SEMIANNUAL REPORT OF THE VETERANS ADMINISTRATION PROSTHETICS CENTER (JANUARY–JUNE 1964)

INTRODUCTION

In the VA Prosthetics Center's semiannual reports, only the salient features of our research, development, and evaluation progress reports are usually summarized. At times, however, when progress in some of these studies indicates a need for detailed treatment, a separate article may appear in the *Bulletin,* as for example, the paper on shoe modifications appearing elsewhere in this issue (1).

Nevertheless, certain topics included in the semiannual report are presented in detail in the hope that close reader scrutiny and, perhaps, trial applications of techniques will stimulate feedback and improvement. One such topic, the specifications on lift aids, is presented in this report.

Also included in the current report are data from the Center's operational programs, which may be of significance for the research-and-development efforts of other groups. Through some of its clinical programs, particularly those nation-wide in scope, the VA Prosthetics Center is able to highlight major difficulties and problems relating to patient service and costs and to focus research and development efforts on improvement.

Since it will be our policy to describe only significant progress in research, development, and operations, a specific item or subject described in one semiannual report may not be followed up until two or three issues later.

This semiannual report was produced by teams of Center personnel. Credit for individual contributions is due Henry Gardner, William McIlmurray, Thomas Pirrello, Jr., Frank Schenck, Peter Thauwald, and Frank Witteck. Assembly of the various contributions and correlation of the individual reports were primarily made under the leadership of Dr. Edward Peizer assisted by Donald Wright and Mrs. Anastasia Keane.

ANTHONY STAROS,
Chief, VA Prosthetics Center,
New York, N.Y.

I. LOWER-EXTREMITY PROSTHETICS

A. Casting Techniques

1. An Instrument for Measuring the Below-knee Stump and Evaluating Socket Fitting. Recent progress in improving below-knee socket fabrica-

tion methods and fitting techniques have created a need for a more precise method of measuring below-knee stumps and evaluating the adequacy of socket fit. For these purposes an instrument (Fig. 1) has been developed to facilitate more accurate measurement of the two most important dimensions in below-knee stumps: (1) Anterior-posterior (A–P), the knee diameter from the patellar tendon to the popliteal area and (2) Medial-lateral (M–L), the knee diameter from the medial to the lateral epicondyle. Improving the precision of these measurements will enable the prosthetist to design below-knee sockets more accurately and to check the adequacy of socket fit more reliably.

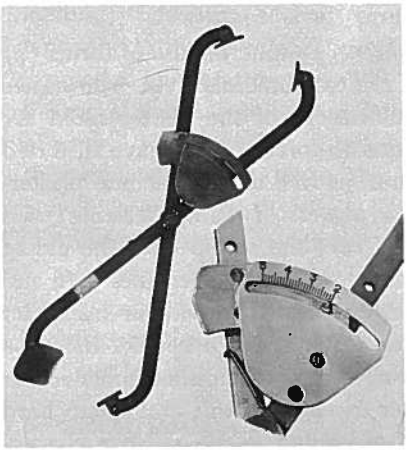

FIGURE 1. Clinical instrument for measuring anterioposterior and mediolateral dimensions of BK stumps. The platens at the top of the figure are used for mediolateral measurements while those at the bottom are used for anterioposterior measurements.

The structure of the knee includes relatively hard and soft tissues, making it difficult to replicate or standardize conventional measurement techniques. A spring-loaded caliper-type device was designed to compensate for variations in stump shape and tissue resiliency and to eliminate the variability due to the forces applied by prosthetists. The spring provides a constant torque of approximately 3 in.-lb., corresponding to the compression that occurs during weight bearing in a properly fitted below-knee socket. To obtain an accurate A–P knee measurement, it was necessary to design the caliper surfaces of unequal size and contour in order to distribute the pressure more evenly on surfaces of varying resiliency, i.e., the relatively hard, resistant patellar tendon and the soft compressible popliteal area. With a constant force distributed over the unequal areas of the caliper surfaces, relatively high pressures are applied to the hard patellar tendon and low pressures are applied to the soft popliteal region, thus tending to equalize the deformation of the tissues.

Because the medial and lateral epicondyles are of similar compressibility, the knee-width measurement is readily achieved with two flat circular discs,

each approximately 1 sq. in. in area at the caliper-measuring surfaces. All measurements are read on a scale graduated in ¼ in. increments. The best results in terms of satisfactory comfort and socket stability have been achieved when the knee measurements were taken over the stump sock or other covering. Clinical checks of socket versus stump dimensions should also be made with the stump sock worn.

Use of the calipers has reduced the number of adjustments required in casting below-knee stumps in the VAPC casting stand. The stump diameter (A–P) obtained with a controlled amount of compression is used to set the distance between the anterior and posterior casting forms.

Bids for a number of calipers for VA use were recently invited and a contract has been awarded. We expect that the caliper will be available very shortly.

2. *Direct Forming of Below-knee Sockets.* Despite the use of forms (2) to shape the stump and to maintain critical dimensions in below-knee plaster-of-Paris casting techniques, modifications of the plaster mold are often required to correct irregularities due to forces which have distorted the wrap cast. To reduce these problems, as well as to eliminate plaster-of-Paris from socket-forming methods, a process of stump "casting" was developed, using a cone which is molded over the stump with the aid of the casting forms. The cone, a thermoplastic acrylic, becomes soft and pliable when heated. It is preshaped and presized to provide even contact over the stump surfaces with minimal distortion. Once heated, the cone conforms to the stump contour, producing a smooth, finished inner-socket surface. The cone contracts to a greater extent over soft areas than hard areas, and thus tends to "correct" itself according to the resistive characteristics of the underlying stump tissues.

Acrylic cones approximately ¼ in. in thickness are now fabricated by laboratory (nonproduction) methods. Five sizes are being used with proximal diameters ranging from 3 to 4 in. in increments of ¼ in. The cone, softened in an oven, deforms readily when pulled over the stump. When properly finished, the acrylic cone becomes the socket.

Before forming the acrylic cone over the patient's stump, A–P stump dimensions at the insertion of the patellar tendon are recorded, and the casting forms are adjusted to these dimensions. The patient is positioned with his stump properly placed between the casting forms. The attitude of the stump relative to the remainder of the body is adjusted and "fixed" in position. The placement of the sound foot is recorded on the floor to facilitate immediate return to the patient's original position while forming the cone (socket).

After checking the dimensions and adjustments on the casting device, the patient's stump is removed from between the casting forms and prepared for casting. The stump is insulated against the heat of the cone by means of two layers of tailored stockinet drawn up to midthigh level.

The friction between the cone and outer stockinet is reduced by lubricating the outer stockinet surface with petrolatum. The petrolatum also acts as insulation between the cone and the stump.

The A–P distance between the two forms is increased to accommodate the double thickness of the stump stockinet and the cone wall. An undersized cone, selected to assure stretching over the stump and thus close contact with all stump contours, is placed in an oven and heated to a temperature of 250° F. for 2 minutes. When soft, the cone is pulled over the stump and knee, and the patient is repositioned in the casting fixture. The patient stands relaxed in the orginal position while the socket material hardens, (approximately 5 minutes).

It is relatively simple to achieve consistently accurate fittings by this method; however, several problems remain. Thermoplastics have a tendency to return to their original shape (in this instance a cone) when exposed to temperatures over 100° F. Hence, a problem arises of stabilizing the socket shape and size to maintain the fitting accuracy under all conditions of wear. The high heat of polymerization with polyester resins deforms the thermoplastic socket when the finishing laminations are applied. The acrylic socket must, therefore, be reinforced before lamination.

Another material having the same basic physical characteristics as acrylic is also being investigated. The new material seems to be considerably stronger and to have less tendency to crack under impact. Efforts to develop a more practical and economical means of fabricating below-knee sockets by direct forming in plastic will continue.

B. Socket Designs

Work to improve below-knee socket design has received a great amount of emphasis. Under consideration at this time are studies of the relative value of hard and soft socket designs and the problem of socket ventilation. The conventional patellar tendon-bearing (PTB) closed-end socket is designed to encompass the entire stump, reducing the pressures upon the tissues; however, the conventional inner-socket liner, formed of a resilient material, permits the stump to slide down into the socket, increasing the vertical shear forces on the stump tissues and also increasing the load concentrated at the distal end. The use of hard-walled sockets with soft ends seems to reduce slipping and uncomfortable wedging of the stump into the socket, making it easier for the patient to tolerate the forces over the end of the stump. Although the hard socket requires an extremely accurate fit, the stump remains stable in the socket for longer periods of wear than with the soft socket liner.

Another variation to be studied is the use of hard, yet very flexible, polyester liners formed to mate with the stump. Sponge rubber is placed on the *outside* of the liner and the final shank formed over the sponge

rubber. This procedure may overcome problems of stump displacement due to the setting of the rubber and perspiration absorption.

An alternative method of ventilating sockets was explored. Socket fenestration, a less costly method of ventilation, was tested by cutting out sections of the socket over selected areas where socket-supporting loads might be reasonably removed. Free passage of air into the socket chamber, however, reduced the negative (suction) pressures, and an increase of "piston action" was noted even though the socket was suspended by a thigh cuff.

A third study has been organized to investigate the use of a pre-ventilated material, a "poromeric" plastic (Corfam), which is prefabricated in sheet form. A technique for molding the material over a cast to form a socket liner is currently being developed.

The material appears to allow the moisture collected in the socket to be released to the outside air. It has been used with reasonable success in the fabrication of shoes here at the Center, and highly encouraging reactions have been expressed by a below-knee amputee who has been wearing a socket lined with this material. The next step is to test Corfam as a liner in a fenestrated socket.

C. Components Evaluation

1. Torque Absorber. The torque absorber for below-knee prostheses, described in BPR 10–1 (3), has been modified to allow interchangeability between pylon and wood prostheses (Fig. 2). We believe that there is a correlation between the torque absorption requirements of the device and the level of amputation; also probably related are length of stump, weight of patient, length of stride, and amount of activity.

While generally satisfactory, the current design has some serious shortcomings. The device functions adequately during the heel contact to midstance phase when the torques developed are relatively low and torque absorption is not critical. From midstance to toe-off, however, excessive external rotation was noted, and subjects complained of the lack of pushoff. Accordingly, the device has been modified. Bumpers of harder rubber have been installed in diagonally opposing quadrants to increase resistance to external rotation.

In general, a comparison of the torques required to rotate the device before and after cycling tests (50,000 cycles at 30 c.p.m.) show that the gum-rubber bumper material (Durometer A 30) has little hysteresis; no permanent changes in elastic characteristics took place. Further testing and clinical evaluation are in progress.

2. Mechanical Knee Frictional Devices. Based on the results of evaluation studies by New York University, the Navy Prosthetics Research Laboratory, Northwestern University, and the VA Prosthetics Center, the Northwestern University Intermittent Friction Device (Model B) for above-knee prostheses has been redesigned by joint efforts of the U.S. Navy, the U.S.

FIGURE 2. The experimental torque absorber is mounted as close to the stump as possible in a pylon BK prosthesis. A cosmetic cover is used to fair the socket to the foot.

Manufacturing Company, and the VA Prosthetics Center. A bench model incorporating the modifications was fabricated and checked out to insure the compatability of the new features with the overall system and to provide detailed production engineering guidance for the manufacturer. The design deficiencies revealed by the earlier studies included:

a. Bending and/or fracture of the anchor bar.

b. Elongation of the leather washers with eventual loss of friction.

c. Deterioration of the rubber bumpers.

d. High and low areas in the resistance due to irregular spacing of the separating plates.

e. Objectionable noise.

f. Cracking of the posterior aspect of the shank. The probable cause of failure was the placement of the anchor bar screw hole too close to the posterior proximal brim of the shank.

In order to overcome these deficiencies, to reduce manufacturing cost, to improve ease of friction adjustment, and to provide a wider range of friction control, the following modifications were suggested:

a. Redesign and fabrication of the anchor bar. Two different, yet func-

760-425 O—65——9

tionally identical, anchor-bar designs were submitted to suggest several methods of fabrication with a view toward reducing production costs.

b. Replacement of the inadequate leather washers by Celastic friction washers, Nylatron, or a brake material suggested by the Navy.

c. Replacement of the lower machine screw with a partially threaded screw to prevent discs from riding on the screw threads. Enlarging the clearance holes in the disc platen and providing extremely soft rubber spacers allowed the required parallel displacement of the discs.

d. Use of a Belleville spring washer in place of the former multiple spring arrangement of the friction adjustment wheel. Two other spring washers were suggested: a finger spring washer and a wave spring washer. Any of these components will provide more adequate frictional adjustments than the multiple spring arrangement.

e. Fabrication of a friction adjustment wheel with wider adjustment prongs and more rounded corners. This makes it easier for the prosthetist to adjust the friction wheel by hand.

f. A fifth disc was added to the four existing friction discs to provide another, larger angular step in the friction-vs.-angle characteristic; more variability in friction patterns is now available using combinations of any three discs.

3. *Wagner Above-knee Assemblies.* Four knee-shank assemblies, manufactured in West Germany and distributed in the United States by the Wagner Orthopedic Supply Company, Salt Lake City, Utah were evaluated.* The four assemblies (Models 98, 200, 204, and 205) were similar in several respects, each featuring:

1. A closed-front knee;
2. Preshaping in lefts and rights;
3. Overall length of complete assembly approximately 18 inches;
4. Controls contained within the unit;
5. Units sized according to calf measurements, with size range from 11¾ to 15½ in. in ¾-in. increments;
6. Facility for use with most foot-ankle combinations currently available;
7. Ability to accommodate long above-knee stumps that come within 1½ in. of knee center;
8. Facility for use with alignment devices such as an adjustable leg, VAPC coupling, and alignment transfer jig. Special features of the Wagner knee are described in Table 1.

*Another company, R & G Orthopedic Appliances Co., Washington, D.C., distributes similar units called Streifeneder Knees. Streifeneder is the name of the German manufacturer of most of the Wagner mechanisms.

TABLE 1.—*Special Features of Wagner Knees*

Model No.	Type of Mechanical Friction	Weight-bearing Friction	Extension Bias
98 (Single-axis knee joint).	Clamp-type split plastic bushing. Constant-friction resistance is provided by increasing the clamping force on the knee axis with a screw adjustment.	No.....	None; external type can be installed.
200 (Floating-axis knee joint).	Pivoted lever with plastic bushing adjusted by thumb-operated disc.	Yes.....	None; external type can be installed.
204 (Floating-axis knee joint).	Pivoted lever with plastic bushing adjusted by thumb-operated disc.	Yes.....	External.
205 (Floating-axis knee joint).	Pivoted lever with plastic bushing adjusted by thumb-operated disc.	Yes.....	Internal.

Model 98 is a single-axis knee joint supplied in either poplar or balsa wood. The friction mechanism provides constant mechanical friction by means of a split plastic bearing which is adjustable for the required resistance. A rubber bumper located on the knee engages the posterior aspect of the shank for hyperextension control. If an extension aid is required, it must be installed externally.

Models 200 and 204, otherwise identical in design, differ only in that Model 200 has no extension aid whereas Model 204 is equipped with an external extension aid. Both units have "floating-knee" axes. When weight is applied, the knee bolt moves upward in relation to the socket to contact a rubber bumper. This excursion of the knee bolt permits the knee to engage a mating surface in the shank, producing a weight-bearing friction brake. Knee swing is controlled by means of a constant mechanical friction device adjusted by a thumb-operated disc located behind the knee. The disc actuates a pivoted lever ending in a plastic bushing which bears against the knee bolt, producing the desired degree of friction. A metal lug, located in the shank, slides in a slot in the knee and contacts a stop to provide hyperextension control.

Model 205 is basically of the same design as Model 200, but features an adjustable internal extension bias control. This device has a plastic, stick-

type yoke that compresses a spring located in the shank. The compressed spring aids extension of the knee.

Evaluation of these devices was based upon mechanical analysis of each of the four models and upon clinical application experience in fitting two subjects with Models 98 and 204, representing all the varying features: with and without floating knee axis and with and without extension aid.

Clinical experience indicated the general adequacy of all samples. Some minor repairs were necessary to eliminate noise in the knee mechanisms, but such problems are common to this type of friction design. Subjects' reactions were generally favorable as regards the functional features of the units. These units required no special fixtures or tools. The preshaped components minimize shaping requirements. The closed-front knee feature improves cosmesis and reduces wear on trousers. Unit cost seems reasonable compared to similar assemblies offered by other manufacturers. Maintenance requirements were minimal. In other respects, the units meet the structural and functional requirements currently in use for *mechanical friction* above-knee assemblies.

4. *Plastic Knee-shank Setup.* An initial evaluation of the Otto Bock 3P4 "lock knee" and plastic foam knee-shank setup indicated that the physical properties of the foam were quite satisfactory for prosthetic application. Studies undertaken to compare the material with conventional poplar generally corroborated the original findings. The impact strength of the plastic foam, however, was quite low. Comparative data on the plastic foam and poplar wood are given in Table 2.

5. *Cosmetic Leg Cover.* The earliest artificial legs were probably pylon-type devices, but modern esthetic concepts and psychological needs led to the development of improved "natural-looking" devices. Pylons are not widely used because of the lack of a cosmetically acceptable design; nevertheless, the pylon or pylon-type design has been used in the development of temporary prostheses and in hydraulically controlled devices, making it increasingly necessary to provide an acceptable cosmetic cover that can be easily removed for adjustment of the mechanical components. Present cosmetic leg covers also lack adequate durability.

The major difficulties encountered in solving the esthetic problem are related to materials. It has been a formidable task to find materials with the desired resilience, weight, and durability.

The VA Prosthetics Center is currently evaluating a flexible, thermoplastic material of medium density and high tear resistance. The material is easily pigmented and color-fast. Several shank covers of varying thicknesses have been fitted over pylon-type hydraulic devices for testing purposes. Other materials of similar characteristics are being investigated in an effort to develop an adequate cover (Fig. 3) for (1) a hydraulic device now in use, and (2) pylon designs generally.

TABLE 2.—*Physical Properties of 3P4 Plastic Foam and Poplar Wood*

Test	3P4 Plastic Foam	Poplar Wood
Impact Strength, ASTM DZ 56–56.	1.9 in.-lb.....	2.5 in.-lb. (perpendicular to grain). 34.7 in.-lb. (parallel with grain).
Compressive Strength, ASTM D695–61T.	1,590 p.s.i....	160 p.s.i. (perpendicular to grain). 1,050 p.s.i. (parallel with grain).
Water Absorption, ASTM D570–59aT.	30.3%.....	28% (fully saturated).
Breaking Load (Destruction by Carver Hydraulic Press).	1,140 lb.....	2,425 lb.
Density.....	2.3 lb./cu. ft..	25 lb./cu. ft.

6. Immediate Postoperative Application of Prosthesis. At the Sixth International Prosthetics Course of the International Society for the Rehabilitation of the Disabled (I.S.R.D.) held in Denmark in 1963, Dr. Marian A. Weiss, Professor and Clinic Chairman of Rehabilitation and Reconstruction Surgery at the Medical Academy "Konstancin" in Warsaw, Poland, described his experiences in applying temporary prostheses immediately after amputation with the patient still under anesthesia. The potential advantages of this procedure lie in the possibility of preventing edema, of reducing the nonambulatory period between amputation and fitting and, perhaps, in reducing the number of sockets required before the stump volume and shape become stabilized. Recognizing the possible medical, psychological, physiological, and economic gains, several groups* in California, Washington, and New York are applying this technique on a research basis.

As first described by Dr. Weiss, a solid, one-piece, total-contact plaster-of-Paris cast was placed upon the stump immediately following surgery before the onset of edema. The patient was encouraged to stand and, if possible, to walk briefly on the day following surgery.

A similar technique has been tried in Boras, Sweden, by Dr. F. Stahl and D. L. Mozer who used a drain in the stump wound and a bivalved cast that permitted size adjustment for stump swelling (4). It was necessary to provide accommodation for swelling since the cast was not applied until the second day when the drain was removed.

*Prosthetics Research Group, Seattle, Wash.; Navy Prosthetics Research Laboratory, Oakland, Calif.; University of California, San Francisco, Calif.; VA Prosthetics Center in conjunction with the Hospital for Joint Diseases, New York, N.Y.

Both the Polish and the Swedish techniques, however, employ the patellar tendon weight-bearing concepts of fitting and associated socket shape. According to the Swedish article, after immediate postoperative prosthesis fitting, circulation and healing both progress as in the amputee with conventional postoperative management.

To date the United States investigators have also indicated that postoperative pain is not increased by use of an immediate postoperative prosthesis, that the time interval between amputation and ambulation is reduced, and that the time spent on one leg is significantly lessened. Important psychologically as well as physiologically is the fact that the patient is kept busy becoming familiar with and learning to use his prosthesis. A temporary BK prosthesis developed by the VAPC for immediate postoperative fitting is shown in Figure 4.

Figures 5 and 6 show two phases in the procedure as employed on one case at the Hospital for Joint Diseases, New York. The technique being used by the New York group follows as closely as possible the principles first outlined by Dr. Weiss.

It is still too early, however, for conclusive statements on the value of the Weiss procedure. Careful planning and organized research are required to obtain valid results. To provide guidance for the various teams working in the field, a selected panel of physicians and others organized

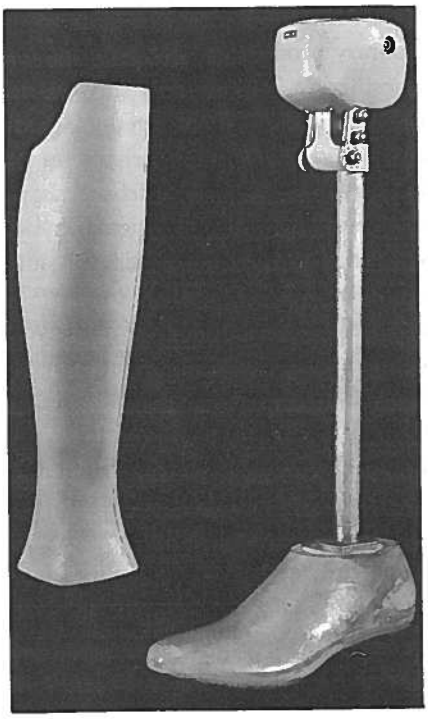

FIGURE 3. A new vinyl cover to be glued to the SACH foot can be used for temporary BK and AK prostheses.

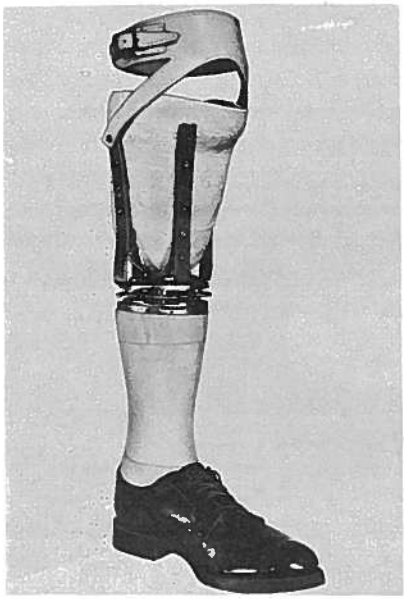

FIGURE 4. Prosthesis for immediate postsurgical application. Metal straps are incorporated into the contoured primary compression cast by an additional plaster-of-Paris wrap or wood screws. Use of an alignment coupling not only permits detachment of the socket and lower leg for bed rest, but allows alignment changes when necessary. Partially shaped lower leg segment and foot provide a temporary adjustable prosthesis with reasonable cosmesis. Lower leg segment of balsa wood conserves weight.

by the Committee on Prosthetics Research and Development and headed by Dr. V. Inman of the University of California at Berkeley have developed a checklist of evaluation criteria. These criteria are now being applied by the various groups.

D. Fluid-controlled Knee Mechanisms

The effects of several fluid-controlled knee mechanisms on the locomotion patterns of above-knee amputees are currently being studied (5). To date, five above-knee subjects have each been fitted with two or more different devices. At the completion of this study, six subjects will have been fitted with four to six different knee mechanisms.

Of particular interest in this study is the effectiveness of these devices in controlling swing-phase phenomena. The prosthetist attempts to control flexion-extension by selecting an appropriate resistance setting which his observation tells him provides the most symmetrical gait pattern and which the patient tells him is not unreasonably uncomfortable. Since all of these devices are designed to control flexion (heel rise) and extension (terminal impact), an interesting question arises as to whether different devices provide essentially similar swing-phase patterns when adjusted for a particular subject.

A preliminary analysis of the data collected on one subject was undertaken in order to learn why specific gait characteristics are displayed and how they are related to the design features of each mechanism. In addi-

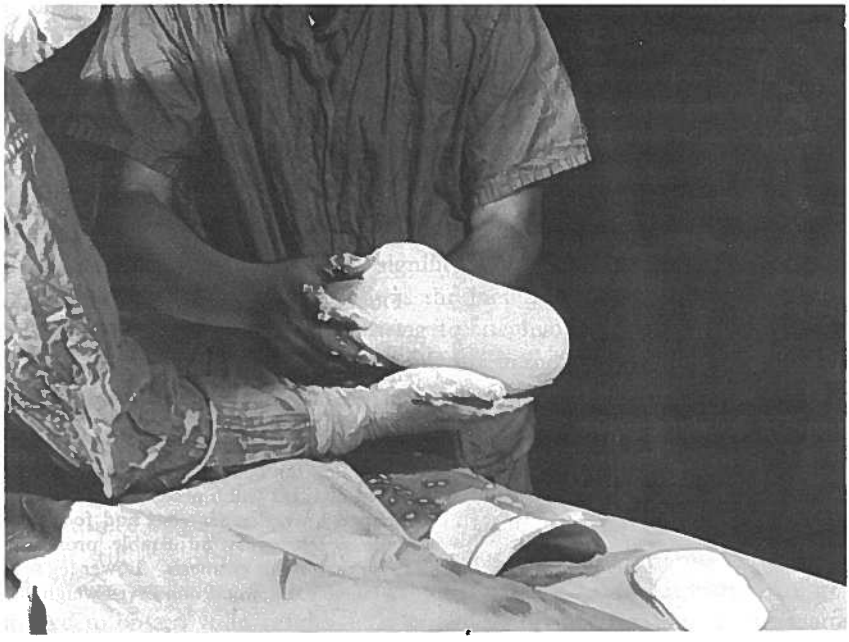

FIGURE 5. Application of an elastic plaster-of-Paris bandage wrap over the BK stump a few minutes after surgery has been completed. Note the extension of the plaster-of-Paris to cover the patella completely. The forms lying on the table will be used to shape the plaster-of-Paris before it sets in order to give the socket some weight-bearing contour.

tion, an attempt was made to determine the most useful evaluation criteria: Is performance with different knee mechanisms better evaluated in terms of "normal" nonamputee criteria or in terms of symmetry, that is, in relation to performance of the amputee's normal leg?

The subject of this analysis was a 39-year-old unilateral above-knee amputee in good health who is employed as a post-office clerk. A supracondylar amputation of his left leg in 1945 produced a 12-in. cylindrical stump with an almost normal range of motion. For 3 years prior to this study, he had worn conventional prostheses. The most recent prosthesis included a suction socket, single-axis, mechanical-friction knee with an extension-bias strap.

The devices discussed in this report are all designed to control swing phase without directly affecting stance phase. Motion about the knee axis is controlled hydraulically; in the Mauch and Dupaco knees, motion is regulated by the application of hydraulic resistance in a program designed to vary the number and size of open orifices during the swing cyle; the Regnell system employs a single orifice.

FIGURE 6. Application of the temporary limb to the plaster-of-Paris socket. Note the socket contouring particularly in the region of the patellar tendon. In this particular prosthesis, a stocking is being used to enhance the cosmesis of the temporary limb.

The standards used for comparing the effectiveness of the three mechanisms were (1) The values for vertical displacement of the body CG and knee angular displacements for normal, nonamputees as originally published by the University of California, Berkeley (6) and corroborated by Murray *et al.* (7) on a larger sample and (2) The values for similar phenomena determined for the unaffected leg of the subject. That these devices are designed to control swing phase means that they should:

1. Effectively control knee function so as not to cause undesirable compensatory adaptations in the motion of the total body or the segments;
2. Control knee *flexion* to limit heel rise after toe-off to a vertical displacement which enhances the appearance and temporal symmetry of the gait;
3. Control *extension* of the knee to provide optimal velocities during swing with properly programmed accelerations and decelerations to make a smooth transition from swing phase to stance phase.

This comparison of knee mechanisms, therefore, includes an analysis of:

1. Maximum vertical body displacement;
2. Gait symmetry as indicated by swing-time ratios;
3. Maximum heel rise;
4. Angular velocity about the knee;

5. Terminal impact as indicated by deceleration at the end of swing phase.

A comparison of the subject's gait while wearing each of the three devices indicated that walking with the Mauch leg produced the least vertical body displacement and temporal asymmetry.

As shown in Table 3, walking with the Mauch leg at cadences of 100–102 steps per minute (s.p.m.), maximum vertical displacements of both hips were nearly equal and corresponded with values given for normal nonamputees at a slightly higher cadence (112 s.p.m.). Although hip displacement on the unamputated side with the Regnell leg was also of normal magnitude, the displacement on the prosthetic side was considerably higher in magnitude. With the Dupaco prosthesis, hip displacement on both the prosthetic and sound sides was considerably higher than normal.

At fast cadences greater hip displacements were clearly evident in all cases. With the Mauch unit, increased asymmetry was noted, with displacement on the prosthetic side substantially higher. In walking with the Mauch leg, swing-phase times for both the prosthesis and the contralateral leg were more nearly equal than with the other combinations (Table 4). In contrast, with the Mauch knee, the stride length (from heel contact to heel contact on the same side) was 4 in. shorter on the prosthetic side than that taken on the unamputated side. While wearing the Dupaco or the Regnell leg, however, the subject walked with nearly equal stride lengths.

TABLE 3.—*Maximum Vertical Hip Displacements at Two Cadences*

Type of Leg	Vertical Hip Displacement (inches)			
	Moderate Cadence (100–102 s.p.m.)		Fast Cadence (120–130 s.p.m.)	
	Unaffected Side	Prosthetic Side	Unaffected Side	Prosthetic Side
Dupaco	2.2	2.8	.....	3.0
Regnell	1.7	2.5	2.6	3.2
Mauch "B"	1.8	1.7	2.8	3.5
Normal Nonamputee Subjects at 112 s.p.m	1.9	.....	.....	.....

It is possible that these three phenomena, vertical displacement of the hip, swing-phase time ratio, and stride length, are an interrelated product of the higher *extension resistances* generally characteristic of the Mauch unit. At approximately equal energy inputs in walking with all three units, the greater

resistance to extension of the knee in the Mauch device produces a shorter stride length, a smaller vertical hip displacement, and hence a swing-phase time that corresponds closely to that of the sound side.

TABLE 4.—*Ratios of Swing-phase Time on Prosthetic Leg to Swing-phase Time on Normal Leg at Two Cadences*

Type of Leg	Cadence	
	Moderate (100–102 s.p.m.)	Fast (120–130 s.p.m.)
Dupaco	1.2	1.4
Regnell	1.2	1.4
Mauch "B"	1.0	1.1

The effect of *resistance to flexion* can be seen in the magnitude of the maximum heel rise just after toe-off. As shown in Table 5, at moderate cadences (100–102 s.p.m.) the heel of the prosthesis did not rise to the same height as the heel of the intact limb regardless of which device was worn; however, with the Dupaco and Regnell legs, heel rise was more symmetrical than with the Mauch leg.

Walking at faster cadences substantially increased heel rise on the prosthetic side. The single-orifice system of the Regnell unit did not seem as effective as the Mauch unit in limiting heel rise; however, this apparent advantage seemed to be offset by the increased heel rise on the unamputated side.

Knee angular displacement curves for all three legs showed a remarkable corrspondence in magnitude and phase (Fig. 7). Maximum knee flexion angles were more symmetrical when the subject wore the Dupaco and Regnell legs than with the Mauch leg, but knee flexion on the Mauch prosthesis was in somewhat closer correspondence with the normal, non-amputee population (Table 6).

Despite differences in resistance to flexion and extension among the three systems, angular velocity about the knee was quite similar in all cases (Fig. 8), indicating perhaps a general optimum for the subject.

As shown by curves of angular acceleration during the last 20 per cent of the walking cycle (Fig. 9 and 10), terminal deceleration at normal cadences (103–105 s.p.m.) is more effectively accomplished with the Mauch and Dupaco devices than with the Regnell unit. The higher extension resistance of the Mauch system produces a smooth deceleration curve corresponding closely to the slope and shape of that obtained for a normal knee, although maximum deceleration occurs earlier; in other words, the shank

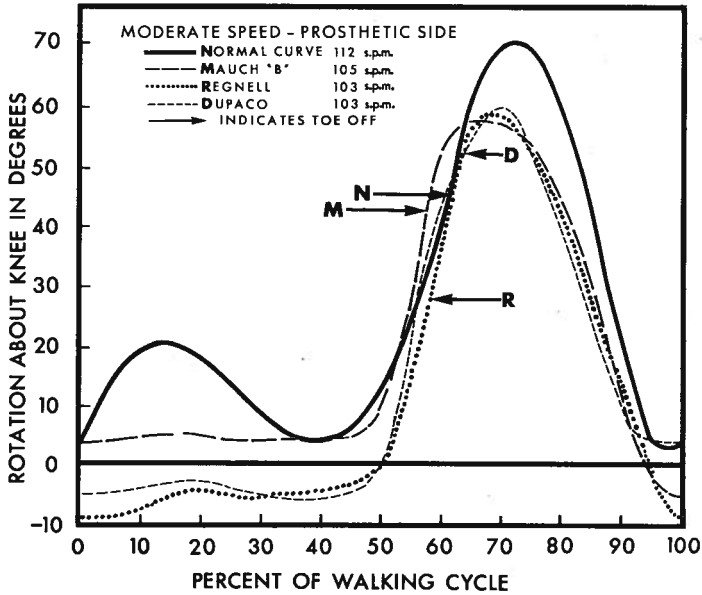

FIGURE 7. Angle of knee flexion and extension plotted against percentage of walking-cycle time.

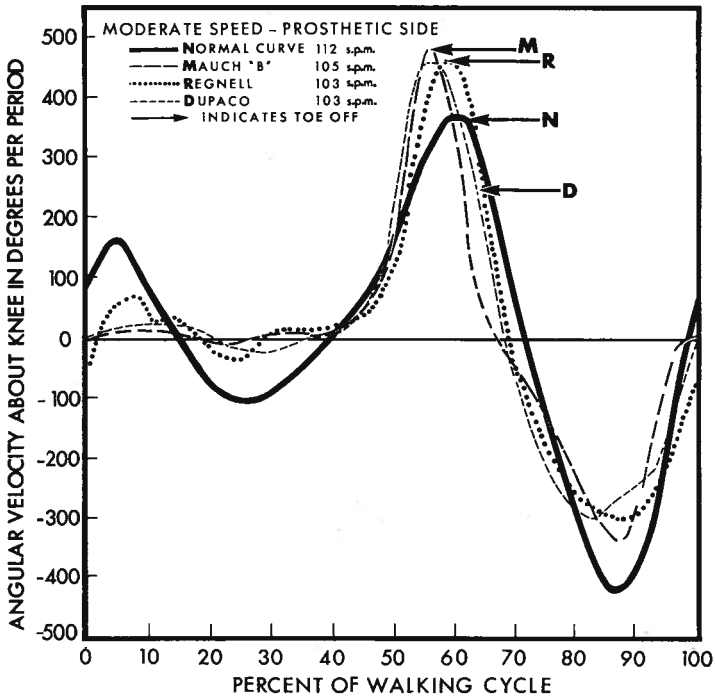

FIGURE 8. Angular velocity of the knee plotted against percentage of walking-cycle time.

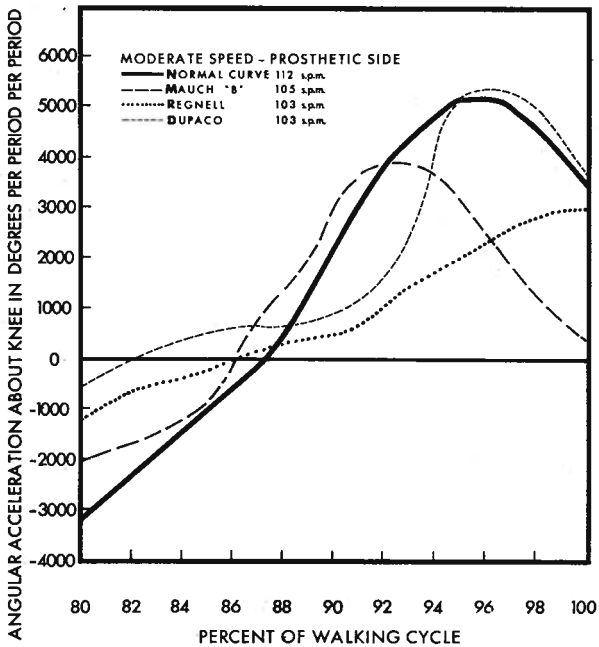

FIGURE 9. Angular acceleration of the knee plotted against percentage of walking-cycle time (moderate speed).

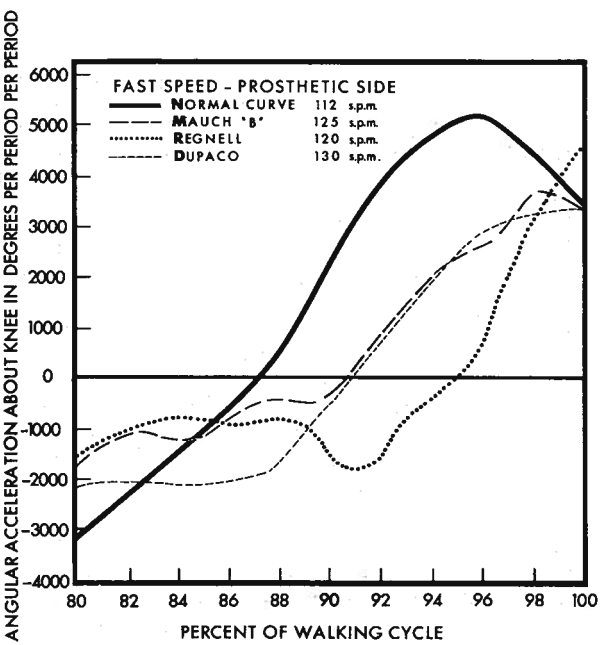

FIGURE 10. Angular acceleration of the knee plotted against percentage of walking-cycle time (fast speed).

loses velocity at a rate similar to a normal leg. Apparently not quite as effective, the Dupaco begins deceleration later in the cycle and consequently attains a higher terminal deceleration magnitude; nevertheless, the peak deceleration with the Dupaco leg corresponds with that of a normal knee. The Regnell leg ends the swing-phase cycle at peak acceleration, indicating perhaps that the single orifice is not capable of the same degree of control as a programmed multi-orifice system.

TABLE 5.—*Comparison of Heel Rise of Prosthetic vs. Normal Leg at Two Cadences*

Type of Leg	Maximum Heel Rise in Inches			
	Moderate Cadence (100–102 s.p.m.)		Fast Cadence (120–130 s.p.m.)	
	Unaffected Side	Prosthetic Side	Unaffected Side	Prosthetic Side
Dupaco	10.2	9.8	.....	11.2
Regnell	10.5	9.9	10.5	12.5
Mauch "B"	11.5	9.0	12.5	10.5
Normal Nonamputee Subjects at 112 s.p.m.[a]	11.4	.....	.....	.....

[a] *Functional and Psychological Suitability of an Experimental Hydraulic Prosthesis for A/K Amputees*, New York University, 1953.

TABLE 6.—*Comparison of Knee Flexion of Prosthetic vs. Normal Leg at Two Cadences*

Type of Leg	Maximum Knee Flexion in Degrees			
	Moderate Cadence (100–102 s.p.m.)		Fast Cadence (120–130 s.p.m.)	
	Unaffected Side	Prosthetic Side	Unaffected Side	Prosthetic Side
Dupaco	59	60	.....	68
Regnell	62	58	64	64
Mauch "B"	70	58	71	69
Normal Nonamputee Subjects at 112 s.p.m	70	.....	.....	.....

At faster cadences (120–130 s.p.m., Fig. 10) both the Mauch and Dupaco legs exhibited almost identical terminal deceleration patterns, but at the time of heel contact all three legs are very close to peak acceleration.

Additional data are being collected on amputee performance with other fluid-controlled knee mechanisms, and other techniques of data interpretation are currently being explored.

E. System for Assembling and Finishing Above-knee Prostheses Off-station

A great deal of time is spent by skilled prosthetists in finishing and assembling above-knee limbs. In an effort to relieve the skilled fitter of many routine tasks so that he may devote more time to the critical aspects of alignment and socket fitting, the VA Prosthetics Center has developed a system to send precise fit and alignment data to commercial facilities for assembling and finishing above-knee legs.

This method of producing limbs lends itself to centralized fabrication and would be of importance in times of national emergency since two or three factory-type fabrication-assembly plants, strategically located throughout the country could be called upon by outlying VA or commercial facilities to furnish prostheses for local amputee populations.

In this procedure, the required fit and alignment data are obtained at the VA Prosthetics Center by means of the Hosmer AK adjustable leg which has been modified to provide finer measurement increments. Of vital importance was the incorporation of a device to permit simultaneous recording of knee flexion and toe-out angles (Fig. 11).

The data, recorded on forms specially designed for the purpose, are sent to a cooperating manufacturer together with the socket and foot. At the manufacturer's plant, the socket is set up on a second, similarly modified, adjustable AK leg according to the data sent from the fitter. The setup is then placed in the alignment duplication jig, and the prescribed components are incorporated into the system. The finished leg is returned to the fitter.

After two months of VAPC testing, three manufacturers (Southern Prosthetic Supply Co., Atlanta, Ga.; United States Manufacturing Co., Glendale, Calif.; Winkley Artificial Limb Co., Minneapolis, Minn.) were asked to cooperate in the further development of the system. Using the Hosmer AK adjustable leg, modified by the VA Prosthetics Center, the three manufacturers are routinely assembling and finishing legs. To date the VA Prosthetics Center has purchased 63 above-knee limbs fabricated by this method. The delivery time varied from 14 to 24 days. Although finishing and shaping were satisfactory, some initial difficulty with height measurements was experienced, a problem easily resolved with the manufacturers.

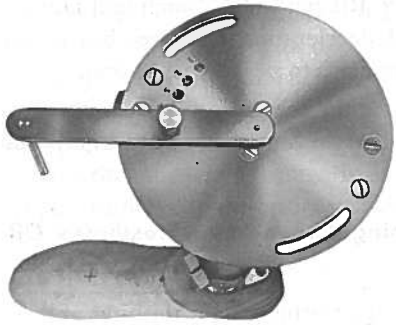

FIGURE 11. Adaptation of the Hosmer AK adjustable leg permits recording of initial knee flexion and toe-out at the same time. Sighting through the tube shown at the left of the illustration, the prosthetist marks the foot at a point which intersects the bore axis. When the alignment transfer is made, the foot is positioned so that the mark appears in the line of sight through the tube. This process replicates the original knee flexion and toe-out attitudes.

II. UPPER EXTREMITY PROSTHETICS

Evaluation of Externally Powered Components

A study has been undertaken to evaluate the utility of CO_2-powered components for upper-extremity amputees. It is designed to provide information on the comparative functional values of individual components, externally powered and conventional terminal devices, wrist units, elbows, and, eventually, shoulder abduction and flexion joints. Selective substitutes of externally powered components for body-powered units will be used to evaluate functional gains through so-called "hybrid" systems of various types.

A bilateral amputee with a left shoulder disarticulation and a very high right above-elbow amputation was fitted with conventional prostheses, trained, and then given an opportunity to wear and use the prostheses for several months. He had not worn prostheses previously. His reactions were reviewed, and his functional status measured; he was subsequently fitted with an externally powered hook-and-wrist rotation unit designed and provided by the American Institute for Prosthetic Research, N.Y. After a period of intensive training, the amputee was returned to his home station to be given additional training and to gain experience in the use of the externally powered components. The study program calls for him to return for evaluation of his reactions and functional status and for fitting, if warranted, with an externally powered elbow.

The American Institute for Prosthetic Research has been extremely cooperative in helping the VA Prosthetics Center launch this study. Mr. Edward Kiessling of AIPR has given us excellent advice and guidance.

III. LOWER-EXTREMITY BRACES

A. Evaluation

1. Below-knee Weight-bearing Brace With PTB-Type Socket. The patellar tendon-bearing below-knee brace design is now undergoing a compre-

hensive evaluation by New York University. It is expected that in the near future that group will begin the first phase of the evaluation, *Review and Examination of Subjects Fitted by VAPC.* We have surveyed all VA Brace Shops throughout the nation to give NYU some indication of the number and types of patients fitted by the Veterans Administration.

Electronic instrumentation developed by the VA Prosthetic Center for a biomechanical analysis will be made available by Bioengineering Research Service, VAPC (Fig. 12).

B. Development

1. Single-bar Brace. The single-bar brace design has been used on a number of leg brace (short leg brace) wearers having flaccid dropfoot conditions. Generally satisfactory results were obtained on those cases where a positive stop was used at the ankle joints. Several patients, however, fitted with commercially available, spring-loaded, ankle braces (single-bar design) found it difficult to clear the floor (toe drag), because the single spring did not provide adequate force to lift the foot.

Various materials and cross sections are being studied for use in single-bar, leg-thigh (long leg) braces. Recently, a failure was reported in a tubular chrome-molybdenum steel single-bar brace worn for 3 months by a paraplegic. The fatigue failure occurred in the lower bar and seemed to have been caused by anterior-posterior bending, a condition which was probably accelerated by the 90° positive stop. The Cr-Mo steel bars on this brace were replaced with aluminum having a modified "I" cross section. The patient has been using the modified aluminum "I" beam brace for 2 months without structural failure.

IV. ORTHOPEDIC AIDS

A. Evaluation

In discharging its responsibility for the care of orthopedically disabled veterans, the Veterans Administration attempts to provide the best possible

FIGURE 12. A strain gage: instrumented, S-shaped stirrup section mounted on a clear plastic shoe. Substituted for the conventional stirrup, this electronic device provides measures of the total axial loads transmitted through the sidebars of the brace.

professional and technical services. Continuing efforts to improve these devices are made through support of extensive research and education programs in orthopedic surgery, physical medicine, prosthetic and orthotic technology, and clinical services. The patient-treatment teams are provided with modern equipment to maintain a treatment level second to none; a variety of newly developed wheelchairs, orthopedic aids, and other devices are continually examined with a view toward replacing older devices with others of proven superiority.

The Veterans Administration undertakes to evaluate the utility of such devices for two cogent reasons: (1) There is always the possibility that a particular development may be a valuable asset to treatment practices; (2) Evaluation results fed back to developers provide guidance for improvement and innovation in the direction of specific VA needs.

Orthopedic aids submitted to the Veterans Administration are evaluated by the Bioengineering Research Service of the VA Prosthetics Center, where a systematic test program is applied in the assessment. Each device is considered from several standpoints: design and material, mechanical function, and biomechanical aspects of operation. In support of the bioengineering program, data on the clinical use of these devices are being provided by two other VA installations, the Departments of Physical Medicine and Rehabilitation Service of the Bronx VA Hospital and of the New York Manhattan VA Hospital.

During this period, 12 orthopedic aids have been evaluated: 4 wheelchairs, 4 types of standing devices, and 4 patient lifts.

1. Wheelchairs. Each of the wheelchairs was evaluated on the basis of the criteria published in the previous issue of this journal (8). As all of these chairs were designed along generally similar lines, a detailed, intensive analysis was required to determine the merit or drawback of specific differences related to weight, wheel design, center of gravity location, and propulsion energy requirements. On completion of each evaluation a report of findings was submitted to the manufacturer, including, in certain instances, specific recommendations for improvement. Experience to date indicates that currently available specifications for wheelchairs are not entirely adequate because of the development and use of newer materials, improvements in design, and the need for functional criteria. We are in the process of developing a set of performance specifications to meet these needs.

2. Standing Bars. Four very different types of standing devices have been evaluated.

The Stand-Alone (9, 10) was designed to provide mobility as well as standing facility for paralyzed patients. It features a unique arrangement

for supporting the patient in an upright position, and provides drive wheels linked to rolling wheels for mobility. A detailed report on the evaluation of this device appears elsewhere in this issue.

The Kim Stander is designed to permit a patient to raise himself from a wheelchair into a standing position with little or no assistance and to support the patient for extended periods. The lift is electrically operated and features a worktable. Our detailed findings have been submitted to the developer.

The Nyquist Portable Standing Bars are ruggedly constructed with no moving parts. They are designed to allow a patient to pull himself out of a wheelchair and to attain an erect position by means of supporting straps.

Several sets of handholds and grabbars have also been evaluated. These devices are primarily designed for permanent installation in hospitals, nursing facilities, and private homes. Their purpose is to improve the independence of disabled persons by assisting them in accomplishing such activities of daily living as rising from a bathtub or commode. In addition to mechanical analyses of structure, materials, and installation requirements, extended clinical experience provided the basis for our evaluation.

3. Patient Lifts. Evaluation studies were completed on four devices designed to transfer patients from beds to chairs, chairs to commodes, and in and out of automobiles. A series of design analyses, mechanical evaluations, and biomechanical tests were performed, and extensive clinical experience was obtained on each of the following: the Iatro American Lift Aid, Invalex Portable Patient Lift, Wolfe Patient Lift, and the Hoyer Travel Lift.

Although all such devices are designed with similar purposes, they employ a wide variety of power sources, mechanical linkages, and configurations. While the development of specifications for materials, dimensions, attachments, construction, and the like would make the evaluation process simpler, it might also tend to stifle creative development in this field; nevertheless, functional standards related to the performance of the device under appropriate conditions are required in the interest of safety, therapy, convenience, and economy.

On the basis of our experience in evaluating these devices, we have proposed a set of functional specifications for this purpose. The functional standards embodied in these criteria were applied in the evaluation of several of the devices listed above. The specifications below, therefore, represent a codification of common sense and utilitarian experience. They are in process of being evaluated in consultation with medical and therapy personnel of physical medicine and rehabilitation services in several stations.

VETERANS ADMINISTRATION SPECIFICATION (PROPOSED) FOR LIFT AIDS, PATIENT, GENERAL

1. Scope and Classification

1.1 *Scope.* These standards relate to mechanical devices for transferring severely disabled persons into wheelchairs, beds, bathtubs, commodes, car seats, and similar aids. Such devices may be used in hospitals or in private homes.

1.2 *Classification.* These devices shall be of one grade, one class, and two general types: (1) Those designed for operation by the patient and (2) Those to be operated by an attendant.

2. Applicable References and Specifications

2.1 *Federal Standard #123,* Marking for Domestic Shipment (Civilian Agencies).

2.2 *Specification #7836100C,* VA Specification for Lift, Patient, Hydraulic.

2.3. *Evaluation of Orthopedic Lift Aids,* Bioengineering Laboratory, VA Prosthetics Center.

2.4. *Federal Specification CCC–C–419,* Cloth, Cotton, Duck, Unbleached, Plied—Yarns (Army and Numbered).

2.5. *Federal Specification QQ–C–320,* Chromium Plating (Electrodeposited).

2.6. *Federal Specification QQ–N–290,* Nickel Plating (Electrodeposited).

2.7. *Federal Specification QQ–P–416a,* Plating, Cadmium (Electrodeposited).

3. Requirements

3.1. *Components and Materials.* The lift shall consist of the following major components: (a) base, (b) mast, (c) lifting assembly, (d) suspension system, and (e) accessories.

Materials: The base, mast, lifting assembly, and other supporting components shall be of a material and construction which do not permit the point of weight application to deflect more than 1 in. for every 100 lb. of load. The sum of all the torque and bending-moment deflections in the base, mast, and lifting assembly shall not exceed this value at any point in the lifting range, i.e., even when the load is suspended at a point farthest away from the mast-base intersection.

3.1.1. *Base.* The lift shall be supported by a stable base which shall:

a. Clear doorways with a minimum width of 23 in.;

b. Provide adequate configuration or adjustment to maintain the load within the area of the base of support at all times;

c. Permit lifting the patient from the floor;

d. Permit a wheelchair to be positioned with the center of the seat under the line of lift;

(e) Include provision for casters and mast attachment.

3.1.2. *Casters.* Each lift shall have:

a. Four detachable and replaceable swivel casters;

b. Easily accessible locks to prevent movement under load;

c. An overall maximum height of 4 in. so as to allow positioning the lift under beds or autos (as measured from floor to top of base).

Materials: Each caster shall be of a material, dimension, and configuration capable of withstanding vertical static loads of 600 lb. without permanent deformation of any component of the assembly. This specification is based on a "worst condition," i.e., a 300-lb. patient plus 100 lb. of inertial forces in lifting, lowering, or swinging applies 400 lb. of force to any two casters at one time. With 3 as a safety factor, 600 lb. per caster must be withstood without deformation. Corrosion-resistant material should be used throughout the device.

3.1.3. *Mast Attachment.* The device to fix the mast and lifting assembly to the base should:

a. Be welded to the base;

b. Permit easy detachment of the mast.

3.1.4. *Mast.*

a. If the mast is of a telescoping type, it should not have obvious lateral movement.

b. If the mast is installed on a castered, movable base, its rotation must be restricted to prevent the load from swinging outside of the base of support.

c. It shall provide attachment for an elevating member, such as a boom or component with similar function.

3.1.5. *Lifting Assembly.* Includes (1) supporting structure such as booms, (2) elevating mechanisms such as screws and jacks, and (3) power sources.

a. The full range of vertical displacement should exceed 35 in.

b. The full range of horizontal displacement should keep the load within the base of support at all times.

c. Clearance must be maintained between the mast and patient (reference: 6 ft.-4-in. man) in all phases of the lift.

d. It shall be capable of moving a 300-lb. load through the complete range, stopping and starting at any point in the range without slippage.

e. The lift shall provide an auxiliary means of lowering the patient in the event of failure in the main power source.

f. It shall be capable of being positioned directly over loads to be lifted or lowered.

g. The elevation control device (handle, button, switch) shall be within reach of the occupant, or at approximately waist height if designed for operation by an attendant.

3.1.6. *Suspension System.*

a. The seat and backrest shall maintain the subject in a sitting position, supporting back and buttocks in a comfortable manner.

b. It shall accommodate individual differences as regards shoulder width, seat width, and distance between buttocks and backrest by adjustment or accessory components.

c. It shall be wide enough to prevent excessive pressures at shoulder and hips.

d. The suspension system shall be attached to the lifting assembly in such a manner that the chains (if used) do not press on the occupant.

Materials: Strength should be equivalent to Federal Specification CCC–C–419 for duck wheelchair seat liners. Materials should be washable, nonshrinking, stain and water resistant, odor resistant, nonirritating, and flame resistant. The design shall permit removal of corrosion-susceptible hardware for laundering of fabric parts. Material shall be fitted with corrosion-resistant grommets or be otherwise adequately reinforced.

3.1.7. *Accessories.* Special items to extend the utility of a lift shall be available as optional equipment for a period of 10 years and should include:

a. Front strap for patients with tremors or spasms;

b. Commode sling;

c. Headrest support for extremely feeble patients;

d. Fixed mounts for bathtub, bed, or automobile.

3.2. *Functional Requirements.* Specifications governing safety, comfort, and operation.

3.2.1. *Safety.*

a. The lift shall provide a uniform rate of elevation and descent. The rate of descent shall not exceed 7 in. per second.

b. It shall be capable of stopping at any point during the lift or descent without slippage.

c. All sharp edges and projections likely to come in contact with the patient shall be padded.

d. Lead screws, gears, and cams shall not be exposed.

e. The lift shall be equipped with a device to limit movement within the prescribed range and prevent overdriving.

f. A decoupling device (swivel or other) shall be provided between the load and the lifting assembly to permit rotation of the load independently of the base.

3.2.2. *Comfort.*

a. No component shall concentrate uncomfortably high pressures on the patient.

b. All hard material coming in contact with the patient shall have adequate padding.

c. During the lift, the patient shall be maintained in a comfortable, upright sitting position.

3.2.3. *Forces to Operate.*

a. Nonelectrically powered devices (hydraulic, pneumatic, or mechanical) shall not require more than 100 in.-lb. of torque to operate under a maximum load of 300 lb.

3.2.4. *Maneuverability.*

a. When assembled, the lift shall be easily maneuverable by one person.

b. It shall be capable of rolling over small obstacles such as thresholds.

c. When disassembled, the lift shall be portable in sections by one person; the heaviest single section shall not exceed 35 lb.

3.2.5. *Ease of Operation.*

a. Positioning the lifting assembly and suspension system over the load shall be possible by a normal person with minimal effort.

b. Placing the means of suspension under the subject prior to lift shall be conveniently accomplished.

c. The patient shall not be deposited in such a manner that he must be manually repositioned.

3.2.6. *Durability, Strength, and Corrosion Resistance.*

a. The lift and accessories shall be of high-quality materials and workmanship and free from any imperfection which detracts from appearance or which may impair serviceability.

b. Finish. All chromium or other corrosion-resistant-type plating shall be in accordance with the requirements of Class I of Federal Specification QQ–C–320, and Class I, Type III (KS) of Federal Specification QQ–N–290, and QQ–P–416a.

c. Workmanship. Welds shall be ground smooth, free from rough areas and pits which may affect appearance or serviceability.

3.3. *Stamp Plate or Product Marking.* Each lift shall be permanently and legibly marked with the name or registered trademark of the manufacturer.

3.4. *Instructions.* Each lift shall be provided with an illustrated description of the assembly, operation, and care of the device.

3.5. *Quality Control.* Each unit shall be subjected to appropriate quality control measures by the manufacturer or distributor to insure compliance with these specifications.

4. Sampling, Inspection, and Test Procedures

4.1. *Sampling.* One randomly selected unit from each lot will be tested for compliance with specifications requirements.

4.2. *Inspection.* Every device (100 percent of sample) will be visually inspected for complete assembly, finish, workmanship, and condition, and be checked for operability.

4.3. *Test Procedures.* One randomly selected unit will be subjected to appropriate testing by the commodities inspector.

4.3.1. *Adjustment Test.*

a. All mechanical adjustments shall be operable with minimal effort.

b. If base is adjustable, it shall function smoothly under the maximum load of 300 lb.

4.3.2. *Capacity Tests.*

a. The lift shall raise a 300-lb. load from the floor to a height of not less than 35 in. and hold it steady at that height.

b. The maximum lowering speed shall not be in excess of 7 in. per second under maximum load of 300 lb.

4.3.3. *Functional Tests.*

a. Transferring a patient from floor to chair and from chair to bed shall be safely and conveniently accomplished.

b. All movable components shall operate smoothly and efficiently and with no appreciable discomfort to the patient.

B. Development

A prototype of a stair-climbing wheelchair has been submitted for evaluation by Project Progress of Virginia, Minn. It is designed generally as a conventional wheelchair with two stair-climbing rubber treads carried beneath the frame between the two large driving wheels. To climb stairs, the treads are lowered to contact the floor, and the drivewheels are raised by means of a hand-operated hydraulic pump. The stair-climbing treads are hand-driven by the large drivewheels through a system of gears and shafts (Fig. 13).

Several other developers have indicated progress along similar lines using, however, portable direct-current power for the stair-climbing function.

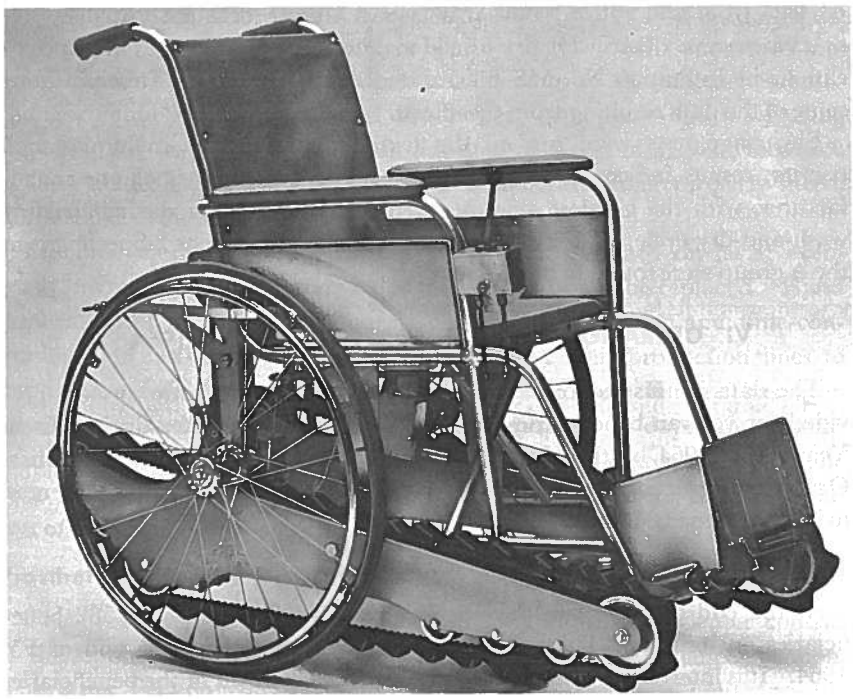

FIGURE 13. The Project Progress (Virginia, Minn.) stair-climbing wheelchair.

V. TESTING STANDARDS

One of the major responsibilities of the VAPC is the inspection and testing of prosthetic devices and supplies to insure that minimum standards of

quality and performance specified in the procurement contracts are being met. A materials-testing program designed for this purpose has been in effect for several years. Continuous efforts are made to reduce the element of personal judgment involved in these procedures and to apply the most precise and reliable methods available. Test results depend on the precision of the methods employed; it is therefore essential to develop, modify and, if necessary, to procure appropriate equipment to maintain a systematic approach to better product development.

All available sources are consulted to improve test methods and specifications. The advice of the U.S. Bureau of Standards, for example, has been of great assistance in our stump-sock test program. It was also quite gratifying to learn that our procedures are the equal of those recommended by the U.S. Bureau of Standards. Continual efforts to upgrade this program are necessary to fulfill our obligation to the disabled by insuring that approval by the Veterans Administration symbolizes excellence. During the period covered by this report, three compliance-test programs were completed.

Stump Sock Specifications Compliance Tests (Interim) were conducted on leg (wool and cotton) and arm (wool) stump socks submitted by five manufacturers. Tests were performed to determine compliance with Specifications in Invitation No. MS–6–65 dated April 13, 1964. Table 7 shows some of the data resulting from these tests.

Shrinkage tests were run on the textile material used in lumbosacral corsets; results indicated excessive shrinkage in the product of one manufacturer, with the greatest amount of shrinkage occurring during the first wash-and-dry cycle. Table 8 shows the results of these tests. Specimens in the A group were found to be deficient.

VI. OPERATIONS REPORT FOR FISCAL YEAR 1964

The data in this section reflect the prosthetic and orthotic services provided for veteran beneficiaries by the VA Prosthetics Center during Fiscal Year (FY) 1964 by the VAPC Orthopedic Shoe Service and Prosthetics-Orthotics Service. When feasible, data of previous years are presented to indicate general trends.

The Orthopedic Shoe Service

At the end of FY 1956, the Orthopedic Shoe Service had 5,582 beneficiaries on its rolls. This number increased to 10,021 by the end of FY 1964. It is interesting to note that in FY 1956, 91 percent of the beneficiaries received new shoes and 37 percent had their shoes repaired through the Orthopedic Shoe Service. In FY 1964, new shoes were furnished to 75 percent of the beneficiaries, while the percentage of shoes repaired through the Orthopedic Shoe Service rose to 95 percent; truly a tribute to the effectiveness of the shoe repair program. It should also be noted that this

savings was accomplished without depriving any beneficiary of those services to which he was entitled.

Although costs for these services have increased, the average costs for the repairing of orthopedic shoes have, fortunately, been relatively stable for the last five years. The magnitude of the present program is indicated by a comparison of the overall cost for new shoes and repaired shoes in FY 1956 and in FY 1964; approximately $534,000 was spent for both new and repaired shoes in FY 1964 as against only $211,700 in FY 1956.

The distribution of protective footwear (rubbers and/or overshoes) is also a responsibility of the Orthopedic Shoe Service. One hundred of the items furnished were rubbers; 214 were overshoes. Careful analysis of measurement of shoes and of lasts for overshoes and/or rubbers often made it possible to use the same last for several different cases, thus saving the cost of fabricating a new last. The cost of the protective footwear program for FY 1964 was $4,033.

The initial fitting of custom-made orthopedic shoes may be maintained for a long period but, as time goes by, changes in the foot will lead to poorly fitting shoes. Prompt and proper feedback from veterans and/or field stations has made it possible to modify lasts so that these "normal" changes could be accommodated for in new orthopedic shoes. During FY 1964, 969 lasts were modified or adjusted at a cost of $1,926; feedback of data made it possible, however, to continue furnishing properly fitted, comfortable orthopedic shoes to veteran beneficiaries.

Machine methods of shoe fabrication tend to be hard on lasts; eventually, no matter how well they are made, the lasts must be replaced. Hand duplication of an orthopedic shoe last is extremely tedious and time-consuming, and last-makers do not like to interrupt their production lines to duplicate a few custom items. The Orthopedic Shoe Service has been fortunate in finding a unique local facility equipped to duplicate custom-made orthopedic shoe lasts and willing to do so at an extremely reasonable cost. During FY 1964, 380 lasts were duplicated and replaced at a total cost of $3,375.

Prosthetics-Orthotics Service

The Prosthetics-Orthotics Service of the VA Prosthetics Center consists of the Limb and Brace Section and the Prefabricated Appliances Section. The latter Section consists of the Prosthetic Components Unit and the Orthotic Components Unit. These units are primarily responsible for the national distribution of lower-extremity prosthetic devices, elastic hosiery, and surgical supports. Though relatively new, these programs are expanding rapidly, as shown in Tables 9 and 10.

TABLE 7.—*Summary of Tests on Stump Socks (Leg, Wool) for Specifications Compliance, Invitation No. MS–6–65 April 13, 1964*

Sample	Interim Check of Fabric Structure				Maximum Shrinkage (%) after 5 Washing Cycles		
	Yarn No. (ASTM 1961 D123)	Ply (No. of strands per sq. in.)	Stitches per Inch	Wales per Inch	Length (in.)	Width at 2 in. from Toe (in.)	Width at 6 in. from Toe (in.)
VAPC Specs	16–18.5	3–7	18±2	12±1	12	10	10
A	18.2	5	17	12	7.8	3.1	6.6
B	18.4	5	17	11	.4	7.1	7.2
C	17.4	5	17	12	9.2	4.8	5.9
D	17.4	5	18	13	9.2	4.3	5.6
E	16.8	5	17	11	4.4	4.1	7.0

TABLE 8.—*Shrinkage Tests on Lumbosacral Corsets VA Invitation, Bid and Award, V1005P–307–A*

Sample (4-in. length)	Average Total Shrinkage [a] After 5 Washing Cycles	
	Inches	Percentage †
Mfg. A (10 samples) [b]	¼	6.1
Mfg. B (10 samples)	⅛	2.6

[a] Maximum allowable shrinkage rate: 3%.
[b] Product did not pass VAPC specifications.
† Average of individual calculations.

Orthotic Components Unit

Elastic hosiery and surgical supports are issued either directly to individuals (based on a doctor's prescription) or to orthopedic shops in VA hospitals for issuance to beneficiaries. The elastic hosiery national distribution program began in FY 1963. Its growth is shown in Table 9.

The distribution of surgical supports was initiated on a limited basis in FY 1961; nation-wide distribution began in FY 1962, and its growth has been increasing steadily (Table 10).

As new items are developed in either elastic hosiery or surgical supports. their clinical evaluation can be expedited through the centralized distribution program. The VA Orthopedic Shops are excellent intermediaries in this distribution program, as evidenced by the numerous repeat orders.

TABLE 9.—*Elastic Hosiery Costs*

[VA Centralized Distribution Program]

Issuance	FY 1963		FY 1964	
	No.	Cost ($)	No.	Cost ($)
Individuals	5,806	33,384.50	9,989	47,460.00
VA orthopedic shops	6,598	37,938.50	12,258	55,130.00
Total	12,404	71,323.00	22,247	102,590.00
Average cost [a]	.....	5.75	.....	[a] 4.61

[a] It should be mentioned that the lowering of the average cost here was primarily due to a change in internal accounting procedures in which certain costs previously chargeable to each item are no longer so chargeable. The true average cost in the two years was about the same.

TABLE 10.—*Cost Figures for Surgical Supports*

[VA Centralized Distribution Program]

Issuance	FY 1962		FY 1963		FY 1964	
	No.	Cost ($)	No.	Cost ($)	No.	Cost ($)
Individuals	1, 153	4, 727	2, 162	11, 568	2, 434	12, 901
VA orthopedic shops	3, 465	14, 206	3, 098	12, 701	3, 995	18, 091
Total	4, 618	18, 933	5, 260	24, 269	6, 429	30, 992
Average cost	.....	4. 10	.....	4. 61	.....	4. 82

Prosthetic Components Unit

Since the Clinical Application Study of the Hydra-Cadence Above-Knee Prosthesis indicated that the limb was acceptable for issuance to veteran beneficiaries, it has been placed on contract. Distribution of Model "D" assemblies (unit, foot, and cover) began in January 1962; from that date through June 1964, 1,052 above-knee amputees have been fitted with this type of prosthesis.

Malfunctions of the Hydra-Cadence, however, continue to be a problem. During FY 1964 it was necessary to replace 535 units, 505 cosmetic covers, and 90 feet. The total cost for replacements and repairs was $150,911. The average maintenance cost per beneficiary for FY 1964 was $143.45. To alleviate this situation, Model "E" unit is now being manufactured which incorporates a number of major design changes. It is still too early to tell the effect these modifications will have on the malfunction problems. Investigation of various materials for the cosmetic cover is being conducted at the VA Prosthetics Center; preliminary results have been encouraging.

Prior to April 1964 all malfunctioning units had to be returned to the manufacturer for repair. This procedure was both expensive and time-consuming since it also necessitated the maintenance of large inventories of units of all sizes. Discussions with the manufacturer led to the decision that many of the repairs could be performed by technicians in the VA Prosthetics Center. Several men were trained, and a repair program started in April 1964. During the last 3 months of FY 1964, 156 units were repaired at the VA Prosthetics Center for a total cost of $1,232; the manufacturer's charges to repair 101 units were $3,306, but it must be understood that their repairs were of a major nature that could not be done at the Prosthetics Center.

Malfunctioning Model "D" units sent to the manufacturer for repair are being converted to Model "E" units; in effect we are receiving updated units with new guarantees. To date, 80 units have been converted to Model E.

During FY 1964, the Prosthetic Components Unit also distributed components for temporary prostheses to the 27 VA orthopedic shops. These components included 109 above-knee units, 71 below-knee units, and 320 SACH feet of assorted sizes. This initial distribution was based on the past usage of temporary prostheses by the orthopedic shops in VA hospitals. Additional items—7 above-knee units, 17 below-knee units, and 32 SACH feet—have been sent to 4 of the 27 orthopedic shops.

Those VA hospitals without orthopedic shop facilities utilize commercial limb establishments to fill their prescriptions for temporary prostheses. Eleven above-knee units, 3 below-knee units, and 5 SACH feet have been shipped directly to commercial limb facilities.

Limb and Brace Section

The fitting, alignment, and fabrication of prosthetic devices and orthopedic appliances for entitled beneficiaries is a prime responsibility of the Limb and Brace Section. During FY 1964, 5,622 beneficiaries were serviced by that facility. Of that number, 1,349 received new items, *i.e.,* a prosthetic device or an orthopedic appliance; the remainder came in for various repair services, including modifications and adjustments of devices or appliances.

Seventy-one total-contact sockets were included in 135 above-knee prostheses delivered to amputees; stump casting for these sockets was accomplished on the VAPC casting stand. Thirteen above-knee prostheses were fabricated using the system for finishing prostheses off-station described earlier in this report. A similar procedure was used for 33 of the 51 upper-extremity prostheses delivered during FY 1964.

Thirty-six below-knee amputees were fitted with patellar tendon-bearing prostheses, and 103 amputees received conventional below-knee prostheses with thigh lacers and knee joints. Of the 10 Syme prostheses fitted, 9 were of the VAPC medial-opening type, and 1 utilized a metal frame.

Knight braces accounted for 85 of the 130 spinal braces fabricated. Of the 62 leg-thigh (long-leg) braces furnished to beneficiaries, 13 were of the weight-bearing type; 112 nonweight-bearing and 8 weight-bearing leg braces (short-leg) were delivered; 1,505 arch supports were fitted.

The VAPC Clinic Team

During 59 meetings in FY 1964, the VAPC Clinic Team saw 108 veteran beneficiaries. These patients came from 19 states and the District of Columbia. Many of those cases were of the "problem" variety referred to the Prosthetics Center by various field stations. Others were volunteer participants in some of the Center's research projects. Some of the "problem" cases were ideally suited for the Center's research needs. One of these was an upper-extremity amputee who had to be fitted as a bilateral shoulder-disarticulation case. He made exceptional progress using con-

ventional prostheses. From time to time, component substitutions are made and the man's performance is re-evaluated to determine the effect of the new component. In another case, a spinal-cord-injury patient had been wearing conventional double-bar braces for years. Upon the Clinic's recommendation, he was fitted, bilaterally, with single or unibar braces. The resultant weight decrease led to increased comfort and better performance and, hence, greater patient satisfaction.

REFERENCES

1. Zamosky, Isidore: Shoe Modifications and Lower-extremity Orthotics. Bull. Prosthetics Reas., BPR 10–2: 54–95, Fall 1964.
2. VA Prosthetics Center: Semiannual Report of the VA Prosthetics Center. Bull. Prosthetics Res., BPR 10–1:106, Spring 1964.
3. *Ibid:* 108–109.
4. Mozer, D. L.: Immediate Postoperative Fitting of Lower-extremity Prosthesis. Orthopädie Technik, 120–128, May 1964; English Trans., Orthopedic and Prosthetic Appliances J., 105–109, June 1964.
5. VA Prosthetics Center: Semiannual Report of the VA Prosthetics Center. Bull. Prosthetics Res., BPR 10–1:111–112, Spring 1964.
6. College of Engineering, University of California, Berkeley: Fundamental Studies of Human Locomotion and Other Information Relating to Design of Artificial Limbs. 1947.
7. Murray, M. P., Drought, A. B., and Kory, R. C.: Walking Patterns of Normal Men. J. Bone and Joint Surg., 46–A(2):335–360, Mar. 1964.
8. Peizer, E., Wright, D., and Freiberger, H.: Bioengineering Methods of Wheelchair Evaluation. Bull. Prosthetics Res., BPR 10–1:77–100, Spring 1964.
9. Bernstock, W. M.: Clinical Application Studies. Bull. Prosthetics Res., BPR 10–1:70–71, Spring 1964.
10. Peizer, E. and Bernstock, W. M.: Bioengineering Evaluation and Field Test of the Stand-Alone Therapeutic Aid. Bull. Prosthetics Res., BPR 10–2: 1–38, Fall 1964.