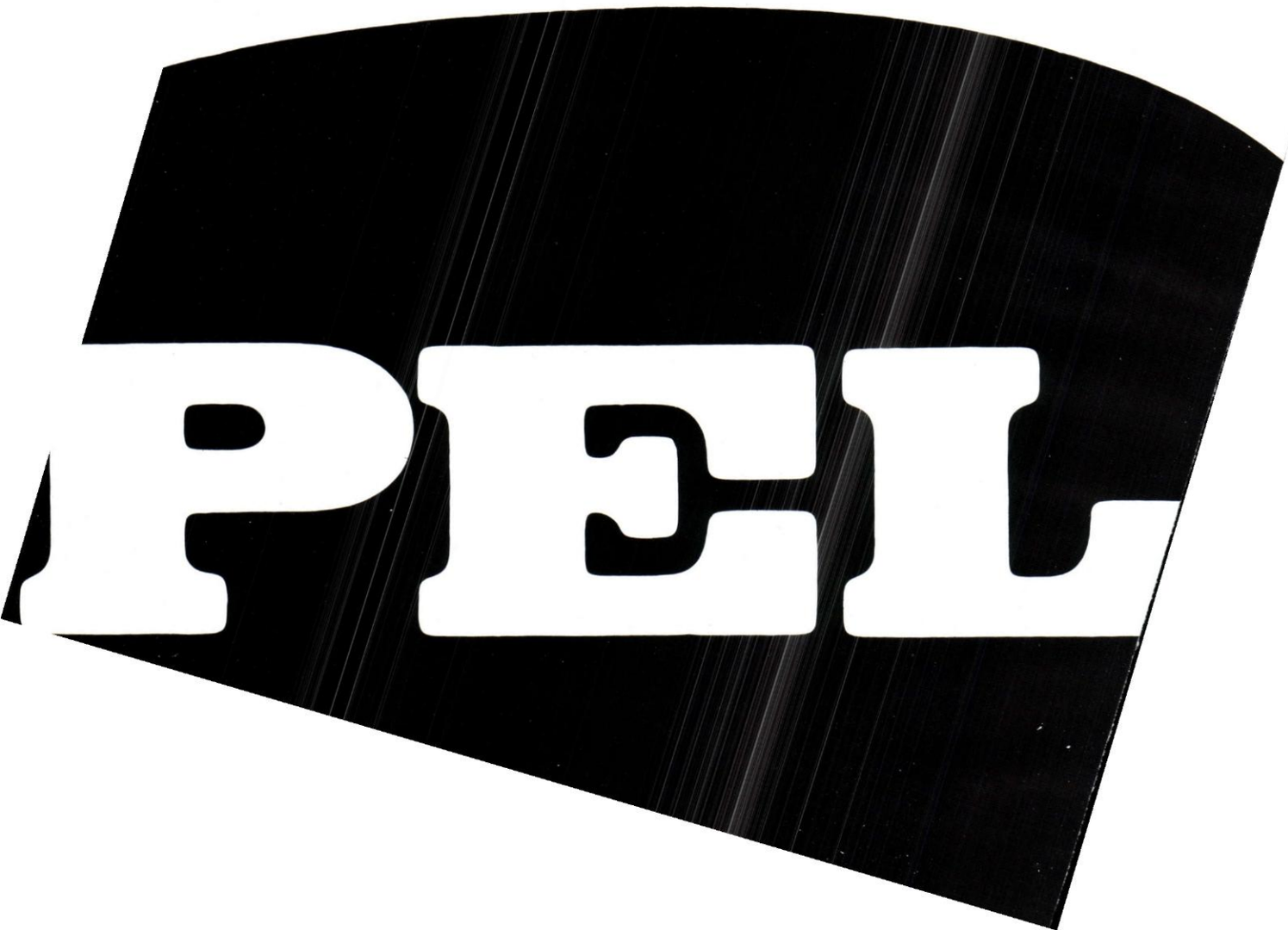

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Please notify the National Headquarters immediately concerning all meeting dates. It is important to submit meeting notices as early as possible. In the case of Regional Meetings, you must check with the National Headquarters prior to confirming date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

1987

February 15–22, Academy Annual Meeting and Scientific Seminar, Hyatt Regency Tampa, Tampa, Florida. Contact: Academy National Headquarters, 703-836-7118.

February 17–21, Academy Annual Meeting and Scientific Symposium, Hyatt Regency Tampa, Tampa, Florida. Contact: Academy National Headquarters, (703) 836-7118.

March 5–7, The 2nd Annual Symposium on Advances in Head Injury Rehabilitation, sponsored by the Dallas Rehabilitation Institute and Dallas Rehabilitation Foundation, Westin Galleria Hotel, Dallas, Texas. Contact: (214) 358-8440; in Texas (800) 441-9199, ext. 8440.

March 9–12, UCLA Total Surface Bearing Suction Below Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

March 16–25, UCLA CAT-CAM/Narrow ML Above Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

March 20–21, Oklahoma Association of Orthotics and Prosthetics, the Texas Association of Orthotists and Prosthetists, and Texas Chapter, American Academy of Orthotists and Prosthetists combined meeting, Dallas, Texas. Contact: Mike Allen, CPO, 2504 W. Ohio, Midland, Texas 79701; tel. (901) 683-5280.

March 25–27, NYU course, The ISNY Below-Knee Flexible Socket, New York, New York. brbontact: Registrar, Prosthetics and Orthotics, 317 E. 34th Street, New York, New York, 10016; tel. (212) 340-6686.

March 25–27, Hosmer Electric Systems Workshop and Seminar, Cleveland, Ohio. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; tel. (800) 538-7748 or (408) 379-5151.

March 30–April 2, UCLA Total Surface Bearing Suction Below Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

April 2–4, AOPA Region IV Annual Meeting, Stouffer's River View Plaza, Mobile, Alabama.

April 9–11, Charleston Bending Brace Seminar, Park Suite Hotel, Altamonte Springs (Orlando), Florida. Contact: Shannon Schwenn or Judy Clark, Charleston Bending Brace Seminars, P.O. Box 2031, Winter Park, Florida 32790; tel. (800) 327-0073 or (305) 645-0414.

April 13–22, UCLA CAT-CAM/Narrow ML Above Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

April 24–25, Spring Meeting, New York State Chapter of the American Academy of Orthotists and Prosthetists, Ramada Inn, Binghamton, New York. Contact: Bryan Finley, CP, Program Chairman, Binghamton Limb and Brace Co., Inc., 142 Harry L. Drive, Johnson City, New York 13790; tel. (607) 797-1246.

May 4–13, UCLA CAT-CAM/Narrow ML Above Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

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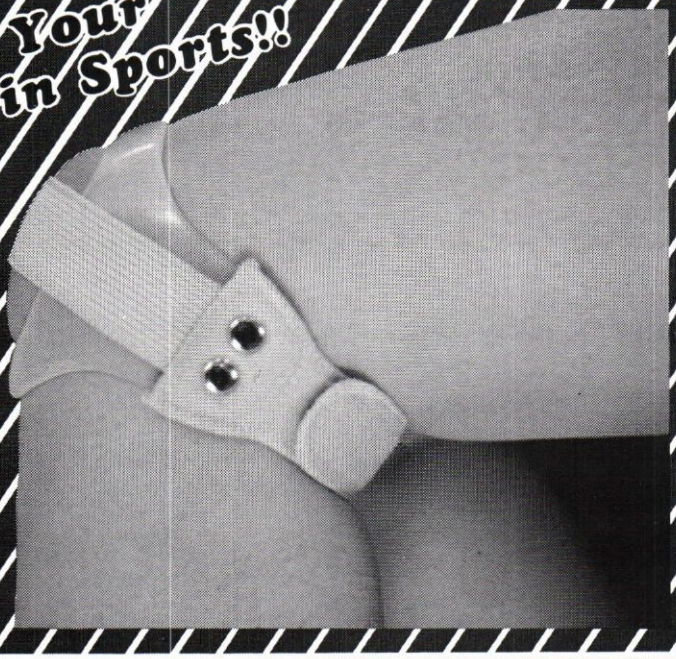

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May 7–10, AOPA Regions II and III Combined Annual Meeting, Hershey, Pennsylvania.

May 7–10, SinoMed '87, an international exhibition and conference program, Shanghai Exhibition Centre, Shanghai, People's Republic of China. Contact: Kallman Associates, Five Maple Court, Ridgewood, New Jersey 07450; tel. (201) 652-7070.

May 13–15, Hosmer Electric Systems Workshop and Seminar, Memphis, Tennessee. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; tel. (800) 538-7748 or (408) 379-5151.

May 21–23, NYU course, the ISNY Below-Knee Flexible Socket, New York, New York. Contact: Registrar, Prosthetics and Orthotics, 317 E. 34th Street, New York, New York 10016; tel. (212) 340-6686.

May 27–29, NYU course, The Narrow ML Above-Knee Socket, New York, New York. Contact: Registrar, Prosthetics and Orthotics, 317 E. 34th Street, New York, New York 10016; tel. (212) 340-6686.

May 27–30, UCLA Total Surface Bearing Suction Below Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

May 28–31, AOPA Region V Annual Meeting, Grand Traverse Hotel, Traverse City, Michigan.

June 3–5, NYU course, The ISNY Below-Knee Flexible Socket, New York, New York. Contact: Registrar, Prosthetics and Orthotics, 317 E. 34th Street, New York, New York 10016; tel. (212) 340-6686.

June 4–6, Annual Meeting of the Association of Children's Prosthetic-Orthotic Clinics, Vancouver, British Columbia. Paper submissions: Francis J. Trost, M.D., Program Chairman, 2545 Chicago Avenue, South, Minneapolis, Minnesota 55404. Registration: Sidney Fishman, Ph.D., c/o NYU PGMS, 317 E. 34th Street, New York, New York 10016. Information: Yoshio Setoguchi, M.D., Child Amputee Prosthetics Project, UCLA Rehabilitation Center, 1000 Veteran Avenue, Room 25-26, Los Angeles, California 90024.

June 5–7, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting, Doubletree Inn, Monterey, California.

June 8–17, UCLA CAT-CAM/Narrow ML Above Knee Prosthetics Course, Los Angeles, California. Contact: Timothy B. Staats, MA, CP, UCLA POEP, Room 22-46, 1000 Veteran Avenue, Los Angeles, California 90024.

June 10–13, AOPA Regions VII, VIII, X, and XI Combined Annual Meeting, Fairmont Hotel, Dallas, Texas.

June 18–21, AOPA Region VI and the Midwest Chapter of the Academy Combined Annual Meeting, Embassy Suite, Indianapolis, Indiana.

June 19–23, RESNA '87, the 10th Annual Conference on Rehabilitation Technology, San Jose, California. Contact: RESNA, Suite 700, 1101 Connecticut Avenue, NW, Washington, D.C. 20036; tel. (202) 857-1199.

June 24–27, Tenth INTERBOR Congress, Barcelona, Spain. Contact: José Ma Camós, Secretary of the Congress, Grau Soler, Buenos Aires, 52, Argentina.

July 5–10, International Conference on Disability Education, Jerusalem, Israel. Contact: Israel Rehabilitation Society, 18 David Elazar Street, Tel Aviv 61901, Israel.

July 12–16, International Conference of Rehabilitation Journalists, Jerusalem, Israel. Contact: Israel Rehabilitation Society, 18 David Elazar Street, Tel Aviv 61901, Israel.

September 6–10, International Seminar on Prosthetics and Orthotics, Dan Accadia Hotel, Herzliya, Israel. Contact: ISPO 1987, P.O. Box 50006, Tel Aviv 61500, Israel; tel. (03) 654 571; TELEX: 341171 KENS IL, Fax: 972 3 655674.

September 11–12, Ohio Orthotics and Prosthetics Association/Ohio Chapter, American Academy of Orthotists and Prosthetists combined meeting, "Bridging the Profeseion," Dayton, Ohio. Contact: Norma Jean Finissi, Executive Di-

rector, O.O.P.A./Ohio A.A.O.P., 4355 North High Street, #208, Columbus, Ohio 43214; tel. (614) 267-1121.

September 21–27, AOPA Annual National Assembly, Hyatt Regency Hotel, San Francisco, California. Contact: AOPA National Headquarters, (703) 836-7116.

September 28–30, Hosmer Electric Systems Workshop and Seminar, Hosmer Dorrance Corporation, Campbell, California. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; tel. (800) 538-7748 or (408) 379-5151.

November 11–13, Hosmer Electric Systems Workshop and Seminar, Hosmer Dorrance Corporation, Campbell, California. Contact: Catherine Wooten, Hosmer Dorrance Corporation, 561 Division Street, Campbell, California 95008; tel. (800) 538-7748 or (408) 379-5151.

November 26–29, Medical/Hospitech 87, Bangkok International Exposition Center, Bangkok, Thailand. Contact: SHK International Services Ltd., 22/F. Tian An Centre, 151 Gloucester Road, Hong Kong.

1988

January 25–31, Academy Annual Meeting and Scientific Symposium, Newport Beach Marriott Hotel and Tennis Club, Newport Beach, California. Contact: Academy National Office, (703) 836-7118.

February 4–9, American Academy of Orthopedic Surgeons Annual Meeting, Atlanta, Georgia.

May 13–15, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting.

May 19–21, AOPA Region V Annual Meeting, Charleston, West Virginia.

June 9–11, AOPA Regions II and III Combined Annual Meeting.

June 14–18, AOPA Regions VII, VIII, X, and XI Combined Annual Meeting, Seattle, Washington.

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.

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

1989

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

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.

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

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

1990

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

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

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

Reprinted below is the application required by ABC for those wishing accreditation for their education programs. Please use this form when applying for credits.

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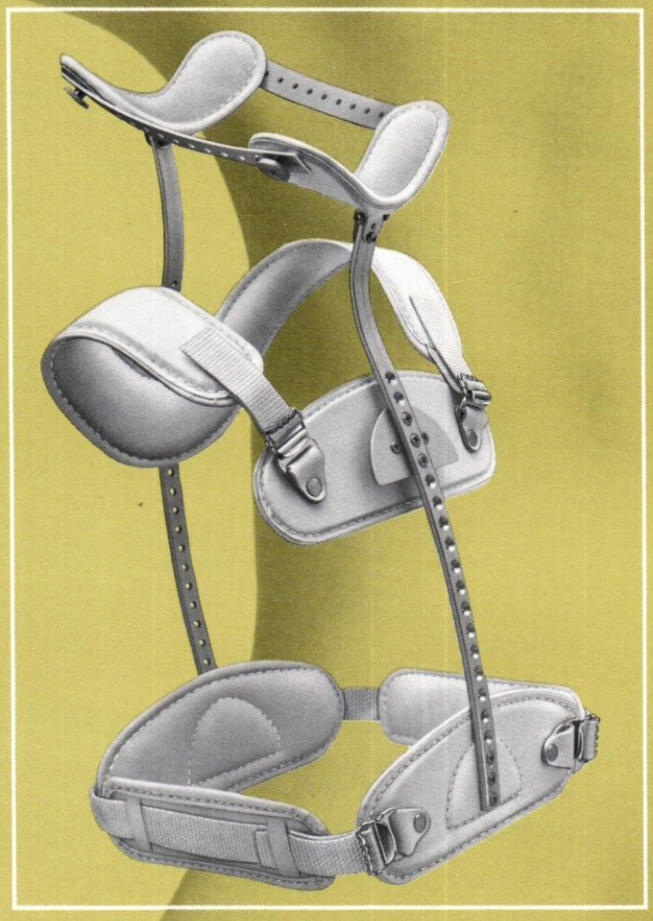

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Case Study: The Use of Bail Lock Knee Joints in the Rehabilitation of the High Level Spinal Cord Injured Below-Knee Amputee

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INTRODUCTION

Prosthetic treatment for a Spinal Cord Injured (SCI) patient has always been, at best, considered to be only cosmetic in design. But when considering the total concept of treatment requirements of the SCI patient, the necessity for standing, simple transfers and the daily maintenance of bodily functions, a prosthesis can be of great value in preventing bone deterioration and assisting in bowel and bladder functions. When viewing the total rehabilitation process of this SCI patient, and removing primary emphasis on ambulation and cosmesis, prosthetic treatment requirements take on a whole new process.

CASE HISTORY

On March 3, 1985, this facility was presented with a 30-year old male, six feet tall, weighing 153 pounds, who had multiple trauma as the result of electrical shock from contacting a high powered wire in early October, 1984. The patient exhibited C7 Myelopathy, incomplete, with neurogenic dysfunction of bowel and bladder. He also had a below-knee amputation on the right resulting from infection secondary to electrical burns, and a partial foot with loss of the first and second toes of the left foot due to injuries received from the exit of the electrical charge.

Upon examination, the patient demonstrated good upper extremity strength, except for weak hand intrinsics. A good grip was exhibited, and it was noted that, because of upper extremity strength, future attempts at ambulation in parallel bars with assistance could be attempted.

The below-knee amputation, performed November 6, 1984, was well healed and of a proper condition for prosthetic consideration. However, a great degree of knee flexor spasticity was present, requiring large amounts of medication. A 90q flexion contracture manually reducible to 45q was also measured. The contractures were of sufficient strength and duration as to pull the patient completely out of a wheelchair. At first, attempts were made to reduce the right below-knee contracture by using a soft knee immobilizer, with no success. A plywood seatboard with an extension was fabricated to maintain extension of the

right below-knee amputation during sitting. Even with four inch foam padding, the rigid surface stimulated the contractures and was withdrawn, and similar treatments disregarded. Serial casting was suggested and accepted on condition that the patient would be closely monitored due to the lack of skin sensitivity, resulting from the level of injury. After initial application of the plaster negative mold, the patient complained of moderate discomfort. The cast was bivalved, and Velcro® adhesive straps were attached for frequent self-examination by the patient.

After two cast changes, results were considered minimal, and the contracture seemed to be fixed. At that time it was hoped that there was no calcification of the joint, and that the contracture was of soft tissue only. At that point, an above-knee amputation was considered to improve and lengthen sitting time and ability. After referral to the Rehabilitation Medicine department for consideration of phenol blocks to limit spasticity, the contracture was measured and determined to be set at 40q. The Rehabilitation Medicine physicians suggested surgical release of contracture after x-rays showed no bony involvement. After evaluation by and consultation with the General Surgery department, the patient himself requested surgical release of the contracture, in hopes of becoming a candidate for prosthetic treatment.

Orthotic and prosthetic lab personnel counseled the patient at great length in an attempt to educate him to the many possibilities and various consequences affecting any degree of prosthetic treatment of SCI patients. After lengthy discussions, it was felt that the patient's expectations were realistic, and a general surgeon released the contracture on July 11, 1985.

The Semi-Membranosus was isolated and resected, as was the Gracilis. The sartorius and biceps femoris were both sectioned. After completion of the surgical procedure, a 30° contracture remained. But the surgeon anticipated full extension after agressive physical therapy.

The Physical Therapy department continued treatment of the patient after surgery with short term goals consisting of strengthening the lower extremity muscles and lengthening standing time on the Blatnik table. With his strength increasing, the patient was referred to the orthotic and prosthetic lab for formulation of a treatment regimen.

ORTHOTIC AND PROSTHETIC TREATMENT

In order to enable the patient to stand with minimal assistance on the Blatnik table, and to prevent any future contracture, he was fit with an ankle-foot orthosis (AFO) with metal bilateral uprights and double action ankle joints on the left. A continual reduction of the right below-knee flexion contracture was also noted at this time. Two weeks later, the patient was transferring to and from his wheelchair, using a slide board with minimal assistance.

One month after fitting of the AFO in preparation for prosthetic care and in anticipation of gait training, the patient was fit with a knee-ankle-foot orthosis (KAFO) on the left lower extremity. The KAFO used bilateral steel uprights with Becker* bail locks and limited motion ankle joints, a full thigh corset, and a soft foam filler for the partial foot amputation.

Simultaneously, the patient was also fit with a PTS temporary below-knee prosthesis (Figure 1). The temporary limb was constructed using standard impression methods and modification principles. It consisted of a vacuum formed Surlyn® socket with a United States Manufacturing** adjustable below-knee Pylon, SACH foot, distal pour pad, full thigh lacer, and Becker bail lock knee joints (Figure 2). Using the PTS design with the full corset combination, we achieved a high level of suspension and medial lateral stability at the knee for this SCI patient.

*Becker Orthopedic Appliance Co., 1776 South Woodward Avenue, Birmingham, MI 48011.

**United States Manufacturing Co., 180 N. San Gabriel Blvd., P.O. Box 5030, Pasadena, CA 91107.

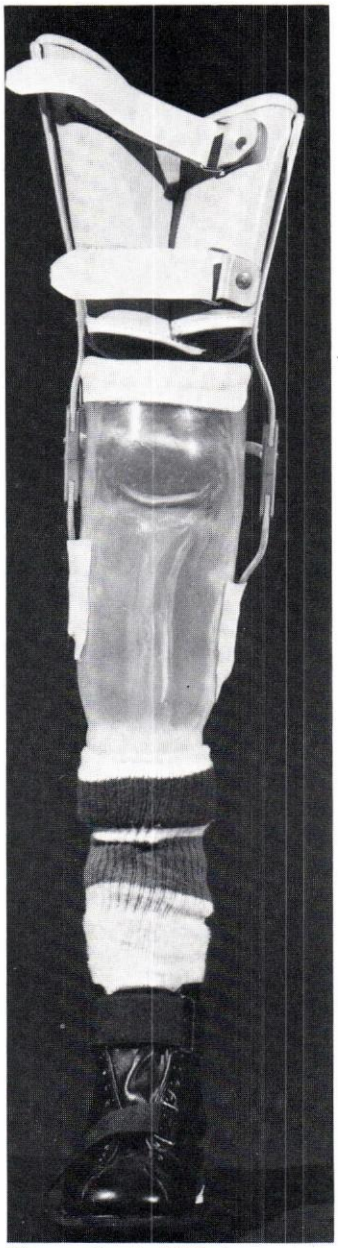

Figure 1. A PTS temporary below-knee prosthesis.

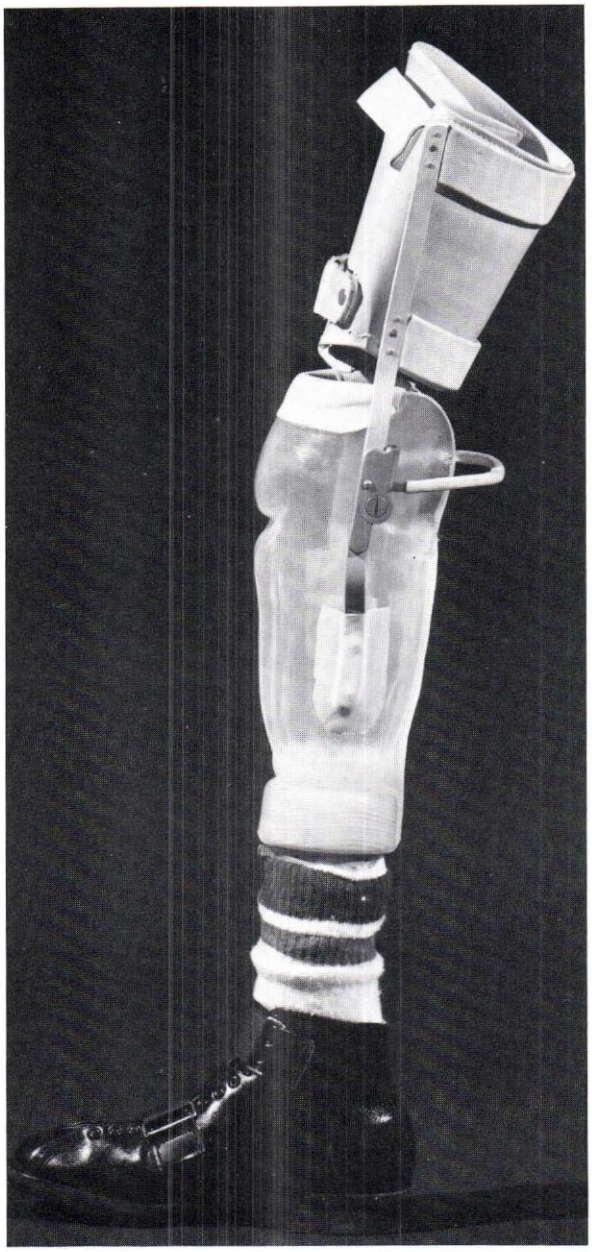

Figure 2. The PTS prosthesis consisted of a vacuum formed Surlyn® socket with a USMC adjustable below-knee pylon, SACH foot, distal pour pad, full thigh lacer, and Becker bail lock knee joints.

RESULTS

The patient accepted the orthotic and prosthetic devices and aggressively pursued physical therapy. Long range goals were to prepare the patient for a definitive prosthesis and to achieve an acceptable amount of independent transfers, in addition to lengthening the period of time in standing and ambulatory exercise. Functional gait was not considered at this time and stress was placed on those protocols which were required for the SCI patient. Although prosthetic treatment of an SCI patient is usually approached with a large degree of caution, this patient experienced minor problems during the fitting and follow-up periods of prosthetic care. However, aggressive physical therapy, extremely high patient motivation, and daily follow-up by orthotic and prosthetic lab personnel all contributed to the ability of a high level SCI amputee to take his first steps on the path of his individualized rehabilitation program.

CONCLUSION

When considering prosthetic treatment of the SCI patient, caution must be exercised, due to the inherent problems of lack or total absence of muscle activity and skin sensitivity. However, with high levels of patient motivation and total commitment of in-house rehabilitation staff, these difficult cases can be attempted with much more success than historical precedents would indicate.

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Case Study: Management of a Severely Involved Cerebral Palsy Patient

Lawrence R. Lange, C.P.O.

INTRODUCTION

D.S. was a four year old cerebral palsy victim. Prior to orthotic care, he exhibited the following physical characteristics (Figures 1 and 2):

1. Marked lumbosacral lordosis
2. Hip and knee flexion contractures
3. Achilles tendon tightness
4. Extensor synergy pattern during ambulation

While attempts were made, prior to seeing us, to reduce contractures, he was measured and found to have 10° to 15° flexion contractures bilaterally at the knees. He also had five to ten degree flexion contractures at the hips. The one promising aspect was the apparent lack of marked spasticity seen in this child.

A prescription was developed in conjunction with the patient's occupational therapist and physician for a molded plas-

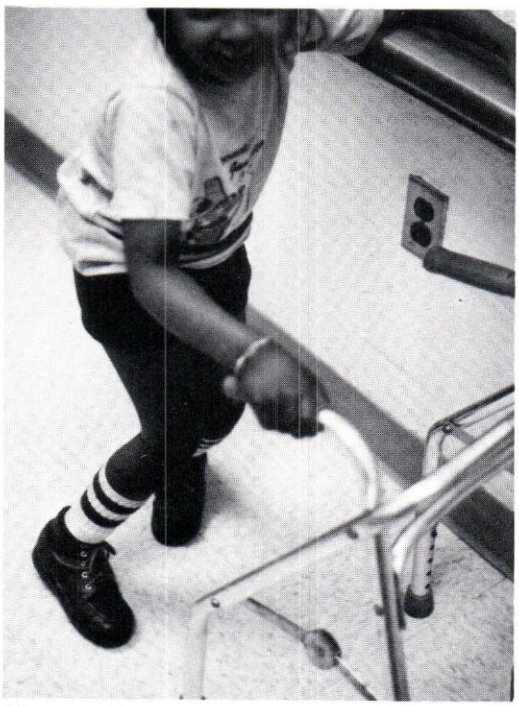

Figure 1. The patient ambulating prior to orthotic treatment viewed obliquely. Note the synergy pattern and lack of stability.

Figure 2. Lateral view of the patient prior to treatment.

tic TLSHKAFO with Becker lightweight aluminum dial lock knee joints.* The most difficult portion of the orthotic treatment revolved around the patient's physical therapy. In order to work on this patient's synergy pattern, it was necessary that there be no rigid hip control joints or band. This required a non-standard approach to his orthotic treatment.

MEASUREMENT

Casting of the KAFO's was done first with the patient in a supine position on the examination table. This was done to enable us to see the effects of locking the knees and ankles, and also to have the patient standing upright for the casting of the TLSO portion of the orthosis. The foot was cast first. It was positioned in slight inversion and as close to a neutral flexion angle as possible. The inversion allowed the feet to be flat on the ground when the thighs were held in the proper amount of abduction. The thigh was then cast, leaving a gap over the knee. The knee was molded last, so the joint could be extended by applying force posteriorly to the proximal and distal portions of the negative mold. We found that this technique reduces distortion to a minimum.

After both legs were molded and the negative models hardened, the patient was placed upright and cast in the traditional manner for the TLSO. It was noted at this time that once the legs were held in normal alignment, without knee and ankle flexion abnormalities, there was reduced hip flexion, and the patient was able to stand virtually upright with little assistance.

MODIFICATIONS AND FABRICATION

Build-ups were made over all bony prominences on the KAFO portions. Slight medial and lateral flattening was done on the thigh and calf areas. The standard

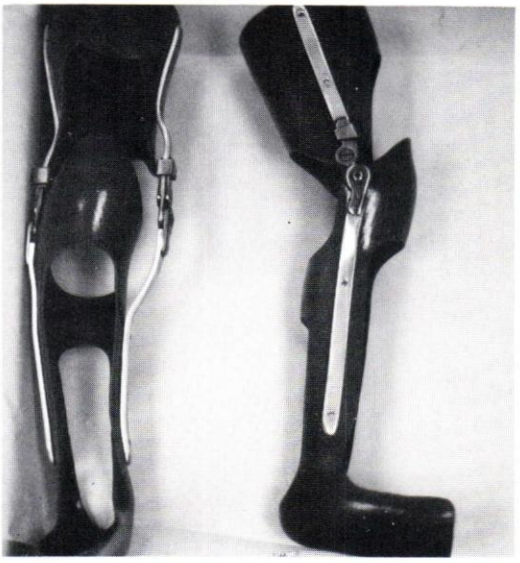

Figure 3. Anterior view of the TLSO component to the system. Note the elastic extension, abduction straps and the articulated posterior section.

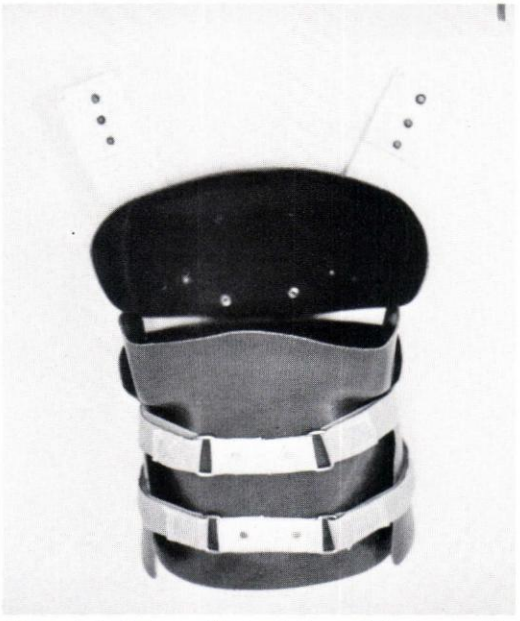

Figure 4. Anterior and lateral views of the knee ankle foot orthoses demonstrating suprapatellar portion and dial lock knee joints.

*Becker Orthopedic Appliance Company, 635 Executive Drive, Troy, Michigan 48083.

TLSO modifications were done on the positive model, with the exception of the lower posterior distal trimlines.

The orthoses were made of lightweight acrylic laminate. The TLSO was fabricated using techniques borrowed from J. Glancy, C.O.[1] The KAFO's were trimmed to allow as much support and comfort as possible (Figures 3 and 4). The latter point was important as the patient had a concurrent skin condition, and perspired very little except on the soles of his feet. All trimlines were adjusted with the patient present. The goal was to ease donning and doffing as well as to achieve all comfort requirements. Below-knee cuff suspension studs and elastic straps were placed in the distal posterior section of the TLSO. The purpose of these straps was to assist in hip extension and abduction by attachment to the thigh sections. As we had seen during measurement, once the knee and ankle portion of the extensor synergy pattern were locked out, the hips were mobile without any apparent spasticity. This allowed the elastic control straps to function and not aggravate the condition. Contoured suprapatellar trimlines assisted in limiting knee flexion and, once the TLSO was in place, lordosis was reduced. The combined effect of the total orthosis was the production of a more upright posture (Figures 5 and 6).

CONCLUSION

It was noted that once the orthoses were removed, the patient remained upright for several steps. The orthoses are designed to hinder any musculoskeletal deformity while also assisting to train the patient in

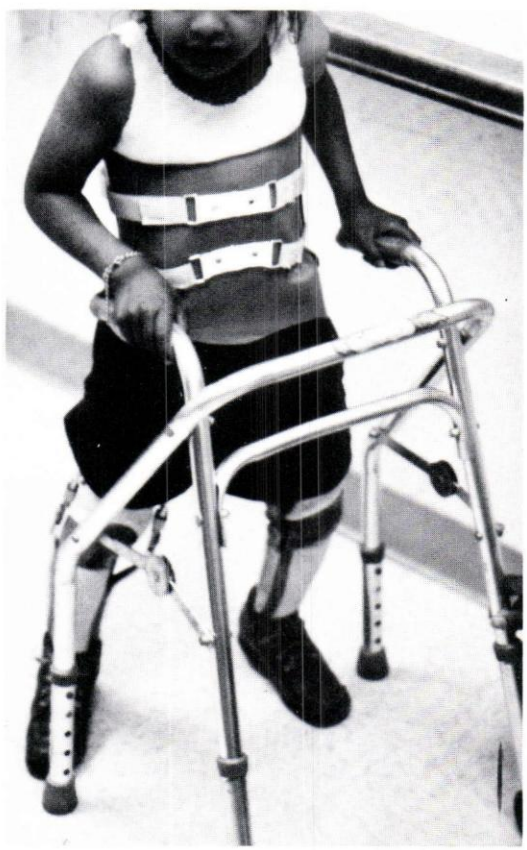

Figure 5. Oblique view of the patient following orthotic treatment demonstrating more upright posture.

Figure 6. Lateral view of patient following orthotic care.

more appropriate swing and stance phase gait. Perhaps, in the future, he will require little, if any, orthotic management as his condition appears to be one of delayed development as opposed to retarded development.

The important features of the orthosis are (Figures 3 and 4):

1. Lightweight, cosmetic construction
2. Control from level of xyphoid to foot
3. Control of hip joint without use of conventional metal hip joints and locks
4. Suprapatellar trimlines to control knee flexion contractures, along with usage of dial lock knee components

ACKNOWLEDGMENTS

I would like to express gratitude in preparation of this case report to the following: E. Douglas Bourgoyne, C.P.O., Lynn Skied, O.T., Joseph M. Cestaro, C.P.O.

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Full Cuff Control

Henry J. Richter, B.S., C.P.

INTRODUCTION

In times where choices are not only welcomed, but literally demanded by patients and rehabilitation teams alike, we are now seeing many exciting and useful new terminal devices and assistive devices for upper extremity amputees. In the fabrication of prostheses, we have been afforded various options in socket design, construction material, and hardware, in order that we may choose the variations most suitable to our patients' needs, desires, and lifestyles.

There have been only two popular options for activation of below-elbow terminal devices—either electric (i.e. myoelectric or switch control), or body powered (i.e. cable and harness). The asset of an electrical control system is its elimination of the irritating and restricting harness, as well as the actuation of the terminal device regardless of the attitude of the sound shoulder or involved humerus and forearm. Some less desirable traits of the electrical control system include weight, reliability, and expense.

The body powered appliances, however, maintain popularity because of their padded sock fits, light weight, ruggedness (less moving parts), and affordability. Their two obvious shortcomings are what make the electrical system an option. They require restrictive harnesses, commonly accentuating uncomfortable pressure on the seventh cervical vertebra and the sound side axilla area. They also require the amputee to maintain his/her shoulders and involved arm in a manner of mechanical advantage in order for the control line to operate the terminal device. Admittedly, supracondylar below-elbow appliances do eliminate some harnessing, but still involve the sound side shoulder. The need for the aforementioned mechanical advantage and a solution for the definite problem with clothing, if worn under a fitted shirt or blouse, are still present.

DISCUSSION

Circumferential cuff ("Full Cuff") control, as applied to body powered prosthetics, is the focus of this discussion. The cuff mechanism works on the same principal as the harness system, in that both can be described as having constant length control lines traveling across active joints; the difference is in simplicity. With the traditional below-elbow harness arrangement, the control line is typically activated through scapular abduction, gleno-humeral flexion and/or forearm extension. The cuff system concentrates on forearm extension only (Figure 1). In this instance, the fewer the variables to consider, the more direct, positive, and reliable the system.

Pivoting on two Hosmer-Dorrance forearm lift posts, each mounted over an epicondyle, the cuff is placed in the same relative position as the typical triceps cuff used with a "Figure 8" harness. These posts may be laminated in position or riveted onto any existing supracondylar below-elbow socket. The cuff plays little, if any, role in suspension; rather, it is the anchor point for the control line itself. The customary

action of the forearm extension for control, as well as the not-so-customary flexion for control, may be employed (note: any terminal device available for traditional body powered prostheses will function with this system).

FABRICATION

The cuff material of choice, at this time, is ⅛" thickness Ortholen®. This has been chosen for its combination of flexibility, strength, and workability. One must remember that not only is the ease of circumferential adjustment crucial after donning and before doffing, but that the integrity of the pivot point must be maintained as well. For these reasons, rigid vertical uprights on a flexible cuff are essential. Although the hinges have little to do with the function of the cuff and attached control line, they have everything to do with cuff placement. Flexible hinges will allow migration of the cuff under load, regardless of the tension, and can only lead to loss of function after only a few articulations.

Residual limb measurements, moulding, and cast modifications are identical to those of accepted supracondylar design. The cuff design and fabrication, on the other hand, is quite different. Considerations for an anterior overlap and underlap should be made, and a positive foundation for the control line, making for the best control action, is planned. Deformation of the anterior cuff, away from the arm, during activation can result in a spongy, unreliable opening of the terminal device. Overlapping will resist such deformation if the control line base plate is mounted on the lateral, underlapping, flap and reinforced by the slotted, overlapping, medial flap. This arrangement allows one to release the Velcro® brand fastener secured medial flap when donning or doffing, easily snugging the cuff to the desired fit once the patient is in the socket.

Measurement of the humeral area is the same as for a traditional triceps cuff. Proximal and distal circumferential measurements along with a height measurement are required. The height of this cuff does play an important role in terms of energy transfer. As the terminal device is activated, control line tension is transferred to the posterior cuff, in the form of pressure, and absorbed by the soft tissue in the triceps area of the arm. Obviously, the more pinch force required by the user, the greater the pressure compensating area required of the cuff. It is interesting to note that the cuff pressure is absorbed by the very muscle group, triceps brachii, initiating this energy transfer through their actions of elbow extension. As these muscles fire and become more tense, they are more resistant to pressure and displacement by the cuff. A broad posterior cuff foundation is necessary for positive and calculable terminal device control. The height of the anterior cuff is less crucial, therefore, a height of 2"-2¼" successfully

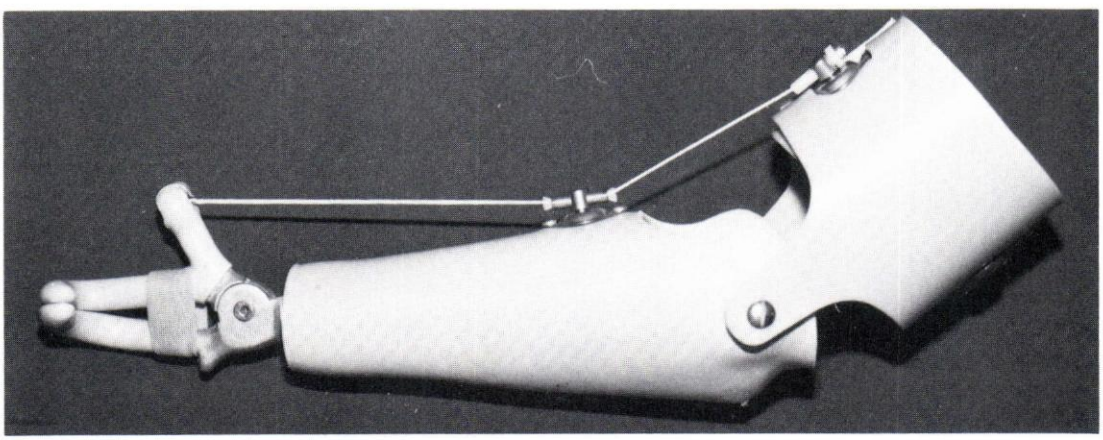


Figure 1. Extension controlled voluntary opening terminal device.

accommodates the 1¼" base plate and two-inch wide Velcro® adjustment band. However, one area of concern in the anterior cuff is in the inferior border. The patient should be able to flex his/her elbow to the same degree, without cubital fold pinching, as she/he would in a supracondylar socket alone.

On folded paper, at a distance equivalent to the height of the posterior cuff desired, mark two horizontal lines representing the proximal and distal edges (Figure 2). Average the proximo-distal humeral circumferences and extend the horizontal lines to half of this value. At midline of this folded tracing, draw a descending vertical line to represent the elbow joint upright. This line should be more than adequate in length, as it can be trimmed during fitting. The uprights should be approximately one inch wide (½" on each side of vertical line). Extend the tracing 1¼"-1½" to create the anterior over- and underlapping flaps.

Note that at this point, the inferior border of the "flap" tracing is raised to provide relief in the cubital fold of the elbow during flexion. One inch is a good starting value. More may be taken away on larger cuffs, but recall the two inch minimum when fabricating smaller cuffs.

Radius the corners of the uprights into the cuff, blending the anterior edge of the upright as one smooth arc into the inferior border of the flap extension. Unfold the tracing onto a ⅛" Ortholen® plastic sheet. Finish the edges with a deburring tool and propane torch glazing (Figure 3).

FITTING

In fitting, we are looking for the hanging angle and range of motion (ROM) characteristics and fit similar to that found in typical supracondylar sockets. Place the cuff around the arm in the relative location selected during the measurements. Mark the inside of the cuff uprights as they lie over the pivot posts (forearm lift posts) in both extension and flexion. Between these marks, center and drill a ¼" hole and mount the cuff (Loctite® or a similar adhesive is advised once a permanent installment is established). Although the cuff tension is a preference adjustment, one should check that it doesn't impinge upon the biceps bunching during full elbow flexion.

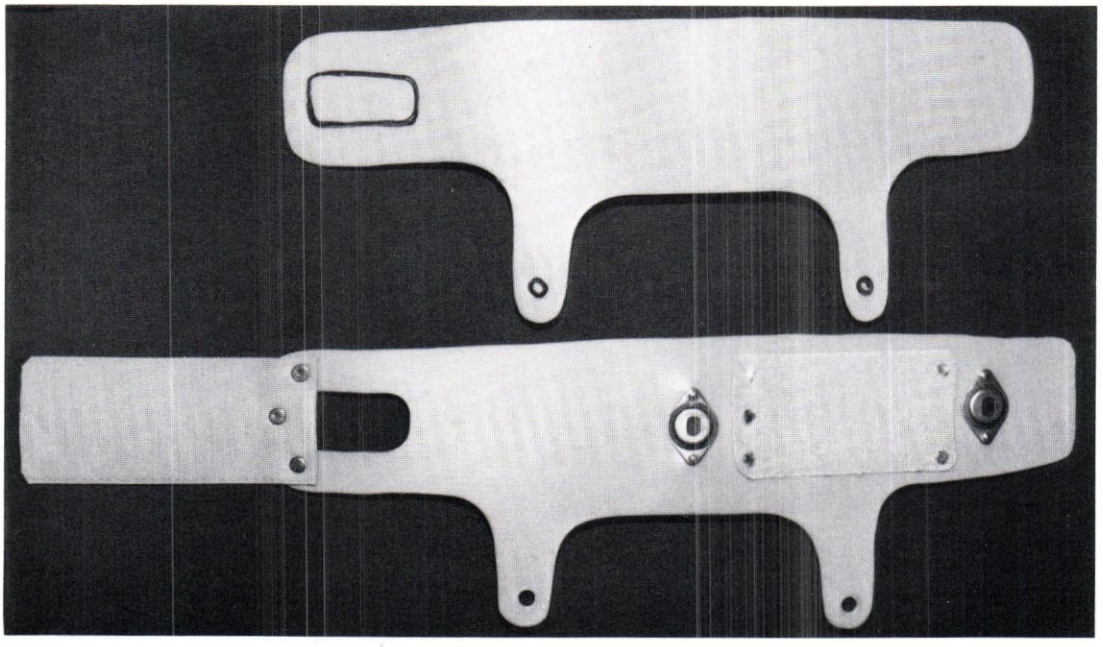


Figure 3. Cuff layout and finished cuff including hardware, slotted flap, and Velcro.®

CUFF LAYOUT (example)

Proximal Humeral Circumference	**10⅛″**
Distal Humeral Circumference	**9⅞″**
Averaged Humeral Circumference	**10″**
Posterior Cuff Height	**4″**

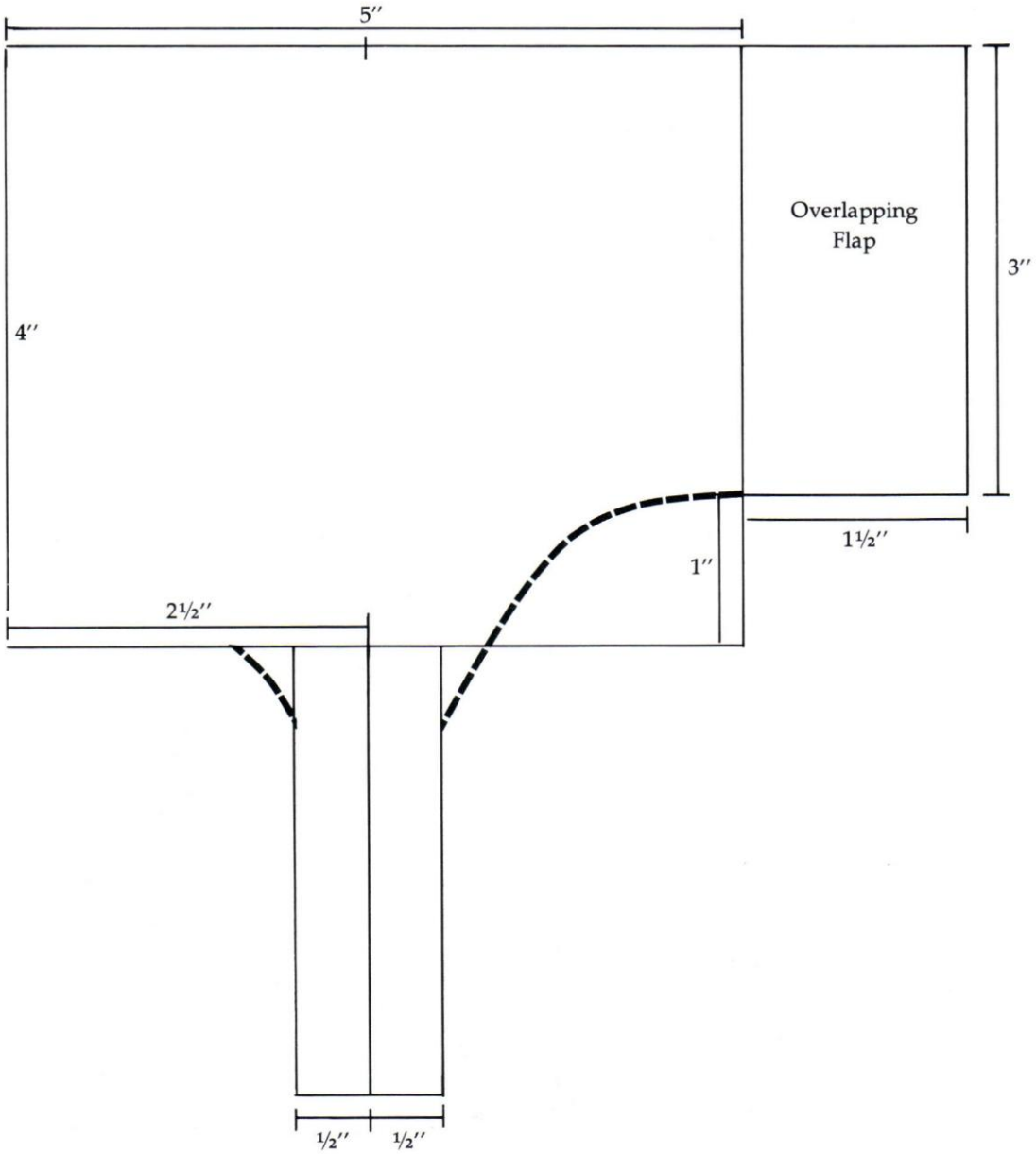


Figure 2.

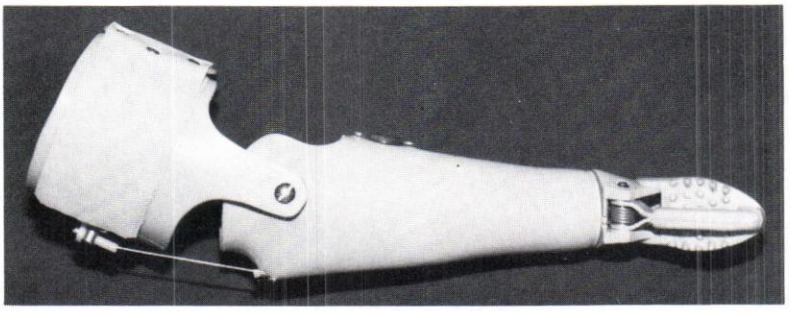
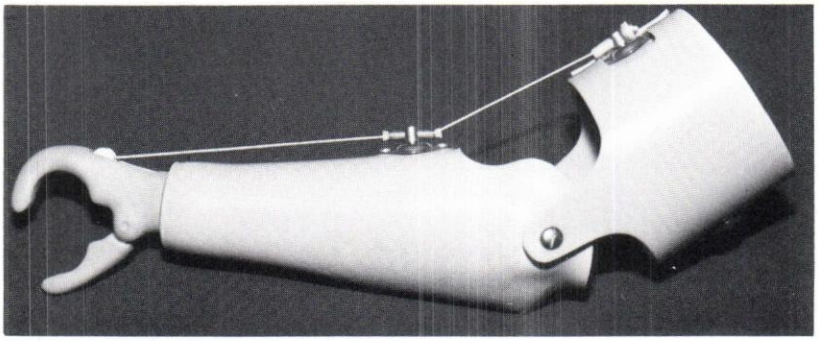


Figure 4A. Flexion controlled voluntary opening terminal device.

Figure 4B. Extension controlled voluntary closing terminal device.



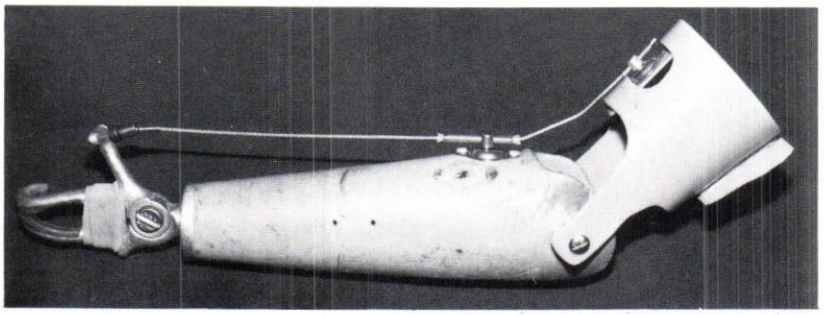
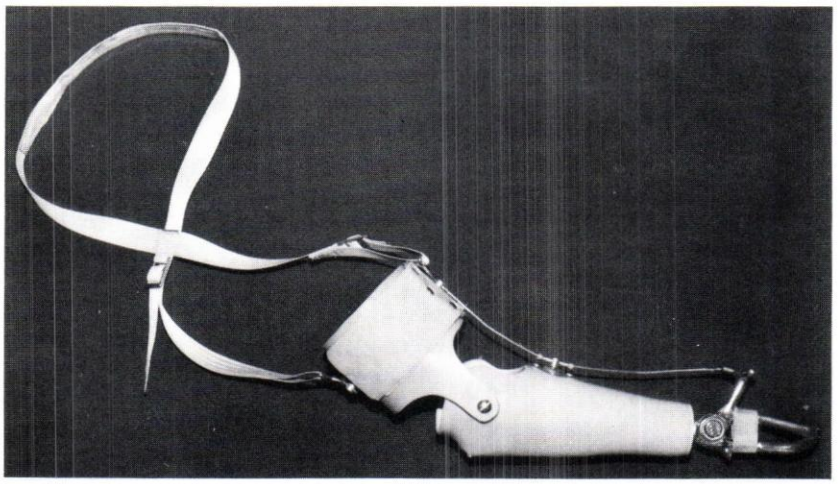


Figure 5. Adaptation of "Full Cuff" control to a pre-existing prosthesis.

Figure 6. Quick connect "Figure 8" harness for heavy axial loading.



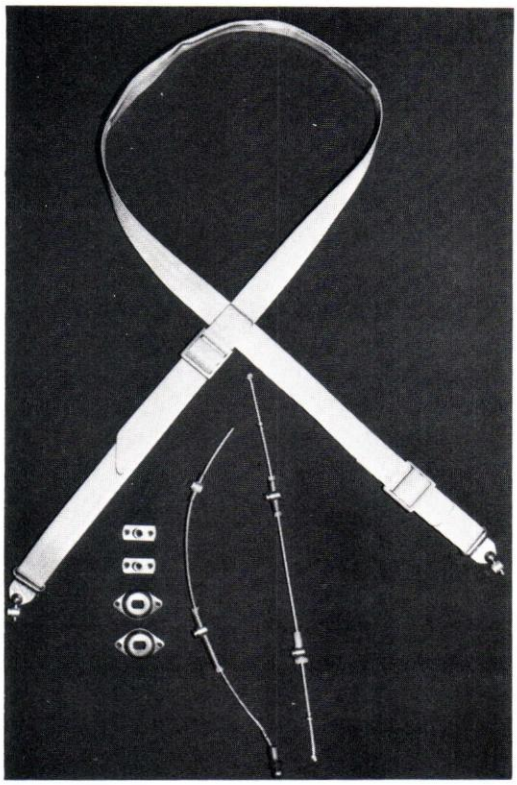


Figure 7. Sample harware: Dacron harness with retainer connections, nylon and wire control lines, base plates, and forearm lift posts.

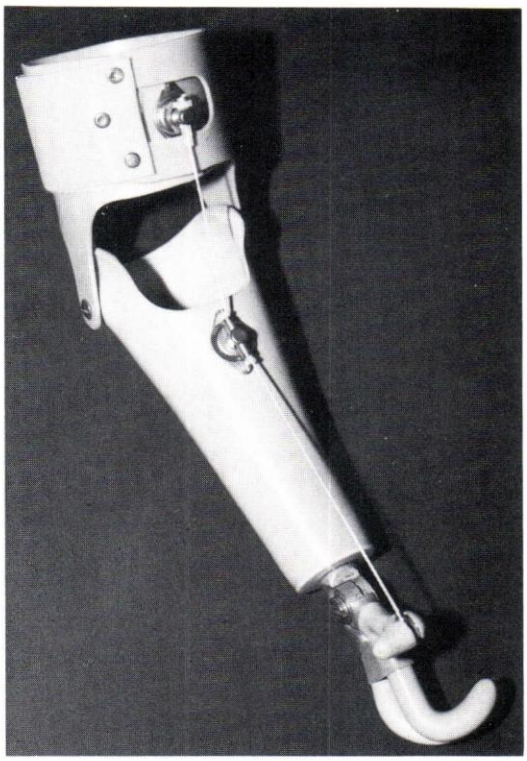


Figure 8. Finished prosthesis with laterally overlapping cuff.

Outfitting the device is simple and direct. With the terminal device pronated and the cuff fixed in extension, pass the control line from the terminal device through to the anterior cuff base plate. Secure the line with a temporary cable hanger and apply the device. Allow the patient to experiment, to his/her satisfaction, with varying combinations of cable tension, terminal device pronation, and elbow extension. Once a definitive cable length is established, install a permanent cable end. Clip the cable flush and buff the end, by wire wheel, for smoothness.

Like all unilateral upper extremity prostheses, its intended capacity is that of an assistive device to potential below-elbow candidates. These candidates should possess sufficient epicondylar prominence, or radio-ulnar styloid prominence in the cases of wrist disarticulations, for suspension. They should also exhibit healthy, active elbow joints and demonstrate that they are not inhibited by the modest limitation to range of motion common to supracondylar socket designs.

As with any device, there are definite limitations. Realistically, this device is seen only as an option: a cost effective, lightweight, and durable alternative to harnesses and battery chargers. Its simplicity makes it compatible with any voluntary opening or closing terminal device, in either an extension or flexion actuating mode, with either braided wire or nylon control line (Figures 4A, 4B).

Its real usefulness lies in the fact that it adapts quickly and inexpensively to any supracondylar below-elbow prosthesis for trial by the amputee (Figure 5). Beyond this feature is the feasibility of instantaneous conversion, back and forth, from a "Figure 8 or 9" harness set-up for work or play, involving heavy axial loading, to the

"Cuff Control" set-up for more comfortable and cosmetic, medium-duty activities. This option is accomplished by means of anterior and posterior base plates on the cuff itself, thus avoiding interruption of the control line during conversion (Figure 6). There are occasions when less can be more, and perhaps this is one.

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Composite Materials for Orthotics and Prosthetics

Dale A. Berry, C.P.(C)

INTRODUCTION

In the ever-changing field of orthotics and prosthetics, recent advancements have been achieved with the use of new materials and resins. In the Spring of 1981, a study project was initiated in an attempt to learn the proper use of these high-tech materials. Data was accumulated from various chemistry and physics texts on the characteristics of composite materials, specifically carbon, Kevlar®, and fiberglass. The next study phase was to fabricate a series of laminated cylinders using composite and stockinette combinations with numerous resins. The cylinders were static-tested for resistance to tension and compression, along with strain and fatigue characteristics. The final research stage was the fabrication of prosthetic appliances for field testing. The lay-up, resin type, total weight, and the author's subjective opinion of every prosthesis was recorded over a 2½ year period. The intent of this article is to present the rationale for specific applications of composites and resins for prosthetic/orthotic appliances, based on field study and the aforementioned research, in conjunction with established West German fabrication techniques and technology.

STRESS

Evaluation of the principal stresses and forces involved in an orthopedic appliance will greatly assist in proper composite choice and application. The principal forces to consider are tension and compression.

Within the socket wall, tensile and compressive resistance forces are responsible for maintaining strength, form, and structure (Figure 1). As body weight is transferred through the socket walls, the outer surface is faced with a specific tensile load, while the inner wall is subjected to an equal and opposite compressive load.

An intermediate layer of material separating the inner and outer wall serves as a transition medium between the opposing forces. The increased distance between the inner and outer wall is directly proportional to increased resistance to fracture, fatigue, and failure. While examining the applied forces on an appliance during the walking cycle (Figure 2) at heel strike, a compressive force is evident at the posterior aspect of the structure, while an equal and opposite tensile force is exerted anteriorly. At flat foot, the forces remain compressive posteriorly and are inverted to a compressive force along the anterior aspect. At toe off, the posterior force transforms to a tensile stress, with the anterior force remaining compressive. Other forces involved with an orthopedic appliance are torque, shear, and impact stress; therefore, they must be considered and appreciated.

With all the specific and individual forces and stress involved with an orthopedic structure, the required properties of a reinforcing composite would be:

- lightweight
- strong under tension
- strong under compression
- flexible, to absorb torque
- stiff, to resist bending and shear stress

- durable, to resist fracture under impact
- capable of resisting stress in all planes
- cost effective
- easy to apply

COMPOSITES

The three composites tested and presently being used in the orthopedic industry are: 1) fiberglass; 2) Kevlar® (Aramid®); and 3) carbon (Graphite). The advantages of one composite over another is due to each material having completely different properties and characteristics (Table 1).

Fiberglass is by far the most common and economical composite. Although the heaviest material of the three, it is easy to saturate with resin and very easy to obtain in many forms and qualities. The principal properties of fiberglass are its durability and flexibility, due to the fibers being twice as strong under compression as compared to the fiber strength under tension.

Kevlar® is the lightest and most expensive composite. It provides an excellent resistance to fracture under impact and can absorb high loads of torque and stress. These desirable properties are, however, compromised, as Kevlar® is very poor in maintaining structure or form under load; it is five times weaker under tension than it is under compression. In addition, Kevlar® is extremely resistant to chemicals and very difficult to saturate with resin.

Perhaps the most valuable composite to orthopedic appliances is carbon. Almost as light as Kevlar®, it is very stiff and able to hold its shape under stress due to its impressive strength under both tension and compression. The structural compromise of the carbon fibers is that the stiffness creates brittleness and a poor resistance to impact.

A very important consideration when working with composites is one of the principles of the fiber—all the available strength and characteristics of a composite fiber are displayed and produced only along the length of the fiber.

To achieve the highest degree of fracture resistance with a composite structure, the angle of the fibers in relation to the applied stress is imperative. With regard to bi-directional (woven cloth) or uni-directional (tape) composites, these materials are excellent for localized strength, but are capa-

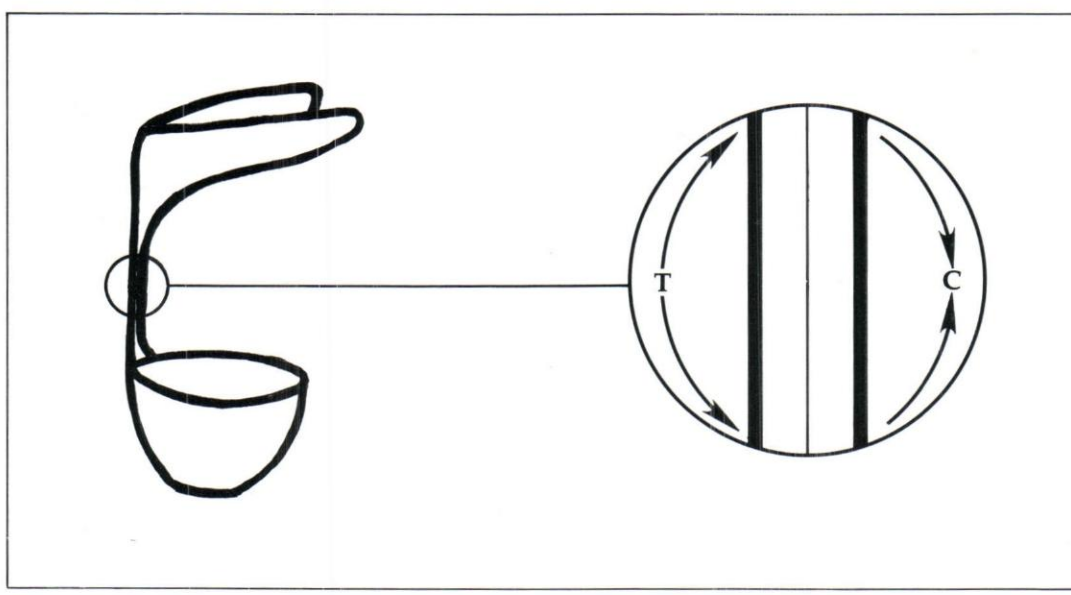


Figure 1. Within the socket wall, a tensile stress is exerted on the outer surface, while a compressive stress is exerted on the inner surface. At the center between the two surfaces, there is an imaginary line called the "Null Zone," which is the transition point of compression and tension. This produces an 'I-Beam' effect. The increased distance between the inner and outer wall is directly proportional to increased resistance to fracture.

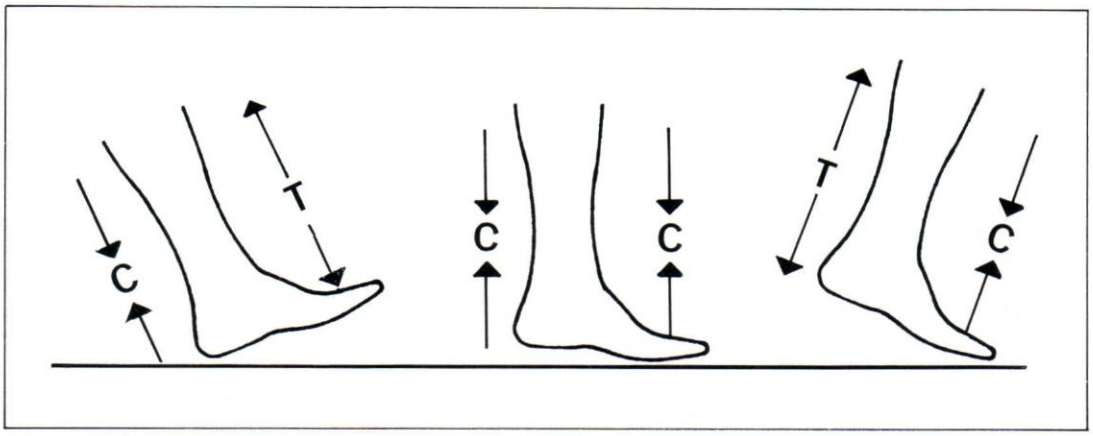
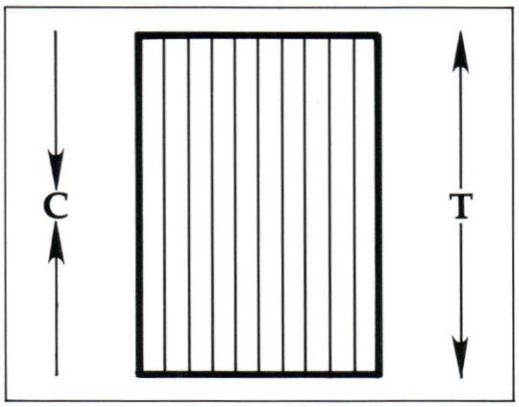


Figure 2. The forces transmitted through the anterior and posterior aspects of an orthopedic appliance fluctuate from tensile (T) to compressive (C) during the walking cycle due to the rotational movement caused by the floor reaction. The stress of torque, shear, and impact have not been included in this diagram, but are present and must be dealt with when reinforcing an appliance.

ble of providing single composite properties in one direction alone, and the fibers of the composite must be positioned perpendicular to the stress plane to be effective. To achieve uniform strength, with equal resistance to fracture in all directions, a "quasi-iso-trophic" composite is required. This can be achieved by applying composites in a mat or knit form, which places composite fibers in a three-dimensional/multi-plane manner. Considering the unique properties of each composite, the most effective application for these fabrics is obtained by blending the materials to produce a "quasi-iso-trophic hybrid" composite. This provides a combination of the most desirable properties of each fiber into a single medium, resistance to torque, shear, compressive, tensile, and impact stress from any possible direction. The ultimate blend is a hybrid of carbon-Kevlar®. Carbon provides lightness and stiffness; Kevlar® provides lightness, impact and torque resistance.

A blend of carbon-fiberglass can also achieve extremely high resistance to fracture with a very good weight to strength ratio, and is somewhat more cost effective. This hybrid displays the stiff and lightweight carbon properties combined with the inexpensive, flexible, and durable fiberglass characteristics.



Tensile and Compressive Values (10^3 PSI) of Static Tested Samples of Kevlar,® Fiberglass, and Carbon

	Fiberglass	Kevlar®	Carbon
Tensile	10	15	13.2
Compressive	5	3	10.6

(Values 10^3 PSI to point of fracture)

Table 1.

SOCKET REINFORCEMENT (Table 2)

The choice of materials to be used for socket fabrication depends entirely upon the patient's demands and requirements.

Varying Lay-up for Prosthetic Appliances

Type of Socket	½ oz. Dacron	Carbon/Glass	Carbon/Kevlar®	Carbon Type	Fiberglass/Nylon Stockinette	Stretch Nylon Stockinette	80/20 Acrylic 2% hardener	Carbon Acrylic 2% hardener
Below Knee Socket	1 inner layer			1 layer at PTB	3 intermediate layers	1 outer layer	300 grams	
Heavy Duty BK Socket	1 inner layer	2 intermediate layers		1 layer at PTB	1 intermediate layer	1 outer layer		300 grams
Super Duty BK Socket	1 inner layer		2 intermediate layers	1 layer at PTB	1 intermediate layer	1 outer layer		300 grams
Below Knee Flexible Socket	*2 layers fiber-glass matting at popleteal fossa	2 intermediate layers		1 layer at PTB	1 inner 1 intermediate layer	1 outer layer		300 grams
Outer Shell for Below Knee					2 inner layers	1 outer layer	250 grams	
Heavy Duty Outer Shell for BK		1 inner layer			1 intermediate layer	1 outer layer		250 grams
Above Knee Socket	1 inner layer			1 layer at ischial level	3 intermediate layers	1 outer layer	400 grams	
Heavy Duty AK Socket	1 inner layer	2 intermediate layers		1 layer at ischial level	1 intermediate layer	1 outer layer		400 grams
*Flexible Socket Frame		2 intermediate layers		*2 layers fiberglass mat	2 intermediate layers	1 inner 1 outer layer		350 grams
Symes Socket	1 inner layer	2 intermediate layers		1 layer at PTB 1 layer at ankle	1 intermediate layer	1 outer layer		350 grams
Heavy Duty Symes Socket	1 inner layer		2 intermediate layers	1 layer at PTB 1 layer at ankle	1 intermediate layer	1 outer layer		350 grams
Knee Disarticulation	1 inner layer	2 intermediate layers		1 layer at condyles	1 intermediate layer	1 outer layer		450 grams
Heavy Duty Knee Disarticulation	1 inner layer		2 intermediate layers	1 layer at condyles	1 intermediate layer	1 outer layer		450 grams
Upper Extremity		1 intermediate layer			1 intermediate layer	1 inner 1 outer layer		250 grams
Heavy Duty Upper Extremity			1 intermediate layer		1 intermediate layer	1 inner 1 outer layer		250 grams
Lower Extremity Childrens Socket		1 intermediate layer			1 intermediate layer	1 inner 1 outer layer		200 grams

Table 2.

All appliances are most durable with five layers of reinforcing composite. The level of durability is controlled by the type of composite used and the method in which it is applied with the applicable resin.

Fiberglass reinforced stockinette will be adequate for the majority of geriatric amputees. If the activity level requires a "heavy-duty" prosthesis, then carbon-fiberglass knit stockinette is preferred. For a "super-duty" socket, carbon-Kevlar® knit composites will provide the necessary strength. When applied with an inner layer of dacron, an intermediate layer of fiberglass nylon, and an outer layer of nylon stockinette, the total thickness for prosthetic sockets in all activity levels remains a uniform five layers of material.

THE BELOW-KNEE SOCKET (Figure 3)

After determining the activity level of the patient and the required blend of composites, consideration must also be directed toward specific stresses in the socket. Due to the thinness of the socket and the fact that significant stress is applied at the patellar tendon level, a strip of two-inch carbon tape is wrapped around the socket at this level to increase the stiffness and to maintain a rigid AP, ML dimension in the socket. For a supracondylar socket, the medial and lateral ears must be stiff and rigid to maintain suspension; thus, the ears are applied with a layer of three inch carbon tape in a vertical direction. For finishing the shin section, the foam (R300 or 10 lb.) is left in place and sealed with Siegelharz resin. The inner layer of composite is dependent upon the patient's activity level. A layer of fiberglass nylon for normal use, carbon-fiberglass for "heavy-duty" use, or carbon-Kevlar® for "super-duty" application.

For "super-duty" and select "heavy-duty" limbs, vertical strips of carbon tape or a layer of bi-directional carbon cloth at the ankle will increase stiffness, tension, and compression resistance. This inner composite layer is then covered with a layer of fiberglass reinforced stockinette and an outer nylon stockinette, and laminated with the appropriate acrylic resin. The average weight of a below-knee socket using this technique is 275 grams, and the average total weight of the finished prosthesis with a SACH foot is 960 grams.

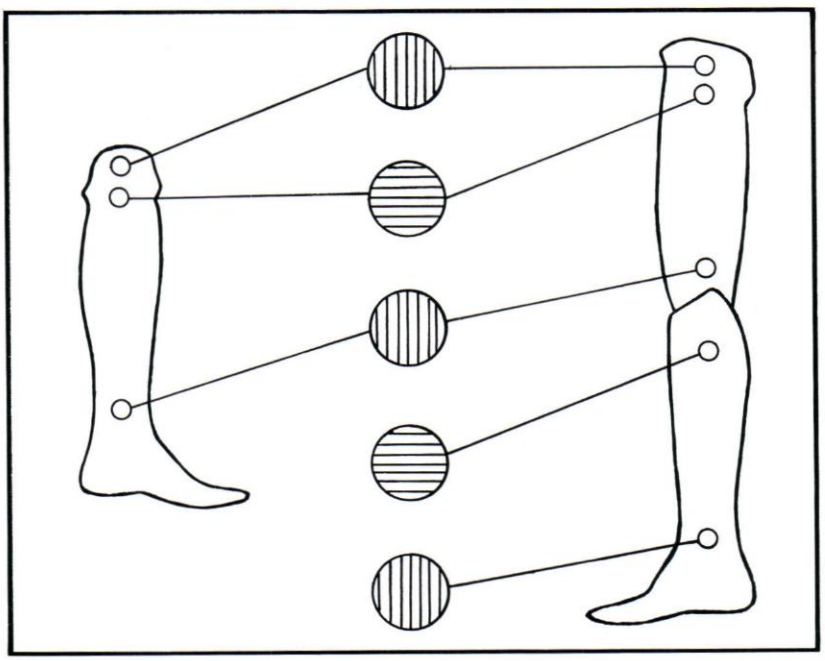


Figure 3. Application of uni-directional Carbon composite on above-knee and below-knee appliances for increased stiffness or specific reinforcement. The Carbon tape is applied to the cast with a small amount of spray adhesive (The acrylic resin will dissolve this during the lamination). The fibers must run perpendicular to the applied stress to hold shape and resist tensile and compressive forces.

THE ABOVE-KNEE SOCKET (Figure 3)

As with the below-knee prosthesis, composite choice for the above-knee socket is dependent upon the amputee's activity level, and a total of five layers of material is preferred. A layer of two or three inch carbon tape is wrapped around the socket at the level of the ischial tuberosity to maintain stiffness and socket shape. In the case of a flexible socket,* the lay-up remains at five layers of composite stockinette, plus two layers of fiberglass matting added between the composite layers to create an "I-Beam," thus increasing strength, stiffness, tension, and compression resistance.

A modular attachment plate is held in place with Siegelharz paste and a lay-up of two layers of fiberglass matting, one layer of fiberglass reinforced stockinette, and a layer of nylon stockinette. For a lower shin section on an exoskeletal type limb, the technique for finishing the below-knee prosthesis is applied over the hollowed-out wood shin portion. The average weight of the above-knee socket is 300 grams; the average weight of an above-knee wood shin prosthesis is 2.5 to 3.5 kilograms, depending upon the foot size and knee unit used (no hydraulic systems were applied).

SYMES AND KNEE DISARTICULATION PROSTHESES (Figure 4)

The disarticulation prosthesis offers the most problems with relation to stress areas and fracture planes. The classic fracture point is the distal anterior and posterior edges of the socket attachment, due to the excessive moments of torque, tensile, and compression stress localized at this section of the prosthesis. Carbon-fiberglass composite is well-suited for the average disarticulation prosthesis, with carbon-Kevlar® meeting the demands for the "heavy-duty" prosthesis.

The localized stress areas at the distal socket attachment points are reinforced with two or three layers of uni-directional carbon tape, ensuring the fibers are running perpendicular to the stress plane. Experience has shown that increasing the total layers of reinforcing carbon tape beyond four layers will only make the prosthesis very stiff and unable to absorb torque and impact. To increase strength, apply one or two layers of fiberglass matting between the carbon layers to produce an "I-Beam" effect better suited to resisting the forces and stresses applied at the ankle or knee.

ORTHOTICS

The major advantages of laminating a composite orthotic device are its lightness, durability, and ability to be stiff and rein-

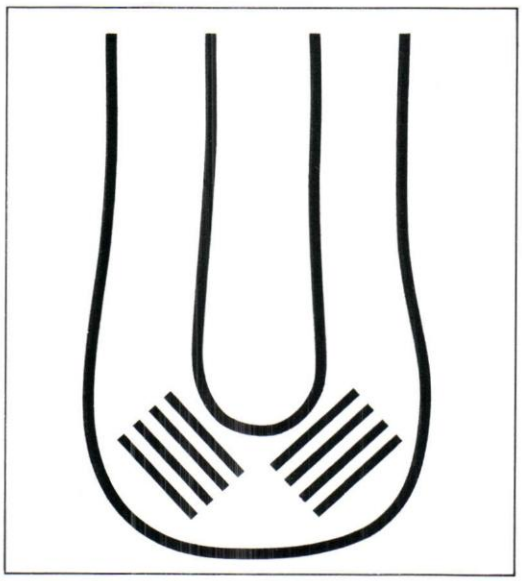


Figure 4. Socket attachment point reinforcement for a disarticulation prosthesis. The Carbon tape is applied with the fibers running perpendicular to the specific stress planes caused by walking (See Figure 2). Overall durability is increased by using a hybrid quasi-trophic composite sandwiched between the Carbon tape.

*Lay-up for the European flexible socket with the width of the medial wall extending from the anterior medial socket edge to the posterior medial socket edge.

forced in any area desired. The disadvantages are that it is time-consuming and expensive when compared to vacuum forming. With this in mind, it seems apparent that a small percentage of orthotic wearers (those patients who are over-active, over-weight, or who need extremely specialized orthotic designs where composite characteristics can benefit the orthosis performance) can justify the application of a laminated orthosis. Because custom orthotic appliances are of varied size and shape, and have to withstand different activity demands, standard lay-up charts have not yet been established. Guidelines that do apply, however, include:

1. Keep lay-ups four to seven layers in thickness.
2. Use carbon uni-directional or bi-directional cloth to increase stiffness and resist stress wherever possible.
3. Design "I-Beams" over stress areas (malleoli, achilles tendon) with one or two layers of fiberglass matting sandwiched between carbon fibers.
4. Evaluate and establish all stress areas and fracture planes so they can be properly and effectively reinforced.

For laminating shoe inserts, arch supports, and UCB inserts, two layers of carbon-fiberglass or carbon-Kevlar®-fiberglass stockinette provided very good results.

SPECIAL SOCKET CONSIDERATIONS

On areas of unusually high stress, a structural design to create an "I-Beam" serves as the best response. Within the socket lay-up, fiberglass matting sandwiched between carbon tape or carbon woven cloth will not add significant weight, but will increase strength up to 20 percent and stiffness up to 40 percent. To provide the ultimate reinforcement, fiberglass matting can be replaced with Kevlar® matting sandwiched between the carbon tape or carbon woven cloth. Care should be taken to identify specific stress planes to ensure the carbon fibers are running perpendicular to it.

In areas where 'grinding' may be necessary to ensure a good socket fit, layers of fiberglass matting are applied over the liner ½ ounce dacron sleeve. The fiberglass matting will provide a very light filler that is completely saturated by the acrylic resin and can be easily ground and buffed to a good cosmetic appearance. To finish the socket edges and relief areas, hand finish with 300 grit sand paper; then apply a thin coat of Acrylic Floor Paste and rub into the plastic.

ACRYLIC RESINS

Acrylic resins are a lightweight thermosetting plastic with excellent wetting properties and good inherent strength, making thin ultra-light orthopedic appliances possible. To achieve the ultimate strength and durability of acrylic, the chemical reaction of the resin must follow a set pattern (Figure 5). It is imperative to shake the tin of resin before use; prosthetic resins are a blend of Methylmethacrylate and citric acid, and will separate in the tin. Failure to stir the acrylic will alter the ratio of chemicals being poured into the cup, creating varying and usually unsatisfactory results (i.e., air holes, improper cure times, boiling laminations, brittle sockets, flexible sockets, soft spots, streaking of color pigment). Acrylic resin pigment is recommended to use with acrylic resin. No more than two percent by weight should be mixed, as the pigment is an active plastics softener, and any mixture over two percent will produce streaking and soft spots. The percentage of Benzol Peroxide hardening powder will provide the best results at two percent by weight. The variances are one percent to three percent, and failure to accurately measure this substance will provide disastrous results (i.e., air bubbles, very brittle laminations, boiling laminations). The blending of acrylic thinners should be avoided at all times. Thinners are non-reactive substances that do not participate in the curing process. Ten percent thinners by weight will reduce acrylic strength by up to 20 percent.

Acrylic resin is available in different

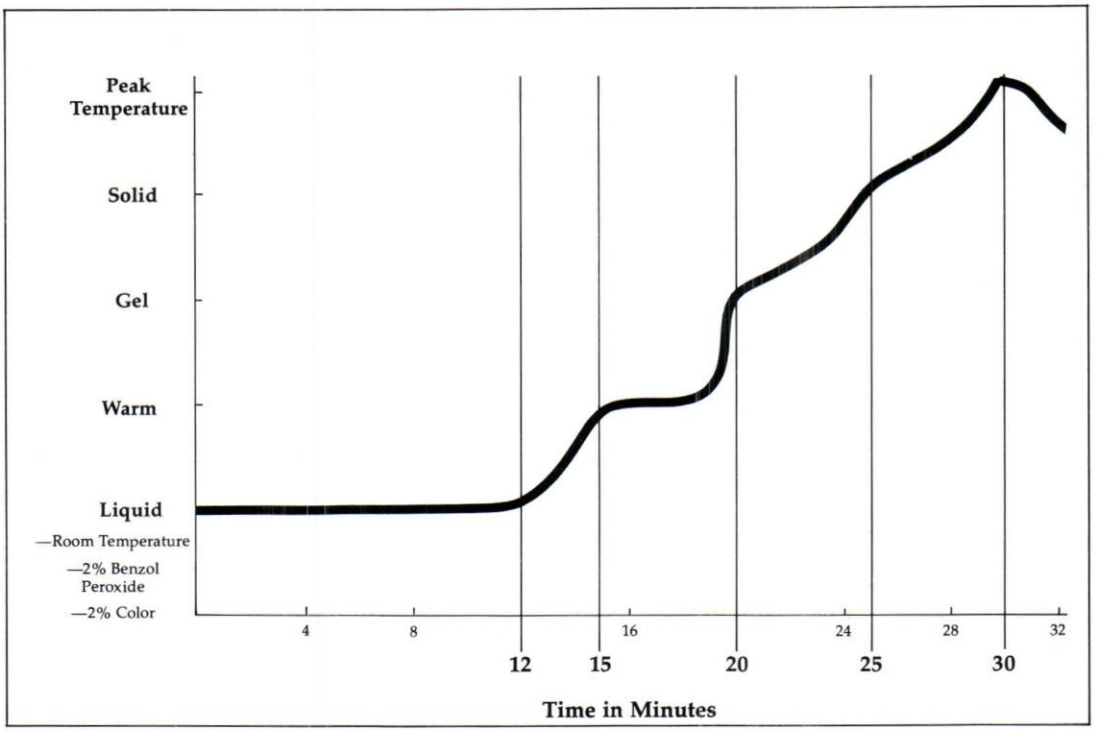


Figure 5. Twelve minutes after 2 percent acrylic pigment and 2 percent benzol peroxide by weight has been blended with the resin, the liquid temperature will rapidly rise to a WARM state. The resin is then poured into the PVA sleeve and impregnated into the material. At the 19 minute mark, the resin will rapidly reach a GEL stage, and the lamination must be completed. The chemical curing process will continue on to the SOLID stage at the 25 minute point and then to a PEAK TEMPERATURE stage at the 30 minute mark. ***To alter this curing pattern in any manner will greatly reduce the strength, durability and working properties of the resin.***

NOTE: This sequence holds true at normal room temperature (21°C.) and normal humidity. If the climate is normal, the resin must be closely monitored and poured into the PVA sleeve when the liquid resin temperature begins to increase. A timer will assist in monitoring the process.

blends, each having its own characteristics and working conditions.

- *80/20 Laminierharz (laminating resin)*—A standard blend of 80 percent rigid and 20 percent flexible resin to be used for vacuum laminations to saturate nylon, fiberglass, and dacron fibers.
- *Elastiharz (flexible resin)*—100 percent flexible resin to be used for vacuum laminations to saturate nylon fibers.
- *Carbonacrylic*—A special blend of 80/20 resin to be used for vacuum laminations to saturate nylon, fiberglass, dacron, and especially carbon fibers (note—carbon acryl will partially saturate Kevlar® up to 85 percent). This resin is designed with a low viscosity for improved saturation and has a higher setting temperature for improved composite bonding. Carbonacrylic resin is not any stronger than regular 80/20 laminierharz; its effect on the carbon fiber is its advantage.
- *Siegelharz (sealing resin)*—A 100 percent rigid resin to be used for bonding common materials and sealing wood and foam. This is the only acrylic resin that can be used without vacuum. For a non-vacuum lamination, blend 30 percent elastiharz with 70 percent Siegelharz with two percent color and one percent Benzol Peroxide paste. This mixture will cure rapidly compared to other acrylic resins.
- *Siegelharz Paste (sealing paste)*—An alternative way to apply Siegelharz, as this is blended into a gel and does not require any fillers. It will set with one percent Ben-

zol Peroxide powder or paste in five minutes to be completely cured in 10 minutes. This material will give an excellent bond between all common materials, will adhere metal joints and attachment plates to sockets, and will serve very well for socket repairs (note—Siegelharz paste will not adhere wood to wood; liquid Siegelharz should be used).

To calculate the amount of resin required for a lamination, a formula has been established for lay-ups consisting of five layers of material:

$$\frac{\text{largest circumference of the cast in centimeters} \times \text{total length of the cast in centimeters}}{\div 3} = \text{the total grams of resin required}$$

The total grams required is then rounded off to the nearest 50 (e.g., a final answer of 333 grams will be rounded off to 350 grams).

CONCLUSION

The opportunities and applications for hi-tech composites and acrylic resins in the orthopedic industry are seemingly infinite. Assessment of every patient's orthopedic appliance, concentrating on structural stresses, composite type, and fiber orientation with proper resin application will increase material performance and provide the numerous advantages "hi-tech" materials have to offer.

ACKNOWLEDGMENT

The Glenrose Rehabilitation Hospital in Edmonton, Alberta, Canada was the location of the initial testing and study on which this paper is based. The support and progressive attitude of the administration and support staff will always be appreciated and remembered.

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Evaluation of High Strength Materials for Prostheses

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John W. Egan, M.S.
Norman G. Gall, M.D.

INTRODUCTION

The weight of a prosthesis has always been a problem for prosthetic researchers.[1] According to Mooney,[2] most below-knee prostheses, laminated in the normal prosthetic laboratory, weigh about five pounds. Below-knee prostheses are usually attached to the limb by a strap around the thigh or with wedges pressing inwards above the condyles of the femur. With normal gravitational forces, this weight creates a friction between the residual limb and the prosthetic socket interface that may cause skin breakdown.

The weight of a prosthesis may cause excessive muscle work that will result in high energy consumption for amputees. Mooney[2] states that, "a standard prosthesis requires approximately 12 percent more energy consumption" and "energy consumption is the key to successful ambulatory activities."

Ganguli, et al.[3] stated that, "with respect to energy expenditure, the degree of departure from normal performance standards in the below-knee amputee fitted with a patellar tendon bearing (PTB) prosthesis is quite high." Cummings, et al.[4] states that, "a distally applied weight of 2½ pounds would be expected to add five to ten percent to the energy requirement of ambulation." Fisher and Gullickson[5] state that below-knee amputees "walk 36 percent slower, expending two percent more Kcal/min and 41 percent more Kcal/mtr than the normal person." Waters, et al.[6] found that vascular below-knee amputees walk 41 percent slower and expended 55 percent more Kcal/mtr/Kcal/kg than non-amputees.

The need for lighter weight prostheses is often cited in the literature[1,2,7,8] and occasionally an innovative procedure will surface;[9] however, when the technology differs from that in current practice, the prosthetic clinic team has difficulty adapting to it. The procedure described by Wilson[9] was not familiar to the prosthetist; as a consequence, this very lightweight prosthesis is not commonly fabricated.[17]

Prostheses are normally excessively heavy, which tends to increase residual limb trauma and energy expenditure with the likelihood of less successful prosthetic function. It is the intent of the clinic team to provide an appliance that will stand up under the strain of constant use. With these considerations in mind, the Rehabilitation Engineering Lab (REL) at the University of Texas Health Science Center at San Antonio (UTHSCSA) proposed to determine if a material could be designed which would utilize normal prosthetic laboratory techniques, yet allow the prosthetist to produce a below-knee prosthesis weighing less than two pounds and having the strength to adequately support normal ambulation loads.

CURRENT STATUS OF WORK IN THE AREA

Aramid™ fibers and carbon fibers were selected as new materials to be used as a reinforcement for the lamination of prostheses because:

- Aramid™ fibers have a very high tensile strength (Figure 1) and the elongation to break ratio is very low (Figure 2).
- Carbon fibers exhibit an excellent modulus and their density is lower than many other materials currently used for strength in prostheses (Tables 1 and 2).

The tensile strength of Aramid™ and carbon fibers is far superior to nylon, the material normally used by many prosthetists. The nearly linear stress/strain curve to failure of Kevlar® 29 (Aramid™ fiber) is similar to that of glass, but unlike those of other organic fibers (Figure 3). Because it is relatively insensitive to fiber surface defects, the tensile strength of Kevlar® 29 is uniform along the length of the fibers.

Research work in the area of orthotics and prosthetics using carbon fibers has been directed primarily toward orthotics. In 1976, N.A.S.A. published a technical brief in which they described a new, lightweight brace constructed of fiber rein-

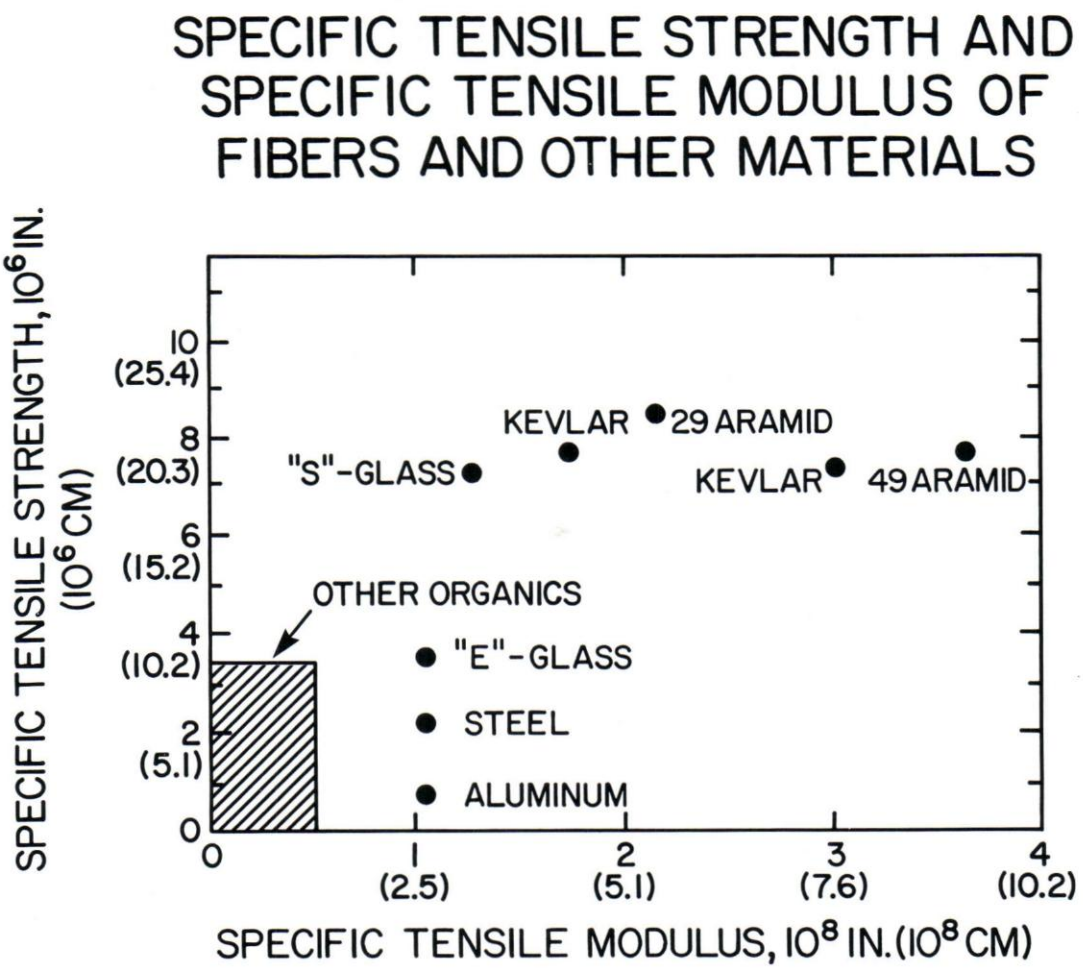

Figure 1. Tensile strength versus tensile modulus.

COMPARATIVE YARN PROPERTIES

	KEVLAR® 29 Aramid	DU PONT Nylon Type 728	DACRON® Polyester Type 68	"E-HTS" Glass	Stainless Steel
Tensile Strength, lb/in²	400,000**	143,000**	162,500**	350,000***	250,000
(MPa)*	(2758)	(985)	(1120)	(2412)	(1724)
Modulus, lb/in²	9,000,000	800,000	2,000,000	10,000,000	29,000,000
(MPa)	(62000)	(5512)	(13780)	(68900)	(199800)
Elongation to Break, %	4.0	18.3	14.5	3.5	2.0
Density, lb/in³	0.052	0.041	0.050	0.092	0.284
(g/cm³)†	(1.44)	(1.14)	(1.38)	(2.55)	(7.83)

*Mpa = MN/m^2 = $lb/in^2 \times 6.895 \times 10^{-3}$
**Unimpregnated twisted yarn test — ASTM D2256
***Impregnated strand test — ASTM D2343
†$g/cm^3 \times 27.68$

Figure 2. Reprinted with permission from Dupont's "A Preliminary Information Memo," Number 375, September 28, 1976.

forced polymer materials.[10] Also in 1976, the Southwest Research Institute published a final technical report prepared by S.R. McFarland and G.C. Grimer[11] in which they reported producing a pair of bilateral long leg braces from carbon fiber filaments. These braces weighed approximately 1½ pounds each, including the footplate which was formed of steel.

The orthoses produced by N.A.S.A. and the braces produced by McFarland at Southwest Research Institute both employed a very lengthy process which requires placing layers of composite materials on an intercore and laminating these materials together to be used as struts for the orthosis. Neelham, in his paper, "Carbon Fiber Reinforced Plastic Applied to Prosthetics and Orthotics,"[12] described a process similar to the one employed by N.A.S.A. and Southwest Research to fabricate a harness for externally powered upper extremity prostheses that were fitted to thalidomide damaged children. He also fabricated a thoracolumbosacral orthosis and a bilateral hip-knee-ankle-foot orthosis.

The fabrication process and the technology needed to fabricate these orthoses and prostheses require extensive retraining in laboratory techniques for prosthetists and orthotists in this country. New machines and tools would have to be installed. Richard Striebinger, in a letter to S.R. McFarland dated February, 1983,[13] stated that his group at the Rensselaer Polytechnic Institute in New York had fabricated an orthosis in a sandwich construction using graphite, Kevlar® 29, and an epoxy matrix along with a foam core. This process, like the others, requires a long, complicated curing process under vacuum at room temperature.

Hittenberger and Putzi,[14] at the V.A.M.C. lab in Seattle, Washington, reported they had developed a laminating procedure for lightweight prostheses which requires one of the laminations to be split and a foam core removed. This produced a prosthesis that weighed approximately 1½ pounds. However, the lab procedures, as described, require the prosthetist to cut the prosthesis posterially along the sagittal line. This would tend to weaken the prosthesis in an area that receives very high stress and might cause it to break. The "Ultralight Below Knee Prosthesis"[9, 15] requires a "hand draped" vacuum formed fabrication procedure and polypropylene polymers. These are split posteriorly and later welded together. While the prostheses are ultralight when compared to conventional systems, the process requires new technology, additional tools and machines, and the end

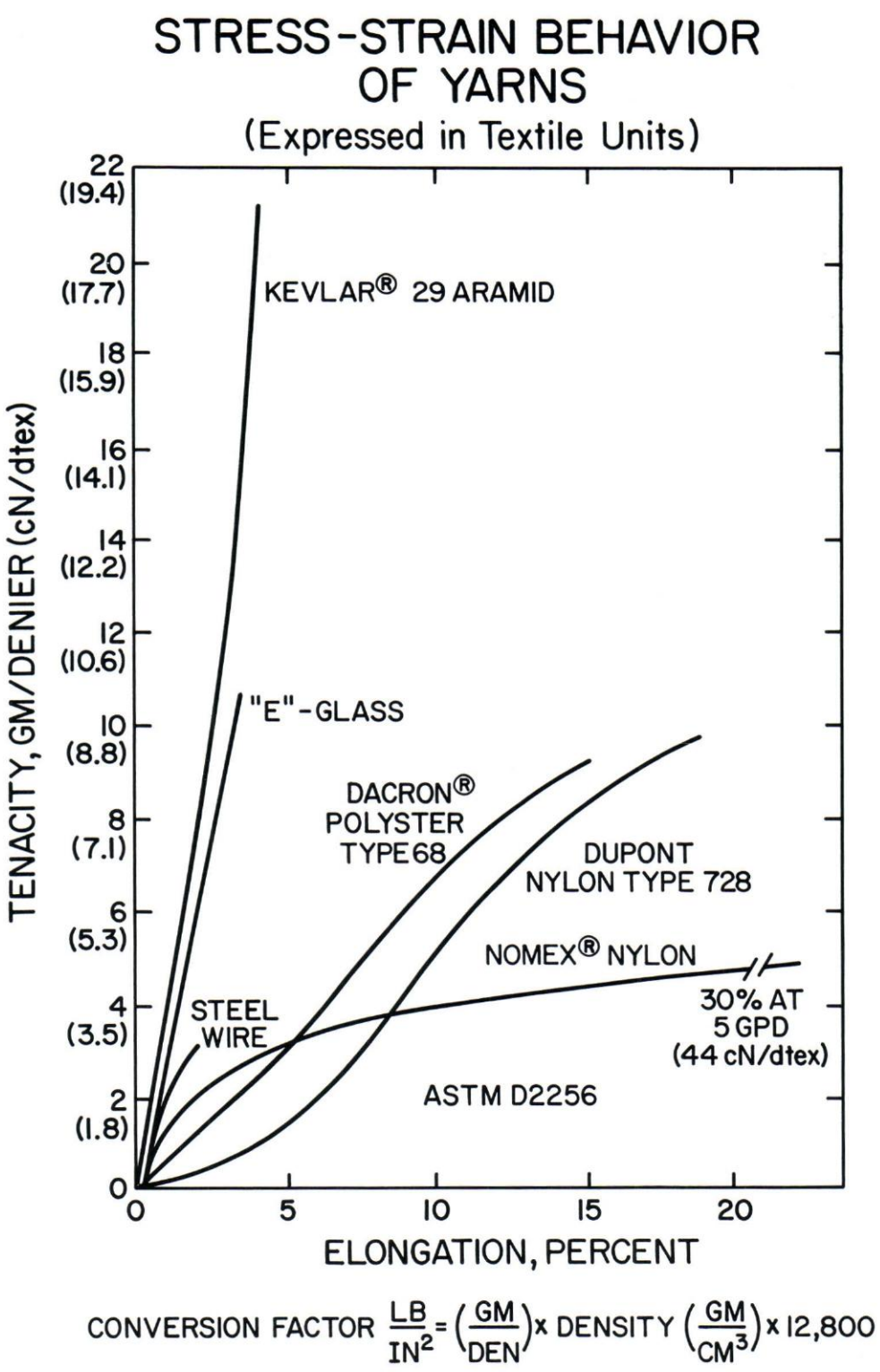
STRESS-STRAIN BEHAVIOR
OF YARNS
(Expressed in Textile Units)
TENACITY, GM/DENIER (cN/dtex)
22 (19.4)
20 (17.7)
18 (15.9)
16 (14.1)
14 (12.2)
12 (10.6)
10 (8.8)
8 (7.1)
6 (5.3)
4 (3.5)
2 (1.8)
0
KEVLAR® 29 ARAMID
"E"-GLASS
DACRON® POLYSTER TYPE 68
DUPONT NYLON TYPE 728
NOMEX® NYLON
30% AT 5 GPD (44 cN/dtex)
STEEL WIRE
ASTM D2256
0
5
10
15
20
ELONGATION, PERCENT
CONVERSION FACTOR LB/IN² = (GM/DEN) x DENSITY (GM/CM³) x 12,800

Figure 3.

	2024-T4 ALUMINUM	POLYPROPYLENE	CARBON FIBER/ EPOXY COMPOSITE
Tensile Strength			
$LB/IN^2 \times 10^3$	55.0	5.1	211.0
Comp. Yield Strength			
$LB/IN^2 \times 10^3$	32.0	5.0	171.0
Tensile Modulus			
$LB/IN^2 \times 10^3$	10.0	0.16	18.5
Density LB/IN^3	0.101	0.033	0.555
Tensile Strength			
Density $IN \times 10^5$	5.4	1.55	38.4
Tensile Modulus			
Density $IN \times 10^8$	1.0	0.15	3.4

*Taken from Table I, NASA Tech Brief 75-10303

Table 1.

Comparison of Tensile Strength of Graphite With That of Other Structural Materials

Type of Material	Strength, ksi		Density lb/in.^{3}
	0°	Isotropic	
Graphite AS/epoxy	225	84	0.056
Graphite HMS/epoxy	165	50	0.059
S Glass/epoxy	260	110	0.072
E Glass/epoxy	120	75	0.072
Aluminum (7046-T4)		55	0.100
Steel (SAE-980)		80	0.289
Nylon (6/6)		12	0.041

Table 2.

product is prone to failure,[17] due to the high stress placed on the ankle-foot components during the forming process, and because of inappropriate heating and cooling of the plastic. This procedure has not gained acceptance by the prosthetic profession.

OBJECTIVE OF THE RESEARCH PROJECT

This project was designed to study the following objectives:

- To establish manufacturing techniques and criteria for knitting Aramid™ and carbon fibers into stockinette materials suitable for lamination in prosthetic laboratories.
- To determine which fiber or combination of fibers would make the strongest and lightest weight prosthesis.
- To determine the best polymer (acrylic—epoxies—polyesters) for laminating these fibers in prostheses.

MATERIALS

Carbon fibers and yarns are made by several companies in the United States, however, most of these products cannot be knitted into materials that are suitable for normal prosthetic applications because the fibers are so soft. In their natural state the fibers must be braided into heavy bulky strands to eliminate breakage during the knitting process. These bulky braids result in an undesirable uneven surface on the completed prosthesis.

Aramid™ fibers and yarns in a variety of sizes are manufactured in the United States. Most of these are suitable for knitting purposes. In addition, the Otto Bock Company of Minnesota has developed a lamination technique using carbon fibers in a mat form,[16] but reinforcement materials in a mat form are not normally used in the prosthetic lab. The superior properties of Aramid™ and carbon fibers have prompted several companies to develop an assortment of fabrics to be used for prosthetic laminating.

Aramid™ fibers are used in Aralon.™* This product is described as a high strength "stockinette" made of high technology fibers next in strength to that of carbon. The manufacturer claims that Aralon™ "produced a prosthesis over 40 percent stronger and almost half the weight of conventional prostheses and that Aralon™ is 2½ times superior in ratio of fiber strength to weight than nylon." It also is claimed to have superior impact and fatigue resistance and excellent thermal stability with little change in dimension over normal temperature ranges. Aralon™ is said to be compatible with both polyester and epoxy resins, and stretches like regular "stockinette." Carbon fibers in combination with glass and Aramid™ were knitted into a stockinette material for this project by IPOS.**

A stockinette material made from a combination of carbon and glass fibers** has been available for several years, but most prosthetic facilities have not used it because it is very expensive, the glass fibers are health hazards to work with, and the knitted material when laminated does not have a smooth appearance. It is claimed this carbon fiber material is compatible with an acrylic resin, trade-named Carbon Acryl.™** According to the manufacturer, Carbon Acryl™ has an additive that makes it very compatible with the carbon fibers and causes a "chemical bond" during lamination.

- The following yarn specifications were obtained for knitting and testing by the Knit-Rite Company of Kansas City, Missouri:
  - —Aramid™: Kevlar® 29—14/1 and 20/1
  - —Carbon: Pyron—4/10 w.c. and 2/32 w.c. Panex (refired)—30Y800, 30Y300 and 30R
  - —Glass: Fiberglass—150—1/0-1
  - —Nylon: Stretch nylon—1/100 Type 66 D-4 Perma-Set

(The above yarns were knit in stockinette and rib stitch by Knit-Rite, Inc., of Kansas City, Missouri, as outlined in Figure 4.)

*Manufactured by Comfort Manufacturing Company of Burlington, New Jersey

**IPOS Komman Ditgesellschaft, Luner Renn Bahn 14.D2120 Luneberg

ARAMID, CARBON, NYLON, GLASS, FIBERS

	Kevlar 1/14	Kevlar 1/20	Pyron 4/10 W.C.	Pyron 2/32	Panex 30 Y 800	Panex 30 Y 300	Panex 30 R
Kevlar 1/14	1/1 lab 1/1 rib						
Kevlar 1/20		1/1 rib					
Pyron 4/10 W.C.	1/1 lab 1/1 rib		1 lab				
Pyron 2/32				4 rib refired			
Panex 30 Y 800		1/1 lab					
Panex 30 Y 300		1/1 lab					
Panex 30 R		1/1 lab					
Nylon 66					1/2 lab	1/2 lab	
Stretch Nylon Y 100		1/1, 2/1 lab 1/1 rib				1/1 lab 1/1 rib	1/1 lab
Fiberglas 150 - 1/0 - 1		1/1 lab* 1/1 rib			*3 stitch settings		

1. Top number refers to ends (strands) of vertically listed fiber.
2. Bottom number refers to ends (strands) of horizontally listed fiber.
3. Rib knit — circular machines.
4. Lab knit stockinette with every other needle out on 6" cylinder.
5. KEVLAR = ARAMID
6. PYRON = CARBON
7. PANEX = GRAPHITE

Figure 4. The top number refers to ends of vertically listed yarn. The bottom number refers to ends of horizontally listed yarn. Lab knit is stocknette; rib knit is knit-pearl stitch on the circular machine.

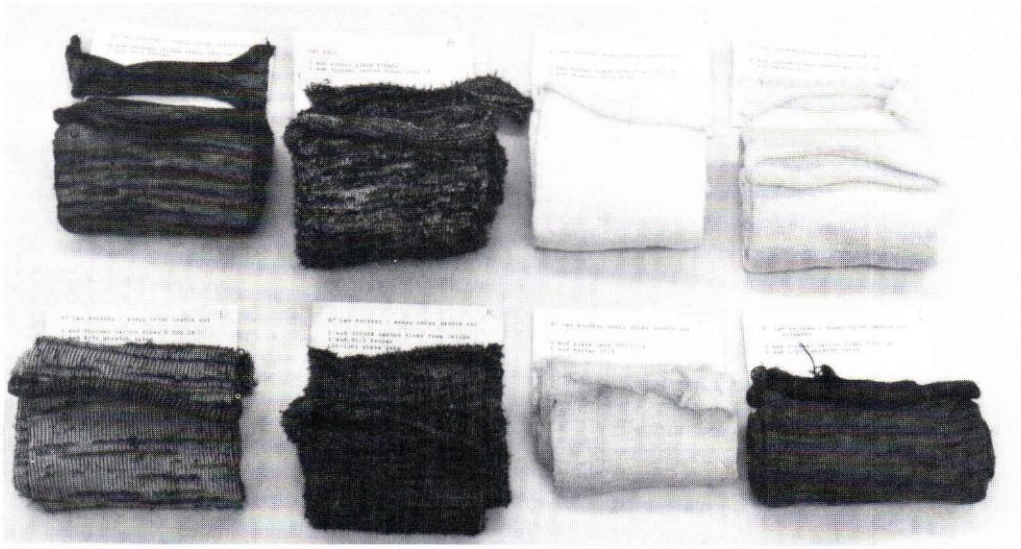

Figure 5. Stockinette knitted for use in this research project.

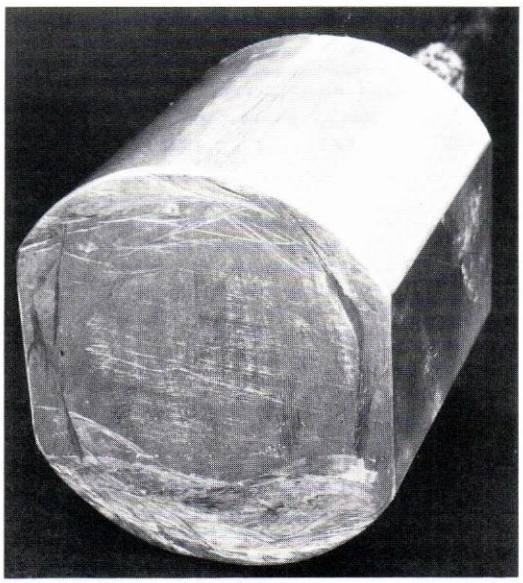

Figure 6. To restrict variability in strength measurements, a cylindrical aluminum mold with two flat sides of equal proportions was machined to be used as the model for laminating all of the test laminations.

Using the knitted stockinette materials from Knit-Rite and IPOS (Figure 5), we laminated a series of test models using the new stockinette material with:

- IPOS Carbon Acryl™ acrylic resin
- Epocast 502 epoxy resin
- Laminac 4110 polyester.

To restrict variability in strength measurements due to physical and geometrical factors, a cylindrical aluminum mold with two flat sides of equal proportions (Figure 6) was machined and fabricated to be used as the model for laminating all of the test laminations. Coupons measuring two and one half centimeters by five centimeters were cut from each of the laminations (Figure 7). These coupons were tested for strength using the Instron.™ The Instron™ conventionally measures strength and flexoral properties of plastics. It conforms to the American National Standard K6575-1971. This testing method has been approved for use by agencies of the Department of Defense to replace Method 1031 of Federal Test Methods Standard 406 and for listing in the DoD Index of Specifications and Standards. The instrument provides a graphic readout of the force (measured in Newtons) required to fracture the coupons.***

***1 Newton =.102 grams.

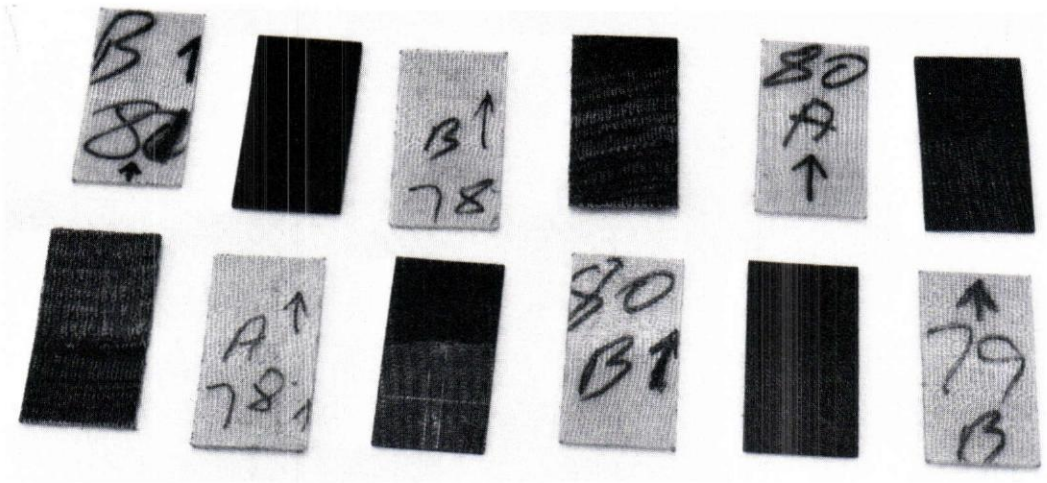

Figure 7. Coupons measuring two and one-half centimeters by five centimeters were cut from each of the laminations.

PROCEDURE

Using each of the different materials with each of the different resins, a series of test models were laminated under vacuum pressure over the custom designed aluminum mold.

- *Acrylic Resins*—Using the custom made mold, we laminated the stockinette made from the Aramid,™ nylon, and glass fibers separately and in combination using the acrylic resins, as follows:
  - —Over the custom mold, we pulled a poly vinyl alcohol sheet (PVA) and applied vacuum under the PVA to insure good mold clarification.
  - —We applied the stockinette, and over this stockinette we pulled a PVA bag to hold the acrylic resin.
  - —Vacuum was applied under this bag to insure good mold conformity.
  - —The laminating resin was prepared by combining 250 grams of carbon Acryl™ with enough hardening powder to effect a cure time of 30 minutes.
  - —This mixture was then poured into the PVA bag, allowed to impregnate the stockinette, and then cured.
  - —From the laminated model, we cut two coupons, 2½ cm by 5 cm.
  - —To test the strength of the coupons, they were placed in the Instron,™ using a three-point bending apparatus on supports spaced 30mm apart. A downward force was applied exactly at the center of the coupon at a rate of 10mm descent per minute. The strength of the material was measured as peak force at fracture.
- *Epoxy Resin*—Using the above described procedures, we laminated a new series over the custom made model using epoxy resin.
- *Polyester Resin*—Using the above described procedure, we laminated a new series over the custom made model using polyester resin.

At this time, our project has produced more than 300 laminated coupons using the various combinations of fibers. The strongest coupons obtained from the various combinations of Aramid,™ carbon, nylon, and glass fibers are listed in Table 3.

RESULTS

Coupons of standardized width and length, but variable thickness, were tested

Strengths of Various Resins and Fabric Combinations

Fiber	Strength N/mm^2	Resin	Std. Dev. + or −
Carbon/Nylon/Aramid	255.567	Acrylic	61.774
Carbon/Glass	193.908	Epoxy	30.604
Carbon/Aramid	176.651	Epoxy	50.791
Carbon/Aramid/Glass	152.781	Polyester	17.492
Aramid	151.851	Polyester	34.166
Carbon/Nylon	148.113	Epoxy	21.675
Aramid/Nylon	131.221	Polyester	10.393
Aramid/Glass	104.945	Epoxy	29.727
Nylon	70.581	Epoxy	5.774

Table 3.

in a three-point transverse loading apparatus using the Instron™ for administering a measured load. Thickness, maximum transverse breaking force, and the standardized width and length parameters were then compiled, and the transverse strength computed according to the formula,

$$S = \frac{F * L}{4 * z}$$

where S— is the maximum stress incurred by an "extreme fiber" most distant from the central bending axis;

F— is the transverse load in Newtons;

L— is the span between the two supports (30mm in this experiment);

z— is the "section modulus" characteristic of the cross-section geometry. For these coupons it is equal to: 1/6 * width * Thickness.[2]

Therefore,

$$S = \frac{3\,F\,L.}{2\,W\,T^2}$$

Coupons were grouped to the type resin and fiber combination; (See Figures 8, 9, and 10).

The appropriate individual transverse strength measurements were then pooled, and means and standard deviations computed (Table 3). The relatively large standard deviations in some of the groups are due in part to the nature of the laminating process currently in widespread use. When woven tubular stockinettes are pulled over a particular prosthesis shape, the orientation and overlap of fiber layers becomes arbitrary within certain bounds set by the stockinette manufacturer's knitting pattern. Accordingly, when test coupons are cut from the laminated prostheses, there is no way to control for direction or degree of offset of fiber layers. Since this element of randomness would creep into all tests, it was concluded that a mean strength estimate would reflect a fairly respresentative number for an "average" prosthesis made in this clinically typical manner.

To illustrate a comparison of "typical" prostheses weights using any of the several possible combinations, we choose a model below-knee prosthesis laminated in nylon/polyester by a local prosthetic facility. The facility was unaware that the below-knee prosthesis was to be used for this research project.

The finished prosthesis, including the socket, was first coated with a castable urethane elastomer produced by Smooth-on, Inc., of Gillette, New Jersey. After curing, the elastomer was carefully removed without stretching, then cut into eleven pieces

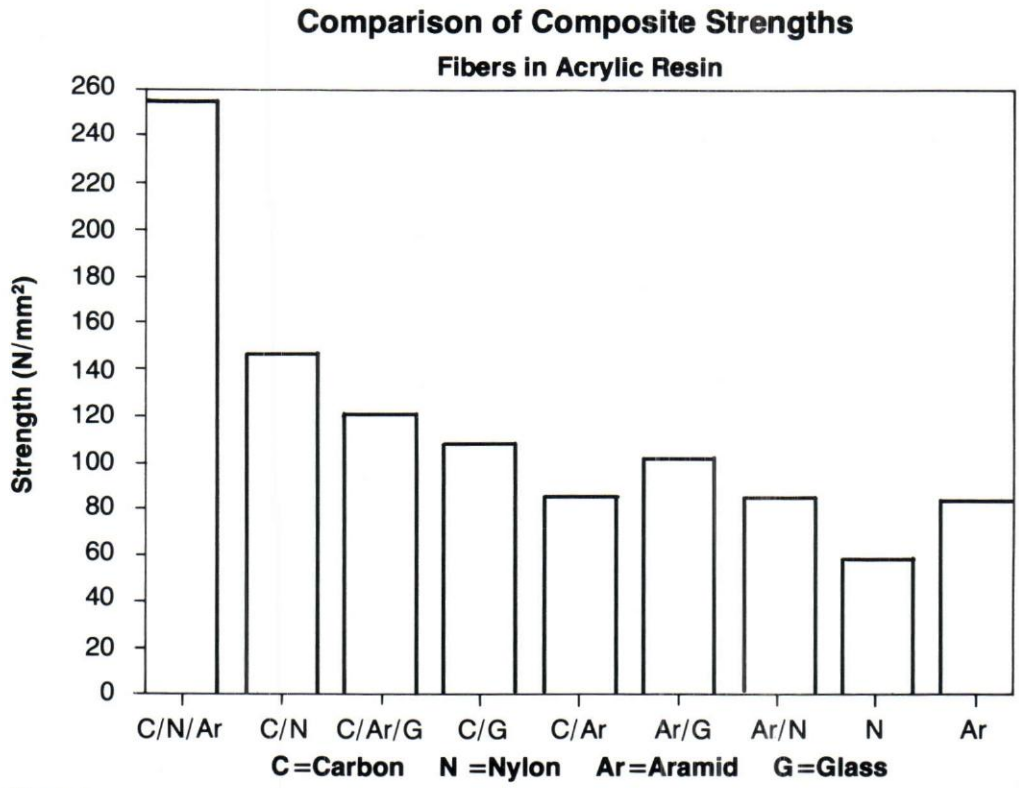

Figure 8.

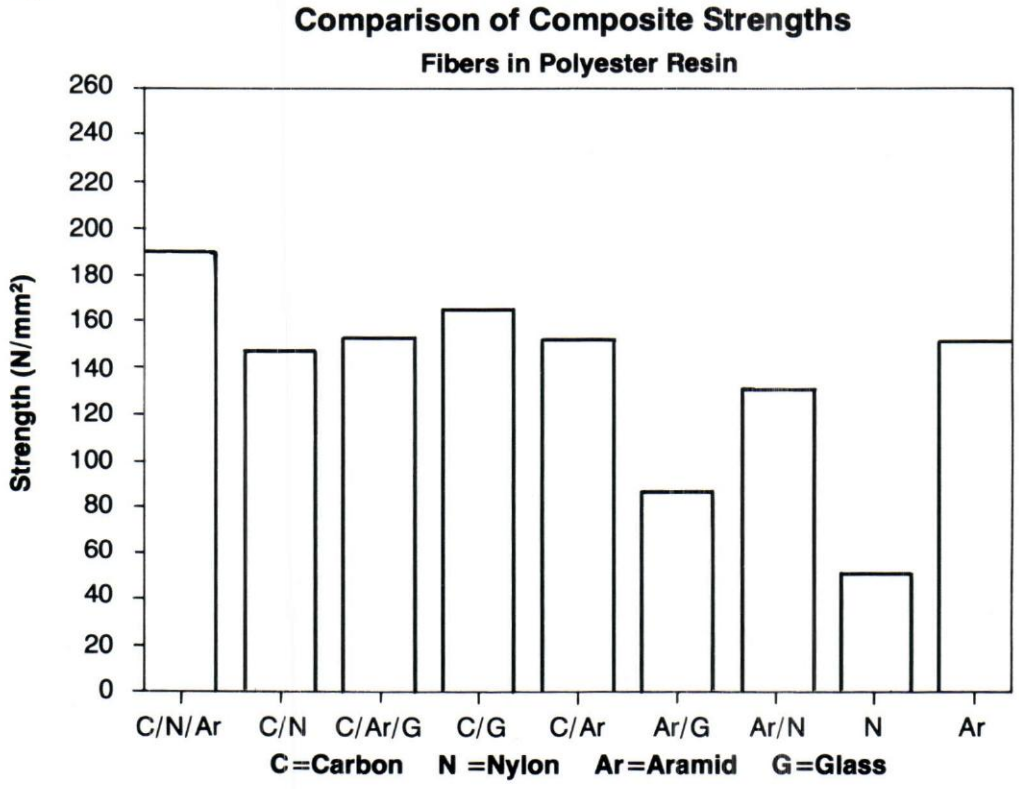

Figure 9.

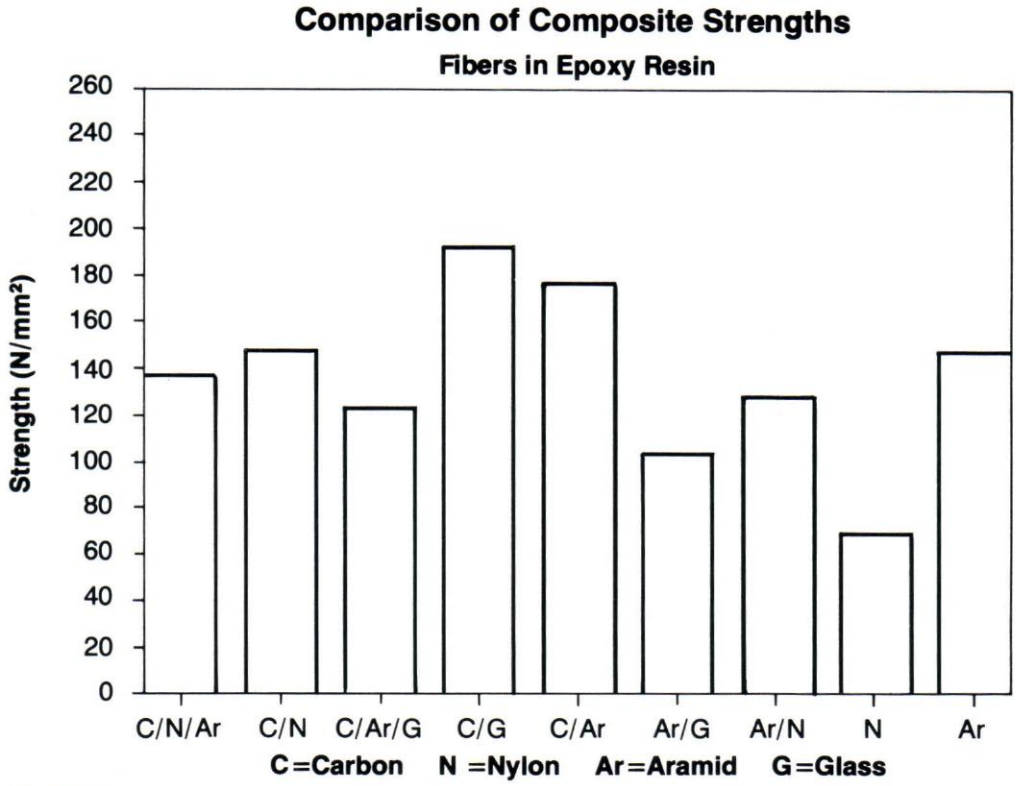

Figure 10.

Characteristics of a Below-knee Prosthesis Made of a Nylon/Polyester Composite

Model B-K Prosthetic Characteristics

Nylon/Polyester Composite

	Area cm^2	Thickness mm	Break Force Nwt
Surface Area Total cm^2:	2073.938		
Zone 1 Inner:	627.138	4.115	1982.024
Zone 1 Outer:	650.000	1.633	123.7539
Zone 2 Outer:	796.800	2.363	375.4693
Coupon Width (mm):	25.000		
Coupon Sep Dist (mm):	30.000	Used for Strength	
Coupon Length (mm):	50.000		
Coupon Tot Area (mm^2):	1250.000		
Coupon Wgt:	Changes		
Nylon/Polyester Stress:	51.200		

Figure 11.

ACRYLIC				
Abbrv.	Equ. Wgt. gm	Ave. Strength N/mm^2	Std. Dev. + or −	Density gm/cc
C/Ar/G	295	121	63	0.710
C/Ar	331	85	15	0.710
Ar/G	479	102	29	1.089
C/G	498	108	8	1.158
N	494	58	16	0.931
C/N	507	147	36	1.303
Ar/N	527	85	9	1.128
C/N/Ar	615	256	62	1.902
Ar	740	83	10	1.574

EPOXY				
Abbrv.	Equ. Weight gm	Ave. Strength N/mm^2	Std. Dev. + or −	Density gm/cc
Ar/G	274	105	30	0.630
C/N	294	148	22	0.757
C/Ar	385	177	51	1.052
N	431	71	6	0.868
C/N/Ar	490	138	14	1.233
Ar/N	525	129	30	1.294
C/G	541	194	31	1.526
C/Ar/G	564	124	16	1.371
Ar	606	149	25	1.564

POLYESTER				
Abbrv.	Equ. Weight gm	Ave. Strength N/mm^2	Std. Dev. + or −	Density gm/cc
C/N	156	147	25	0.402
C/N/Ar	292	190	24	0.818
C/Ar/G	414	153	17	1.080
C/G	422	166	58	1.131
Ar/G	499	87	23	1.078
C/Ar	518	152	50	1.350
Ar/N	605	131	10	1.500
Ar	544	152	34	1.451
N	691	51	6	1.250

Table 4.

in such a way that they would lay approximately flat. These pieces were then measured using a 2-D digitizing planimeter. This area figure was taken to represent the total surface area of the prosthesis, excluding the plantar surface of the prosthetic foot. For this prosthesis, the area totaled some 2,300 square centimeters.

Next, circular corings were taken in various areas or "zones" on the prosthesis: four in the socket wall, and three in various places down the leg. For the socket cores, which penetrated both the outer prosthetic wall and the socket inner wall, two distinct layers of hardened composite were visible. Measurements of layer thickness were made. Three zones emerged: Zone 1i, the inner lay-up thickness of the socket wall itself; Zone 1o, the outside wall thickness of the socket; and Zone 2o, the outside wall thickness everywhere else in the leg.

A set of nylon/polyester coupons were tested to obtain a figure for the material's strength (as maximum stress). By calculating the equivalent breaking force required to break a nylon/polyester coupon with thickness equal to that of each zone in the prosthesis, a "Design Break Force" figure was obtained for each zone (Figure 11). Then, using the stress numbers determined for each test material, an estimated thickness could be calculated for any new material used to build a prosthesis having a similar "Design Break Force" for each zone. Furthermore, knowing the density of each composite, the surface area (2,300 cm^2) and thickness of material requred, a weight figure was generated giving the minimum weight of composite materials required in an equivalent "typical" prosthesis (Table 4).

CONCLUSION

Knitted combinations of high-strength yarns were laminated with different resins and laboratory tested in order to obtain a material which could be used for making lightweight, high-strength prostheses and orthoses by facilities using techniques and equipment readily available to them.

This project has established knitting specifications for stockinette manufacture using Aramid™ and cotton yarns. These yarns and the combinations tested may not be the most suitable for prosthetic laminations because of the many variables, i.e., price, availability, combinations not tested, and the fact that newer and stronger fibers are waiting to be discovered.

Although prototypes of prostheses have been made by the Rehabilitation Engineering Laboratory, the actual clinical work still needs to be done. However, the results of this research indicate that materials have been identified which have potential and should be tested further using controlled experimental designs.

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Technical Note: A Technique for Prosthetic Nipple Restoration

R.L. Engelmeier, D.M.D., M.S.

INTRODUCTION

Many techniques have been presented in the prosthetic literature concerning the restoration of oral-facial defects. This is primarily because of the tremendous functional and psychological impact that these defects impose on their unfortunate hosts. Individuals place varying degrees of importance on the appearance of different parts of the body, and because of these varying priorities, they react differently to the loss and subsequent restoration of these different body structures. Some patients are quite happy wearing a black eye patch following orbital exeneration while others demand the ultimate in an orbital prosthesis. Some will hide the loss of an ear by simply growing their hair longer while others will insist on short hair and a prosthetic ear. For some, a small defect, which could easily be hidden, can impart the same feeling of being unwhole that very large facial defects do for others. Often a patient's defect, no matter how small, becomes the focus for all that's wrong in his or her life.

Patients with breast cancer, who have undergone mastectomy surgery, suffer tremendous psychological pain. In addition to the frightening diagnosis, they have to cope with what they feel is a loss of their femininity. At present, plastic surgeons can reconstruct the female breast by means of a Latissimus Dorsi Myocutaneous flap procedure and the use of a silicone gel mammary implant. The implants come in various sizes and usually do well to restore breast contours. An attempt to recreate the areola and nipple of the breast involves another surgical procedure. Grafting necessitates harvesting properly pigmented mucosa from a donor site, such as the vaginal labia. Therefore, it requires two surgical sites to reconstruct a nipple during this second surgical procedure. As healing proceeds, however, the protrusion of the nipple is usually lost due to scarring (Figure 1). The whole purpose of this second surgical procedure is to achieve nipple protrusion to allow for symmetry when the patient is wearing sheer clothing or a bathing suit. Until a surgical technique is perfected to achieve this goal, a very simple prosthetic technique can be used to achieve the same result and eliminate the need for the second surgical procedure.

TECHNIQUE

Sculpturing and positioning guidelines are drawn on the patient with an indelible ink pencil before making the impression (Figure 2). A horizontal line is drawn at the level of the protruding nipple of the normal breast. In the case of a bilateral reconstruction, this line is drawn slightly below the center of the breast at the point of its greatest curvature. A vertical line is drawn perpendicular to the horizontal line at the same distance from the mid-line as the normal nipple (this vertical line should be 1½″ to 2″ medial to the distal end of the clavicle depending on the size of the breast and age of the patient). The opposite normal breast is the best guide. In addition, four dots are placed around the circumference of the opposite normal areola to mark its extent. This

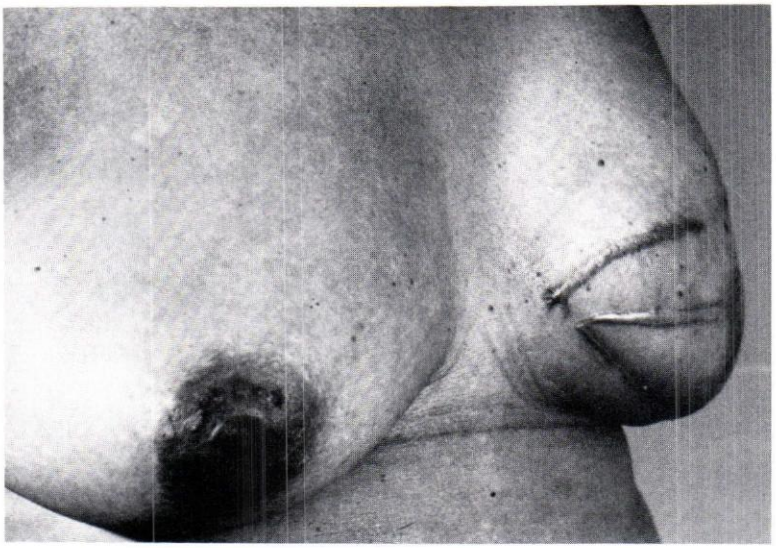

Figure 1. Breast contours restored with Latissimus Dorsi: Myocutaneous flap and silicone gel mammary implant. Nipple protrusion was lost during healing.

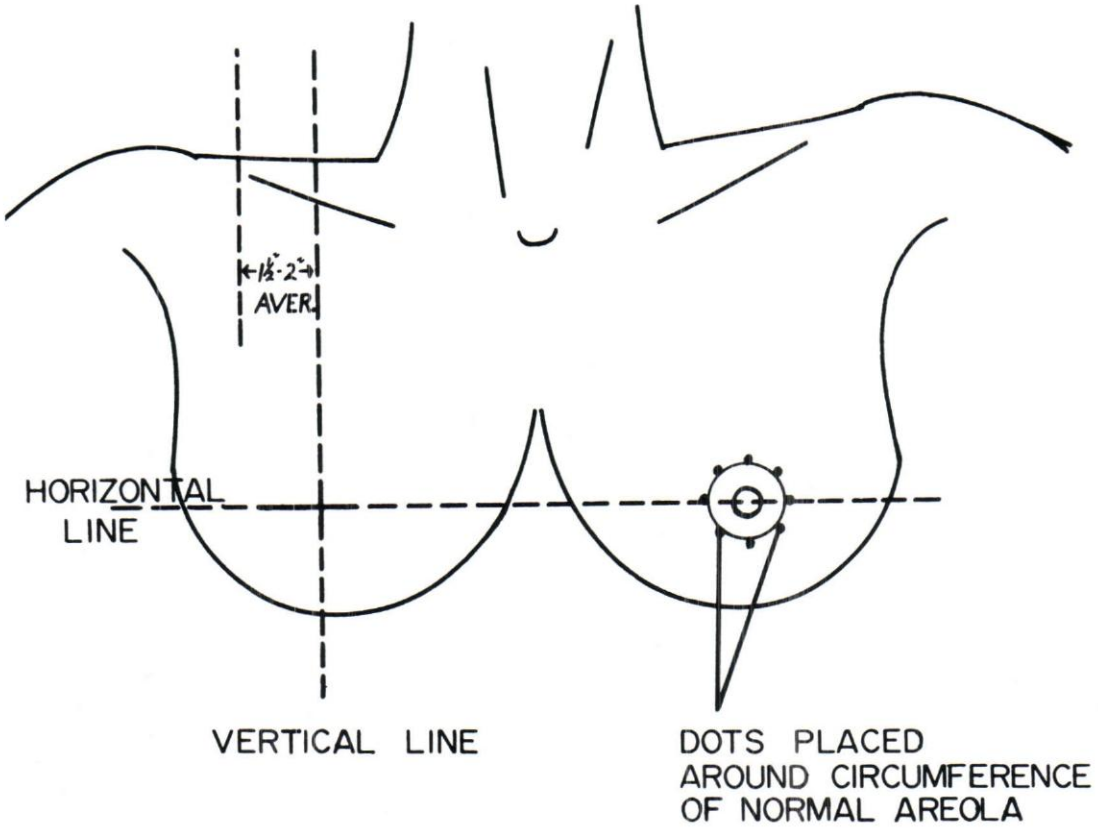

Figure 2. Diagram of guidelines drawn on the reconstructed breast before making the impressions.

border is not always apparent on the positive model.

An impression is then made of each breast[1] using Irreversible Hydrocolloid,* gauze, and accelerated dental plaster.** The patient is kept in an upright position and the impression is kept as thin and light as possible to minimize skin distortion. The indelible pencil guidelines will be transferred to the negative impressions. Molten Base Plate wax*** is poured into the negative impression of the normal breast to the level of the border of the areola (Figure 3). This will yield the initial sculpture of the prosthesis. After this wax-up is recovered, the impression is filled with dental plaster, to produce a positive model. Before the patient is dismissed, a basic shade is mixed by blending MDX-4-4210 Silicone† with earth tones and opaquers.†† Flocking material can be added to simulate fine vascularity. Surface characterization can be accomplished at the time of delivery to further customize the shade.[2]

The wax-up is oriented on the positive model by centering the protrusion of the nipple over the intersection of the two lines on the model. The wax-up is properly adapted, the margins are sealed to the model, finished to a knife edge, and textured to blend in with the rest of the areola area. By making the small areola prominences more defined in one area, the operator can make it easier for the patient to tell the top from the bottom of the prosthesis (or left from right in a bilateral case). If a "try-on" is desired, it must be done before the margins are sealed to the positive mold.

The plaster mold is created by trimming and boxing the positive model, and then pouring the upper half of the mold with plaster. After the wax is eliminated from the mold by boiling water, the mold is

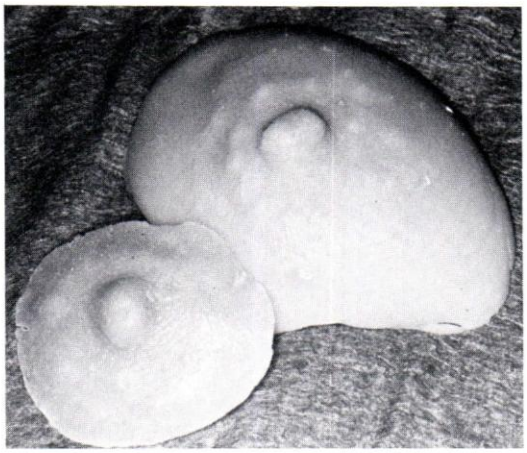

Figure 3. Wax-ups recovered from impressions of normal breasts.

packed with appropriately shaded MDX-4-4210 silicone and cured for 45 minutes in boiling water. After the shade has been mixed and the hardener added, the silicone can be placed under a vacuum to eliminate porosity. In cases where the nipple and areola are two different shades, the two shades are mixed separately. The nipple is first packed into the mold and feathered out over the areola to prevent a line of demarcation. Then the areola shade is packed in on top of it. After processing, the mold is cooled, and the prosthesis is recovered and trimmed (Figure 4). The diameter is checked with the index dots on the refer-

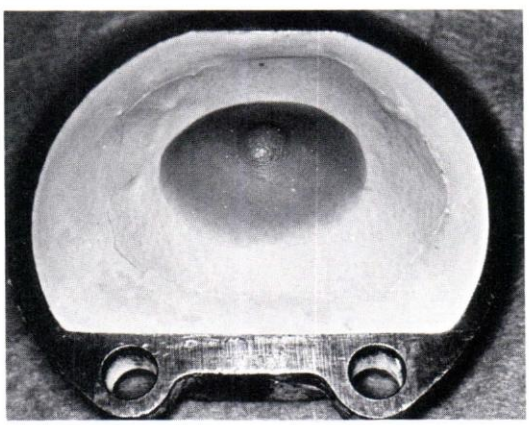

Figure 4. Processed and trimmed prosthesis on drag portion of mold.

*eg. Jeltrate: L.D. Caulk Co.; Milford, Delaware 19963.

**eg. Dental Modeling Plaster; Whip-Mix Corp., Louisville, Kentucky 40217.

***eg. NeoWax: Dentsply/York Division, York, Pennsylvania 17405.

†eg. Dow-Corning: Midland, Michigan 48604.

††Earth Tone Pigments, Kaolin Opaquers, and Flocking Material available from Factor II, P.O. Box 1339, Lakeside, Arizona 85929.

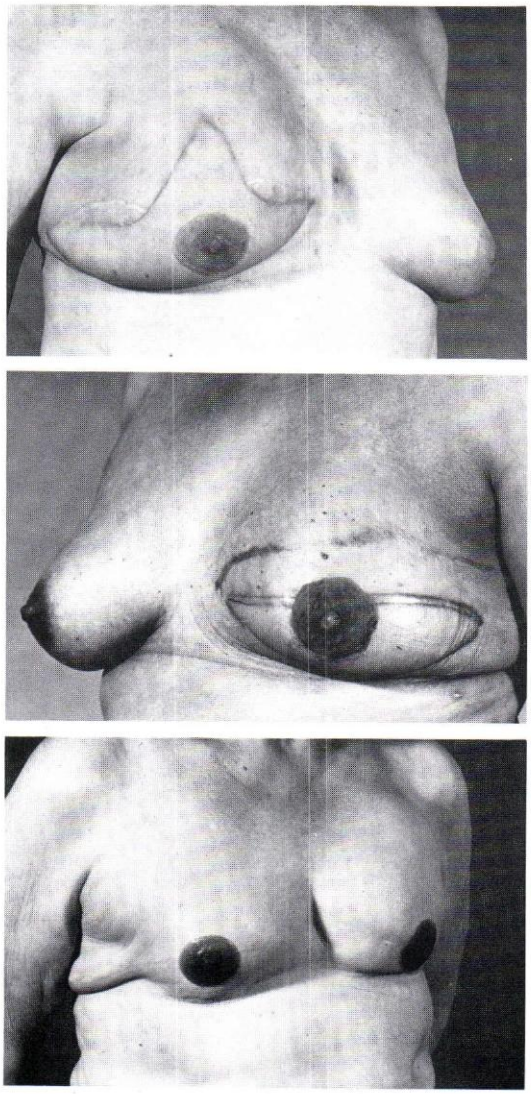

Figure 5. Delivered prostheses on three different patients.

ence model. These prostheses adhere well with Daro Adhesive.††† Patients of the author report showering and swimming with them with no dislodgement.

Sealing of the margins of a nipple prosthesis is really not necessary. But if further characterization or shade correction is desired,[2] this can be achieved with Xylene, Type-A Medical† grade adhesive and Artists Oil Paint (Figure 5).

AUTHOR

Robert L. Engelmeier, D.M.D., M.S., is Assistant Chairman, Department of Prosthodontics, Travis Air Force Base, California.

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The opinions expressed herein are the personal views of the author and do not necessarily reflect the views of the United States Air Force.

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††† Factor II Products, Lakeside, Arizona 85929.

Orthotics and Prosthetics, Vol. 40, No. 4, pp. 63–66.

Technical Note: Improved Techniques in Alginated Check Sockets

Mark A. Abrahamson, C.P.
Keith E. Vinnecour, C.P.O.
David F.M. Cooney, P.T., C.P.O.

INTRODUCTION

One recent prosthetic advancement, the total surface bearing socket design, has been shown to enhance the function and comfort of the amputee. Alginate is one material used by our profession to accomplish this improvement.

Alginate as a prosthetic fitting tool was introduced to the prosthetic profession by Bob Hayes, C.P., in 1975[1] and its use was updated in 1985.[2] Alginate's continued benefit in practice is undeniable. No matter how accurate your negative impression, or detailed your modification, there's always room for improvement. At Beverly Hills Prosthetics Orthotics, as well as in many other facilities across the country, alginated check sockets have become a standard fitting procedure. The alginate procedure is a valuable tool for obtaining optimum prosthetic fittings. Not only does this procedure give the practitioner the opportunity of achieving a true total surface bearing type fit, but also serves as an excellent education and evaluation tool. Each time you complete the alginate procedure, your impression taking and modification deficiencies become apparent. With this information, prosthetists are better able to advance their own skills.

Practitioners who employ this technique have all experienced the frustration of having the alginate come loose from the wall of the socket as the patient removes his/her residual limb. Sanding the inside of the check socket and cleaning it with solvent has in the past been the method of choice, but this is too time-consuming and does not always yield the desired results. A search was therefore initiated for a product or methodology which would keep the alginate adhered to the wall of the socket. A product called HOLD Spray-On Tray Adhesive®* (Figure 1) was located. This was developed for use in dental practices to assist in taking alginate, or hydrocolloid mouth impressions. It has completely re-

Figure 1. HOLD Spray-on Tray Adhesive.®

*Catalogue No. 11461. Teledyne Dental Products, Elk Grove Village, Illinois 60007.

solved the socket/alginate interface problem.

PROCEDURE

It should be emphasized that the alginate procedure is not a cure-all for a poorly fitting test socket. Any deficiencies in the socket weight-bearing characteristics must be rectified before advancing to the alginate step.

The procedure we use for alginating check sockets is an adaptation of that used by Jan Stakosa, C.P. of the Institute for Advanced Prosthetics in Lansing, Michigan. The patient is fit with a clear Uvex® check socket.[3] All check socket fittings are done on bare skin coated with Otto Bock Insulating Cream®** (this is preferred over Vaseline® because of its improved feel and ease of removal from the patient's skin). The check socket is then statically aligned. Because the residual limb is now more closely positioned in its proper orientation in the socket for weight bearing, this ensures a more accurate alginating process. When these steps are completed, the check socket is ready for close visual inspection. Skin coloration is used as a visual cue for accuracy of fit. Probing with thin corset stays may be done to obtain further evaluative information. A china marker is then used to map out intended alterations. Excessively tight areas should be relieved; loose areas are marked.

Once the evaluation has been completed, the check socket is removed and the insulating cream cleaned out completely with a solvent. Injection holes are now drilled in the check socket (prior to our use of alginate adhesive, we devised a special tool for this purpose). This drill bit*** (Figure 2), makes a 5/32" hole in the socket and simultaneously drills a 3/8" diameter countersink in the outer half of the socket wall (Figure 3).

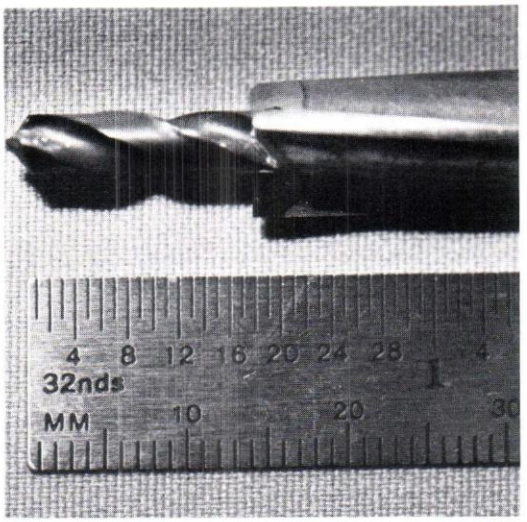

Figure 2. The Otto Bock drill bit makes a 5/32" hole in the socket and simultaneously drills a 3/8" diameter counter sink in the outer half of the socket wall.

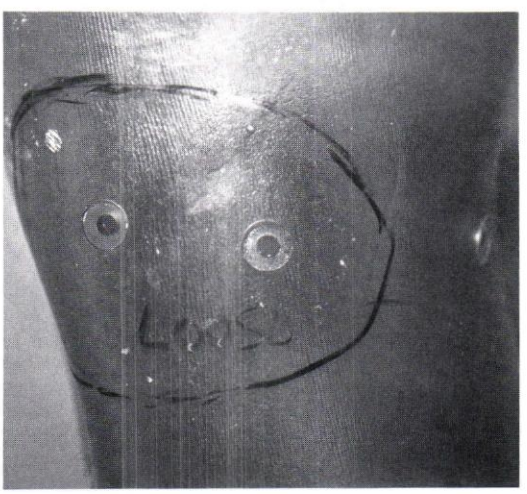

Figure 3.

During the procedure, the alginate collects in the countersink and forms a lock which helps to hold the alginate in place (we continue to use this tool for added insurance). At least three holes are drilled 1" proximal to the most distal point of the socket. These holes act as pressure relief valves and ensure that the patient is completely down in the socket. This also pre-

**Catalogue No. 640Z5. Otto Bock Orthopedic Industry Inc., 4130 Highway 55, Minneapolis, Minnesota 55422.

***This tool is a custom made combination of a 3/8" spot facer with the pilot removed and 5/32" drill put in its place.

vents the patient from getting excessive distal end pressure that results from too much alginate. Additional holes are drilled in the socket over previously marked loose fitting areas (Figure 3). Alginate will be injected into these loose areas with a 60 c.c. syringe while the patient is weight-bearing in the test socket. Relief holes are also drilled over any bony prominences or sensitive areas. All holes should be deburred on the inside of the socket.

The socket is now sprayed with a thin coat of HOLD® tray adhesive. The socket will be ready for application in approximately two minutes and will remain sufficiently tacky for 10-15 minutes thereafter. At this point, a parting agent such as insulating cream may be applied to the residual limb if it is unusually hairy.

The alginate is now mixed according to manufacturer's specifications, and a thin coat poured over the entire surface of the inner socket (Figure 4). This ensures that nothing will come into contact with the HOLD® adhesive and inhibit bonding with the alginate. The residual limb is now placed into the check socket, with the patient bearing approximately 70 percent of his body weight on the prosthesis, maintaining a vertical load, as evidenced by a perpendicular pylon. This will ensure that the patient is properly seated in the socket. If this precaution is not taken, the alginate could hold the patient out through hydraulic pressure and thereby void the fitting. Once the patient is down in the socket, additional alginate is injected through the previously drilled portholes in the socket (Figure 5). Patient feedback is important in order to pinpoint areas that feel loose or tight. All patients report improved comfort once the alginate is injected and has jelled.

At this point, the patient is asked to sit, while keeping the knee extended. A small amount of water, poured into the socket about the knee will help break the suction, thereby allowing the socket to be slowly and gently removed from the patient. Once the socket has been removed, it should be locked in the correct alignment and filled with plaster as soon as possible. The purpose of this is to circumvent the tendency

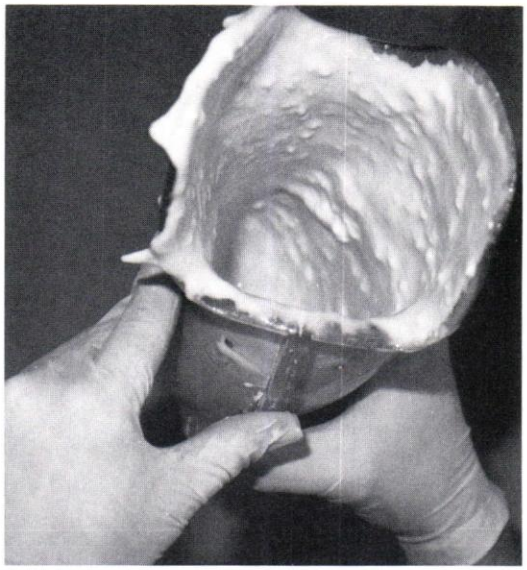

Figure 4. A thin coat of alginate is poured over the entire surface of the inner socket.

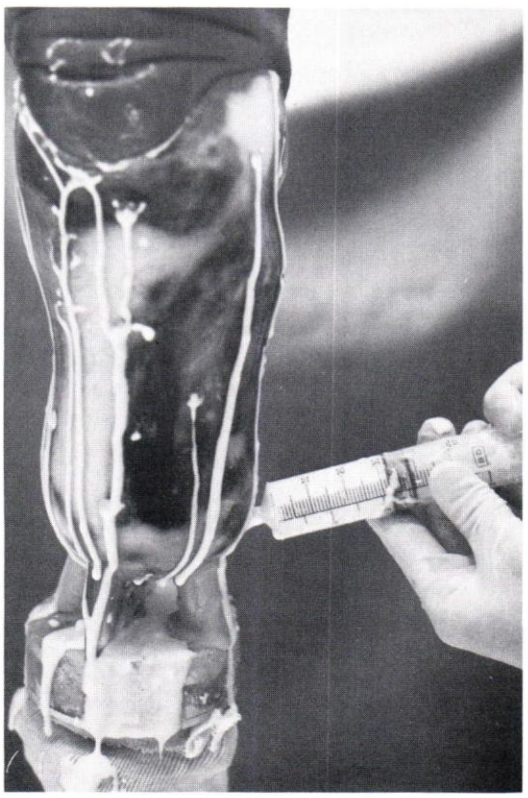

Figure 5. Once the patient is down in the socket, additional alginate is injected through the portholes.

of alginate to shrink and dry out, thereby altering the fit.

After the plaster has hardened, the check socket and alginate are removed from the positive model. The resultant positive model will require minor smoothing before another test socket is fabricated in order to verify the fit resulting from the alginate procedure.

CONCLUSION

Since its introduction to the field, the alginate procedure has proven to be a valuable aid in prosthetic fittings. Unfortunately, the alginate's poor adhesion to test socket surfaces has no doubt discouraged many prosthetists from using this technique on a regular basis.

It is hoped that the solution presented in this paper will encourage more practitioners to discover or rediscover this technique. Once the benefits are seen, it is difficult to avoid using this procedure. This technique will create a prosthesis that provides increased comfort, minimizes tissue atrophy, and improves ambulatory endurance. In so doing, the definitive limb will require fewer post-delivery adjustments and ultimately reduce the necessity for replacement.

AUTHORS

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David F.M. Cooney, P.T., C.P.O., is Chief Prosthetist/Orthotist at Beverly Hills Prosthetics Orthotics, Inc.

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ACKNOWLEDGMENT

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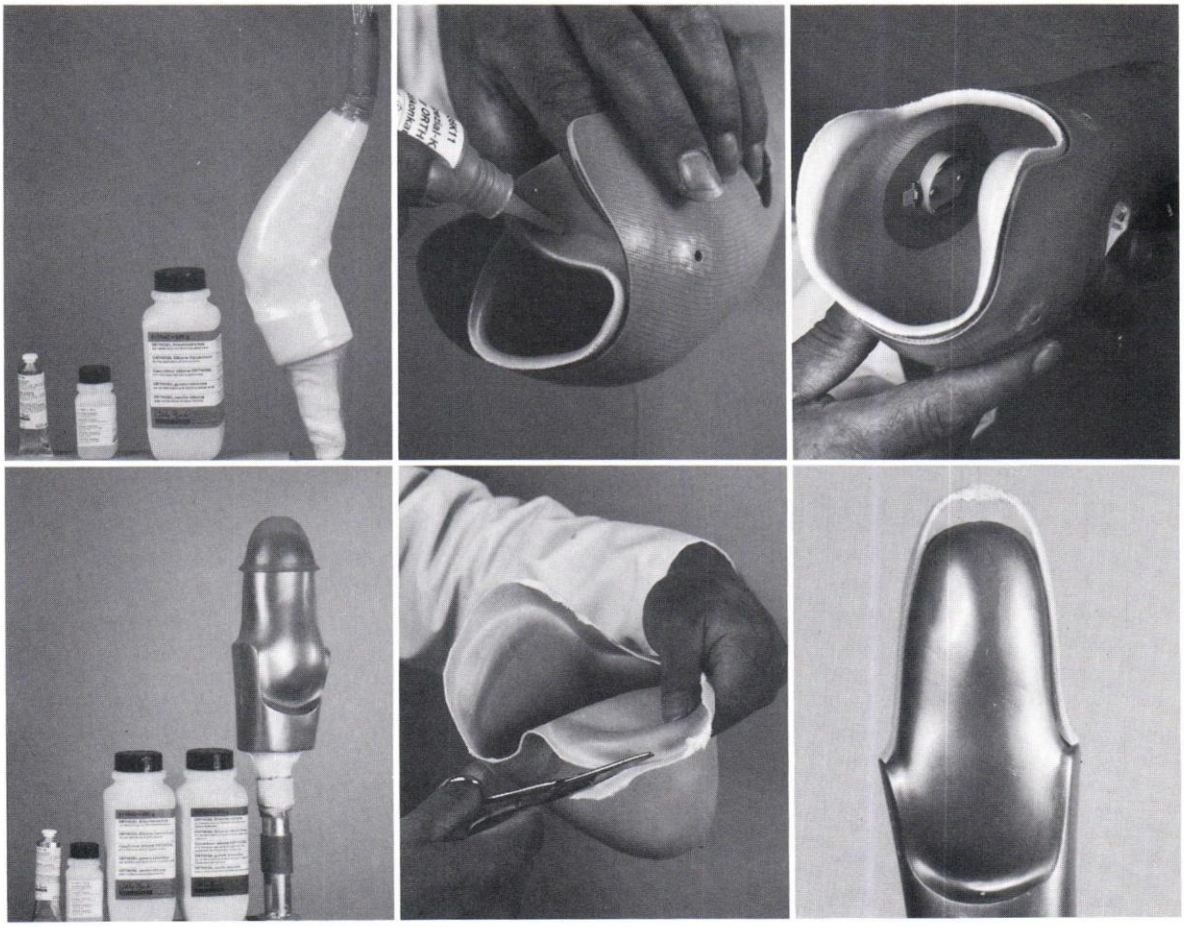

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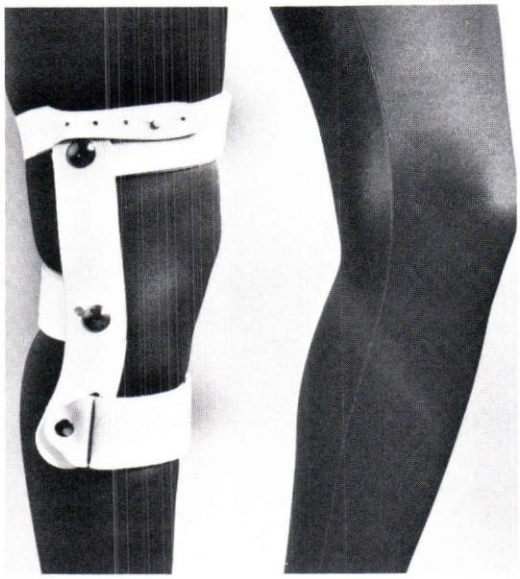

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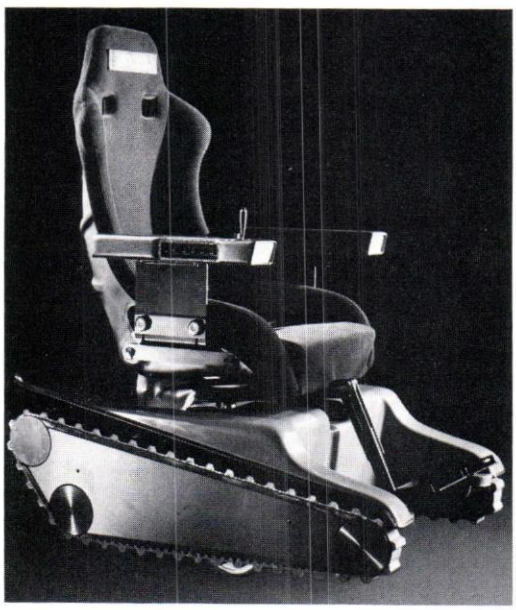

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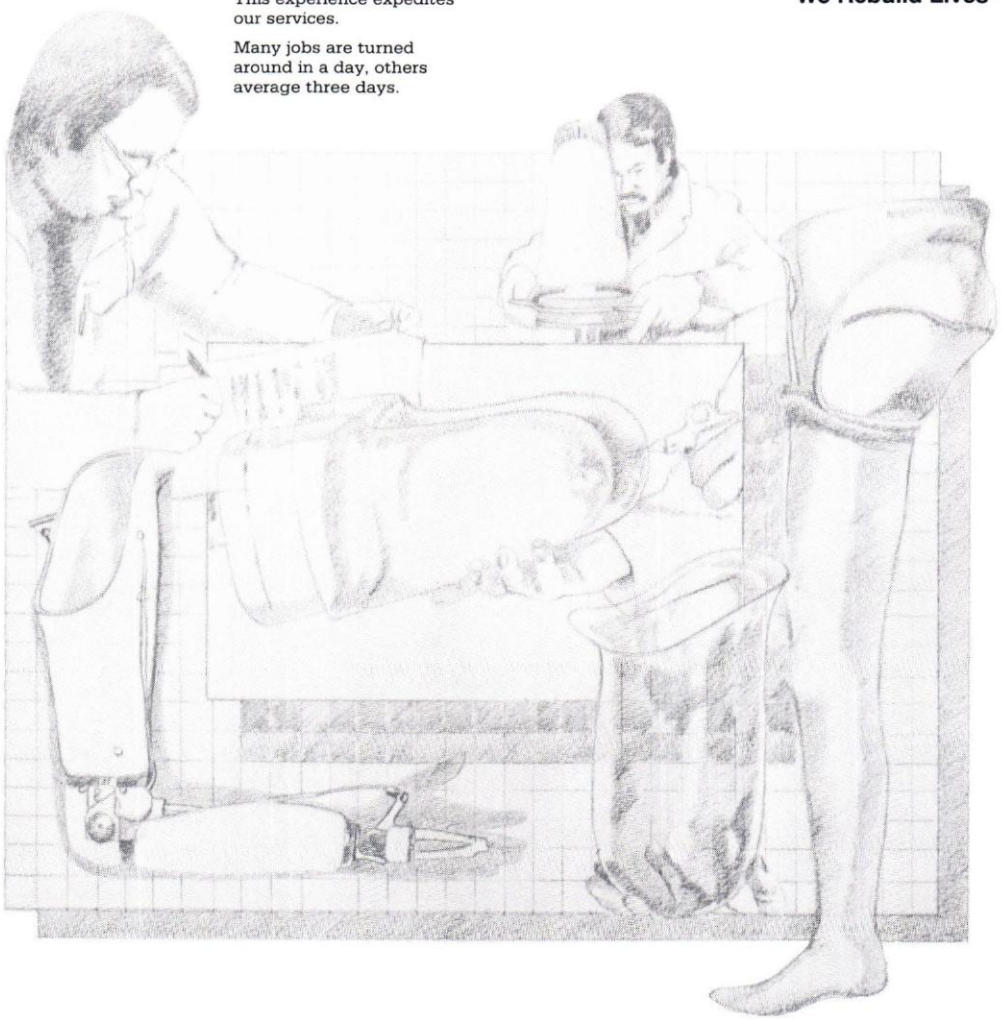

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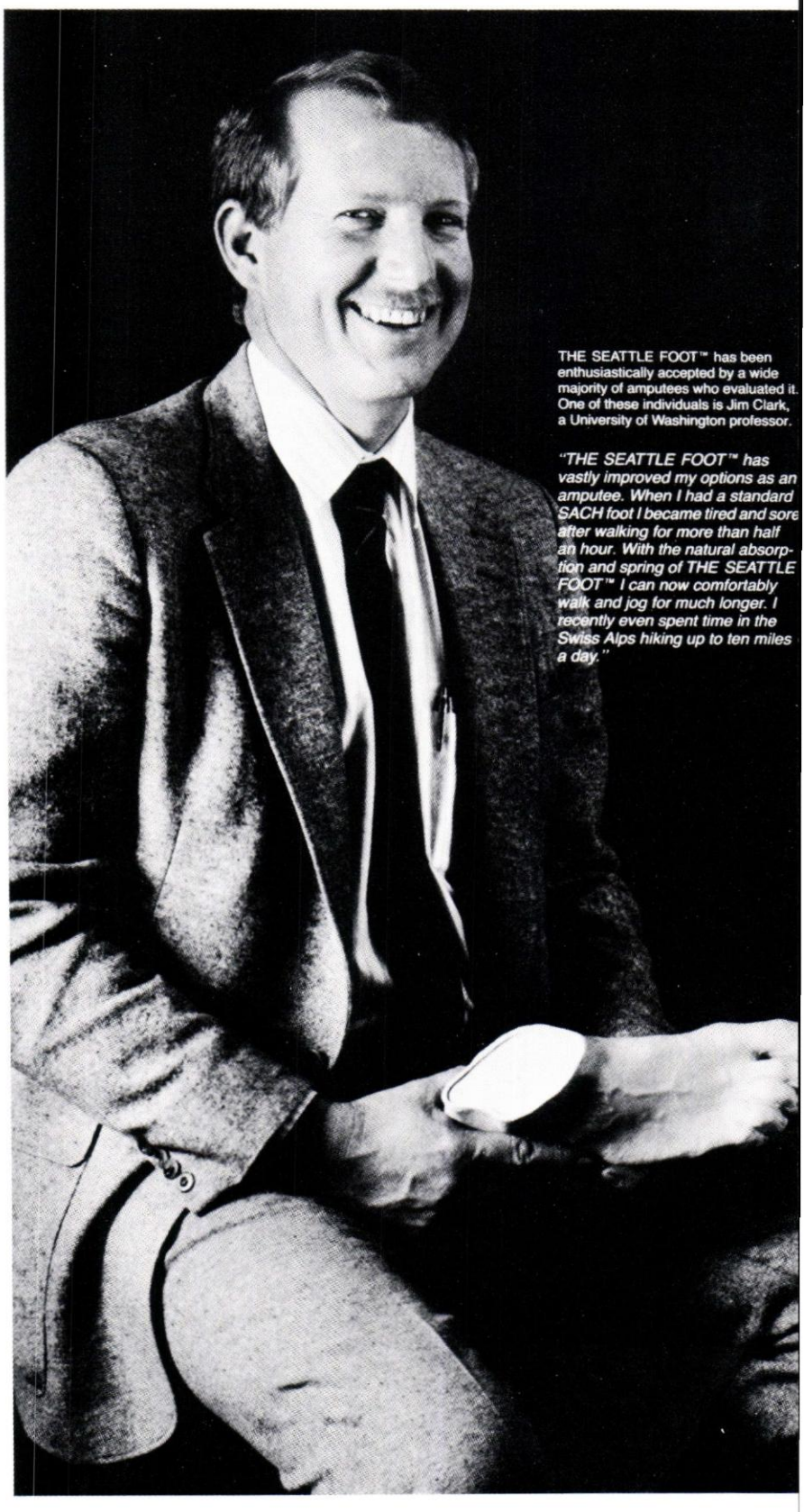

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Heavy Compression (green top) 25-30mm Hg at 50% stretch
Medium Compression (gray top) 10-15mm Hg at 50% stretch
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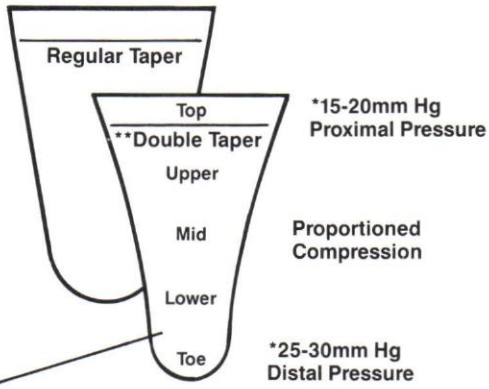

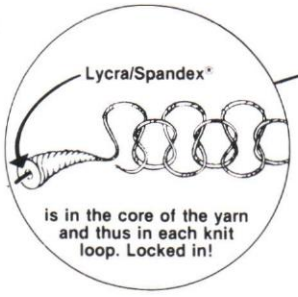

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