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REPORT

I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

1. Compression of the Lower Extremity. In continuation of our efforts to develop adequate methods for measuring pressures applied to the lower limb by elastic hosiery and other orthopedic devices, we have employed two new methods. Both the "Komprit," a pneumatic pressure gage of German manufacture, and an electronic pressure transducer have been used to measure pressures applied by elastic hosiery to the lower leg. The results of these studies are reported in Section V.A.1. Of special interest, however, is the electronic pressure transducer and its potential value in other studies of pressure applied by orthotic and prosthetic devices.

The basic transducer consists of a metal capsule 0.5 mm. thick and 6.35 mm. in diameter. One face is a diaphragm incorporating a full wave, temperature-compensated, semiconductor strain gage bridge. The output is 20 millivolts at full scale pressure, with ± 0.5 percent full scale linearity and hysteresis. The gage is compensated for temperature at .05 percent full scale per degree Fahrenheit. The basic Sensotec transducer manufactured by Scientific Advances, Inc., of Columbus, Ohio, was modified

according to our specifications. The original gage was inadequate for other than purely hydrostatic conditions. The entire unit (Fig. 1) is now encapsulated in a silastic rubber droplet giving it a concave surface on one side that provides intimate contact with the skin. With its modified radius of curvature due to the silastic coating, linear and repeatable results were obtained. A calibration curve (Fig. 2) for the unit was obtained with the pressure being applied through a simulated "skin." In this test the inactive side of the gage was placed against a rigid backing (as it is when used in a socket), and the active side of the gage was covered with a sheet of Sylgard 51 that was exposed to controlled pneumatic pressures.

Additional tests using Polysar as the socket material, showed the gage to be reliable in measuring the interface pressure between the skin and the socket. In Figure 3 are the results of interface pressure measurements when outside pressure is applied to the Polysar by a pneumatic sleeve.

The development of a small, thin, reliable pressure transducer is a significant event in prosthetics. For many years the need for such a device has frequently been expressed at meetings of the National Academy of Sciences—National Research Council, Committee on Prosthetics Research and Development. Several universities and other agencies have been granted prosthetics research funds for the development of such a pressure transducer.

During the past 5 years many attempts have been made to solve the difficult problems of accurately and reliably measuring pressures at the interface between stumps and sockets. Among all the laboratories engaged in this work, only the group at New York University under the direction

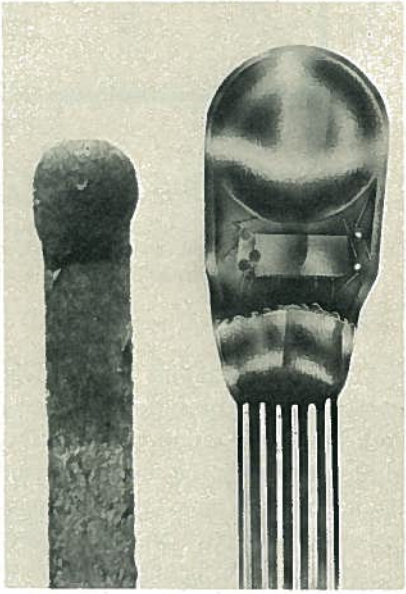

FIGURE 1.—VAPC-Sensotec semiconductor strain gage pressure transducer shown beside paper match.

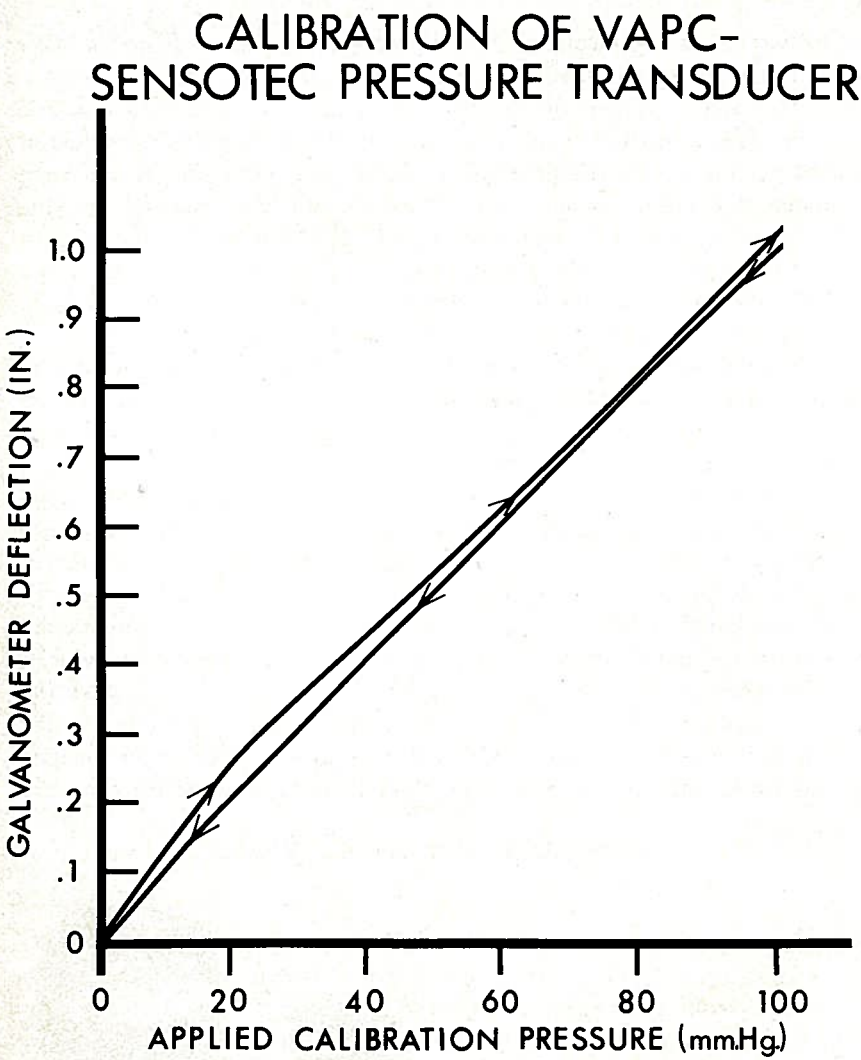

FIGURE 2.—Calibration curve of VAPC-Sensotec pressure transducer shows that response is linear. Hysteresis is negligible as indicated by similarity in the loading and unloading curves.

of Mr. Renato Contini has actually measured stump/socket pressures and correlated these pressure patterns with gait events. The success of the New York University group is a breakthrough of great magnitude. It has placed within our hands the means for studying and evaluating socket design and fit, perhaps the most crucial element in prosthetics service and one which had defied objective analysis to date. Credit for this breakthrough and for the scientific excellence of their work in measuring pressure with great

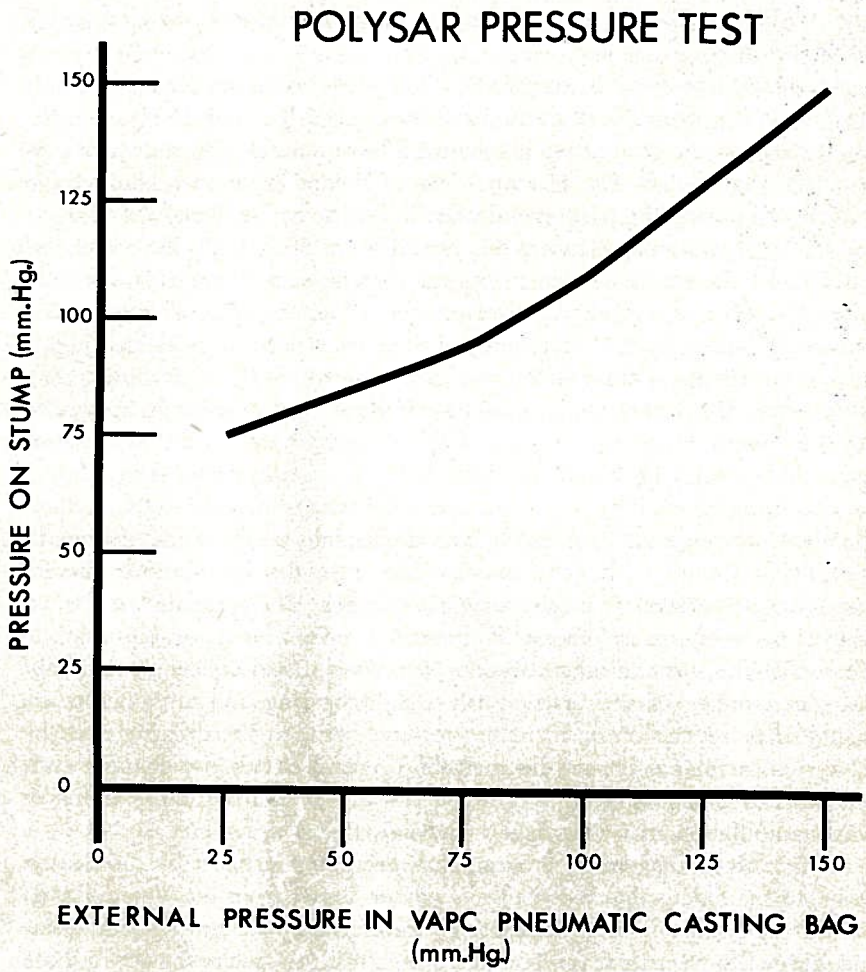

FIGURE 3.—VAPC-Sensotec transducer indicates that pressures applied externally to Polysar casts produce less-than-proportionate increments on the stump, up to approximately 90 mm. Hg. Higher external pressures produce equal internal increments.

precision and reliability belongs to NYU. Their pioneering work established the basic criteria for the study of pressure and only they have measured the pressures at such critical points in the socket as the lateral wall, the ischial seat, and the distal end. Their studies employed extremely accurate and reliable pressure gages, but due to their large size, they were inserted through the wall of the socket. They could not be applied to all sockets on a clinical basis without destroying the socket structure. Based on their work, we have redesigned a commercially available pressure transducer for use as a clinical evaluation tool. Due to its small size, durability, and reliability,

the VAPC-modified Sensotec pressure transducer is now being used in several studies.

Currently underway in cooperation with the Prosthetics Research Study in Seattle is a program to study pressures applied by rigid dressings in the immediate postsurgical fitting procedure. The amputation management procedures suggested by Dr. Marian Weiss of Poland have been studied and elaborated during the past several years in Seattle by Dr. Ernest M. Burgess under VA sponsorship. During this period over 120 patients have been the subjects of the recommended myoplastic and myodetic surgical procedure, the application of a rigid plaster-of-paris dressing, and early weight-bearing and ambulation. At the Seattle project these procedures have yielded highly beneficial results. Pain and edema are reduced to the vanishing point, stumps are firm and clean, hospital time is reduced, and problems associated with fitting of permanent prostheses are minimized. According to Dr. Burgess, these results depend on the triad of fundamental procedures employed in the amputee management program: myoplasty, rigid dressing, early ambulation. Myoplasty and myodesis are assumed to be the causal factors in producing stumps with good muscle tone since the traumatized muscles continue to perform in a physiological manner. The rigid dressing is assumed to be the causal factor in preventing edema and possibly pain, by reason of the mechanical restriction of excessive fluid collection and subsequent swelling of the tissue. Early weight bearing and ambulation are assumed to reestablish or increase pressures between the dressing and the stump and enhance normal healing. The general circulatory changes associated with change of posture or activity of the traumatized muscles during early ambulation may also have beneficial effects on healing.

This basic management program has been standardized by the Seattle team to the extent that it is being taught to teams from the Veterans Administration and other institutions. In transferring the knowledge and experience of the Seattle team to others, it is difficult to specify the appropriate pressure to be applied by the rigid dressing. The Seattle group plans to measure pressures between the stump and the socket to determine the pressure magnitudes normally applied by the rigid dressing and to monitor pressure vicissitudes until the cast is removed. These data on pressure therefore will serve as a guide in the initial application of the rigid dressing and in subsequent reapplications. The utility of these data may depend on the capability of controlling pressures while applying the dressing, a difficult problem with plaster-of-paris. The use of Polysar and the pneumatic casting techniques (BPR 10–3) designed to control pressure may overcome the difficulty.

In addition the Seattle group plans to obtain objective measures of other effects of these procedures. The EMG activity of the stump musculature will also be studied as an indication of its "tone" or physiological condition.

The electrical activity of the gastrocnemius and the tibialis anterior muscles will be monitored before and after amputation. Temperature will be measured in an effort to chart the course of the inflammatory response to trauma.

At the VA Prosthetics Center we are also using the pressure transducer in studies of the biomechanics of socket design. The UC–BL quadrilateral socket design calls for shaping the socket to maintain the stump adducted and flexed to prevent lateral instability of the hip in stance phase and to optimize the extension capability of the hip in early stance phase. The ischial seat and the critical anterior-posterior socket dimension are designed to distribute body weight between the ischium and the proximal portion of the thigh. Total contact designs are intended to distribute the body weight somewhat differently. We are now studying above-knee socket pressure patterns with the VAPC-Sensotec pressure transducer in an effort to validate the biomechanical theory underlying the UC–BL quadrilateral and the UC–BL patellar-tendon-bearing socket as originally expressed by Professor Charles Radcliffe of the University of California at Berkeley (1). In addition we are attempting to develop a clinically useful system employing pressure gages to evaluate socket fit.

2. *Gait Patterns of the Aged.* With the increasing age of World War II veterans and the increasingly older groups coming to the attention of civilian agencies, a great deal of interest has been generated about the special considerations which may be required in the prosthetic and orthotic fitting of the so-called geriatric patient. At the present time, prosthetics and orthotics services for these patients consist of modifying conventional systems to make them lighter, reduce the effort required to use them, and to stabilize patients to prevent falls. In essence, components designed for the young or middle aged, relatively vigorous amputee and paralytic, are modified to serve the aged. The underlying assumption, obviously, is that the aged, but otherwise normal person, walks in a fashion similar to younger persons—perhaps more slowly but nevertheless with similar temporal, kinematic, and kinetic characteristics. A corollary asumption is that the aged amputee can or does walk in a manner similar to the younger amputee. The tenability of these assumptions seems questionable in the light of several bits of evidence.

According to Drillis (2), cadence, step length, and speed of walking generally decline among the *non-amputee* aged population beginning at approximately 45 years. Moreover, his data indicate that the variability of these factors is also smaller than in the normal population. In general, average cadence declines approximately 10 percent in the 60–70 year old group and the range of speed is halved. Average step length in this age category is reduced approximately 20 percent and the range is reduced by approximately 30 percent. Average speed of walking is reduced by approximately 20 percent and the range of speed is reduced by 30 percent. Bearing in mind that these data described a *normal non-amputee* population (approximately 750 sub-

jects), we can expect that the performance of the *geriatric amputee* population will be even further modified.

A hypothetical but realistic geriatric above-knee amputee 60–70 years of age might be expected to walk at a cadence between 60 and 70 steps per minute, with a step length of approximately 25–28 in. and at a velocity of 2.0 to 2.5 m.p.h. Furthermore, increasingly reliable clinical observations indicate that aged (non-amputee) persons walk with a flat-footed gait, i.e., they do not exhibit the "normal" heel-toe transition. They also tend to walk over a flexed knee in stance phase without exhibiting the normal knee flexion in early stance phase followed by extension and subsequent knee flexion prior to toe-off. These and other observations lead us to believe that the normal locomotion pattern of the aged but otherwise normal population is significantly different from the normal pattern for younger people. Identifying and quantifying the gait factors which distinguish the aged from the younger population may be extremely valuable in providing guidelines for the design and development of special prosthetic and orthotic components for older disabled people.

With this in mind we have undertaken a study of the locomotion patterns of the aged. This study is being carried out in cooperation with Dr. H. Lippman of the Albert Einstein Hospital and Dr. Morris Brand of the Sidney Hillman Health Center, New York City, both of whom have had long experience with "geriatric" patients. We have deliberately tried to avoid the term geriatric since in current usage it has become so inclusive as to defy operational definition. As presently used, the term geriatric conjures up a patient perhaps over 65 years of age with a syndrome of infirmities which may include poor hearing, deficient sight, balance problems, arthritis, and other disease processes. To simplify the approach to this problem we are selecting subjects above 65 years of age who are otherwise unimpaired beyond the normal diminution of physical capacity and who are not under treatment for significant diseases.

The research program is designed to follow the general pattern of the earlier studies at UC–BL which produced the current standard reference data on normal human locomotion. Because instrumentation development has advanced since those studies were completed, additional data are being collected. The principal outcomes of this study will be a quantitative description of the temporal, kinematic, and kinetic factors in the gait of older people and an analysis of the significance of the differences between their gait and those of the younger population.

B. Development (Components)

1. Adjustable Below-Knee Standard Prosthesis. As previously reported (BPR 10–8) 100 of these units were manufactured and are being distributed to VA stations on request for use as temporary prostheses. We are continuing our efforts to bring to a completion the final design of cosmetic covers for

this unit. When an adequate cosmetic cover becomes available, adjustable below-knee standard prostheses will serve in both a temporary and a permanent capacity (Fig. 4).

Additional units are being ordered with several minor design changes. The socket adjustment plate of the quick-disconnect system will be made of aluminum since the previously used phenolic adjustment plate could not be readily machined to the required tolerances. This plate is also drilled and tapped to receive the socket attachment strap to permit interchangeability and reuse of the socket of the component. The maximum tube length has been increased from 8 to 12 in. to provide sufficient length for tall amputees with short below-knee amputations.

2. *Adjustable Standard Above-Knee (Multiplex) Prosthesis.* In early 1965 we described our original design concept and the potential advantages of this device (BPR 10–3). Since then we have periodically reported our progress in the development of the unit which has now reached a stage where we are considering bids for production engineering and fabrication of a limited number of devices. In its present configuration, to be known as the Model I, the unit includes a knee lock, variable mechanical friction knee control, and a complete alignment capability including individual adjustment by means of simple screw systems as in the Adjustable Below-Knee Standard Prosthesis. It offers options for a torque absorber, and it is capable of accommodating almost all of the commercially available hydraulic knee control mechanisms. Conceptually, this device is the core module of an above-knee prosthetic system whose applications range from the simplest immediate post-operative prosthesis to a permanent, sophisticated prosthesis incorporating the most advanced foot, shank, and knee mechanism. This versatility

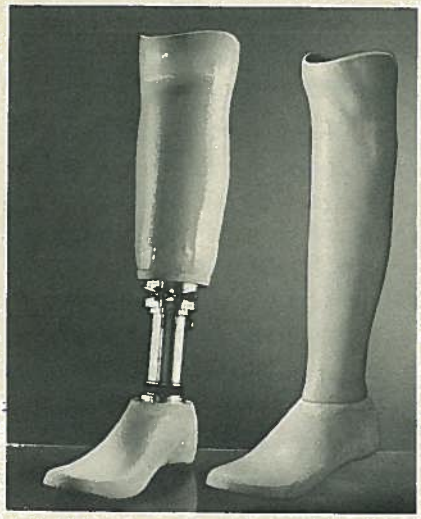

FIGURE 4.—VAPC BK Standard prosthesis (left) and most recent model of its cosmetic cover (right). The current production model includes the VAPC torque absorber just above the foot attachment plate.

stems from the modular design in which the basic unit conveniently accepts additional modules ranging from the simplest to the most complex available. Thus, it is conceivable that the basic unit may be furnished the patient on the operating table when the needs are crude alignment and perhaps knee stability. Within a short period as walking skill develops, more precise alignment is available and the required control of knee swing can be provided. The series of sockets from the original plaster-of-paris wrap to that of the final permanent prosthesis is easily installed. As a patient progresses to the point of discharge from the hospital, almost any of the available swing-control mechanisms and other components prescribed for the final prosthesis may be substituted. This gives the clinician the capability of continuously and conveniently modifying his prescription in accordance with the patient's progress and developing needs (Fig. 5).

We plan to issue these units on a routine basis in a program similar to that now in effect for the below-knee standard prosthesis. Initial use may be limited to temporary prostheses. As soon as adequate cosmetic covers are procured the full potential of these units will become available.

As the same time we are designing a second generation device to be called the Model II. While in the drawing board stage the Model II will permit us to accommodate newly developed but not yet commercially available

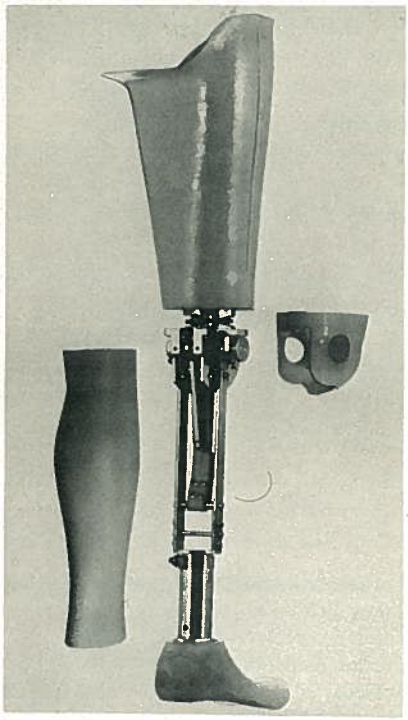

FIGURE 5.—VAPC Standard AK Prosthesis with cosmetic knee and shank covers.

devices such as the modified Mauch Model A knee control and perhaps a modified UC–BL pneumatic knee mechanism. Useful feedback from the field experience with the Model I will also be reflected in the design of the Model II.

C. Development (Techniques)

Direct Forming of Below-Knee Sockets. Polysar extruded into appropriately sized tubes has been increasingly used to mold below-knee sockets directly over patients' stumps (BRP 10–8). Twenty-four patients fitted during the past year with prostheses including Polysar sockets are still wearing them.

Fifteen patients have been wearing Polysar sockets for periods up to one year; nine additional patients have been fitted with temporary sockets of Polysar. The prescriptions for these patients varied according to need including all the conventional prosthetic feet, thigh lacers (three cases), and total contact sockets with foam ends. The nine temporary prostheses were fabricated with pylon-type shanks, and the permanent prostheses were fabricated with wood shanks. All the sockets were constructed on the PTB model. In some cases modifications of the socket were necessary due to stump shrinkage. This was routinely accomplished by reheating and reforming the Polysar socket according to the new requirements. In addition to the twenty-four unilateral amputees, one bilateral amputee with an above-knee and below-knee amputation has been wearing a below-knee Polysar socket and a conventional above-knee socket.

We are continuing to fit patients with sockets made of Polysar since no significant contraindications have been noted and the potential advantages of Polysar, detailed in previous issues of this Bulletin, are being realized.

Polysar in the form of extruded tubes is now being made available to suppliers of prosthetic and orthotic components and materials. For those who may wish to use Polysar to form below-knee sockets, a manual detailing the VAPC procedure is available on request.

D. Evaluation (Components)

Revised UC-BL Pneumatic Above-Knee Swing Control Mechanism. Fifty units, manufactured by the A. J. Hosmer Corporation of Campbell, California, have been delivered. Each has been inspected, and a representative sample has been retested for compliance with the original specifications. These units will be distributed to selected VA field stations as part of a clinical application study.

E. Evaluation (Techniques)

None.

II. UPPER-EXTREMITY PROSTHETICS

A. Development

VAPC Upper-Extremity Control System. In the previous issue of this Bulletin we reported an analysis of nine externally powered hands. Among our findings we noted that many brilliant concepts were embodied in some of the nine hands but that inadequate attention had been focused on the harnessing and control problems involved in utilization by a patient. With increasing frequency, development projects seem to be directed toward new or improved hardware without adequate consideration of, or relation to, a clearly defined problem of patients.

We believe that it is far more fruitful to base development projects on specific and precisely analyzed patient problems. A case in point is the standard above-elbow unilateral amputee whose catalog of problems is lengthy. The conventional above-elbow prosthesis usually consists of a terminal device, wrist unit, forearm, mechanical elbow with lock, socket, and standard above-elbow figure-8 harness. To operate his terminal device in a meaningful way, the amputee is required to flex his shoulder in order to flex the prosthetic elbow. Then he abducts and extends his shoulder in order to lock the elbow, an awkward motion which displaces the terminal device. After locking the elbow he continues flexing his shoulder in order to open or close his terminal device. Normally, he needs 2 to 2½ in. of excursion to flex the elbow and an additional 2 to 2½ in. of excursion to close or open the terminal device fully. After he has "used up" 4 to 5 in. of excursion he grasps or releases objects. At the extreme range of available excursion, control over terminal device force and position may not be optimum. It is also generally impossible for the above-elbow amputee to operate his terminal device simultaneously with elbow flexion or extension; the elbow must be locked prior to operating the terminal device. The amputeee, therefore, must operate sequentially, three separate "machines"—elbow lift, elbow lock, and terminal device.

A reasonably designed, electrically powered elbow could easily eliminate the need for locking and unlocking and reduce by half the total excursion required. Control over terminal device operation by means of muscles might well be superior to control by externally powered sources since much of the original feedback system including the residual tactile and proprioceptive receptors would be employed. From this viewpoint it would seem potentially advantageous to design an above-elbow prosthesis consisting of conventional components and an electrically powered elbow. Moreover, if this hybrid system could be controlled by the same number and type of control motions used in current conventional systems, a significant step forward in service to above-elbow amputees may be taken.

With this in mind we have designed a control system to extend and flex an electrically powered elbow and to open and close a terminal device by the

same control motion conventionally used—shoulder flexion (Fig. 6). The device is designed to be attached to the control attachment strap (CAS) of the conventional above-elbow figure-8 harness and connected to the conventional Bowden control cable. Flexion of the shoulder exerts a pull through the CAS. An excursion of approximately ¼ in. and a force of 4 or 5 oz. closes a microswitch which causes an electrical elbow to extend. Continued

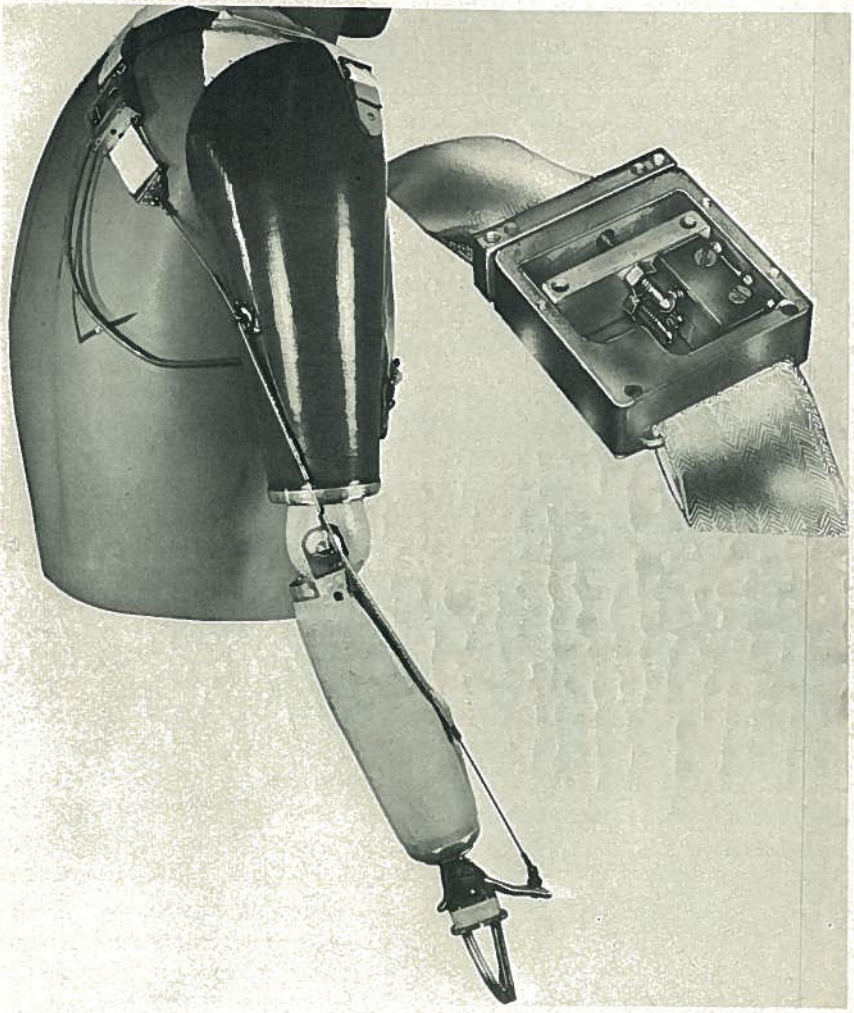

FIGURE 6.—Mounted on mannequin is prosthesis with electric elbow and VAPC control system installed in control attachment strap of conventional AE figure-8 harness. Inset shows position of two control rods and two microswitches which they depress and release as they travel under the pull of the control cable. Shown here is "neutral" position in which the two microswitch buttons are positioned in the detents of the conrol rods.

shoulder flexion, producing an additional excursion of ¼ in. and a force of approximately 10 oz., overrides the first switch and closes a second microswitch which causes the electric elbow to flex (Fig. 7). Continuing shoulder flexion beyond this point transmits force exerted by the muscle through the switch to the conventional control cable to actuate the terminal device (Fig. 8).

The principal advantages of this system are: 1. reduction of excursion required to operate the terminal device by approximately 2½ in.; 2. elimination of the elbow locking body motion—abduction-extension of the shoulder, elimination of the displacement of the terminal device during the elbow locking and unlocking cycle; 3. improvement of terminal device control owing to the greater length of the muscles while they are actuating the terminal device. Secondary advantages are: 1. extensive retraining of the patient is unnecessary since the same basic control motion—shoulder flexion, is employed; 2. the control switch is inserted between the control attachment strap of the conventional above-elbow harness and the Bowden cable system; 3. the elbow lock control strap is eliminated; and 4. the system can be used with any electrically powered elbow and any conventional terminal device.

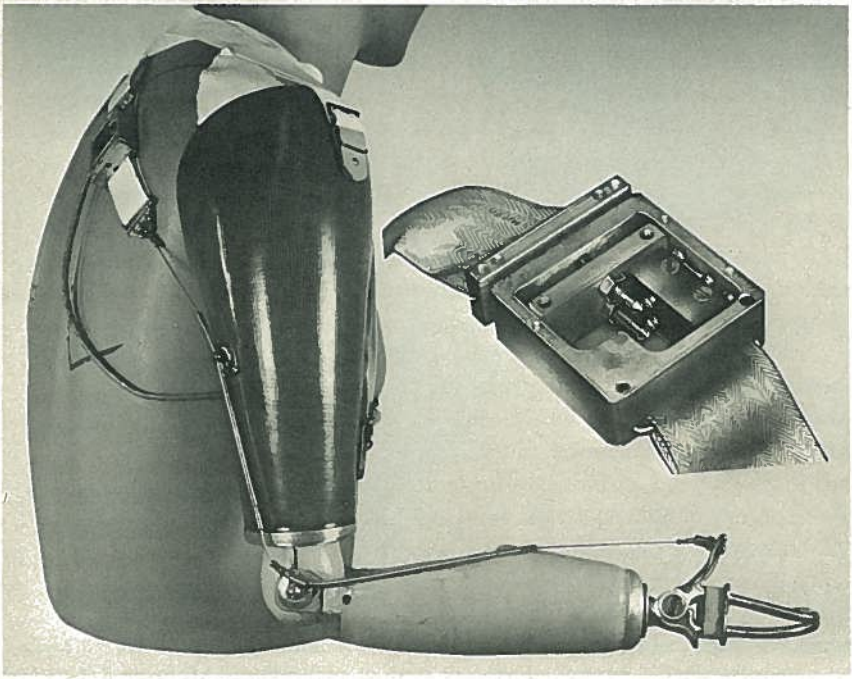

FIGURE 7.—A low force small excursion pull draws the control rod past the upper microswitch button (controlling extension) and into position to actuate the lower button to produce elbow flexion.

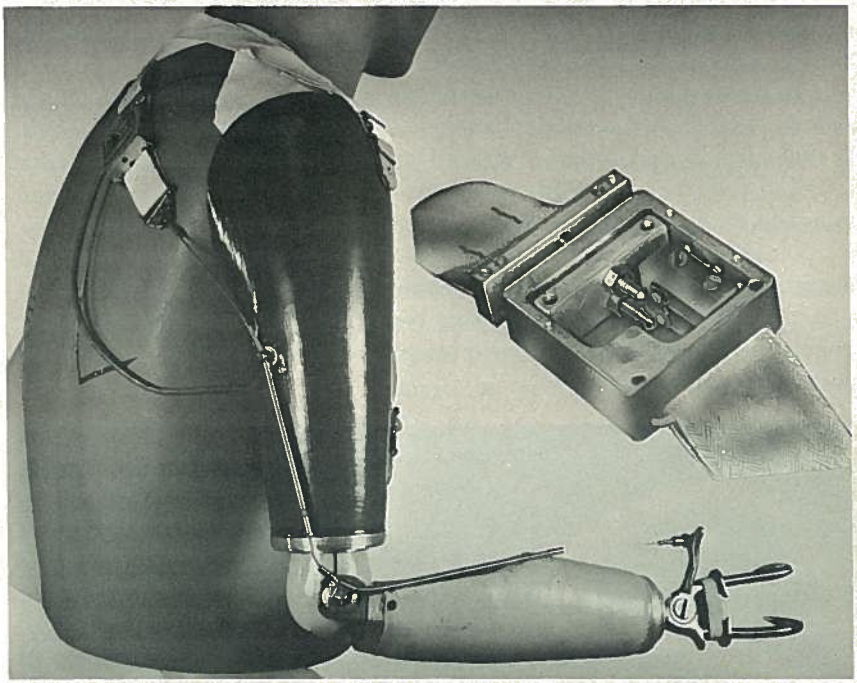

FIGURE 8.—Continued pull on cable moves both control rods off the microswitch buttons and transmits pull through the entire unit to control attachment strap and to control cable to actuate terminal device.

The model pictured in Figures 6, 7, and 8 is a prototype to enable us to determine the most advantageous force and excursion for controlling the elbow. Several models are being fabricated for laboratory studies with patients.

B. Evaluation (Components)

1. AIPR (American Institute of Prosthetic Research) Externally Powered Components. As reported in the previous issue (BPR 10–8) our research and evaluation program on the AIPR system has been completed. A report detailing several recommendations was submitted to the developer. Due to the accelerated pace at which the desired information was obtained, the developer has requested funds from his sponsor for a clinical field study to determine the utility of the AIPR pneumatic prosthetic system for adult bilateral arm amputees including shoulder-disarticulations and above-elbows. The aim of this project is to determine whether the current system is sufficiently practicable to be effectively applied in typical amputee centers, or whether the more highly specialized procedures previously employed in laboratory studies are required. VAPC is cooperating with Duke University Medical Center and the developer, Dr. Edward Kiessling, in con-

ducting biomechanical studies and evaluation programs on new and improved models of the AIPR components.

2. *Evaluation of Externally Powered Hands.* In the previous issue of the Bulletin, we reported a detailed analysis of nine externally powered hands. As indicated in that report several of the hands were available to us for testing in the laboratory. Others were analyzed on the basis of information published by other agencies which evaluated one or another of the hands. We pointed out that certain of the apparently negative features we noted in particular hands may have been eliminated in subsequent models. One such case is the Beograd hand, an early version of which was discussed in our report. Since then we have in our possession a late, presumably production model of this unit (Fig. 9).

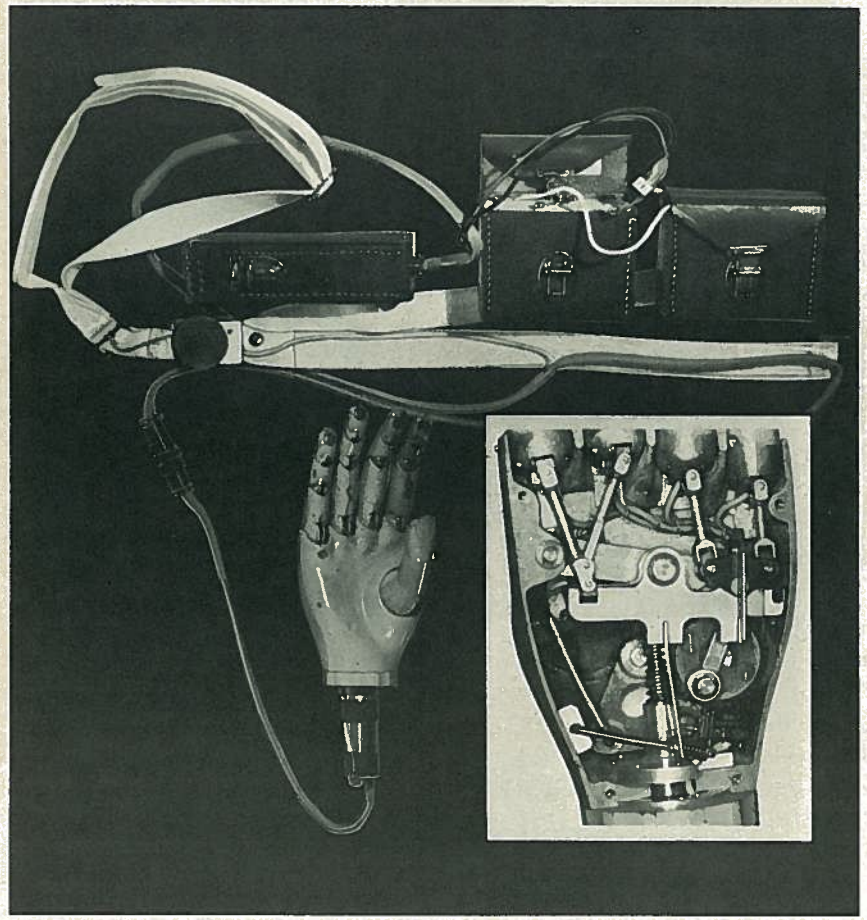

FIGURE 9.—Most recent model of Beograd hand with battery packs and amplifier section.

The current hand is reasonably anthropomorphic and can be covered with a cosmetic glove. The first metacarpal of the thumb is adducted, and the proximal phalanges of the fingers are longer than in a normal human hand.

The revised model retains the two modes of prehension: palmar and grasp. In palmar prehension the first and the index finger approximate the thumb as the ring and little finger continue closing and approximate the palm. In grasp, the thumb partially abducts and all four fingers approximate the palm.

The hand alone weighs approximately 25 oz. with another 44 oz. added by the battery and control package. Electrical power is supplied by four stacked silver-cadmium cells for a total voltage of 20 volts, and a current capability of 2 ampere hours. The maximum opening range is about 2¾ in. with a maximum prehension force of 5 lb.

The hand provides voluntary opening as well as voluntary and automatic closing in both modes of prehension. Information for the control unit comes from five sources: two in the harness and three in the hand itself. A microswitch and a potentiometer are mounted in one strap of the figure-8 harness, and as the wearer stretches the harness he releases the microswitch and turns the potentiometer; the harder he pulls the more the potentiometer turns. When he releases tension in the harness, the potentiometer returns and the switch resets. A second microswitch activated by the wiffletree push rod on the worm-gear drive is located in the hand. This switch is actuated whenever the fingers are in the extended position. On each of the fingertips is a pressure switch that is activated by contact with any solid object. The fifth source of information for the control unit is evidently obtained by monitoring the motor stall current.

In voluntary operation of the hand the direction of motor rotation is controlled by the microswitch in the harness which operates as an alternator; i.e., the first time actuated the motor will turn clockwise, the second time counterclockwise, the third clockwise, etc. The potentiometer in the harness controls the power available to the motor; i.e., under a weak pull the motor goes slow and with little force and as more force is applied to the potentiometer the speed increases.

The relationship between direction of motor rotation and hand operation is as follows: with the hand closed in palmar prehension, if the motor rotates counterclockwise, the hand will extend fully (actuating the microswitch in the hand) and then close in the grasp position; when the motor is rotated clockwise, the reverse sequence will happen. Thus, for the wearer to close the hand in palmar prehension from the open position, he must remember the last direction in which the hand was operated. If it was palmar, then he pulls the harness once with his force controlling the rate of palmar closure; if it was grasp he must "double shuffle" the harness to make the hand close in grasp. For grasp a similar situation occurs. Every

time the motor is actuated it starts in the opposite direction to which it last ran.

Operation in the automatic condition has the same limitations as above, but the microswitch in the hand acts to preset the direction (the reverse of the last action), and actuating the pressure switches in the fingers make the hand close at maximum speed.

The Beograd hand remains a sophisticated design concept that attempts to more fully simulate the normal hand. Fully articulated fingers that close in two adaptive prehension modes is a unique feature. Closing on contact with an object is a characteristic of the normal hand and also unique among powered devices. The present operating sequence is fairly simple and should not require much training.

As an extension of these design studies of externally powered terminal devices, we have expanded the program to fit patients with currently available powered hands in order to evaluate their function. In this portion of the program we plan to fit below-elbow and above-elbow patients successively with each of the available externally powered hands. After a suitable period of training and use, the performance of each patient with each hand will be tested objectively. A comparative analysis of relative skill in prehension, positioning, and practical tasks will be obtained. The VAPC test procedure previously described in BPR 10–3, pp. 128–129, is being employed.

To date, the Canadian hand and the Viennatone hand have been installed in the prosthesis of a below-elbow amputee who previously wore a Becker Imperial hand, a "Muenster" type below-elbow socket, and an axilla loop below-elbow harness.

Both of these externally powered hands are operated by EMG signals from the wrist flexors and extensors, which respectively close and open the fingers of the terminal device. New below-elbow prostheses of the Muenster type were required in order to incorporate the electrodes of each system into the sockets. In the Canadian hand the entire fitting and preliminary training of the subject was done at the Montreal Rehabilitation Institute, Montreal, Canada. The Viennatone hand was fitted and the patient trained at the Bioengineering Research Service, VA Prosthetics Center. The subject's initial performance with each of the new terminal devices and his performance 1 month afterward were evaluated.

As shown in Table 1, there was little significant difference in the speed of performance regardless of the device used. Significantly more grasp and compression errors were made with the externally powered hands. However, clinical observation and patient reaction clearly indicated that the performance with both externally powered hands was far less tiring than with the Becker Imperial hand. Although the Viennatone hand was "slower" (two seconds versus one second) than the Canadian, the patient preferred it because of the "firmer grasp" (12–20 lb. versus 4+ lb.).

TABLE 1.—*Analysis of Hand Functions Performed with Three Different Hands*

TD test	Becker		Canadian		Viennatone	
	Position	Prehension	Position	Prehension	Position	Prehension
Time (min.)	3.88	2.00	3.87	2.42	3.82	2.20
Errors (numbered)	2 grasp	3 compression	7 grasp	10 compression	6 grasp	8 compression, 2 grasp
Subjective rating	Good	Fair	Good	Fair	Good	Good
Observation and patient reactions.	Required 2 rest periods during test. "Really have to work."	Failure on metal disk, metal square, metal triangle. Difficult to release—wedging in TD ball.	In hurry to grasp—would knock over objects.	Failed on *all* flat metal pieces. Content to grasp compressible objects with no thought to force required.	"This test required *more effort* than Canadian did."	Failed to grasp large metal triangle, metal disk, metal square. "Slower than Canadian, less sure than Canadian, but prefer Viennatone for *firmer* grasp."

These data are insufficient evidence to support conclusions on the relative functional utility of the three types of hands. During the continuation of the evaluation program the same subject will be fitted, trained, and tested with other devices and additional subjects will be recruited.

An example of other devices which will be evaluated is the Collins hand (Fig. 10), which demonstrates very versatile prehension and grasp patterns.

3. Fillauer Low Friction Transmission System. A sample of an upper-extremity cable transmission system manufactured by Fillauer Surgical Specialties, Inc. of Chattanooga, Tennessee, was evaluated. Designed to

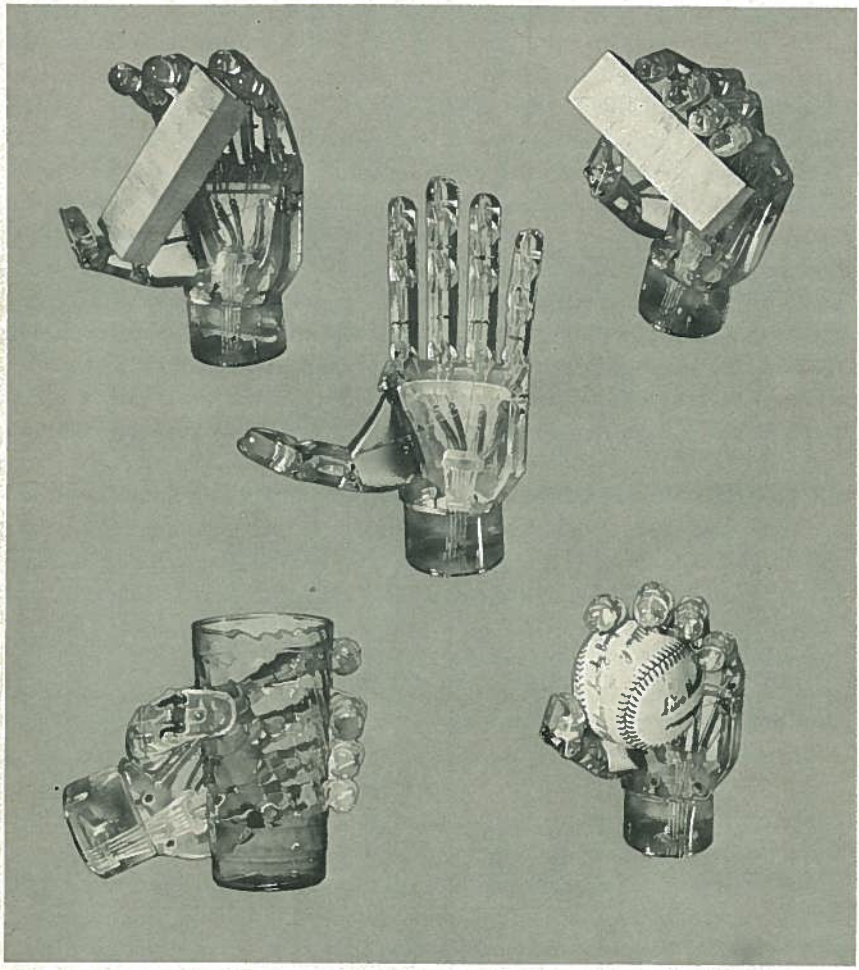

FIGURE 10.—Collins hand (center) has five active digits which enable it to grasp irregular objects at any angle of approach (above) and to "wrap around" cylindrical and spherical objects (below).

operate with very low friction the system consisted of: a. plastic transmission cable of spiral construction, .051 in. in diameter; b. Teflon liner with a wall thickness of .017 in.; c. heavy-duty steel housing; d. American Cable Company (ACCO) terminal fittings; and e. heavy duty hanger and cross bar (Fig. 11). The test procedures employed in this evaluation were the same as those used previously to evaluate cable transmission systems (BPR 10–8).

As shown in Table 2, the Fillauer Low Friction Transmission System elongated approximately 11 percent under 50 lb., a load which might normally be applied during use. When the 50 lb. load is removed, the cable material shortens but remains approximately 2 percent longer than its original length. These characteristics are quite similar to those noted in the Perlon Cable Transmission System in which similar elongation apparently did not constitute a deficiency when used by patients. Efficiency of the Fillauer Cable System in flexion and extension is comparable to the Bowden system and slightly higher in flexion than the Perlon system.

The tensile strength of the *total system* is comparable, if not slightly superior to, conventional Bowden cable systems. This may be due to the fact that the Fillauer system utilizes a heavier ball receiver (ACCO), the site of failure when the complete systems are tested. When tested without the ball receiver, the Fillauer cable breaks at the ACCO terminal fitting under approximately 115 lb. of tension. Under the same conditions, the standard Bowden cable withstands approximately 220 lb. The Fillauer cable without fittings breaks when loaded with 130 lb. which is comparable to Perlon. The Bowden cable breaks at 305 lb.

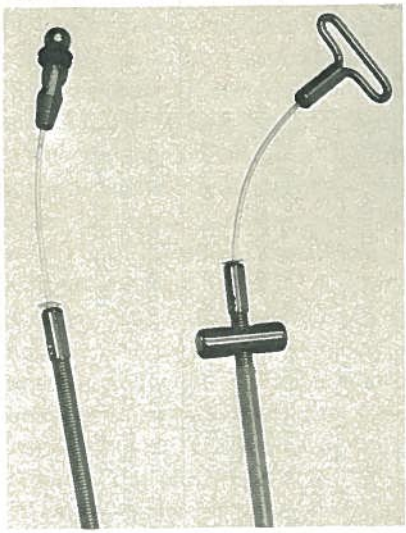

FIGURE 11.—Fillauer Low Friction Transmission System showing plastic transmission cable with Teflon liner and steel housing.

TABLE 2.—*Elongation, Efficiency, and Tensile Strength of Three Upper-Extremity Cable Transmission Systems*

System	Elongation percent (approx.)	Efficiency percent		Tensile strength, lb.		
		Flex.	Ext.	Test #1 Total system	Test #2 Without ball receiver	Test #3 Cable only
Bowden, conv. (standard) 3/64 in. steel cable, Teflon Spaghetti Tubing No. 15 AWG, standard steel housing, heavy-duty steel components.	0–1.5	79	93	92 Steel ball pulled out of receiver.	220 Cable pulled out of swaged harness connector fitting.	305 Cable broke at point of attachment to lower test grip.
Perlon cable 1/16 in. Nylaflow tubing 7/64 in. ID by 3/16 in. OD, heavy-duty steel components.	11–16			70 Cable pulled out of swaged ball fitting.	70 Cable pulled out of swaged ball fitting.	125 Cable broke within fixture attached to lower test grip.
A) Nylaflow tubing flared with cold tool.		68	91			
B) Nylaflow tubing countersunk on drill press.		68	91			
C) Nylaflow tubing flared with hot tool.		50	87			
Fillauer low friction transmission system. Transmission cable 0.051 in. in diameter. Teflon liner 0.017 in. thick. Heavy-duty steel housing.	11–13	80	94	105 Steel ball pulled out of receiver.	115 Cable pulled out of terminal fitting (ACCO).	130 Cable broke at point of attachment to test grip.

Since the tensile strength of both the Fillauer cable and Perlon is approximately the same, the greater strength of the *complete system* may be attributed to the way the cable is connected to the ACCO fittings. The slight increase in efficiency, 10 percent in flexion and 3 percent in extension, may indicate that the Teflon of the Fillauer system is superior to Nylaflow of the Perlon system in reducing friction. However, the Teflon liner deteriorates more under the grating effect of the braided cable used in the Bowden and Fillauer systems (Fig. 12).

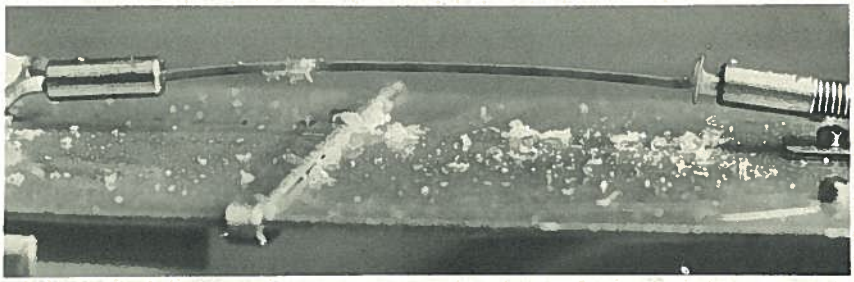

FIGURE 12.—Abrasion of the Teflon liner by the cable.

Both the Fillauer and the Perlon cables take a permanent set under tension, a matter which may affect long term durability. Several Fillauer Systems will be fitted to patients to determine its utility.

4. Humeral Rotator. In continuation of our studies to determine the utility of active humeral rotation for above-elbow and shoulder-disarticulation amputees, two samples are under investigation. As described in BPR 10–8, a laboratory model of an electrically powered, shear switch-controlled, humeral rotator was fabricated and installed in the prosthesis of an above-elbow amputee. The subject was trained to use the device and his performance of routine tasks is being studied.

The second system is a mechanical humeral rotator designed by a bilateral above-elbow amputee, Mr. Joseph Ivko, for his own use. As shown in Figure 13a, the device consists of two disks, the lower of which rotates with the elbow and forearm. Rotation is adjustable through a range of approximately 180 deg. A spring-loaded plunger locks the unit in any of seven positions. The humeral rotator is controlled by the same body motion used to lock and unlock the elbow; it is unlocked by a greater range of abduction-extension than used for locking the elbow. The elbow therefore must be unlocked to operate the humeral rotator. Internal rotation is actively controlled by shoulder flexion and external rotation is passively controlled.

Control cables for the elbow and the humeral rotator are mounted on the anterior portion of the humeral section and attached to the anterior sus-

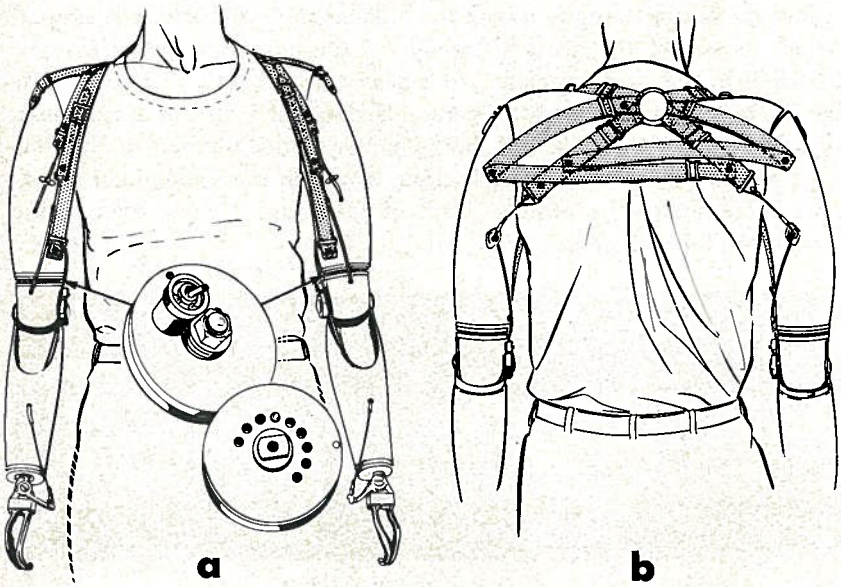

FIGURE 13.—(a) Mechanical humeral rotator, designed by Mr. Joseph Ivko, permits manual locking and unlocking of two plates mounted between elbow turntable and elbow to provide simulated humeral rotation. (b) Posterior view shows medial control cables which serve to rotate forearm internally when humeral rotator is unlocked.

pension strap of the harness. Several models of this device have been ordered for evaluation.

C. Evaluation (Techniques)

Direct Forming of Below-Elbow Sockets. On the basis of our experiences in fabricating below-elbow sockets of Polysar it has become apparent that forming a Polysar socket directly on the stump is a very simple technique. The use of Polysar in fabricating a *conventional* below-elbow prosthesis eliminates the need to: a. make a plaster-of-paris wrap cast, b. pour a positive cast, c. modify the positive cast, d. make a wax check socket, and e. laminate the socket. All of the other fabrication procedures with respect to wrist fittings, elbow hinges, harnesses, and control systems, are unchanged. New York University has undertaken a program to evaluate the fabrication procedure and the general utility of below-elbow prostheses constructed with Polysar sockets (Fig. 14).

Polysar in the form of extruded tubes is now being made commercially available through suppliers.

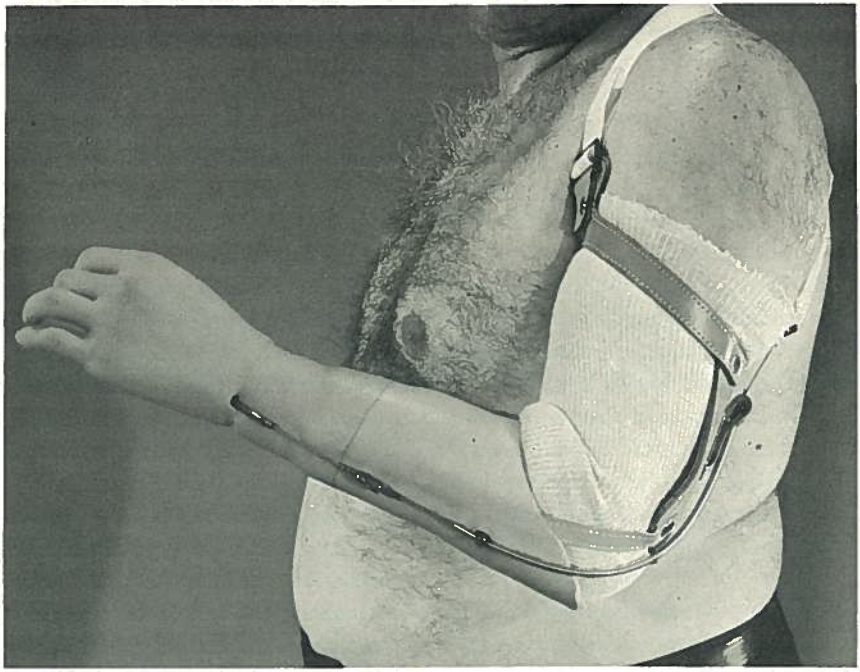

FIGURE 14.—Completed BE prosthesis with Polysar socket.

III. LOWER-EXTREMITY ORTHOTICS

A. Development

1. Survey of Brace Components. In the previous issue we published a list of lower-extremity brace components now in general use as indicated by their appearance in manufacturers' catalogs and prescription forms. The list was arranged in categories relating to the segment of the lower limb to which they are applied. To determine the extent to which these components are used in their purported applications we have conducted a survey of 28 VA Orthopedic Brace Shops. The results of the survey, detailed elsewhere in this issue, were highly significant.

2. Arch Supports. In a previous issue (BPR 10–7) we reported our efforts to utilize Spence-Gel in foot appliances designed to reduce plantar discomfort and calluses. Spence-Gel is a stable, chemically inert, silicone gel whose consistency is similar to that of human fat tissue. Foot appliances constructed with sheets of Spence-Gel were fabricated for two subjects. One had a history of callosities under the third metatarsal head that required the attention of a podiatrist at least once a month, and the other patient previously wore Schaeffer combination arch supports with two longitudinal steel shanks and a high metatarsal elevation.

Both subjects were provided with innersoles fabricated with ¼-in.-thick sheets of Spence-Gel in compartments under the heel, medial arch, and metatarsal region. These supports were worn by both patients for approximately 5 months with positive results. Both found the Spence-Gel supports extremely comfortable. The visits of one subject to his podiatrist were sharply reduced and the second subject reported substantial reduction of the pains he typically experienced in the lower calf and the plantar surface of the foot.

However, after 4 to 5 months of wear the Spence-Gel material extruded between the top and bottom sections of the support and through the stitches of the separate compartments into adjacent areas between the top and bottom sections. It has been impossible by stitching the top and bottom layers of the support, to contain the load-absorbing material which is subjected to high forces in standing and walking. With each step the full body weight, plus the inertial forces of movement, is applied to relatively small areas of the sole creating high pressures which cause the material to displace between adjacent stitches.

Due to the clearly positive results obtained from our experience with patients, we are continuing our efforts to package the material in a way to prevent unwanted displacement. Two methods are being investigated. In one we employ vinyl sheeting on which we deposit the Spence-Gel to conform to the required size of the areas under the heel, medial arch, and transverse arch. The vinyl sheet is then folded over and heat-sealed to enclose the Spence-Gel in compartments and to finish the lateral border of the appliance. The second method consists of depositing the material on a rubber sheet which is then vulcanized to enclose the gel to maintain it in its proper position.

B. Evaluation (Components)

1. VAPC Modular Single-Bar Braces. Our efforts in designing and developing single-bar (unilateral upright) braces have been described in previous issues. In BPR 10–1 we reported a measure of success in fitting single-bar leg-thigh braces to victims of polio, multiple sclerosis, paraplegia, and hemiplegia. The obvious advantages of the single-bar construction include the reductions of weight and bulk with a consequent conservation of energy by eliminating the medial bar.

Furthermore, the alignment of a single-brace joint to coincide with the anatomical knee center is a far simpler procedure than aligning two brace joints. Tibial torsion is more easily accommodated, and unwanted transverse rotation at the hip is more effectively controlled. Relative motion between the sidebar and the limb is minimized by specially designed thigh bands and calf cuffs. The thigh and cuff bands, with anterior openings, encircle about two-thirds of the extremity, providing more support than the half band generally used with double-bar braces. In effect, the tibia and the femur

substitute for the medial bar due to the fit and configuration of the bands and cuffs.

In BPR 10–2 we reported our work in developing a single-bar brace construction for leg braces ("short leg brace"). We described a brace consisting of a band, and as an innovation, a lateral spring-loaded round sidebar and a 90 deg. plantar-flexion stop. The design of the spring-loaded sidebar permits axial displacement, transverse rotation at the mechanical ankle joint, and possible eversion and inversion of the foot.

The key components in these systems are the leg-thigh brace calf and thigh bands and the spring-loaded sidebar (a component that might also be useful for some patients fitted with leg-thigh braces as well as leg braces). We have therefore designed a modular brace system, some of whose components are interchangeable and serve for both leg and for leg-thigh braces. Moreover, the system permits a wide latitude in brace prescription since it offers options of solid or spring-loaded sidebars, solid or spring-loaded ankle joints, and lever lock or free knee joints. Stirrups, calf bands, and thigh bands are standardized, and a hip joint is available that offers the option of free motion or provides a positive stop.

As shown in Figure 15 the leg brace and the leg-thigh brace are constructed of the basic stirrup and ankle joint. The leg brace is complete with conventional calf band. The leg-thigh brace shown incorporates the spring-loaded lever lock knee joint.

The VAPC modular single-bar brace system has been developed to the point where the Pope Brace Company was awarded a contract to make a limited quantity of the components for evaluation in several VA Orthopedic Brace Shops and in the facilities operated by several members of the CPRD Workshop Panel on Lower-Extremity Orthotics. As received from the manufacturer, the ankle joints, knee joints, and all joint bushings are made of hardened 410 stainless steel, and except for the lever lock knee joint, are designed to accommodate needle thrust bearings (Fig. 16). The stirrup, the pelvic upright, and the calf and thigh bands are made of 304 stainless steel because these components usually require extensive bending and cold working. All of the joints, including the stirrups and bands are neuter, that is, they can be used for either right or left extremities. The joints are designed to accommodate 5/16 in., round, stainless steel, aluminum or synthetic bar uprights; the joints can be modified to accept a 3/8-in.-round bar for heavier patients. General purpose Loctite, Type B (7–10) is used to secure the joint bushings and screws. Components for both leg-thigh and leg braces will be made available to other facilities for limited field evaluation.

2. *AMBRL Fiberglas-Epoxy Drop-Foot Brace.* The U.S. Army Medical Bioengineering Research Laboratory has designed a drop-foot brace in which Fiberglas-epoxy rods form the uprights. These rods, fabricated from Scotchply Reinforced Plastic, Type 1008, manufactured by the 3M Com-

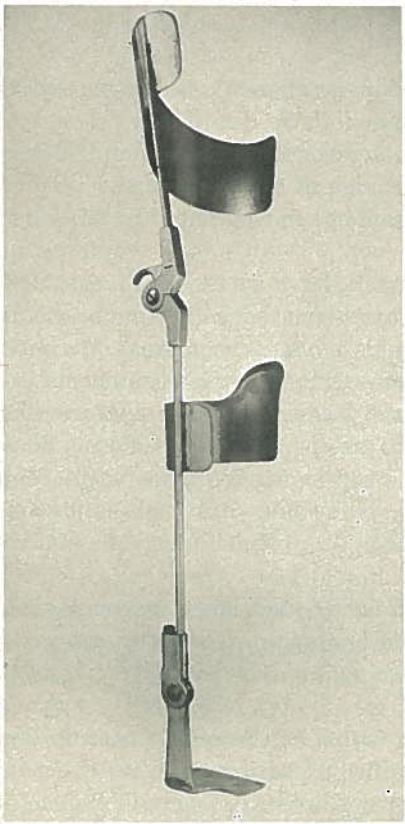

FIGURE 15.—Leg and leg thigh brace assembled.

pany of St. Paul, Minnesota, are pre-impregnated epoxy Fiberglas available in rolls of various widths.

A strip of the material is cut to the desired length. The tape is then rolled longitudinally as tightly as possible until a round rod is obtained (five strips of ¾-in.-wide tape will produce a rod of ¼ in. diameter; four strips will produce a rod of 7/32 in. diameter; three strips, a rod diameter of 3/16 in.). The uncured rod is then spirally wrapped with Tedlar, a polyvinyl alcohol; slight but uniform tension is applied during the winding. After sealing the rod at both ends with Tedlar release film, the wound rods are laid on a flat surface and cured for approximately 2 hours, at 160 deg. ±5 deg. C.

After consultation with AMBRL, we fabricated braces with straight Fiberglas uprights and a flexible leather cuff. In addition, we modified the shoe attachment using a length of 3/16 in. ID stainless steel tubing to receive the Fiberglas uprights. The center section of the tubing is flattened and a stainless steel tongue is soldered to the flattened portion (Fig. 17).

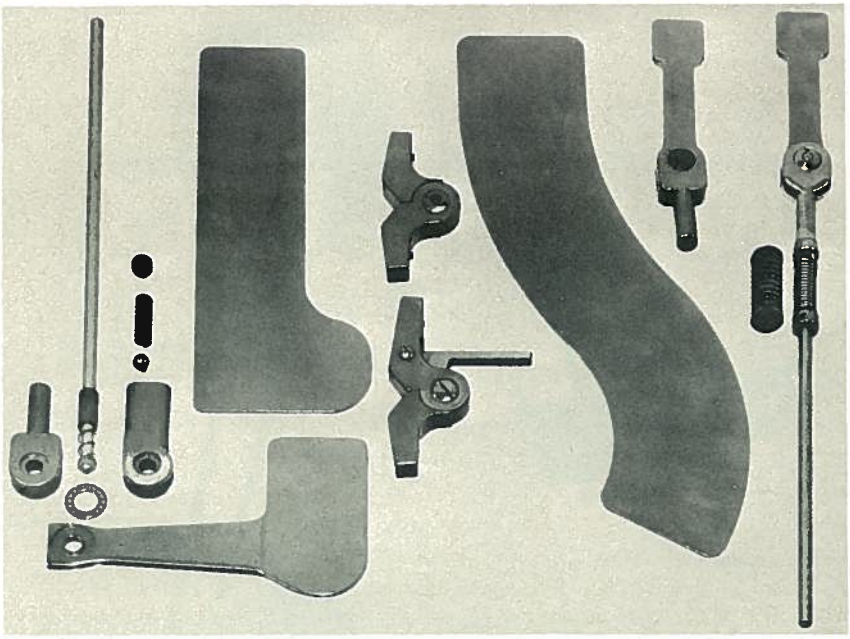

FIGURE 16.—VAPC Single-Bar Brace modules.

To date, we have fitted six drop-foot braces of the Fiberglas-epoxy type. The patient reaction to these braces has been favorable and we have not experienced fabrication difficulties or failures.

Since the procedure for handling the pre-impregnated tapes is a hand process subject to variation, we are now studying the use of prefabricated rods of the same, and possibly other materials.

C. Evaluation (Techniques)

Measurement of Foot Parameters. It is generally acknowledged that research and evaluation relating to the design and utility of appliances for the foot have not kept pace with prosthetics and orthotics research in general. One of the major impediments to progress in this field has been the difficulty of measuring and evaluating foot function, including motion at the ankle and subtalar joints as well as within the foot. Design and development in this field as in any other must be based on an adequate body of reference data to describe problems objectively and to make reliable observations for comparison. The collection of useful reference data depends on the existence of adequate instrumentation for recording appropriate parameters. To date, we have been handicapped by our inability to measure and record significant data at the shoe-ground and foot-shoe interfaces, and at the subtalar and ankle joints as well.

FIGURE 17.—AMBRL Fiberglas-epoxy sidebar.

Traditionally, kinetic factors related to the interaction of the foot and the ground have been measured by the use of force plates. Force plates are particularly effective in studying normal gait patterns and in evaluating the utility of prostheses and in some cases braces. Force plates do not accurately reflect, however, the instantaneous transition of weight across the entire surface of the foot; they only provide the "average" center of pressure at any instant during the stance phase. Areas of high pressure and shifts of pressure foci from one point to another are not readily revealed.

Photometric techniques in the form of "stick" pictures, cyclograms, and motion picture photography have also been used to study the effects of various pathological conditions on angular motion about the major joints on the lower extremity. Angular rotation about the ankle, knee, and hip has been measured by means of potentiometers mounted at the approximate centers of rotation at each of the major joints.

In dealing with problems of the foot, we are considering principally problems related to pressure distribution on the plantar surface of the foot and with foot balance, i.e., attitudes and motions of the subtalar and ankle joints. We are also concerned with the *effects* of foot/ankle dysfunction on the total gait pattern. Most of the currently prescribed appliances for the foot are intended to relieve or redistribute pressures on the sole of the foot and/or to control motion of the subtalar and ankle joints by restricting unwanted motion. To describe adequately the function of the foot and to evaluate the effects of foot appliances, we need to measure the pressure patterns on the sole of the foot both statically and dynamically. Adequate pressure data would provide the most direct indications of both the requirements for, and the effectiveness of, appliances. However, dysfunctions of the foot of any kind have second order sequelae in disrupting the normality of the total locomotion pattern. It is also necessary therefore, to study such significant gait factors as the pattern of weight application to the ground and rotations about the major joints. At the present time, the force plate and the photometric "stick" pictures provide adequate measurements of the latter two parameters. Pressures on the sole of the foot during walking have been extremely difficult to measure.

In the past, Ped-O-Grams, foot switches, and glass walkways did not provide adequately sensitive and reliable measures of the pressures applied between the foot and the shoe, between the shoe and the ground, or between the foot and the ground.

In BPR 10–4, pp. 161–164, we reported our efforts to develop appropriate instrumentation and techniques for objectively measuring weight-bearing patterns, foot balance, and other gait factors of patients fitted with appliances designed to support the foot and the arch. Pointed out were the difficulties of collecting objective data on such matters as "foot balance," pressure distribution, and weight transition during stance phase. We described our early efforts to utilize for this purpose the barograph, an instrument developed initially by Elftman (3) and refined at the University of California at Berkeley (4). The barograph permitted us to photograph the pressure distribution pattern over the entire surface of the foot or shoe but the data were difficult to quantify. Since that time we have designed a new barograph based on the original principle. The sensitivity and reliability of the new model approach the limits for such an instrument (Fig. 18).

The barograph consists essentially of a plate of glass with a high intensity light source around its perimeter. A modified polyvinylchloride (PVC) mat cut in a pattern of precisely dimensioned pyramids is placed on the upper surface of the glass plate. The entire plate, covered by the mat, is set above a parabolic mirror angled at 45 deg. to the lower surface of the plate.

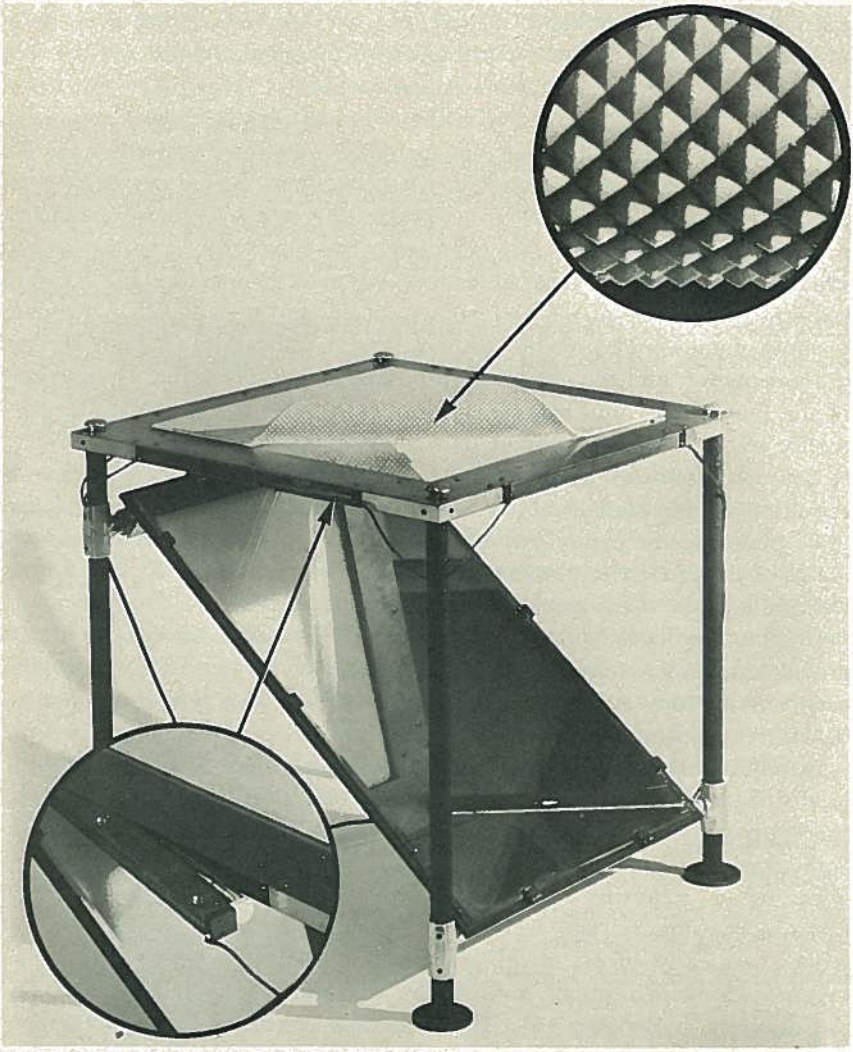

FIGURE 18.—Portable version of barograph designed to be placed between two tables to provide an appropriate walkway without elaborate installation. Lower inset shows high intensity lights. Upper inset is detail of mat.

Light is introduced through the edge of the glass in such a manner that it strikes the surface of the glass at a greater angle than the critical angle for a glass-air interface. The light "bounces" between the two parallel sides of the glass being reflected internally with minimal loss. When a pyramid of modified polyvinylchloride is pressed against the glass by the weight of the foot, the local critical angle is greatly increased. This results in light

crossing the interface, diffusing into the pyramid, and then being reflected down through the bottom surface of the glass plate. The higher the force compressing the pyramid onto the glass the larger and the more intense the lighted interface area becomes. The critical angle is changed so markedly because the ratio of the indices of refraction from glass to PVC approaches unity while the ratio of the refraction indices of glass to air is above 1½.

Used in conjunction with force plates and stick pictures, the barograph provides us with the most useful and complete set of data on foot function obtained to date. While very far from perfect, this system enables us to record pressure between the foot or the shoe and the ground and to relate these pressures to other kinetic and kinematic factors. For example, in order to study the effectiveness of an appliance for the foot, we can record the pathological condition as a frame of reference by obtaining force plate recordings, stick pictures, and barograms of the subject standing and walking (if possible) without the appliance. This procedure is repeated when the patient stands and walks with the appliance and the data obtained can then be compared to the previous record. If the device was prescribed to redistribute pressure on the sole of the foot, the barogram would reveal the changes due to the appliance. The effects of the device on the kinetic and kinematic factors of gait can also be determined by comparison.

This system was recently employed to evaluate the effects of an orthopedic appliance worn by one patient (Subject A). Shown in Figure 19 is a static barogram of the foot pattern of a 42-year-old, 146-lb. patient with traumatic amputation of phalanges II, III, IV, and V including portions of the metatarsal of the right foot.

The right shoe is built over a Levy mold, which includes a toe filler made out of soft cork. There is a long steel spring (extending from heel to toe) in his right shoe. A ⅛-in.-thick sponge rubber insole was placed on top of the Levy mold and lasted with the shoe. The lateral quarter and vamp of the right shoe have been padded with foam rubber. After the shoe was completed, a ⅛-in.-thick sponge rubber piece was cemented on the inner side of the toe filler. The left shoe was made over a custom-fitted last and had no orthopedic features.

The upper half of Figure 19 shows pressure patterns of the patient without shoes on which is observed a relatively normal weight-bearing pattern on the left foot, although somewhat less than normal weight is carried on the phalanges in general. The right foot shows obvious concentrations of pressure on the heel and under the head of the first metatarsal. Clearly revealed is a compensatory adduction of the great toe of the right foot in an effort to maintain lateral stability in the absence of the four smaller toes. When full weight is borne on the right foot alone, there is a marked increase in the pressure on the extreme distal tip of the right

SUBJECT A

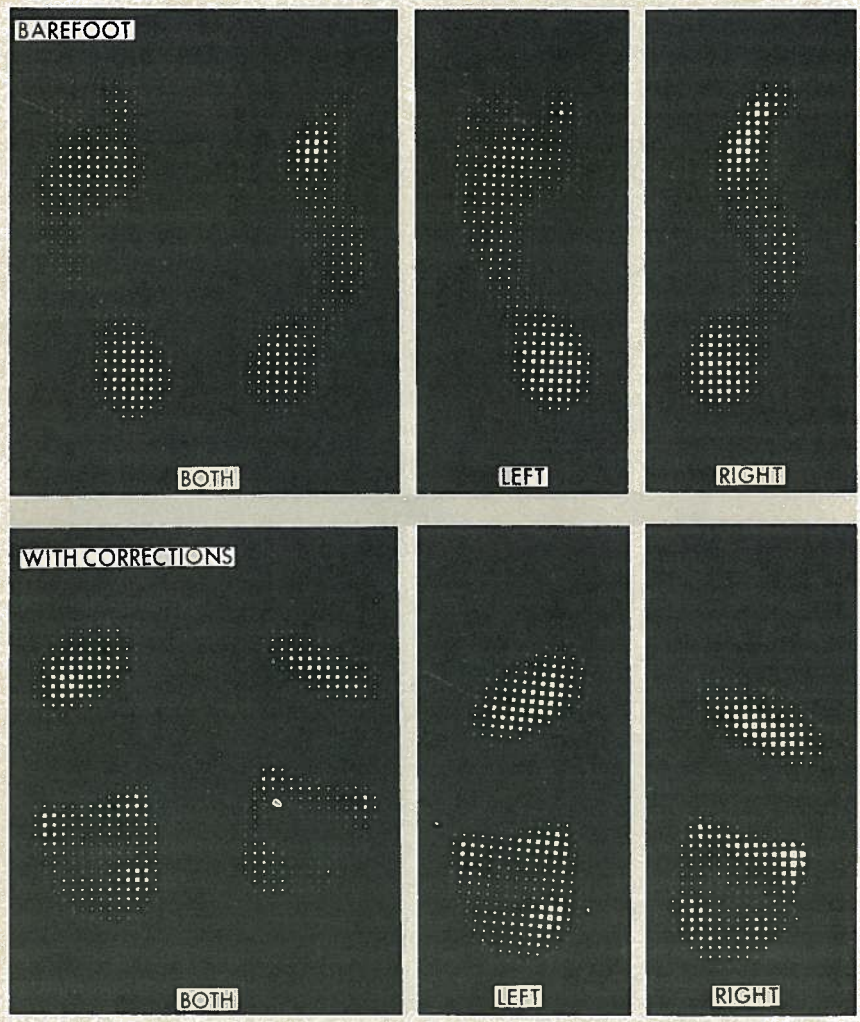

FIGURE 19.—Static barogram of Subject A with amputation of phalanges II, III, IV, and V standing barefoot (top) and with shoes (bottom). Barograms marked "left" mean all weight placed on left foot; those marked "right" indicate weight only on right foot.

toe. The lower half of the illustration shows the weight-bearing pattern of the patient wearing his orthopedic shoe. The distribution of weight seems relatively normal with perhaps more weight borne on the unamputated foot when standing erectly. When standing on the right foot alone, the barogram reveals a very high concentration of pressure on the right breast of the heel.

The upper half of Figure 20 shows the pressure pattern of the patient's right foot and the transition of weight onto the left foot. Reference to frames 20–40 reveals a high concentration of pressure on the first metatarsal head of the right foot (frame 20) as compared to the more evenly and centrally distributed pressure on the sound foot (frame 40) at toe-off. The lower half of the illustration, a barogram of the patient walking with his appliance, shows a reasonably normal redistribution of weight across the plantar surface of the foot. Despite the correction, very little weight is transferred forward to the metatarsal ridge. In contrast, the left sound foot (upper portion of frames 26–46) transfers the weight more evenly from the heel to the ball to the tip of the shoe.

In Figure 21 the linear displacements of the patient's hip, knee, and ankle and their angular rotations are compared with those of a typical normal subject. Of particular interest are the differences in the displacement of the hip and the flexion of the right knee when the patient walked barefooted and with his correction provided by the orthopedic shoe. Stance-phase time and knee-flexion range on the right foot were reduced when he

SUBJECT A

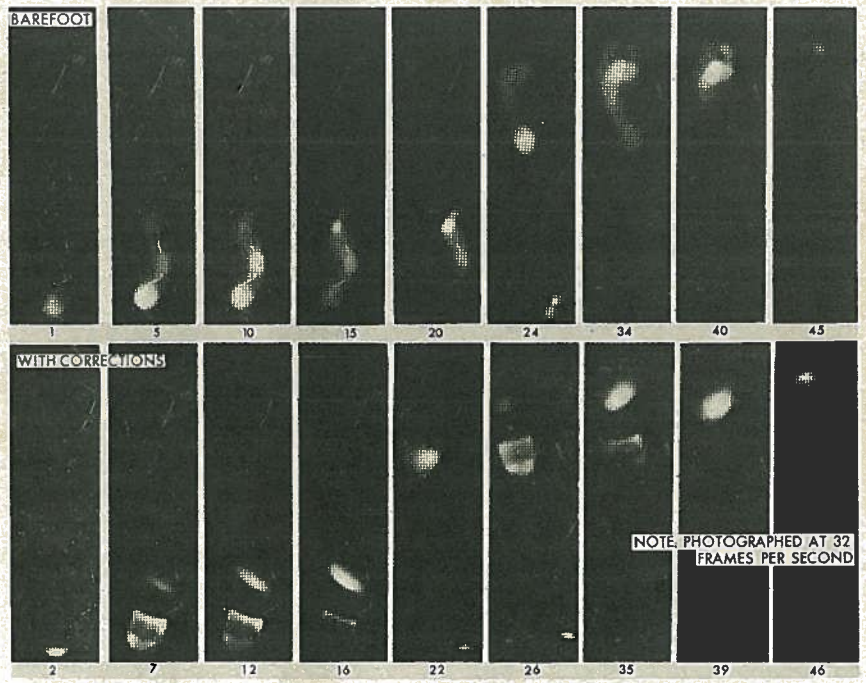

FIGURE 20.—Dynamic barogram (selected frames) of Subject A barefoot (top) and with shoes (bottom).

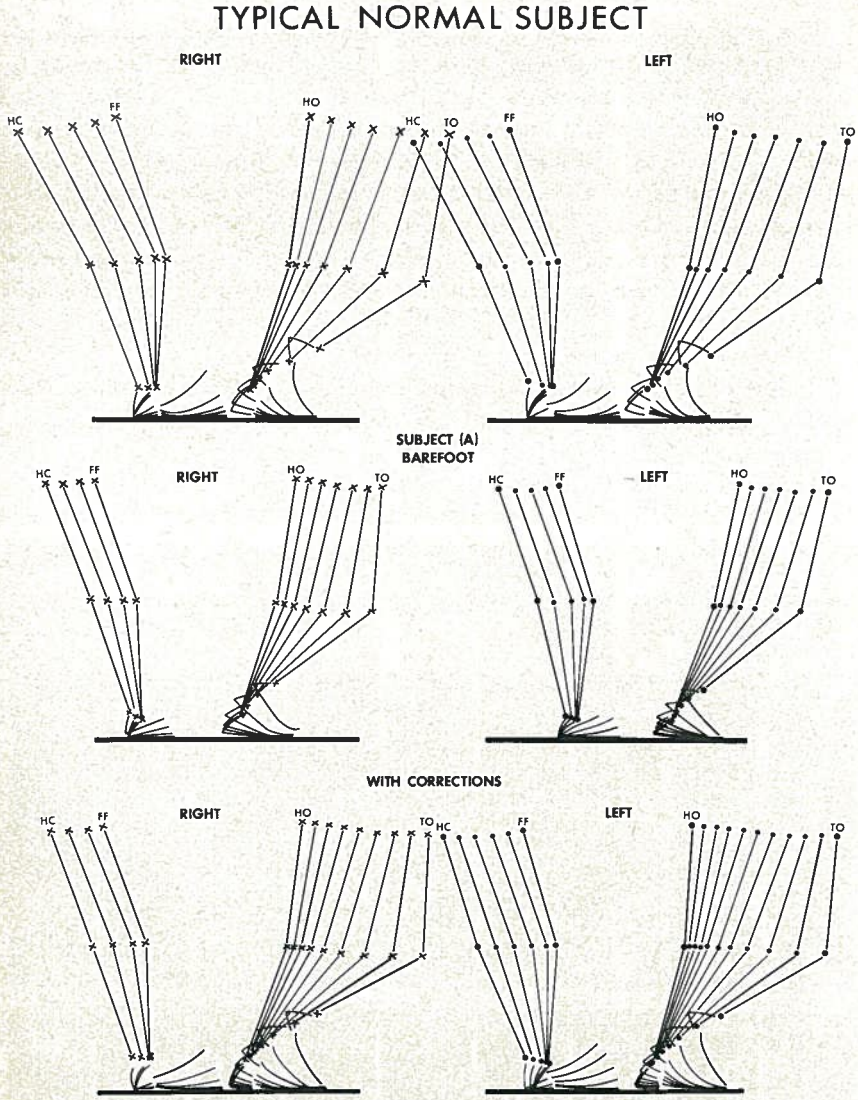

FIGURE 21.—Motion of Subject A's hip, knee, ankle, and foot compared to typical normal person (top).

walked barefooted. With the correction, he achieved a more normal pattern as shown by comparison with the left foot and with the typical subject.

Figure 22 shows the force plate record of the loads applied by the patient to the ground in comparison with the record of a normal typical subject. As expected, in the absence of four toes on the right foot there was very little, if any, push-off. Of special interest is the jog in the first peak of the

left or normal foot indicating an uneven transition of weight off the great toe of the right foot.

Despite the difficulties of reducing data of this type and the special equipment required, the system provides the most definitive picture obtainable at this time. Efforts are being made to refine the data collection battery by the use of newly developed pressure gages which can be inserted in the shoe.

A portable version of the barograph is being routinely used in the Orthopedic Shoe Service of this Center to record the weight-bearing pattern of patients fitted with custom orthopedic shoes. Permanent records obtained

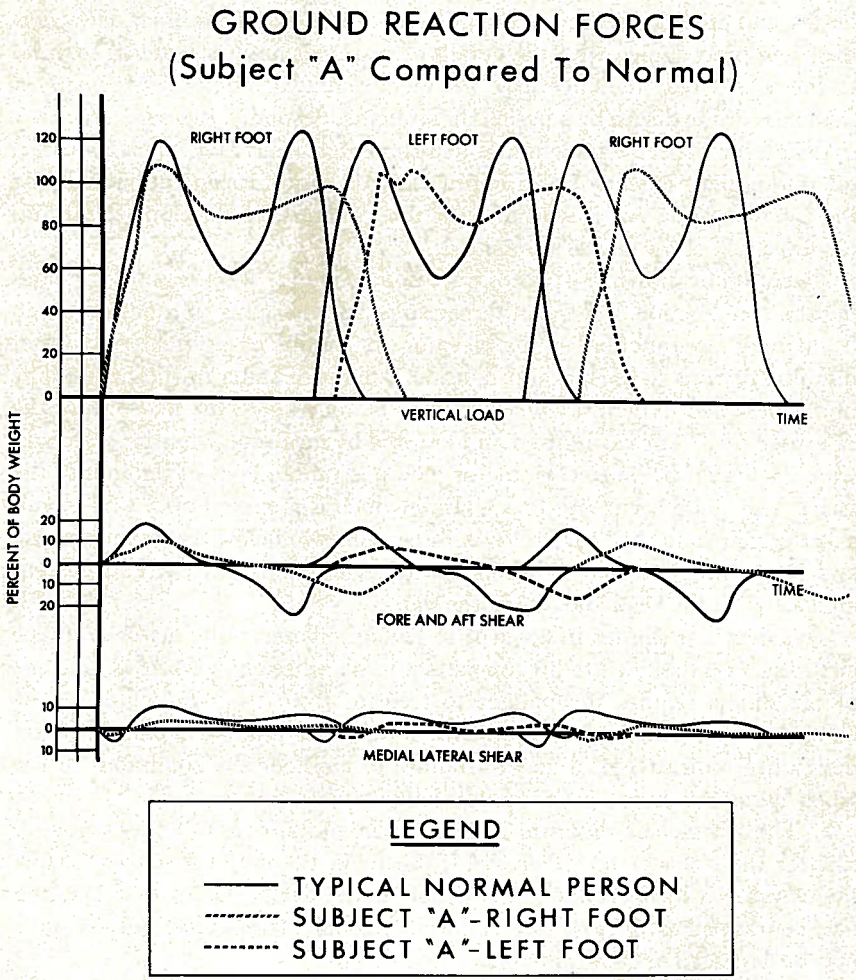

FIGURE 22.—Vertical load and shear forces of typical normal person compared to Subject A.

by new prescription and on worn shoes are being used to study the effects of wear on foot balance and weight distribution.

IV. ORTHOPEDIC AIDS

A. Development

None.

B. Evaluation (Components)

1. Home Stairway Lift. This report is presented with a view toward describing some of the fundamental problems associated with the design of home stairway lifts. We are frequently asked to evaluate designs of this nature and our analyses reveal the prevalence of two basic failings. Designers are often preoccupied with the details of materials and mechanical features to the extent that they overlook the biomechanical requirements involved in the use of the device by a patient in his home. Secondly, there is a tendency to design for a specific patient and to assume that what may be highly useful for one patient extends to an entire class of handicapped individuals. As the following analysis is intended for the guidance of developers, the full identification of the device is withheld.

Twenty-four drawings (photos were not available) were examined detailing the construction, materials, dimensions, and assembly of a device which is intended to function as an elevator to enable paraplegics and other disabled persons to ascend and descend a flight of stairs in the home. The unit is designed to permit a wheelchair to be rolled onto the platform of the "elevator car." The car is lifted and lowered by means of a continuous chain which is driven by an electric motor mounted on the floor at the foot of the stairs. The chain is encased in a housing which is mounted on a wall alongside the stairs and the "car" rolls on three rails, two below the chain housing and one within the chain housing. Switches to control ascent and descent are located at the head and foot of the stairs.

This device is similar in concept to several commercially available home elevators. It is designed to lift the patient in his wheelchair while all but one of the others require the patient to transfer from the wheelchair to the elevator seat. The following comments, noted in our analysis of one particular design are indicative of a general failure to consider the conditions of use when designing home elevators for handicapped persons.

a. The wheelchair platform is 36 in. wide and although it may swing up to clear the stairway for other use, it precludes the use of a stairway while the elevator is in operation. This is potentially dangerous in that children may be trapped on the stairs while the chair is moving or objects left on stairs may be damaged or cause damage.

b. There is no apparent provision for easy placement of the chair on the chair platform. Since the platform is approximately 5 in. off the floor, it would appear that the chair and patient must be lifted and placed

on the platform. A similar problem would be faced at the head of the stairs in rolling down from the platform to the landing.

c. The system is unsafe in that there appears to be no provision for automatically maintaining the car's position during ascent or descent in the event of a mechanical failure of one of the attachment or support links.

d. The control system is unsafe. Failure of the limit switch, either mechanically or electrically, would cause the car to continue moving beyond its intended limits and to follow the moving chain. As a consequence the device might be damaged and/or the patient injured. Additional safety devices such as continuous limiting switches or other means are required.

e. The unit's design does not permit a patient to stop the motion of the car at any point between the foot and head of the stairs. This is a potential danger in that the patient cannot control the motion of the car to avoid objects falling in its path or persons unknowingly walking in its path. The device also precludes the return to the top or the bottom of the stairs without first going through the complete ascent and descent cycle.

f. The wheelchair platform has no provision for securing the chair during ascent and descent. The patient must depend on the wheelchair brakes to prevent the chair from rolling while it is moving on the elevator.

g. The structure is unevenly designed in that certain elements appear to be extensively overdesigned with respect to strength.

h. In general the unit is not designed economically. It would probably cost four or five times more than other commercially available devices. Although most of the others are not designed to lift the patient in his chair, the problem of transfer from the chair to the elevator seat is not formidable. Even the use of a second chair to make one available at head and foot of stairs seems to be a safer and more economical solution to this problem.

i. The use of this device would be restricted to homes which have a single flight of stairs from one level to the other without intervening landings or a platform at the foot of the stairs.

In summary, the sole special function of the unit is to permit a paraplegic or other wheelchair user to ascend and descend a flight of stairs while sitting in his wheelchair without requiring him to transfer to the elevator seat. In this respect, it is similar at least to one other commercially available home elevator for lifting both chair and occupant. The device is unsafe because the operator does not control its movement at all times. It lacks back-up safety devices. It has no provisions for securing the chair on the platform while it is being lifted. In addition, it is likely to be unnecessarily expensive due to overdesign of several sections. Even if available, the device would be restricted to homes with a single straight flight of stairs between floors without a platform on the lower end. This lengthly catalog of apparent deficiencies stems from a failure to consider the full range of patients and the variations in the environments in which the device is to be used.

2. *Revised Model Power Aid.* As previously reported (BPR 10–8) the Power Aid for wheelchairs developed by California Medical Aids, Montrose, California, was evaluated. In general, the unit was found potentially useful for several classes of patients due principally to its versatility, ease of installation, and relatively light weight. However, the evaluation brought out the need for the following modifications:

a. Improve portability by packaging it in three reasonably-sized parcels.

b. Improve the traction of the drive wheel in ascending slopes by redistributing the weight.

c. Do not ground circuit to frame.

d. Use positive-type locking electrical connectors.

e. Provide higher capacity battery charger.

f. Provide a device for uncoupling the motor from the drive wheel to permit operating the chair manually.

g. Provide means for reversing the motor to avoid rotating the tiller 180 deg.

h. Reduce force to operate the power control.

Several models of the Power Aid incorporating some of the suggested modifications were procured and subjected to a limited clinical evaluation at Rancho Los Amigos; New York VA Hospital; East Orange VA Hospital; Bronx, N.Y., VA Hospital; and the Bioengineering Research Service. For test purposes, the Power Aid was mounted on several different wheelchairs.

In general, wheelchairs equipped with a Power Aid were found to be beneficially employed by patients suffering from cerebral palsy, hemiplegia, Parkinsonism, ataxia, cardiac conditions, multiple sclerosis, certain types of triplegia and quadriplegia, and emphysema. Although several negative reactions were noted, the unit was considered quite useful on the basis of its ease of operation, versatility, and attractive appearance.

The specific positive comments were:

1. Battery easy to change.
2. Battery easily removed for folding the chair.
3. Highly maneuverable.
4. Light weight.
5. Attractive appearance.
6. Versatile application.

The following negative comments were made:

1. Control lever too hard to operate.
2. Chair moves too fast.
3. Too much power.
4. Needs clutch to facilitate manual operation or pushing by an attendant.
5. Cannot approach closely enough to or easily back away from table.
6. Loss of traction on inclines and uneven ground.

Since the initial evaluation the manufacturer has redesigned the Power Aid and furnishes it only as a unit mounted on a Howmet wheelchair with desk-type arms (Model #6488–14–15–16) (Fig. 23). The current model includes modifications which have reduced substantially the problems presented in the negative comments 1–5 above. The force required to squeeze the operating lever has been reduced. A speed control has been incorporated which permits setting of maximum speeds below the maximum available. A clutch has been incorporated which disengages the motor from the drive wheel reducing sharply the force required to propel the chair manually.

An electrical toggle switch which reverses the direction of the motor has also been incorporated. Together with the desk-type arms of the Howmet wheelchair, this permits a patient to approach a table and to back away easily. However, the vertical section of the steering post is mounted forward of the lower rail of the desk-type frame and only the side of the chair on which the Power Aid is *not* mounted can actually

FIGURE 23.—The Howmet wheelchair.

be driven under a table. The patient is thus positioned obliquely with respect to the side of the table. Although the arm which controls the unit is farther away from the table, it is well within range.

Loss of traction on inclines (the sixth negative comment), may well remain a problem in special instances. Loss of traction going up hills is due to the rearward shift of the center of gravity of the entire man-chair system resulting in descreased weight on the drive wheel. At inclines above 6 deg., the loss of traction is significant and may in fact prevent ascent unless the patient leans forward in the chair as he climbs the hill. Leaning forward may reduce the problem to some degree but this form of compensation may not be available to some patients. In addition, the attempt to isolate the frame from the electrical system by insulation was inadequate leaving it still possible for a patient to receive an electrical shock.

Prescription indications for the most recent model require that the patient be capable of squeezing the manual speed control lever with a force of 5 lb. To enable a patient to steer efficiently, residual motion in the upper extremity should permit the hand to go through a range of motion corresponding to 100 deg. of internal/external humeral rotation and approximately 20 deg. of humeral flexion/extension. The new model requires that the patient be capable of operating a toggle switch at approximately the thigh level when seated. For *independent* operation, a patient must be able to reach two switches mounted on the unit at approximately ankle height (Fig. 24). However, as these switches govern motor speed and the disengagement of the motor from the wheel, they can be preset by an attendant.

The current model of the Power Aid is a substantially improved version. Two deficiencies however are still present: 1. The frame must be isolated from the electrical system. The manufacturer has accepted a suggested modification for this purpose and is currently taking steps to install the appropriate components. 2. The second deficiency relates to the location of the vertical projection of the c.g. of the entire system. It was noted in earlier models mounted on chairs other than the Howmet, that the unit lost traction on inclines due to the rearward displacement of the c.g. With the Power Aid mounted on the Howmet chair, the battery is placed farther aft than in other chairs, intensifying the problem slightly. However, the significance of this shift is best determined by more experience in patient utilization.

The Howmet wheelchair, an integral element of the system, is not presently a VA acceptable item. It was evaluated against current specifications (report in Section IV. B.4). Three deficiencies were revealed: 1. the lower end of the footrest support bar is too long (approximately 1 in.), 2. the force required to fold the chair was too high, and 3. the seat rail displaced under load to a point where it moved completely off its supporting bracket. Consultations with the manufacturer of the Howmet chair produced agreement on modifications to eliminate these deficiencies.

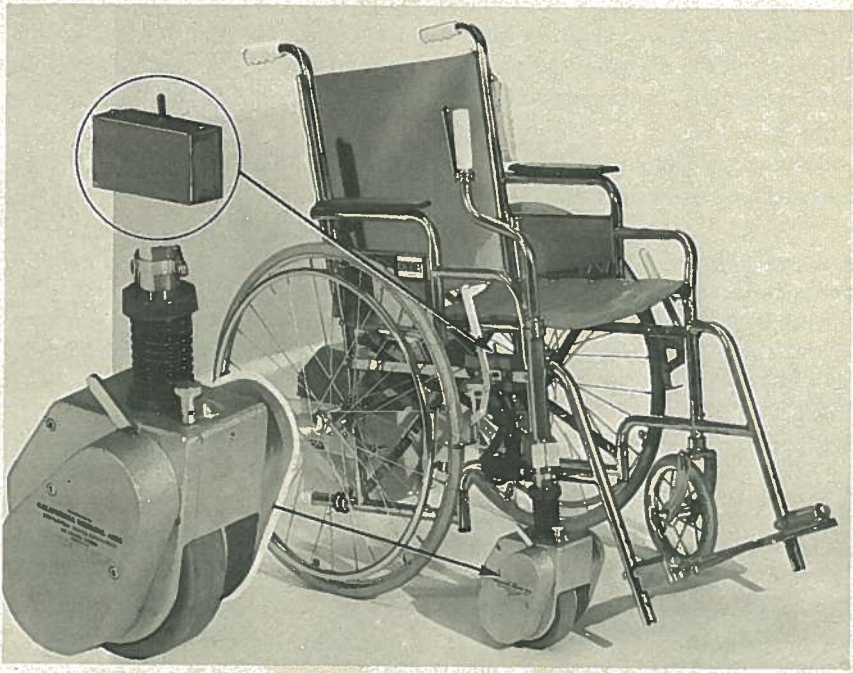

FIGURE 24.—Power Aid mounted on Howmet wheelchair. Upper inset shows switch which controls forward and backward motion. Lower inset shows lever by which drive wheel is disengaged from the motor to facilitate manual propulsion. Also shown is power control knob.

3. *Wheelchair Pads.* The results obtained in a limited clinical evaluation of the Stryker Floatation Pad (BPR 10–8) have stimulated interest in wheelchair pads in general. More information is needed about other commercially available wheelchair pads, the patients for whom each should be prescribed, and the time in the treatment regimen when certain pads are best utilized. Accordingly, we are continuing our limited clinical study to determine the relative value of several wheelchair cushions.

Inpatients and/or outpatients at designated VA installations, who have a history of and susceptibility to ulceration, and/or generally impoverished skin have been selected to participate. The clinic teams have chosen patients in wheelchairs who are users of a particular pad. Subjects who are "satisfied" users of, for example, the Stryker Pad or the "Bye Bye" Decubiti Pad, have been issued another pad. At the time of delivery, a preliminary questionnaire designed to elicit pertinent facts about time and use of wheelchairs and wheelchair pads was administered by the clinic team. After 1-month, 2-month, and 3-month periods, follow-up questionnaires are being administered by the clinic team. This information will provide data for the final report.

4. Howmet Wheelchair. A Howmet wheelchair of recent design, manufactured by the Howmet Corporation Hospital Division of Archbold, Ohio, was evaluated for compliance with current standards for wheelchairs (Fig. 23).

Deviations from standards are recorded below in accordance to their specification numerical listing.

Specification Requirements	*Deviations*
Leg rest Length (3.2.2)—The lowest point of the footplate to be at least 2½ in. from floor surface.	1¾ in.
Wheels (3.2.3.2)—Chairs equipped with 24 in. diameter drive wheels should weigh no more than 45 lb.	23⅞ in. diameter drive wheels with the chair's weight being 46 lb.
Dust Cap (3.2.3.2)—To be provided.	None provided, but the flanged hub may have been intended to serve as such.
Tires, Solid (3.2.3.2)—Compressibility at least 1 in. under a 200-lb. load.	¾ in.
Comfort (3.2.5.2)—With an occupant, gapping between seat and backrest should be no more than 1 in.	1⅞ in.
Comfort (3.2.5.2)—The chair shall provide some means to prevent occupant's feet from sliding off footplate.	No provision.
Ease of Operation (3.2.5.4)—The force to fold chair with one hand shall not exceed 15 lb.	25 lb.
Instructions (4.0)—Illustrations, operation, and care to be provided.	Attached tags describe nominal service operation.

Noted during durability tests was rearward deflection of the dual cross-frame members which prevented the seat rails from properly engaging their respective support brackets.

The deviations from the standards have been brought to the attention of the manufacturer.

C. Evaluation (Techniques)

None.

V. TESTING

A. Standards Development Program

1. Elastic Hosiery. In a previous issue (BPR 10–8, pp. 208–210), we described our efforts to develop adequately objective methods and instruments for measuring the compression forces applied to the lower limb by elastic hosiery. We pointed out that there was extreme variability in both the magnitude and the pressure gradients applied by stock samples of commercial elastic hosiery as measured by the Circumferential Tension

Approximation Method (CTAM) developed at the VA Prosthetics Center and described in BPR 10–6. Moreover, by this method we could not replicate the compression patterns described by manufacturers of custom elastic hosiery.

During the present reporting period, two new methods for measuring the compression forces applied by elastic hosiery have been tested. The first employs an electronic pressure transducer (VAPC Sensotec) consisting of a metal capsule 0.5 mm. thick and 6.35 mm. in diameter described in detail in Section I.A.1. The pressure transducer was placed directly on the lower extremity at a point 3 in. from the floor and the elastic stocking was carefully donned to avoid displacing the instrument. Pressure was recorded at this level and subsequently at points 1 in. apart above the malleoli up to the top of the stocking (Fig. 25). Shown in Figure 26, are the results of measuring the compression applied by a custom elastic stocking and a stock one-way stretch elastic stocking both of which were specifically ordered for the same subject. In general, higher pressures were recorded by the electronic transducer than previously by the CTAM. From a point approximately 10 in. above the ground to the top of each stocking, a desirable negative gradient was recorded for both the custom and stock hose. The pressure drop over his region was approximately 25 mm. Hg. or ½ lb./in. ^{2}. However, in the region between the malleoli and the mid-calf, undesirable positive gradients were recorded with pressure increasing approximately 25 mm. Hg. from the malleoli to the mid-calf.

The second method employed the "Komprit Compression Tester," a German product designed specifically for measuring the compression of elastic hosiery. It consisted of a rubber bag which is placed beneath the sock and a microswitch located outside the sock directly over the bag (Fig. 27). When air is pumped into the bag it deflects outward. This deflection is sensed by the microswitch which causes a light to go on and a buzzer to sound. At this signal the manometer on the tester indicating the hose compression is read.

When measured by this instrument, both the custom and the one-way stretch hose appeared to apply a neutral pressure gradient from 6 in. above the floor to the top of the sock (Fig. 28). (Below 6 in., the contour of the leg precluded the use of the Komprit; in order to obtain an accurate measure, the surface has to be relatively flat.) The magnitude of pressure was approximately 40 to 50 mm. Hg.

Three significantly different pressure readings were obtained by the three test methods applied. Of the three, the electronic method seems to be the most accurate and reliable.

2. *Comparative Analysis of Eight Wheelchairs.* In continuation of our program to develop functional specifications for wheelchairs, a large scale field evaluation program was completed. To control the quality of the

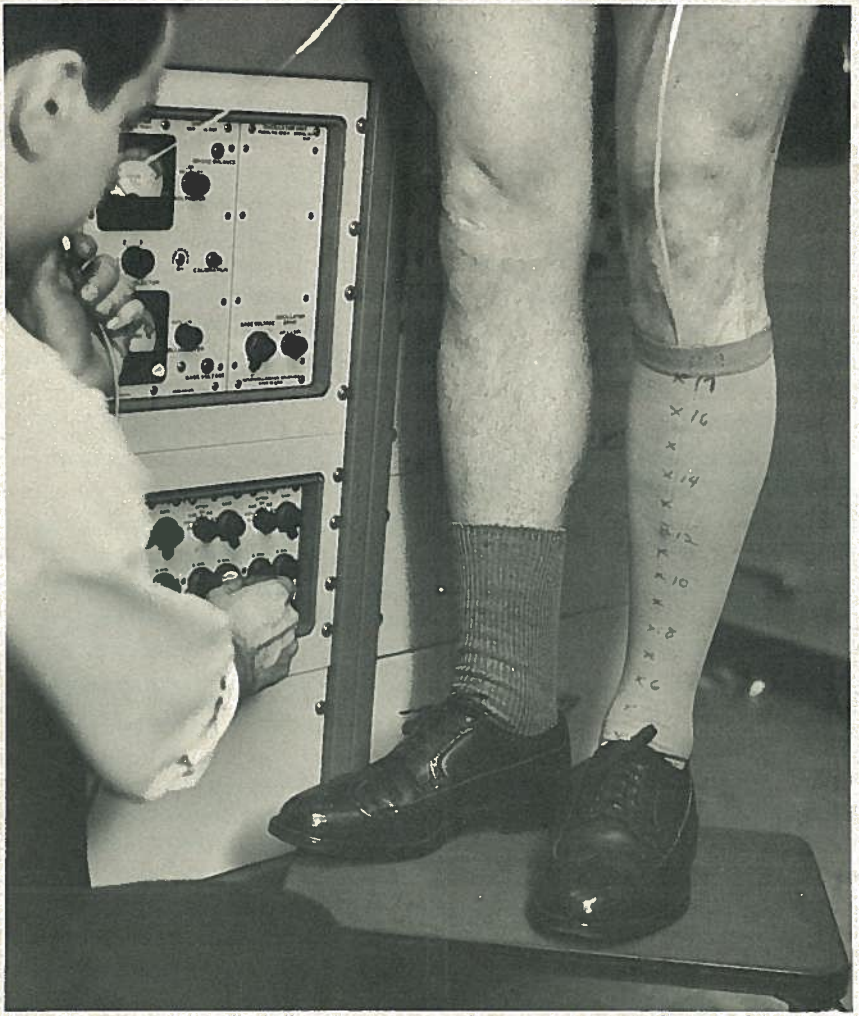

FIGURE 25.—VAPC-Sensotec pressure transducer used to measure pressure applied by elastic hosiery.

80 chairs (10 chairs each of 8 types) a physical inspection of each chair was performed prior to shipment to participating clinics. In addition one sample of each type (8 chairs) was checked against the complete range of tentative VA functional specifications. A test program for physical evaluation in the field was also developed together with appropriate equipment and instruments. The results of these tests will be reported in a future issue of the Bulletin.

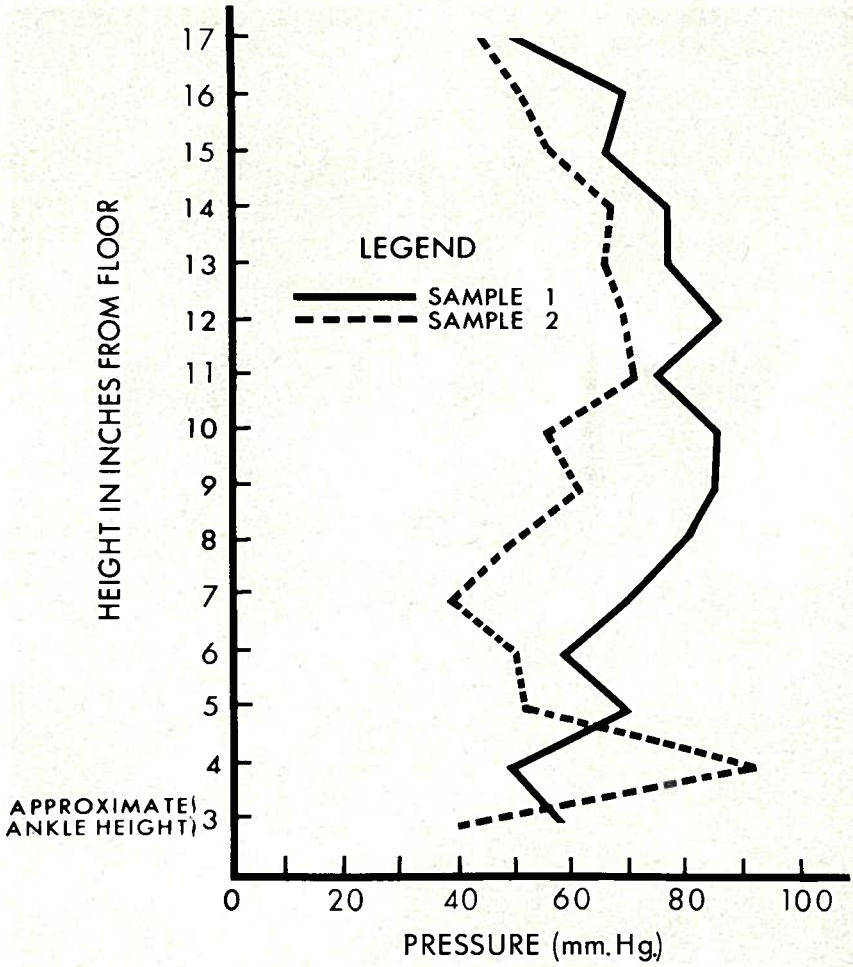

FIGURE 26.—Pressures applied by two types of elastic hosiery as measured electronically. Sample 1—custom elastic hosiery; sample 2—stock one-way stretch elastic hosiery.

B. Compliance Testing

1. Stump Socks

Knit Rite Post-Operative Stump Socks. Two samples of stump socks designed especially for immediate postsurgical application were evaluated against appropriate VA standards for stump socks and to determine compression characteristics. These tests were applied as a first step in our effort to develop standards for stump socks used in postsurgical prosthetics management.

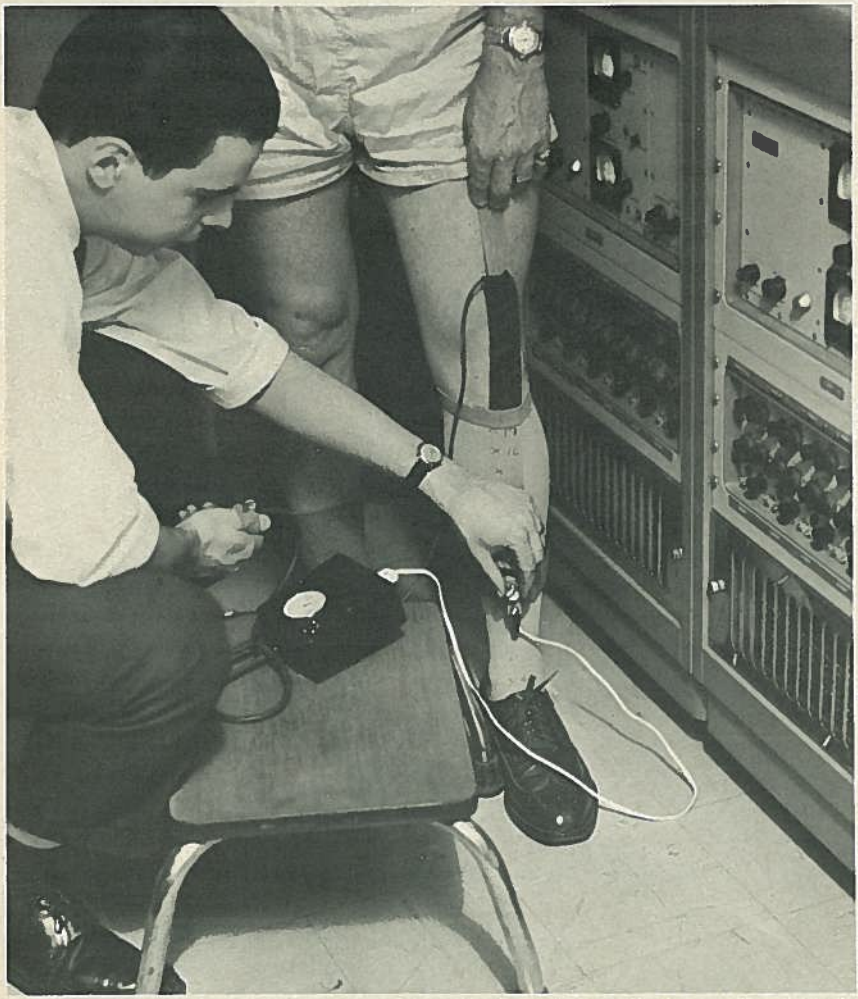

FIGURE 27.—"Komprit" pneumatic pressure meter used to measure pressure applied by elastic hosiery.

Both samples shrank excessively at a point 2 in. from the distal end and one sample in addition shrank excessively at a point 6 in. from the distal end. The fabric content of both samples is Orlon-acrylic fiber (95.25 percent) and Lycra-Spandex fiber (4.75 percent); thus, it is questionable whether these standards designed for 100 percent wool socks could be applied. Moreover, this type of stump sock is currently used in a one-time application in immediate postsurgical fitting. Its shrinkage therefore, after five washing cycles, is of less consequence than as in wool stump socks designed for continuous use.

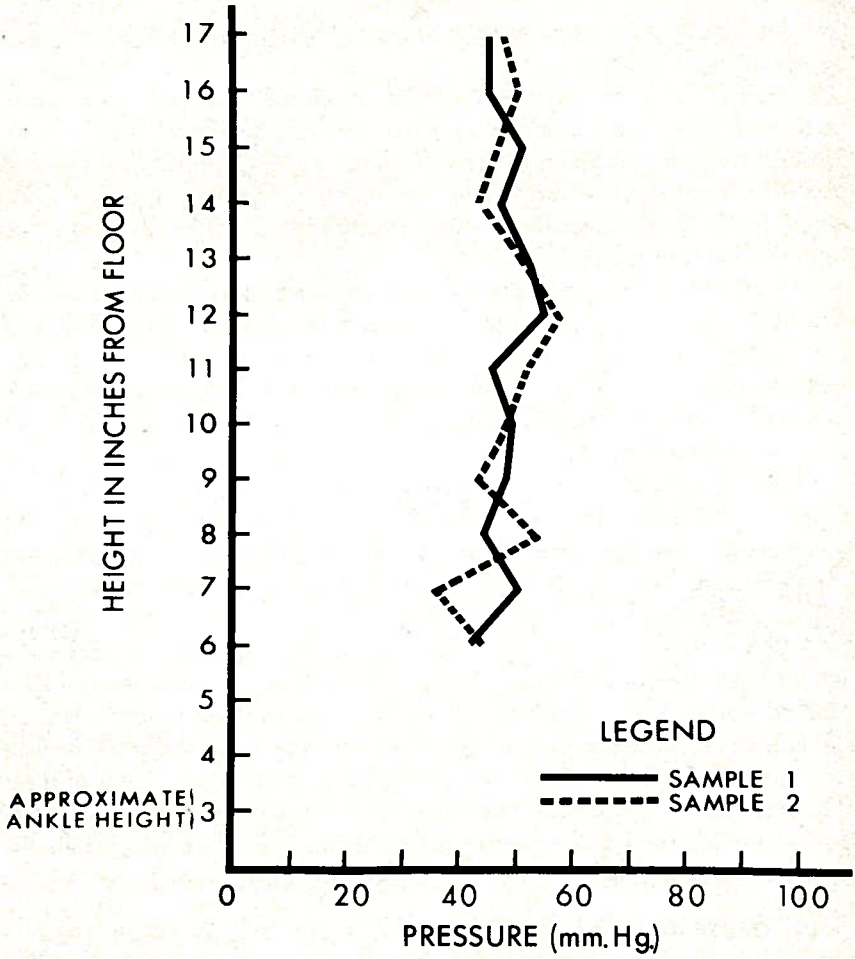

FIGURE 28.—Pressures applied by two types of elastic hosiery as measured pneumatically. Sample 1—custom elastic hosiery; sample 2—stock one-way stretch elastic hosiery.

Compression forces exerted by these socks are being measured by means of the VAPC-Sensotec pressure transducer.

2. *Upper-Extremity Components*

a. APRL #44 Voluntary Closing Hand. One model of the APRL #44 hand was checked for compliance with the Tentative Specifications for "Hands, Adult Size, Voluntary Closing." This terminal device, manufactured by the Sierra Engineering Company and distributed by the A. J. Hosmer Corporation, generally conformed to specification requirements. The force required to move the active fingers through their complete range

was marginally adequate however, ranging between 7 and 9.5 lb. The requirement is 6 to 8.5 lb.

b. APRL Voluntary Closing Hook. One model of the APRL hook manufactured by the Sierra Engineering Company and distributed by the A. J. Hosmer Corporation failed to meet currently applied specifications in that excessive force to close the fingers was required to obtain a secure grasp between the fingers (functional requirement III.B.5). The distributor has been apprised of the deficiency.

3. SACH Feet. During this period, four manufacturers submitted samples of SACH feet for annual compliance testing. The results of the tests indicated that three manufacturers' products were in substantial compliance with the requirements. The other sample could not be rated as acceptable in view of two significant deviations from the specification requirements: a. toe-break too short and b. toe-curl in excess of allowable limits.

In addition, samples of a new SACH foot were submitted by the Navy Prosthetics Research Laboratory for evaluation. The foot is constructed of a laminated wood keel and a molded sole and toe section. The unit was tested by comparison with applicable standards for SACH-type feet (VA Specifications No. VAPC–L–5902–3).

With two minor exceptions the foot met all the functional requirements set forth in the specifications. Toe-break fell short of specifications—feet marked size 13 were actually size 12 and the A and B reference lines were not indicated. The discrepancies were of very small magnitude and, apprised of them, the developer will have little difficulty in modifying them to meet requirements. At the present time these feet are not commercially available, but we are informed by the developer that arrangements are being made for commercial fabrication.

VI. OPERATIONS REPORT FOR FIRST HALF, FISCAL YEAR 1968

The VA Prosthetics Center experienced a significant increase in activity in the 6 months ending December, 1967, when "services rendered" totaled 44,459 as compared with all of Fiscal Year 1967, ending June 30, 1967, when total services provided (for the full year) were 63,821. Issues of new orthopedic shoes and shoe repairs were increased (Table 3). The use of temporary prostheses increased somewhat, as did the production of permanent prostheses, as reflected in Tables 5, 6, and 7. In all probability, the increases were due to the influx of wounded veterans from the Vietnam era and the increasing frequency of disabilities of age associated with veterans of World War II and the Korean conflict.

The data reflected in the following pertain to major services rendered veteran beneficiaries, during the first half of Fiscal Year 1968, by the VA Prosthetics Center's Orthopedic Shoe Service, Prosthetics-Orthotics Service,

and the Restorations Service. In most instances fiscal year comparisons are included.

A. The Orthopedic Shoe Service

Table 3 reflects full year fiscal comparisons of our national Orthopedic Shoe Program from 1964 through 1967 and including the first half of Fiscal Year 1968. We experienced an increase of 123 beneficiaries for the first half of Fiscal Year 1968. In addition we provided 159 veterans with ortho-inlay shoes, at a savings in excess of $7,000. We also issued 50 pairs of overshoes and rubbers.

TABLE 3.—*VAPC National Orthopedic Shoe Program*

	Fiscal year				
	1964	1965	1966	1967	First half 1968
Beneficiaries on rolls	10, 021	10, 322	10, 587	10, 861	10, 984
New shoes, prs.	7, 317	7, 336	7, 688	7, 181	3, 789
Prs. of new shoes issued per benef. on rolls per yr.	. 73	. 71	. 73	. 66	[a]. 69
Repaired shoes, prs.	9, 484	9, 422	10, 432	8, 932	4, 963
Prs. of shoes repaired per benef. on rolls per yr.	. 95	. 91	. 99	. 82	[a]. 90

[a] Projected for full fiscal year.

B. The Prosthetics-Orthotics Service

Table 4 reflects the activity in the Orthotic Components Unit of the Prefabricated Appliances Section relating to the distribution of surgical supports and elastic hosiery. Table 5 reflects a distribution of prosthetic components. In addition 8 cervical collars and 12 Knight spinal braces were distributed.

The Limb and Brace Section of the Prosthetics-Orthotics Service fabricated and delivered lower-extremity prostheses and braces as indicated in Tables 6, 7, and 8. Other items fitted are as follows: Arch Supports, 825; Artificial Arms, 17.

TABLE 4.—*VAPC Surgical Support and Elastic Hosiery Programs*

Issuance channel	F.Y. 1964		F.Y. 1965		F.Y. 1966		F.Y. 1967		First half F.Y. 1968	
	Surg. supp., ea.	Elas. hos., ea.	Surg. supp., ea.	Elas. hos., ea.	Surg. supp., ea.	Elas. hos., ea.	Surg. supp., ea.	Elas. hos., ea.	Surg. supp., ea.	Elas. hos., ea.
Directly to veterans	2, 434	9, 989	2, 482	9, 554	2, 634	9, 801	2, 697	10, 866	1, 145	5, 591
VA Orthopedic Shops	3, 995	12, 258	3, 722	11, 049	5, 885	14, 517	5, 056	15, 924	2, 243	7, 708
Totals	6, 429	22, 247	6, 204	20, 603	8, 519	24, 318	7, 753	26, 790	3, 388	13, 299

TABLE 5.—*Distribution of Prosthetic Components by VAPC, First Half F.Y.* 1968

Issuance channel	Temporary prostheses [a]			SACH feet	Hydraulic systems [b]			
	AK	BK	Total		Initial issue	Replacement issue	Total	Active wearers [c]
For specific beneficiaries	42	64	106	76	223	603	826	2287
VA Orthopedic Shops	51	63	114	172	—	—	—	—
Totals	93	127	220	248	223	603	826	2287

[a] In F.Y. 1967 we distributed 151 above-knee temporary prostheses and 212 below-knee prostheses respectively.
[b] In F.Y. 1967 we had 2081 active wearers of hydraulic systems.
[c] Seventeen wearers changed from one hydraulic system to another hydraulic system or were issued spares.

TABLE 6.—*Complete Below-Knee Artificial Limbs Fitted by VAPC, First Half F.Y. 1968*

Type	Permanent	Temporary	Total
PTB			
Cuff	14	4	18
Lacer	6	4	10
Carved wood	8	—	8
Molded socket, non-PTB	23	6	29
Syme	10	—	10
Chopart	1	—	1
Totals	62	14	[a] 76

[a] 150 below-knee artificial limbs were fitted in F.Y. 1967.

TABLE 7.—*Complete Above-Knee Artificial Limbs Fitted by VAPC, First Half F.Y. 1968*

Type	Permanent	Temporary	Totals
Molded socket, non-total contact	17	8	25
Molded socket, total contact	29	16	45
Carved wood socket	6	—	6
Hip disarticulation	7	—	7
Knee bearing	4	—	4
Totals	63	24	[a] 87

[a] 153 above-knee artificial limbs were fitted in F.Y. 1967.

TABLE 8.—*Braces Fitted by VAPC, First Half F.Y. 1968*

Item	Number
Below knee	73
Above knee	46
Arm	3
Spinal, prefabricated but custom fitted	52
Cervical collars	14
Total	[a] 188

[a] In F.Y. 1967, 539 braces were fitted.

C. The Restorations Service

Table 9 reflects the activity of the Restorations Service in its major prosthetics responsibilities.

TABLE 9.—*VAPC Production of Restoration Items, First Half F.Y. 1968*

Item	Number
Artificial eyes	176
Body restorations	21
Cosmetic gloves	28
Facial restorations, ear	2
Facial restorations, nose	3
Facial restorations, orbital	4
Plastic hands, full	12
Plastic hands, partial	10
Tinting of gloves	6
Repairs to appliances (all)	14
Other items or services	82

D. Special Clinic Team

The Special Clinic Team met 23 times during this period. Sixty veterans were referred by 19 VA field stations.

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NOTE: This report was prepared by Dr. Edward Peizer with the assistance of Donald W. Wright, Henry F. Gardner, William J. McIlmurray, and James Cohen.