SEMIANNUAL REPORT OF THE VETERANS ADMINISTRATION PROSTHETICS CENTER

(*JULY–DECEMBER 1963*)

INTRODUCTION

Presented in this section of the Bulletin is a report from the VA Prosthetics Center on its research, development, and evaluation activities. Henceforth, it will be the policy of the Center to publish its semiannual research and development progress reports in the Bulletin.

For this inaugural issue of the Bulletin, background information has also been included to orient readers who are not acquainted with some aspects of the Center's work. In a few instances, this has necessitated the use of material previously published, such as in the section on casting techniques.

Future research and development reports will be as comprehensive as presented here, but even more emphasis will be placed on data and results.

Acknowledgments are due to everyone in the VA Prosthetics Center for the report presented here, but special credit should be given to Thomas Pirrello, Jr. and William J. McIlmurray for assembling the various contributions.

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I. LOWER EXTREMITY PROSTHETICS

A. Casting Techniques

1. Development of Casting Stands. A casting stand has been developed for implementation of the new concept of total-contact fitting. The stand includes five adjustable components: (1) the stand assembly, (2) the crossmember assembly, (3) a posterior socket form, (4) an anterior socket form, and (5) a lateral socket form for above-knee casting (Fig. 1). For below-knee casting, the stand consists of four adjustable components: (1) the stand assembly, (2) the crossmember assembly, (3) an anterior socket form (left and right

models), and (4) a posterior socket form (Fig. 2). By simply changing socket form components, the same stand assembly and cross-member assembly can be used for both above-knee and below-knee casting.

Essentially, the adjustable form components represent the socket walls and are used to contour the elastic plaster-of-Paris wrap of the stump under conditions of weight-bearing. The plaster-of-Paris wrap helps to stabilize the stump tissues by placing them under pressures simulating firm socket contact.

The size of the casting forms already developed has proven satisfactory for most adult amputees. By scaling down the present forms, sizes for children can be readily obtained.

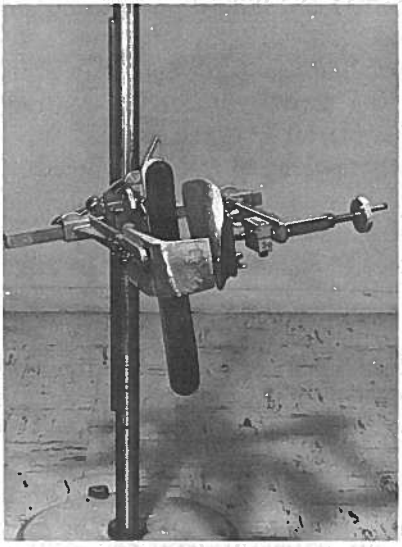

FIGURE 1. *Above-knee casting stand with socket forms.*

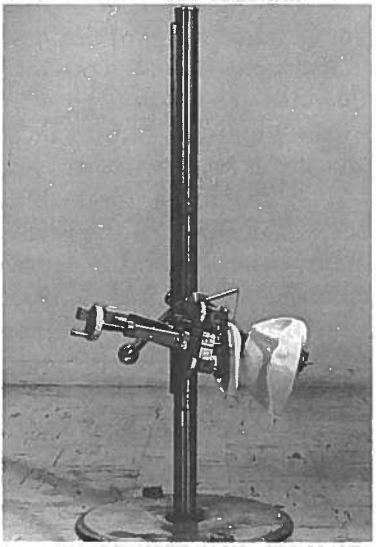

FIGURE 2. *Below-knee casting stand. Note absence of lateral socket form.*

2. Above-knee Stump Casting. To overcome the circulation problems of the above-knee stump, which occur even with the quadrilateral socket shape, a new concept of socket fitting based on total contact has been developed. The purpose of this method is to reduce the sharp contrast of forces acting on the distal and proximal ends of the stump. To overcome this force differential, a pressure gradient is computed into the socket to give low pressure at the bottom and high pressure at the top of the socket where the main control and weight-bearing occur. To provide stabilization of the stump under high force concentrations, such as are present in the stance or weight-bearing phase, the socket must be cast and shaped to accommodate the stump changes that occur during those periods of high force.

The contour of the above-knee posterior form is based upon concepts of socket shape developed from successful quadrilateral fittings and anthropometric measurements. The medial section of the form helps to establish the medial-to-lateral width of the socket and the proper position of the ischium upon the posterior socket support. The posterior form itself is adjustable with respect to the stump in a medial-lateral plane. This adjustment is used to orient the adductor longus tendon in its proper position relative to the anterior socket form.

The anterior socket form contains a proximal anteromedial flare and a rectus femoris channel. The lateral socket form consists of a flat surface adjustable for varying degrees of adduction and abduction. Adjustments can also be made for medial and lateral positioning and for a limited range of flexion-extension.

During casting, the pre-set forms provide stump support only in the intended weight-bearing areas, and the nonweight-bearing contours of the stump are formed by the elastic plaster-of-Paris wrap. When the cast is set, the fixture is then slowly lowered to free the stump.

This technique for making above-knee stump impressions is described in detail in the revised Prosthetics and Sensory Aids Service Fabrication Manual F–1, "Above-Knee Prosthetics: Stump Casting," published in October 15, 1963. The method lends itself to the fabrication of most types of ischial-gluteal-bearing above-knee sockets. The sockets can be of total-contact design, hard or soft end, and made of various materials such as wood, plastic, or leather. This casting technique has also been used in forming sockets for above-knee ischial weight-bearing braces.

3. Below-knee Stump Casting. The major weight-bearing areas of the below-knee stump have been utilized by prosthetists in the past. Clinical experience has indicated, however, that improper socket size and contouring occurring in the original fabrication process continued to produce stump-fitting problems, particularly on various bony prominences. A method was therefore developed for stump casting under weight-bearing conditions, i.e., with support of the weight limited to those areas which could tolerate it. Below-knee casting forms were therefore designed for use with the VAPC above-knee casting stand. The B/K casting stand includes an anterior form and a posterior form; no lateral form is used with the B/K casting stand.

The proximal portion or the anterior socket form is contoured to provide a patellar-tendon type of socket support. The anteromedial and anterolateral oblique walls of that component are contoured to establish firm stump contact on both sides of the tibia; these contours prevent socket contact along the anterior crest of the tibia. The concavity produced by the form contour immediately below the

patellar-tendon protuberance is designed to prevent contact with any of the bony anterior tibial protuberances.

In the posterior socket form, the proximal portion is contoured to provide the upper posterior socket flare, which is further shaped to provide for the hamstring tendons. To prevent formation of steps in the cast, the posterior form is contoured distally to follow the gastrocnemius bulge.

The below-knee casting technique is similar to the above-knee procedure. Pre-setting of the stand and form components is based on patient prosthetic information. During the trial fitting, the stump is positioned with the patellar on the tendon protuberances of the form. The optimal position is established when the stump can no longer "slip" through the socket but can still be pulled up between the forms. Alignment reference is then marked for later restoration of the patient's optimal position.

As in the above-knee procedure, the elastic plaster-of-Paris bandage is then wrapped around the stump to provide the tension necessary to maintain the volume of the stump during shaping in the casting fixture. The nonweight-bearing contours are formed by the wrap alone, and the pre-set forms provide support only in the intended weight-bearing areas.

The stump replica taken from this cast represents the shape, size, and contours of the below-knee stump under weight-bearing conditions; therefore, the PTB hard socket with a closed end and the PTB conventional socket can usually be fabricated over this cast *without* further modification. This casting technique is also used for fabrication of the below-knee weight-bearing "sockets" for braces.

A draft copy of text and illustrations for "Below-Knee Prosthetics: Stump Casting with the Use of a Casting Stand," Fabrication Manual FM–2, has been prepared for review. During April of 1964, the Center will indoctrinate the university prosethetic instructors in below-knee casting with the casting stand.

B. Socket Designs

A control study is in progress on two variations of the PTB below-knee socket design, namely, a total-contact hard socket with a soft end and a fenestrated hard socket with a soft end.

The total-contact hard socket was designed with a chamber in the distal end (Fig. 3). At a later stage of the fabrication and fitting process, with the amputee standing in the socket, the chamber is filled with a soft resilient silicone foam. Although evaluation had at first been restricted to patients who had worn PTB prostheses, subjects who have worn other types of prostheses (including side joint and lacer) are now being fitted.

The fabrication of the fenestrated socket is similar to the total-contact soft end described above, except that a section of the socket along the crest of the tibia is removed (Fig. 4). This opening (or window) provides for increased stump ventilation. Patient studies are continuing, and case reports on both socket designs are being prepared for publication.

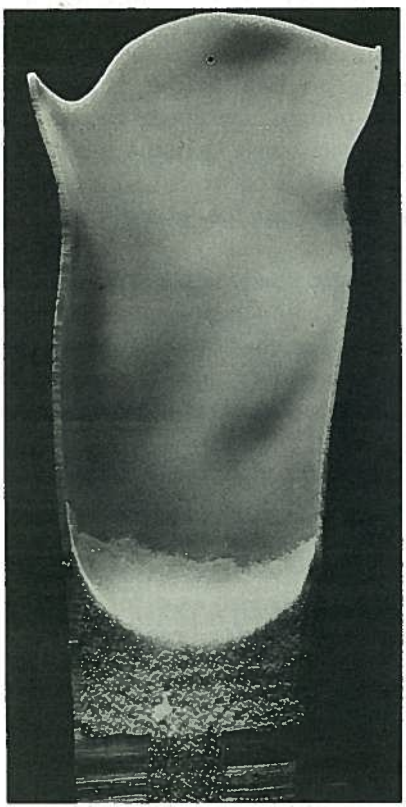

FIGURE 3. *PTB below-knee total-contact hard socket with soft end.*

FIGURE 4. *PTB below-knee fenestrated socket with soft end.*

C. Components Evaluation

1. Torque Absorber. Pelvic and tibial rotation are dynamic factors of walking that often produce serious problems for the leg amputee. Gait analysis studies have shown that the greater the restriction of axial rotation, the greater the shear stresses between stump and socket. In an effort to minimize these stump and socket shear stresses, the VA Prosthetics Center has developed a torque absorber for use in above-knee and below-knee prostheses (Fig. 5 and 6). Essentially, the device provides a means of controlling axial rotation

within the prosthesis during the weight-bearing (stance) phase of walking. This is of special value to the bilateral amputee in compensating for both pelvic rotation and tibial rotation. The main rotational element of the torque absorber is a standard bearing which is held between two aluminum plates joined by four setscrews. The top plate carries a vane that rotates against four elastomer bumpers that are set into the bottom plate. The type of elastomer used, as well as shape and size, determines the amount of resistance to rotation. The experimental model has a resistance of 40 inch-lb. per 15° deflection off center. The device is about 11/16 inch in height, 2¾ inches in diameter, and weighs about 6 oz. The torque absorber can be "sandwiched" with epoxy resin and machine screws within two pieces of wood, which can then be glued to the standard wood shank for finishing. Evaluation studies of the device are now in progress to determine its optimum location and resistance for use in above-knee and below-knee prostheses.

2. *Mechanical Friction Devices.* Two prototypes of mechanical friction devices for above-knee prostheses were received from the Prosthetics Research Center of Northwestern University for objective testing and clinical evaluation. Both are intermittent friction models that utilize a multidisk brake principle to produce mechanical friction for regulators of the velocity and travel of the shank. The friction adjustment in both devices is controlled by a spring-loaded disk, which when rotated, increases or decreases the friction. Model A, the knee-bearing device, requires straps in the friction system, but Model B, the above-knee unit, does not.

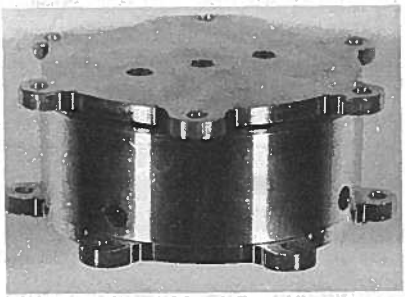

FIGURE 5. *Torque absorber for above-knee and below-knee prostheses.*

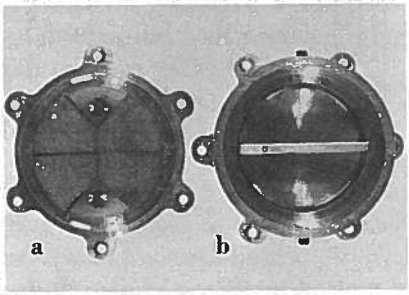

FIGURE 6. *Internal view of torque absorber. (a) Top view, elastomer inserts; (b) bottom section, vane attachment.*

The knee-bearing device is designed for mounting in the shank of a knee-disarticulation prosthesis or a supracondylar-type prosthesis. Two dacron straps are attached to the socket, one anteriorly and the other posteriorly, to control heel rise and terminal deceleration.

The intermittent friction device is designed for mounting in the shank of an above-knee prosthesis and is restricted to those knee-shank setups with a 9/16-inch diameter knee bolt. Model B is designed with a three-step adjustable friction buildup in both directions of swing, with varying degree of free motion. Optional installation of an extra disk decreases the degree of free motion. Two bumpers (stops) mounted in the knee control slot (one anterior, the other posterior) contact the friction disks, successively varying the resistance to flexion and extension of the shank. Preliminary evaluations and biomechanical analyses have been conducted; Figures 7 and 8 illustrate resistance curves for both devices.

A limited number of the devices has been issued to the field for further clinical data. A period of 90 days of wear is proposed as sufficient time for patient adjustment to socket fit, alignment, and functional characteristics of the device. At the end of this period, clinical experience will be recorded, and biomechanical evaluations will again be conducted on amputee and prosthesis.

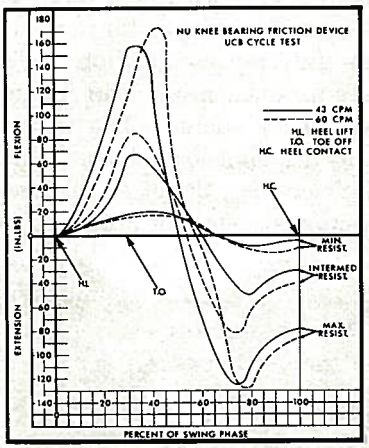

FIGURE 7. *Resistance curves of N.U. knee-bearing friction device.*

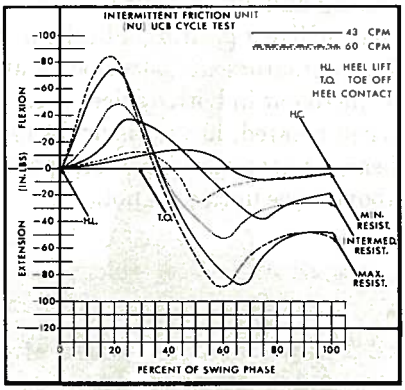

FIGURE 8. *Resistance curves of N.U. above-knee intermittent friction device.*

3. Plastic Knee-shank Setup. Evaluation was conducted and completed on the 3P4 Safety Knee with a newly developed plastic foam knee and shank setup, supplied by the Otto Bock Co. Since the hardware and locking mechanism were of the same design and material as the earlier accepted model, 3P4 safety knee (wood), the primary objective of this evaluation was the analysis of the plastic foam substituted for poplar wood in the setup. The hardware and plastic knee-bolt bushings were molded in or inserted directly into the foam. The overall weight of the complete plastic knee-shank setup as received

from the manufacturer was 3 lb. 12 oz. Test data included density, water absorption (ASTM 24 hr.), impact strength (ASTM and IZOD impact strength), and compression yield stress, as well as destructive testing (Carver hydraulic press) of the entire plastic knee-shank setup.

Results indicated that the physical properties of the plastic foam were more than satisfactory in the areas of testing, relative to the prosthetic application. Although no clinical evaluation was required, the four units purchased will be used in the fabrication of prostheses for selected patients in order to obtain supplementary data.

4. Air-exchange Valve for Stump Sockets. Studies have been initiated on three samples of an impulsion-type socket valve submitted by the Navy Prosthetics Research Laboratory (NPRL) for VAPC evaluation. According to NPRL, this air-exchange valve was designed to ventilate the stump and stump sock within a total-contact or closed-end-type socket. The device is a two-piece unit, consisting of a valve chamber with air exchange holes and a ball made of polyurethane. The valve is installed flush with the distal inner surface of the stump socket so as not to impinge upon the stump end. During the walking cycle, air is drawn into the socket as the stump moves upward; as the stump moves downward, the ball drops to the valve seat, forcing the entrapped air upward and out of the socket brim. This action is continuous throughout the walking cycle.

According to the NPRL, the valve has been used with relative success in patellar tendon-bearing prostheses, slip-socket below-knee prostheses, and Syme's prostheses; smaller valves have also been used in sockets for above-elbow amputees.

The VAPC sample valves were installed in standard PTB prostheses in a minimal amount of time. Evaluation continues pending meaningful reaction and response from the test patients.

D. Fluid-controlled Knee Mechanisms

A relatively long-term study has been initiated to investigate the effects of several fluid-controlled knee mechanisms on the locomotion patterns of above-knee amputees. Data from these tests will be used as a basis for (1) more complete prescription indications and (2) for determining whether special fitting techniques are necessary to obtain optimum value from a specific device.

Currently, 10 above-knee amputees are being fitted with a succession of eight fluid-controlled knee mechanisms. Criteria for selection of amputee subjects include the active use of a conventional prosthesis with single-axis, mechanical-friction knee joint, and freedom from medical complication. Each subject is also screened by the staff physician whose examination is used to provide baseline data on somatotype, nutritional status, condition of stump with respect to skin,

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muscular status, range of motion, vascularity, sensitivity, and similar noteworthy features of the sound limb.

The mechanical characteristics of each knee unit are evaluated before fitting. Bench control tests to determine uniformity and reliability of flexion-extension characteristics under various temperatures, torque, and adjustment settings are run on each mechanism as it is received from the manufacturer. A second test procedure cycles thigh-knee shank assemblies under predetermined inputs at the simulated hip joint. This test was designed in an attempt to correlate knee mechanism function as determined on a mechanical "gait" simulator with corresponding functions when the unit is worn by an amputee. This design problem, though formidable, warrants further exploration.

After mechanical testing, the knee unit is installed in the subject's conventional prosthesis, which has been carefully evaluated as to foot, knee, and socket alignment; repairs are also made on the prosthesis if needed. The subject's performance on the conventional leg, prior to installation of the knee mechanism, is then recorded. If any deviation from conventional alignment is required to obtain optimal fit, this information is recorded, as well as data to determine the effects of (1) functional characteristics of the unit on socket fitting; (2) weight distribution (center of gravity of leg) on resistance characteristics and alignment of prosthesis; (3) resistance characteristics on alignment; and (4) subject reaction to the limb during the wear period.

In the biomechanical analysis of each fluid-controlled device, the variables most directly related to the function of the knee-joint mechanism in level walking are of primary interest. For this reason, velocity of the center of gravity, acceleration of the shank at varying cadences, and metabolic cost rates are determined. Knee mechanisms also feature control of swing and stance; therefore, biomechanical analysis of the fluid-control unit in ascending and descending slopes and stairs is very important. Evaluation procedures include the measurement of displacement, velocity and acceleration of the center of gravity (CG), and angular displacement, as well as velocity and acceleration of the ankle, knee, and hip during ascent and descent at various degrees of ramp inclination. Observations are made on the method of descent (e.g., one-step, jackknife, or step over step), and objective determinations are made of the vertical and horizontal CG displacement and velocity, stance phase time, and angular acceleration of the knee joint.

As the performance evaluation on each device is completed, the fitting procedure is repeated with another device, which is randomly selected. To date, nine knee mechanisms have been fitted, and biomechanical runs have been completed on eight of these.

II. LOWER EXTREMITY BRACES

A. Evaluation

1. Below-knee Weight-bearing Brace With PTB-type Socket. Studies have been continued on the use of a below-knee weight-bearing brace for unweighting the leg is cases with a physiologically sound lower extremity except for the B/K pathology. The socket, brace bars, and shoe constitute the three major components of this brace (Fig. 9 and 10). The socket closely resembles the PTB design but is hinged posteriorly to provide entrance for the leg. The bars and stirrup are of conventional design and may be purchased commercially. The shoe is modified to incorporate a rocker bar and a compressible SACH-type heel insert. This type brace permits the patient with below-knee involvements to ambulate with reasonably normal knee function.

Since its inception, this orthesis has been used to unweight legs for patients with pathologies that included malunions of the tibia and fibula, partially ankylosed ankles, and arthritic ankles and feet.

As a result of a recent Committee on Prosthetics Research and Development meeting, evaluations on the use of the below-knee weight-bearing brace for pathologies indicated above are to be conducted by New York University. It is likely that this study will include congenital leg anomalies, but only those cases with sound knees. For such cases, modifications of the brace uprights will be necessary for assembly of the prosthetic foot to the system.

2. Electronic Muscle Stimulator. A portable electronic muscle stimulator, the Theratronic Muscle, designed primarily for patients with residual drop-foot conditions caused by hemiplegia or hemiparesis was evaluated. The muscle stimulator is designed to be used in place of a leg brace. The apparatus consisted of the following four major components (Fig. 11):

a. A stimulator and battery pack fitted with belt loops. The nickel-cadmium battery is rechargeable; the pulsed signal range is between 20 and 80 volts and is adjustable. (The adjustment to be made by a physician or physical therapist.)

b. A pulse applicator, consisting of a small flexible electrode covered by an absorbent pad, which is worn over the motor point of the paralyzed muscle. The pad is saturated with electrolyte and replaced daily.

c. A reference electrode, which is worn over the calf of the leg to complete the circuit. It too must be saturated with electrolyte and replaced daily for proper conductivity.

d. A heel switch enclosed in rubberized cork. The heel switch is turned automatically off and on during the normal walking cycle, thus

governing the muscle-activating pulse. It is worn in the heel of the shoe.

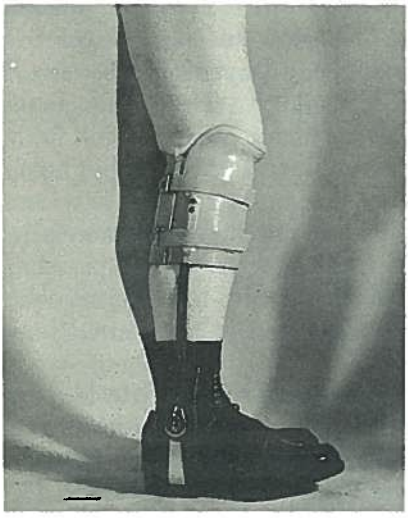

FIGURE 9. *Lateral view of below-knee weight-bearing brace with adjustable PTB-type socket.*

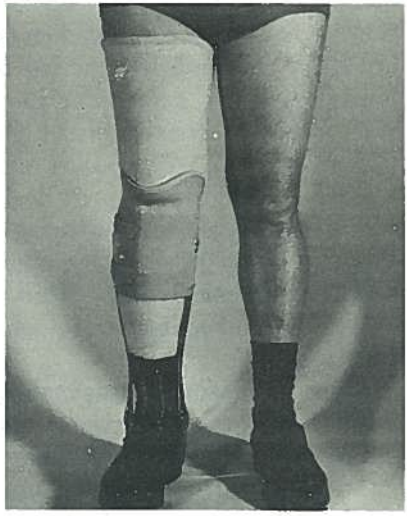

FIGURE 10. *Anterior view of the below-knee weight-bearing brace with PTB-type socket.*

Throughout the walking cycle, the activating pulse to the paralyzed muscle is turned on when the heel is raised and stays on until the foot is set down again. When the patient stands, the switch is closed and the current stays off.

Experience to date with this device included clinical observation of three patients and a biomechanical analysis on one.

In the Bioengineering Laboratory, an evaluation was made of the subject's performance (1) without assistive devices, (2) with the conventional drop-foot brace, and (3) with the muscle stimulator (Fig. 12 and 13). Performance in each of these three conditions was compared with that of the sound limb in respect to (1) step time; (2) foot slap (time from heel contact to foot flat); (3) toe clearance of the floor in midswing; (4) average vertical hip oscillation; (5) average vertical knee oscillation; (6) knee angle during midswing; and (7) ankle dorsiflexion angle in midswing. Comparative results of walking performances without device, with brace, and with stimulator, were as follows:

1. Step time: Almost identical for the sound limb and the affected limb under all conditions.

2. Foot slap: The slowest heel-contact-to-foot-flat time was with the drop-foot brace; the fastest, with the muscle stimulator. This in-

dicated that wearing the stimulator resulted in the greatest tendency for foot slap.

3. Toe clearance: Identical at midswing for the sound limb, the affected limb without any device, and the affected limb with the drop-foot brace. With the muscle stimulator, toe clearance seemed excessive, being 2.2 inches from the floor to the most distal portion of the shoe.

4. Average vertical hip oscillation: Approximately equal with the stimulator and without any device, both cases being significantly greater than for the sound limb and the affected limb with the conventional brace.

5. Vertical knee oscillation (the height to which the knee is raised during swing phase): Greatest with the stimulator (3.08 inches in contrast to 2.2 inches in the other conditions).

6. Knee angle during midswing: increased with the muscle stimulator, possibly indicating that as the heel rose off the ground, the stimulator was energized and maximum dorsiflexion was produced (approximately 20 degrees, with 15½ degrees remaining through midswing range). Figures 12 and 13 corroborate the extensive dorsiflexion attained after midswing. These curves also suggest that a certain amount of functional regain was attained since, without any device, the patient achieved almost 12 degrees of dosiflexion between midswing and heel contact.

The electronic muscle-stimulating device appeared to be a potentially valuable orthotic aid. Nevertheless, experience to date has indicated that the electronic muscle stimulator tested was not yet suitable for general use.

Recommendations to the manufacturer included suggestions for: (1) a better means of locating and fixing the electrode over a motor point; (2) a redesigned heel switch; (3) a more convenient means of voltage regulation and switch control; and (4) a means of replenishing electrolytic jelly without removal of the electrode. At the request of the developers, further investigation of the electronic muscle stimulator may be continued by this Center.

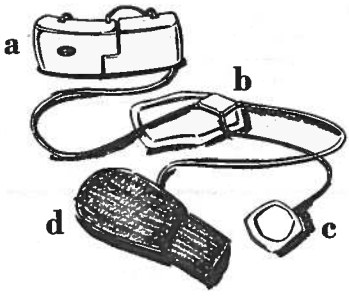

FIGURE 11. *Theratronic muscle stimulator: (a) stimulator and battery pack; (b) reference electrode; (c) pulse applicator; (d) heel switch.*

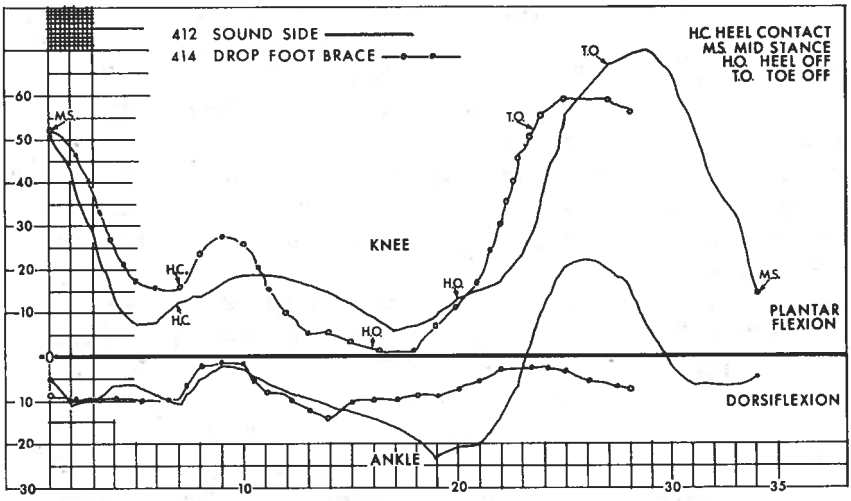

FIGURE 12. *Gait curves of patient on sound leg and on leg with drop-foot brace.*

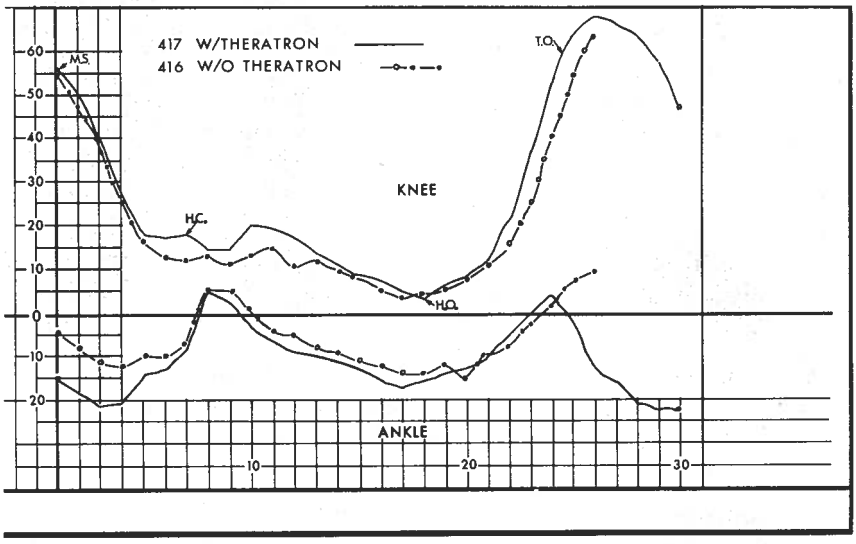

FIGURE 13. *Gait curves of patient walking, with and without electronic muscle stimulator.*

B. Development

1. Single-bar Brace. The work of others in the field stimulated the VA Prosthetics Center to investigate the designs of a single-bar brace (Fig. 14 and 15). Although the unibar brace principle had previously been applied only to cerebral palsy cases, the VA Prosthetics Center has had reasonable success fitting braces of this design to polio, multiple sclerosis, and paraplegia and hemiplegia patients.

There are many advantages to single-bar bracing. By eliminating the medial bar, the weight and bulk of the device are reduced. This reduction of weight helps to conserve the energy of the patient, who naturally has below-normal muscle power. The bilateral brace wearer benefits particularly from the reduction in bulk since he may often need to wear other devices, such as urinary appliances, along the medial aspect of his leg.

FIGURE 14. *Single-bar brace.*

FIGURE 15. *Anterior view of patient wearing single-bar braces. Note the lack of brace interference with the urinary appliance.*

The unibar is an improvement over double-bar devices because the brace joints are more easily related to the anatomical joints with a single bar. Normal or abnormal tibial torsion can also be accommodated more readily than in a double-bar brace. Adjustments are also

relatively simple for talipes valgus and talipes varus, foot deformities in which the heel is turned outward from the midline of the leg and inward from the midline, respectively. Corrections for toe-out may be adjusted relatively easily without the strain on metal joints and band inserts that might occur with double-bar braces.

The prevention of rotation of the extremity is effected not only by the uprights but also by the bands and cuffs on thigh and calf. Indeed, the uprights would be of no support value whatsoever if the cuffs and bands were not fitted securely. The design of the thigh band eliminates bracing on the medial upper portion of the thigh, and the design of the calf band eliminates the necessity of having a knee-cap form. The thigh and cuff bands, with anterior openings, encircle about two-thirds of the extremity, giving more support than the half bands generally used in the double-bar braces.

Experience with 10 patients indicated that most commercially available brace parts are not usually adequate for single-bar braces. Ball bearing joints at the ankle, the knee, and hip are anticipated as an innovation for single-bar braces.

Bars of different cross sections and shapes are to be studied, and tests will be performed on different types of metals. Because of inherent advantages to both patient and orthotist, the fitting of single-bar braces will be continued, utilizing parts currently available.

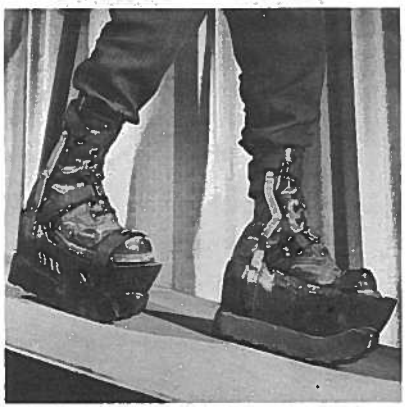

FIGURE 16. *Sabots being tested for hill descent.*

III. SHOES

A. Evaluation of Protective Footgear (Sabot)

At the request of the Army Medical Biomechanical Research Laboratory (AMBRL), the Center evaluated the Sabot, an armored footgear designed to reduce the incidence and severity of injury to personnel engaged in mine-clearing operations (Fig. 16).

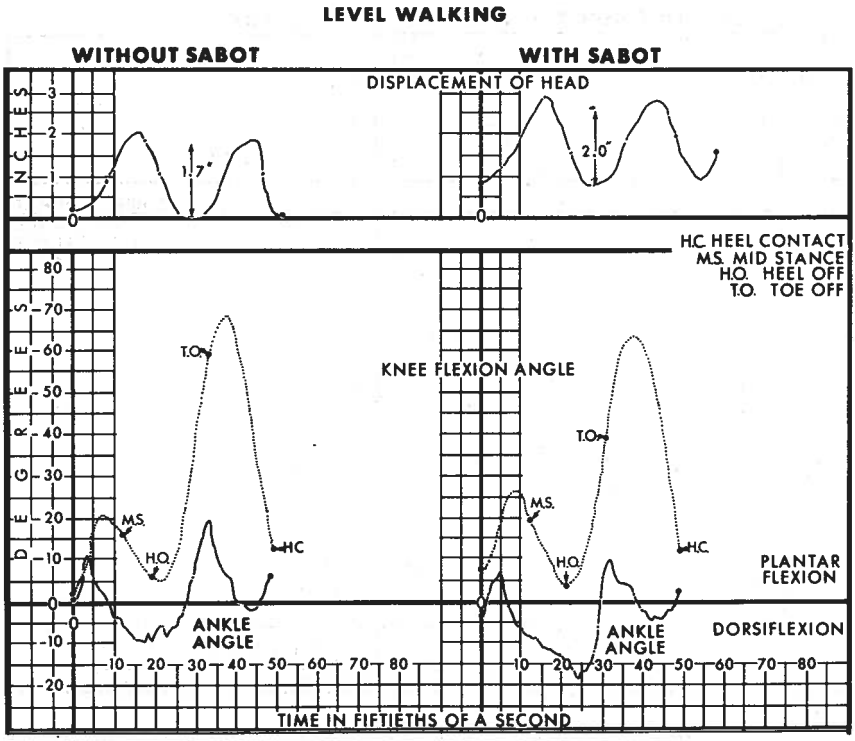

Figure 17. *Gait patterns in level walking, with and without Sabots.*

The device, originally developed by the Marine Corps (Naval Medical Field Research Laboratory), was improved at AMBRL through redesign of the bracing system by which the Sabot is attached to the wearer's leg. The two major modifications for VAPC evaluation were (1) relocation of the ankle-joint axis to approximate the average vertical and mediolateral positions of the anatomic ankle axis, and (2) increase in the width of the calf band to reduce unit surface pressure at the calf-band-boot interface.

A study was designed to compare the performance of a subject walking with and without the Sabot while wearing a conventional military field uniform, including field boots. Three areas of performance evaluation were considered:

1. Analysis of level walking in which center of gravity oscillations and limb segment displacements were compared;

2. Hill ascent and descent in which heel-toe transition and phasic knee angular displacement were compared;

3. Energy costs in which the relative stress of walking with the Sabot was determined.

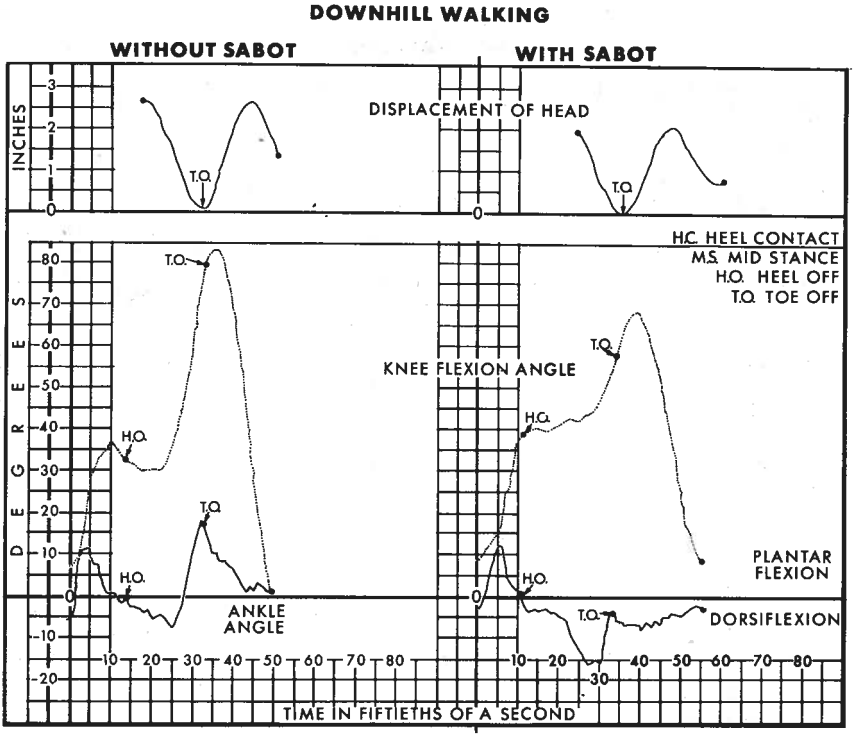

FIGURE 18. *Gait patterns in downhill walking, with and without Sabots.*

By means of force plates, ground reaction force data were recorded. The cyclograph provided interrupted light photographs at the rate of 50 exposures per second. The tachograph was used to measure instantaneous horizontal velocity of the center of gravity (CG). All of these data were obtained simultaneously. Energy costs were calculated on the basis of oxygen utilization measures obtained by the open-circuit method.

The subject of this study was a 26-year-old male soldier in good health. He was 72¾ inches tall and weighed 163¾ lb. He participated in the development of the Sabot and was thoroughly trained in its use.

Wearing the Sabot in level walking did not produce significant gait alterations as regards vertical CG oscillation or knee-flexion patterns; ankle action was modified more extensively with increased ranges of dorsiflexion between midstance and heel-off and decreased plantar flexion from heel-off to toe-off (Fig. 17). This alteration, however, did not appear to affect the effort or symmetry of the total gait pattern nor to change significantly the temporal pattern of the stance-swing cycle.

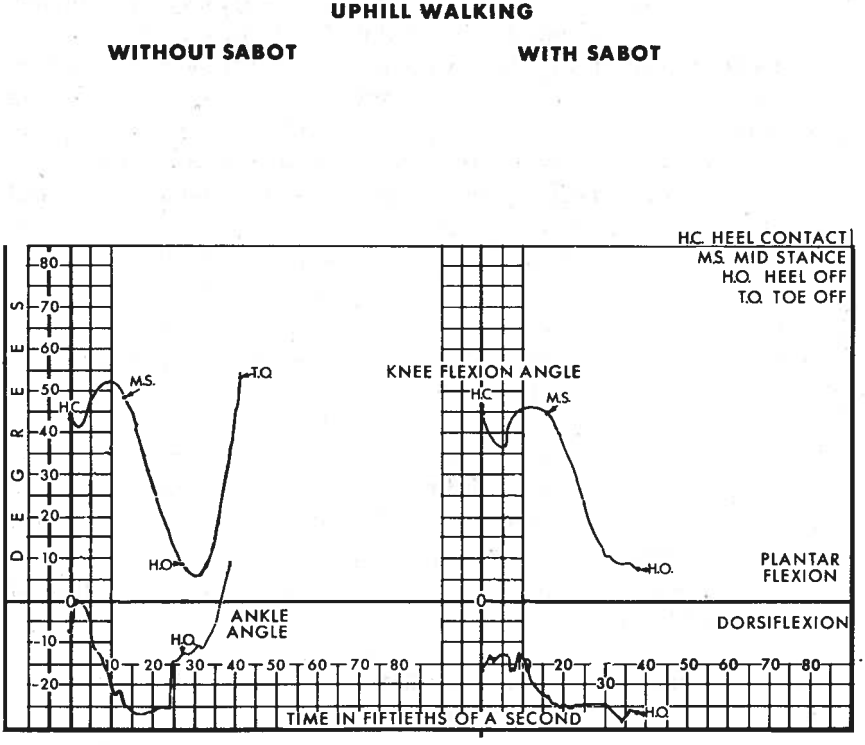

FIGURE 19. *Uphill-walking curves, with and without Sabots.*

Wearing the Sabot in descending a slope (12.5° slope) produced some adaptive alteration of the subject's normal gait pattern, but the sum of the difference was not of great consequence (Fig. 18).

The most general effect of the Sabot in ascending an incline (12.5° slope) was an increased step time. With the Sabot, heel-contact to heel-off required $^{38}/_{50}$ second, as against $^{28}/_{50}$ second without Sabot. Knee flexion ranges were similar (Fig. 19).

Wearing the Sabot did not significantly affect the high-energy consumption portion of gait, the vertical CG oscillation, or the character of knee action.

The test procedure required the subject to walk 14 laps on a rectangular track 172-foot in circumference, a total of 2,408 feet. He performed at three cadences, 80, 110, and 120 steps per minute, while wearing a standard military boot and while wearing the standard boot with Sabot.

Prior to each trial, the subject remained seated for a period of 30 minutes. During the last 9 minutes, the resting metabolic rate was determined. Immediately following the exercise, the subject was

seated and his metabolic rate was again recorded to determine whether an oxygen debt had been incurred. (In this study, oxygen debt was taken as a criterion of undue stress; the absence of oxygen debt was taken as an indication of a level of activity that could be maintained over long periods of time.) Pulmonary ventilation was measured during the walking activity. Relative costs were estimated, utilizing an oxygen equivalence ratio. In addition, pulse rate and blood pressures were recorded before exercise, immediately after exercise, and again at the end of the recovery period.

No oxygen debt was incurred by the subject when walking with the Sabot at cadences of 80 and 110 steps per minute or when walking without the Sabot at any cadence. When walking with the Sabot at 120 steps per minute, an oxygen debt of 94 cc. 0_2 per minute was incurred. The relative energy costs of walking at each of the selected cadences both with and without the Sabot are shown in Table 1.

At the low cadences generally utilized by a soldier on patrol or mine-clearing operations, the Sabot did not increase the normal energy cost of walking; at 120 steps per minute, however, the Sabot incurred a 21 percent increase in energy expenditure.

The device seems adequate in its present configuration, in spite of its deviations from normal ankle action.

TABLE 1

ENERGY COST

(cc. 0_2 per minute)

Cadence (steps per minute)	With Sabot	Without Sabot	Percentage difference
80	795	809	2
110	925	865	7
120	[1] 1146	943	21

[1] Oxygen debt included.

B. Preliminary Study on Conventional Shoe Wedges

The objective of this study on conventional shoe wedges was to determine the effects of external shoe wedges on the pronation and supination of the foot. It was specifically designed to measure changes in the attitude of the foot effected by conventional wedges of varying thickness. Each of the following types of external wedges in ⅛-, 3/16-, and ¼-inch heights was tested: Lateral heel, medial heel, lateral heel and sole, and medial heel and sole. The effects of these applications were observed while subjects stood and walked while (1) barefooted, (2) wearing shoes without wedges, and while (3) wearing shoes with wedges.

Clinically, the effects of wedging are evaluated by observation of heel-cord deviations from the vertical. Karpovich and Wilklow [1]

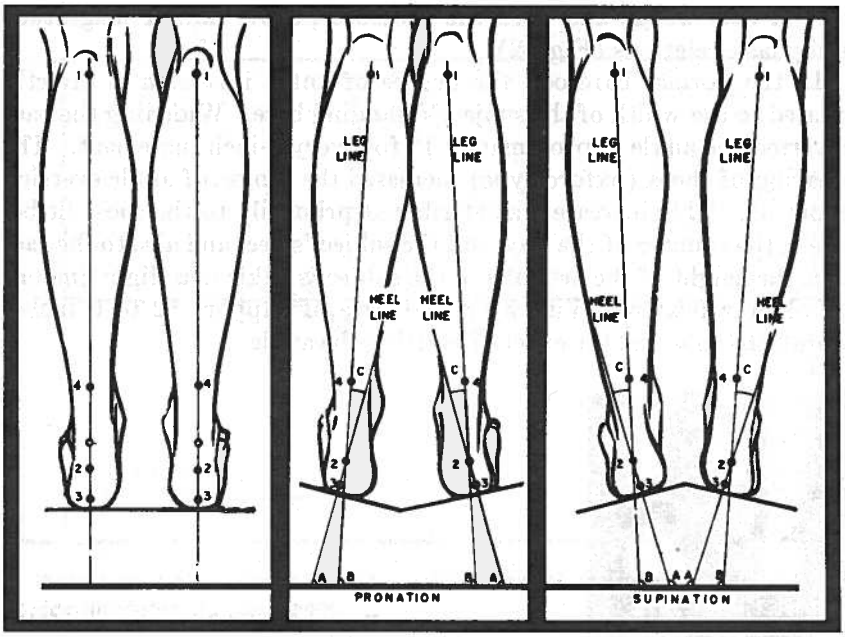

Figure 20. ***Angle C used by Karpovich and Wilklow as the point of measuring motion around the subtalar axis.***

have used a somewhat more objective technique to measure motion around a point roughly approximating the subtalar axis, which they have designated angle *C* (Fig. 20). This technique is extremely difficult under dynamic conditions, however, so another method was used in the initial phase of the VAPC study, pending the development of more precise techniques.

Although patients with various types and degrees of foot deviations will be the subjects of this study, normal subjects were used in the initial phase because of the immediate need to develop appropriate instruments and testing techniques. One of the first testing devices developed was a wedge simulator designed to reduce the time required for installation and removal of a large number of wedges throughout the study. This instrument was designed as a scaled-down modification of the VAPC coupling mechanism (Fig. 21). Installed in the hollowed-out heel of the subject's shoe, the device produces a convenient and consistent control method of raising and lowering the lateral or medial sides of the heel.

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[1] Karpovich, P. V. and L. B. Wilklow, A goniometric study of the human foot in standing and walking, *U.S. Armed Forces Medical Journal*, X:8:886, Aug. 1959.

To validate the use of the device, a comparison was made of the measurements obtained with both the wedge simulator and actual wedges. Curves of changes at angle *C* obtained with a leather ¼-inch medial heel wedge and with the simulator, show similar magnitude and phasic relations (Fig. 22).

In the normal barefoot, the degree of ankle inversion is directly related to the width of the subject's standing base. Widening the base inverted the ankle approximately 1° for every 2-inch increment. The wearing of shoes (oxford type) increased the ranges of ankle eversion about 3°. This increase was attributed primarily to the loose fit between the counter of the shoe and the subject's heel and also to the fact that the height of the heel placed the subject's ankle in a slight amount of plantar flexion. With a normal base of support (2 to 6 inches wide), the shoe had the effect of everting the ankle.

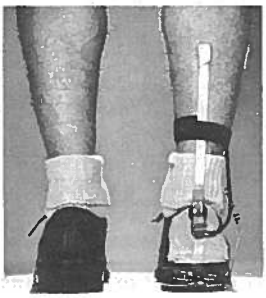

FIGURE 21. *VAPC wedge simulator. The vertical tape on leg facilitates measurement of angle C.*

The addition of ⅛- and 3/16-inch medial heel wedges tended to invert the ankle approximately 4° to 5° per ⅛-inch increment. The ¼-inch wedge was not as proportionately effective, producing only a very small additional change.

The subject's stance, whether with feet parallel or in a "comfortable" position with the natural toe-out, produced no measurable effect on the results. It was also noted that only 2 of the 11 cases had the same amount of toe-out in both feet. The average difference was about 1° more of external rotation of the right foot than of the left.

In spite of differences in magnitude effected by wedge thickness, a consistent curve pattern of eversion-inversion has been demonstrated. A typical curve is shown in Figure 23.

At normal speeds, the foot at heel contact is in 6° of inversion. As the vertical load is applied over the foot, eversion increases sharply, reaching a maximum of almost zero degree of inversion (neutral position). During midstance the foot begins to invert gradually until push-off when it inverts sharply, returning to the original 6° of inversion at toe-off. At slow speeds, the pattern is the same, but the inversion is more gradual; at fast speeds, inversion is rapid.

Validation of these techniques has resulted in the establishment of standards for further study of patients with various types and degrees of foot deviations.

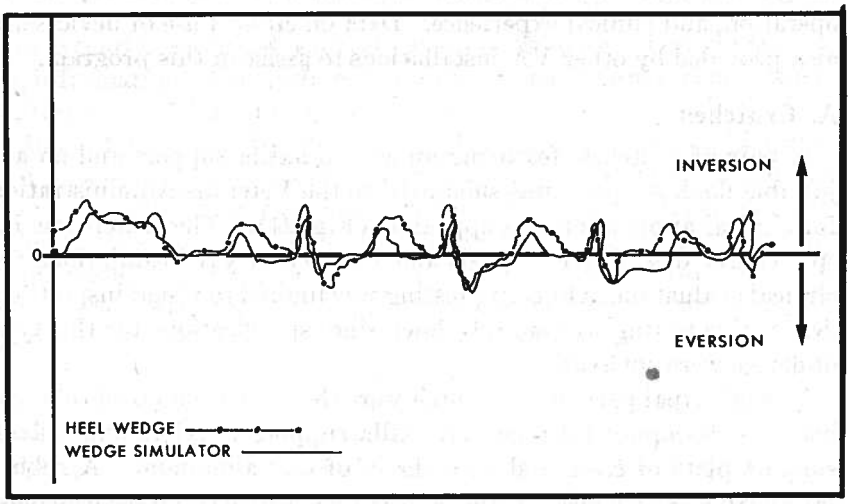

FIGURE 22. *Gait patterns obtained with standard test wedges and with the wedge simulator at normal pace.*

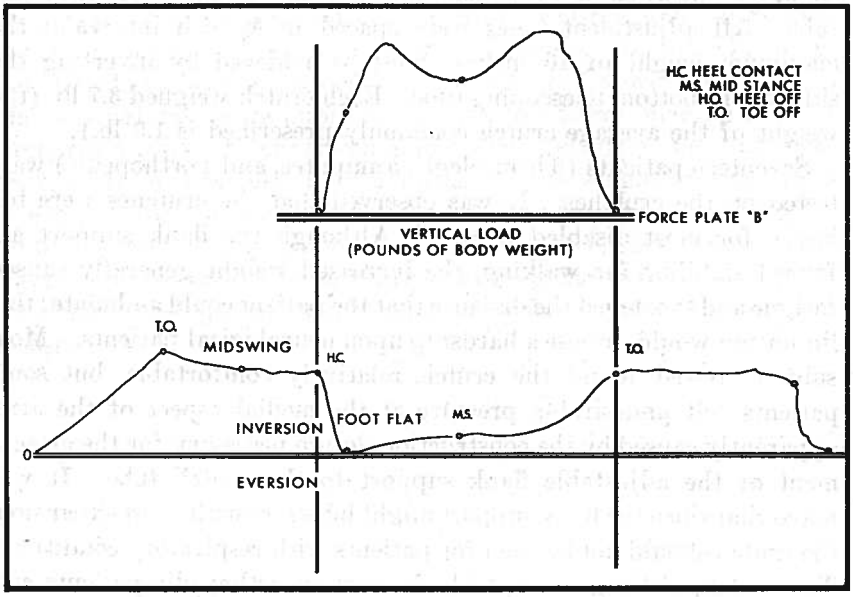

FIGURE 23. *Typical pattern of eversion and inversion of the foot during the gait cycle, walking west at normal speed.*

IV. ORTHOPEDIC AIDS EVALUATION

As part of its continuing program of systematic evaluation of orthopedic aids, the Center considers such devices from several aspects, i.e., design and material, mechanical function, biomechanical aspects of operation, and clinical experience. Data on clinical use of devices are also provided by other VA installations to assist in this program.

A. Crutches

A pair of crutches, featuring an arched axilla support and an adjustable flank support, was submitted to the Veterans Administration for clinical application and appraisal (Fig. 24). The crutch was inspected for descriptive analysis and issuance to VA installations for clinical evaluation. Objective testing was limited to visual inspection; destructive testing was not scheduled since specifications for this type of device were not available.

The principal parts of the crutch were the main tube (steel tubing), bottom telescoping tube (steel), axilla support (steel tubing), flank support plate of steel, and a handgrip of cast aluminum. A rubber safety tip was provided for the bottom tube, which could be telescoped within the main tube, giving a total adjustment range of from 42½ to 52½ inches. One side of the telescoping tube housed a spring clip, engaging one of eight adjustment holes on the bottom end of the main tube. All adjustment holes were spaced in ¾-inch intervals; the maximum height of 10 inches could be achieved by inverting the sides of the bottom telescoping tube. Each crutch weighed 3.7 lb. (the weight of the average crutch commonly prescribed is 1.9 lb.).

Seventeen patients (6 hemiplegic, 5 amputee, and 4 orthopedic) were tested on the crutches. It was observed that the crutches were too heavy for most disabled persons. Although the flank support afforded stability for walking, the increased weight generally caused fatigue and shortened the distance that the patient could ambulate; this limitation would impose a hardship upon neurological patients. Most subjects tested found the crutch relatively comfortable, but some patients felt undesirable pressure at the medial aspect of the arm, apparently caused by the construction design necessary for the attachment of the adjustable flank support to the crutch tube. It was noted that since the flank support might interfere with chest expansion, the crutch should not be used for patients with respiratory conditions. The crutch, although acceptable for certain orthopedic patients and amputees, would require a decrease in weight and an increase in range of height adjustment to be satisfactory for general use.

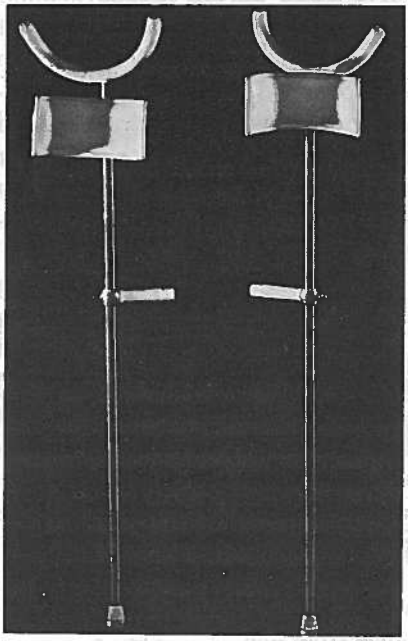

FIGURE 24. *Newly designed crutches with arched axilla and adjustable flank supports.*

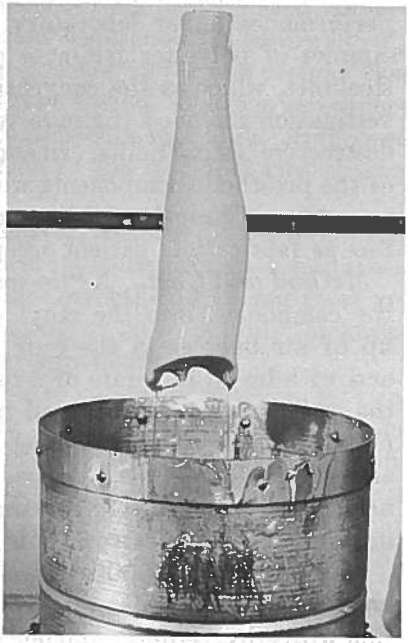

FIGURE 25. *Test component dipped in ethyl cellulose; note the drying rack in background.*

B. Patient Lifts and Standing Bars

Three patient-lift aids and one standing parallel-bar apparatus were recently received for evaluation. The various patient-lift devices were designed to transfer patients from beds to chairs, into and out of automobiles, or to permit the patient to rise with little or no assistance in order to stand for extended periods. Each device was powered differently: one was mechanically operated; one was hydraulically operated; and the third was electrically operated. The portable standing apparatus, which provided support for knee and hip extension, was designed to permit a patient to stand for prolonged periods.

Evaluation reports were submitted on each of the above-mentioned orthopedic aids to the developers. The Bioengineering Laboratory is currently setting up performance standards for future evaluation of all such orthopedic aids.

V. MATERIALS INVESTIGATION

A. Ethyl Cellulose External Finishing

A new technique for plastic-finishing wood or plastic prostheses has been developed and preliminary evaluations completed. Components externally finished with ethyl cellulose plastic and reinforced

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with Helanca yard stockinet were drip-dried before testing. Their performance under laboratory testing was compared with that of samples of polyester laminate reinforced with two layers of nylon stockinet, which is the conventional fabricated external finish. Investigation included comparative weights, moisture absorption, and destructive testing limits. In addition to the laboratory tests, a number of the prosthetic components with test external finishes were worn by amputees. Clinical experience data were then obtained on such factors as fabrication, patient acceptance, and durability.

Method and Cost. A "tie-coat" sealer is necessary before dipping the component into the ethyl cellulose. To minimize the stirring up of air bubbles in the tank, the component should be immersed and withdrawn at a rate of 5 to 9 inches per minute. After drying, the sealed socket is immersed into pigmented ethyl cellulose plastic (Fig. 25); A Helanca stockinet is then stretched over the component, and the reinforced socket dipped again into the ethyl cellulose. Alternate dipping and drying continues at the recommended rate until the desired thickness of 50 to 60 mils is achieved.

On this experimental basis, costs (including 15% for waste) were about $3.45 to dip coat the average foot, shank, and socket combination with ethyl cellulose and one Helanca stockinet.

Fabrication and Testing Results. Seventeen test specimens were prepared with six types of external finishes over tapered wood cylinders. Weights of the wood cylinders before and after finishing were determined as well as the moisture absorption (moisture meter, model DC–2) and vertical breaking load (Dillon tensile tester). In the destructive tests, a vertical load was applied to a tapered mandrel, which was inserted into a test specimen until failure. The wood failed in exactly the same section of the unit, i.e., a quarter break parallel with the grain of the wood. This was true even for cylinders with small knots and irregularities.

In this series of tests, specimens reinforced with one layer of Helanca stockinet showed the highest compressive strength characteristic, and those without any stockinet reinforcement, the second highest. Increasing the number of layers of stockinet did not increase the tensile strength. Instead, it was noted that increasing the number of stockinet layers decreased the maximum breaking load at which test samples failed. The compressive strength of test samples of ethyl cellulose reinforced with one layer of Helanca stockinet and without reinforcement was greater than that of the conventional polyester nylon laminate specimen.

Moisture absorption variation of all test specimens was small enough to be considered negligible for weight change calculations.

The results of the clinical evaluation of the test-finished components conducted over a one-year period were good. There was only one failure out of more than 80 components of various designs that were finished and worn throughout that period. Patient acceptance of the dip-finished ethyl cellulose components was high. Improved cosmesis resulted from the elimination of the stockinet ribs characteristic of the polyester laminate.

The ethyl cellulose dip method, however, requires the use of a metal or plastic container large enough to prevent spillage of the liquid when a component is totally immersed. Since several dips (about five) are required to build up the finished coating of 50 to 60 mils thickness, drying racks are necessary to hold the components.

Since the process is readily adaptable to volume production, fabrication cost savings could be significant. One of the advantages of the dip method is that semiskilled labor can be employed to produce consistent results in contrast to the conventional plastic laminate process, which requires the skill and time of a highly experienced craftsman to produce a properly reinforced and cosmetically acceptable prosthesis.

An illustrated manual on the ethyl cellulose dipping procedure is currently being prepared.