SEMIANNUAL REPORT OF THE VETERANS ADMINISTRATION PROSTHETICS CENTER

(July-December 1964)

INTRODUCTION

Presented below are the latest reports on major activities in the VA Prosthetics Center's research, development, and evaluation programs. Again included are data about several of the Center's operational programs. These data can be compared to those published in the Bulletin of Prosthetics Research, 10-2 (1) containing the previous semiannual report of the Center.

The semiannual report does not cover all activities of the VA Prosthetics Center. Certain other research, development, and evaluation projects now underway have not as yet produced significant findings for reporting in this issue. Also, the detail varies for the material covered in this report depending upon the progress made in each study.

Following is an outline of the contents of the semiannual report. It should be noted that work of a broad spectrum has been assigned to the Center. The effort in lower-extremity prosthetics, for example, includes activities from basic studies through evaluation to clinical practice. Even though under some of the other major headings the same scope of work is not now suggested, future reports might include such other aspects of research. For example, in upper-extremity prosthetics, a program involving some basic studies of motion is now being developed.

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Again, the semiannual report was produced by direct involvement of many Center personnel. The special activities of Dr. Edward Peizer for

the research portion and Mr. Frank A. Witteck and Mrs. Anastasia S. Keane for the operations portion should be recognized.

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I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

Below-Knee Joint Placement. The problems of placement of single axis knee joints and the functional requirements for thigh corsets have not been adequately resolved to date. The need for an adequate solution to this long standing question was emphasized recently by a workshop panel of the Committee on Prosthetics Research and Development (2). Radiological studies have been undertaken by the VA Prosthetics Center to investigate the relative motions of femur and tibia and to identify the single, most useful axis of rotation. Corollary studies are in progress to develop a method for locating the optimal axis of rotation by reference to external bony landmarks. In addition, we are evaluating the shape, size, and function of current thigh corset configurations.

The knee is generally classified as a hinge-type joint, implying a single axis of rotation throughout its range. However, it is well known that the knee is a combination hinge and gliding joint whose instantaneous center of rotation throughout the range of flexion-extension shifts as a function of the changing radii of the curve of the femoral condyles. Thus, the position of the center of rotation varies, and single-axis knee joints can only be aligned coaxially with the knee center of rotation at one point in the range. The anatomical knee joint center will deviate from this coincident alignment at all other points throughout the range. The location of the mechanical knee joint therefore represents a compromise, and optimal alignment is a matter of locating the joint centers to minimize relative motion in the greatest portion of the useful range. Failure to find an adequate compromise location can result in excessive piston action of the stump, undue stresses on the thigh corset with resulting discomfort, and diminution of function.

Based on limited X-ray studies, the instantaneous center of knee rotation has been plotted over the entire range of knee flexion. These analyses revealed a point about which the axis migrates minimally for approximately the first 100 deg. of knee flexion. This point, representing the best compromise, seems to bear a reasonably constant relationship to several bony landmarks (Fig. 1a, b). Varying slightly with knee width, its position is approximately 1⅛ in. above the lateral tibial plateau at a point approximately three quarters of the distance between the anterior tibial tubercle

and the posterior border of the head of the fibula measured from anterior to posterior. The validity of this hypothesis is being checked in an independent study undertaken by a New York University student working on a Master's thesis. Measurements will be obtained on a sample of approximately 90 normal males to determine the practicability of locating the functional knee center by reference to external bony landmarks. The sample will include an anthropometrical cross section with respect to nine categories of body type related to height and weight.

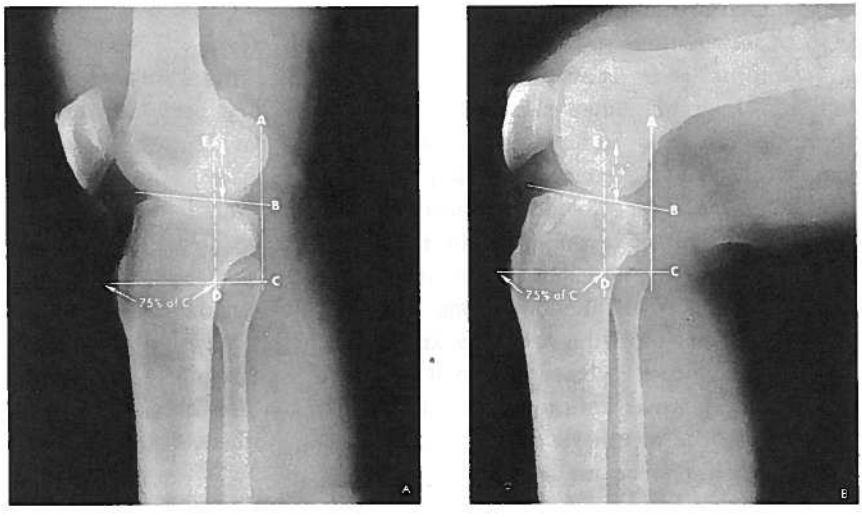

FIGURE 1. The indicated anatomical joint center bears a relatively constant relationship to the posterior borders of the shafts of the femur and tibia in extension (a) and at 90 deg. of flexion (b), a relationship which remains constant through this range. Shown are A—alignment of the posterior femur, tibia, and fibula; B—location of the lateral tibial plateau; C—an A-P line between the anterior tibial tubercle and the posterior head of the fibula; D—a point approximately ¾ of the distance between the anterior tibial tubercle and the head of the fibula; and E—a point approximately 1⅛ in. above the lateral tibial plateau representing the minimally displacing center of knee rotation.

Additional validation of the hypothesis is being sought in dissection studies on cadavers at Duke University by Dr. Frank Clippinger.

B. Components Development

1. Torque Absorber. The torque-absorbing device for lower-extremity prostheses has been redesigned to increase the resistance to external rotation by installing harder rubber bumpers in diagonally opposite quadrants as described in the Bulletin of Prosthetics Research, 10-2 (3). Three units were fabricated of which two have been installed in below-knee prostheses worn by patients. During the past several months these patients have

reported their general satisfaction and strong desire to continue using the torque absorbers. On the basis of this preliminary application, biomechanical studies are being initiated to determine the phasic pattern of function, excursion in the internal-external rotation range, resistance torques, and the influence of this device on other aspects of gait. Ten additional units are being fabricated for use in these studies. They will be installed in prostheses worn by several below-knee, above-knee, and possibly hip-disarticulation patients. Bilateral cases in each amputation category will be included in the sample.

2. *Above-Knee Pylon Designed for Use with Various Swing-Control Mechanisms.* There is available today an assortment of approved mechancal and hydraulic knee swing-control systems. They vary in order of sophistication and the amount of resistance they provide, ranging from a simple, mechanical, constant-friction device through a programmed hydraulic system. This provides physicians with a relatively broad armamentarium from which to prescribe the most appropriate device for an individual patient. But judicious prescription depends on a clear understanding of the differences in function among the systems. Even the most informed person may not have the necessary knowledge at his fingertips. While not irrevocable, once a selected device is installed, exchanging it for another device is expensive and time consuming. Aligned for one knee-control mechanism, the prosthesis does not readily accept others, and an entirely new setup has to be made.

We are developing a device to accept various mechanical and hydraulic swing-control units (Fig. 2). It consists essentially of a pylon attached to a yoke which is designed to receive knee-control mechanisms. Fitted with a foot attachment fixture, the pylon permits alignment adjustment and can be enclosed in a cosmetic cover.

This device can be used in hospitals as a temporary prosthesis without a cosmetic cover when early ambulation is indicated. It can be used as a clinical test instrument, aiding the physician in his final determination of the most appropriate knee mechanism by permitting a patient to try several easily interchanged components. Used with a cosmetic cover, it also serves as a permanent prosthesis which still permits interchange of knee-control components.

Fabrication of several test models is under way.

3. *Caliper for Measuring Below-Knee Stumps and Evaluating Socket Fitting.* This instrument, described in the previous issue of this Bulletin (4), was designed to permit more accurate measurement of the anterior-posterior knee diameter from the patellar ligament to the popliteal area and the medial-lateral knee diameter from the medial to the lateral epicondyle. Enabling the prosthetist to measure these dimensions more precisely is a significant step in the improvement of below-knee socket

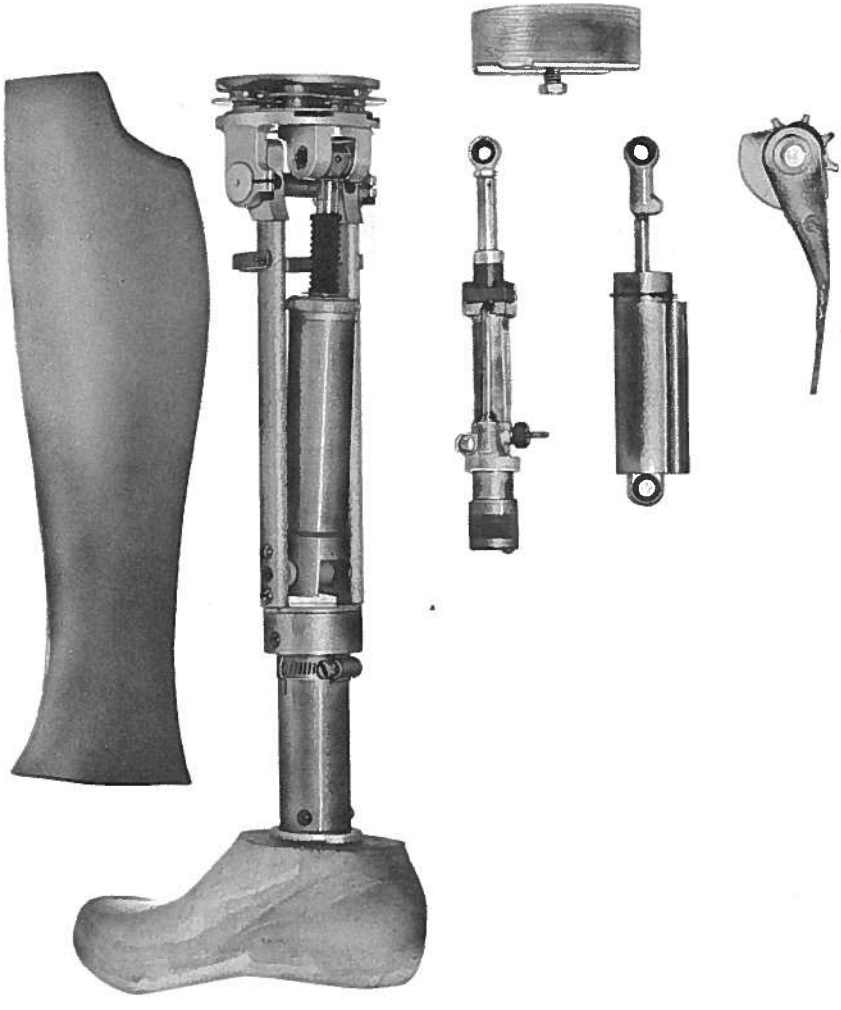

FIGURE 2. A developmental model of the pylon with a Mauch B swing-control unit installed. The knee-control mechanism is easily interchanged with several other hydraulic or mechanical friction units (adapters not shown): The Hydra-Knee, the Dupaco, and the Northwestern University Disk Friction System. The device, shown here attached to the VAPC knee coupling, can be fitted with the cosmetic cover shown on the left.

design and provides the clinic team a convenient means for reliably checking the adequacy of socket fit.

After final modification of the original caliper design, 50 units were fabricated. These were distributed to VA field stations and to universities together with a set of instructions for standardizing the measurement tech-

nique.* The calipers may now be purchased from the United States Manufacturing Company, 623 South Central Avenue, Glendale, California.

C. Technique Development

1. Direct Forming of Below-Knee Sockets. We have not as yet solved the problem of forming satisfactory below-knee sockets directly over a stump. Actually, two separate problems are involved: (a) finding a material which is dimensionally stable after being formed into a socket and which withstands the forces normally applied to sockets in standing and walking, and (b) developing a method of socket formation in which appropriate external forces are applied to produce a biomechanically sound design without requiring the conventional mold modification.

Originally, special casting forms were developed to apply forces in the anterior-posterior direction in order to mold a plaster-of-Paris wrap to the desired contour. This method had several disadvantages but the primary problems were the unsuitability of plaster-of-Paris for the final socket, and the molding pressures were not uniformly distributed around the entire stump, tending to distort the medial and lateral contours. Pre-shaped acrylic cones were unsatisfactory because the high temperature needed to "soften" the thermoplastic cones required the use of a protective stump cover. As a result, the sockets were oversized. In addition, since it was a thermoplastic material, the socket had a memory; i.e., it tended to return to its original conical shape and it was prone to "creep" or deform when subjected to body heat and to the forces exerted during weight bearing. Moreover, regardless of the material used, the patient was required to stand in a fixed position for periods of approximately 20 to 30 minutes.

Several other materials are now being considered for this direct forming on the stump. One which shows promise is a high viscosity resin which is applied with a spatula to a reinforcing fabric. A stump sock and thin rubber sleeve are used to protect the stump from the resin and to provide insulation against the thermal effects of polymerization of the plastics. In addition, we are investigating the application of a pneumatic-pressure casting technique to improve cast or direct socket forming (discussed below).

2. Pneumatic-Pressure Casting for Below-Knee Stumps. The adequacy of below-knee weight-bearing sockets depends on accurate replication of the stump—a difficult task. Casting by current methods does not allow precise control of the forces over the stump periphery; subsequent modifications to control these forces are similarly inaccurate. The resulting forces have no known relationship to those applied during standing and walking.

Casting methods which apply equal pressures all over the stump provide

*Available on request from the editor.

good replicas but fail to take advantage of those areas that can better tolerate high forces; modification of the replica is necessary to form a socket mold. Ideally, high pressures should be concentrated in the more tolerant areas with gradually reduced pressures applied to surrounding areas tapering to lower uniform pressure over the remainder of the stump. We concentrate the major portion of the forces of the patellar ligament and the flares of the tibial condyles and distribute the remainder over the softer and fleshier areas. Previous methods of accomplishing these goals included manual deformation of a plaster-of-Paris wrap and the use of casting forms by which pressure was applied to small areas of the stump by means of the VAPC casting stand. In both of these methods the forming forces were applied in the anterior-posterior direction with the resulting distortion of the fleshy masses in the medial-lateral direction. Casts obtained in this manner required major modification.

We are now employing a pneumatic-pressure casting system to control the distribution of external forces. Although fluid-pressure systems have been used in the past, ours requires only a pneumatic-pressure sleeve and a forming sleeve with specially contoured sections designed to concentrate the externally applied forces over the patellar ligament and the flares of the tibial condyles (Fig. 3 and 4). In this procedure all other external forces are uniformly distributed over the entire stump surface; the areas designed to bear force concentrations are formed by the contours within the forming sleeve. To prevent the distal migration of soft tissue, the end of the stump is capped with hardened plaster-of-Paris bandage. In an effort to prevent proximal migration of tissues and fluids during casting, the pressure sleeve was designed to encompass the entire knee and the femoral condyles. The circumferential pressures on the thigh just above the knee are of the same magnitude as those acting about the entire stump. By means of an air pressure gage, the casting pressure is monitored. At the present time, we are experimenting with casts formed under pressures of 2.0 lb. per square inch. The contour of the final product is dependent only on the resistive characteristics of the underlying tissue and the external pressure used to form the cast. To date, a total of twenty patients have been fitted with temporary sockets by this method. Approximately one half the number are now being fitted with permanent prostheses after using temporary prostheses for periods of 4 to 6 weeks. These patients, averaging 58 years of age, were fitted approximately 2 months after surgery. Most of the amputations were due to diabetic gangrene and arteriosclerosis. Stump lengths ranged between 5 in. and 6 in. below the tibial plateau.

We are providing other research groups at Duke University, the Prosthetics Research Study associated with the Seattle VA Hospital, and Northwestern University, with the pneumatic-pressure casting equipment

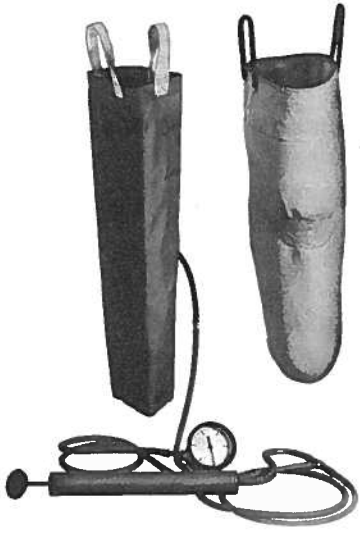

FIGURE 3. The equipment used for pneumatic casting. Shown are the Jobst bag (left), air lines, pump, and pressure gage used to apply controlled pressures to the specially contoured forming sleeve (right).

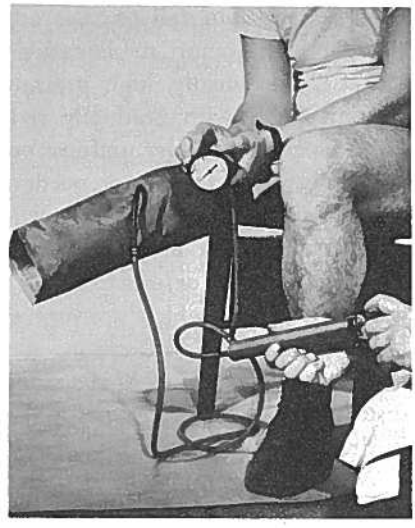

FIGURE 4. Pneumatic casting technique in which pressure is controlled in the pressure sleeve. Between the pressure sleeve and the stump are the rubber forming sleeve and the plaster-of-Paris wrap.

and appropriate instructions. Efforts by these groups will help us to determine the adequacy of sockets formed by pneumatic pressure upon seasoned below-knee stumps.

3. Hemipelvectomy and Hip-Disarticulation Stump Casting with the VAPC Casting Stand. The Canadian hip-disarticulation prosthesis is generally conceded to be a significant improvement over previously used tilting-table-type prostheses. Casting for the Canadian-type prosthesis, however, was difficult. Overhead slings to support the patient's weight and to maintain the "stump" in a weight-bearing condition were developed at Northwestern University and at Kolman Prosthetics in Whittier, California. These systems employ an overhead pulley arrangement which supports the sling vertically at a single point. As a result, a sling of insufficient anterior-posterior depth is formed, tending to distort the tissues undesirably.

We have designed a sling which offers several advantages. It is used with the VAPC casting standing and eliminates the need for additional, permanently installed equipment where the VAPC casting stand is already in use. It provides more convenient working conditions, permits easier height adjustment, improves patient stability during the casting procedure and, due to the flatter curve of the sling, does not excessively distort the tissues (Fig. 5).

The sling is a double thickness of canvas, 6 in. wide with attachment loops at each end. It is suspended from the arms of the casting stand by means of the loops. It provides a flatter weight-bearing surface and does not compress the surrounding tissue. Details on construction and application are published in FM-3, June 1, 1965: Casting Hemipelvectomy and Hip Disarticulation Amputees with the VAPC Casting Stand.

D. Evaluation (Components)

1. Northwestern University Disk Friction System. This device recently received the approval of the Veterans Administration for clinical use on its above-knee amputee beneficiaries after appropriate prescription. The present system (Fig. 6) is a simple, multiple disk, friction brake installed in the conventional wood knee-block and shank of an above-knee prosthesis. Variations in the angular size of the disks provide a stepped alteration of the resistance pattern during the swing phase (Fig. 7). The larger the disk angle, the longer it will provide resistance during the swing phase. The total frictional resistance can be altered by turning an adjustment wheel mounted concentric with the knee axis.

Earlier models of this device, originally developed at Northwestern University under VA contract, have been the subject of extensive testing which revealed certain deficiencies and provided guidance for redesign and improvement.

FIGURE 5. Casting for Canadian hip-disarticulation prosthesis. The casting sling is suspended from the VAPC casting stand. The relatively flat curve minimizes tissue distortion.

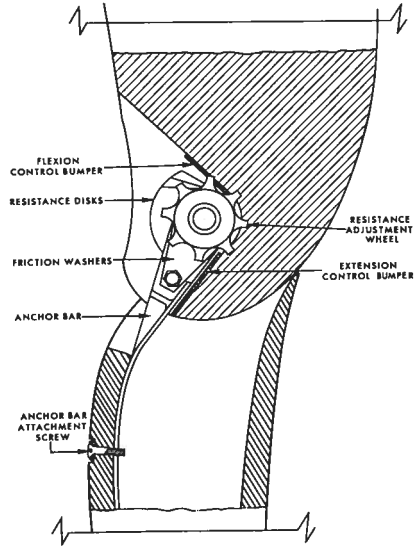

FIGURE 6. Northwestern University Disk Friction System in Setup.

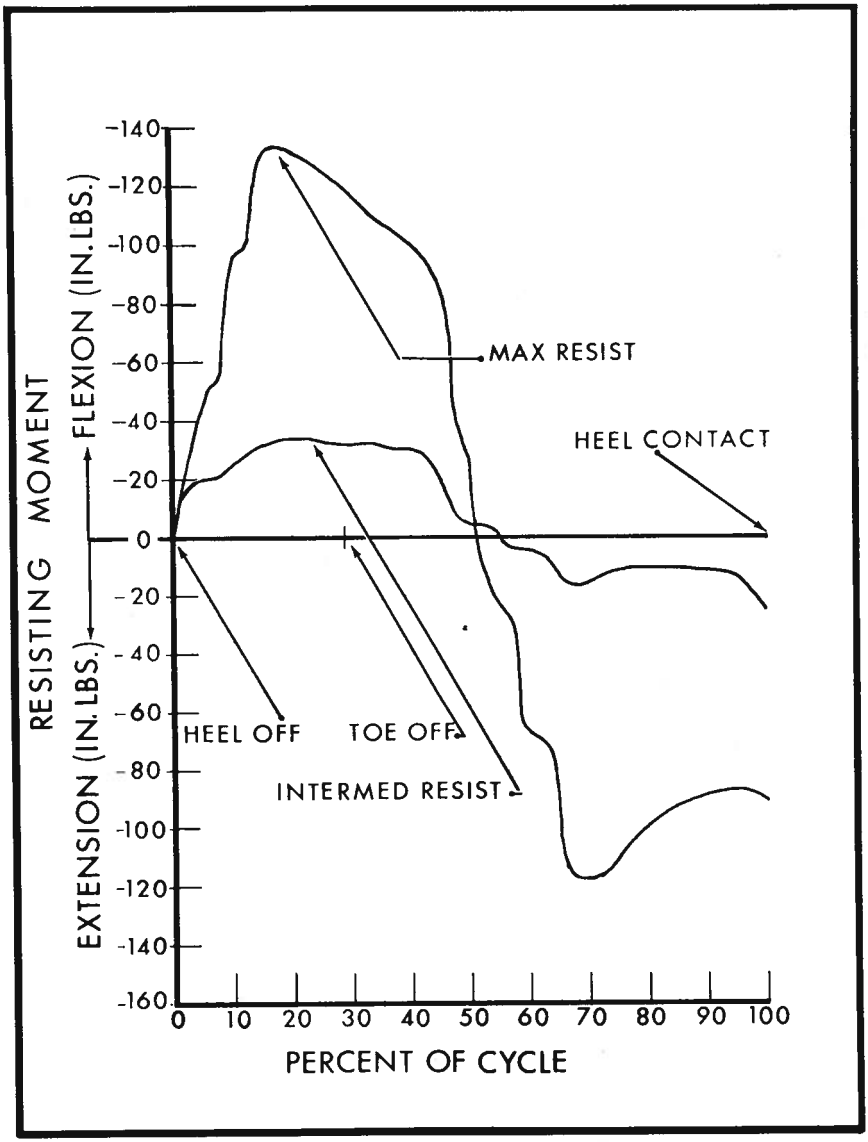

FIGURE 7. Flexion-extension resistance moments provided by the Northwestern University Disk Friction System adjusted for maximum resistance and for intermediate resistance. The stepped alteration in resistance can be seen clearly in the ascending and descending slopes of the flexion-extension curves at maximum resistance. These changes are partially obscured at the intermediate setting.

Despite a series of minor modifications made by the manufacturer, two major problems persisted. Early testing revealed a weakness of the anchor bar. The sharply curved bar was subject to microscopic fractures in fabri-

cation, leading to failure. The leather friction washers were totally inadequate, and it was suggested that the manufacturer substitute a brake-lining material which would not extrude as did leather.

Studies to determine the probable causes of failure in the anchor bar indicated that substantial forces were applied at terminal impact in the swing phase (Fig. 8).

A series of tests to evaluate the utility of the resistance mechanism revealed a short adjustment range between the minimum and maximum friction settings.

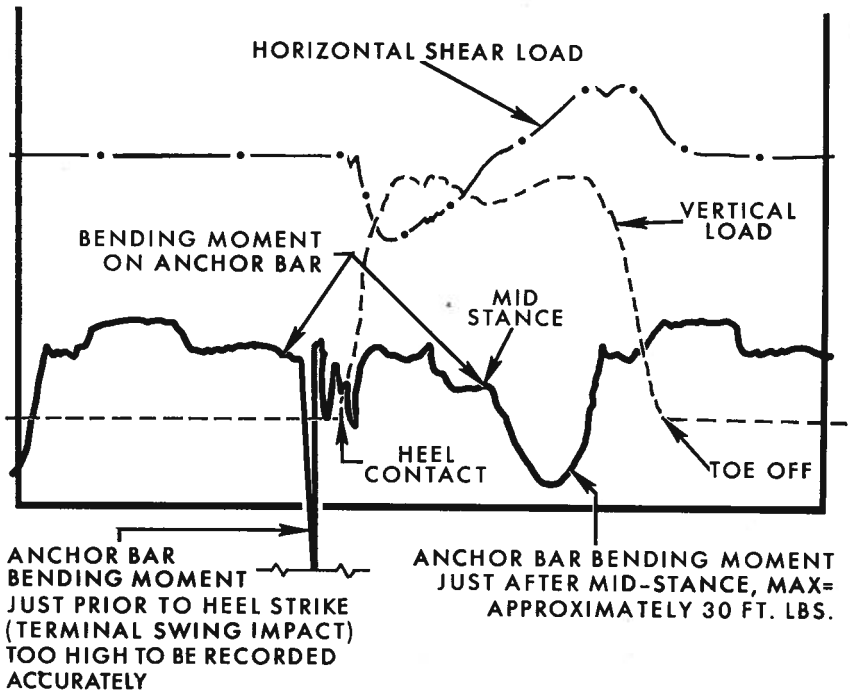

FIGURE 8. Bending moment on anchor bar of Northwestern University Disk Friction System.

Nevertheless, the unit provided adequate resistance forces. For example, the resistance magnitude fell just below the order of minimum resistance values of several hydraulic units. At minimum resistance settings, drop-times* among five hydraulic systems in flexion ranged from 0.20 seconds to 0.32 seconds and in extension from 0.19 seconds to 0.31 seconds. Comparable drop-times for the Northwestern system were 0.22 seconds in

*One measure of the resistance of knee mechanisms is the time required for the knee to travel the flexion or extension ranges under a constant torque load.

flexion and 0.24 seconds in extension. Similarly, measures of knee moment obtained at minimum resistance settings on four hydraulic units fell below the maximum values for the Northwestern device.

These results were obtained with hard-anodized friction disks and a flat, woven asbestos brake-lining material. Other materials, including Nylotron, Celastic, and a laminated brake material were tested in the same way and found to have serious deficiencies. The woven brake lining provided a higher coefficient of friction and more resistance to bending and tension than all other materials tested.

As a result of these tests, the manufacturer improved the anchor bar, adopted the recommended brake material, installed a more resilient extension bumper, and substituted a heavier compression spring. The Northwestern University Disk Friction System now available is an easily adjustable, variable swing-phase control which is simple and relatively inexpensive.

2. *Immediate Post-Operative Prosthesis.* The technique of fitting patients with prostheses immediately after amputation, developed by Dr. Marian Weiss of Poland, has generated a great deal of interest. Several medical centers in this country are employing these procedures on a research basis, and their experiences to date have revealed a need for special prosthetic devices to facilitate fitting immediately after surgery. Two devices designed for this application have been submitted for evaluation. The study included a mechanical analysis of design, materials, and function as well as clinical experience.

The units, one manufactured by the A. J. Hosmer Corporation and the other by the United States Manufacturing Company, are used with a below-knee temporary socket in a procedure in which the stump is cast in plaster-of-Paris bandage and the device is attached to the cast (socket) immediately after surgery (Fig. 9 and 10). Both units provide rotary (long axis), angular, and linear adjustments to facilitate alignment under *dynamic* conditions since it is often difficult to obtain proper alignment with the patient lying down.

In general both the United States Manufacturing and the A. J. Hosmer devices are quite similar in design and function. Both units are attached to the plaster socket by means of straps connected to the upper attachment plate. The attachment straps on the United States Manufacturing device are corrugated while those on the A. J. Hosmer device are smooth. Both provide a quick disconnect to detach the pylon and foot from the socket during bed rest. Their weights are approximately the same and both can accommodate stumps of about the same length. They both provide adequate ranges of adjustment.

Laboratory analyses, however, revealed major differences: (a) linear movement and angular attitude of the pylon are independently adjusted

in the United States Manufacturing device while in the A. J. Hosmer unit, linear displacement of the foot (outset-inset) is accomplished by angular rotation of the pylon and readjustment of the foot attitude; (b) the United States Manufacturing device will maintain a setting under greater loads than the A. J. Hosmer device; and (c) it is easier to make and maintain a given setting accurately with the United States Manufacturing device due to the independence of linear and angular adjustments.

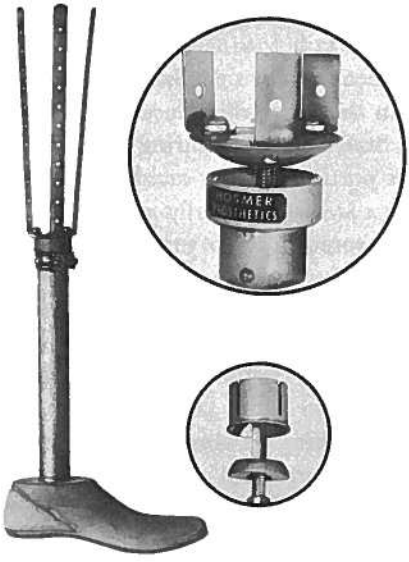

FIGURE 9. A. J. Hosmer Corp. prosthesis designed for early ambulation. Shown in the insets are the spherical sections for adjustment at the knee (above) and adjustment at the ankle (below).

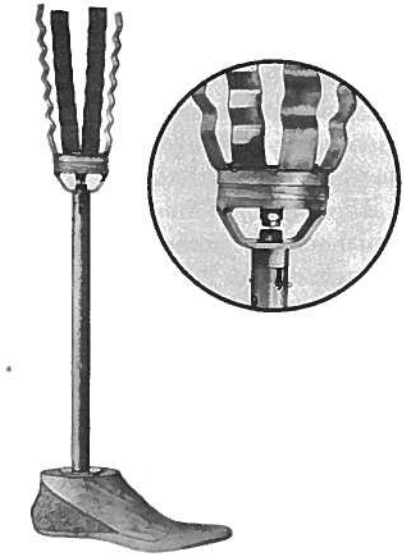

FIGURE 10. United States Manufacturing Company prosthesis designed for early ambulation. Shown in the inset are the mating wedge-shaped plates which rotate to provide angular adjustment at the knee and which also provide linear adjustment by means of slots, not shown in this illustration.

Both of these devices have been used in several clinical facilities across the country, including University of California at San Francisco, Duke University, Prosthetics Research Study associated with the VA Hospital in Seattle, and the Waukesha and Oconomowoc Memorial Hospitals in Wisconsin. Each of these installations reported its initial experiences with respect to difficulties, advantages, and unusual circumstances surrounding the use of both devices. However, their responses, summarized below, represent far more experience with the A. J. Hosmer (Fig. 9) than with the United States Manufacturing (Fig. 10) device.

Attachment of the Device to the Cast. Some difficulty was experienced with the metal adjustment straps of the A. J. Hosmer device. They loosened in the plaster-of-Paris casts. In many instances, it was necessary to rivet the straps to the socket to maintain them in position. This problem did not occur with the United States Manufacturing device, perhaps as a result of the corrugations.

Utility and Convenience in Detaching Components During Bed Rest. In certain instances it was difficult for the patient to detach the components of the A. J. Hosmer unit. The ball in the quick disconnect could not be depressed sufficiently, and on occasion it failed to return after being depressed. It was also difficult for the patient to align the ball and detent. Noise developed due to wear or "slop" in the quick disconnect.

A major difficulty with the United States Manufacturing device was that patients could not detach the lower portion without assistance.

Alignment for Standing and Walking. It was generally reported that initial settings were lost when attempts were made to realign the A. J. Hosmer device. Also reported were difficulties in adjustment and inconvenience due to the need to remove the shoe in order to realign the foot. Range of adjustment, however, was considered adequate.

No problems of this type were reported with the United States Manufacturing device.

Maintenance of Alignment after the Beginning of Ambulation. A need for realignment due to the loosening of the proximal adjustment of the A. J. Hosmer device was reported by only one facility. However, readjustment in the alignment of both devices was more often required to accommodate more active gait, changes in shoe heel height, and as patients' knee stability improved.

Mechanical Design. These units were generally considered an improvement over previous models. However, additional improvements were recommended by the clinical facilities as follows:

United States Manufacturing Company Device

a. Index the adjustment plates to permit reorientation to a previous alignment;
b. Reduce the size of adjustment plates;
c. Serrate the foot attachment plug to prevent involuntary rotation;
d. Modify or redesign the quick disconnect.

A. J. Hosmer Corp. Device

a. Improve attachment straps to prevent loosening in the plaster wrap;
a. Improve the locking mechanism of the spherical adjustment segment;
c. Modify or redesign the quick disconnect;
d. Calibrate the system to permit reorientation to a previous alignment.

This evaluation suggests that while both units have been successfully applied, use of the United States Manufacturing device resulted in fewer problems than the A. J. Hosmer device. However, it should be borne in mind that far fewer models of the United States Manufacturing unit were used and that until more information is available, final conclusions cannot be drawn.

Recommendations for improvement of both devices have been submitted to the manufacturers.

3. *Reference Data on Swing-Control Mechanisms.* A chart is being prepared in which are summarized functional characteristics, properties, and clinical experience data on five fluid-controlled swing-phase units. Designed for distribution as an aid in prescription, it provides information on the following systems: Hydra-Cadence, Henschke Mauch Model B, Dupaco, Hydra-Knee, and the Hydra-Nu-Matic. Details on swing-phase characteristics include: resistance curves, effects of cadence on resistance, adjustment range, and type of temperature compensation. Data on maintenance requirements include: working time to replace the unit, lubrication requirements, and anticipated repairs in terms of type, cost, and frequency. In the category of fitting considerations are special tool needs, adjustment capabilities, shank designs and sizes available, knee block designs and sizes, limits on stump lengths which can be accommodated, foot limitations, availability of installation manuals, alignment require-

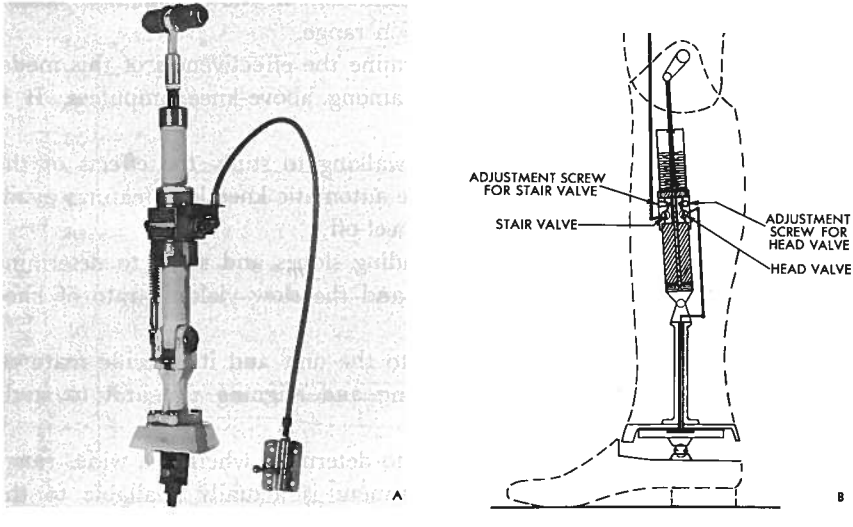

FIGURE 11. Regnell swing- and stance-control unit showing (a) complete device including the manual stair valve control and the universal joint-type foot attachment, and (b) schematic of the hydraulic system showing how the head valve controlling knee flexion is closed by a rod which is actuated by force from the foot during the period between heel contact and mid-stance.

ments, shank length limitations, and finishing needs. Also included are special features and prescription criteria. We hope that this tabulation as well as others we have been developing on mechanical frictional systems will constitute a basis for functional standards for prosthetic knees.

4. Regnell Hydraulic Knee, Model A (Swing and Stance Control). The present Model A which incorporates hydraulic control of both swing and stance phase (Fig. 11a, b) is now being evaluated.

This unit features:

a. Hydraulic control of swing phase.
b. Automatic knee lock in stance phase controlled by force on the heel which closes a head valve and prevents flow of oil and, thus, movement of a piston. (In the process of aligning an above-knee leg, the prosthetist attempts to achieve an optimum condition by trading off stability for desirable swing-phase characteristics. Stability, of course, is essential and in effect, the minimum stability requirement predetermines the location of the knee center. The Regnell unit offers automatic knee lock and, therefore, the location of the knee center presumably is not dependent on stability requirements.)
c. A manually operated "stair valve" which acts as a shunt and permits a flow of oil when the head valve is closed due to pressure on the foot. It allows the knee to flex but at a slow yielding rate which is adjustable to zero.
d. A four-way ankle which provides motion in the dorsiplantar flexion range and in the inversion-eversion range.

This evaluation is designed to determine the effectiveness of this model of the Regnell unit for general use among above-knee amputees. It is based on an analysis of:

a. Amputee performance in level walking to study the effects of the swing-control mechanism and the automatic knee lock features available between heel contact and heel-off.
b. Amputee performance in descending slopes and stairs to determine the effects of the "stair valve" and the slow yielding rate of knee flexion.
c. Subjective reactions of patients to the unit and its specific features.
d. Prosthetic experience in installing and aligning the unit to study time/cost factors.
e. An analysis of alignment factors to determine whether a wider range of anterior-posterior knee adjustment is actually available to the prosthetist and if so, whether the added range contributes to improved amputee performance.

To date, three subjects have been fitted with the device. Preliminary findings indicate that if properly fitted, well aligned, and functioning adequately, the automatic stance-phase control permits patients to walk

down hill without the same degree of shortening of their prosthetic stride length as with other units. In addition, patients responded favorably to the knee stability from heel contact to mid-stance in level walking. However, the foot attachment is inadequate. With relatively little use, the screws by which the foot is attached back out of the lower ankle plate. This results in displacement of the ankle block, increasing noisiness, alteration of alignment, and most important, failure of the automatic knee stability control. This problem has been noted in two of the three prostheses to date; in both cases the condition occurred so frequently that the subjects discarded the units. The manufacturer has been notified of this difficulty.

This study is planned as the first part of a more general program to evaluate other available stance-phase units.

5. *Hydra-Knee.* The Hydra-Knee, a simple hydraulic knee swinging control mechanism, has undergone a series of evaluation studies. Early models of this unit provided insufficient resistance through the first half

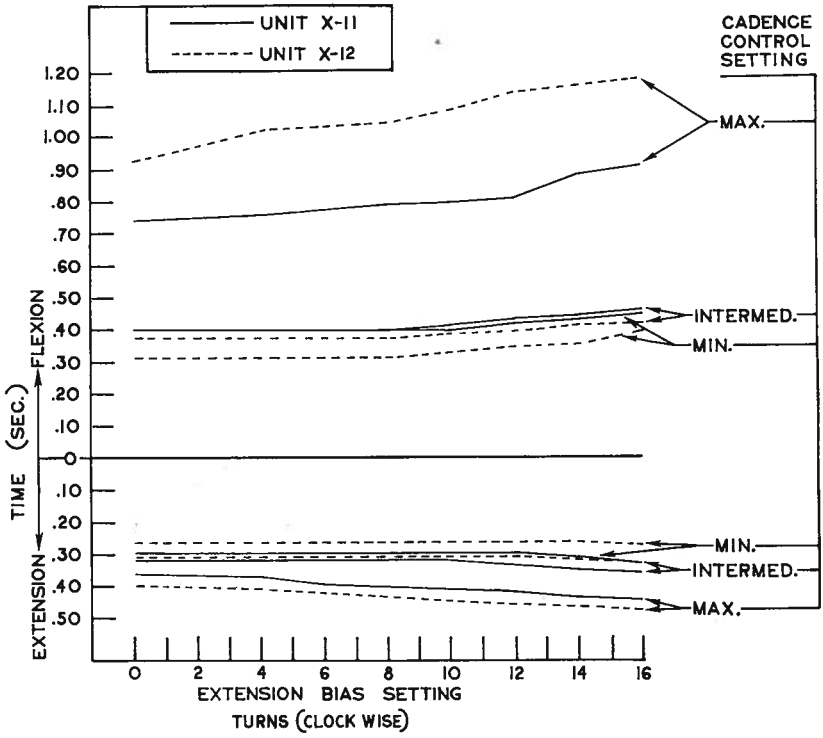

FIGURE 12. Resistance characteristics (in terms of drop-time) of the Hydra-Knee indicating little change in resistance at minimal and intermediate settings and showing the relative independence of extension-bias magnitude.

of the resistance control range and required an excessive number of turns of the resistance control valve for easy adjustment (Fig. 12). Despite these deficiencies, this device provides a useful range of resistance to flexion-extension in the latter portion of the adjustment range. Moreover, it also features a unique adjustable extension-bias system which operates throughout the flexion-extension cycle. As extension bias has little or no effect on flexion-extension resistance in terms of swing time, this device may be particularly useful for subjects with short stumps or for those who walk with very short steps. Suggestions have been made to the manufacturer for reducing the number of adjustment turns and improving resistance over the entire adjustment range as well as changing the knee block bushings and introducing an external extension stop.

II. UPPER-EXTREMITY PROSTHETICS

Evaluation (Components)

1. Externally Powered Components. The development of an externally powered arm prosthesis is an object of great interest to prosthetics research centers in this country and abroad. Several systems* powered by compressed gas and controlled by mechanical linkages have been available, but how these devices compare with well-fitted conventional components is not clear. More information is needed on the specific types of upper-extremity amputees for whom completely powered systems are indicated, and those who can profit most by mixed systems employing one or more externally powered components in an otherwise conventional, body-powered prosthesis. Such systems are commonly referred to as hybrid systems.

We have undertaken a two-phase study whose preliminary step is an analysis in depth of the performance of one bilateral amputee, above-elbow (shoulder-disarticulation) using both conventional and powered prostheses. The first phase is limited to an analysis of the application and effectiveness of the American Institute of Prosthetics Research CO_2 powered components. The second phase includes a systematic evaluation of all externally powered components (gas as well as others) as they become available.

The 40-year-old unemployed subject of this initial investigation is a bilateral arm amputee with an amputation through the neck of the humerus on the right dominant side and a shoulder disarticulation on the left side (Fig. 13a, b), the result of a farming accident in April, 1962. He had never worn a prosthesis.

*The "Heidelberg Arm" of Heidelberg, Germany; the Sierra System of Sierra Engineering, Sierra Madre, California; and the American Institute of Prosthetics Research System of New York, developed by Dr. E. Kiessling.

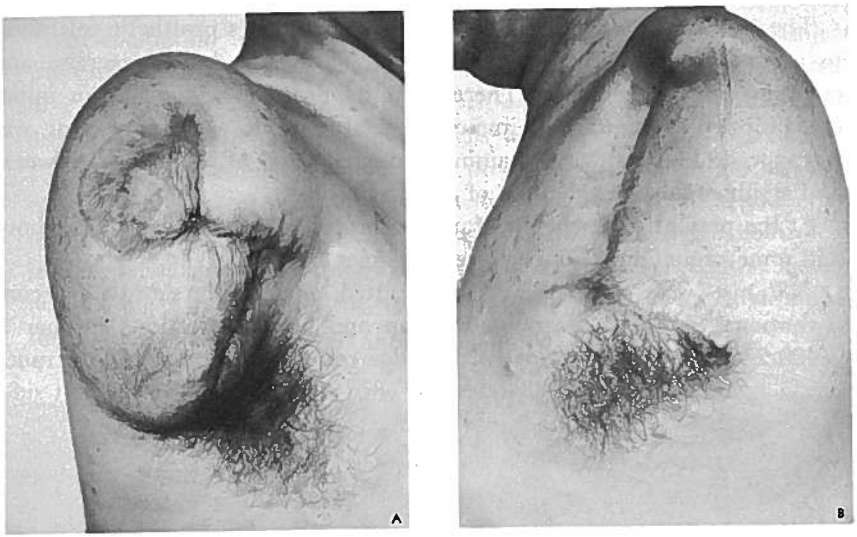

FIGURE 13. Bilateral short above-elbow and shoulder-disarticulation patient showing (a) right above-elbow stump, and (b) residual of left shoulder disarticulation.

The evaluation program involves a comparison of the subject's performance with "optimally" fitted conventional bilateral prostheses and with several American Institute of Prosthetics Research externally powered components. It consists of a series of steps in which, first, the more mobile, dominant right (humeral neck) side is fitted with a succession of powered units. Then, after the full range of available components has been applied to the right side, the left prosthesis will be similarly powered. The procedure includes:

a. Initial fitting with conventional prosthesis on right humeral neck side, shoulder-disarticulation cap on the other side.
b. Fitting of conventional shoulder-disarticulation prosthesis on left side with shoulder abduction joint replacing shoulder cap.
c. Substitution of AIPR CO_2 powered hook and wrist rotation unit for conventional units on the right humeral neck side.
d. Refitting with Münster-type socket and conventional components on the right humeral neck side.
e. Substitution of AIPR CO_2 powered hook and wrist rotation unit.
f. Substitution of a powered elbow for the conventional one on the right humeral neck side.
g. Substitution of a passive shoulder joint (humeral abduction-adduction and flexion-extension) with an externally powered lock coupled to the elbow.
h. Systematic addition of power to the left shoulder-disarticulation prosthesis.

The first step was designed to develop the subject's prosthetic tolerance, to assess his motor ability, and to allow shrinkage due to exercise and compression to take place. Thereafter, the procedure involved an initial evaluation of performance immediately after each new component was introduced, followed by a training-and-use period of at least four weeks and a subsequent reevaluation of performance.

At the present time, the program has proceeded through step (e) above and some preliminary findings are available:

Step a. The subject was initially fitted on the right side with a conventional shoulder-disarticulation cap and humeral section, Hosmer E 400-2 (spring assist) elbow with nudge control, friction wrist, Dorrance 555 hook, and a manually-operated wrist flexion unit. On the left side, he was fitted with a shoulder cap only (Fig. 14a, b).

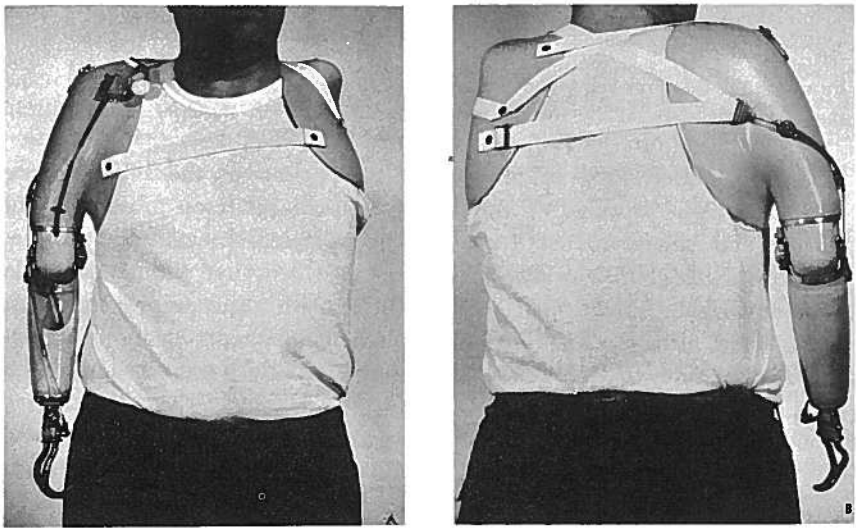

FIGURE 14. Step a: Initial fitting of patient showing (a) a front view of the shoulder cap above-elbow prosthesis on the right side and shoulder cap initially fitted on the left side, and (b) a rear view.

After a 3-day period of training at the VA Prosthetics Center with emphasis on donning and removing the prosthesis and on *controls* practice (i.e., operation of the components), he used the prosthesis for 3 months at home. During this home period, his training continued on a regular basis at the Buffalo, New York VA Hospital emphasizing *use* training (i.e., the performance of self-care activities and other meaningful tasks).

An evaluation after this 3-month period indicated that the subject was quite adept, had a high prosthetic tolerance, and very strong motivation toward prosthetic rehabilitation. In addition, tissue shrinkage had terminated. On this basis, the next steps were undertaken.

Step b. A shoulder-disarticulation prosthesis was added to the left side. It consisted of a shoulder cap with a passive shoulder abduction joint, a Hosmer E 400-2 (spring assist) elbow with nudge control, a friction wrist, and a Northrop-Sierra two-load hook. His control system was modified to include a sheave mounted externally as an extension of the control attachment strap (Fig. 15a, b).

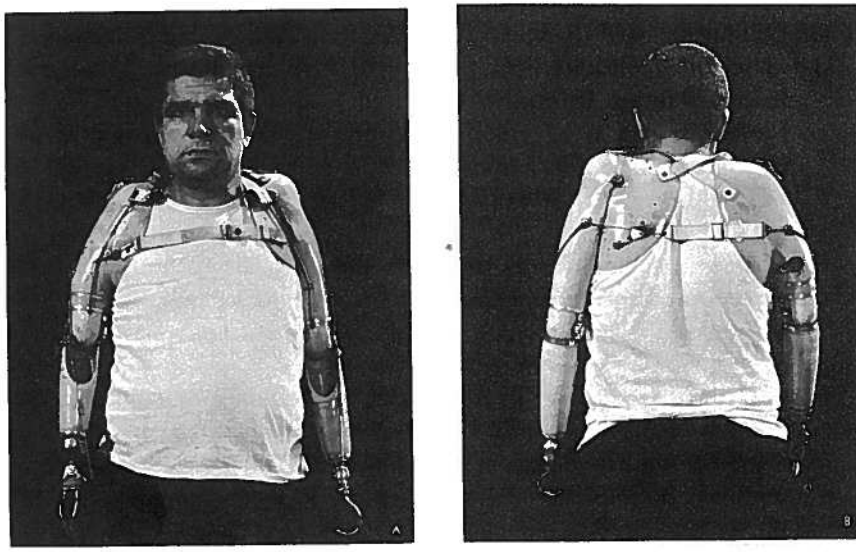

FIGURE 15. Step b: Shown in (a) are bilateral nudge controls for operating elbow locks. The cable attached to cross-chest retention strap is only utilized to operate pressure valve when powered components are substituted. Shown in (b) are bilateral Hosmer E400-2 spring assist elbows and sheave to increase the force to operate right terminal device and prevent inadvertent opening when left prosthesis is being operated. The extra holes in the plastic laminate are used as alternate harness-control points.

After a 3-month period of additional training and use, his performance was reevaluated. At that time, the subject reported that he was not able to accomplish two-handed activities because: (a) he was unable to bring the terminal devices together, and (b) even when adjusted for light load, the left terminal device could not be operated without inadvertently opening the right terminal device. The lack of shoulder motion precluded all dressing and most personal hygiene self-care activities.

Observation in the laboratory tended to bear out the subject's experience. A self-care activity evaluation revealed that he was only able to accomplish those eating activities involving the above-elbow prosthesis on the right side. Nevertheless, his control in positioning the prosthesis and in grading the prehension force used in grasping was excellent. However, he failed to take advantage of pre-positioning the hook and used instead, exaggerated body motions.

With his right above-elbow prosthesis, the subject performed a series of functional tests involving prehension, positioning of the terminal device, and practical activities. The prehension test involved grasping, transporting, and releasing 12 objects of various sizes, shapes, weights, and compressibility. The positioning test involved the removal of 24 similarly shaped wooden blocks placed in a three dimensional array of

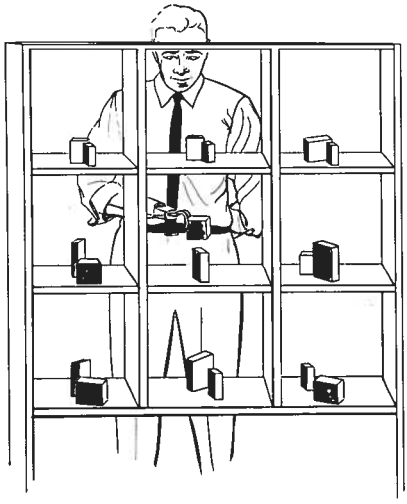

FIGURE 16. Apparatus for testing control of prehension and terminal device positioning in three vertical and four horizontal planes and at varying distances in front of the body.

pigeonholes (Fig. 16). The top row extended beyond his left and right sides at eye level; the second, at approximately breast level; the third, at waist level; and the bottom row, at the mid-thigh level. Objects were placed in the pigeonholes at different depths. Both prehension and positioning tests were scored in terms of the amount of time taken to complete the tests; the number of grasp, release, and compression errors made; and a subjective estimate by the tester of the overall proficiency in terms of normality of appearance. Appearance was graded on a 5-point scale ranging through excellent, good, average, fair, and poor with interpolation by means of plus and minus in each category. The test results are shown in Table 1.

TABLE 1. — *Performance with Body-Powered Prosthesis (Step b)*

Tests	Total time, min.	No. of errors	Appearance
Prehension	3.0	3	good +
Positioning	3.7	1	good +

The practical activity test consisted of seven items selected from a list of 20 common tasks representing both one- and two-handed activities that could be easily tested and retested in the laboratory. These tasks are also practical in the sense that they involve the fundamental motions required for a great number of other tasks. The selected activities included drinking from a glass, combing hair, sharpening a pencil, grasping a briefcase, turning the pages in a book, lighting a match, and picking up packages. Those activities which required more than 4 minutes to complete were scored as failures.

The subject successfully completed five of the seven practical activities and failed to complete two. All of these test results, which compare favorably with those of other bilateral above-elbow shoulder-disarticulation (5) patients, were used as the basis for comparison with all subsequent performances.

In addition to the above tests, the energy cost of completing a second prehension test was determined. The total (net) energy expenditure was 1.088 liters of oxygen. It took the subject 3.7 minutes to complete the task. He committed no errors and had an overall rating of good plus.

Step c. An AIPR CO_2 powered hook and wrist rotation unit was substituted for the conventional components in the subject's right above-elbow prosthesis. A chest expansion strap to control wrist rotation was the only modification made in his basic control system. The control of the terminal device was achieved through humeral flexion: the stump depressed a control valve mounted in the anterior portion of the socket. The power pack (CO_2 manifold) was carried in the back trouser pocket. The left prosthesis was not modified at this time (Fig. 17a, b).

Following a 3-month period of training and wear, the subject was reevaluated. He reported that use of the wrist rotation unit made several activities (brushing teeth, shaving) much more convenient and made others possible (turning door knobs, locking and unlocking doors). The powered terminal device offered more prehension force (approximately 10 lb. vs. 5 lb.) allowing him to bring heavier objects up to his mouth. In addition, the powered unit provided full opening of the terminal device at all angles of forearm flexion. With external power applied

to the right terminal device, the opposite shoulder was no longer involved in providing reaction for forces for its operation. Thus freed, the right prosthesis had a greater range of motion and, in addition, inadvertent opening when the left terminal device was operated was no longer a problem.

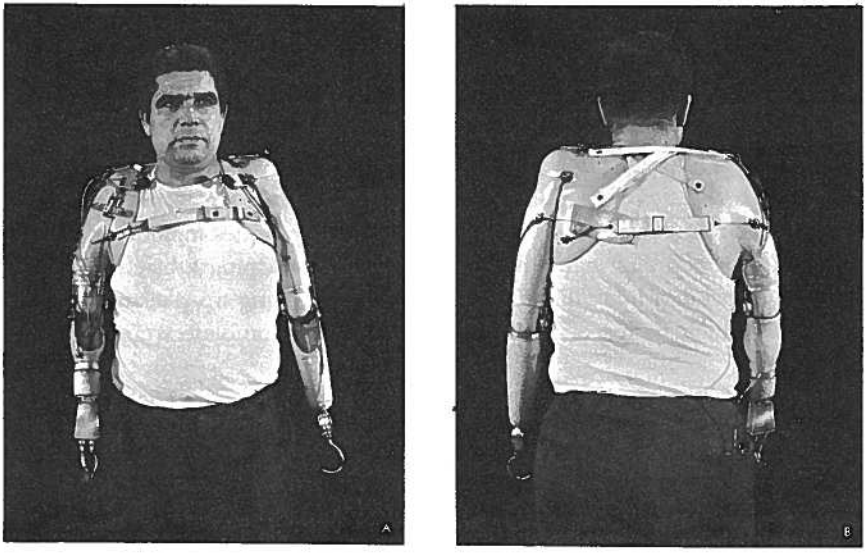

FIGURE 17. Step c: AIPR powered hook and wrist rotation unit on right side. The following control elements are shown in: (a) pressure valve actuated by the short right AE stump to control terminal device operation, chest expansion strap to control wrist rotation, and nudge control to operate the elbow lock. In (b) are compressed gas manifold inserted in the patient's hip pocket and harness arrangement with sheave to flex the elbow by scapula abduction.

Several disadvantages were also reported. Occasionally, he struck the terminal-device control valve accidentally with his stump. Carrying a heavy briefcase would inadvertently actuate the wrist rotation unit, a problem he overcame by completely pronating his hook. The subject was unable to get his terminal device close enough to his body to perform dressing activities.

As observed in the laboratory, the subject was generally slower in accomplishing a given activity because he tended to open his terminal device excessively each time he grasped or released an object. He lacked fine control over the amount of force he used, often distorting compressible objects. He utilized the wrist rotation device almost exclusively to handle door locks and knobs rather than terminal device pre-positioning. The results of the functional tests are presented in Table 2.

TABLE 2. — *Performance with American Institute of Prosthetics Research Powered Hook and Wrist Rotation Unit (Step c)*

Tests	Total time, min.	No. of errors	Appearance
Prehension	4.4	10	Average —
Positioning	7.0	1	Average —

In general, the subject required more time to complete the tests with the powered units, and he committed more errors—notably compression errors. His performance on both tests was rated slightly below average.

Of the seven practical tasks, he performed five successfully and failed in two. He was still unable to do those requiring the use of both prostheses due to limitation of humeral movement and the absence of a wrist flexion device.

The total (net) energy cost of performing a second prehension test was 1.205 liters of oxygen requiring 4.6 minutes to complete the task. During that time he committed 12 errors and his overall performance was rated as poor. Actually, he used more energy (approximately 15 percent) with the powered units but this was attributed primarily to the greater error incidence which increased the time required to perform the task.

Step d. The subject was fitted with a new socket on the right side. This socket, of the Münster type was designed to permit more freedom of motion at the shoulder and increase the range of terminal-device movement. His conventional, body-powered components were replaced in the system, but the nudge control was eliminated and shoulder depression was used to lock the mechanical elbow. The harness was a figure "X" type for bilaterals with elastic straps anchored on the left shoulder cap (Fig. 18a, b).

At the end of a 3-month wear period, the subject was pleased with the system. He was able to get close enough to his body with the terminal device to operate the zipper on his trousers. He preferred shoulder depression to actuate the elbow lock because it was faster and did not require him to turn his head. Also, the new socket and elastic harness allowed him to open the terminal device wider at his mouth than he was able to do with his previous conventional system (2⅝ in. vs. ⅛ in.).

He was again limited to one-handed activities due to inadvertent opening of his right terminal device when operating the left one. He felt that the prehension force was inadequate; he dropped things as he brought them up to his mouth. He was unable to enter his pants pocket

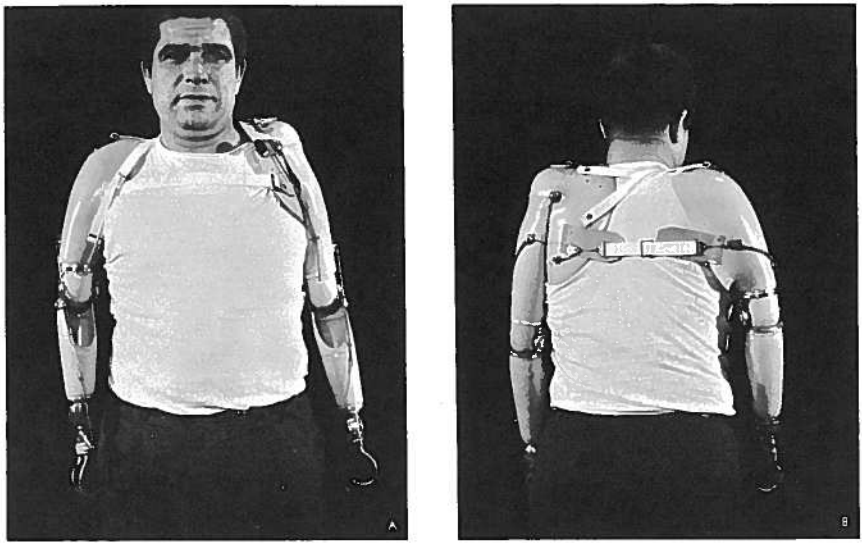

FIGURE 18. Step d: Münster shoulder cap to improve shoulder mobility by elimination of the "wings." Conventional components have been reinstalled to determine the effect of the new socket apart from the components. Shown in (a) is a harness change to permit elbow lock by means of scapula depression. As shown in (b), the harness is otherwise unchanged.

and he had difficulty operating the wrist flexion device as the wrist tended to rotate until the control button was under the control cable.

The subject was quite adept at operating the elbow lock, the terminal device, and in controlling prehension forces. Except for feeding and a few other activities, he still needed help for dressing and toilet tasks.

The laboratory tests, conducted during this visit, are summarized in Table 3.

TABLE 3. — *Performance with Body-Powered Prosthesis with Münster Socket (Step d)*

Tests	Total time, min.	No. of errors	Appearance
Prehension	2.8	1	Excellent —
Positioning	5.4	0	Good +

Performance on the prehension test was better than with the previously used body-powered system (Table 1). He took 1.7 minutes longer to complete the positioning test. This was attributed to the use of the elbow lock

with each grasp and release attempt, whereas previously he had kept the elbow locked and used his body to position the terminal device.

On the practical activities tests his score remained the same—five successfully completed and two failures.

The net energy cost for a second prehension test was a total (net) of 0.992 liters of oxygen. Completion required 3.8 minutes; he committed three errors and his performance was rated as good minus. This was roughly comparable to his previous performance with the conventional prosthesis and shoulder cap system.

Step e. An AIPR CO_2 powered hook and wrist rotation unit was again substituted for the body-powered components of the subjects right prosthesis. The control system remained the same except for the addition of a chest expansion strap to control the wrist rotation unit. Operation of the terminal device was accomplished by placing a control switch in the lateral portion of the socket where *abduction* of the stump actuated the control valve. The sheave on the left side was removed to allow him to utilize that terminal device more fully (Fig. 19a, b).

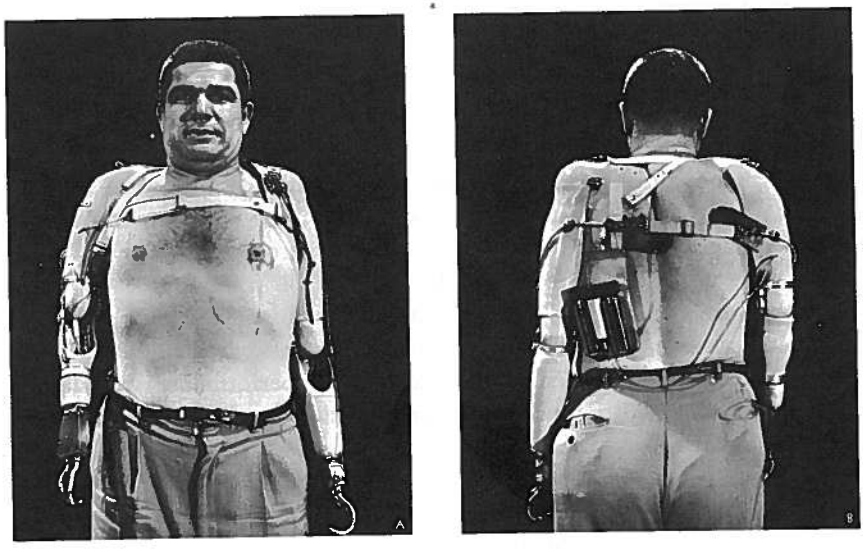

FIGURE 19. Step e: AIPR powered hook and wrist rotation unit and Münster Socket. (a) Terminal device operation is provided by abduction of stump against a pressure valve inserted in lateral wall of socket; wrist rotation is accomplished by a pull valve actuated by a chest expansion strap, and elbow is locked by shoulder depression. Shown in (b) is sling suggested by patient to attach compressed gas manifold to prosthesis permitting him to remove complete system without assistance. With external power, inadvertent operation of terminal device was no longer a problem and sheave was removed.

The subject's initial reactions were that he could operate both terminal devices independently for two-handed activities, and that he could now enter his pants pocket.

However, wrist rotation could not be accomplished when both terminal devices were close to the body since he could not utilize chest expansion with both scapulas abducted. He reported another disadvantage. When the right arm was flexed to 135 deg. the socket tended to rise making operation of the terminal device by abduction of the stump difficult. It was still impossible for the subject to get his terminal device close enough to his body for personal hygiene and dressing activities.

Initial tests performed *prior to the wear period* are reported in Table 4.

TABLE 4. — *Performance with American Institute of Prosthetics Research CO_2 Powered Hook and Wrist Rotation Unit and Münster Socket (Step e)*

Tests	Total time, min.	No. of errors	Appearance
Prehension	3.1	9	Fair
Positioning	6.1	5	Average —

Performance times in the prehension and positioning tests were substantially faster than on previous tests with gas powered components and shoulder cap prosthesis (Step c). However, more errors were made in positioning, a fact attributed to the lack of controls training.

His performance on the practical activities tests included six successfully completed and one failure. He was able to do one practical activity that he previously had not been able to do and his performance was slightly better than that with powered units and shoulder cap.

The net energy cost of the prehension test was 1.27 liters of oxygen during a period of 4.6 minutes. He committed 11 errors and his performance was slightly better than his previous performance with powered units (Step c). However, as the tests in Step e were performed immediately after refitting, improvement may be expected after the scheduled wear and additional training period.

A summary of all the test results is presented in Table 5 for comparative purposes. Early experience reveals that the subject performs primarily as a unilateral amputee and that his proficiency with conventional prostheses is well above average for a patient with high bilateral upper-extremity amputations. The Münster socket significantly increased his range of

motion on the right side by eliminating the restricting walls of the cap. The use of a powered terminal device and wrist rotation unit did not substantially reduce the effort required to use the prosthesis. His performance was somewhat slower and he made more errors with the powered components. However, with training and practice a better appraisal of these matters can be made. The powered components have operated in a highly reliable fashion to date without breakdown.

TABLE 5. — *Summary Table of Test Results (Right Prosthesis)*

Tests	Shoulder-disarticulation cap		Münster socket	
	Body-powered components (Step b)	Gas-powered components (Step c)	Body-powered components (Step d)	Gas-powered* components (Step e)
Prehension				
Total time, min.	3.0	4.4	2.8	3.1
No. of errors	3	10	1	9
Appearance	Good +	Average —	Excellent —	Fair
Positioning				
total time, min.	3.7	7.0	5.4	6.1
No. of errors	1	1	0	5
Appearance	Good +	Average —	Good +	Average —
Practical activities (7 tasks)	5 completed 2 failures	5 completed 2 failures	5 completed 2 failures	6 completed 1 failure
Net energy cost of prehension test	1.088 liters O_2	1.205 liters O_2	.992 liters O_2	1.270 liters O_2
Total time, min.	3.7	4.6	3.8	4.6
No. of errors	0	12	3	11
Appearance	Good +	Poor	Good —	Poor

*These test results were obtained prior to a period of wear and training.

This program is continuing through the additional planned steps including the use of powered components on the left side.

Plans for the second phase in which a larger sample of patients will be fitted with powered components are being completed.

The purpose of the second phase is to compare the effects of externally powered and body-powered components on the performance of selected high level bilateral amputees and to develop prescription criteria for externally powered upper-extremity systems.

2. *Child Amputee Prosthetic Project Terminal Device.* A novel design for a terminal device is currently being evaluated simultaneously at the

VA Prosthetics Center and several other laboratories. Designed by Mr. Carl Sumida of the University of California-Los Angeles Child Amputee Prosthetic Project, this terminal device has no "thumb" or projecting operating rod and is actuated by a push-pull rod passing through a standard ½″ x 20 mounting stud permitting 360 deg. of rotation without affecting control (Fig. 20).

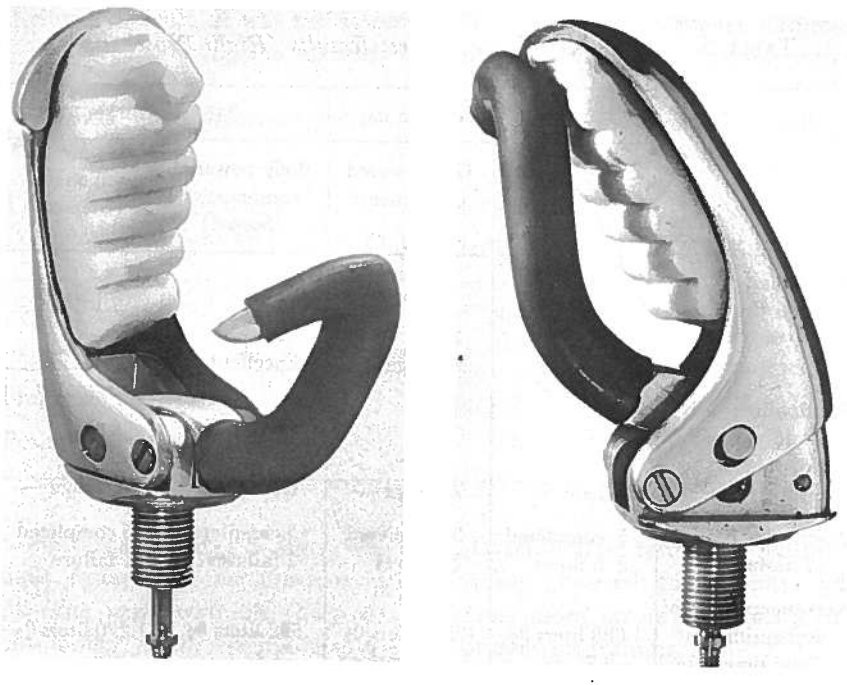

FIGURE 20. Sumida experimental terminal device developed at the Child Amputee Prosthetic Project, University of California at Los Angeles showing the central actuation rod and one of several experimental inserts.

The novel features include a single broad finger and a conventionally shaped finger both of which move in opposition. The larger finger is designed to provide a degree of "palmar prehension" in addition to the functions of conventional split-hook terminal devices of finger tip prehension and grasp. The "palmar" surface contains a resilient insert whose function is to offer a larger contact surface when depressed by irregularly shaped objects.

The evaluation program is aimed at assessing the functional utility of the concept and determining the most applicable material, shape, and resiliency of the insert.

III. LOWER-EXTREMITY ORTHOTICS

A. Development

1. Below-Knee Weight-Bearing Brace. During the past four years, approximately 80 veterans have been fitted with this device throughout the Veterans Administration. On a clinical basis, we consider this treatment successful because patients continue to wear and prefer the below-knee weight-bearing brace, returning at regular intervals for replacement. The Orthotic Study Group of New York University is now independently evaluating these VA cases in an attempt to determine the general utility of the brace.

Until recently, all the patients fitted with this brace at the VA Prosthetics Center were unilateral brace wearers. However, in July 1964, a veteran was fitted bilaterally and the results were highly successful, if not dramatic. Complete details on this experience are expected to be published in the Spring 1965 issue of Artificial Limbs. The report that follows constitutes a summary.

The patient, first examined at the VA Prosthetics Center in July 1964, was 46 years old, 5 ft. 4 in. tall and weighed 129 lb. with braces. Although

FIGURE 21. Bilateral below-knee weight-bearing braces worn by subject.

unemployed, he was active in working around his home and the community. Both legs were deformed as a result of an explosion in 1944 during World War II. The severe damage included incomplete union of the right tibia with bone spurs, shortening of the left leg, complete ankylosis of both ankles, and loss of muscle power in the thighs due to disease atrophy. He experienced severe pain when he bore more than a small fraction of his body weight on his feet. For 17 years he walked with the aid of two conventional below-knee braces and two axillary crutches using the swing-through gait.

Clinical evaluation resulted in prescription of bilateral below-knee weight-bearing braces with unitary sidebar-stirrups providing no ankle motion. Also prescribed were orthopedic shoes with high quarters, soft inner molds, soft cushion heels, and metatarsal rocker bars. During the fitting process, it was noted that he could walk short distances without the aid of crutches.

The patient had been wearing the newly prescribed below-knee weight-bearing braces (Fig. 21) approximately one month when he reported that he walked for about 3 to 5 hours per day with the assistance of one axillary crutch; he wore the braces 12 hours per day. He had used the below-

FIGURE 22. Use of cork build-ups on soles of shoes to duplicate alignment and elevation of original stirrup.

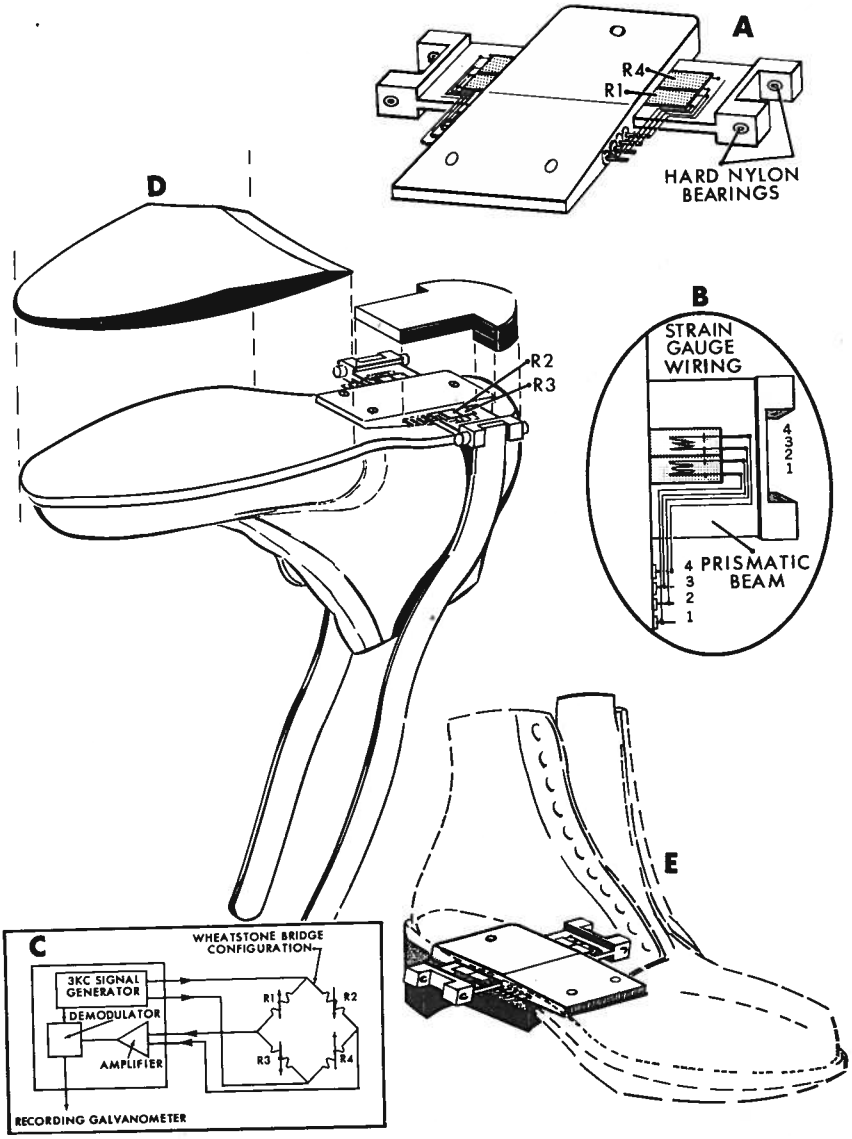

FIGURE 23. Schematic of stirrup with instrumented strain gages. Shown in A are the tongue and cantilever beam on which are mounted strain gages (R1 and R4) connected to terminal posts. Inset B is a wiring diagram of connections between strain gages and posts. C is a complete circuit diagram showing direction of resistance change in bridge circuit excited by a 3 kc signal generator. Amplified and fed through a demodulator, outputs are recorded. D and E describe method of attachment to simulate conventional stirrup and side bar configuration.

knee braces without crutches during the first 3 to 5 days after delivery, walking for relatively long distances with intervening rest periods.

This improvement generated several significant questions:

a. To what extent was the below-knee brace unweighting the foot during the period from heel contact to toe-off?
b. What was the character of the gait pattern without crutches in relation to standards of normal locomotion?
c. What were the patient's reactions and experiences?

To answer these questions, the subject's performance was evaluated in a series of biomechanical analyses after approximately four months of wear of the below-knee braces. The forces transmitted to the ground and the forces in the side bars were measured to determine the extent to which the braces "unweighted" the legs. A specially designed stirrup instrumented with strain gages was used for this purpose (Fig. 22 and 23). A second aim was to describe the character of the gait exhibited by the patient in relation to normal locomotion patterns. Following the test, patient reactions and experiences were obtained by interview.

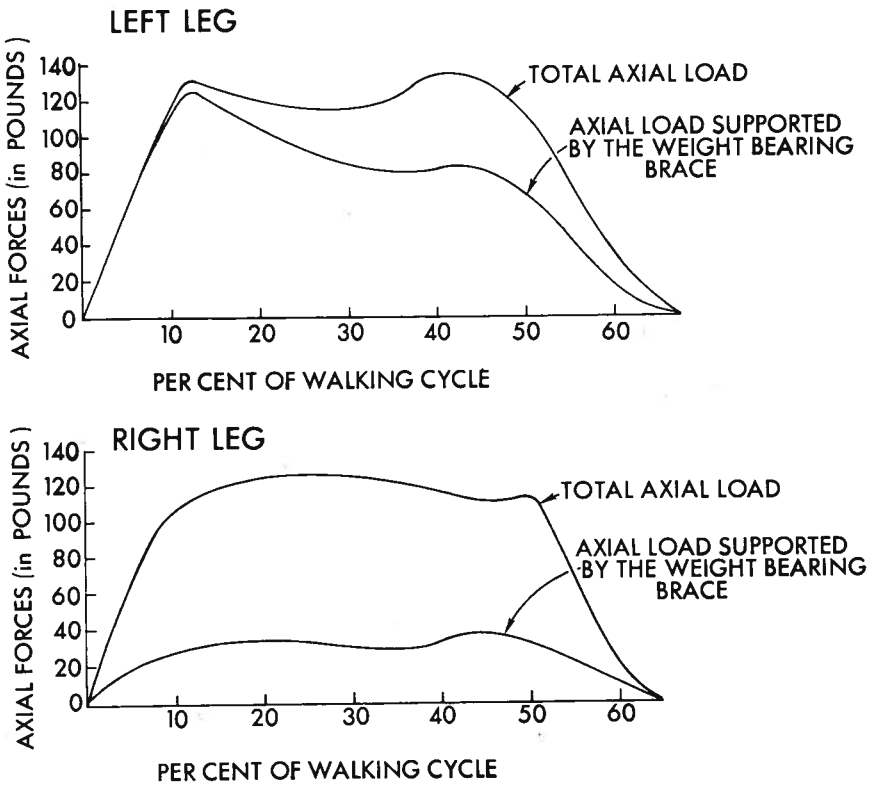

FIGURE 24. Distribution of forces applied to the ground through the tibia and brace side bars.

Reference to Figure 24 indicates that the brace on the left leg carried approximately 96 percent of the total axial load applied to the ground during the first fifth of the stance-swing cycle, a point well past foot-flat and within the region of mid-stance. Thereafter, the distribution of axial load changed with the brace bearing decreasing loads through the next fifth of the cycle (just past heel-off) where it bore about 60 percent of the axial load. Sixty percent of the declining axial load was carried by the brace for the next fifth of the cycle after which the swing-phase portion of the cycle began. The brace on the right leg continuously supported approximately 25 percent of the axial load during the entire stance phase of the stance-swing cycle.

From the foregoing, it can be seen that the subject was capable of tolerating approximately 75 percent of the load on his right foot but could only tolerate loads ranging from 4 to 40 percent on his left leg. This distribution of loads (body weight plus inertial forces) between the foot and the brace permitted him to walk comfortably and in a reasonably normal manner.

Adjusting the load distribution between the foot and the brace can be accomplished by changing the length of the sidebars or by changing the resiliency or the depth of the innersole.

Motion picture and cyclographic studies revealed that the subject walked slower and more deliberately than normal non-brace-wearing subjects and exhibited the following gait deviations: (a) moderate abduction, (b) slight side sway, and (c) excessive knee flexion during stance phase.

Neither the abduction nor the side sway was of a magnitude to detract substantially from the normality of his appearance. The excessive knee flexion was quite noticeable. However, in the absence of ankle-joint function and the inability to plantar flex, this adaptation permitted him to achieve a smoother transition from heel contact to foot flat.

More detailed analysis of the temporal factors of his gait revealed that he attained optimum symmetry at approximately 83 steps per minute with a swing-to-stance time ratio of 0.562 as opposed to 0.552 for a normal person at the same cadence.

The patient expressed his opinions and reactions to the braces in terms of comfort, amount of use, and the range of activities in which he engaged. These responses indicated that use of the below-knee weight-bearing braces substantially increased his mobility by permitting him to discard the crutches. As a result, his entire range of activities broadened and his sense of independence improved.

2. *Single-Bar Brace (Unibar Brace).* Seventeen patients have been fitted during the past several years with the single-bar brace (6). Included in the group are eight patients fitted with the single-bar *leg* (below-knee) brace of which four are unilateral and four bilateral. Nine patients are

fitted with the *leg-thigh* (above-knee) brace of which four are unilateral and five bilateral. These patients have been wearing single-bar braces for periods ranging from 6 months to 2½ years as shown in Table 6.

TABLE 6. — *Numbers of Patients Wearing Unibar Brace According to Period of Wear*

Wear (in months)	Type of unibar brace	
	Leg	Leg-thigh
24-30	0	4
12-23	5	4
6-11	3	1

Studies are still in progress to determine the optimum material and cross section for use in the single-bar, leg-thigh brace. The original modified "H" or "I" section (AL.2024-T3) reached its elastic limit after application of a bending moment. Therefore, it will not deform permanently at loads below this value. Previous studies of the forces applied to brace side bars during gait revealed that in a below-knee weight-bearing brace, for example, the maximum bending moment was approximately 360 in.-lb., less than half the resistance value of the original "I" section. With the view that a safety factor of 2 or 2.5 is desirable, it would appear that the strength afforded by the present section should not be substantially reduced. However, several other cross sectional configurations and materials are being studied.

B. Evaluation (Components and Techniques)

1. Custom versus Stock Orthopedic Shoes. Most physicians and orthotists recommend the use of orthopedic shoes in cases requiring shoe modifications or braces. However, there are two vastly different types of shoes labeled "orthopedic," the *custom* and the *stock* orthopedic shoe. The latter usually contains a steel shank and in certain instances it also includes a long medial counter and Thomas heel. It does not include reinforcement required to prevent "breaking" of the sole at undesirable points and to prevent lateral bulging of the uppers.

Stock orthopedic shoes are frequently prescribed or selected by patients and in this connection, cost is probably a significant if not decisive factor since typical costs for stock orthopedic shoes average half, or less than half, the cost of custom orthopedic shoes.

Stock orthopedic shoes are also selected because the patient may seem initially to walk in much the same manner as when wearing new custom

orthopedic shoes. What cannot be seen is the quick deterioration and shorter life of the stock orthopedic shoe and the longer life and functional value of the custom orthopedic shoe.

A recent experience illustrates these points. A young man with congenital deformities of the feet for whom orthopedic shoes had been prescribed was tested in our laboratory.* He was considered an excellent subject for this type of analysis because of the remarkable adaptations he had made to his deformities.

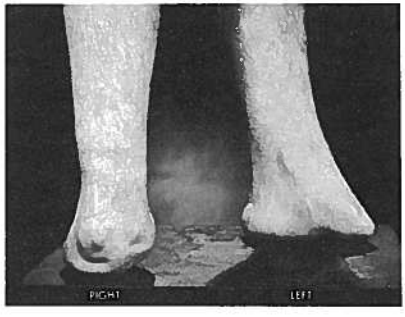

FIGURE 25. Bilateral congenital "amputation" with tarsal structures on the left side and a condition similar to a Syme's stump on the right side.

The subject, a 19-year-old congenital amputee with partial hands and feet (Fig. 25) was wearing previously prescribed stock orthopedic shoes with steel shanks as the only special feature. Added externally were sole and heel extensions. After approximately 12 months of wear, the shoes exhibited a severe "break" in the tarsal region of the right and another break though less severe, on the left shoe. The lateral walls of both shoes bulged excessively resulting in permanent deformation and reduction of support (Fig. 26). The short steel shank protruded through the sole at the point corresponding to the "break" and the wear of the soles revealed a pattern of little or no support anterior to the shank which terminated at a point corresponding to the tarso-metatarsal joint line.

He was fitted with shoes custom-shaped to lasts of his feet (Fig. 27). The shoes were reinforced with long, flat, steel springs and steel shanks installed between inner and outer soles to increase the resistance to dorsiflexion after mid-stance and to shift the "toe-break" farther anteriorly. They also featured stiff, high, long counters and a wider heel base with a reversed Thomas heel on the right shoe to increase lateral support. An

*Complete details on this study are expected to appear in the Spring 1965 issue of Artificial Limbs. The subject, who also had hand deformities, was referred to the VA Prosthetics Center by Mr. J. Donald Smyth, Rehabilitation Counselor, Maryland State Department of Education, Division of Vocational Rehabilitation, Baltimore 1, Maryland. A prescription was suggested to Mr. Smyth who arranged for fabrication by a commercial custom shoe maker.

inside cork extension was prescribed to accommodate leg shortening. After four months of use, the wear pattern of the soles indicated that the patient was receiving support, that is, resistance to dorsi-flexion or "shoe-break," projected out as far as the toe.

FIGURE 26. Previously worn stock orthopedic shoes showing extensive deformation.

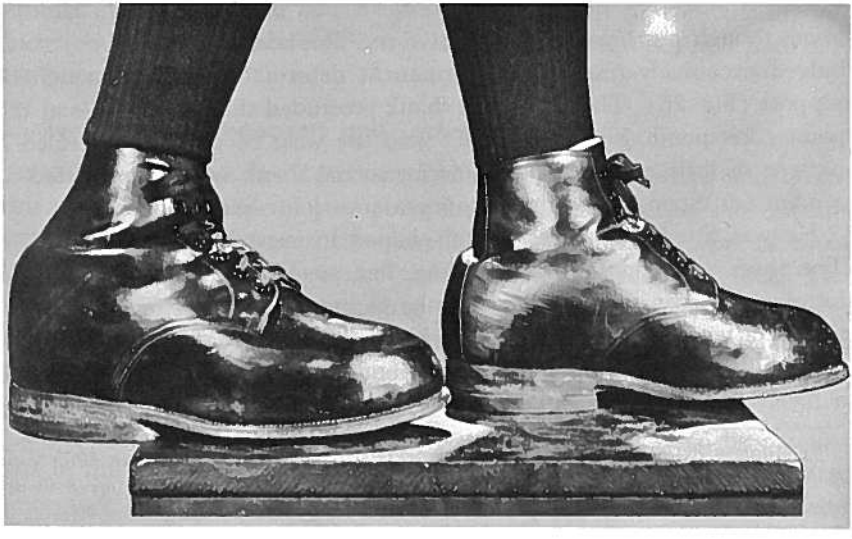

FIGURE 27. Custom orthopedic shoes showing very little deformation after four months of wear.

A biomechanical analysis showed that flexion of the left knee during early stance phase was reduced substantially with the custom shoes. This was attributed to the increased support provided by the custom shoes in the tarso-metatarsal region with a consequent reduction of the "drop off" on the right leg during late stance. As a result of the excessive "drop off" due to the "break" of the stock orthopedic shoes, the hip remained at a lower elevation than it would otherwise have attained and made necessary additional compensatory flexion of the left knee. Reducing the "drop off" maintained the hip at a higher elevation and made these compensations unnecessary.

Also demonstrated was an ability to maintain higher fractions of his body weight on the supporting foot after heel-off which resulted in a more equitable application of body weight forces to the ground.

The subject stated unequivocally that the custom orthopedic shoes were far superior to the stock shoes that he had previously worn. They were more comfortable, they provided better support, and the inside buildup was more cosmetically desirable. The subject wore the custom shoes home and refused to take the stock shoes with him, discarding them on the spot.

There is very little question of the superiority of custom orthopedic shoes over presently available stock orthopedic shoes just as custom prosthetic sockets are superior to "stock" sockets. Even in the case described above, when superficially the need might be considered minimal, clear advantages were provided. On this functional basis alone, preference should go to custom orthopedic shoes. Further study of the life expectancy of custom and stock orthopedic shoes should serve objectively to clarify where real economy in this matter lies.

2. *Arch Supports Formed Dynamically.* In recent years, there has been an increasing interest in the molding of appliances for the foot under dynamic conditions. This type of molding is generally used for forefoot imbalance to reduce the weight applied to plantar excrescences. At the same time, this type of mold might support the long arch. Alignment of the rearfoot, however, requires other means.

We are currently evaluating the materials and methods developed by Army Medical Biomechanical Research Laboratory for the purpose of fitting patients with arch supports expeditiously under dynamic conditions, that is, while the patient walks. The molding process develops an impression in Silastic of the plantar area of the foot at the metatarsal-phalangeal joints. A crest is formed which fills the void between the heads of the metatarsals and the first phalangeal joints permitting the retracted toes to react against a solid surface. The crest also provides an additional weight-bearing area, relieving the pressure on the metatarsal heads and producing functional extension of the toes.

However, successful application of the AMBRL method requires that shoes be comparatively new, of proper last, and fitted with an undeformed innersole to avoid distorting the invested materials. An excessive amount of molding material tends to accumulate at the toe end of the innermold, and modification may be necessary to prevent excessive dorsi-flexion pressure on the toes during rollover and toe-off. Sanding or buffing leaves the Silastic without a binding agent making it doubly essential to start with good quality new shoes and to avoid modifications. When longitudinal or metatarsal arches are required, they must be bonded individually to the innermold.

Our investigation is being continued with a view to compare the relative merits of all the currently used types of molded innersoles and arch supports.

IV. ORTHOPEDIC AIDS

Evaluation (Components)

1. Lightweight Wheelchairs. Two improved models of a lightweight wheelchair previously described in Bulletin of Prosthetics Research 10-1 (7), were evaluated to determine the effectiveness of modifications made by the manufacturer as a result of earlier testing. These models were improved by: (a) the addition of rubber bumpers, (b) a heavier cross brace of 20-gage chrome alloy steel tubing, and (c) deeper anodized aluminum hand rims and wheels to prevent staining. In one of the models, the large drive wheel axles were offset posteriorly to improve stability (Fig. 28). The test program included a check on compliance with specifications, and studies of durability, operational force requirements, stability, and patient reactions obtained during clinical use.

In an effort to achieve lightness, the manufacturer has used materials, methods, and dimensions that deviate from current VA specifications for wheelchairs. The long term effect of these modifications on the life expectancy of the chair remains to be determined. However, in laboratory tests both models satisfactorily completed 100 cycles of simulated curb descents with negligible changes in the deflection of the crosspiece, a site of failure in the previous model.

This newer model required somewhat higher forces to start than the original model, but they were about the same as the forces needed to start a conventional chair. Wheelchair users generally commented favorably on the operation on level ground, and particularly when "hopping" up curbs. The light weight was found to be a decided advantage over conventional models in getting the chair in and out of cars.

The new models did not display the instability noted in the earlier version. In the "offset" model the center of gravity was shifted substan-

FIGURE 28. A special model of the lightweight wheelchair showing the offset drive wheel axles.

tially forward and provided highly increased resistance to the reaction forces of vigorous stroking which tended to topple the earlier models. In the non-offset model, the additional weight of the heavier frame sections appears to have had the effect of maintaining the center of gravity at a lower level with a passenger aboard thereby increasing its resistance to toppling forces.

The lightweight wheelchair may not be indicated for general use outdoors where the narrower, caster tires have a tendency to sink into soft ground, or for heavy (over 175 lb.) patients due to the relatively narrow seat and the light construction. The more stable "offset" model is desirable where increased stability is desired.

2. *Prototype Stair Climbing Wheelchair.* A great deal of interest has been expressed from many quarters in the development of a wheelchair capable of ascending and descending a flight of stairs. This capability is

seen as a solution to problems faced by wheelchair users in entering public buildings and homes without ramps or other special modifications. However, the design of a device to meet these needs must also meet the mobility needs of conventional wheelchair usage. This problem has generated sufficient interest to have become the subject of cash prizes awarded by the National Inventors Council, United States Department of Commerce, for feasible designs.

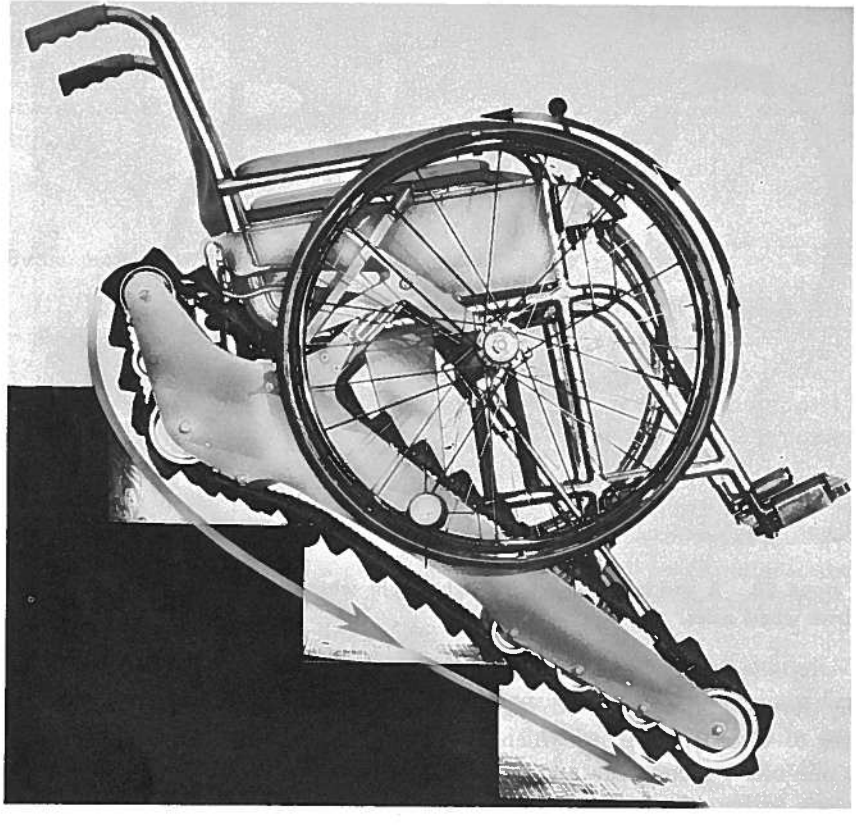

FIGURE 29. The Project Progress stair climbing wheelchair. Arrows indicate direction in which stair climbing tracks move when ascending stairs. Note the semi-reclining attitude of the chair during this mode of operation.

Although many designs (drawings) have been studied in this laboratory during this period, the first prototype to be submitted has been evaluated (Fig. 29). This device consists essentially of two main sections: (a) a conventional heavy duty Hollywood wheelchair, with (b) a stair climbing assembly consisting of two cleated rubber tracks mounted on idler and drive wheels.

In this configuration it is designed to operate on level ground as a conventional wheelchair carrying the stair climbing assembly beneath the frame. For stair ascent and descent, the climbing tracks are lowered to the ground and the conventional driving wheels, linked by a system of levers and cams, are raised off the ground; at the same time the seat reclines. Once in position, the tracks are operated by means of the conventional drive wheels through these linkages.

To climb stairs, the patient backs the chair against the first step until a rubber cleat catches the top of the step. He rotates the drive wheel which causes the track to climb the step, slipping to readjust itself every time the *one* cleat that is supporting the load reaches the end of its travel and another load-bearing cleat becomes engaged. At the top of the flight of stairs, the chair tilts backward and the track slips until a second cleat engages the top step. The chair pivots about the top step until the track is flat on the landing. The patient drives the chair clear of the stairs and repositions the seat to a level attitude.

Evaluation of this prototype model revealed a series of grave shortcomings as regards both mechanical function and the functional hypotheses underlying the design. In view of the serious mechanical deficiencies, suggestions were made to the developer for improvement and for extensive redesign. The most serious shortcomings were related to the hydraulic system, the mechanism for suspending the stair-climbing assembly, the absence of safety features, and the high energy requirements.

The precise patient-need underlying this design is not clear. A device which employs body power for conventional wheelchair use on level ground and as a stair climbing device is of doubtful value. The power required to raise a 100-lb. chair up a flight of stairs in a reasonable time, with a 150-lb. man aboard is very high. For example, if the vertical height of the stairway is 10 feet, the required work is approximately 2,500 ft.-lb. at 100 percent efficiency, or as is more likely, at approximately 15 percent efficiency, 17,000 ft.-lb. This amount of work is equivalent to the work done by a normal person (30 percent efficiency) walking up a flight of stairs with a vertical rise of approximately 40 ft., or more than 4 full flights of stairs. The wheelchair user is again penalized in operation on a level floor, being required to stop and start the device with the added weight of the stair climbing gear which accounts for approximately half the total weight.

Also to be considered is the utility of such a device indoors in the home where other means of negotiating stairs are available, e.g., lifts, assistance by others, ramps. For ordinary outdoor use, climbing both curbs and stairs must be considered. Paraplegics with no involvement of the upper extremities rarely have difficulty in managing curbs with conventional wheelchairs. Complete quadriplegics should not be required to travel in

streets unaccompanied by another person who can offer assistance in curb climbing as well as in level movement. The most significant area of need seems to be for the otherwise capable patient; i.e., paraplegics who may have occasion to travel outdoors in a wheelchair and perhaps to enter public or other unmodified buildings. A flight of stairs is an insurmountable barrier for this type of patient without the assistance which he normally does not require.

The requirements for such a device should include:

a. Operation on level ground in much the same fashion as current conventional wheelchairs.
b. Easy collapsibility for loading and unloading from an automobile by the patient himself, implying a total weight not significantly in excess of conventional wheelchair weights (approximately 50 lb.).
c. The stair climbing mode should not require energy inputs greatly in excess of level ground operational requirements (approximately 50 ft. lb. per arm per stroke).

3. *Wheelchair Propulsion Bars.* A device designed to increase the mechanical advantage in propelling a wheelchair was evaluated (Fig. 30). It consists essentially of two steel bars which pivot independently on the axle housing between the wheels and the frame of the wheelchair. Mounted on the upper end of each bar is a bevelled dog which bears on the tire surface and transmits the driving force to the wheels. Also provided are two adapter rods which make it possible to reverse the direction of arm

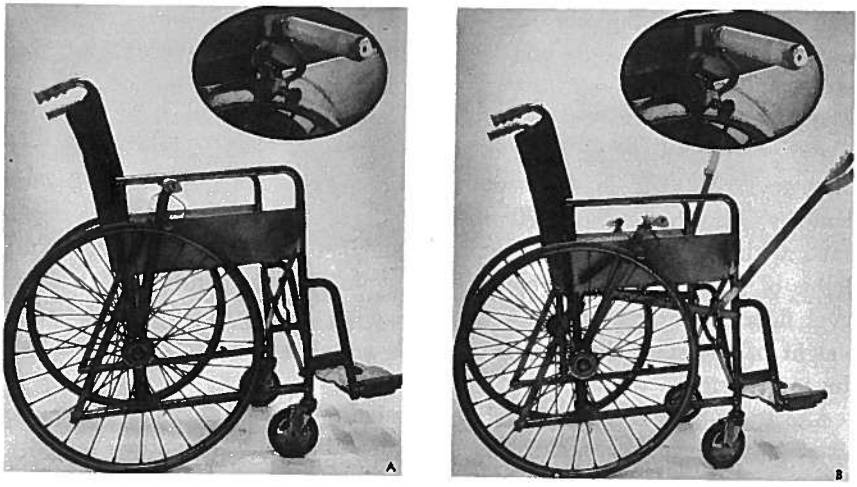

FIGURE 30. Propulsion bars fitted to conventional wheelchair. Shown in (a) are the operating handles and the spring-loaded dogs which drive the wheels. Shown in (b) are the adapted rods which reverse the drive direction.

movement required to propel the chair, i.e., elbow flexion or elbow extension.

The concept of increasing the mechanical advantage in propelling a wheelchair and positioning the hands more conveniently to stroke the chair has much merit and it could be of great service to a particular group of patients. However, this design was found to be inadequate as regards materials and mechanical construction, and unsafe due to a poor braking system that did not stabilize the chair adequately for patient entry and exit.

4. *Adapter for Improvement of Wheelchair Safety.* This device consists of a bracket with a caster wheel mounted under an extension board attached to the framework in the front of a wheelchair. Its purpose is to provide additional stability in getting in and out of the wheelchair for patients who require the use of an extension board. It acts as an anti-tipping feature by extending the base of support and accommodating the forward shift of the vertical projection of the center of gravity when the patient's knees are extended. This device appears to be potentially useful for hospitalized wheelchair patients with legs immobilized in knee extension. However, caution is recommended for use on slopes greater than 5 deg. since, in the absence of a means to retract the caster, the wheelchair becomes unstable.

5. *Wheelchair Specifications.* Existing VA specifications, established a number of years ago, for folding type wheelchairs do not provide for the increased use of new and lighter metals, synthetic materials, and improved designs. We are reevaluating current specifications with a view toward recasting them in a form that recognizes recent developments. In this connection, we are attempting to translate as many of the requirements as possible into *functional* performance standards. Based extensively on patient needs, criteria of this nature are more meaningful and do not unnecessarily restrict creativity in design or use of materials.

6. *Special Crutch Designs.* Two types of an axillary crutch for unilateral use that can be concealed beneath the user's clothing are being evaluated. The lower end fits into a bracket clamped to the patient's heel. One type of crutch folds to facilitate sitting. This is accomplished by pulling up on the crutch handle. When the standing position is assumed by the patient, the crutch automatically locks in a normal vertical position for bearing weight (Fig. 31).

Several basic problems were noted in the concept underlying the design of these prototypes. Clinical experience indicated that in their present configuration both models were capable of very limited vertical adjustments. Normal adult axillary height ranges from 43 to 62.5 in. with an average of 52.5 in. (8). With adjustability limited to three specific lengths, this crutch can accommodate only those patients who measure 48.5 to 50.5

in. from the axilla to the floor, considerably less than average size. Moreover, the points at which the folding crutch flexes at the hinged joints do not match the joint locations for many taller patients.

These crutches were not adaptable for any of the four normally taught crutch gaits, limiting the user to an atypical pattern that involved full, or partial weight bearing on the affected side. Medically, this is undesirable as excessive pressures under the axilla could lead to crutch paralysis.

More significantly, these crutches present a potential safety hazard in that the patient is unable to discard them if he should fall, a problem which is magnified if the crutch is worn under the clothing, as suggested by the developer.

FIGURE 31. The folding crutch in extended and folded positions. At the bottom is the heel attachment clip with receptacle for the crutch.

FIGURE 32. Battery powered, telescopic Hydra-Crutch in extended and retracted positions.

7. *Hydra-Crutch.* A pair of electrically powered (battery) hydraulic crutches were submitted for evaluation by FAR-PRIT Associates, Inc. of Greenville, Texas (Fig. 32). This crutch represented by the developer as a prototype, consists of a telescoping steel tube sealed at the ends acting as a piston inside a second tube, a two-piece Sacrolic auxilliary support, a two-phase motor driven by two nickel-cadmium batteries, and a hydraulic lift system. They are designed to lift and lower patients from and into chairs and to serve as crutches for ambulation.

Combining these two functions in a single pair of crutches is a novel conception and this device may be a substantial aid to crutch users who have considerable difficulty in rising from a chair and in sitting down, as for example, patients with arthritis, multiple sclerosis, or other generally debilitating diseases.

The test procedures were suspended temporarily due to failure of the hydraulic system. This may have represented a maintenance rather than a design problem, since both crutches had been used extensively for demonstration purposes. At the developer's request, these units were returned for renovation or replacement. The evaluation of these crutches will be continued as the refurbished or new models become available.

V. TESTING

A. Specifications Compliance

During this period the annual test program for SACH feet was completed and other tests of compliance were applied to lumbosacral corset samples and stump socks.

1. *SACH Feet*
   a. Annual Compliance Tests. Current production models of SACH feet were submitted by six manufacturers for evaluation to determine acceptability for VA contracts. Of this number, the products of two manufacturers were rated as highly acceptable and that of a third, as acceptable. The feet submitted by the other three manufacturers were not acceptable in that they failed to meet the functional standards set forth in the Specifications No. VAPC-L-6401-1. The primary deficiencies in this connection were excessive toe curl and improper location of the toe-break. In a few instances, excessive toe deflection under load indicated that they were too "soft."
   b. SACH Feet Development. One model of a Type II (external keel) SACH foot evaluated during this period failed to meet specifications. The deficiencies included inadequate resistance to flexion of the toe, evidence of poor workmanship (notches, rough edges, poor finish), and an unacceptable laminated heel.

Two models of a Type I SACH foot presently under development were also evaluated. One model was of conventional design and apart from an excessive resistance to toe flexion at loads above 65 lb., was found to be within acceptable functional limits. The other model appeared to be functionally adequate as regards compliance with the specifications. However, it is of an unconventional design in that it is attached to the shank by gluing an extended "ankle" portion to the shank. No attachment bolt is provided. Clinical application studies were recommended.

2. *Lumbosacral Corset Material.* In view of the rather wide variations in construction and knit geometry of the material used in lumbosacral corsets by different manufacturers, we are developing a comprehensive compliance testing program similar in purpose to our SACH foot and stump sock programs. During this period, the products of two manufacturers were tested for shrinkage due to laundering.

A VA supplier submitted several coutil samples used in the production of appliances. Tests were conducted in accordance with Federal Specification CCC-T-191b, Part 5550.2 (DOD), September 17, 1963, *Shrinkage in Laundering, Cotton, Linen, and Mixed Cotton and Linen Cloth.* However, facilities for injection of steam were not available. Our test results indicate that average shrinkage in the warp direction was 2.2 percent while average shrinkage in the filling direction was 2.8 percent.

Comparing average shrinkage rates to similar laboratory tests conducted previously, we concluded empirically that the shrinkage characteristic of this material is within acceptable limits.

Three test samples received from another supplier revealed notable differences in knit geometry among the back, side, and lace sections of the finished product suggesting that different grades of coutil material are used. Tests for shrinkage due to laundering indicated that although the average shrinkage values of these appliances range from 2.1 percent to 2.7 percent after 5 wash and dry cycles, each of the three types of coutil had a different *rate* of shrinkage. Each reached its maximum shrinkage after a different number of wash-dry cycles.

3. *Stump Socks.* Three samples of wool leg stump socks and one sample of a wool arm stump sock were submitted by a foreign company for evaluation as described in Invitation No. MS-6-65, dated April 13, 1964.

As shown in Table 7, the results of these tests indicate that the samples fell below VA specifications in the number of wales per inch, a factor reflected in the excessive shrinkage in width.

With the exception of the number of plies, the arm stump sock falls below VA standards in knit geometry (Table 8). Although shrinkage characteristics are well within acceptable limits, the small size of the arm

TABLE 7. — *Compliance Test Data on Three Wool Leg Stump Socks*

Specifications		No. 15	No. 16A	No. 0
Yarn Size	16-18.5	17.3	17.2	17.3
Ply	3, 4, 5, 7	5	5	5
Stitches/in.	18 ± 2	16.2	16.2	16.1
Wales/in.	12 ± 1	10.0	10.2	10.0
Material	100% Wool	NOT TESTED FOR WOOL CONTENT		
Maximum Shrinkage in Percent of Initial Length and Width After 5 Wash-Dry Cycles				
Length	12	1.7	4.5	7.3
Width at 2 in. from toe	10	15.0	13.6	13.6
Width at 6 in. from toe	10	13.6	11.2	8.4

TABLE 8. — *Compliance Test Data on One Wool Arm Stump Sock*

Specifications		Test sample No. 202
Yarn Size	18.5-20.0	17.6
Ply	3, 4, 5	5
Stitches/in.	19 ± 2	15.5
Wales/in.	13 ± 1	10.4
Material	100% Wool	NOT TESTED FOR WOOL CONTENT
Maximum Shrinkage in Percent of Initial Length and Width After 5 Wash-Dry Cycles		
Length	12	4.8
Width at 2 in. from toe	10	8.3
Width at 6 in. from toe	10	7.2

sock and the fact that only one sample was available may limit the significance of these results.

4. *Trans-Aid.* A heavy-duty-type patient lifter designed specifically for institutional service has been evaluated both in the Bioengineering Laboratory and clinically in a nearby VA hospital (Fig. 33). It featured a four-point suspension system that provided full body support for severely

disabled patients, safety straps for spastic patients, and a unique wrap-around body sling that greatly facilitated the placement of the sling under a patient prior to the lift. Except for a minor deviation, the device complied with the recently approved VA Specifications for Lift Aids.

B. Materials

1. Corfam. Early in 1964, E. I. DuPont de Nemours and Company, Inc. introduced Corfam, a synthetic leather-like material that is a composite of several common resins or polymers. While no one expects that

FIGURE 33. The Trans-Aid, manufactured by the Iatro Dynamics Corporation, has several unique features: the U-shaped base, the wrap-around sling, and the mechanical screw-type lifting mechanism.

Corfam will replace leather completely, there are many instances where synthetic alternatives will be very desirable.

In laboratory tests, Corfam was found to be the equal of fine calf leather, resembling it in appearance. It was unaffected by moisture, weighed a third less than leather, cleaned easily, kept its luster, was scuff-resistant, and retained its shape. However, Corfam does not spread or conform to dynamic changes of the foot and must be "heat set" after lasting or molding.

The VA Prosthetics Center does not look upon Corfam as a broad-spectrum substitute for leather but as a supplement for use in orthotic and prosthetic devices. On an experimental basis, shoe uppers, stitched to all leather soles, are being made of Corfam by conventional machines and techniques. Corfam is available in thirty-six inch rolls of uniform quality and weight. As a result, many inspection and cutting steps required for leather hides are eliminated, and excessive waste due to irregular qualities and shapes is reduced.

Two claims of practical value made for Corfam are its porosity or "breathability" and its durability. Interestingly, it has proven difficult (and to our knowledge, not yet adequately accomplished) to demonstrate the structure of the "pores." The photomicrograph in Figure 34 obtained at 300 magnifications using hematoxin and eosin stains indicates a structure somewhat related to porous epoxy or polyester laminate in which the "pores" do not communicate directly between the inner and outer surfaces. Rather, they inter-communicate with each other extensively and reach the surface at disparate points.

FIGURE 34. Photomicrograph of Corfam structure (300 magnifications) indicating discontinuity of the "pores."

Physical test results at the Army Medical Biomechanical Research Laboratory to compare the porosity of Corfam with that of leather have been equivocal. Some of the currently available data indicate that certain types of Corfam are slightly superior to leather in porosity and other data show the reverse.

Nevertheless, there is some evidence that the claim for durability is valid. Data derived from physical tests conducted at the Army Quartermaster Laboratory, Natick, Mass., indicate Corfam's superiority in endurance. It was flexed 190 thousand times before damage occurred as opposed to approximately 60 thousand times for leather. Furthermore, eight VA beneficiaries have been wearing shoes made of Corfam for periods of six to twelve months. While these studies have not demonstrated the superiority of Corfam, they do indicate that it is at least the equal of leather for shoe uppers.

In addition, Corfam has been increasingly used to replace leather for calf bands and cuffs in braces. The potential gain in this connection is high. The leather components of a brace must be replaced at least once during the life of a brace. If Corfam proves durable enough to last for the life of the brace, the net savings in materials and labor would be substantial. Studies on this material are continuing.

2. *Pressona Gypsum Bandage.* This product, a rolled non-elastic standard 32/28 crinoline with serrated edges to prevent raveling, is bonded with gypsum.* Individual rolls are packed in polyethylene bags. For test purposes negative impressions were taken of six normal legs and six below-knee amputation stumps. Among the advantages noted during these tests were that this material saturated quickly, the layers bonded readily with minimal working, wrinkles and bridging were minimal, and it covered more area per roll than comparable products.

Several disadvantages were noted. The Pressona bandage must be held firmly over as much of its width as possible while applying to prevent loss of the plastic core. If the core is lost, the plaster roll collapses and the bandage telescopes, becoming unmanageable.

VI. OPERATIONS REPORT FOR FIRST HALF, FISCAL YEAR 1965

The data in this section reflect the prosthetic and orthotic services provided for veteran beneficiaries during the first half of Fiscal Year (FY) 1965 by the VA Prosthetics Center's Orthopedic Shoe Service and Prosthetics-Orthotics Service.

A. The Orthopedic Shoe Service

As shown in Table 9 and Figure 35, the workload of the Orthopedic Shoe Service has been steadily increasing. It is gratifying to note the continued large-volume use of the shoe repair program is resulting in a trend toward the issuance of fewer pairs of new shoes to our beneficiaries.

*Gypsum is a hydrus calcium sulfate ($CaSO_4.2H_2O$); plaster of Paris is a hemihydrate of calcium sulfate ($CaSO_4.\frac{1}{2}H_2O$).

TABLE 9. — *Numbers and Costs of Shoes in VAPC National Orthopedic Shoe Program*

	Fiscal Year				
	1961	1962	1963	1964	First Half F.Y. 1965
Benef. on roles	9,017	9,344	9,681	10,021	10,152
New shoes, prs.	7,154	7,465	7,249	7,317	3,607
Prs. of new shoes issued per benef. on roles per yr.	.80	.80	.75	.73	.71*
Cost, $	372,798	394,222	387,757	440,136	218,312
Repaired shoes, prs.	8,390	8,660	9,716	9,487	4,436
Prs. of shoes repaired per benef. on roles per yr.	.93	.93	1.00+	.95	.87*
Cost, $	91,760	92,514	97,778	93,520	62,293
Total cost, $	464,558	486,736	485,535	534,056	280,605

*Projected for full fiscal year.

In the first half of FY 1965, 506 lasts were modified or adjusted at a cost of $976. During this same period, 154 lasts were duplicated at a cost of $1,463.

Only 58 pairs of protective footwear (rubbers or overshoes) were issued during the reporting period. Twenty-three of these were rubbers; the other 35 pairs were overshoes. The total cost for these items was $1,555.

Corfam, a poromeric material, has been used in place of leather in uppers of orthopedic shoes. Eight beneficiaries have shoes with Corfam uppers but only three were made aware of the change from leather to Corfam. Seven pairs of shoes were fabricated by orthotists of the Orthopedic Shoe Service; the eighth pair was fabricated by one of the VA Prosthetics Center's contractors. Wear experience is not sufficient as yet to present meaningful data.

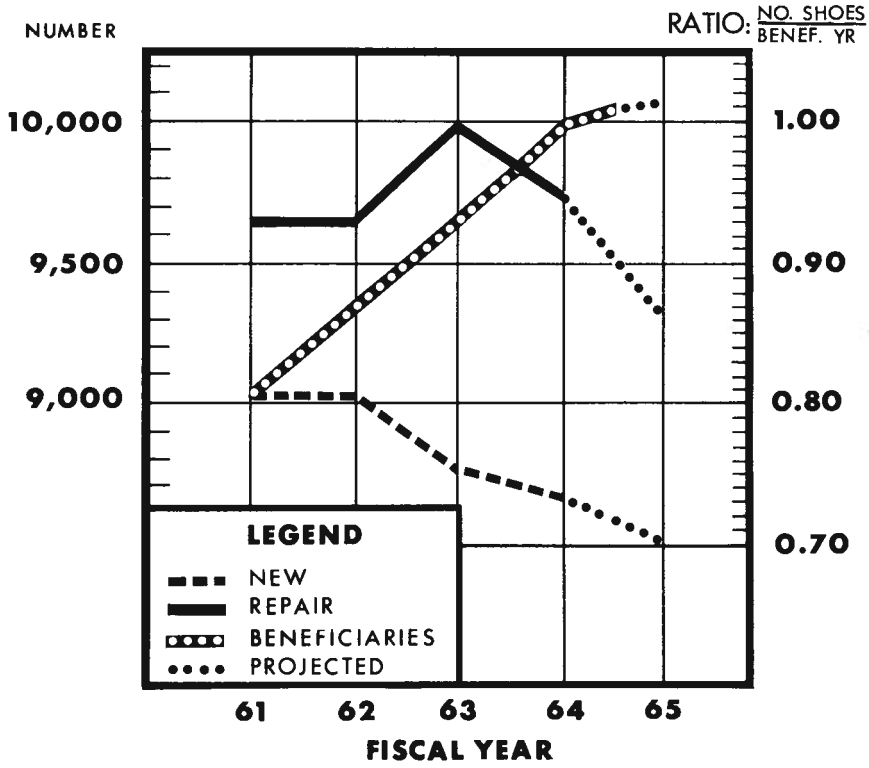

FIGURE 35. As shown, number of new shoes issued and number of repairs for each beneficiary are decreasing, while the number of beneficiaries are increasing.

TABLE 10. — *Numbers and Costs in the VAPC Elastic Hosiery Distribution Program*

Issuance Channel	F.Y. 1963		F.Y. 1964		FirstHalf F.Y. 1965	
	No.	Cost($)	No.	Cost($)	No.	Cost($)
Individuals	5,806	33,385	9,989	47,460	4,628	37,548
VA orthopedic shops	6,598	37,938	12,258	55,130	6,492	29,866
Total	12,404	71,323	22,247	102,590	11,120	67,414
Average cost, $		5.75		4.61		6.06

B. The Prosthetics-Orthotics Service

The Orthotic Components Unit of the Prefabricated Appliances Section, Prosthetics-Orthotics Service, has the responsibility for the distribution of surgical supports and elastic hosiery. As shown in Tables 10 and 11 and Figure 36, the growth of these programs is continuing.

TABLE 11. — *Numbers and Costs in the VAPC Surgical Support Distribution Program*

Issuance Channel	F.Y. 1963		F.Y. 1964		First Half F.Y. 1965	
	No.	Cost($)	No.	Cost($)	No.	Cost($)
Individuals	2,162	11,768	2,434	12,901	1,206	9,954
VA orthopedic shops	3,098	12,001	3,995	18,091	1,569	7,625
Total	5,260	24,769	6,429	30,992	2,775	17,579
Average cost, $		4.61		4.82		6.34

The fluctuations in average cost figures are mostly due to changes in the internal accounting procedures.

Since the last report (for the FY 1964), the Prosthetic Components Unit of the Prefabricated Appliances Section distributed an additional 124 Hydra-Cadence Above-Knee Systems, bringing the total number of wearers to 1,176.

During the first half of FY 1965 it was necessary to replace 405 Hydra-Cadence (H-C) units, 455 H-C cosmetic covers, and 70 H-C feet. During the same period, 98 units were repaired by the manufacturer at no cost to the VA since these units were covered by the guarantee provisions of the contract. VAPC technicians repaired 178 units at a total cost of $1,690. One hundred and eighty-one Model "D" units were repaired and converted to Model "E" units at a total of $27,150.

The distribution of components for temporary prostheses continues at a steady pace. VA orthopedic shops received 18 above-knee units and 31 below-knee units. Thirteen above-knee units and 8 below-knee units were sent to commercial limb facilities. Ninety-two SACH feet were issued to VA orthopedic shops and 18 to commercial limb facilities.

The total output of lower-extremity prosthetic devices and lower-extremity orthotic appliances produced by the Limb and Brace Section

during the first half of FY 1965 is shown in Figure 37. In addition to the items shown in the figure, 826 arch supports were fitted and 26 spinal braces were furnished to beneficiaries.

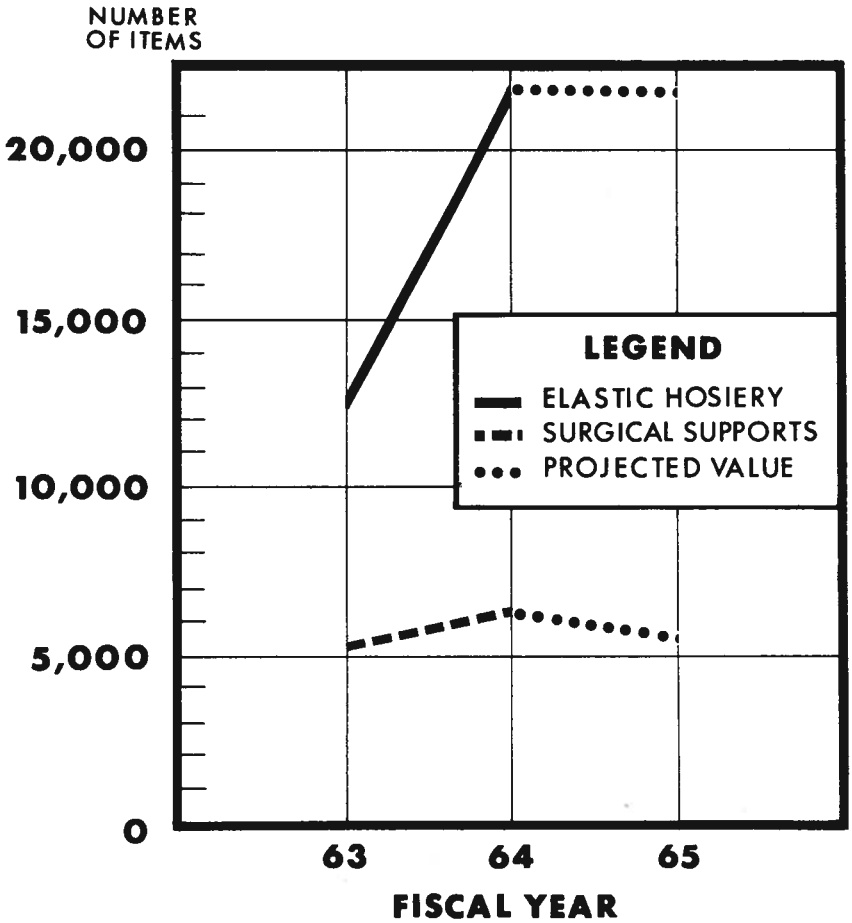

FIGURE 36. Initiation of the distribution of elastic hosiery and surgical supports by the VAPC caused increased issuances in FY 1963 and 1964, with future distribution expected to level off.

In the first half of FY 1965, the VAPC Clinic Team held 38 meetings during which 91 veteran beneficiaries were seen. These patients came from 8 states and the District of Columbia. As previously reported, this team handles mostly cases of the so-called "problem" variety but some cases are participants in several VAPC research projects.

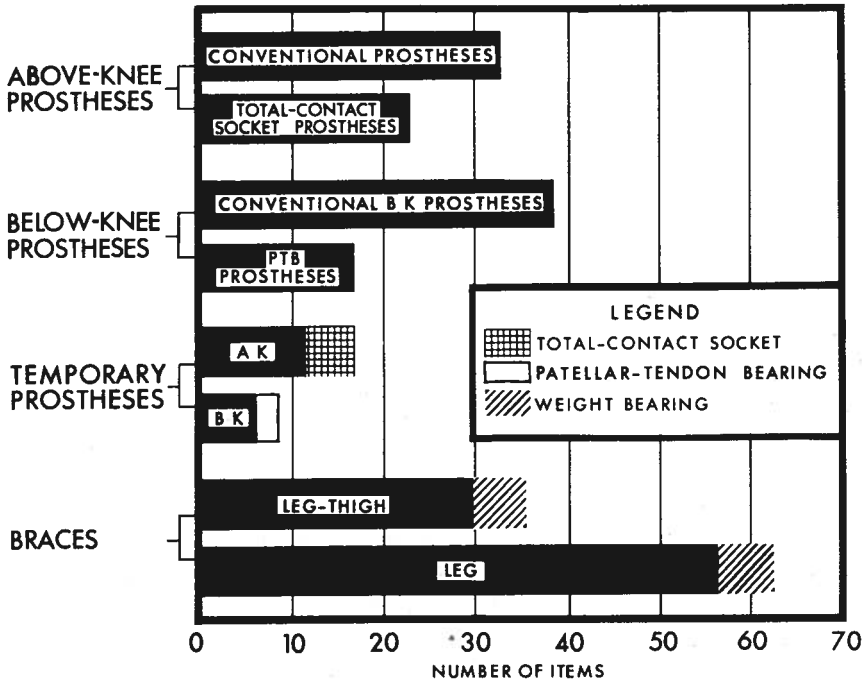

FIGURE 37. Total output of lower-extremity prosthetic and orthotic devices and appliances produced during first half of FY 1965.

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