MARCH 1976

Orthotics and Prosthetics

Orthotics and Prosthetics

Volume 30, No. 1 March 1976

Orthotics and Prosthetics is issued in March, June, September and December. Subscription price, payable in advance, is $10.00 a year in the U.S. and Canada. Rate elsewhere is $11.00 a year. Single issues, $3.00 each. Publication does not constitute official endorsement of opinions presented in articles. The Journal is the official organ of the publisher, The American Orthotic and Prosthetic Association in collaboration with the American Academy of Orthotists and Prosthetists, and serves as the U.S. organ for Interbor. All correspondence should be addressed to: Editor: **Orthotics and Prosthetics,** 1444 N St., N.W., Washington, D.C. 20005. Telephone, Area Code 202, 234-8400.

Orthotics and Prosthetics is indexed by Current Contents/Clinical Practice.

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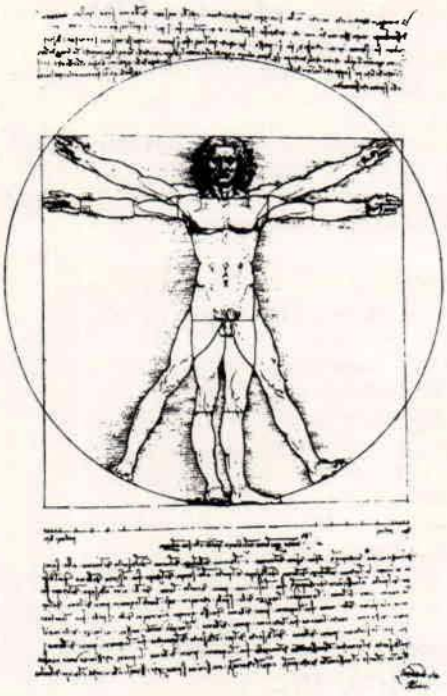

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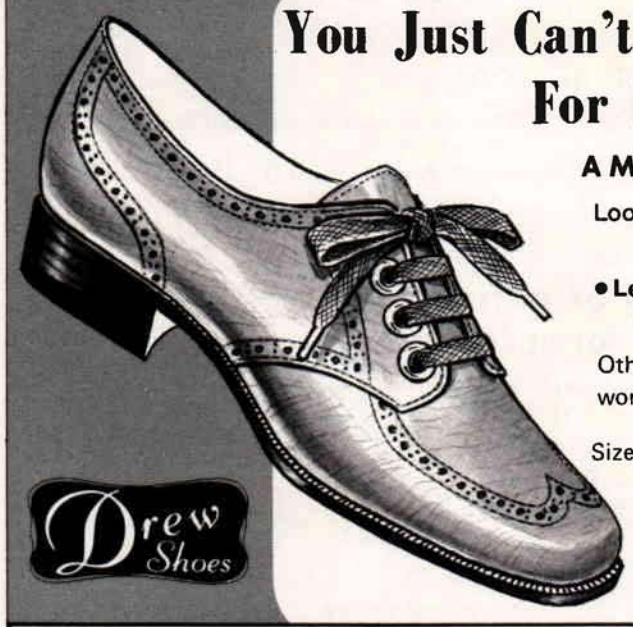

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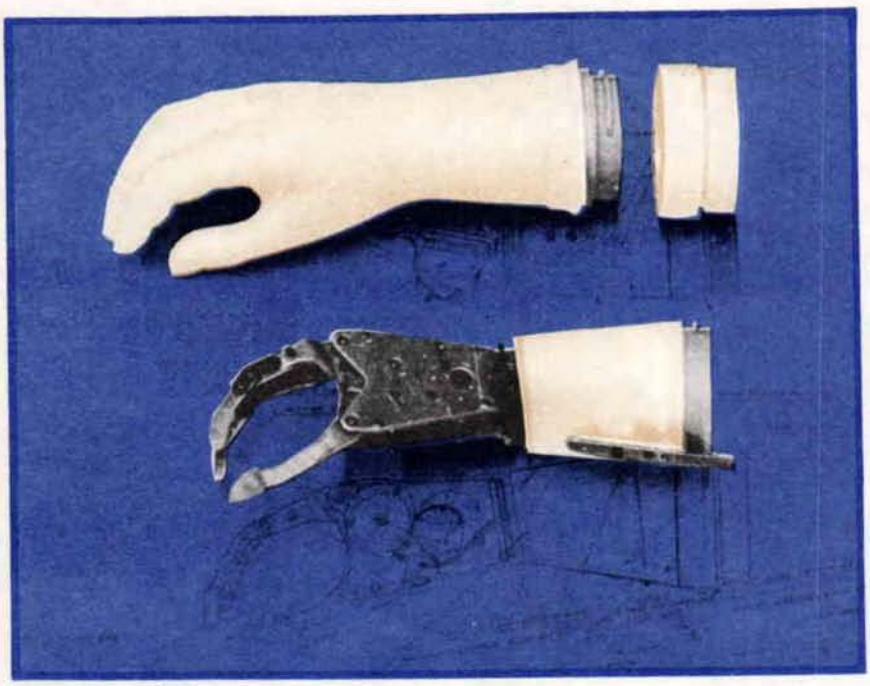

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1976

WITH THE TIMES

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are taken routinely quite for granted.

For the first ten years or more the education program was designed to disseminate the results of research to practicing prosthetists, therapists, physicians, and surgeons. However, the following generations gradually demanded preparatory education programs, and the education programs were modified to meet these logical demands. Today it is difficult to visualize a prosthetics and orthotics service in this country without the contributions made continuously by the various components of what we refer to as the Prosthetics–Orthotics Education Program.

The American Board for Certification in Orthotics and Prosthetics, has progressively demanded more stringent requirements in basic education before allowing orthotists and prosthetists to qualify to stand for their board examinations.

Unfortunately virtually all of the efforts of the formal education program are devoted to the basic training of young people, and almost no effort is being put forth to offer continuing education for the certified practitioners. To compound this problem the "schools" seem to have ignored the advances achieved by various research groups and in many areas are lagging behind the more advanced practitioners. Outstanding examples of this are vacuum forming techniques in orthotics and prosthetics and the use of check sockets in prosthetics.

The American Academy of Orthotists and Prosthetists has done much to fill this void and perhaps will, in the future, meet the needs of prosthetics and orthotics clinic teams for continuing education. If we, as practitioners, are to continue to improve our services to patients we must somehow see that the various education programs are kept up to date. The responsibility seems to be a most appropriate one for the American Academy of Orthotists and Prosthetists to assume.

Howard R. Thranhardt

Orthotics and Prosthetics, Vol. 30, No. 1, pp. 3–11, March 1976

MEETING THE CHALLENGE OF PARTIAL HAND AMPUTATIONS

Leonard F. Bender, M.D.[1], and
Richard D. Koch, C.O.[2]

Amputation of a part of a hand may create significant functional limitations for the amputee that are difficult to ameliorate by either orthoses or prostheses. We have attempted to provide both function and cosmesis to many persons with partial hand amputations and have been confronted by a variety of patterns of amputations and many different vocational and avocational needs. To help us develop a logical approach to devices for partial hand amputations we have studied the losses and divided them into categories.

Loss of a portion of a hand occurs primarily through trauma. In Michigan, a mixed industrial and farming state, we frequently see partial hand amputations resulting from accidents with punch presses, metal shears, conveyor belts, meat grinders, corn pickers and threshers. Less commonly, explosions, bullets, shredders, and thermal injuries cause amputations; rarely are fingers and hands lost because of vascular disease and malignancies.

Traumatic amputation of a part of the hand in an industrial accident usually occurs along a straight line through the phalanges or metacarpals but may be at any angle to the line through the distal row of carpal bones. It may remove, spare, or partially involve the thumb.

Farm accidents are more often uneven at the amputation site due to the tearing, clawing or shredding action of farm implements.

Congenital skeletal deficiencies of the fingers and hand account for a small number of patients who seek hand prostheses. In these cases, the pattern of absence is also variable but usually quite different from that of traumatic amputation.

The outcome for each patient has been dependent first upon the level and extent of amputation and then upon the expressed preference for either function or cosmesis plus specific requirements made of the device. An amputee who is interested predominantly in cosmesis should get what he wants and not have some more functional but less cosmetic device forced upon him. His present vocation and his avocational desires must be considered carefully. The device must always be designed primarily to meet the amputee's needs. If it is obvious that one device will not fulfill all his needs, but two or three will do so, several devices should be prescribed.

The primary purpose of this paper is to discuss devices which have been fabricated for persons with partial hand amputations.

We have found it helpful to divide partial hand losses into several categories. Our classification of partial hand amputations results in four categories (Fig. 1):

1. Transphalangeal, thumb spared.
2. Thenar, partial or complete.
3. Transmetacarpal, distal; thumb spared or involved.
4. Transmetacarpal, proximal; thumb spared or involved.

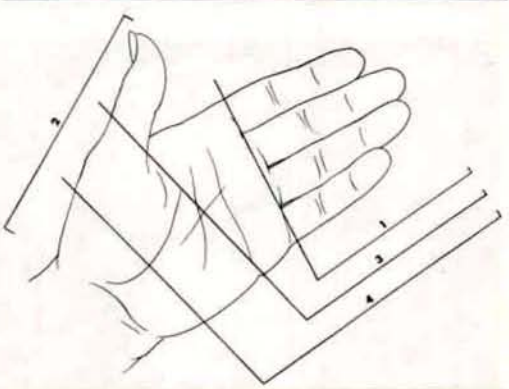

Fig. 1. Levels of partial hand amputation: 1. Transphalangeal; thumb spared. 2. Thenar, partial or complete. 3. Transmetacarpal, distal; thumb spared or involved. 4. Transmetacarpal, proximal; thumb spared or involved.

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[2]Director, Department of Orthotics and Prosthetics, University of Michigan Medical Center, Ann Arbor, Mich.

TRANSPHALANGEAL AMPUTATIONS

Transphalangeal amputations usually involve the four fingers in similar or slightly varying degrees. Most often these patients need no device but they may desire cosmetic fingers. These can be fitted easily when at least half of the proximal phalanx remains for each finger. Cosmetic fingers may be held on by suction or by special adhesives. They can be cut from a cosmetic glove made for prosthetic hands or may be custom made. Occasionally such a person needs and desires additional function.

CASE ILLUSTRATION

A thirty-year old male lost the fingers on his right hand in an industrial accident involving metal shears. A portion of the proximal phalanx of digits two through five remained with good range of motion and strength at the metacarpophalangeal (MCP) joints. The thumb was spared. He was right dominant and wanted to continue to use tools with his right hand. However, the proximal phalanx of the fifth digit was considerably shorter than that of the second digit so power grip was not effective. A stainless steel orthosis with plastisol covering was designed and fabricated (Figs. 2, 3). The device was designed especially to provide power grip to hold a cutting torch, wrench and hammer (Fig. 4). The side MCP hinge moves freely and permits the strength of the finger flexor muscles to flex the opposition bar and provide a satisfactory grip on handles and tubular objects.

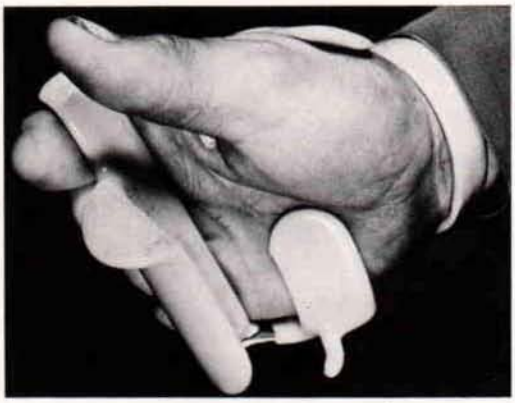

Fig. 2. Transphalangeal amputation of the second through fifth digits. Orthosis with dorsal metacarpal and phalangeal bars, side rivet pivot hinges at the MCP joint.

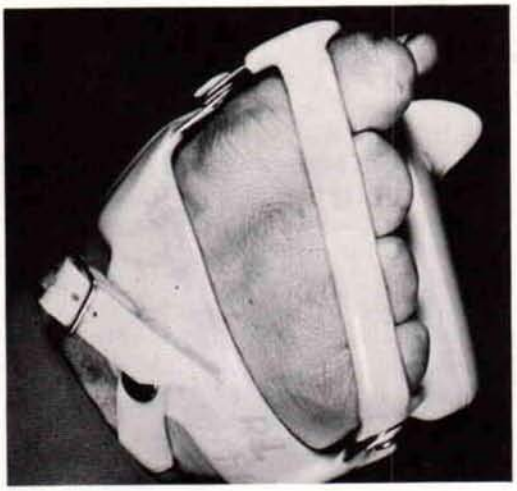

Fig. 3. Palmar view of orthosis designed to improve grasp for a person with trans-proximal-phalangeal amputation.

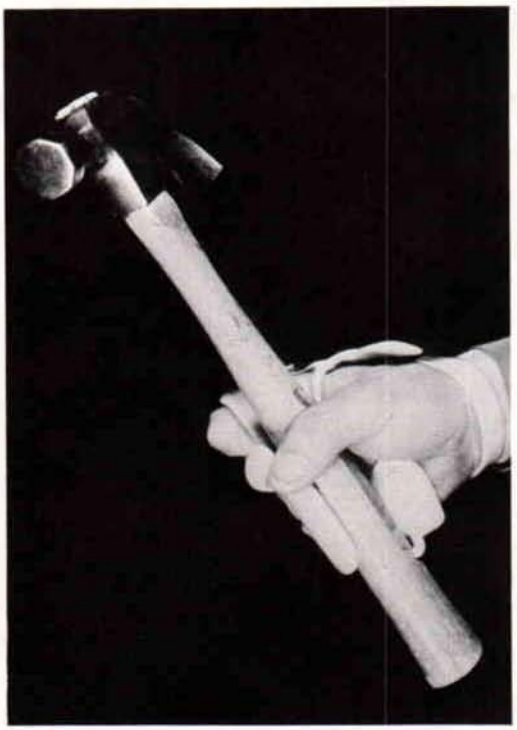

Fig. 4. Effective grasp of a hammer handle between thumb and orthosis.

THENAR AMPUTATIONS

Amputation of the thumb alone is relatively rare in our experience. When it does occur and involves the metacarpal as well as the phalanges, a prosthetic thumb can be made. Plastic laminates can be shaped to resemble closely a thumb and to provide a shallow socket that fits against the second metacarpal and is held on by either a Velcro or plastisol strap around the hand. If other digits adjacent to the thumb are amputated also, the prosthesis can be made with a wider socket and still provide a satisfactory prehension post (Figs. 5A and 5B).

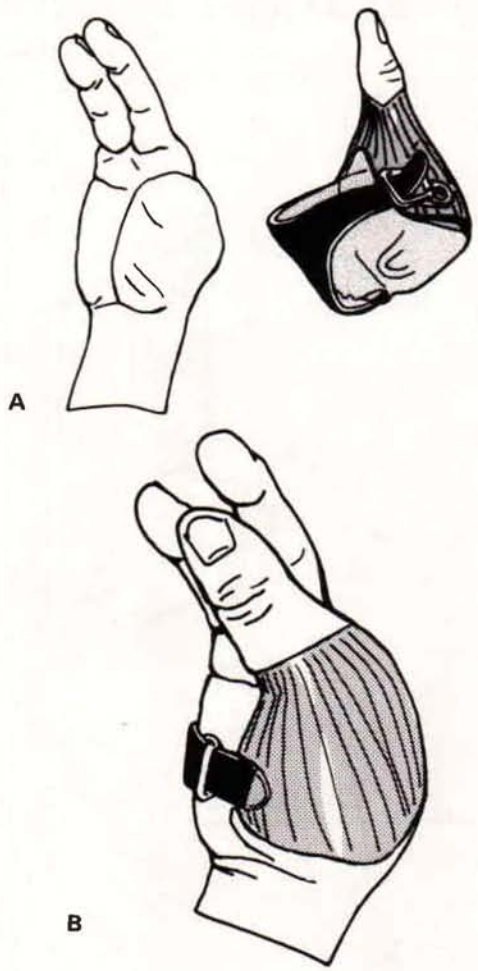

Fig. 5A. Amputation of the thumb, index, and middle fingers.
Fig. 5B. Prosthetic thumb fits over stump of metacarpals and straps to hand.

Congenital absence of the thumb occurs rarely and does not require replacement by a device. These children become very adept with the remaining digits and tend to use the index finger as a thumb. Surgical pollicization of the index finger generally appears to be unnecessary.

DISTAL TRANSMETACARPAL AMPUTATIONS

Amputations through the distal transmetacarpal area are quite common in our industrial location. Because most of these persons have been involved in strenuous occupations and probably will continue to do skilled and semi-skilled labor we must make stout devices for them. Some persons will wish a cosmetic hand in addition to a functional device. A custom made cosmetic hand is generally necessary; standard cosmetic gloves with fillers are not satisfactory.

Basically, two alternatives are available to the orthotist-prosthetist in providing functional devices for persons with distal transmetacarpal amputations. A prosthesis shaped like a mitt can be constructed (Fig. 6) or an open steel prosthesis can be designed to fill a specific need (Fig. 7).

The mitt prosthesis must be particularly strong in the distal portion. We recommend using six layers of glass cloth plus one layer of one-half ounce Dacron felt over a sheet of 0.080 inch thick stainless steel. The palmar area is lined with neoprene to provide friction. This prosthesis provides a large prehension area for gross activities such as lifting furniture (Fig. 8), and also permits finger tip prehension along the rim of the mitt. The rim is placed where the finger tips would normally be located. This prosthesis will withstand much abuse and is very practical for farmers and persons in laborious occupations.

Stainless steel prostheses are fabricated to the anatomy of each patient; they are not formed over a plaster mold or cast. Each piece is formed, fitted and silver soldered in place individually. To construct an opposition post for an amputation at the transmetacarpal level through the thumb and third, fourth, and fifth digits with only the entire second digit remaining one begins by shaping a palmar piece out of 0.050 inch thick stainless steel (Fig. 9). A steel rod is added from the palmar piece over the distal stump to the dorsal edge of the palmar piece. This rod prevents undesired

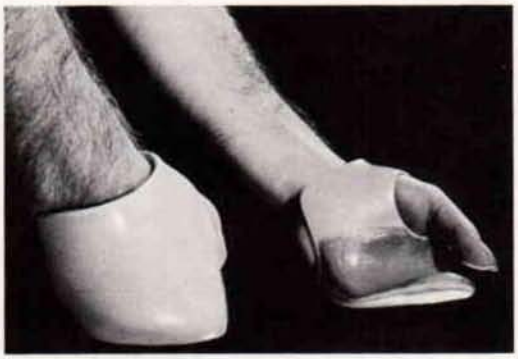

Fig. 6. Mitt-shaped prostheses for bilateral distal transmetacarpal amputations with sparing of the thumbs. (Reproduced from (*1*) by permission.)

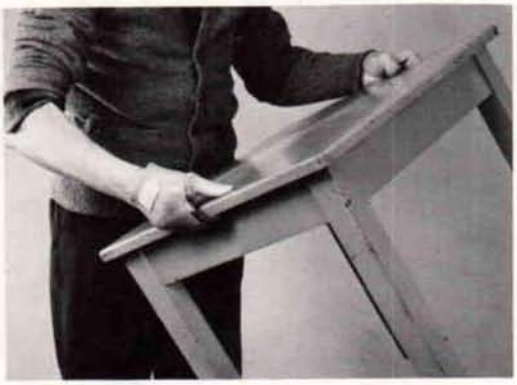

Fig. 8. A mitt-shaped prosthesis can provide adequate prehension force to lift heavy objects.

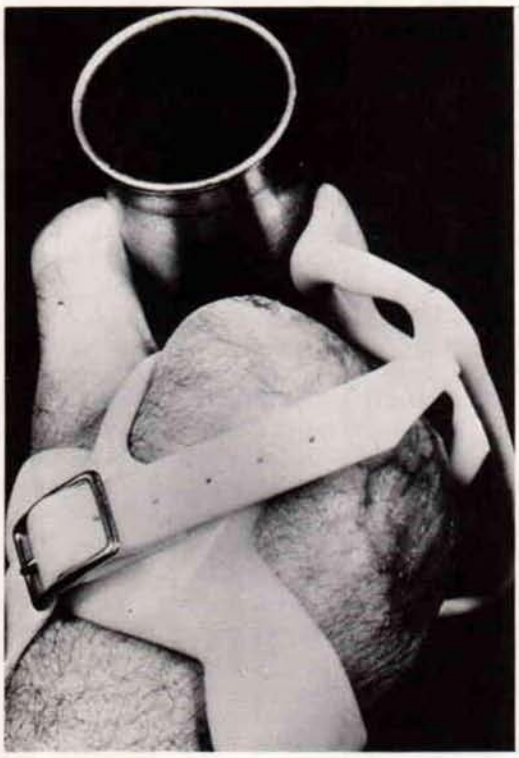

Fig. 7. Plastisol covered, stainless steel prosthesis for distal transmetacarpal amputation with loss of the distal phalanx of the thumb. (Reproduced from (*1*) by permission.)

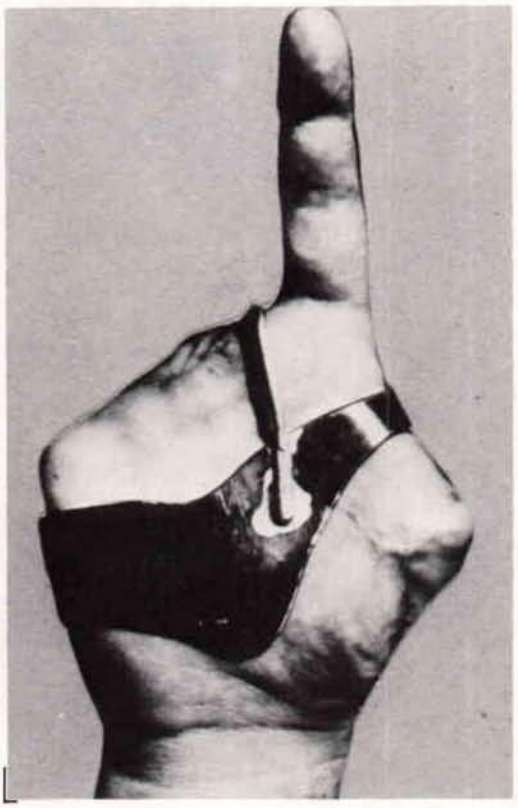

Fig. 9. Palmar piece shaped from stainless steel to serve as a foundation for an opposition post.

movement of the device proximally. The opposition post is then constructed and soldered in position (Fig. 10). Finally, a contoured steel plate is added over the end of the third metacarpal to avoid excessive pressure from the rod and the whole device is covered with flesh-tone plastisol by a dipping and curing process (*2*) (Fig. 11). A plastisol strap around the base of the metacarpals keeps the prosthesis from slipping off the hand distally.

Poor skin coverage of the amputation site is an unusual problem. Surgical techniques permit satisfactory skin coverage of distal amputation sites.

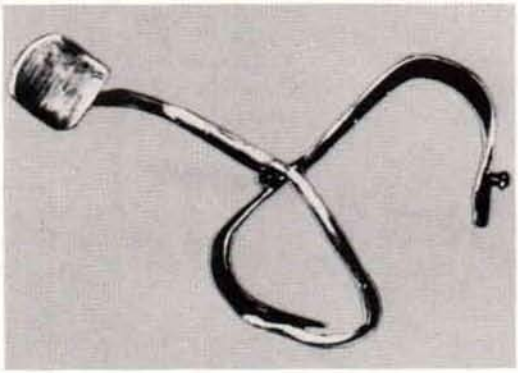

Fig. 10. Steel rod is placed over distal stump to prevent proximal migration of the prosthesis. Opposition post is soldered in position.

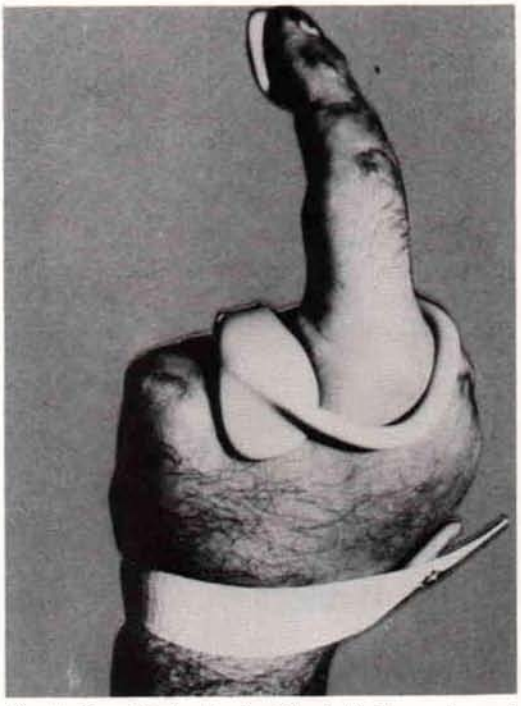

Fig. 11. Completed orthosis with plastisol covering and strap.

CASE ILLUSTRATION

A sixty-year old farmer lost all fingers including the thumb at the distal transmetacarpal level in a cornpicker machinery accident. Adequate skin coverage was not achieved because he did not permit additional surgery to cover the stump with skin graft (Fig. 12). The amputation site was quite sensitive and needed protection. He also desired function; he needed to grasp mechanical milkers firmly so he could attach them to his cows.

A stainless steel prosthesis was designed with a contoured piece of 0.050-inch stainless steel on the dorsal area which wrapped around the fifth metacarpal and stopped at the fourth metacarpal (Fig. 13). A small steel rod was positioned in the web space of the thumb for stability and to prevent proximal movement of the prosthesis. Stainless steel rods, 3/16-inch in diameter, were con-

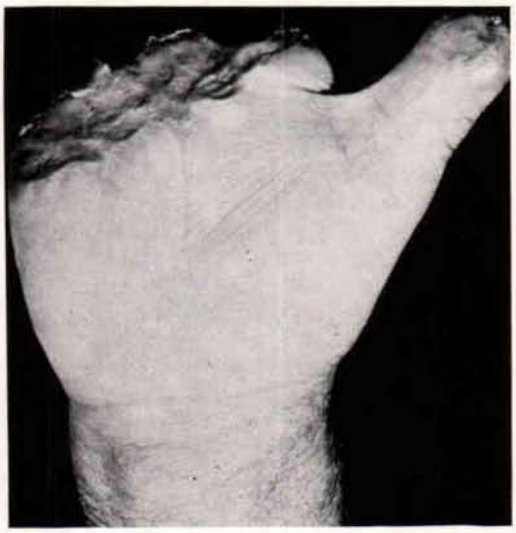

Fig. 12. Amputation through the distal transmetacarpal area of digits two, three, four, and five with loss of the distal phalanx of the thumb. Skin coverage is inadequate for hand activities without protection. (Reproduced from (*1*) by permission.)

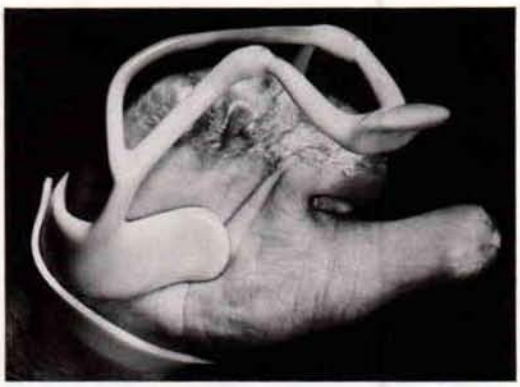

Fig. 13. Stainless steel prosthesis designed for hand shown in Fig. 12. It protects the amputation site and permits prehension. (Reproduced from (*1*) by permission.)

toured on the dorsal aspect and slightly distal to the sensitive amputation site to prevent bumping it, and to add strength to the unit. A second set of stainless steel rods were then positioned so that they would hold a semicircular opposition post in place. The post is silver-soldered to the rods at the appropriate angle to hold tubular objects such as milkers. Proper design and positioning of the post requires especially good judgement. One must consider carefully the length, range of motion, and strength of the thumb, as well as the size, shape, and weight of the objects to be picked up. Plastisol coating (*2*) provides a non-absorbent, resilient, pleasing appearance but the plastisol must be roughed up on the prehension surface of the opposition post to provide adequate friction for good function.

PROXIMAL METACARPAL AMPUTATIONS

The thumb may be spared, partially amputated, or absent in amputations through the proximal metacarpal area of the hand. Either cosmetic hands or functional devices or both may be used to restore appearance and function. The functional devices and, to a lesser extent, the cosmetic hands are more difficult to construct at this level of amputation because the short stump provides a less stable base. The device may need to extend above the wrist to gain stability or assist function.

CASE ILLUSTRATION

A twenty-nine year old female lost all fingers and the thumb on the right hand in a punch press accident (Fig. 14). A right-dominant person, she desired cosmesis and sufficient function to hold small objects and feeding utensils. Appearance of her hand was restored by a cosmetic plastic glove with semi-rigid fillers in the fingers and a zipper fastening on the volar aspect of the wrist (Fig. 15). An opposition post, similar to the one shown in Figure 11, was constructed with a short post to oppose the stump of the first metacarpal (Fig. 16). It provided the desired function.

In some traumatic amputations of the hand, damage may be done to proximal structures by the violence of the injury; the peripheral nerves and brachial plexus are particularly prone to such injury.

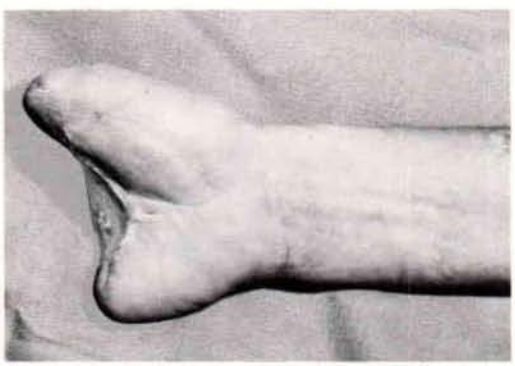

Fig. 14. Amputation through the proximal area of all metacarpals as the result of a punch press accident.

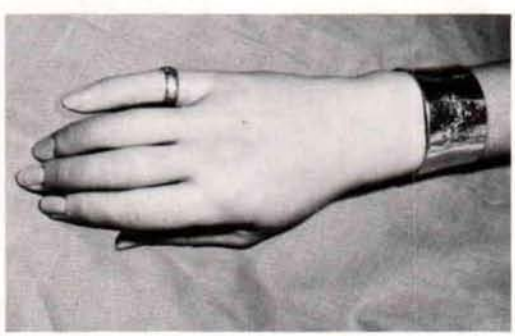

Fig. 15. Cosmetic hand and glove for hand shown in Fig. 14.

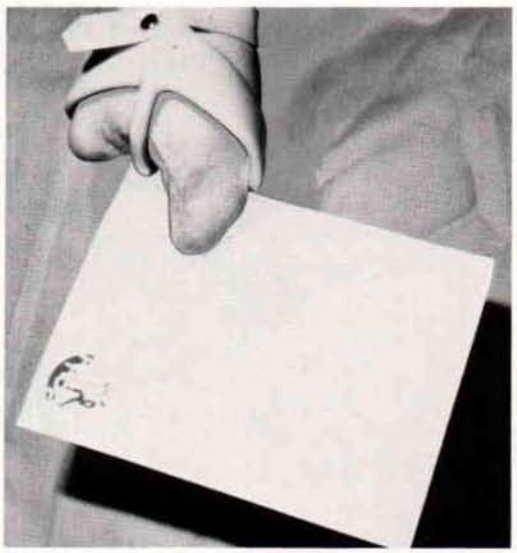

Fig. 16. Prosthesis with opposition post provides prehension for hand shown in Fig. 14.

CASE ILLUSTRATION

A twenty-five-year-old machine operator caught his left hand and clothing in a grinding machine that twisted his arm and mangled his fingers. The thumb was spared but all fingers were amputated and weakness of muscles supplied by the radial nerve below the elbow was noted. Wrist stability was added to a steel palmar band with opposition post by means of a plastic forearm cuff and 0.065-inch thick stainless steel straps along the radius and ulna incorporating single pivot hinges at the carpal level (Fig. 17). The hinges must be properly aligned with the axis of wrist flexion and extension. Teflon washers are used in the hinge joint to reduce friction. If wrist extension assist is needed, hooks can be soldered proximally and distally to the wrist joint on the dorsal side and rubber bands or springs used to provide the assistance. Spring tension is adjusted most easily by using rubber bands.

An alternate solution in proximal transmetacarpal amputations with an intact thumb is to fabricate an epoxy resin mitt similar to that shown in Fig. 6. To gain stability the mitt will usually have to encase the carpal area. It can be made sufficiently flexible, yet semi-rigid, so that it can be slipped over the stump and held in place with a single Velcro strap (Fig. 18). The rim of the mitt can be made to any desired shape so utensils and small objects can be picked up as well as providing sufficient opening to grasp larger objects like an electric razor.

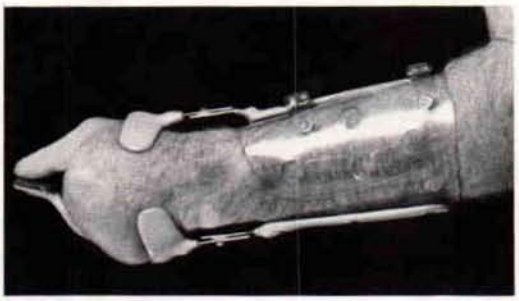

Fig. 17. Prosthesis designed to provide increased wrist stability and prehension between opposition post and thumb when all fingers are amputated but thumb is spared.

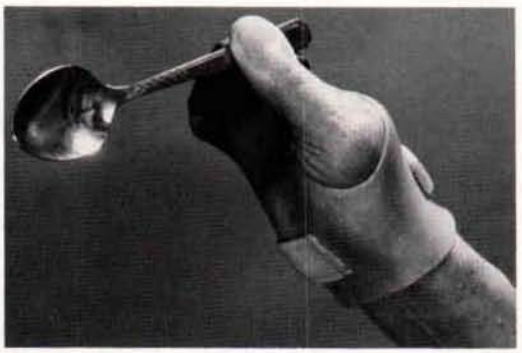

Fig. 18. Epoxy resin, mitt-shaped prosthesis for amputation of fingers at the proximal transmetacarpal area with sparing of the thumb.

CONGENITAL DEFICIENCIES OF THE HAND

Persons with congenital skeletal deficiencies of the hand, other than those with only the thumb missing, can sometimes be assisted by orthotic and prosthetic devices. The pattern of congenital skeletal deficiencies of the digits and rays of the hand is quite variable and even difficult to classify. The proposed International terminology for the classification of congenital limb deficiencies is helpful and those interested in orthotics and prosthetics should be aware of it (*3*).

CASE ILLUSTRATION

A twelve-year-old female with only one digit of the left hand, absence of a portion of the carpus and limited range of motion of the radiocarpal joint desired a new opposition device to replace a worn out volar splint. Active extension at the wrist was possible only to a neutral position but flexion was good. An opposition post was built so that fine prehension and large grasp were both possible (Figs. 19–22).

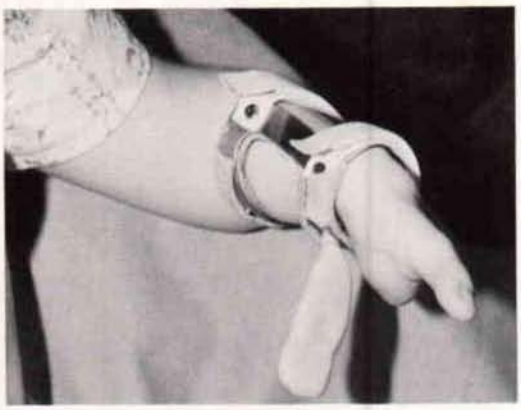

Fig. 19. Opposition platform for congenital skeletal deficiency of the left hand with only one functional digit and limited range of carpal motion.

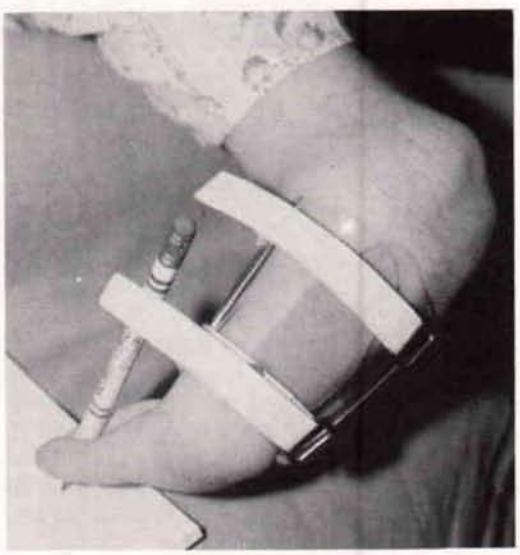

Fig. 20. Use of the opposition platform in writing.

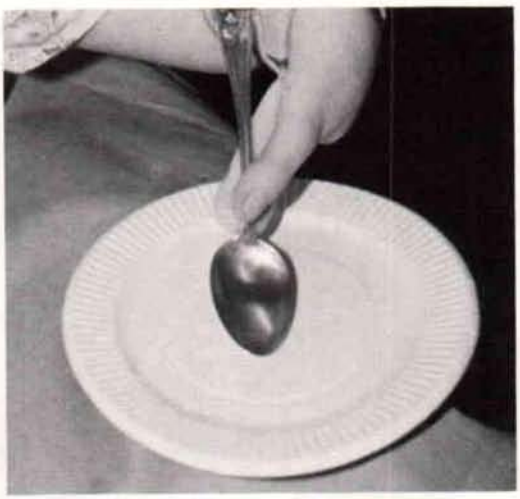

Fig. 21. Use of the opposition platform in eating.

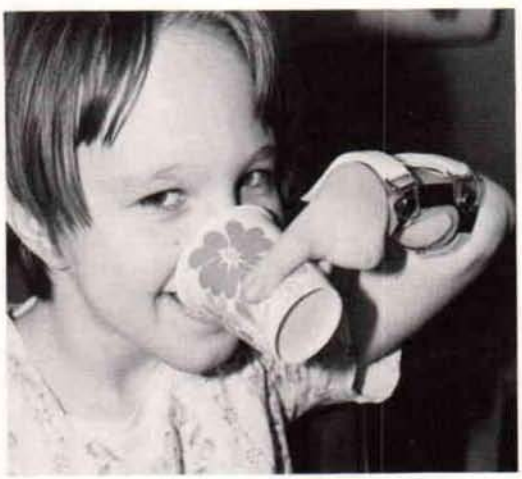

Fig. 22. Use of the opposition platform in drinking.

CONCLUSION

By utilizing basic principles of orthotics and prosthetics and by exercising considerable ingenuity it is often possible to provide considerable improvement in function and/or cosmesis to individuals with partial hand amputations.

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Orthotics and Prosthetics, Vol. 30, No. 1, pp. 13–21, March 1976

ELASTIC MATERIALS AS A SOURCE OF EXTERNAL POWER IN ORTHOTICS

A PRELIMINARY REPORT[1]

John Glancy, C.O.[2]

Sources of external power for use in orthotics have been limited for many years. Unfortunately, it appears that an 'ideal' source will not be available for a generation or two. What practical alternatives can be offered to a large proportion of today's handicapped population and those of the next generation? The use of elastic materials as a source of external power seems to hold promise as a practical alternative for the interim.

Elastic materials, such as shock cord, while limited in meeting requirements in any sense of the ideal, nevertheless can store and deliver surprising amounts of dependable energy. Furthermore, the material can be adjusted easily, is noiseless, lightweight, commercially available, durable and inexpensive.

The ability to store energy is the most important characteristic of elastic materials for orthotic use. They can be made to produce a continuous force that can be put to work in a manner that introduces a new element to spinal and lower-limb orthotics: *dynamics*. The fact that these dynamic forces can be varied in magnitude and made to serve a variety of purposes allows a versatility that portends exciting possibilities.

Some examples of the possibilities afforded by elastic materials in orthotics are:

*A small dynamic force across the axis of an anatomic joint can provide constant stretching of soft tissue to prevent the development of contractures without inhibiting useful motion.

*A preset dynamic force can be applied across an anatomic joint that has imbalanced musculature crossing it in a manner that simulates the missing opposing 'tone' to prevent involuntary swaying, thus aiding balance.

*A single, preset dynamic force can be introduced so that it can change from an extension moment, through dead center, and then to a flexion moment, or the reverse, in a 'programmed' sequence, as angular changes occur in the anatomic joint to which the force is being applied. Thus, a single force can be used to oppose or assist motion in *two directions* in the same plane.

*A present dynamic force can be increased gradually by allowing its fixed ends to be stretched further apart as an anatomic joint moves. The resulting additional force can be used to act as a 'guide' or control for motion activated by muscle power.

*An easily adjusted dynamic force, in the form of an extension moment, can be used postoperatively to maintain and preserve surgical releases of flexion contractures of the hips or knees.

*When the weight of a body segment is not too great, available elastic materials can substitute for absent muscle power and move the segment through space, especially when gravity is used to assist the desired motion.

*In instances where elastic materials are being used to activate segmental motion, adjustment of the dynamic force can also affect the velocity of a segment's movement.

*The dynamic forces that available elastic materials offer are of sufficient magnitude to be of exceptional usefulness to growing children, primarily because of children's relatively lightweight bodies, in the prevention of musculoskeletal deformities that develop after birth.

By providing dymanic resistive force, elastic materials offer stability and control without inhibiting useful motion. The current state-of-the-art is limited almost exclusively to static control. As a consequence, useful motion is constantly—albeit inadvertently—being subjected to interference. It now appears that longterm protection for the musculoskeletal system, without sacrificing freedom of mobility, is orthotically possible.

[1]This project was supported by a grant from the James Whitcomb Riley Memorial Association

[2]Indiana University/Purdue University at Indianapolis, School of Medicine—Department of Orthpaedic Surgery—Division of Orthotics

A SPECIFIC CLINICAL APPLICATION— A PRELIMINARY REPORT

A new look at elastic materials as a source of external power was undertaken as part of an ongoing effort to develop a practical orthotic system of ambulation for paraplegics that would enable them to walk with greater ease and confidence than present-day techniques allow. A practical means of ambulation, with a good deal less dependency upon a wheelchair, continues to be our ultimate goal, and with this in mind, it was evident that first priority must be given to finding an easier way for the paraplegic patient to rise from and return to a wheelchair. This portion on the use of elastic materials as a source of external power in orthotics will limit the *specifics* of their use to our experiences in attacking this multifaceted problem.

The most obvious improvement to the current technique for rising and sitting seems to be a system that will enable the paraplegic patient to get to his feet in a quite conventional manner, starting with the knees flexed in a normal seated position. The more the overall problem was studied, the more the "obvious" seemed to be the most practical.

For a paraplegic to rise from his wheelchair in a smoother, less exhausting way it is necessary to compensate for the lack of sufficient height provided by the arm rests (Fig. 1). Folding "crutches" (Fig. 2) that provide the extra height were designed and installed on a wheelchair as permanent components. The pinlocks on the front casters and the rear wheel locks are engaged before rising to stabilize the wheelchair.

Inability to prevent 'out-of-phase' anterior rotation of his pelvis when attempting to leave the chair while facing forward is another factor that prevents the paraplegic in achieving a standing position easily. The latissimus dorsi muscles used in raising the trunk inadvertantly cause anterior rotation of the pelvis because their insertions are located on the spinous process of the lower thoracic and lumbar vertebrae. Since the antagonists that normally prevent such out-of-phase pelvic rotation are paralyzed, an increase in forward rotation and an accompanying increase in lordosis are under no restraints other than the range limitations of the joints of the vertebrae and hips.

To stabilize the pelvis and lumbar spine against the forces of the latissimus dorsi and gravity while

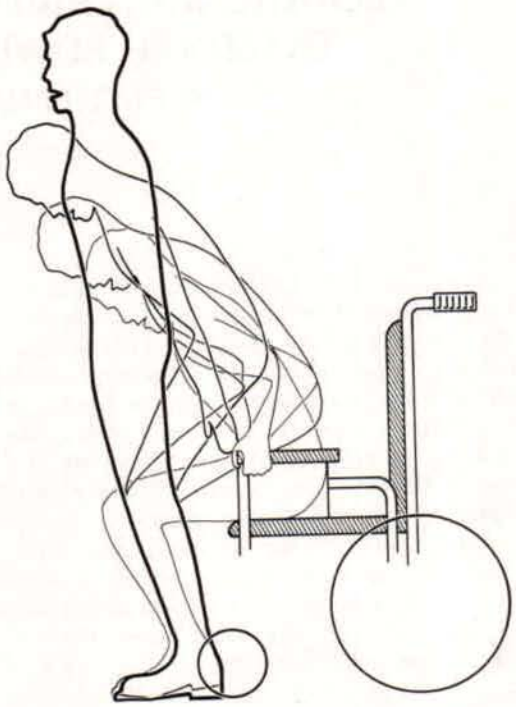

Fig. 1. A schematic view demonstrating the lack of sufficient height of supports for the paraplegic's arms in order for him to lift himself to his feet from the seated position.

Fig. 2. Prototype folding wheelchair crutches: The crutches are designed to be permanent components. Note that the hand grips are turned downward, below the level of the armrests, when not in use.

rising and returning to the wheelchair a modification of a system reported earlier (*1*) was used (Fig. 3). Since paraplegic patients must wear bilateral knee-ankle-foot orthoses (KAFO's), the posterior elastic panels that are attached to the movable pelvic portion of the polypropylene thoracopelvic unit are attached distally to the thigh cuffs of the KAFO's. This arrangement permits the initial, or 'preload', extensor moment to be of greater magnitude because the thigh cuffs are fixed to the uprights of the KAFO's and therefore an excessive buildup of shear forces on the surface of the thighs (of some concern when the thigh cuffs are not attached to fixed points) no longer need be a consideration. However, the characteristics of various elastic materials *do* place a limit on the magnitude of the preload, in that the setting must allow sufficient additional stretching to permit enough excursion of the elastic panels for the patient to sit.

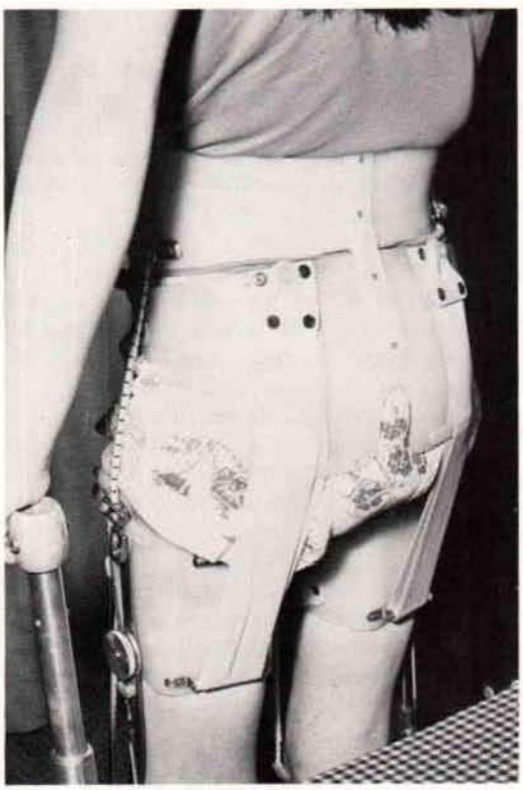

Fig. 3. Closeup of the modified polypropylene thoracopelvic unit with dynamic pelvic extension elastic assists: The elastic is 2" wide and is doubled over nylon rollers attached to the distal ends of the polypropylene quadrilateral thigh cuffs to ensure an even pull throughout. Note the plexidur studs used for the proximal attachments. To don, the patient passes the elastics through the nylon rollers and slips the ends, with their slotted polypropylene pieces, over the studs while lying on her side.

The extensor moment generated by the bilateral elastic panels provides the paraplegic patient with the following functions:

A. While rising from a wheelchair:
   1. Prevention of the lumbar region from going into excessive lordosis;
   2. A resistive force to out-of-phase, anterior rotation of the pelvis,
   3. Substantial assistance to the patient's arms when lifting his body.

B. When standing:
   1. Additional stability by preventing involuntary swaying of the trunk in an anterior direction about the hips' axes. Thus a 'back-up' safety feature is added to the technique of 'hanging' upon the iliofemoral ligaments in order to maintain anteroposterior balance.
   2. Sensory feedback provided by the intimate fit of the thorax portion of the polypropylene thoracopelvic unit because its upper border extends 2 inches above the level of sensory deficit. Any tendency for the trunk to 'jackknife' is felt in the form of increased pressure as it begins to occur, and the motion is readily checked by his arms and crutches before it is beyond control. Related to this feature is the fact that as the trunk rotates forward, the extensor moment (simultaneously generated by both elastic panels), increases proportionally to angular changes occurring in the hip joints and thereby slows down the trunk's forward rotation. The net result is additional time for the patient to react to the 'reading' he is receiving via the thoracopelvic unit.

C. While lowering the trunk into the wheelchair:
   1. An increase in magnitude of the extending moment about the hip throughout the descent. The out-of-phase rotation of the lumbar spine and pelvis is being resisted by the extensor moment generated by the bilateral elastic panels.
   2. Sensory feedback that enhances control of the rate of descent is received via pressure on the thoracopelvic unit.

The folding crutches and the polypropylene thoracopelvic unit with its pelvic extension assist

permitted the patient to rise to a standing position, but, though up, he could not extend his knees fully so as to engage the knee locks of the KAFO's. Consequently, his knees buckled at each attempt to bear weight upon his lower limbs. It was clear that further progress would not be possible without a mechanism capable of automatically extending the knees and locking them in extension.

It was felt that the mechanism would have to extend the tibii fully and automatically, because both arms of the patient are needed for support.

The external power needed would be substantial. Because the weight of the lower limb (including the orthosis), with assistance from gravity, does not produce full extension at the knee joints of a paraplegic patient while his arms support his extended body, a resistive force must be present. The weight of each lower-limb below the knee is calculated to be 1/15 of the body's weight (*2*). The weight of the below-knee portion of each KAFO is about 1 1/2 pounds (.68 kg). A reinforced shoe of the Hush Puppy (R) variety weighs about 1 pound (0.45 kg). For example, the total weight involved, per lower-limb, for a patient who weighs 100 pounds would be about 9.2 pounds (4.1 kg). Ruling out the presence of spasticity in the hamstrings and/or gastrocnemius muscles, or the presence of fixed contractures in either, a 'guesstimate' must be made of the magnitude of force needed to overcome the resistance. Because we do not have information as to the amount of resistive force we are dealing with, clinically, it is necessary to set the initial extension force somewhat arbitrarily. To complete the example given, we chose 6.5 pounds (2.95 kg) (approximately the weight of the lower limb below the knee) as the minimum force that the knee extension assist unit would have to exert upon each lower limb to achieve full extension.

Because of the power limits of available elastic materials to supply an external extension force to the knee joints, an attempt was made to determine just how many *specific* functions that are credited to the quadriceps muscle group would have to be mechanically produced to accomplish the task. An analysis was made of a number of the many functions that the quadriceps muscles normally perform under a variety of conditions. All but four of the functions studied were rejected as either irrelevant to the immediate needs of paraplegic patients or, if a substitute could be delivered, would be impossible for them to manage because of the general limitations of current orthotic development. The four functions of the quadriceps muscles which were felt to be crucial to a solution and therefore would somehow have to be mechanically produced are:

1. The force they generate to resist and/or prevent knee flexion whenever the body's CG moves behind the knee axes when standing. (This force varies from minimal—when checking normal involuntary swaying in the anterioposterior plane—to much greater magnitudes when serving as a control mechanism for any activity that involves flexing of the knees, during which the body's CG is behind the feet.)

2. Their crucial contribution to the complex control forces that enable the normal individual to lower his trunk into a chair at varying rates of speed.

3. Their active contribution to the lifting of the body from a seated position by extending the thighs about the knee axes.

4. Their ability to extend the lower leg about the knee axis, at varying rates of speed, when the lower leg is not bearing weight.

The rate of acceleration of extension of the lower leg must be regulated, if possible. This is important in order to keep to an absolute minimum the time that the patient must support his full weight upon his arms, once he has achieved full extension of his hips and trunk at the completion of his rise from the wheelchair.

A dynamic, automatic knee extension assist that facilitates full extension of the lower leg and allows the automatic spring knee lock of a KAFO to engage is shown in Figure 4. The illustration also shows the type of KAFO that is used with the new system. A stainless steel Becker automatic springlocking off-set knee joint is modified to enable the patient to put the spring-lock in the disengaged position when standing, a prerequisite to sitting. When he wishes to sit down, the off-set knee joints prevent the patient's knees from buckling as he flips the handles (which are connected to the locking cables) to the disengaged position, one leg at a time. Thus, he has a hand free to support himself on either of the extended crutches attached to the wheelchair. As the patient leans against the anterior panels of the thigh cuffs and the anterior panels of the solid-ankle shells below, (*3,4*), the combination of the mechanical knee center being 1 inch behind the an-

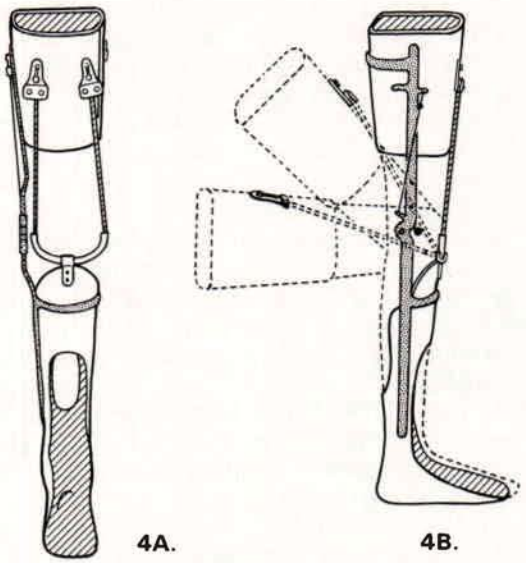

Fig. 4. The dynamic knee extension assist: A. Anterior view showing the unit attached to a KAFO. B. Lateral view illustrating the pivotable action of the knee extension assist unit that shows how the force changes from an extension moment in the upright position to a flexion moment in the seated position. Note the cosmesis of the unit—no part protrudes above the flexed knee. (The solid dot represents the axis of the anatomic knee joint.)

atomic knee center (*5*) and the dynamic extension moment being generated by the looped elastic shock cord, results in a remarkably stable pair of *unlocked* knees.

As the patient lowers himself into the wheelchair, the combined bilateral knee extension force generated by the elastic shock cords offers a substantial resistive force that, in turn, relieves an appreciable amount of the weight that the arms must support. For example, we return to the sample case of the patient who weighs 100 pounds (45 kg). In order to extend each lower limb, it was estimated that each would have to have a dynamic extension force of 6.5 pounds (2.95 kg). Therefore, the total force resisting the weight of the body as it is lowered into the wheelchair would be 13 pounds (5.9 kg)—the sum of both shock cord assists. Since both lower portions of the limbs are resting upon the floor and no motion is occurring below the knee axes, their weight need not be considered as part of the mass being supported by the patient's arms. In the example, the weight of the two lower limbs (exclusive of the KAFO's), totals 13.4 pounds (6 kg). Subtracting 13.4 pounds (6 kg) from the total body weight leaves 86.6 pounds (39.3 kg)—the actual weight that the patient's arms are supporting (exclusive of the thigh and trunk portions of the orthotic system). By dividing the actual weight being supported, 86.6 pounds (39.3 kg), into the total resistive force of 13 pounds (5.9 kg), we find that the arms are being relieved of approximately 15% of their burden. Conversely, the arms are provided the same assistance (about 15%) when lifting the body up from the wheelchair.

Figures 4, 5, and 6 show some details of the components of the system. The knee extension assist unit weighs about 3 oz. (105 gms). The semicircular polypropylene tubing is allowed to swivel so that the force of the shock cord provides an extensor moment in the extended position and a flexion moment in the seated position.

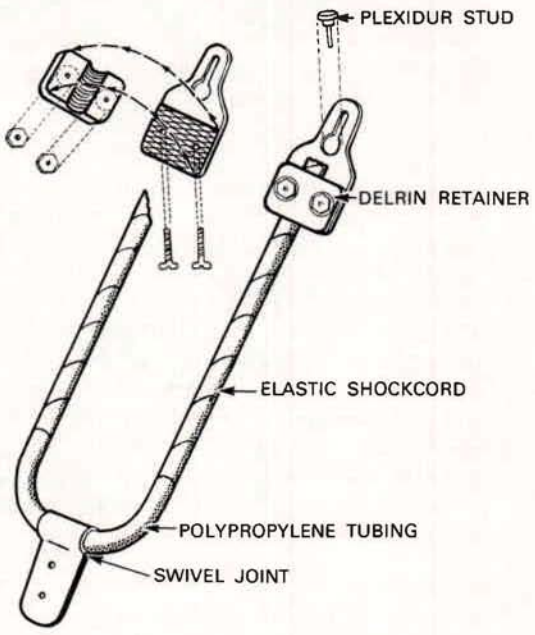

Fig. 5. The dynamic knee extension assist unit: Detail of the components.

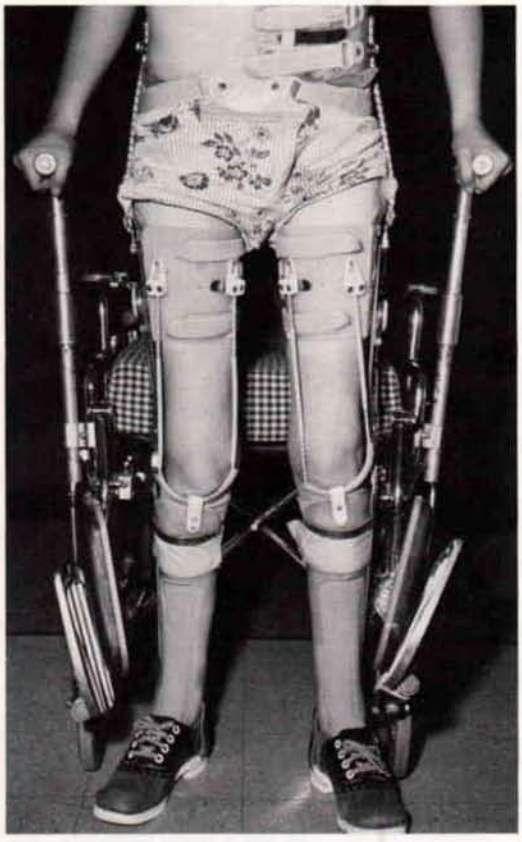

Fig. 6. The dynamic knee extension assist unit; Close-up showing the units on a patient's KAFO's.

Figure 7 shows a patient rising from her chair while using the folding crutches, the thoraco-pelvic unit with extension assists, the knee extension assists, and the dual locking, single, lateral upright KAFO's with offset knee joints. The patient is a 16-year-old, T10 level, traumatic paraplegic with a surgical fusion of L2-3 and with hip flexion contractures of 15 degrees on the right and 10 degrees on the left. It is 3 1/2 years since her accident. The patient weighs 115 pounds. Note the degree of pelvic control and how the dynamic knee extension assists change from a flexion to an extension moment as she rises.

ADVANTAGES TO THE PARAPLEGIC PATIENT

As the patient rises from his wheelchair the shock cord provides a gradually diminishing flexion moment for approximately the first 25–30 deg. of his ascent. This insures that his lower legs will remain under him until he gets enough of his weight over them to the point that they will not slide out from under him as he is rising.

For the remaining 60 deg. to full extension the shock cord provides an extension moment of increasing magnitude, with the minimum at approximately the 35 deg. point. The axis of the offset knee joints being 1 inch behind the anatomic knee axes add to the magnitude of the extensor moment as the patent's thighs press against the anterior panels of the quadrilateral cuffs. The pressure of the thighs is transferred down the uprights to the mechanical axes and increases the radius perpendicular to the extension force generated by the shock cord. The net effect is additional assistance to the lifting of the trunk and greater stability of the lower legs as the trunk is lifted up and over them.

When the patient is seated the force generated by the shock cord is constant. Therefore, the femurs are placed in compression. The pelvis is pressed against the femoral heads by the force generated by the elastic panels of the thoraco-pelvic unit and the condyles are being pressed toward the pelvis by the force of the shock cords. These two forces, combined with the thoraco-pelvic unit, provide trunk stability while seated. It should be noted that when the offset knee joints are fully extended the force of the shock cords is absorbed by the uprights and does not increase the vertical load on the lower limbs when standing.

SUMMARY

A preliminary report is presented on the potential benefits of elastic materials as a source of external power for orthotic use in general. In particular, a description of the experiences, to date, of an ongoing effort to devise a practical orthotic system of ambulation for paraplegics demonstrates the application of dynamic forces for two specific activities that were given first priority; getting up from and sitting down into a wheelchair. An orthotic system has been devised for these two activities that consists of four newly-developed components:

*A pair of folding crutches that are permanently attached to the wheelchair;

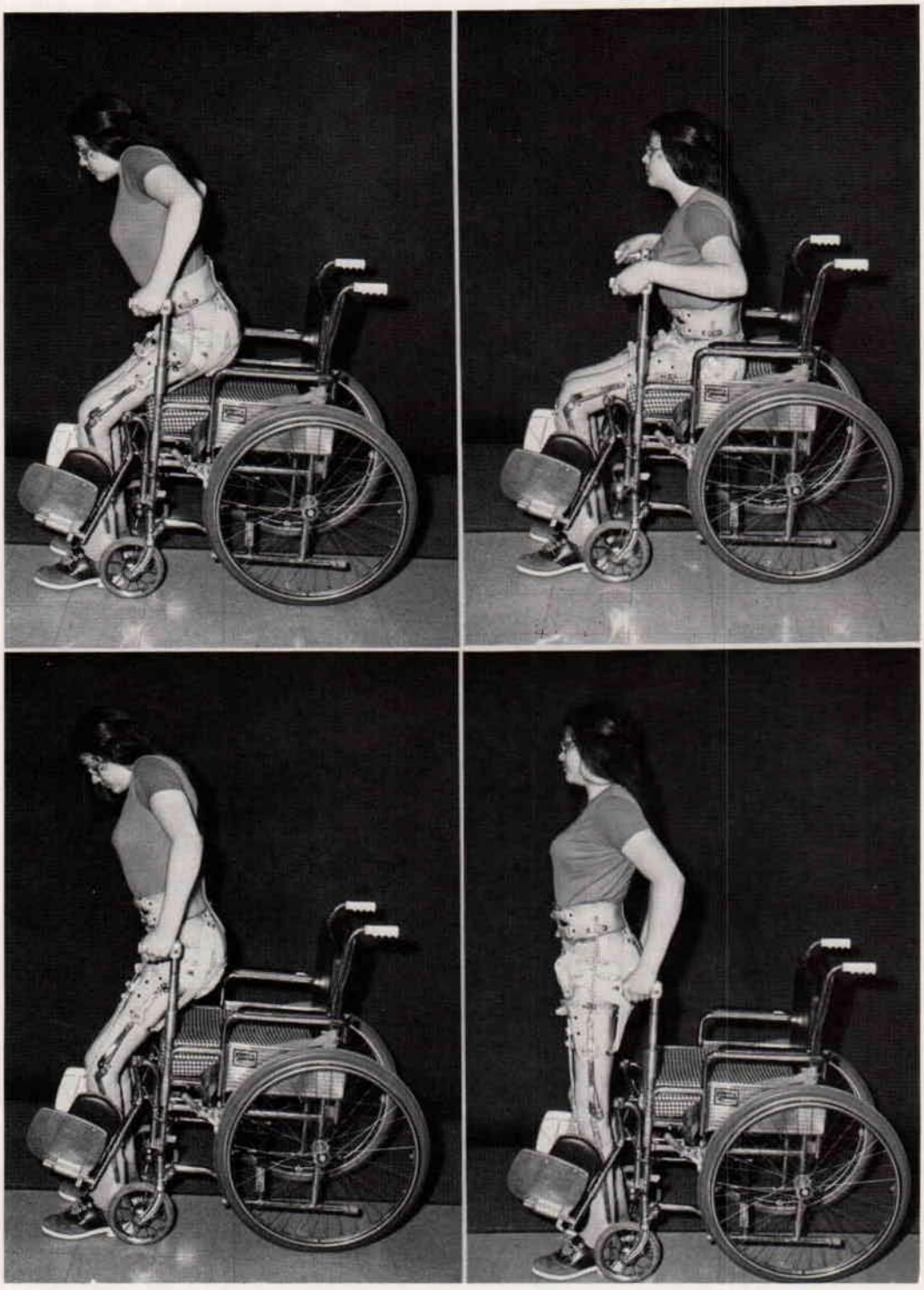

Fig. 7. Patient rising from specially equipped wheelchair.

*A modified use of the polypropylene thoracopelvic unit with dynamic pelvic extension assist that was previously developed for low-level myelomeningocele patients.

*A dynamic knee extension assist unit that automatically extends the lower limbs before the patient lowers himself to the floor, after having achieved an upright position;

*The use of a single lateral, spring-locking, offset knee joint for the patient's KAFO's. The spring-locks are modified so that they can be locked individually in the disengaged position in preparation for sitting down into a wheelchair.

In Figure 7, left, the patient is shown releasing a knee-lock and locking it in the disengaged position in preparation for sitting. The shock cord at lateral midline, with its proximal attachment to the thorax portion of the thoracopelvic unit and its distal attachment to the lateral panel of quadrilateral thigh cuffs, provides mediolateral stability of the trunk without the use of metal bars. In figure 8, right, the patient is shown lowering herself into the specially equipped wheelchair.

The rationale for these developments is described in detail. It should be emphasized that these developments are the results of an *ongoing* project. Their preliminary use is reported at this early stage because it is believed that their apparent potential value can readily be clinically tested and evaluated by others and, if accepted, speed up their refinement and further define applications for general orthotic use.

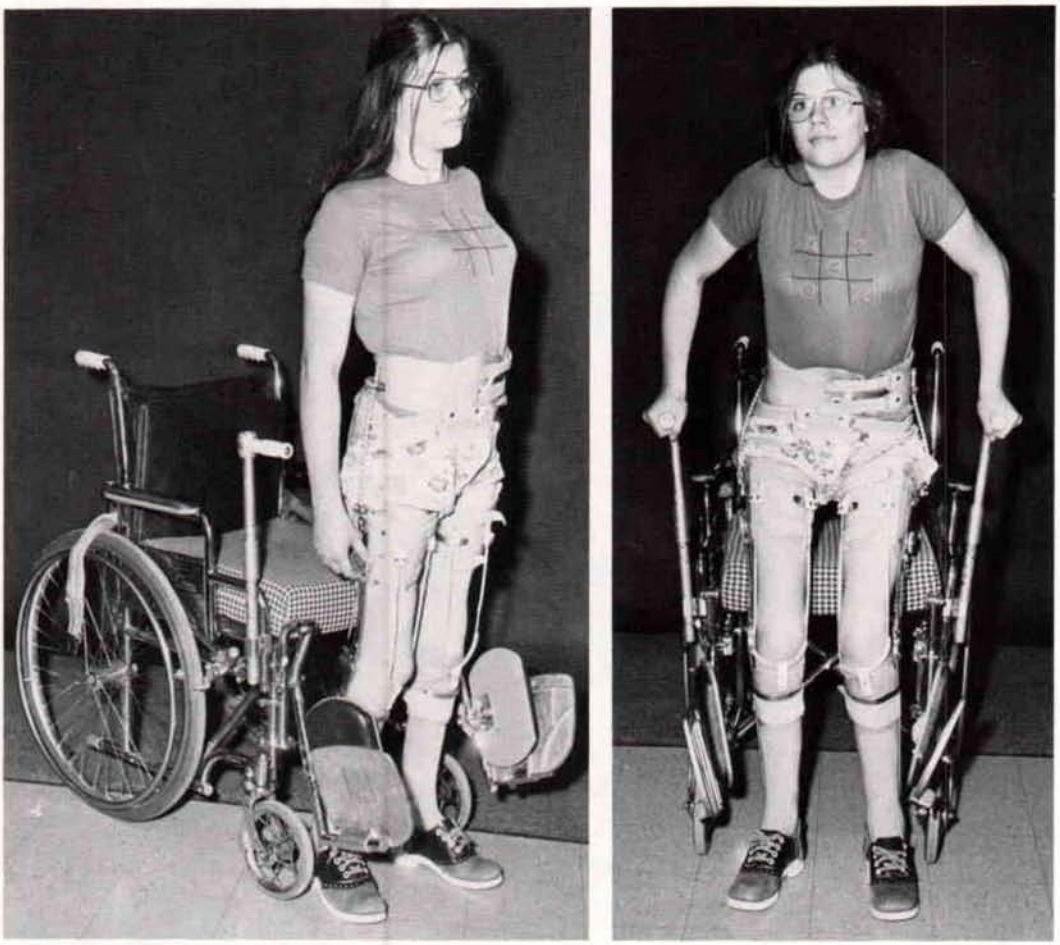

Fig. 8. Patient lowering herself into a specially equipped wheelchair.

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Orthotics and Prosthetics, Vol. 30, No. 1, pp. 23–29, March 1976

A LOW COST VACUUM-FORMING SYSTEM[1]

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Edward A. Bianchi, B.S.[2], and
Richard A. Foulds, M.S.[2]

Vacuum-forming is an excellent method for molding sheets of plastic into complicated shapes. It is just beginning to be used in the field of rehabilitation medicine where the need to make devices that fit the human form is great. This article describes a new, inexpensive apparatus which enables orthotists and prosthetists to use the vacuum-forming process in their work with a very small outlay of capital. Very little training is required to use the apparatus, and it is now being made available in limited quantities.

In the vacuum-forming process a sheet of hot, pliable plastic is drawn either into or around a mold with the use of suction provided by a vacuum pump. When the plastic cools and hardens, it retains the shape caused by the mold. An example of the usefulness of a molded orthosis is shown in Figure 1.

The process, though simple, when adapted to the needs of mass-production, requires very expensive machinery. Until recently only industrial vacuum-forming equipment was available, with prices ranging from $4,000 to over $125,000. Because of the large investment in money and space required to obtain and use the machines designed for mass production, very few medical facilities have made use of the vacuum-forming process.

For mass-production expensive equipment results ultimately in low unit costs. But for the specialized, one-of-a-kind world of rehabilitation, expensive equipment does not necessarily justify itself, and a low-volume, low-cost, non-automated system seems much more suited to the needs of orthotists and prosthetists. Such a machine is feasible when the inherent simplicity of the vacuum-forming process is fully exploited.

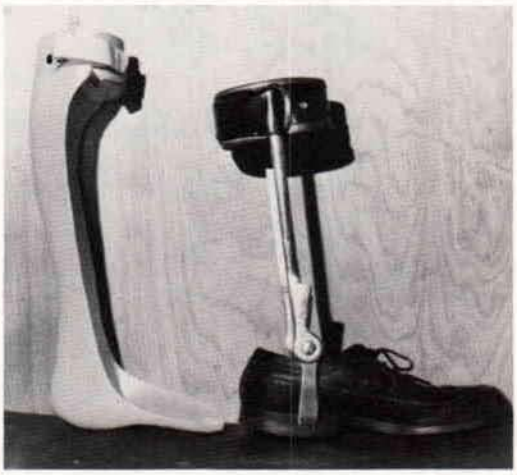

Fig. 1. A molded ankle-foot orthosis (left) is contrasted to the conventional metal and leather orthosis that it replaces. Besides being lighter in weight, the plastic orthosis requires no modification to the shoe. This feature makes it possible for the patient to interchange shoes easily.

With these thoughts in mind we designed and built a vacuum-forming apparatus called the "Bracemaker" (Fig. 2), which is simple, functional, versatile, and, above all, inexpensive. In accordance with the terms of the federal grant which provided part of the funds for this work, we are making the design of the "Bracemaker" available to the medical community.

Why should orthotists and prosthetists be interested in vacuum-forming, when they already have many conventional techniques to use? The advantages of plastics are part of the reason. Plastics are often found in rehabilitation devices where light weight, cosmesis, flexibility, durability, and close fit are needed. Until recently, it has been necessary to form plastics by hand, a process that requires considerable skill, and often repeated attempts, to achieve good results on any but the simplest shapes. With vacuum forming very little skill is needed, results can be quite

[1]This work was supported in part by Rehabilitation Services Administration Grant #16-P-56800/1-10 (RT-7).

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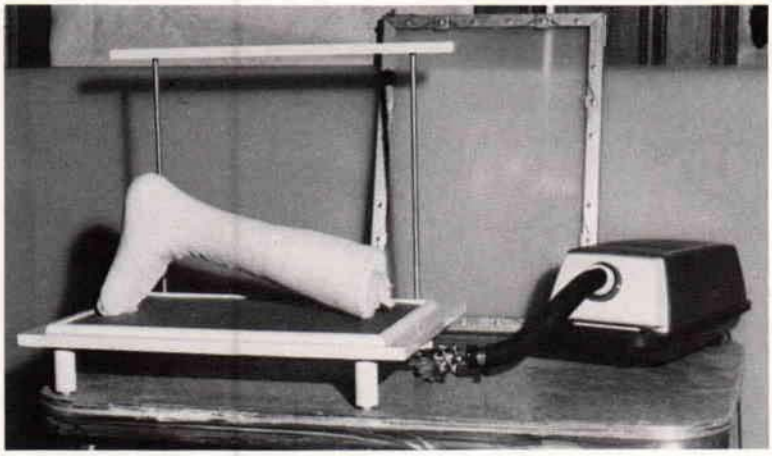

Fig. 2. The "Bracemaker." The vacuum cleaner is used for the pump to evacuate air through the table top from the space between the plastic sheet and the table. Note the two upright guides at the rear edge of the table. The rack, or frame, for holding the plastic sheet during heating and forming is shown in the rear.

consistent, and parts can be formed which cannot be done at all by hand. Moreover, the technique can be used with a large variety of plastics—almost all of the thermoplastics. This eliminates the need for special, low-temperature forming plastics, and opens up a wide range of new plastics that are extremely useful in the provision of rehabilitation devices.

THE PROCESS

The type of vacuum-forming most useful in prosthetics and orthotics can be broken down into six distinct steps:

1. *Mold Preparation*—A mold in the shape of the part to be formed must be prepared. In prosthetics and orthotics plaster-of-Paris is usually

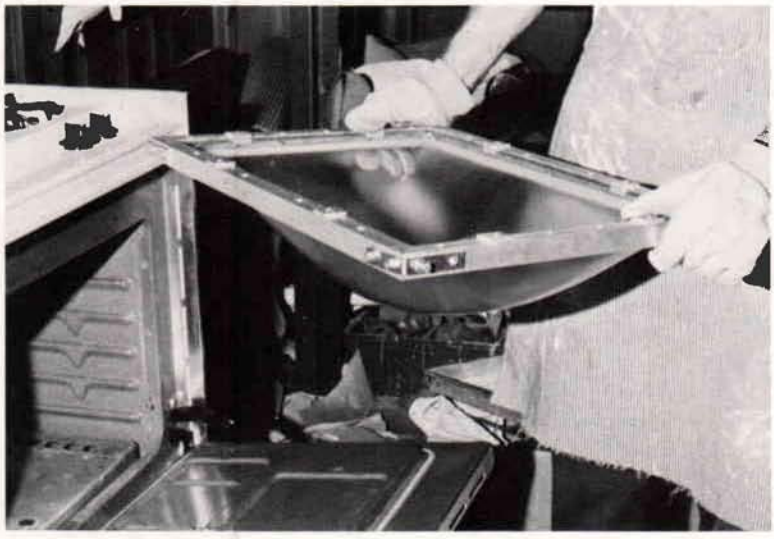

Fig. 3. The molten plastic sheet being removed from the oven prior to placement over the male model. Note the special frame, or rack, for holding the plastic sheet around the periphery.

the material of choice. The mold is placed on a special table (Fig. 2), and supported for the best wrapping action of the plastic.

2. *Heating*—The plastic sheet must be brought to a temperature that makes it soft and pliable, yet leaves it strong enough to resist tearing (Fig. 3).

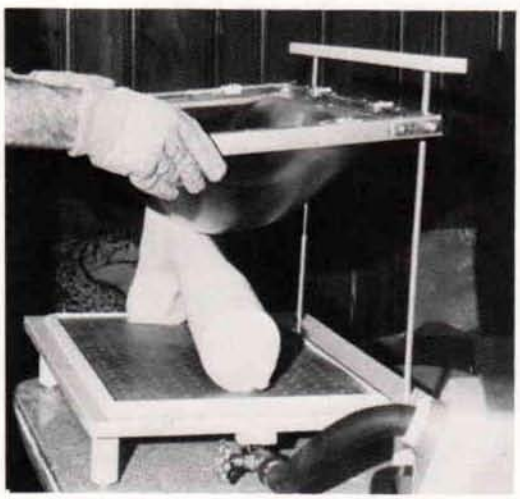

Fig. 4. The plastic sheet is draped over the mold. The uprights are used to assist the operator in keeping the molten plastic sheet aligned with the table.

3. *Draping*—The plastic sheet must be draped over the mold and then sealed at its edges, so that the mold is totally enclosed (Figs. 4 and 5).

4. *Drawing*—The air underneath the plastic is pumped out, allowing the pressure of the atmosphere to wrap the soft plastic around the mold, taking its shape (Fig. 6).

5. *Cooling*—The plastic is allowed to cool while the vacuum is maintained. The plastic hardens and regains its normal properties.

6. *Extraction*—The finished part (Fig. 1, left) is cut out of the plastic. (Unfortunately, the excess plastic is waste which can rarely be reclaimed, even though it is inexpensive relative to other materials). The mold may also be removed at this time. Usually it can be removed intact, but ocassionally it must be broken and removed in pieces.

THE BRACEMAKER

The basic elements of the "Bracemaker" vacuum-forming system (Fig. 2) are:

1. an oven, for heating the plastic
2. a rack, for handling the plastic
3. a vacuum table, to support the mold and seal the plastic against air leakage during drawing
4. a vacuum pump, for withdrawing the air

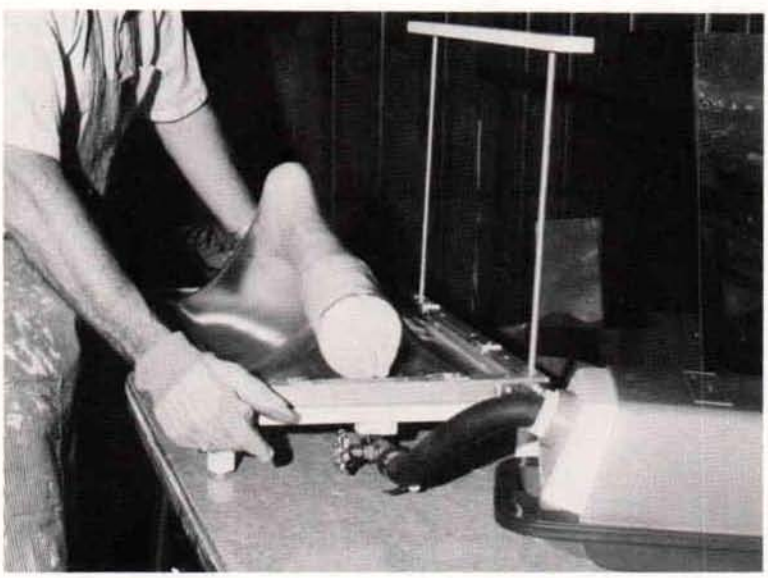

Fig. 5. The plastic sheet and frame are brought in contact with the vacuum table so as to seal the enclosed area with respect to the atmosphere.

Fig. 6. The air underneath the plastic is pumped out and the plastic follows the form made by the model and table. The form is retained faithfully by most sheet plastics upon cooling.

The oven is an ordinary household oven, either gas or electric. Because they are mass-produced, these ovens are far less expensive than the acceptable alternatives. Yet, they are just as adequate as laboratory and industrial ovens are for this purpose. One can often be bought new for $200.00 or less.

The rack (Fig. 7) has been designed especially for the vacuum forming process, and it holds the plastic sheet firmly about its perimeter even when the plastic is in the molten stage. Its special clamping devices, or "traps" (Figs. 8 and 9), are designed to permit easy insertion and removal of the plastic, and to obviate the need for special

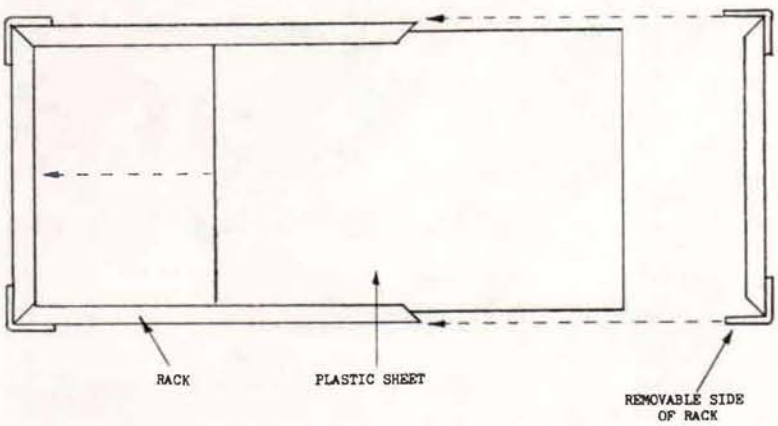

Fig. 7. The rack for holding the plastic sheet during heating and forming. See Figures 8, 9, and 10 for further details.

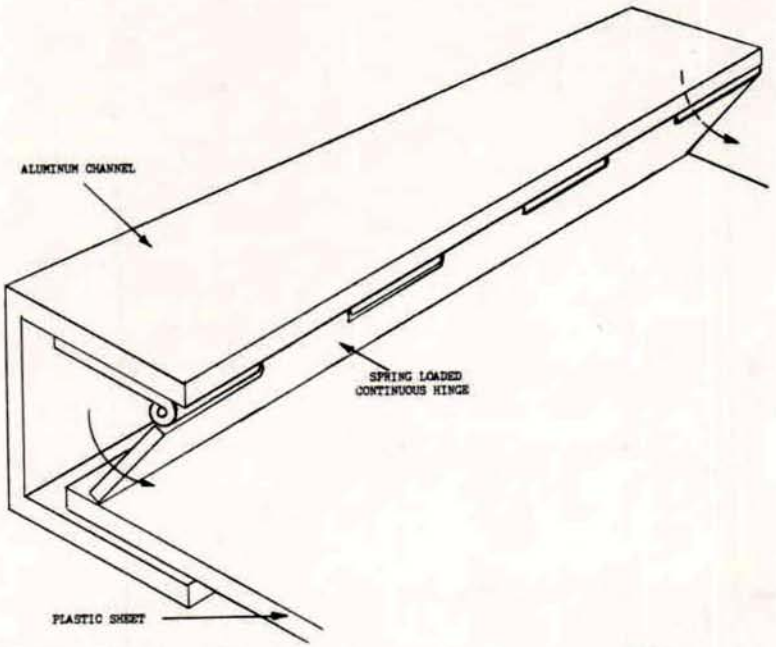

Fig. 8. Details of the trap in the frame to hold the plastic sheet during heating and forming.

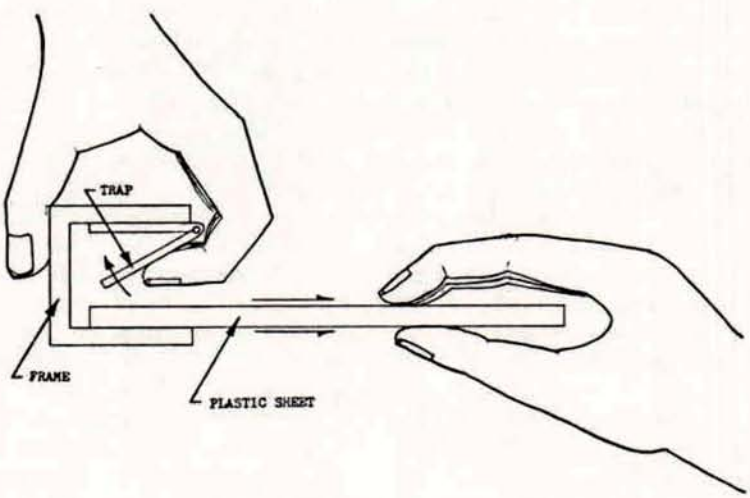

Fig. 9. The plastic can be removed from the rack only when the hinge-traps are released.

preparation of the plastic, such as drilling holes for bolts or pins. The spring-loaded continuous hinge digs into the plastic when tension is applied to pull it out. The traps must be released by hand, as shown in Figure 8, in order to free the plastic sheet. One end of the rack is removable to permit insertion of the sheet. To heat the plastic, the rack can be placed in an oven instead of one of the regular wire shelves. This arrangement makes it possible for the plastic to be placed in the oven, removed, and carried to the vacuum table without the need for handling the plastic itself. Because

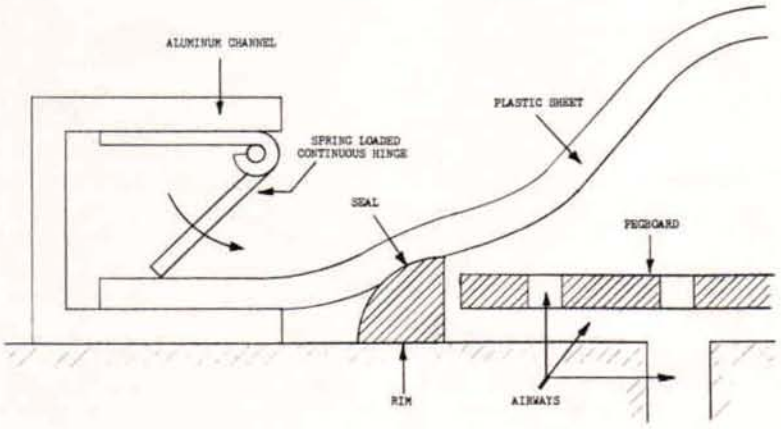

Fig. 10. Cross-section view of plastic sheet, rack, and vacuum table during a forming operation.

the plastic sags several inches when heating, and becomes sticky and pliable, this handling ability is quite necessary for success.

The vacuum table, also a special piece of equipment, is the basic work-surface of the "Bracemaker." It is a low table, designed to sit atop a workbench. It incorporates a perforated surface on which the mold is placed and through which the air is withdrawn, a rim or vacuum seal which provides a seal against leakage (Fig. 9), a valve to control the rate at which the plastic is drawn around the mold, and a set of guides to aid in draping the plastic over the mold.

The vacuum pump is nothing more than a simple household vacuum cleaner with fittings to attach the suction hose to the vacuum table. Surprisingly, it is more than adequate for the job, and in fact is far better than vacuum pumps costing four to twenty times as much because a vacuum cleaner pump has a far greater tolerance for leakage than most other pumps, while still being able to draw the vacuum needed.

The peripheral equipment needed varies with the raw material used and the end product desired. Tools for cutting out sheets of raw plastic, smaller cutting tools for extraction work, cast cutters, finishing tools, are all needed to some extent in vacuum-forming. But there are many different tools which can do these jobs, and most shops have a number of suitable tools already. For this reason such items are not supplied with the "Bracemaker," but recommendations as to what is needed and what can be used are available.

OPERATION

A good example of the use of vacuum forming in orthotics is the procedure used to make an ankle-foot orthosis (AFO). Figure 1 shows a conventional steel-and-leather orthosis on the right and a vacuum-formed plastic orthosis on the left. This type of plastic orthosis is, to date, the most widely used and highly developed of the vacuum-formed appliances.

The ankle-foot orthosis is usually made out of polypropylene sheet, most often 3/16 in. thick. Polypropylene is often called an "engineering plastic," and is noted for its ability to stand up to constant flexure, its resistance to "creep," its strength, its springiness, and its low cost. These are all useful properties for the AFO. Other useful available plastics are Lexan, ABS, and polyethylene. The cost of plastic for each brace varies with material and supplier, but, at the time of this writing the average cost in the Boston area for polypropylene for one orthosis is $4.00.

To make the mold for the AFO, a plaster cast of the patient's lower leg and foot is taken, using standard techniques. The cast is then cut down the anterior wall with a cast cutter, and removed from the patient. It is then dusted thoroughly inside with talcum powder, sealed again usually with strips of plaster bandage, and filled with plaster of Paris.

When the plaster has cured, the cast is stripped away, to reveal a positive model of the patient's lower leg and foot. This model is finished to re-

move rough spots, and then trimmed at the toe and shin so that it will stand heel upwards on the vacuum table (Fig. 2). The mold is then covered with cotton stockinet, and placed on the vacuum table.

A sheet of plastic is placed in the rack. The rack and plastic sheet are then inserted in the oven, which has been holding at 500°F. The rack should be placed in the top of the oven, with plenty of clear space underneath to permit the plastic to sag as it is heated. Within five minutes the plastic will begin to turn clear and sag. In two or three more minutes it will be clear all the way to the edges of the rack, and it will have sagged six to eight inches in the center. At this point it is ready to be removed from the oven (Fig. 3).

The vacuum pump is turned on, and the vacuum control valve is opened to one quarter-turn. The rack containing the heated plastic is removed from the oven, carried to the vacuum table, lifted high over the mold, and draped smoothly down over it (Fig. 4). The rack is pressed down onto the table (Fig. 5), stretching the plastic over the rim which surrounds the vacuum surface, and forming the vacuum seal (Fig. 9). The plastic will begin to draw down over the mold, and the rate of draw can be adjusted by further opening or closing the vacuum valve. Once the plastic is properly drawn down, the valve should be adjusted so that it does not draw further, but still remains tight on the mold (Fig. 6). The vacuum is left on until the plastic hardens—in two or three minutes.

Once the plastic has cooled thoroughly, it can be removed from the rack. The orthosis can then be rough-cut out of the plastic, and the mold removed. (The mold can normally be re-used, should that be necessary.) The orthosis is then finished to the proper shape, but left a little oversize at the ankle. The ankle area is where the orthosis flexes the most, and the size of this section is critical to the corrective force the orthosis provides to the patient. It is necessary to leave the ankle stiffer than is thought to be necessary, so that material can be trimmed away during patient trials until the correct amount of stiffness has been reached.

A strap with a Velcro closure is riveted at the top of the orthosis, so it can be fastened to the patient's leg (see Figure 1). The lower part of the orthosis inserts into the patient's shoe as if it were part of his foot. No other attachments are necessary.

Should the patient find the orthosis to be irritating at some point, or some other flaw is discovered, minor changes in the orthosis can be made. Small areas of the plastic can be re-heated with an electric heat gun or a propane torch, and bent by hand to a new configuration. However, major modifications by this method are rarely successful, and usually a new molding is required.

The final finishing of the orthosis should include fire-polishing. This process extends the life of the orthosis by removing microscopic cracks left on its edges by previous finishing operations. These cracks are where the stresses in the plastic become highest, and where breakage is most likely to start. Playing the blast (but not the actual flame) from a propane torch along the edges of the brace will cause a small amount of plastic to melt there. When the melted plastic cools it forms a smooth bead, which should be devoid of cracks.

PROJECT STATUS

In May of 1974 we began the distribution of a small number of pre-production "Bracemakers" to hospitals and brace shops in the New England region and in New York. The object was to test the machine in the field and to gain a base of experience outside of our own facilities. As of this writing we have had machines in the field and in operation for over a year, with encouraging results. We estimate over three hundred orthoses have been made on our machines. Most of these have been AFO's, but a number have been experimental orthoses of other varieties. The private orthotics facilities using our machine have shown a definite ability to custom-manufacture plastic orthoses.

Our future plans for the "Bracemaker" project aim at the development of a self-perpetuating technology. This includes the development of a production model of the "Bracemaker," the distribution of machines to facilities throughout the country, and the establishment of a clearinghouse for the assembly and distribution of information on vacuum-forming. In addition, we intend to continue our research into new uses for vacuum-formed parts in rehabilitation medicine.

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KINEMATIC ANALYSIS OF THE HUMAN ELBOW

C. W. Beadle[1] and M. E. O'Brien[1]

This note describes a simple method for making a kinematic analysis of the elbow for the purpose of confirming how closely the elbow can be considered a pure hinge joint and for determining the location of this hinge point in order to properly design and fit a lower arm orthotic device. An anatomical description of elbow motion has been given by Taylor (*2*) and many others. The radius and ulna are generally considered to rotate about the elbow axis in the manner of a simple hinge.

Assuming planar motion, the relative motions of two rigid bodies such as the forearm bones can be described by two curves called the fixed and moving centrodes. The moving centrode rolls without slip on the fixed centrode as shown in Figure 1. There is a unique pair of centrodes for any given relative motion. While there are other methods for describing relative planar motion between two rigid bodies such as a generating curve and an envelope, the centrode method proves to be the easiest to obtain experimentally and a simple apparatus was designed which measures and records the motion of the elbow joint. From this data, the centrodes can be determined. The method is similar to that described by Freudenstein and Woo (*1*) for the analysis of knee motion.

One of the two members (rigid bodies) is stationary while the other is allowed to move with its natural motion. At least two marking pens attached to the moving member at two different positions such as A and B in Figure 1 trace out different paths (A – A′ and B – B′) while the member moves. There must be a means of identifying points on the two curves which were traced at the same time. A point on the fixed centrode is determined by erecting a normal (i.e., a perpendicular to the tangent) to each curve at any such pair of points. The intersection of these two normals determines one point on the fixed centrode. This process is repeated for many pairs of points on the two curves traced out by the moving member. The curve fitted through the resulting series of intersection points becomes the centrode.

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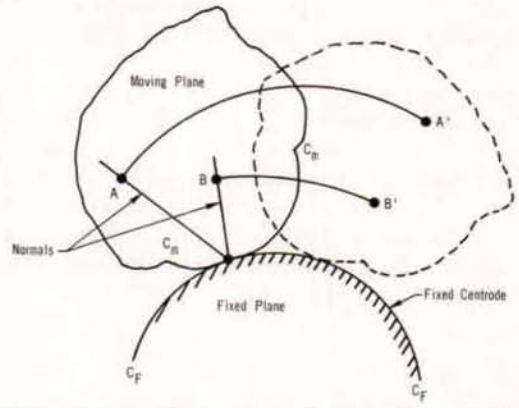

Fig. 1. Fixed and Moving Centrodes

EXPERIMENTAL DETERMINATION OF THE CENTRODE

The apparatus is shown in Figure 2. The upper arm is held rigidly against a horizontal board while the lower arm sweeps from full flexion to full extension along side a vertical board. A pen

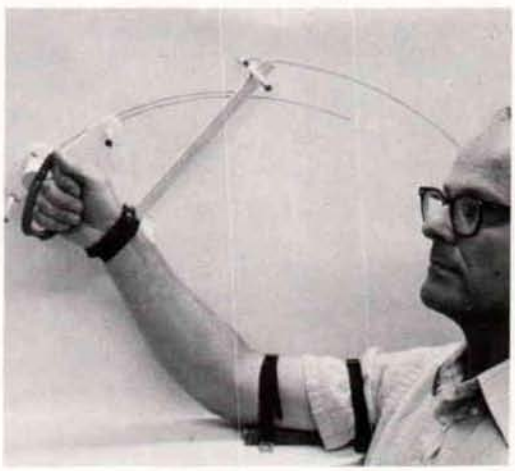

Fig. 2. Tracing Apparatus

holder strapped to the lower arm records three point paths on the board. The wrist joint is wrapped to prevent motion. The elbow is placed against a dowel for initial positioning with the arm in full flexion, and the upper arm is strapped in place. When the dowel is removed, the lower arm can be placed in full extension. While the three point paths are being traced, the board is vibrated lightly so that reference marks are left spaced along the three paths and it is then possible to know the simultaneous positions of the three pens at several positions along the paths. The intersection of the normals determines a point on the centrode of the upper arm (Fig. 3). While only two normals are necessary, the third serves as a check on accuracy. Eighty percent of the intersections of all the normals were found to be in a circle of diameter 7.95 mm (.313 inch). The centrode of the lower (moving) arm could be determined by the inverse process but this was not carried out for reasons described below.

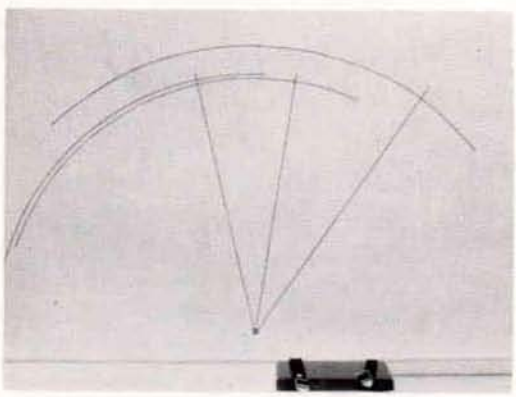

Fig. 3. Determination of a Point on the Fixed Centrode

DISCUSSION AND RESULTS

Since the fixed centrode lies within a 7.95 mm (.313 inch) diameter circle, it is clear that regardless of the shape of the moving centrode, the elbow joint can be closely approximated by a pure hinge. Indeed some of the deviation from pure hinge motion may well be caused by cartilage compression, muscle action, stress on the elbow due to load, etc. A check on the accuracy of this approximation was made by retracing the three point paths with a compass placed at the center of the 7.95 mm circle. The greatest deviation from the original paths was .76 mm (.030 inch).

Since the elbow may be reinserted into the measuring device after the data has been reduced and the pivot point found, the axis of the elbow hinge can be clearly marked on the test subject. If the patient is to be fitted with an orthotic device, he should of course be wearing the cuff or sleeve on the upper arm that will be used in the device during both the determination of the centrode and during the determination of the hinge point.

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Orthotics and Prosthetics, Vol. 30, No. 1, pp. 33–39, March 1976

THE ORTHOTIC CARE OF THE DENERVATED FOOT IN HANSEN'S DISEASE

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Paul W. Brand, C.B.E., F.R.C.S.[2],
Joseph K. Reed, Jr.[3], and
David Welch[4]

Problems of the denervated foot in Hansen's disease (Leprosy) are due to both sensory loss and motor impairment with deformity resulting from peripheral nerve paralysis. Neither feature need be harmful providing prophylactic measures of orthotic care are instituted. It is still another factor, namely trauma or infection, or both, imposed upon insensitive and deformed areas that is responsible for destruction of the foot. The control of these problems therefore is based upon avoiding extrinisc trauma and particularly upon elimination of intrinsic trauma resulting from the repetitive stresses of walking and standing in unsuitable footwear. The latter feature is the basic principle of the orthotic care that is indicated (*1*).

The total management of the denervated foot of Hansen's Disease is carried out in three phases. All wounds, particularly those affecting the plantar surface, must be healed. This is accomplished by eliminating weight-bearing through bed rest or by applying a walking cast. When healing has been obtained, gross deformities are corrected surgically with restoration of the form and action of the foot to as nearly normal as possible. In the last phase, because of persisting residual anatomical and functional impairments, orthotic care is provided by altering the footwear to compensate for these changes (*5*).

When patients possess gross deformity which cannot be corrected because surgery is contraindicated, these deformities are more difficult to manage because modified footwear may not relieve completely the high pressure points on the plantar surface. Such patients may remain confined to a wheelchair a greater part of the time; however, some degree of benefit is obtained by bearing weight intermittently for short distances during which time trauma is of short duration and relatively less intense, with recovery more likely to occur within the periods of rest obtained with use of the wheelchair.

The U.S. Public Health Service Hospital at Carville, Louisiana, is devoted to the care of patients with Hansen's Disease. The foot care program here includes a preliminary study of clinical, radiographic, and footprint examinations. The findings are correlated to determine if the problems are static or dynamic or both, in origin, if there are indications for surgical correction of pathologic deficits, and to determine the appropriate prophylactic measures to preserve the foot by preventing the recurrence of complications.

Manifestations of paralysis of the peripheral nerves vary from minimal isolated changes of sensory loss to patterns conforming to the distribution of the affected peripheral nerves, or the more extensive pattern of "stocking" distribution, and the deformities of claw toes and drop foot. These changes may exist alone or be associated with secondary changes of absorption and contractures. Also, even after surgery, varying degrees of deformity that require orthotic care may exist.

In Hansen's Disease, the foot may be classified arbitrarily into four categories: (The same classification applies to the denervated foot from peripheral nerve paralysis from any cause.)

Category I – the foot is grossly normal but possesses loss of plantar sensation.

Category II – the grossly normal foot possesses loss of plantar sensation and plantar scarring, commonly affecting the ball of the foot.

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Category III - the foot is deformed with loss of plantar sensation and plantar scarring; however, the length and width of the foot is not affected appreciably.

Category IV - the pathologic short and/or narrowed foot is due to metatarsal phalangeal or lateral ray absorption or to amputation.

TYPES OF FOOTWEAR FOR THE DENERVATED FOOT

Four general types of footwear have been developed for the denervated foot at this hospital. These include the sandal, a "healing shoe", the modification of a regular shoe including the extra-depth shoe, and a custom fabricated shoe.

SANDALS

Sandals are prescribed to be worn immediately following the healing of plantar ulcers or post-operative wounds while the patient is waiting for shoes being fabricated or for the delivery of purchased shoes. Some patients like these sandals and choose to wear them most of the time even when definitive shoes have been provided. Two types of sandals have been made; a simple "quickie" and a modified "clog".

The "quickie" sandal is simple in construction, and is made available to the patient in a matter of 20 to 30 minutes. It consists of a soft, flat insole of expanded polyethylene foam backed with a 6 mm-thick microcellular rubber, which is attached directly to an outer sole of neoprene crepe rubber (Fig. 1-A). The patient stands on the heated foam before it is attached to allow the foam to conform to the contour of the foot.

The "clog" sandal (Fig. 2) requires a twenty-four hour period to make. It has a thick base consisting of a polyethylene insole fully molded in its entire extent, and is supported by a neoprene crepe sole with a mixture of wood flour and latex placed between the soles to maintain the molding of the insole (Fig. 1-B). The "clog" sandal possesses significant practical advantages over the simple "quickie" type, and is therefore preferred whenever possible. The sandals are held to the foot with straps made of cotton webbing and Velcro tape.

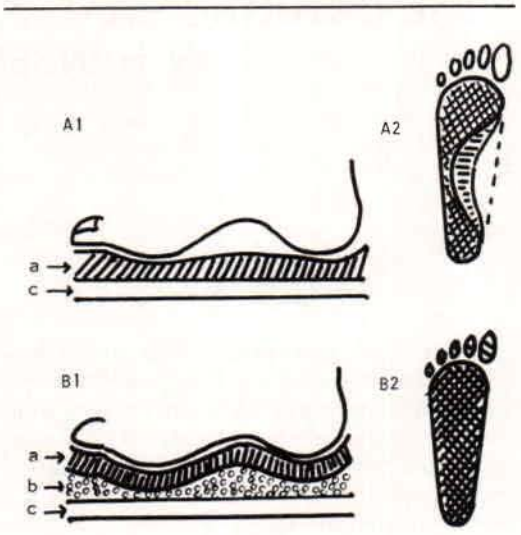

Fig. 1A. The soft non-molded insole (a) which does not distribute the weight bearing pressures evenly and completely to the plantar surface (A2).

Fig. 1B. A molded insole (a) fitting against the plantar surface to distribute the weight bearing pressures evenly to the plantar surface.

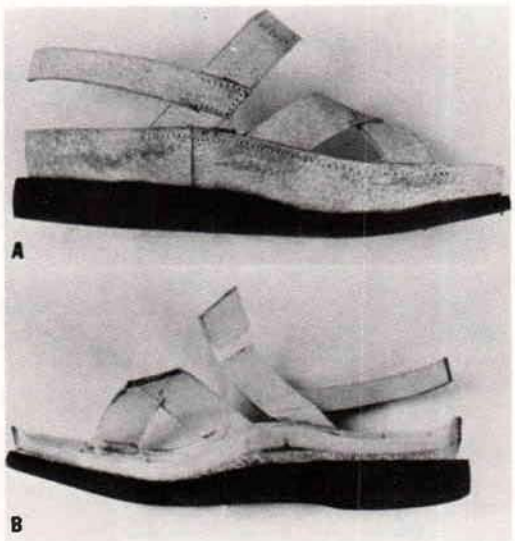

Fig. 2A. Lateral and, B, sagittal views of the molded clog sandal.

THE HEALING SHOE

The "healing shoe" was developed as a compromise when the patient refuses absolute bed rest or will not tolerate a plaster cast. It is made over a recently made plaster model of the foot and ankle. A molded insole of polyethylene is then made and fastened to the model by tape. The entire model and insole are then covered with polyethylene, from the heel forward to include the quarter and vamp of the shoe. It is made in a manner so that the insole can be removed for adjustment or replacement. This is reinforced by lamination with polyester resin, nylon stockinette, and a layer of fiberglass upon which a crepe rubber sole is applied (*3*). The shoe is taped to the foot so that it can not be removed except by a staff member. Since the introduction of Lite Cast II[5], it is now possible to use this material instead of the polyester resin, with a considerable savings of time.

THE MODIFIED REGULAR SHOE

A third type of footwear is the modified regular shoe. The shoe is available for purchase with regular or extra depth. Alterations are made mainly of the insole and outer sole, for the purpose of relieving high pressure points by distributing weight-bearing forces evenly to the plantar surface of the foot while standing and during propulsion. The extra depth shoe is designed so that the insole, which may vary from 6 to 12 mm. in thickness, can be removed and a custom-made insole of polyethylene can be substituted.

Soft insoles are inexpensive but require frequent replacement. They may be molded or not. To be effective, the non-molded insole must be of sufficient thickness so that the foot can sink completely into the material yet leave some resilient padding under the prominent parts of the foot. This type of insole is used only in open sandals because the movements of the foot in a shoe with such an insole makes it prone to damage from friction. The effectiveness of a thin, soft, non-molded insole is questionable, and thus it is used only for feet without deformities which do not need relief for high pressure points. Perhaps the best insole of this type is the Spenco insole, a microcellular rubber material covered with nylon to reduce friction and shear.

A molded insole has the advantage of distributing weight-bearing pressures over the entire plantar surface, but it occupies space which necessitates an extra-depth shoe to accommodate it. The need for this type of insole and shoe can be predetermined by using the thin "Harris" foot mat (*4*) or the microcellular slipper sock (*2*) placed within a shoe and tested while walking.

Materials used for the construction of insoles fall into two categories, a closed cell polyethylene foam material and microcellular rubber. The polyethylene foams can be molded in a heated condition directly on the foot or on a model of the foot. Three polyethylene materials have been employed; Plastizote[6], Pelite[7], and Ali Plast[8]. Plastizote is available in 3 mm., 6 mm., and 13 mm. thicknesses. We have employed the 6 mm. sheet as thin material and the 13 mm. as thick material. The 3 mm. thick Plastizote is too thin for therapeutic use. The 6 mm. ("thin") layer is generally used when the insole is supplemented with a supporting material, whereas the 13 mm. ("thick") layer is often used without additional support.

Pelite is available in medium and firm densities, and is available in 3/16-in., 3/8-in., and 1/2-in. thicknesses. The popular thickness used here is the 3/8-in. firm layer of Pelite which is usually combined with Plastizote to give it added support. It is used as an alternative to microcellular rubber because it can be bonded to Plastizote at the time of initial heating and then molded directly over the foot.

Ali Plast has been recommended as a substitute for Plastizote. It is available in two forms. Ali Plast 4E is similar in density to Plastizote. It is supplied in 24 in. x 36 in. sheets of 1/4 in. and 1/2-in. thicknesses. Ali Plast 6A is slightly more

[5]Available from the Orthopedic Co., Inc., Merck, Sharp & Dome, West Point, Pennsylvania 19486.

[6]Patented by Bakelite Xylonite, Ltd. Available from Smith and Nephew, Ltd., Bessemer Road, Hertfordshire, England; or The Knit-Rite, Inc., Paramedical Distributors, 1121 Grand Avenue, Kansas City, Missouri 64106; or the Apex Foot Products Corp., 118 West 22nd Street, New York, New York 10011.

[7]Available from Fillauer Orthopedics, 936 E. Third Street, Chattanooga, Tennessee 37401; or the Apex Foot Products Corp., 118 West 22nd Street, New York, New York 10011.

[8]Available from Ali Med., 11 Concord Square, Boston, Massachusetts 02118.

dense than Plastizote, and is recommended as a backing for Ali Plast 4E. It is supplied in 21 in. x 36 in. sheets of 1/8 in., 1/4 in., and 1/2 in. thicknesses. Its use here has been limited in deference to Plastizote.

Two kinds of microcellular rubber are employed, plain Neoprene[9] and a nylon reinforced neoprene, the Spenco insole[10]. Plain neoprene is classified as R-425-N, and R-431-N. The R-431-N is slightly denser than R-425-N and both are available in the 3-mm. and 6-mm. thickness. The material is used mainly to support the molded Plastizote insole, provide additional softness, and can be bonded to the underside of the insole by contact cement. The 3-mm. sheet is used because there is minimal distortion and less difficulty attaching it to the edges of the insole. Several layers can be added to increase the softness of the insole. The Spenco insole is microcellular rubber covered with a thin layer of nylon which gives the rubber stability with a capability of stretching in all directions, yet upon release it resumes its original state. The addition of nylon makes this an expensive item but it preserves the microcellular rubber to give it longer wear as well as producing less friction. It is used alone for the grossly normal foot with an insensitive sole, or it may be used to complement the molded Plastizote insole when it is placed over it.

Plastizote has been used almost exclusively as the basic layer for soft molded insoles. It is molded by heating to 140°C for three minutes after which it is applied directly to the foot or to a plaster model of the foot. Other materials, Pelite and microcellular rubber, have been used to complement the molded Plastizote. A combination is used when weight-bearing forces are excessive and, although tests may suggest an even distribution of the plantar pressures, real-life activity demonstrates that the Plastizote alone does not provide adequate protection. Every model of the foot must be of recent origin, preferably being made immediately before it is used, as a model with a long "shelf life" may not represent the present deformity exactly, and molded insoles prepared from them should not be expected to protect adequately the plantar surface.

[9]Obtainable from Rubatex Corp., Bedford, Virginia 24532.

[10]Obtainable from Spenco Medical Corp., P.O. Box 8113, Waco, Texas 76710.

Though patients with normal plantar sensation may wear conventional shoes possessing leather insoles without problem, they are not recommended for the sensory denervated foot as they are potentially harmful. The addition of a Spenco insole is a simple measure that provides protection for the foot that has no deformity, but lacks sensation.

CUSTOM FABRICATED SHOES

The outer sole of footwear may be hard as reflected in the use of leather or it may possess a resiliency in the form of a neoprene crepe sole. When the outer sole is modified, it is altered to alleviate plantar pressure as the foot "rolls" or "rocks" forward during propulsion. Two modifications are made to provide a forward roll on the foot in gait. The metatarsal bar (Fig. 3) is placed under the metatarsal shafts to receive some of the weight ordinarily transmitted to the metatarsal heads which is now vulnerable to injury on account of plantar scarring and underlying bone deformity invovling the metatarsal heads. The second modification is the 'rocker sole' which is used when the foot is short.

The short foot is most difficult to manage due to shortening of the toe lever which makes the

Fig. 3A. Lateral and, B, sagittal views of a regular shoe with a deep toe box and a metatarsal bar.

distal ends prone to damage during the push-off phase. This foot presents an even greater problem when a concealed drop foot with functioning plantar flexors exists. Grossly, the short foot is classified as having either one-third or two-thirds loss of the forefoot. Where there is loss up to one-third, the foot may be fitted with a regular or extra depth shoe but with specific modifications (Fig. 4). A molded soft insole and a rocker sole

Fig. 4A. Lateral and, B, sagittal views of a "rocker" mechanism applied to a shoe of regular length.

are basic, with a molded filler for the toe box. The part of the shoe beyond the end of the foot must be prevented from taking weight by keeping it turned upward and held by a steel shank (Fig. 5).

The markedly short foot requires a custom fabricated shoe to fit the foot (Fig. 6). It is made on a model of the foot. The soft insole is molded on the entire length of the plantar surface, and an outer rocker sole occupies the entire length of the shoe to receive the body weight with an even and wide distribution of its pressures to the plantar surface (Fig. 2). It is also necessary to turn the forepart of the shoe upward thirty degrees or more so that the foot "rocks" forward. The pressure is still taken on the proximal part of the shoe, but not distally since the long lever would multiply the thrust on the end of the foot. The upward turning of the forepart of the shoe also serves to help the foot to clear the gound as it is raised to enter the swing phase. The rocker shoe "rocks" forward when there is unilateral involvement. However, with both feet shortened, requiring custom fabricated rocker shoes, the patient's gait becomes modified with flat steps and short strides. The steel shank prevents collapse of the sole complex and provides an even transmission of body weight to the foot through the rocker mechanism.

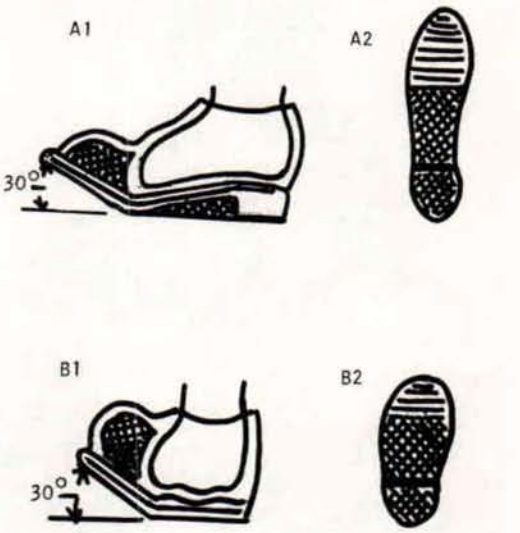

Fig. 5A-1&2. A rocker mechanism applied to a regular length shoe fitted to a foot with 1/3 loss of forefoot.

Fig. 5B-1&2. A short shoe fabricated with rocker mechanism to fit an extremely short foot.

CORRELATING THE SHOE TO FOOT DEFICITS

The Category I denervated foot presents the problem of only one deficit, the loss of sensory perception on the plantar surface. This type of foot requires a soft insole within a proper fitting shoe, with the addition of a Spenco insole worn on a prophylactic basis. If the footprints or the

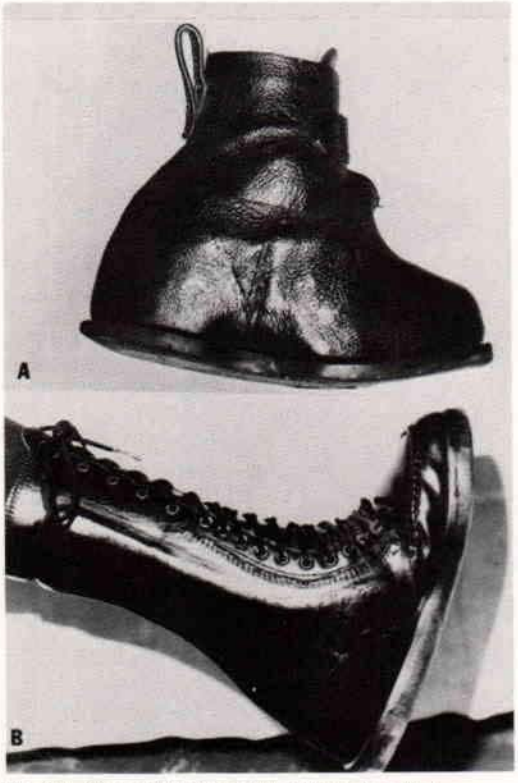

Fig. 6A. Custom fabricated shoe for a short foot.

Fig. 6B. Short boot required to stabilize the ankle to maintain balance.

microcapsule sock demonstrates an uneven distribution of the plantar pressures, a molded Plastizote insole is indicated. These insoles are used instead of the insole of a factory-made extra depth shoe, and because they are removable replacement when indicated because of wear is facilitated.

The Category II denervated foot possesses two deficits, loss of plantar sensation and loss of subcutaneous tissue with scarring. This foot requires a molded insole, which is preferably made of Plastizote. Should Plastizote alone be inadequate to relieve the plantar high pressure points, the Plastizote insole is complemented by the addition of either Pelite or microcellular rubber. These insoles require an extra depth shoe to accommodate them.

In the Category III denervated foot, there is in addition to loss of plantar sensation and scarring, the presence of deformity. Gross deformities such as claw toes and drop foot are remediable by surgery unless contraindicated for general health reasons. Absorptive bone changes, particularly at the metatarsal phalangeal joint level, which may result in abnormal bony prominences on the plantar surface can usually be managed with orthotic measures. The basic provision is the molded Plastizote insole alone or supplemented by a layer of microcellular rubber. In the case of rigid claw toes, an extra depth shoe with a deep toe box usually provides adequate space. In advanced cases of metatarsal phalangeal damage with scarring, a metatarsal bar placed proximal to the joint level transfers part of the body weight to the metatarsal shafts to alleviate the metatarsal heads from compression and shear forces (Fig. 5A-1&2). When ulceration recurs in spite of adapted footwear, surgery may be advised to remove a bony prominence or to move the whole line of metatarsal thrust to a more proximal and less scarred level by metatarsal head excision. This will be of permanent benefit only if the patient continues to wear specially modified footwear.

In Category IV, the denervated foot is short, because of either a complete transverse metatarsal phalangeal absorption or amputation. A regular shoe, usually requiring extra depth, can be modified to fit a foot with loss of less than one-third of the forefoot. This will result in only little change in the gait pattern. However, an extremely short foot is best managed with a shoe fabricated to fit the foot exactly. Here the custom fabricated shoe with a molded Plastizote insole and a "rocker" mechanism is employed (Fig. 5B-1&2).

METHODS FOR ASSESSING PLANTAR PRESSURES

The "Harris" footprint mat is used for assessment in diagnosis and treatment. It is a simple practical method that is inexpensive and provides a permanent record. The thick mat is used with bare feet for routine study, whereas the thin mat is used within the shoe (*4*). Its main limitation is that it does not record pressures on the sides and dorsum of the foot. When measurement of pressures are desired of the lateral and/or dorsal aspects of the foot, these can be determined by use of the microcapsule sock (*2*).

SUMMARY

The use of modified types of footwear have been effective in preserving the denervated foot of Hansen's Disease. Their prophylactic value is especially significant when they are employed for the insensitive and deformed foot. Of equal importance is use of them following corrective surgery to obviate further damage to the foot.

Corrective orthotic measures are never employed for the purpose of overcoming deformity because abnormal forces may be created that are potentially dangerous to the denervated foot. The objective of orthotic care for the denervated foot is to provide a soft molded insole to fill spaces that are void, thereby increasing the surface area receiving pressure yet provide relief of points of high pressure. Once this is accomplished, it can be maintained through recheck examinations at regular intervals when worn parts of the footwear are replaced. This type of patient is never discharged, but instead remains under the constant surveillance of the physician and orthotist.

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Orthotics and Prosthetics, Vol. 30, No. 1, pp. 41–42, March 1976

THE MODIFIED OR MINIHALO

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Louis A. Hazouri, M.D.[2]

A 14-year-old, white female was hospitalized from an automobile accident which resulted in a fracture-dislocation of C-4 and C-5 vertebrae. The patient was on her way from church, riding in the rear of an automobile, when it was struck from behind. She was seen in Langdale, Alabama, where she was evaluated by a physician who referred her after it was certain she had a fracture-dislocation of C-4 and C-5 vertebrae. The patient had complained of neck pain as well as tingling in the upper limbs.

She was placed in complete bed rest and was fitted with a hard cervical collar, before being transferred to the Columbus Medical Center where examination revealed that all limbs had some degree of weakness with poor coordination of the upper limbs, even though she was able to open a package of chewing gum. The patient was cooperative, alert, well-developed, well-nourished, oriented as to time and place, but complained of headaches as well as generalized neck pain. Her heart and lungs were clear. Her blood pressure was 110/70. Her pulse was regular. There was no evidence of any damage to the lungs. There was evidence of contusion about the right scalp. The cranial nerves revealed no evidence of any intracranial pressure. The pupils were round, regular, equal, and reacted to light and accommodation.

On 5/14/75 Gardner tongs were applied.

On 5/14/75, X-rays revealed a vertical fracture line through the posterior aspect of the C-4 vertebral body. Good head position and alignment were evident. Follow-up at 6:30 p.m. showed similar findings.

On 5/15/75 X-rays revealed slight subluxation. The equipment was readjusted and rechecked two hours later, revealing satisfactory abduction and good alignment.

At 7:30 a.m., 5/16/75, X-rays again showed fracture of the vertebral body. However, at this time there was some anterior slipping of the C-4 vertebra with respect to C-5, amounting to five or six millimeters, which indicated fracture instability in the arch. A repeat lateral view taken with better extension shows good position and alignment as noted in the intial two studies.

On 5/21/75 the patient was taken to surgery. Disc excision and fusion was carried out on the C-4 and C-5 vertebrae. Subsequent to disc excision and fusion, the patient was maintained in traction.

On 6/9/75 consideration was given to providing the patient with an orthosis. Because we had had quite a bit of trouble with the four-bar cervical orthosis, its use was ruled out if we could find something better. A halo was discussed, but it, too, was ruled out.

It was decided that we would make a mini or modified halo, with the use of the Gardner tongs.

A molded Plastizote collar was securely and firmly fitted to the patient. Plaster-of-Paris bandages were wrapped around the collar, not only to secure it, but also to provide a firm base for application of the modified halo. The Gardner tongs were left intact but the weights were removed from the traction. Four turnbuckles were secured to the Gardner tongs: one on each side attached to the anterior aspect of the collar, and one on each side attached to the posterior aspect of the collar. These turnbuckles were placed on the Plastizote collar in the position where the plaster-of-Paris had been wrapped around, and were secured to the collar with the plaster-of-Paris both anteriorly and posteriorly. The patient was placed in slight hyperextension to maintain the position that had been held during traction.

When the plaster-of-Paris had dried, the turnbuckles were opened slightly to reduce the pressure on the patient's chin and also to give some traction. That afternoon the patient was allowed to stand and walk a few steps. The next day she went to physical therapy. The only discomfort the patient complained of was a small abrasion on

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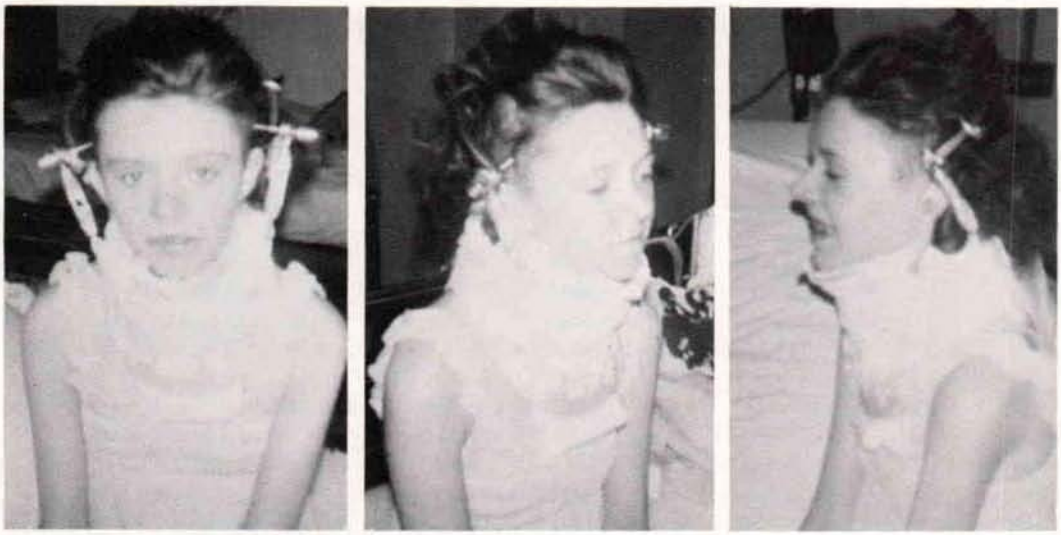

Fig. 1. Three views of the patient with the minihalo.

her chin which was corrected easily by opening the turnbuckles.

On 6/14/75 the patient was dismissed from the hospital. Her response had been excellent and she and her mother had been instructed carefully in reference to the care of the patient and the orthosis.

Contact with the patient on June 25, 1975 revealed that she was having no problems with the orthosis. It was very comfortable, and she was able to go about her normal activities each day. She was very pleased with the results.

We have concluded that use of this modified or minihalo has proved to be most comfortable, effective and simplest to manage of any of the cervical braces that we have ever used. This method does not preclude the use of the "Halo," but can serve as an alternative in most cases.

Orthotics and Prosthetics, Vol. 30, No. 1, pp. 43–47, March 1976

ULTRA-LIGHT PROSTHESES FOR BELOW-KNEE AMPUTEES

A PRELIMINARY REPORT[1]

A. Bennett Wilson, Jr.[2]
Melvin Stills, C.O.[2]

It seems self-evident that for the older lower-limb amputees the lightest limb possible, consistent with safety, is desirable. Recommendations to this effect have been made from time to time but little attention was paid to them. Dr. Joseph Barredo, a physicist retired from the Atomic Energy Commission, and a below-knee amputee owing to trauma, has been emphasizing the importance of weight reduction to research and development laboratories, since 1972 (*1*). Actually a method for producing an extremely light below-knee prosthesis was reported by Wollenstein in 1972 (*3*) (Fig. 1), but the method is expensive in that the labor required is quite high relative to conventional practice.

At first glance, it appears that the so-called pylon, or endoskeletal, prostheses would be lighter than the crustacean type, but closer scrutiny reveals that the crustacean type, when properly engineered and fabricated, can be much lighter and provide the same strength characteristics.

Two factors make the crustacean type superior with respect ot the strength-weight ratio. First, the further away the outer fibers of a structure are from the central axis, the larger is their lever arm, and therefore the greater is their resistance to bending about the central axis. Furthermore the need for a separate foot-ankle unit can be eliminated, resulting not only in weight reduction but in less need for maintenance. Obviously, weight reduction reduces suspension problems.

Advances in vacuum forming of sheet plastics in orthotics and prosthetics show promise of making it practical to provide an extremely light, all plastic, crustacean-type, below-knee leg similar to the Wollstein design. Some development work to achieve this goal has been started at the Rehabilitation Engineering Center at the Krusen Center for Research and Engineering.

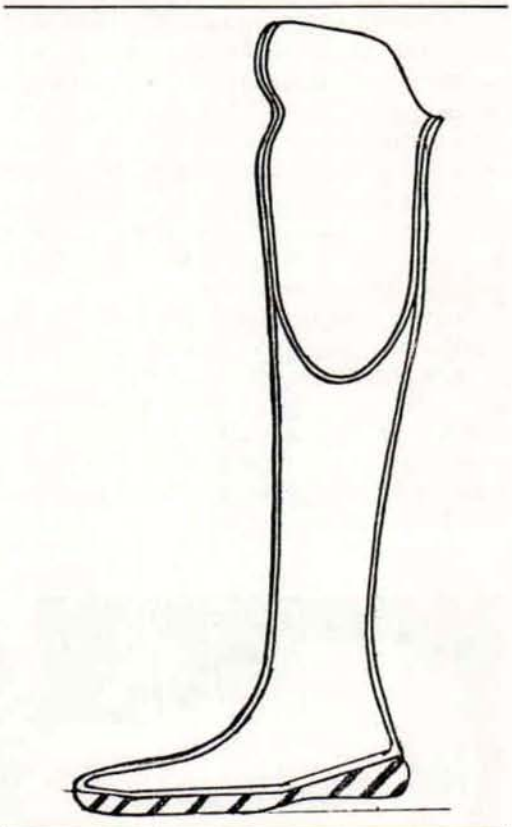

Fig. 1. Cross-section view of the Wollstein below-knee prosthesis. Wollstein used standard laminating techniques to achieve a crustacean shell. The functions of the SACH foot are provided by the cushion heel and sole that are added. The shape of the foot section can, of course, be altered to provide any function that can be provided by the SACH foot concept.

[1]This work was carried out with partial support from the Rehabilitation Services Administration under the terms of Research Grant #23P-55518

[2]Krusen Center for Engineering and Research, Moss Rehabilitation Hospital—Temple University, 12th Street and Tabor Road, Philadelphia, Pennsylvania 19141

Although, it was felt that the older patients with medical problems other than amputation would appreciate the reduction in weight the most, two young healthy male below-knee amputees were chosen as subjects because they were reliable, and could be counted upon to give their reactions clearly and without bias.

The basic approach of Wollstein has been used, but new materials and fabrication techniques have been introduced.

A hollow polypropylene foot-ankle unit (Fig. 2) with removable sole and heel is used for alignment trials with a vacuum-formed polypropylene socket (*2*) and an adjustable shank (Fig. 3). The aligned assembly is placed in a vertical transfer fixture (Fig. 4), the shank is removed, the top of the foot-ankle unit is cut off (Fig. 5), and the space between the sole to the proximal area of the socket is filled with a rigid foam that is foamed in place. The polypropylene foot shell is cut away and the foam is shaped to match the contralateral shank (Fig. 6). The anterior and posterior sections of the foot and shank are molded of polypropylene (Fig. 7). These sections along with the polypropylene socket are welded together and the sole and heel wedge are secured to the foot section to form a strong but extremely light PTB-type prosthesis (Figs. 8 and 9).

The foot-ankle unit for walking trials was also made of polypropylene by vacuum forming mainly because it offers an easy way to fabricate such a unit in the laboratory. Once the shapes for various sizes are standardized, they can probably be mass-produced for sale to individual facilities to be used with an adjustable shank from patient to patient.

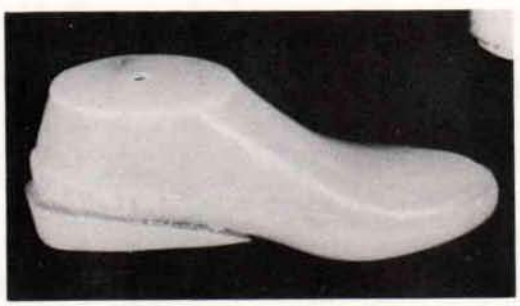

Fig. 2. The hollow polypropylene foot-ankle unit used in the experimental work. Units that will not require fabrication in the individual facility will be practical when the various techniques involved in the total system are refined.

Fig. 3. Walking trials using the vacuum-formed polypropylene socket. The weight of the currently available adjustable legs requires use of the conventional supracondylar strap.

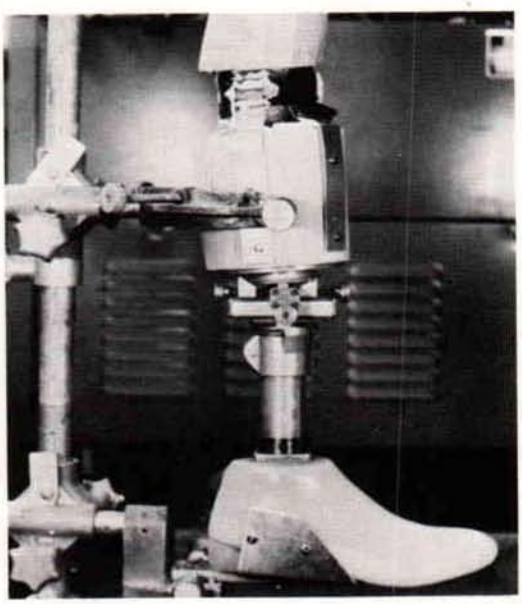

Fig. 4. Trial unit in the vertical transfer fixture.

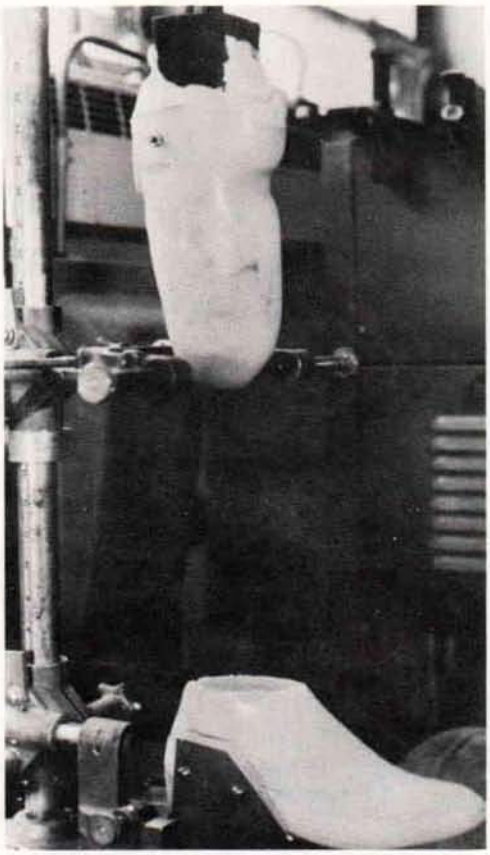

Fig. 5. Polypropylene socket and foot unit in the vertical transfer fixture after the shank has been removed and the top of the "foot" has been cut off. The top of the foot unit is removed to receive the rigid foam that will be foamed in place to provide a positive model for forming the prosthesis.

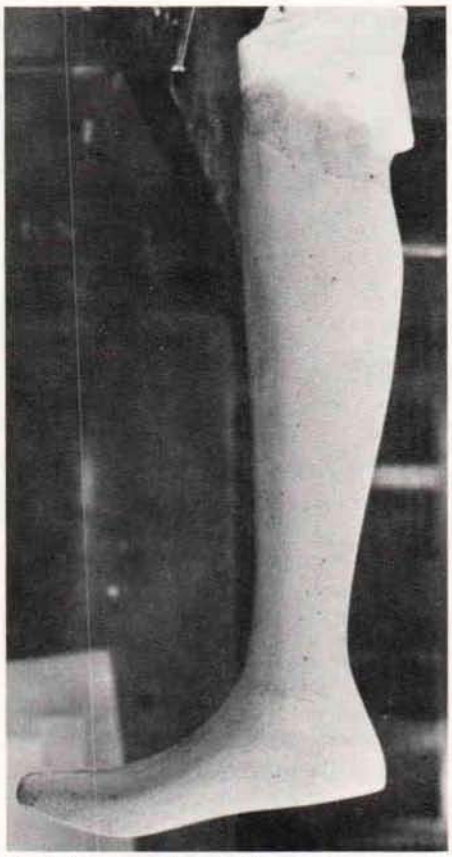

Fig. 6. Positive model ready for the foaming process. Note that the polypropylene shell for the foot has been removed.

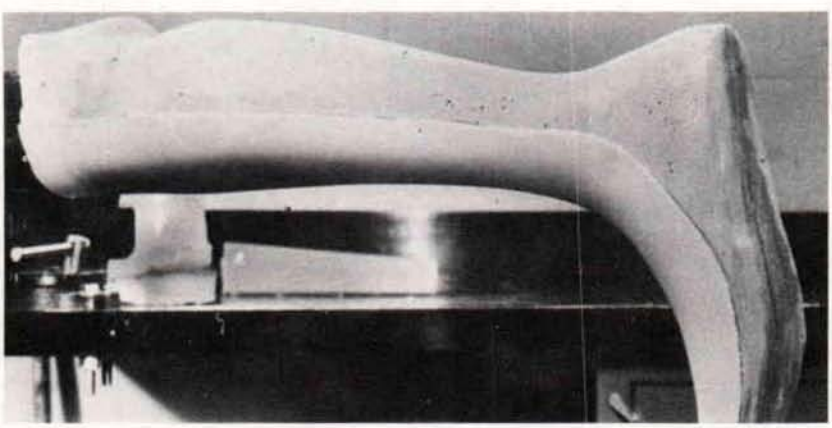

Fig. 7. Positive model with the anterior portion of the polypropylene shank trimmed and reinstalled in preparation for formation of the posterior portion.

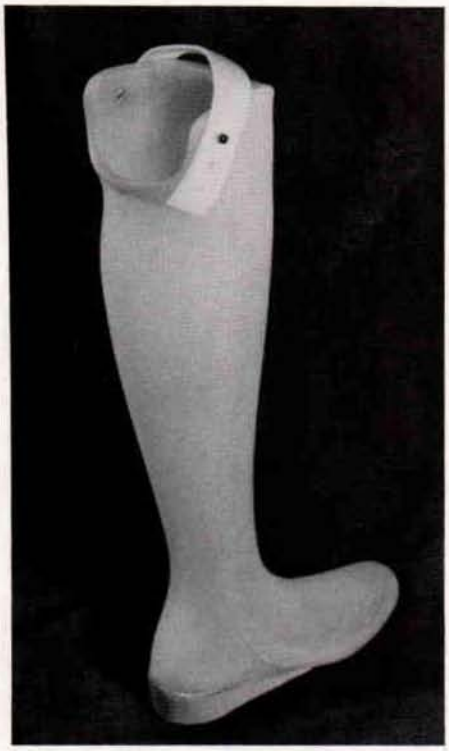

Fig. 8. The completed polypropylene ultra-light below-knee prosthesis ready for fitting.

Fig. 9. The ultra-light below-knee prosthesis worn by patient. Note the light suspension strap.

SOME RESULTS

Experiences with the two subjects to date follow:

Case No. 1

Case No. 1 is a 21-year-old male, 5′10″ in height, 145 lbs. in weight, with a right below-knee amputation as a result of a motorcycle accident. The stump is scarred and has been difficult to fit comfortably. His conventional prosthesis weighs 4.25 lbs and he has worn it for 2 1/2 years. He has been provided with 3 different polypropylene prostheses during the development period so far. The first experimental prosthesis failed structurally because of poor forming technique. The second experimental prosthesis weighs 2.7 lbs, and he has worn it 3 months because he prefers it to his conventional PTB. The third prosthesis weighs 1.5 lbs. but the socket does not fit as well as the socket in the second one, so he wears it only on special occasions. The subjects initial reaction was that it felt much lighter and required less energy to use. The subject wears his prosthesis daily for work and during sports activities such as football and swimming. Although waterproof, the buoyancy causes difficulty in swimming. This problem can probably be overcome without too much difficulty.

Case No. 2.

Case No. 2 is a 21-year-old male, 6′4″ in height, weighing 181 lbs., with a left below-knee amputation as a result of an industrial accident. He has a well healed stump with no scar tissue. He has been wearing a conventional BK prosthesis weighing 4.6 lbs, for 2 1/2 years. His first and only light weight prosthesis weighs 2.2 lbs. and he has worn this prosthesis for 3 months. His initial reaction was that the prosthesis was so light that he might have trouble controlling it, but he has accommodated to it nicely. Subject is a basketball player and he was instructed not to play basketball with the prosthesis because fatigue levels of the material in this configuration had not

been determined. At last report, subject was playing basketball without apparent damage to prosthesis.

Each of the light weight prostheses is suspended with a simple elastic strap worn above the patella (Figs. 8 and 9). Attempts will be made to develop a socket with supracondylar suspension. Expert prosthetists are being consulted in order to develop techniques that will insure quality fit between the limb and the socket.

DISCUSSION

A reduction in weight in prostheses seems to be highly appreciated by active, young unilateral below-knee amputees. Aside from energy savings, less suspension problems are encountered and control seems to be improved.

It seems obvious that the older below-knee amputee will benefit as well, if not to a greater degree. To what extent weight reduction is useful, of course, is not known, and needs to be determined. It might well be that a below-knee prosthesis can be too light, but weight reduction over standard practice certainly seems to be in order.

Polypropylene is an adequate, easily available, relatively inexpensive material for use in making below-knee prostheses. The fabrication technique needs further refinement. Wall thickness can be reduced, and better welding techniques will provide for improved appearance.

A series of patients—children, young adults, and older patients—should be provided with ultra light limbs and the results compared with the current "best" practice.

A study of the effects of the change in configuration of the foot needs to be made. This study would involve gait analysis. Should reduction in weight for some reason not be the goal, the introduction of an integral foot-ankle-shank seems still to be desirable in order to eliminate problems encountered with present foot designs.

The Rehabilitation Engineering Center will welcome the cooperation of other centers who would like to participate in a collaborative study.

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Orthotics and Prosthetics, Vol. 30, No. 1, pp. 49, March 1976

TECHNICAL NOTE

TWO COMMON PROBLEMS IN APPLICATION OF THE "MILWAUKEE BRACE"

No matter how careful the orthotist is in forming the relief for the iliac crest for the "Milwaukee Brace", the anterior bar often leans laterally in one direction or another owing to the actual scoliotic deviation, tighter muscles, or rib deformities on one side of the body.

One way of solving this problem is to provide the anterior bar with a slot at the location of the lower hole, and position the bar on the inside of the hinge. In this way the bar can be moved laterally and tilted as well. Adjustments possible by this method are small and are limited by the pelvic hinge, but the amount of adjustment needed usually is so slight that this method is adequate. When a plastic girdle is used the amount of adjustment is not limited. In this case the anterior bar may be mounted on a plate so that it is adjustable. At the same time, the anterior bar can be made to follow the midline as the correction progresses (Fig. 1).

Another problem arises in adjusting the thoracic, lumbar, and kyphosis pads under clinical conditions to obtain an instant evaluation of the positioning. Many methods of securing pads are used. One approach that has worked well for me is the use of Velcro tape. The male Velcro section is attached to the pad and the female section is sewn to the webbing or leather strap (Fig. 2). The pads may now be moved about easily to facilitate better positioning without time-consuming alterations.

Andy Parmley, C.P.O.
239 Jefferson Ave.
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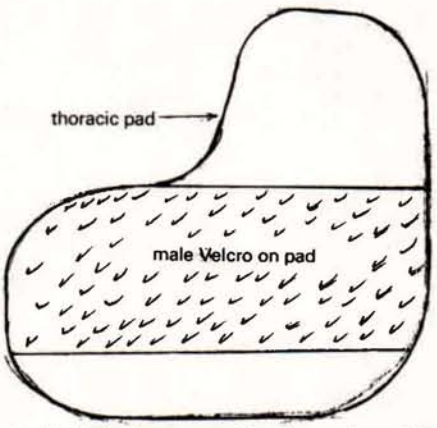


Fig. 2. Typical placement of male section of Velco tape on thoracic pad to provide for evaluation of location of the pad during clinical trials. The female section is placed on the strap.

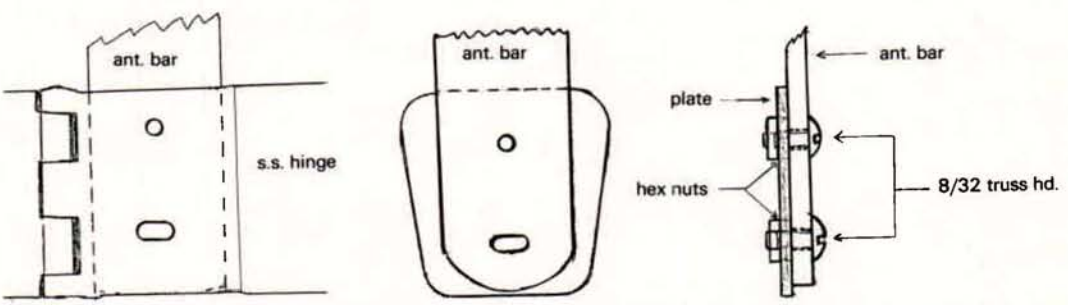


Fig. 1. An elongated lower hole is provided in the anterior bar of the "Milwaukee Brace" to allow for adjustment. The view on the extreme right illustrates this method with the conventional pelvic girdle, while the other two views show how this arrangement is used with plastic pelvic girdles.

Orthotics and Prosthetics, Vol. 30, No. 1, pp. 51–52, March 1976

NEW PUBLICATIONS

BULLETIN OF PROSTHETICS RESEARCH, BPR 10–23, Spring 1975, 399 pp., 207 illustrations, $4.25.

The Research Center for Prosthetics of the Veterans Administration announces that the Spring 1975 issue of the Bulletin for Prosthetics Research is now available from the Superintendent of Documents, U.S. Government Printing Office, Washington, D.C. 20402. The price is $4.25 postpaid.

FEATURED IN THIS ISSUE:

INDEX TO THE BULLETIN OF PROSTHETICS RESEARCH, BPR 10–21, Spring 1974, and BPR 10–22, Fall 1974.

REMITTANCES FROM ABROAD:

No additional charge is required if the address is within the United States, its possessions, Canada, Mexico, and all Central and South American countries except Argentina, Brazil, British Guiana, French Guiana, and Surinam (Dutch Guiana), and British Honduras. Where foreign mailing is required, add 25% to the total cost of the number of Bulletins ordered. Remittances from other countries should be by international money order or draft on a United States bank, payable to the Superintendent of Documents. UNESCO coupons may also be used. *Foreign money orders and postage stamps are not acceptable.*

IMPORTANT:

Please do not send any remittance to the Veterans Administration or to the Editor of the Bulletin. To do so will only delay receipt of your Bulletin, since all sales are handled by the Superintendent of Documents.

EQUIPMENT FOR THE DISABLED. The 4th Edition of EQUIPMENT FOR THE DISABLED—a series of illustrated booklets—will be published by the Oxford Regional Health Authority on behalf of the Department of Health & Social Security; the compilation and editing continues at Mary Marlborough Lodge, Nuffield Orthopaedic Centre, Oxford.

Communication, the first booklet in the NEW (4th) edition was published in September 1975 to replace COMMUNICATION (3rd edition —1971).

The next booklet will be *Clothing & Dressing* for adults. It will be followed by *Home Management*.

The booklets have been completely revised and contain new material. They are intended for use by those professionally concerned with selecting aids and equipment for handicapped persons of all ages, and "guidelines" are given to assist in the selection of an appropriate aid for a particular individual. Brief details and illustrations are included, together with an indication of the price area and manufacturers'/suppliers' addresses.

It is essential that those working in this field should keep abreast of new developments by replacing their present booklets with new ones as they become available. Details will be given in the professional journals at the time of issue.

In the meantime, the 3rd edition and its binder are available from stock. Enquiries concerning orders or information about the 4th edition should be sent to:

Equipment for the Disabled
2 Foredown Drive
Portslade
Sussex BN4 2BB

UK prices remain unchanged at £1.50 per booklet, £1.50 per binder (to hold 10 booklets), postage extra. Overseas prices quoted on request.

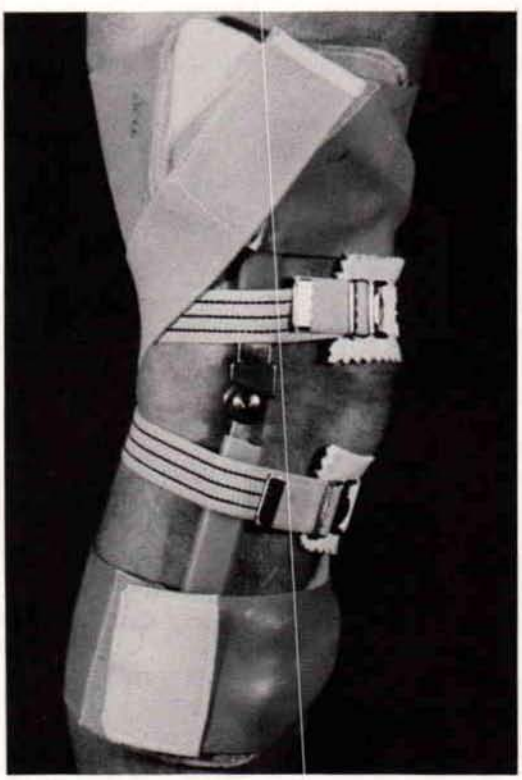
Orthotics and Prosthetics, Vol. 30, No. 1, pp. 53, March 1976

ATTENTION

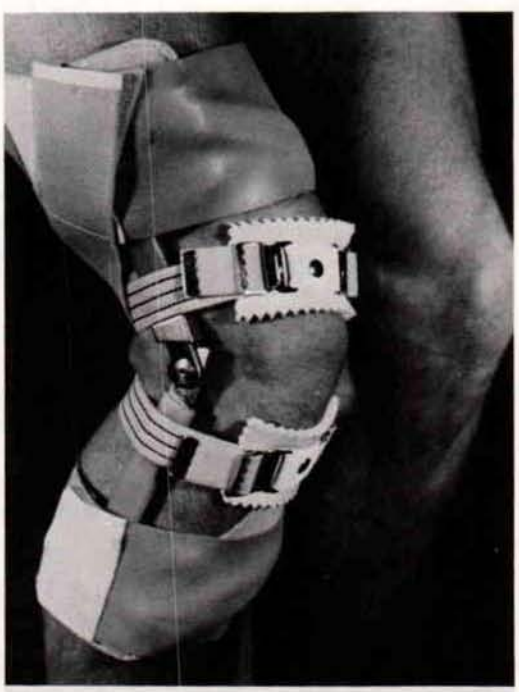
Soon after distribution of the September issue of Orthotics and Prosthetics the editor received a letter from Mr. Jack Castiglia, President of the Lenox Hill Brace Shop taking exception to the statement "The Lenox Hill derotation brace was prescribed with a slip-lock at the knee. It was learned later that the Lenox Hill device cannot be fabricated with a slip lock," which appeared in the article "Polypropylene Knee Orthosis with Suprapatellar Latex Strap" by Malcolm Dixon and Robert Palumbo. Mr. Castiglia enclosed photographs to show that it is indeed possible to incorporate a slip lock in the Lenox Hill design.

When asked about this Mr. Dixon replied that the VA clinic team had been informed by a representative of the Lenox Hill Shop, that a lock was not available at the time referred to in the article (two years ago) and went on to say: "We realize that we were remiss in using the terminology "could not" rather than "would not" with respect to the Lenox Hill slip lock." For this, we apoligize to Mr. Castiglia. Mr. Dixon also went on to state "If Mr. Castiglia or any member of his staff has published references to the slip lock in available scientific journals we would appreciate receiving reprints of such publications."

The editor also apologizes, and trusts that this mistake has not mislead any potential user of the Lenox Hill Brace.



The Lenox Hill derotation brace with slip lock.



INFORMATION FOR AUTHORS

ORTHOTICS AND PROSTHETICS
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EDUCATION

All submitted manuscripts should include:

1. THE ORIGINAL MANUSCRIPT AND TWO COPIES. If possible, the duplicate manuscripts should be complete with illustrations to facilitate review and approval.
2. BIBLIOGRAPHY. This should be arranged alphabetically and cover only references made in the body of the text.
3. LEGENDS. List all illustration legends in order, and number to agree with illustrations.
4. ILLUSTRATIONS. Provide any or all of the following:
   a. Black and white glossy prints
   b. Original drawings or charts

   Do *not* submit:
   a. Slides (colored or black & white)
   b. Photocopies

PREPARATION OF MANUSCRIPT

1. Manuscripts must be TYPEWRITTEN, DOUBLE-SPACED and have WIDE MARGINS.
2. Indicate FOOTNOTES by means of standard symbols (*).
3. Indicate BIBLIOGRAPHICAL REFERENCES by means of Arabic numerals in parentheses (6).
4. Write out numbers less than ten.
5. Do not number subheadings.
6. Use the word "Figure" abbreviated to indicate references to illustrations in the text (. . . as shown in Fig. 14)

PREPARATION OF ILLUSTRATIONS

1. Number all illustrations.
2. On the back indicate the top of each photo or chart.
3. Write the author's name on the back of each illustration.
4. Do not mount prints except with rubber cement.
5. Use care with paper clips; indentations can create marks.
6. Do not write on prints; indicate number, letters, or captions on an overlay.
7. If the illustration has been published previously, provide a credit line and indicate reprint permission granted.

NOTES:

—Manuscripts are accepted for exclusive publication in ORTHOTICS AND PROSTHETICS.
—Articles and illustrations accepted for publication become the property of ORTHOTICS AND PROSTHETICS.
—Rejected manuscripts will be returned within 60 days.
—Publication of articles does not constitute endorsement of opinions and techniques.

—Authors will be supplied with 25 reprints.

RESOLUTION CONCERNING THE METRIC SYSTEM

The following resolution was adopted by the Board of Directors of the American Orthotic and Prosthetic Association at its meeting in San Diego October 3, 1973:

WHEREAS by Act of Congress it has been determined that the United States should proceed towards adoption of the metric system as used almost universally throughout the rest of the world, and

WHEREAS the technological professions and many segments of the health professions have commonly used the metric system over an extended period of time, and

WHEREAS it is important for members of the orthotic/prosthetic professions to interact with their colleagues in the medical and technological communities for optimum patient service be it hereby

RESOLVED that the American Orthotic and Prosthetic Association endorses the use of the metric system by its members and other orthotic and prosthetic practitioners in the United States, and in witness of this endorsement and Association urges the editors of its journal *Orthotics and Prosthetics* to commence the dual reporting of weights and measurements in both the English and metric systems at the earliest possible date with the objective of employing the metric system solely by the time of the 29th Volume in 1975.

METRIC SYSTEM
Conversion Factors

LENGTH

Equivalencies

angstrom	= 1 x 10^{-10} meter (0.0 000 000 001 m)
millimicron*	= 1 x 10^{-9} meter (0.000 000 001 m)
micron (micrometer)	= 1 x 10^{-6} meter (0.000 001 m)

To Convert from	To	Multiply by
inches	meters	0.0254†
feet	meters	0.30480†
yards	meters	0.91440†
miles	kilometers	1.6093

AREA

To convert from		
square inches	square meters	0.00063616†
square feet	square meters	.092903

VOLUME

Definition

1 liter = 0.001† cubic meter or one cubic decimeter (dm^3)
(1 milliliter = 1† cubic centimeter)

To convert from	To	Multiply by
cubic inches	cubic centimeters	16.387
ounces (U.S. fluid)	cubic centimeters	29.574
ounces (Brit. fluid)	cubic centimeters	28.413
pints (U.S. fluid)	cubic centimeters	473.18
pints (Brit. fluid)	cubic centimeters	568.26
cubic feet	cubic meters	0.028317

MASS

To convert from	To	Multiply by
pounds (avdp.)	**kilograms**	**0.45359**
slugs‡	kilograms	14.594

FORCE

To convert from	To	Multiply by
ounces-force (ozf)	newtons	0.27802
ounces-force (ozf)	kilogram-force	0.028350
pounds-force (lbf)	newtons	4.4732
pounds-force (lbf)	kilogram-force	0.45359

*This double-prefix usage is not desirable. This unit is actually a nanometer (10^{-9} meter = 10^{-7} centimeter).
†For practical purposes all subsequent digits are zeros.

STRESS (OR PRESSURE)

To convert from	To	Multiply by
pounds-force/square inch (psi)	newton/square meter	6894.8
pounds-force/square inch (psi)	newton/square centimeter	0.68948
pounds-force/square inch (psi)	kilogram-force/square centimeter	0.070307

TORQUE (OR MOMENT)

To convert from	To	Multiply by
pound-force-feet	newton meter	1.3559
pound-force-feet	kilogram-force meters	0.13826

ENERGY (OR WORK)

Definition

One joule (J) is the work done by a one-newton force moving through a displacement of one meter in the direction of the force.

1 cal (gm) = 4.1840 joules

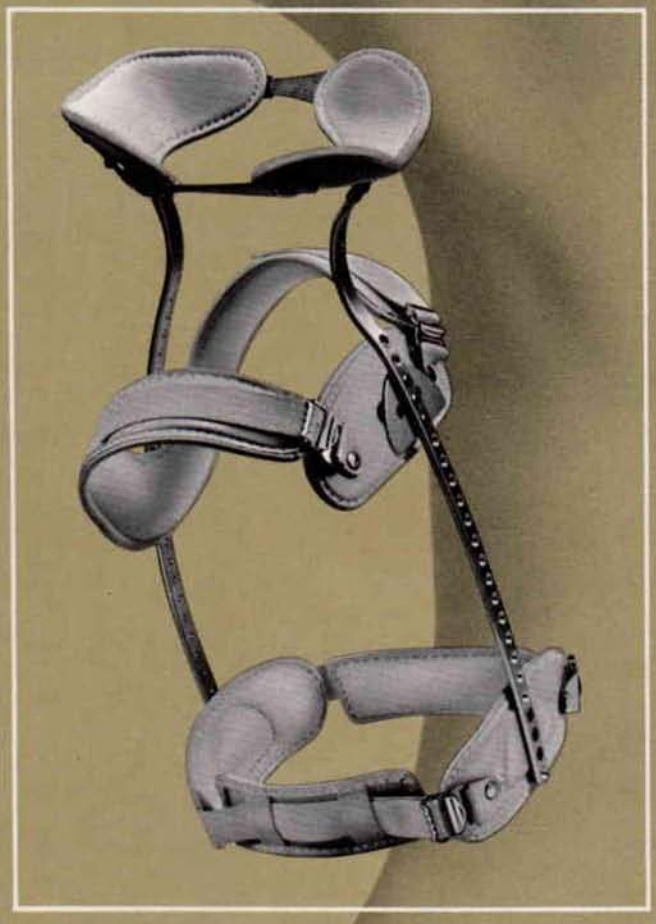
To convert from	To	Multiply by
foot-pounds-force	joules	1.3559
foot-pounds-force	meter-kilogram-force	0.13826
ergs	joules	1×10^{-7}†
b.t.u.	cal (gm)	252.00
foot-pounds-force	cal (gm)	0.32405

TEMPERATURE CONVERSION TABLE

To convert °F to °C $\quad °C = \frac{°F - 32}{1.8}$

°F	°C
98.6	37
99	37.2
99.5	37.5
100	37.8
100.5	38.1
101	38.3
101.5	38.6
102	38.9
102.5	39.2
103	39.4
103.5	39.7
104	40.0

*A slug is a unit of mass which if acted on by a force of one pound will have an acceleration of one foot per second per second.



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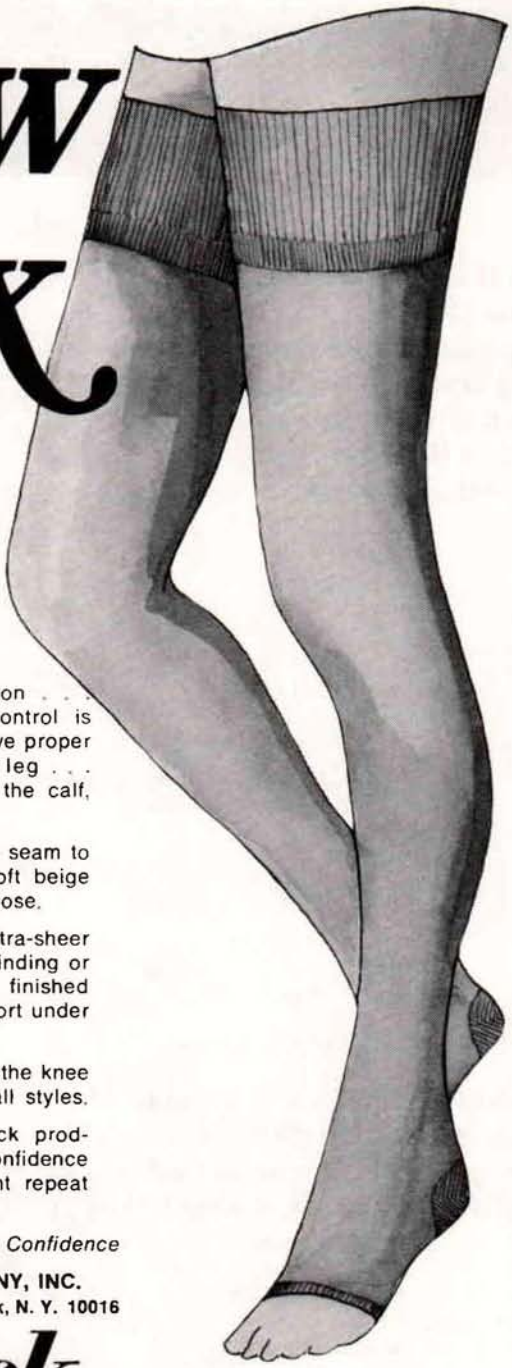

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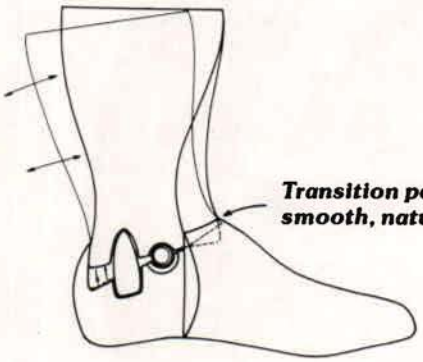

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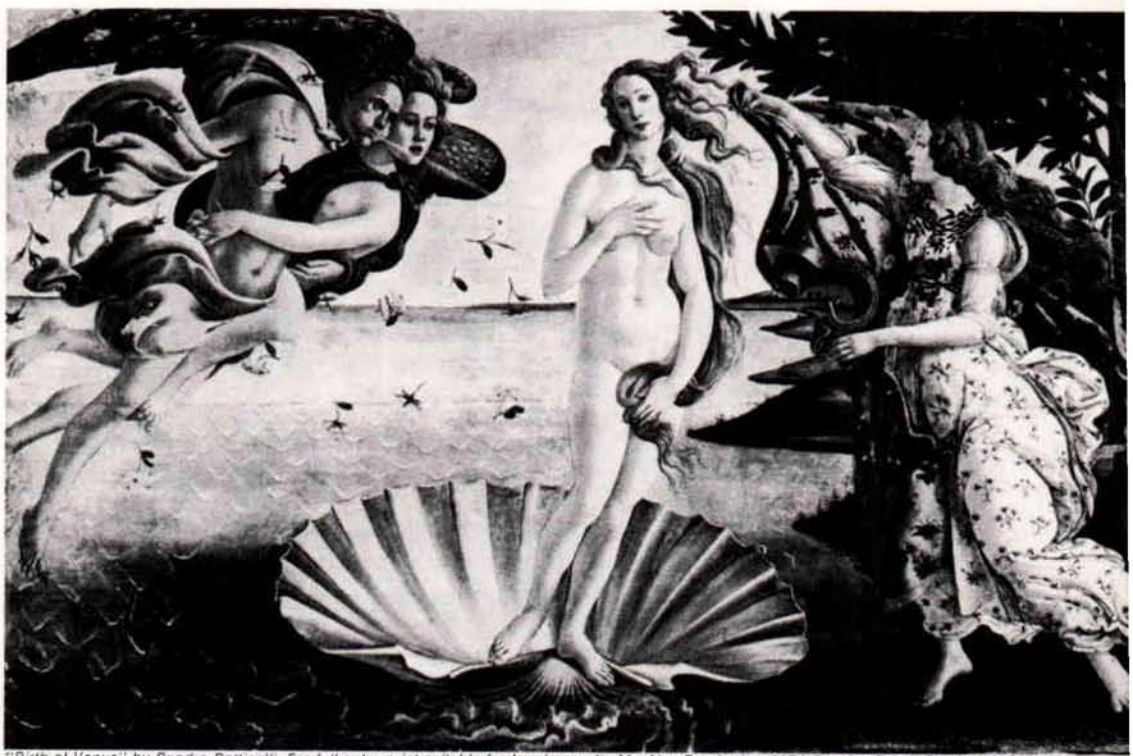

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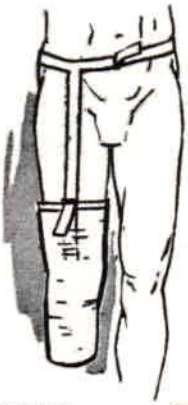

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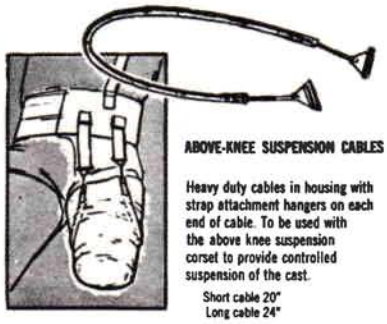

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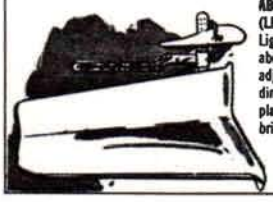

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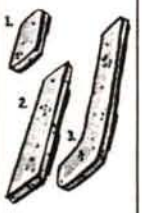
18"	6"	5"
18"	8"	6"
18"	9"	7"



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