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REPORT

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This report details the progress and status of studies conducted in various laboratories and services of the VA Prosthetics Center. It was compiled with the assistance of Donald W. Wright, Research Physiologist.

I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

1. Effects of Compression of the Lower Extremity. As reported in the previous issue we have been attempting to relate arterial blood pressure to gravitational and compressive forces applied to the leg. The purpose of these studies is to develop a simple measure of the physiological effects of compression applied by elastic hosiery. To date we have been able to demonstrate a fairly close correspondence between diastolic pressure in the dorsalis pedis artery and the normal gravitational forces applied to the limb. Diastolic pressure was reduced as a patient, initially positioned vertically on a tilt

table, was rotated to a horizontal position with respect to the ground (BPR 10–6). However, localized compression forces applied to the lower extremity by means of a pneumatic pressure bag (simulating the effects of elastic hosiery) did not produce corresponding quantitative changes in arterial pressure (BPR 10–7). Subsequent efforts were made to monitor arterial blood pressure continuously in the hallux by means of a Pressure Pulsometer using digital electrodes and digital compression during gravitational changes produced by tilting the subject.

These efforts have been unsuccessful principally because pressures in the hallux were too high to be recorded (300+ mm. Hg.) with the apparatus used. Even with the subject at a 45 deg. angle (midway between a horizontal and vertical position), pressures in the hallux reached approximately 260 mm. Hg., the maximum pressure that could be recorded. However, a definite relationship, not observed when measured manually, was noted between pressure in the dorsalis pedis artery and compression forces applied to the lower limb. As shown in Figure 1, there was a reduction in dorsalis pedis pressures as the applied compression forces were increased.

These results indicate no discernible effects after applying compression forces up to 25 mm. Hg. Between applied pressures of 25 mm. Hg. and 75 mm. Hg., a mild but definite effect is noted. At 100 mm. Hg. arterial pressure falls almost 20 percent, a drop which may represent a significant reduction in arterial blood flow and hence a limiting point.

To date these studies have not yielded results which could form the basis for a practical procedure for testing the effects of elastic hosiery in a clinical

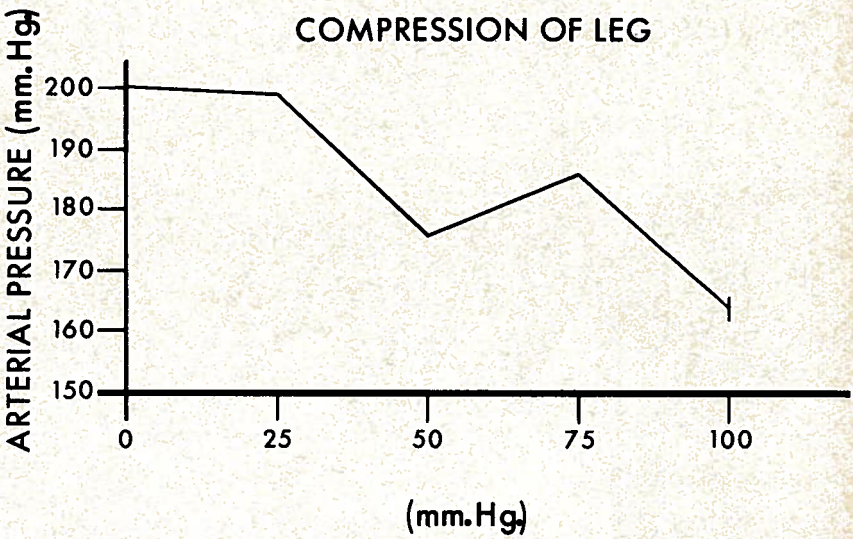

FIGURE 1.—The effect of localized compression of the lower extremity on the blood pressure in the dorsalis pedis artery.

setting, our original goal. Intra-patient variations in all of the measures employed often exceed inter-patient differences. Serious questions are thereby raised about the validity of the measures and the reliability of the techniques employed. Nevertheless, there remains the practical necessity for objectively measuring the compressive forces applied by elastic hosiery. To develop methods for evaluating and guides for prescribing elastic hosiery we need to know: What pressures do various types of elastic hose apply? What are the most appropriate pressure ranges? Are gradients necessary?

Since we have not succeeded in adequately measuring the *physiological effects* of elastic hosiery we have shifted the emphasis of these studies to the development of a practical clinical instrument for measuring pressures at the interface of the limb and elastic stocking. A relatively crude technique has been devised for this purpose. The apparatus employed consists of a standard sphygmomanometer and a digital compressor bag ordinarily used with the Pressure Pulsometer. The bag is inflated by means of a bulb to 10 mm. Hg. and placed on the medial blade of the tibia avoiding the surrounding soft tissue. The elastic stocking is donned pulling it over the bag carefully. The pressure in the bag is increased by the circumferential forces exerted by the stocking and the difference is read on the manometer. Initial experiments with this crude model indicated the feasibility of this technique. A refined model employing three bags is being fabricated. In addition, a similar device is being purchased from a commercial source (Fig. 2).

2. *Work and Energy in Walking.* As part of an extensive study to relate energy expenditure of walking at various velocities to physical and physio-

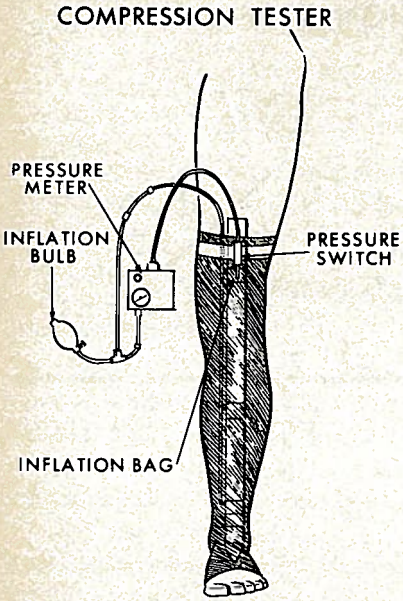

FIGURE 2.—Device (Kompritest) for measuring the compression effects of elastic hosiery at the ankle, midcalf, and knee.

logical variables (BPR 10–7), we are studying the efficiency of walking when limitations are imposed on joint motions. Among the very obvious joint limitations attendant on above-knee amputation is the absence of normal knee flexion in stance phase. Eberhart et al. (1) have described normal knee flexion in stance phase as a major energy-conserving factor in normal gait. The substructure of this theory related vertical oscillation of the body center of gravity (c.g.) to work done in walking and to its corollary energy cost. The normal knee is described as flexing approximately 20 deg. between heel-contact and foot-flat, an event which permits the body c.g. to traverse a rather flat sinusoidal path (Fig. 3), whose peak-to-trough magnitude is approximately 2 in. (2). Twenty degrees of knee flexion during the heel-contact to foot-flat period should reduce the vertical displacement of the body c.g. by approximately ½ to 1 in. The work done should also be reduced to an extent roughly equal to the body weight times ½ to 1 in. at each step. When the knee does not flex in stance phase, the vertical displacement of the body c.g. is greater. This may account for a substantial portion of the approximately 100 percent higher energy costs of walking for the above-knee amputee (BPR 10–1). These data are of interest in pointing to the possibility of designing prostheses which require less energy on the part of the patient. There is no device available today which is designed to provide the *function* of the knee flexion in early stance phase.

A great deal of work has been done on the design and development of knee mechanisms to increase stance phase *stability, e.g.,* Mauch, Regnell, Bock, etc. Most such mechanisms are designed either to lock positively or

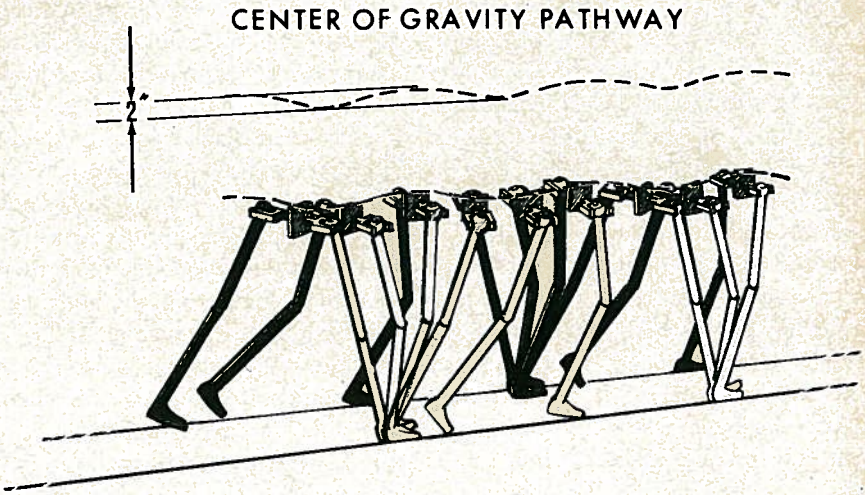

FIGURE 3.—The effects of several factors on the vertical pathway of the center of gravity (after Eberhart et al.).

to develop high resistance under weight bearing in early stance. They are often described as providing "increased security" and "more stability." By providing high resistance to flexion moments about the knee in stance, they reduce dependence on alignment for stability and/or require the patient to exert less force to stabilize the knee by extension of the hip. None is designed specifically to permit flexion of the knee in early stance. Yet there is evidence to indicate the possibility of significantly reducing energy cost of walking if this were possible.

It is a difficult design problem to duplicate the range and phase of normal knee flexion in early stance. Since the effect of flexion in early stance is to reduce the magnitude of the vertical motion of the body c.g., the same effect may be achieved by permitting the prosthesis to telescope, i.e., to shorten without requiring rotation about the knee joint. A limited experimental program was undertaken to test the concept that a reduction in the magnitude of the vertical c.g. oscillation may be reflected in reduced energy cost of walking. A simple device called a flexion compensator (Fig. 4), installed in the pylon-type shank of an above-knee prosthesis, permitted approximately 1 in. of vertical displacement by the telescoping action of its distal and proximal sections. Body weight applied through the compensator overcame the resistance of a calibrated spring adjusted to deflect under the patient's body weight at a rate corresponding to the application of the first peak of the vertical load curve. Force plate and cyclographic records were made of the subject's gait patterns.

Reducing the vertical rise of the body c.g. to a magnitude which the normally flexing knee produces resulted in significant differences in the gait pattern of an above-knee amputee. As shown in Figure 5, the manner in which the patient applied his body weight to the ground in stance phase appeared to become more normal. His characteristic trimodal vertical load pattern became a more normal bimodal phenomenon.

Figure 6 shows that the typical pattern of vertical oscillation of the body c.g. in the anterior-posterior plane was different when the effects of normal knee flexion were simulated. Instead of the two typically symmetrical peaks of approximately equal amplitude, peaks of unequal magnitude were recorded on both the prosthetic side (Fig. 6A) and the sound side (Fig. 6B) when using the flexion compensator. As shown in Figure 6A, the body c.g. did not rise as high when simulated knee flexion was available. At heel-contact the compensator began to compress preventing the c.g. from rising to the same height it reached when the compensator did not function. After heel-contact on the sound side (Fig. 6B) (double support period) the hip on the prosthetic side rose sharply as weight shifted to the sound side and the flexion compensator spring extended. However, the hip on the prosthetic side did not rise as high as it did without the compensator. Figure 6B shows the effects of the flexion compensator on the vertical

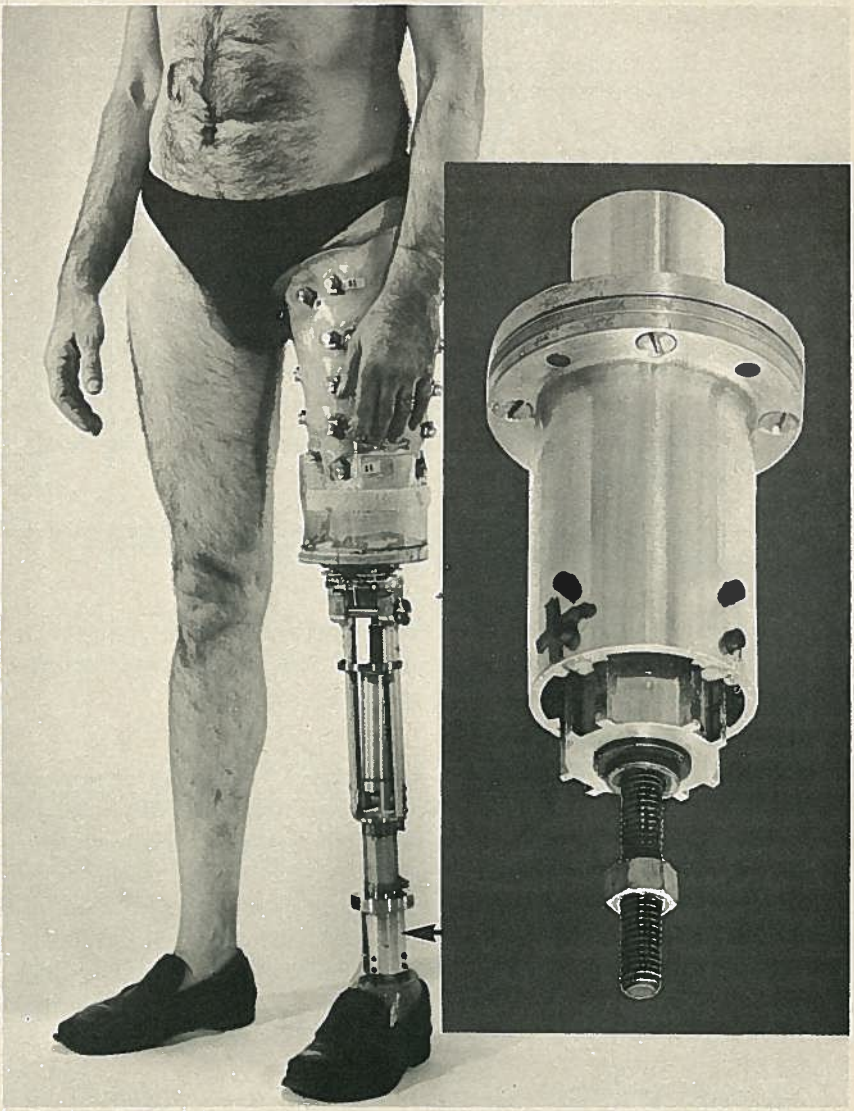

FIGURE 4.—The VAPC flexion compensator installed in the pylon of an AK prosthesis (socket shown was used in N.Y.U.'s pressure studies described elsewhere in this issue) simulating the function of normal knee flexion during stance phase.

motion of the sound hip. The path of the c.g. remained consistently lower when the compensator was operating. The rapid oscillation noted between heel-contact and foot-flat may result from the combined effects of the spring extension and the unaccustomed mode of weight transfer. Whether these alterations were of permanent or temporary nature is unknown since

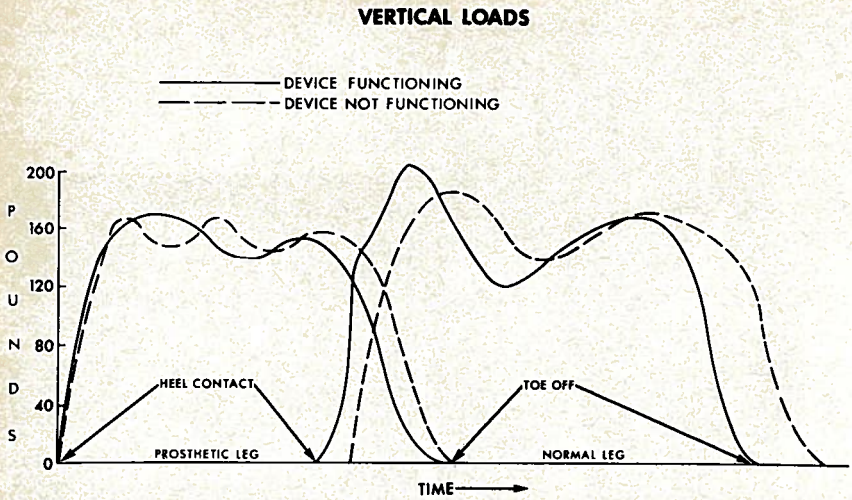

FIGURE 5.—Simulated knee flexion in early stance (flexion compensator) produces a more normal bimodal vertical load pattern on the prosthetic side.

the subject was only permitted to wear the experimental device in the laboratory. These studies are currently being continued to determine the reliability of these changes and to relate them to energy costs.

B. Development (Components)

1. Adjustable Below-Knee Standard Prosthesis. The design of this unit was frozen at the point described in the previous issue, i.e., with a conventional socket attachment plate and with another plate especially designed for immediate postsurgical fitting. The first production run of the below-knee prosthesis (100 units with accessories) has been received from the Pope Brace Company. These units are being distributed to VA Orthopedic Shops throughout the country for use as temporary and immediate post-operative prostheses. Although several types of cosmetic covers have been fabricated, we have not yet completely standardized this component (see Section I.B.3.).

2. Standard Above-Knee (Multiplex) Prosthesis. Three additional units of the prototype model of the standard above-knee prosthesis, described in the previous issue, are being fabricated. On the basis of limited testing, production engineering of the current design is being undertaken. It is anticipated that a completed above-knee prosthesis, incorporating the Standard Above-Knee (Multiplex) unit, will weigh significantly less than similar conventional units. The main weight reducing element is the elimination of the heavy knee block and wood shank used in conventional and hydraulic setups. Thus, the multiple special features of the device will be available without cost in weight.

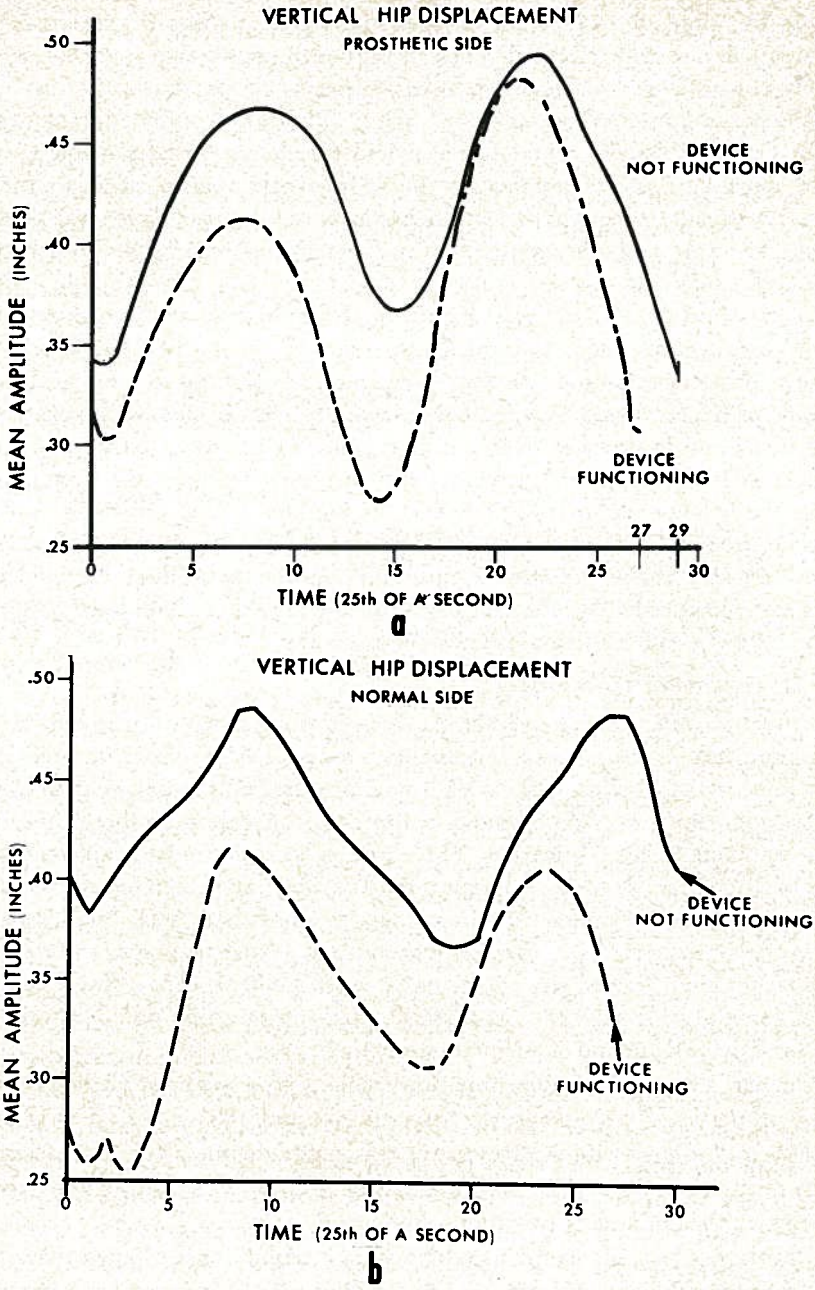

FIGURE 6.—(a) Simulated knee flexion (flexion compensator) did not permit the body c.g. to rise as high in stance phase as walking with an unflexing knee. The vertical height of the c.g. was approximately the same in swing phase. (b) The vertical rise of the body c.g. was smaller with simulated knee flexion.

At the time of production engineering, stress analyses of the current design will also be performed. These may result in additional weight reductions. Nevertheless, this device, besides representing a standard structure for a number of knee and foot-ankle functions, will result in a lighter prosthesis.

3. Cosmetic Covers. Standardized pylon-type shanks for both below-knee and above-knee prostheses have reached a stage of development where they are functionally adequate and are being increasingly used in general practice. However, a significant problem still remains as regards the design of cosmetic covers. It has been difficult to find materials with suitable stain resistance and durability. Designing adequate methods of attachment to the foot-ankle section and standardizing sizes have been particularly troublesome. As described in the previous issue, tentative solutions to some of these problems have evolved. A vinyl cover has been chosen which is thicker at the distal end in order to maintain its shape over the distal portion of the pylon. It is thinner at the proximal end to permit heating and stretching over the below-knee socket. The cover of the above-knee prosthesis is attached proximally by means of a retaining ring. We have tentatively standardized cosmetic covers in basic adult male sizes for the right and left sides. We have procured 12 molds, six each for below-knee and above-knee covers. Three sizes in both rights and lefts will be fabricated.

C. Development (Techniques)

Direct Forming of Below-Knee Sockets. Forming sockets directly on amputation stumps is a potentially valuable procedure which offers possibilities for improved socket fit, easier socket modification, substantial reduction in fabrication time, and more readily mastered techniques than those used in conventional socket fabrication. The entire process depends completely on use of a material which: a. is highly ductile at moderately high temperatures, b. is easily worked under conditions found in most limb shop facilities, c. has a "poor memory," i.e., once set it should not change its shape, and d. exhibits minimal "creep" in conventional environments. Apart from these fundamental properties the required material must also be nontoxic, odorless, relatively light, and of adequate strength.

Polysar, a synthetic similar to natural rubber, possesses most of the necessary properties. At temperatures between 160 and 180 deg. F. it is quite ductile and it cools rapidly. It remains reasonably ductile after its temperature drops 20 to 30 deg. F. and thus can be applied to a stockinet-covered stump within a minute or two after softening.

Laboratory tests indicate that after setting it maintains its shape even under stress in ambient temperature up to 120 deg. F. Other tests have shown that conventional fastenings, rivets and screws, are adequately retained so that it is possible to use all conventional components and accessories with Polysar sockets (BPR 10–7). The raw granular material, purchased from the

Polymer Corp. of Sarnia, Ontario, is extruded by a commercial firm to form tubes for fabricating sockets.

Clinical experience to date has yielded satisfactory results in fitting patients with both temporary and permanent sockets *made* of Polysar and formed directly on the stump.

The material lends itself conveniently to local relief and appears to be sufficiently stable in below-knee applications to maintain its dimensions. As a result of these applications, a fabrication technique employing the VAPC pneumatic casting system has been standardized. To date twelve patients have been satisfactorily fitted. An illustrated manual has also been prepared detailing the steps in the current fabrication procedure. This preliminary manual is designed to assist other research groups to evaluate this technique independently. On the basis of these experiences the manual will be modified as required for general use.

D. Evaluation (Components)

1. Follow-up Evaluation of UC–BL Pneumatic AK Swing Control Mechanism. Since the basic report (BPR 10–6), several modifications in this unit have been made by the developer, and additional laboratory test experience has been obtained. Of the three modifications, two (replacing the attachment bushings and increasing the diameter and hardening of the piston rod) did not relate directly to the function of the unit. The third modification established a critical length for the piston rod of 7.589 in. ±0.010 in. To date, six veteran amputees have been fitted with the earlier model of this device and two of the modified units have been tested in the laboratory.

All six patients reacted favorably to the use of the UC–BL pneumatic knee with respect to ease of walking, resistance characteristics, weight, and extension aid. The ranges of useful resistance at several adjustment settings and at moderate cadences of approximately 86 steps per minute are shown in Figure 7. Biomechanical analyses of knee function indicated that the unit provides a smooth, normal appearing swing pattern which closely corresponds to the angular displacement pattern of the normal knee with respect to phase, magnitude, and shape (Fig. 8).

Optimum resistances to extension and flexion are adjusted somewhat differently than in other similar mechanisms. Resistance to extension (terminal impact) is adjusted by means of the Level Valve. Once extension resistance has been adjusted adequately, it may be necessary to make further adjustments for flexion resistance. This adjustment is also done by means of the Level Valve. However, when the Level Valve is used to adjust for flexion resistance, the original setting controlling extension resistance is changed. This change is compensated for by means of the Ratio Valve. To obtain maximum functional efficiency from this unit, adjustment instructions must be carefully followed.

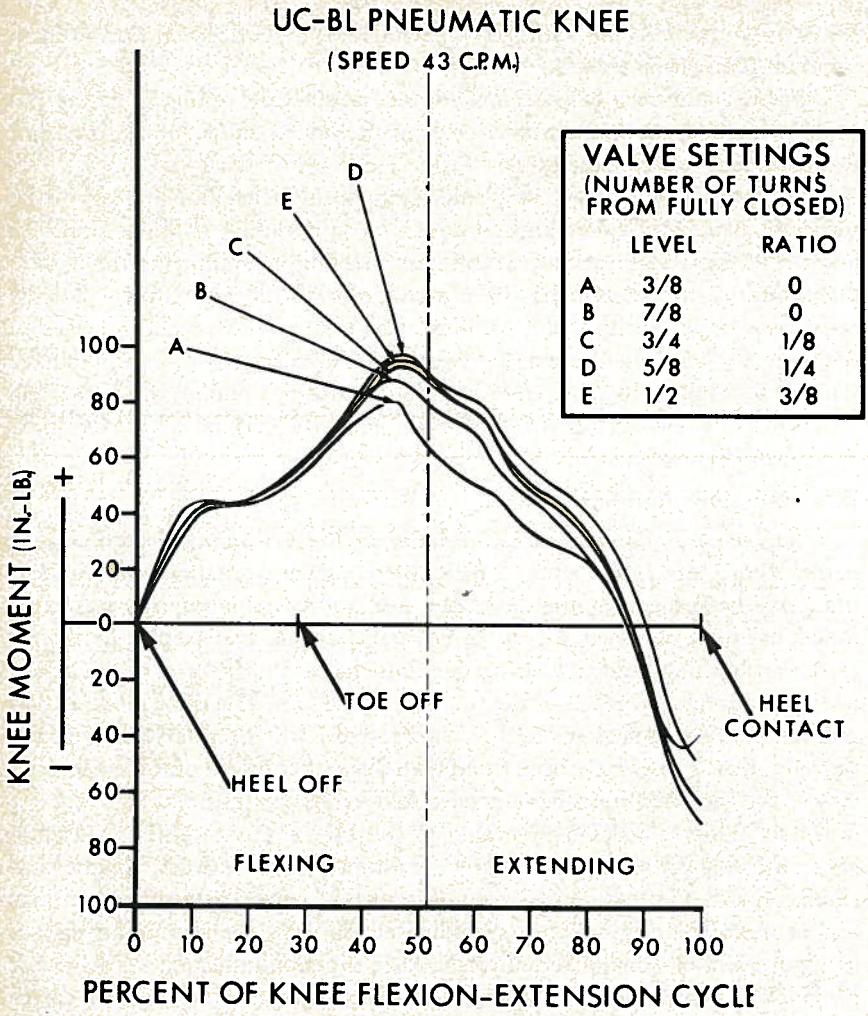

FIGURE 7

2. *Follow-up Evaluation of the Navy Intermittent Friction Knee.* Submitted by the Navy Prosthetic Research Laboratory, this unit was previously evaluated on the basis of mechanical tests and clinical fitting experiences. Reported in BPR 10–6 were the results of the inspection procedures and of the functional testing to determine resistance characteristics. At that time no clinical data were available. Since then, one subject who had worn the Navy Intermittent Friction Knee for 5 months was evaluated. He had previously worn a single-axis knee with hickory-stick extension-control, a conventional socket with flexible hip joint, and waist belt. Except for the

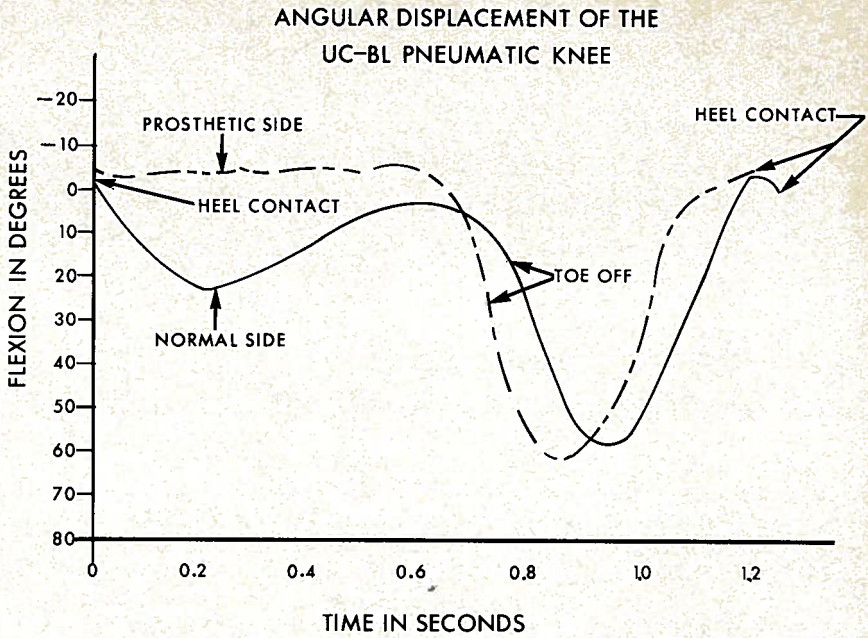

FIGURE 8

knee unit, the new prosthesis incorporated the original components and alignment. No additional accessories were required.

Installation of this knee unit was performed in the conventional manner without difficulty. The range of resistance adjustment was satisfactory, and once adjusted no additional resistance adjustments were required during the first 3 months of wear.

A biomechanical analysis of knee angular displacement patterns during swing phase indicated that the Navy Intermittent Friction unit provided a consistently smooth swing pattern in both flexion and extension (Fig. 9). As finally adjusted for the subject, knee angular displacement patterns with the prosthesis were quite similar to those of the sound leg with respect to shape and phase. However, the range of rotation, i.e., the magnitude, was significantly greater in the prosthesis, the maximum flexion range being approximately 80 deg. with the prosthesis as against approximately 60 deg. for the sound leg. This variation was attributed to the rather atypical knee flexion pattern of the subject's sound leg. He did not flex his sound knee significantly in stance phase and appeared to restrict the range of flexion of the sound knee in swing phase. The Navy unit could well have been adjusted to reduce this asymmetry but the subject preferred the observed pattern.

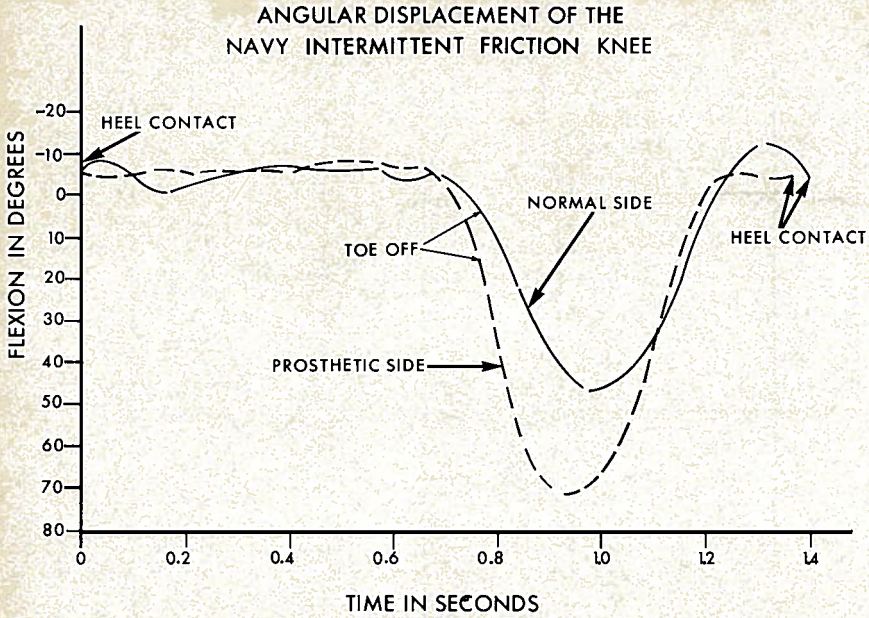

FIGURE 9.—The Navy Intermittent Friction Knee mechanism provides a relatively smooth, normal appearing knee flexion-extension pattern.

After more than 90 days of wear, the subject stated unequivocally that he preferred the Navy unit to his previously worn conventional unit. He had found it "less tiring," it permitted him to walk with "a more natural gait," and he no longer required intermittent use of a cane. During the period of wear, he made no adjustment of the resistance setting, and the unit functioned without breakdown.

The Navy Intermittent Friction Knee is a reliable, easily installed and relatively maintenance-free device. It provides a significant improvement over conventional mechanical friction knee mechanisms.

3. Evaluation of Teufel DAW Protective Nylon Sheath. The sheath is essentially a nylon stump sock designed to be worn under conventional wool stump socks as a means of reducing friction between the stump and the sock, thereby avoiding transmission of shearing forces to the skin of the stump. A European product supplied by the Teufel Company of Stuttgart, West Germany, the sheath is distributed in this country by Southern Prosthetics Supply Company of Atlanta, Georgia. It is identical to another product known as the Rem-DAW Sheath and consists of a rather finely woven nylon stump sock. The Teufel sheath is ordered in appropriate sizes on the basis of stump length and girth measurements. The proximal portion of the sheath extends above the below-knee or above-knee socket rim, and the wearer is instructed to fold the protruding material carefully over the

prosthesis after the stump covered with the nylon sheath and the conventional wool sock are inserted into the prosthesis.

According to the supplier's literature the benefits to be derived from the use of the sheath are that it: 1. reduces friction, 2. protects the stump, 3. prevents sores or irritations, and 4. aids evaporation of perspiration.

To evaluate the utility of this product, 14 veteran amputees wore the Teufel sheath for periods ranging from 2 to 60 days. The subjects ranged in weight from 135 lb. to 235 lb., and in height from 5 ft. 3 in. to 6 ft. 3 in. Included were four above-knee, six below-knee, one above-knee/below-knee, two below-knee/below-knee and one below-knee/Syme amputees. Of the 12 below-knee *stumps* in this sample, five were fitted with conventional below-knee sockets and seven with patellar-tendon-bearing sockets. In general the sample was a fairly typical cross section of World War II veterans with long experience in the use of stump socks and prostheses. All the subjects normally wore wool stump socks. Twelve among them wore one stump sock each, while the other two wore 2 each, indicating that they all had relatively good well-fitted sockets. None of the subjects had known allergies; five among them had extensively scarred stumps.

Reactions of the subjects to wearing the Teufel sheath were quite mixed. Seven of the 14 subjects rejected the sheath after periods of wear ranging from 2 to 30 days. Five veterans who had worn the sheath from 20 to 90 days reacted positively and indicated their desire to continue using it. Two veterans were undecided but did indicate their willingness to continue wearing the Teufel sheath if, in one case, a smaller sized sheath could be provided, and in the other, if the top of the sheath could be modified to accommodate the hinge of the hip joint he wore.

The five subjects who preferred to use the Teufel DAW Sheath substantiated several of the claims made for it. They experienced less friction between the stump and the sheath, they were more "comfortable"; sore spots and irritations on their stumps seemed relieved; and three of the five indicated that they had perspired less. All of these patients were unilateral below-knee amputees, three of whom wore patellar-tendon-bearing sockets. In general, the reduction of perspiration and abrasion was the major reasons for preference of the sheath.

Among the seven subjects who rejected the Teufel sheath, the principal reason for rejection was that wrinkles formed in the sheath due to its inability to conform to sharply tapering stumps. One patient complained of excessive piston action of the stump in the socket, another indicated that the sheath did not alleviate irritations, and a third, that it tended to cause ingrown hairs.

Use of the Teufel sheath seems to be of significant value to unilateral below-knee patients, especially during periods of heavy activity. In addition, its use seemed to reduce the effects of perspiration in this group. In general

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the above-knee amputees and bilaterals of various types in this study were not adequately served by the Teufel sheath. Since wrinkling was a recurrent reason for rejection in this group, further study seems indicated to determine if the sizing procedures and donning instructions are adequate.

E. Evaluation (Techniques)

Objective Checkout Procedures. There is a general and strong tendency to think of improvement in prosthetic and orthotic service principally in terms of hardware design and development. Equally important in high quality service to patients is the matter of evaluating the quality of a finished appliance and the ability of the patient to use it to its full potential. This is undoubtedly a matter of good clinical practice involving personnel, facilities, and procedures. All responsible clinicians take great pains to insure adequate training of their personnel, and to acquire the best available facilities to aid them in prescribing and checking out prosthetic and orthotic devices. Despite these advantages, and through no fault of clinical personnel, currently applied checkout procedures for artificial limbs may be inadequately applied or unintentionally neglected.

The several systems for checkout of artificial limbs are all rooted in the basic clinical evaluation concepts developed early in the artificial limb program at the University of California at Los Angeles. These checkout systems are valid instruments, having stood the test of time under the exacting scrutiny of patients and clinicians alike. No serious argument can be raised against the content of any of the current checkout systems, but much can be said against their length. Due to innovation in hardware and fitting techniques, minor changes have been wrought over the years in matters of content and format. Although there is little disagreement about the content of the systems in use today at the University of California at Los Angeles, Northwestern University, New York University, and generally throughout the country, they all remain excessively time consuming, tedious and inconvenient.

The heart of all of these systems is the evaluation of three fundamental features which determine the value of the integrated prosthetic-patient system: function, appearance, and comfort. All current checkout systems are organized to elicit information enabling the clinician to judge whether the prosthesis in use by the patient meets these minimum standards of function, comfort, and appearance. The simplest complete checkout guide used by many VA installations contains 73 items, each of which must be examined and noted in checking out a below-elbow prosthesis. The checkout guide for above-knee prostheses contains 134 separate items. It is generally conceded that a complete checkout of such a limb can easily occupy an hour of a trained clinician's time. With increasing centralization of prosthetic service in amputation centers, the case-load volumes often make it impossible to check each limb thoroughly by means of these procedures. Nevertheless,

checkout of artificial limbs is the final and perhaps the most crucial step in adequately serving the patient. We are, therefore, faced with the problem of reexamining the structure of current checkout procedures with a view toward consolidation and integration without compromising the standards.

A checkout program in which a patient performed a small number of relatively simple tasks while a trained clinical observer noted and recorded the prosthetic flaws and training inadequacies would be ideal. The closest approach to this situation is the subjective gait analysis performed as part of the lower-extremity checkout. The clinician observes the patient's gait from front, rear, and side and identifies "gait deviations" which he then relates to training requirements or prosthetic deficiencies. This procedure is entirely adequate in the hands of highly trained observers, but even then it remains excessively subjective, especially as regards evaluation of the *degree* of deficiency noted. After all, merely noting a "gait deviation" is valueless unless it leads to a decision about the probable cause and what action, if any, is required to remedy the deficiency. Even more difficult, and more subjective, is the matter of judging the adequacy of appearance, a condition which often leads to lowering of standards.

A preliminary analysis of the problem indicates that there is little hope of significantly reducing the time requirements and the special skills needed for adequate checkout by merely juggling or restructuring the existing checkout guides. It may be possible to evaluate patient performance objectively and to base an evaluation of the man-machine system on the obtained data. The key elements in prosthetic construction and function might be analyzed in terms of a small series of simple tasks which a patient can readily perform in the clinic. The tasks themselves may in some cases be practical, everyday operations. In others, they may be abstract tasks designed to require the patient to produce a variety of forces and motions which indicate prosthetic adequacy.

In any case this is a formidable task in which success can certainly not be predicted. It will deeply involve the contributions of prosthetists, developers, test construction specialists, engineers, and scientists. Nevertheless, the VA Prosthetics Center will undertake a study of this problem.

II. UPPER-EXTREMITY PROSTHETICS

A. Development

Humeral Rotator. The electrically driven humeral rotator described in the previous issue of the Bulletin has been fabricated, installed in an above-elbow prosthesis, and fitted to a highly experienced patient. The subject is a unilateral above-elbow amputee with a short, conical stump, who represents a typical though certainly not ideal patient for this application. After initial fitting of the device, a training program was undertaken to teach the amputee to employ his limited residual humeral rotation to actuate two shear switches,

one mounted anteriorly and the other posteriorly through the socket wall (Fig. 10 and 11). The switches are adjusted so that they bear lightly on the surface of the stump inside the socket. Internal rotation of the stump produces clockwise shear forces on both switches, a condition which actuates the electrical motor and rotates the elbow turntable internally. External rotation of the humerus inside the socket moves both switches in a counterclockwise direction, reversing the direction of the motor and producing external rotation of the forearm at the turntable. Inadvertent forces applied

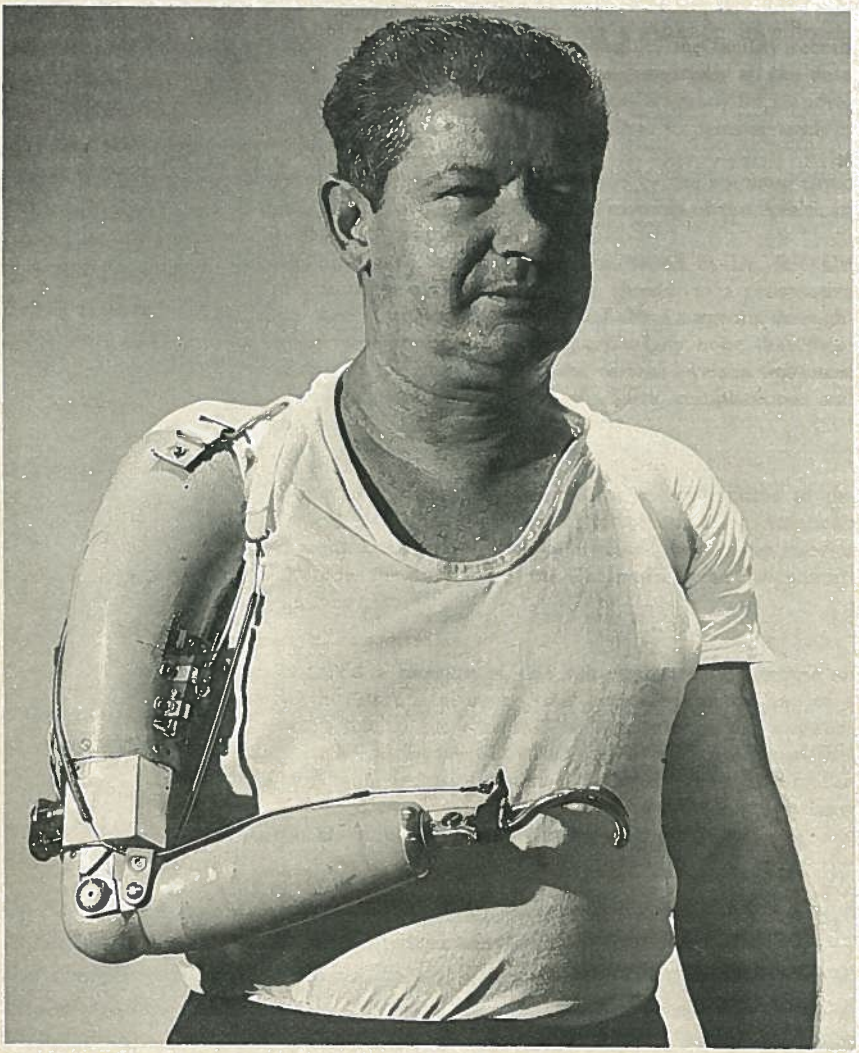

FIGURE 10.—A laboratory model of the VAPC Humeral Rotator designed to study the need for humeral rotation among AE amputees.

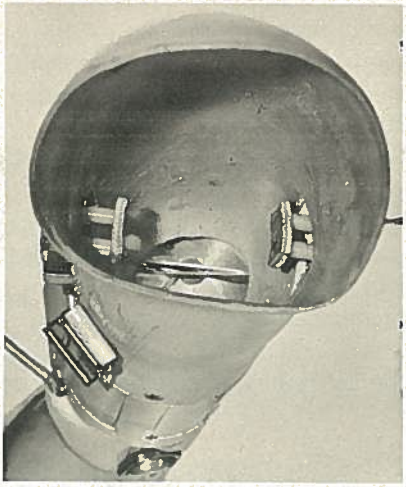

FIGURE 11.—Shear switches controlling humeral rotation.

to either switch alone, or those forces which move the switches in opposite directions, have no effect.

The training program is aimed at teaching the subject to coordinate elbow flexion-extension and internal-external rotation of the forearm in the performance of both abstract tests and practical tasks. When training now in progress has reached an appropriate level, a testing program will be initiated to determine the potential utility of providing humeral rotation in above-elbow prostheses.

B. Evaluation (Components)

1. AIPR (American Institute of Prosthetic Research) Externally Powered Components. As reported in BPR 10–7, we have completed a comprehensive study on one bilateral shoulder-disarticulation amputee fitted with the AIPR externally powered system. A comparative analysis was conducted over a two-year period of the amputee's subjective reactions and functional capabilities with bilateral conventional prostheses and bilaterally fitted AIPR systems. All the available AIPR components were systematically substituted in the conventional arms in a series of steps (a through h) until both arms were complete AIPR systems. Subjective reactions were obtained and performance evaluations were carried out at each step. Also included were fundamental kinematic studies and performance analyses by advanced photometric techniques. All the test results obtained in these studies and the conclusions drawn therefrom are based on comparisons with a baseline represented by the original fitting of the patient with conventional prostheses. To test the validity of these comparisons after almost two years, the patient was refitted bilaterally with conventional arms, retrained and asked to wear them for a period of three months, to reestablish the baseline reference (step h).

In review, the chronology of the steps a through h follows:

a. Initial fitting and use of conventional prosthesis on right humeral-neck side and shoulder-disarticulation cap on the other side—December 18, 1963 to May 16, 1964 or 5 months.
b. Fitting and use of conventional shoulder-disarticulation prosthesis on left side with shoulder abduction joint replacing shoulder cap, no change on right side—May 16, 1964 to July 7, 1964 or 2 months.
c. Substitution of AIPR CO_2 powered hook and wrist rotation unit for conventional units on the right humeral-neck side, no change on left side—July 7, 1964 to September 24, 1964 or 2½ months.
d. Refitting with Münster-type socket and conventional terminal device, wrist rotator, and elbow on the right humeral-neck side, no change on left side—September 24, 1964 to February 15, 1965 or 5 months.
e. Substitution of AIPR CO_2 powered hook and wrist rotation unit on right, no change on left side—February 15, 1965 to May 24, 1965 or 3½ months.
f. Substitution of a powered elbow for the conventional one on the right humeral-neck side, no change on left side—May 24, 1965 to June 28, 1965 or 1 month.
g. Completely powering the left shoulder-disarticulation prosthesis—June 28, 1965 to March 29, 1967 or 21 months.
h. Refitting with Münster-type socket and conventional components on the right humeral-neck side and a conventional shoulder-disarticulation prosthesis on the left side—March 29, 1967 to September 6, 1967 or 5 months.

Three and one-half months after refitting the subject with the conventional prostheses, laboratory testing indicated that his performance was indeed comparable to his previous performance using body-powered components (step d) (see Table 1). The subject repeated his previous experiences, viz. 1. the conventional (rubberband) terminal device had not provided sufficient prehension force to permit the "live-lifting" of certain objects to the area around his face, and 2. the absence of powered wrist rotation precluded certain self-care activities, and made others much more difficult (brushing teeth, shaving). In summarizing his experiences, he expressed a very strong preference for the AIPR system, particularly on the dominant (right) side.

With these findings, the laboratory portion of this program has been terminated. The patient will be refitted bilaterally with new AIPR prostheses.

This study resulted in five clearcut indications for design changes in current AIPR components or for the design of others, as follows:

a. Increase speed of elbow flexion-extension.
b. Design a wrist flexion unit.

TABLE 1.—*Summary of Test Results*

Tests	Body-powered components (Step d) February 1965	Body-powered components (Step h) June 1967
Prehension		
Total time, min. No. of errors Appearance	2. 8 1 Excellent—	3. 3 6 Good
Positioning		
Total time, min. No. of errors Appearance	5. 4 0 Good+	5. 4 1 Good+
Practical activities (7 tasks)	5 completed 2 failures	5 completed 2 failures
Total net energy cost of prehension test	.992 liters 0_2	2.489 liters 0_2
Total time, min. No. of errors Appearance	3. 8 3 Good—	4. 5 9 Poor

c. Design a humeral rotator.
d. Redesign the valve mounting system.
e. Study feasibility of correlated functions: elbow flexion and wrist rotation; elbow flexion, wrist rotation, and wrist flexion; humeral rotation and wrist flexion.

Other significant findings relate to fitting techniques, training methods, and evaluation procedures. The results of our evaluation have been made available to the developer who has made a proposal to his sponsor for evaluating the AIPR system in clinics throughout the country.

2. *Functional Analysis of Externally Powered Hands*

A. Introduction

The hand is often philosophically referred to as one of the three major determinants of human evolution. Together with man's large convoluted brain and binocular vision the hand allowed him to develop as a tool user and tool maker. Elaboration of these fundamental capacities has led him from the use of the stick to reach food and the club for hunting and defense to the building of machines for direct interplanetary communication.

The hand is at once a very powerful and extremely delicate tool. *Grip* or prehension grasp forces of 100 lb. or more are possible, yet the hand is capable of positioning objects meaningfully in a microscopic field. The hand is also a sensory organ through which information is transmitted and received. It is difficult to evaluate the relative importance of the hand as a tool and as

a sense organ; functionally it is both or either. In modern society neither survival nor personal fulfillment is as dependent on the hand as they were in more primitive cultures. Today loss of one hand reduces the motor and sensory capacities of an individual but need not seriously impair his ability to earn a livelihood and to derive personal satisfaction. Prosthetic replacement of a lost hand restores some of the lost function and some, although a good deal less, sensation.

Among the conventional terminal devices in use today, are rubber-band-powered voluntary-opening hands or hooks [a] which provide from 1 to 4 lb. of prehension force between hook and fingers. Spring-loaded terminal devices provide up to 7 or 8 lb. and voluntary-closing devices may provide as much as 40 to 50 lb., a figure approximating the forces applied between the normal finger tips. Although it is possible to approximate the forces of the normal hand, it has only been possible to provide a very small fraction of the vast number of ways in which the normal hand exerts these forces. Only pinch forces between the fingers and the thumb or between the fingers and the palm are available.

Conventional artificial hands and hooks also provide two kinds of sensory feedback. Proprioceptive feedback based on the relationship between the stump and the position of the fingers is one form. In addition tactile feedback is available. Objects pushed, pulled or hooked produce reaction forces which are transmitted through the terminal device to the socket and to the stump where they are converted to tactile sensations—a far less sensitive feedback than the normal physiological pattern.

To operate a terminal device with conventional hands both below-elbow and above-elbow amputees use the force and motion remaining at the shoulder. There, flexion furnishes energy to the terminal device through a transmission system consisting of a reaction point on the opposite shoulder, linked to a cable which is attached to the operating lever of the hand or hook. A well-trained amputee grades the opening or closing range of the terminal device by "hold off" in the case of elastically loaded voluntary-opening devices, or by directly applied forces in voluntary closing devices. Thus, conventional terminal devices replace lost hand function to the extent that they furnish forces approaching those of the normal hand and some small, but significant, degree of sensory feedback. A prevailing view is that the function of the remainder of the artificial arm is to position the terminal device and to act as a power and sensation transmission link between the terminal device and the man.

Conventional artificial hands are often heavy, and lack eye appeal. They provide either two or three finger prehension and some models permit man-

[a] Voluntary-opening terminal devices are those which are opened by muscular exertion against a closing force furnished by rubber bands or springs. Voluntary-closing devices are closed by inputs of muscular force.

ual adjustment of finger position. In general all auxiliary functions other than opening and closing are manually operated. Except for one or two voluntary-closing devices, they lack adequate prehension force.

During the past 10 years there has been a spate of development in this field attempting to overcome some of these deficiencies by introducing external power sources and by improving the control of terminal device function through more direct linkage with the neuromuscular system.

The basic requirements are a power source for the hand and a control system to regulate the application of power to the hand. In general, development in this field has not been a systematic process. There are several developmental hands which are driven by external power sources but which do not incorporate related control systems. Other developments ignore currently accepted principles of upper-extremity prosthetic management, e.g., we do not ordinarily involve the sound limb to control a prosthesis, yet several externally powered hands depend on this type of control.

Given below is an analysis of nine externally powered hands. Of this number, four have been evaluated in our laboratory and three others have been available for physical inspection but not for testing. The other two are described on the basis of demonstrations and evaluation reports of other laboratories. The format of this report is specifically designed to describe the developmental stage of each of the known externally powered hands, as of the publication date of the Bulletin. The merits and deficiencies of each hand are discussed in the perspective of patient needs and current upper-extremity prosthetics management practices. Only the principal design features are considered. We assume that adequate product engineering could eliminate minor design and construction inadequacies. Thus, we have stripped away unnecessary jargon and we have employed the set of criteria generally used in evaluating conventional devices in this country.

In addition to these nine hands, several others have been designed, developed and tested in the past, as for example, the IBM hand. Still other newer hands are described in current literature by Suzuki and by Kato in Japan and by Lymark and by Hiertonn in Sweden. These have not been included in this analysis because their development is not as advanced as those described here or they are no longer in development. Their status therefore is not of imminent interest to clinicians or research administrators who are concerned with practical applications.

B. Analysis

Much is written and said about two special features of externally powered hands—feedback and proportional control. Those terms are used in several different contexts. Unfortunately they mean different things when used in relation to the patient and in referring to the function of a device.

Feedback, for example, is classically defined as "the return of a portion of an output to the input *for controlling the output.*" Feedback therefore

is a characteristic of hands, in which information about the behavior of the fingers is fed back to the motor which then modifies or adjusts finger behavior. The information from the fingers may describe their behavior in terms of position, velocity, or the force they apply. The information may be in the form of electrical current or mechanical forces. The information is used to alter the output (speed or power) of the motor to control the fingers in a precise way. For example, the USAMBRL hand, discussed below in detail, includes a classic feedback feature. By turning on a switch, the patient simply actuates the motor to close the fingers. The fingers close on and grasp an object with a specific force. If the initial force is so inadequate that the object begins to slip when lifted, a device in the thumb senses the motion of the object and sends an electrical signal (information) to the motor causing it to close the fingers further or to increase the prehension force. This *automatic control* of prehension force is based on a feedback system completely contained within the hand and does not require the patient's knowledge or effort.

Often overlooked in considering the control of a prosthetic hand are more conventional kinds of feedback ordinarily required to control conventional mechanical hands. In defining the feedback feature of the USAMBRL hand we described its "output" as the behavior of the fingers and its input as the power furnished by the motor. If we enlarge this context somewhat, we can consider a mechanical hand in which the output is still the behavior of the fingers, but the input is the central nervous system (CNS) of a patient which controls the hand through a musculo-skeletal link and a prosthetic link, the "motor."

This system also depends on feedback. Information about the position or the velocity of the fingers is "fed back" to CNS via visual cues. The information is used by the CNS to alter the output of the muscles which transfer power to the hand to modify or adjust finger behavior. In this system the information may be in the form of visual or auditory cues. Information may also be "fed back" in the form of tactile sensations received by skin receptors. *Conventional control* systems also depend on feedback, but the CNS is one link in the feedback system. Both a conventionally harnessed APRL VC hand and the EMG controlled Soviet-Canadian hand depend heavily on visual and auditory cues for the control of finger position, velocity, and force.

The principal difference therefore between *Automatic* and *Conventional* Control systems lies in the fact that in one case the feedback path runs from the fingers of the hand to the motor which responds automatically while the second case involves the conscious effort of the man. Figure 12 schematically describes the feedback path of both Automatic and Conventional Control systems.

The relative merits of Automatic and Conventional Control systems for

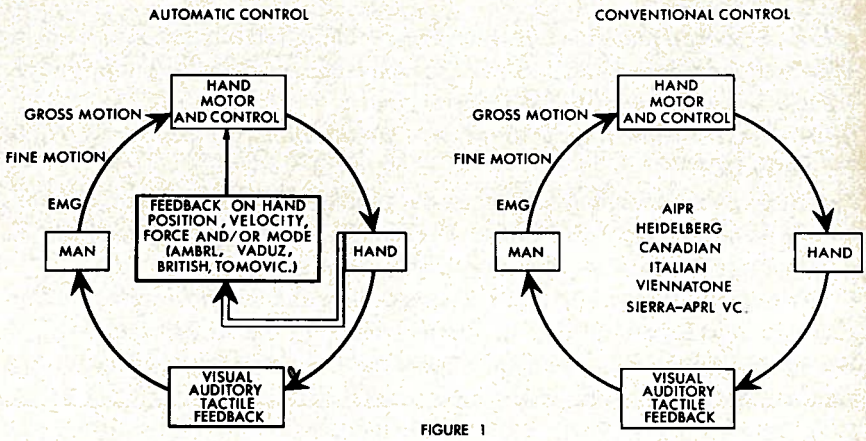

FIGURE 12.—Schematic description of automatic and conventional controlled externally powered hands.

typical arm amputees are unknown at this time. A partially sighted bilateral arm amputee might conceivably locate an object but have difficulty in controlling finger position and force to grasp it properly. An automatic control system might be highly useful. Except for such extreme examples it is difficult, in the absence of more clinical experience, to determine the value of automatic control for other more typical cases.

A great deal of discussion in this field centers about the concept of proportional control. Classically, proportional control refers to the correspondence of duration and magnitude between an input and an output. For example, the position, velocity and force applied by the fingers are proportional to the respective inputs, i.e., the position, velocity and force of the cable in a conventional prosthesis. The time duration and/or the magnitude of an input EMG signal may be proportional to the position, velocity and/or force applied by the fingers of the hand (output).

In these terms which are admittedly unorthodox with respect to control engineering terminology all conventional and externally powered hands feature proportional control. Some hands feature direct proportional control in which 1. the force of an output is related to force of the input or 2. position of the output corresponds to position of the input. This is considered superior to indirect proportional control in which for example the time duration of the input signal determines the prehension force.

With the exceptions cited above, the terminology employed in the following analysis is in general use. Each hand is discussed in eight categories: a. source and availability, b. shape, c. type of prehension, d. weight, e. power source, f. opening range and prehension force, g. type of control and feedback. Another section contains a discussion of the specific features of each device. Each hand is discussed in the context of general requirements for

terminal devices, man-machine interface problems relating to control, operational requirements, clinical value of special features and, where appropriate, comparison with conventional artificial hands.

1. *Montreal Rehabilitation Institute—Modification of the Soviet Hand (Fig. 13)*

a. *Source and Availability.* The hand is produced by the Montreal Rehabilitation Institute, Montreal, Canada. It consists of the original Soviet electric hand and significantly improved Canadian electronic systems.

b. *Shape and Size.* The hand bears a reasonable resemblance to the human hand being comparable in size and shape to the APRL #44 hand. A Soviet rubber glove or a standard American cosmetic glove will fit on the unit.

c. *Types of Prehension.* A form of palmar prehension is provided. The four fingers of the hand move as a unit with the index and middle fingers meeting the moving thumb.

d. *Weight.* The entire hand, that is the powered hand and integrated circuitry, weighs approximately 13½ oz.

e. *Power Source.* The complete system (both motor and amplifier) is powered by a series of rechargeable nickel-cadmium batteries rated at 13.5 volts and 450 milliampere hours capacity which weigh 10 oz. A miniature battery charger is provided with the unit. The Soviet system uses heavier, vented, wet-cell nickel-cadmium batteries.

f. *Prehension Forces and Opening Range.* The hand normally applies a pinch force of 4 lb. This may be increased to 6 lb. by further muscle activity (pulsing the muscle). The maximum opening range is 3¾ in.

g. *Type of Control.* The hand provides voluntary opening and closing (VO–VC), automatically locking in any position. EMG signals from the wrist flexors or extensors are picked up by electrodes on the inner surface

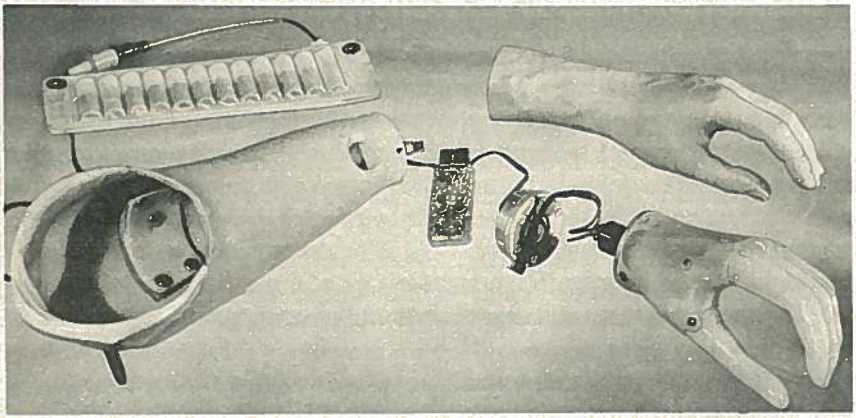

FIGURE 13.—Montreal modification of the Russian myoelectric prosthesis.

of the socket to close or open the hand. Specifically, muscle activation potentials on the skin over the flexor carpi ulnaris (closing) and the extensor digiti minimi (opening) are sensed. To create these signals the subject is taught to flex or extend his phantom little finger. The sensing electrodes are connected to two high-impedance integrated-circuit differential amplifiers with high common mode rejection. The amplified signal is conditioned by voltage doubling, rectifying, and smoothing. The conditioned signal drives a pair of impedance-matching transistors that control the opening or closing relay. A load resistor in the B+ supply for the relays permits only one relay to operate at a time. Once activated, these relays require a lower holding voltage and permit a smaller EMG signal to continue an action once initiated. This characteristic makes it difficult for a signal from the opposing muscle to reverse the action accidentally. The motor is dynamically braked; i.e., when a command signal ceases and the relays are inactivated, the motor stops. This hand depends on conventional visual feedback for proper operation. The relays click softly (but audibly) when an attempt is made to drive the hand past its normal opening or closing range.

h. *Discussion.* The Montreal modifications represent a substantial improvement over the original Russian hand. Some inadvertent activation of the hand occurs due to very high ambient electric fields.

The Canadian myoelectric hand does not differ significantly from the Soviet myoelectric hand in appearance and general operating characteristics. It is, however, a highly significant improvement with respect to smaller packaging, neater design, lighter batteries, and improved safety.

It is the classic version of a system consisting of an electrically driven hand whose motor is turned on and off by EMG signals. The principal advantage of this type of system lies in permitting a below-elbow amputee to open and close the hand using a neural path similar to the original one without displacing the terminal device. Coupled with this is the ability to operate the terminal device in almost any position of shoulder flexion, extension, or abduction without regard to harness tension. The system is rather light at 30½ oz. including glove, hand, electronic components and socket. The appearance and size of the hand are acceptable by current standards. It provides a three jawed chuck grip generally considered superior to two finger grips. Normal maximum prehension forces of 4 lb. are considered inadequate but by pulsing the motor, a marginally adequate 6 lb. maximum can be achieved.

Environmental electromagnetic signals occasionally cause inadvertent operation. The control system is designed for below-elbow amputees whose need for external power is questionable. Above-elbow amputees and certainly shoulder-disarticulation amputees are already encumbered with harnesses to control elbow flexion and extension and to lock and unlock the elbow. It is difficult to identify another useful but unusued control site. Auditory and visual cues provide feedback in a conventional control system.

The negative and positive features of this unit are:

POSITIVE	NEGATIVE
1. Relatively light weight; well designed and packaged.	1. Control by EMG limits use to BE patients.
2. Sensitive to low magnitude signals.	2. Low prehension force.
3. Satisfactory appearance.	3. Only conventional auditory and visual feedback.
	4. Subject to inadvertent activation.

2. (*Italian*) *I.N.A.I.L. Italian Miodyn Hand* (*Fig. 14*)

a. *Source and Availability.* The hand is commercially available from the Instituto Nazionale per l'Assicurazone contro gli Infortuni sul Lavoro, (I.N.A.I.L.) Bologna, Italy.

b. *Shape and Size.* The hand, produced in three sizes, is adequately anthropomorphic except for a prominent thenar eminence. It is designed as a three digit skeleton (thumb and first two fingers) with the thumb support and motor mount enlarging the thenar eminence. The framework is covered with a rubber shell that includes two small floater fingers and gives the hand its shape which is roughly similar to the Sierra APRL adult size hand. Cosmetic gloves to cover the rubber shell are available.

c. *Type of Prehension.* The hand provides palmar prehension. The index and middle fingers and the thumb close in a three jaw chuck. The ring and

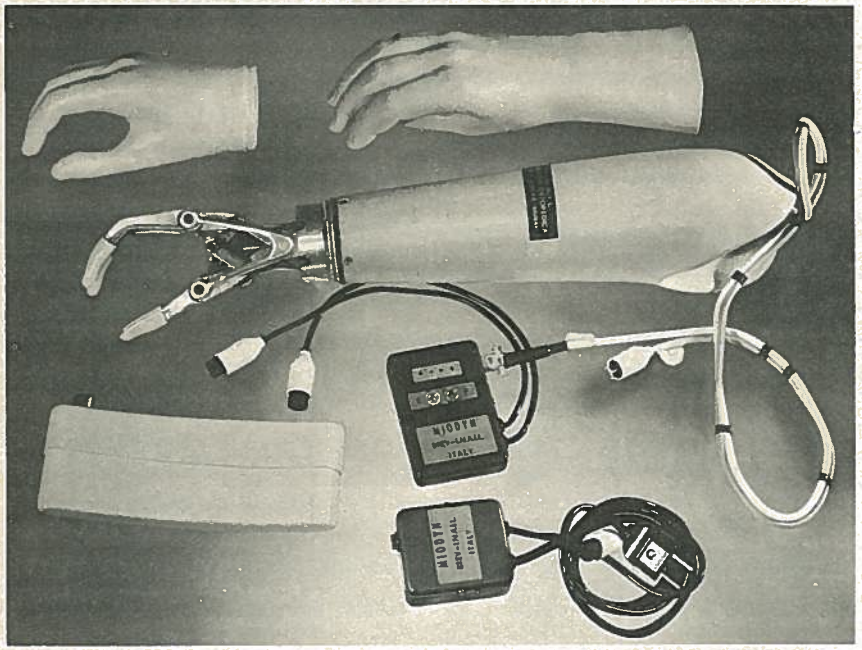

FIGURE 14.—Italian myoelectric system.

little fingers tend to follow the index and middle fingers being part of the same rubber form, but they apply no significant force to an object being grasped.

d. *Weight.* The hand, glove and electrical cabling weigh 18½ oz. The amplifier fits in a patient's pocket and the batteries are belted around the user's waist.

e. *Power Source.* The system (both motor and amplifier) is powered by a series of rechargeable nickel-cadmium batteries at 13.5 volts and 450 milliampere hour capacity. In its neatly designed container it weighs 9½ oz. A miniature battery charger is supplied with the unit.

f. *Prehension Force and Opening Range.* The maximum prehension force available in this hand is 17–20 lb. The hand is self locking due to its lead screw power transmission system, so it will maintain its grip after the input signal stops. The maximum opening is 2¾ in.

g. *Type of Control.* The hand requires an EMG signal from the wrist flexor in the patient's forearm to close and a signal from the wrist extensor to open. Although EMG signals may be sensed from both muscles, the stronger of the two signals determines which function will be performed.

Specifically, the muscle activation potentials in both flexor carpi ulnaris and extensor carpi radialis longus are sensed by electrodes in the socket directly over each muscle. Two amplifying systems (capacitor-coupled, discrete-component, differential amplifiers) amplify and rectify the signal to close or open the relays which control the small electric motor in the hand. Adept operation depends on conventional control including auditory and visual feedback.

h. *Discussion.* The Italian myoelectric hand is another much improved version of the original Soviet design similar in appearance and function to the Canadian model. It is cleanly designed and well packaged. Although heavier than the Canadian model, it is relatively light in weight and it operates reliably. It is quite sensitive and responds to the EMG signals from light to moderate muscle contractions. A clear advantage over the Canadian and original Soviet models is the rather high prehension force available—17–20 lb.

Its automatic locking feature, achieved by means of a lead screw, is available at the cost of low electromechanical efficiency (Fig. 15). At 2¾ in., its opening range would be considered marginally adequate. It is subject to inadvertent operation in the presence of a strong electrical field. A new model however, is designed to eliminate this problem by using amplifiers mounted on the electrodes. Auditory and visual feedback are used to control operation. The significant features of this device are classified below:

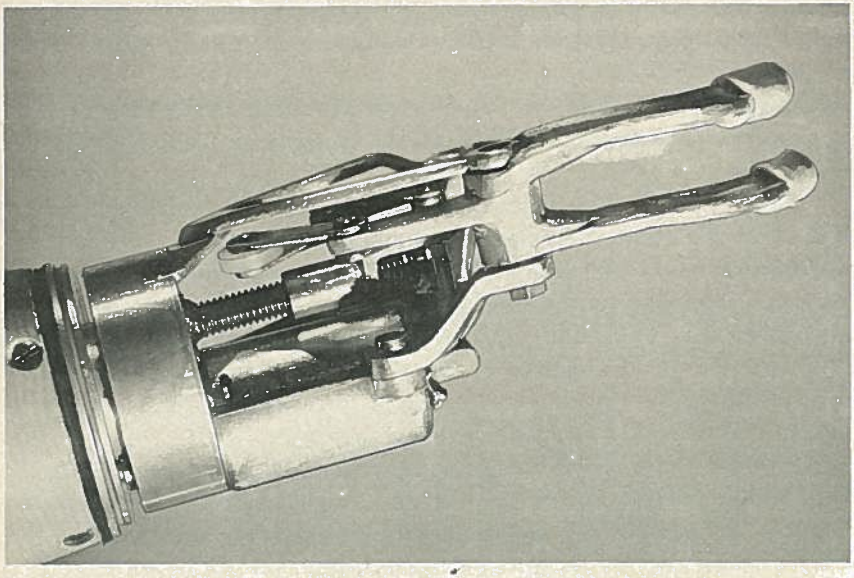

FIGURE 15.—Internal mechanism of the Italian myoelectric hand.

POSITIVE	NEGATIVE
1. Well designed, neatly packaged, adequate cosmesis.	1. Control system limits use to BE patients.
2. Adequate prehension force.	2. Patient must be fitted at source.
3. Sensitive to signals from light to moderate muscle contractions.	3. Conventional auditory and visual feedback.
	4. Subject to inadvertent operation.
	5. Small opening range.

3. *Heidelberg Pneumatic Hand* (*Fig. 16*)

a. *Source and Availability.* The Heidelberg Pneumatic Hand is available through the Otto Bock Orthopedic Industry, Inc., Minneapolis, Minnesota.

b. *Shape and Size.* The device is a sculptured wood hand that does not closely approximate the shape of the normal hand. It is not designed to receive a cosmetic glove.

c. *Type of Prehension.* The hand closes with the active thumb closing against one stationary finger (fingertip prehension). The tips of the thumb, index, and middle fingers are covered with rubber pads to aid prehension.

d. *Opening Range and Prehension Force.* Maximum opening is 2 in. With a newly charged CO_2 tank the maximum prehension force is 22 lb.

e. *Weight.* The exact weight is unknown but considering its construction it should not be excessive.

f. *Power Source.* A lightweight aluminum CO_2 pressure canister and reduction valve are used in Europe. A device such as this does not meet safety requirements in the United States, where steel canisters are required

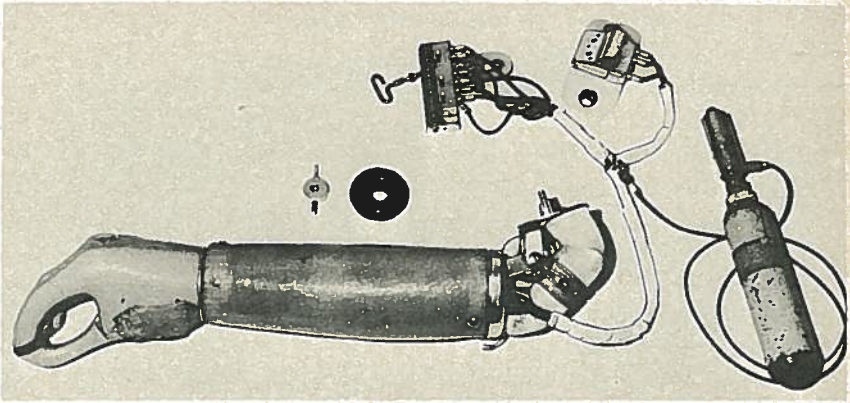

FIGURE 16.—Heidelberg pneumatic system.

to meet the standards. The canister is recharged from a non-portable larger CO_2 tank.

By actuating a control valve, CO_2 is directed into a bellows in the hand providing power for closing. The valve is actuated by a gross body motion, i.e., any overt movement of a body segment. Actuating a second control valve exhausts the bellows into the atmosphere, and a return spring opens the hand.

g. *Type of Control.* A gross body motion of low force and displacement opens the on-off closing control valve and closes the hand. The length of time, it is achieved by on-off action. Opening is achieved by a second on-off contact, the prehension force applied. Although control is thus related to time, it is achieved by on-off action. Opening is achieved by a second on-off valve also actuated by another gross body motion. No automatic control is provided. The user depends on conventional auditory and visual feedback for control.

h. *Discussion.* The pressure canister does not meet United States safety standards. Cosmetically the hand is unacceptable. Its maximum opening range of 1.95 in. is inadequate. It provides only finger tip prehension, a motion which is normally used less than one fifth of the time. No positive lock is provided.

The significant features are listed below:

POSITIVE

1. Wrist rotation available.
2. Part of a complete system for use by BE, AE, and SD patients.

NEGATIVE

1. Conventional auditory and visual feedback.
2. Poor type of grasp.
3. Insufficient opening range.
4. Poor cosmesis.
5. Aluminum tanks unacceptable by U.S. Safety Standards.
6. Filling and exhaust noise.

4. *The Viennatone Hand, "Myomot MM 2" (Fig. 17)*

a. *Source and Availability.* The Viennatone Hand is mass produced and commercially available. It is available in the United States through the Fidelity Electronic Ltd., 5245 Diversey Avenue, Chicago, Ill. 60639, or from the manufacturer, Viennatone Hörgeräte Vertriebsgesellschaft m.b.H., 1164 Wien XVI, Fröbelgasse 28–30.

b. *Shape and Size.* The hand is reasonably anthropomorphic. It is finished with a cosmetic glove. Due to the skeletal method of construction, a three finger framework covered by a PVC shell and a cosmetic glove, it has a prominent thenar eminence. It is similar in size to the Sierra APRL adult size hand.

c. *Type of Prehension.* The hand provides palmar prehension with the index and middle fingers approximating the thumb. The third and fourth fingers are hollow rubber floaters.

d. *Weight.* The hand alone (including cosmetic glove) weighs approximately 17 oz. The amplifier/battery package adds another 17½ oz.

e. *Power Source.* Ten rechargeable Nicad cells, supplying 450 milliampere hours at 12 volts, power the amplifier and electric motor. Two differential EMG amplifiers pick up muscle signals which are fed to switching transistors to control the flow of battery power to the motor. The motor drives a four stage spur-wheel gear reducing system with a very high operating efficiency connected to two segmented gears (Fig. 18). One gear is placed coaxially with the center of rotation of the thumb. The other is aligned with the common axis of rotation of the first and second fingers. A unidirectional lock is incorporated in the third stage of the spur gear system

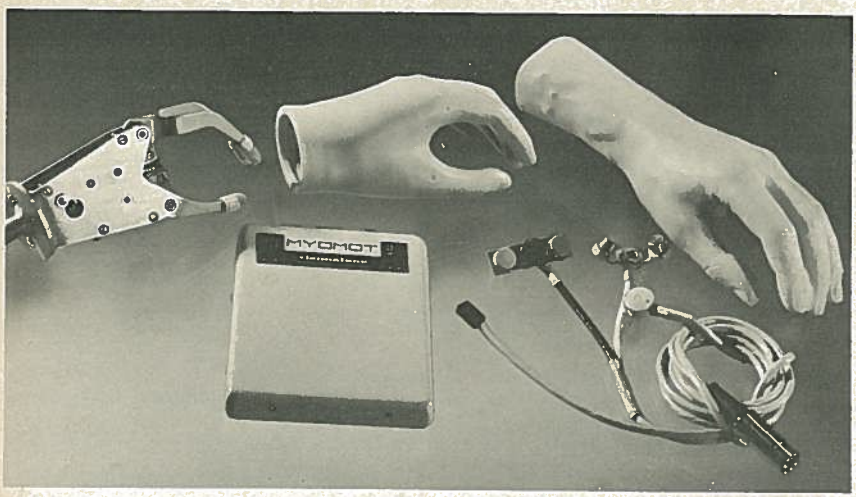

FIGURE 17.—Viennatone system, Myomot MM 2.

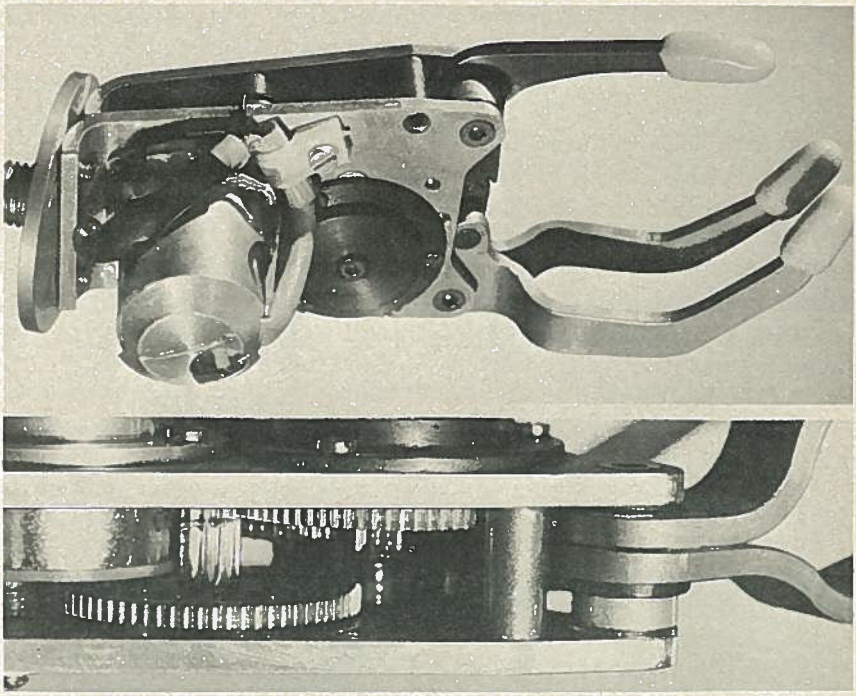

FIGURE 18.—Internal mechanism of the Viennatone hand.

assembly permitting the motor to drive the hand in either direction yet, preventing inadvertent opening by external forces. The high efficiency of the entire system results in very little power drain. A battery charger is supplied with the unit.

f. *Prehension Force and Opening Range.* Normal maximum prehension force is 18 lb. This may be increased by pulsing the motor. The opening range is 2¼ in. The opening signal must be maintained to keep the hand fully opened or the bias of the rubber hand shell would partially close the fingers.

g. *Type of Control.* As in the Canadian and Italian hands, EMG signals from the wrist flexors close the hand and the EMG activity of the extensors open the hand. This unit uses switching transistors instead of the relays used in the others. Prehension force is controlled by starting and stopping the system.

Visual feedback is the primary source of control information. Some motor noise provides auditory feedback but in the stalled condition no noise is heard.

h. *Discussion.* The "Myomot MM 2" hand is an extremely well-engineered device. According to the manufacturer it has been cycled in excess of

a million times without significant wear or a failure. It is built as a modular system for easy replacement of components in case of damage. It is moderate in weight and offers very high prehension forces. Light to moderate muscle contractions operate it easily. It is electromechanically efficient. Strong electromagnetic fields will cause inadvertant actuation as in any EMG system with appreciable space between the electrodes and amplifiers. The opening range at 2½ in. is smaller than desirable. A testing device, the "Myotest MY 21" (Fig. 19), is available for locating the best control site on the muscle for the snap-in electrodes. An installation booklet is provided. The use of standard American threads on the wrist fitting would help domestic installation. The significant features of the "Myomot MM2" are listed below.

POSITIVE

1. Well designed, neatly packaged and adequately cosmetic.
2. Adequate prehension force.
3. Sensitive to light to moderate muscle contraction.

NEGATIVE

1. Control system limits use to below-elbow patients
2. Small opening range.
3. Conventional feedback.
4. Inadvertent operation.

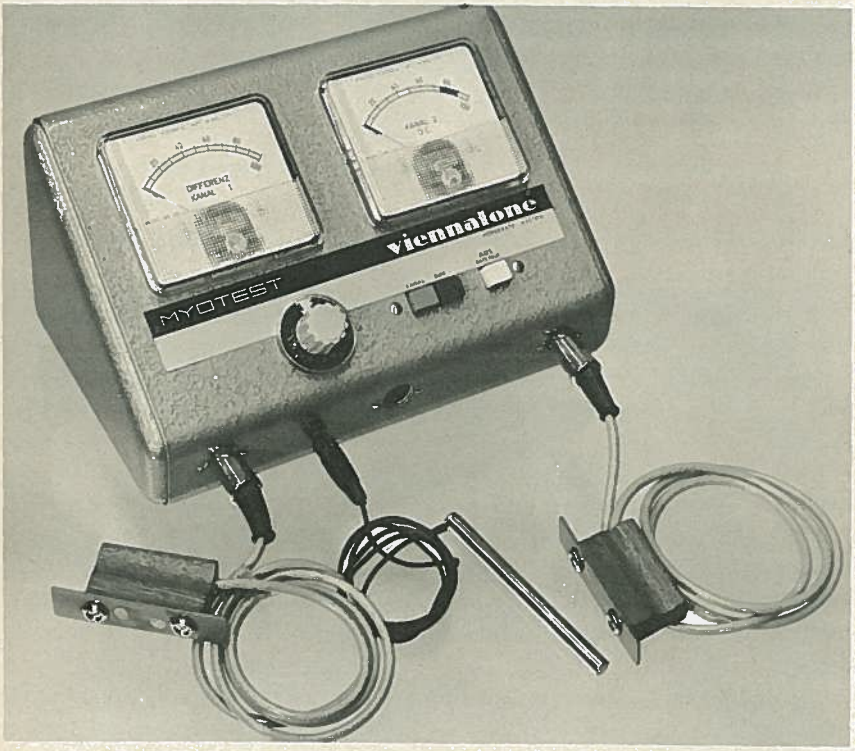

FIGURE 19.—Myotest MY21 testing device used to locate the best control site.

5. *Beograd, Belgrade, Tomovic Hand*

a. *Source and Availability.* A prototype of this hand was evaluated by UCLA and USAMBRL (3) (4). It was produced by the Mihailo Pupin Institute, Belgrade, Yugoslavia. It is our understanding that 20 units of the revised model have been manufactured for further testing but no new units are available to us at this time.

b. *Shape and Size.* The hand is not adequately anthropomorphic to meet current American standards for hand cosmesis. Although not much larger than American hands, it lacks natural looking contours and appears bulky. It is covered by a non-cosmetic rubber glove.

c. *Type of Prehension.* Two types of prehension are available: palmar and grasp. In *palmar* prehension the thumb and either the index or the index and middle fingers approximate each other. In *grasp* prehension the four digits approximate the palm without the intervention of the thumb. This hand offers both modes; actually, the four fingers always flex toward the palm but in palmar prehension, the thumb moves to intercept the first two fingers.

d. *Weight.* The device weighs approximately 35 oz. not including the battery. By comparison, the APRL VC Hand, one of the heavier conventional devices, weighs 13 oz.

e. *Power Source.* The unit is powered by a 12 volt battery but no specific information is available on either type or size of the power source for the new units.

f. *Opening Range and Prehension Forces.* No specific information is available as to the precise range of opening nor the apprehension forces available. However, an evaluation report (3) indicates that prehension force in the available unit was approximately 4.0 lb.

g. *Type of Control.* This hand provides voluntary control of both opening and closing (VO–VC). The details of its construction with respect to mechanical linkages, electrical system, and electronic components and circuitry are described in (3) and (4). The operational cycle requires activation (by pressure) of two switches and one or more of eight sensors (pressure) on the volar surfaces of the four fingers. The eight sensors are arranged in two rows of four each with one on each distal finger pad and one each on the pads of the second phalanx.

Pressure on a distal sensor preconditions the hand to close in the palmar prehension mode; pressure on a proximal pad preconditions the hand to close in the grasp prehension mode. However, before the hand will close in either mode, a pressure switch, (closing and opening) mounted in the harness above the bulge of the trapezius must be actuated. A second switch mounted in a "biceps" cuff must also be actuated to reverse the direction of the motor before a second actuation by the trapezius switch (opening-closing) will open the hand.

Both the motor reversing and the opening-closing switches are mounted on the sound side.

The designed operational sequence is:

1. Activate switch (closing or opening) on sound side trapezius. Hand begins to close (palmar or grasp type of prehension) determined by pressure on distal or proximal finger pad.
2. Activate motor reversal switch on sound side biceps.
3. Activate switch (opening or closing) on sound side trapezius. Hand opens.

The unit is controlled by conventional feedback sources, visual and auditory cues plus the unique "adaptive" features provided by the finger sensors.

h. *Discussion.* The Beograd design is based on a highly sophisticated concept to simulate one of the extremely important functions of the normal hand—selection of either of two prehension modes in a semiautomatic manner. Voluntary selection is possible if desired by positioning the terminal device to touch the appropriate pad or the object to be grasped.

As a prototype this device should not ordinarily be judged on the basis of shape, bulk, and possibly weight. However, the key feature of this design requires the thumb to remain in an abducted position while the hand is open. Improving the resulting unattractive overall shape would require extensive redesign. The thumb position may preclude insertion of the hand into pockets, small spaces, and it may catch on objects in the immediate environment.

Prehension forces of 4.0 lb. are considered inadequate particularly for an externally powered device. Pinch forces of almost equal magnitude are available with 4 rubber bands on conventional hooks and conventional VC hands allow appreciably higher forces.

The basic operational cycle requires three independent sequential operations to accomplish the simple task of grasp and release. The closure or opening switch must be actuated, and the motor must be reversed twice. This is achieved in a conventional terminal device by two control motions: shoulder flexion to close or open and shoulder flexion again to release.

A device of this type would seem highly impractical for the above-elbow or shoulder-disarticulation amputee. In addition to the three control motions which the hand requires even for the simplest task, at least four additional control motions would be required to operate a conventional elbow, far exceeding the tolerance of most patients. Furthermore, typical operation of the conventional elbow requires abduction and extension of the humerus with resulting tension on an axilla loop. This may cause closure of the trapezius switch mounted on shoulder and/or inadvertent change in the direction in which the motor will run when next actuated.

The control concept is also faulty from the standpoint of involving the sound limb in the control of the prosthesis. Proper prosthetic management

of the unilateral amputee avoids harnessing or other involvement of the sound side. Conventional harnesses are designed to avoid active involvement of the sound side. Training procedures are aimed at reducing overt action of the sound shoulder.

Both positive and negative features are listed below:

POSITIVE	NEGATIVE
1. Choice of two useful prehension modes, normally semiautomatically.	1. At present limited to BE patients where the needs are minimal. Excessive weight may contraindicate use by AE patients.
	2. Excessive number of controls.
	3. Protruding uncosmetic thumb.
	4. Low prehension force.
	5. Provides only conventional auditory and visual feedback.

6. The Army Medical Biomechanical Research Laboratory Mechanical Hand with Automatic Proportional Control of Prehension (Fig. 20)

a. *Source and Availability.* The hand is not commercially available at present. Six redesigned hands are being fabricated under the direction of the U.S. Army Medical Biomechanical Research Laboratory, Walter Reed Army Medical Center, Washington, D.C.

b. *Shape and Size.* The hand, covered by a standard cosmetic glove, is adequately anthropomorphic by current standards. It is built in the APRL #44 hand shell with a special detector in the thumb. A lead screw drive mechanism is mounted in the hand, and the motor and speed-reducing gear are mounted in the wrist.

c. *Type of Prehension.* As in the other APRL designs, the hand features palmar prehension with the index and middle fingers approximating the thumb. The ring and little fingers are floaters.

d. *Prehension Force and Opening Range.* The hand applies pinch forces in the range of 2 to 12 lb. Pinch force magnitude is automatically controlled to stop slippage of an object between the fingers. The maximum opening range is either 1¼ in. or 2½ in. depending on the position of a manually controlled thumb.

e. *Weight.* The weight of the hand alone (exclusive of socket) is 15½ oz., and the battery weighs 9¾ oz.

f. *Power Source.* Electrical power is supplied by rechargeable Nicad batteries (Gould 12.0 volt, 450 milliampere hours). Power is supplied to the standard APRL finger linkage and worm drive by a D.C. motor (Globe Industries #C43A106) with 14.8/1 gear reducer (Globe Industries C–1A–911). The prehension logic system consists of a phonograph cartridge (Astatic #18) mounted in the thumb and an electronic control package. The phonographic cartridge detects slip between the thumb and the fingers commanding the motor to close the hand further when slippage occurs.

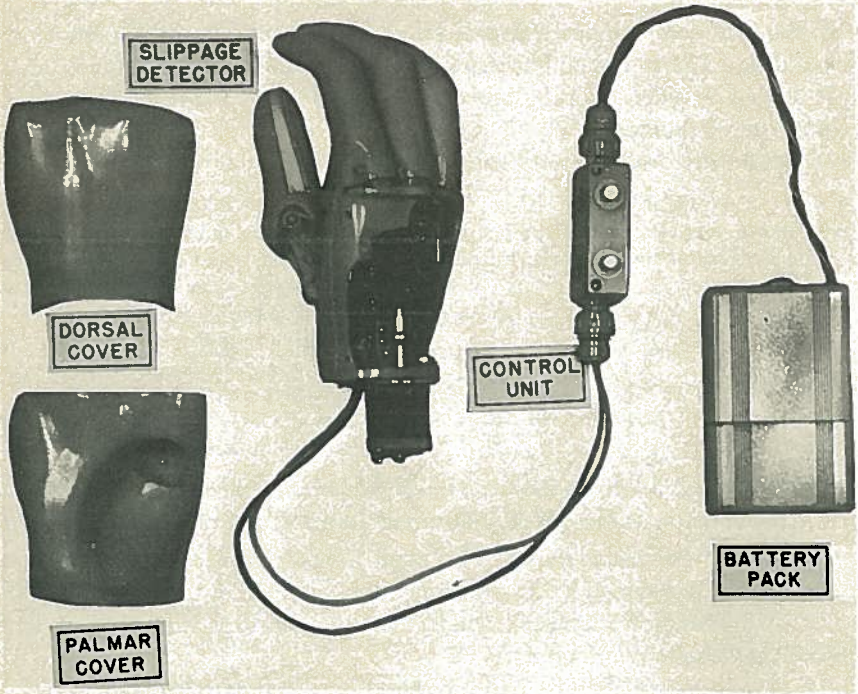

FIGURE 20.—USAMBRL System showing mechanical hand with automatic proportional control of prehension.

The electronic package consists of an amplifier (Fairchild μA702), integrator, Schmitt trigger, and a switching transistor.

g. *Type of Control.* By means of a continuous control motion (pull on control cable or other) the patient actuates the motor to close the hand to a preset force level (2 lb.). If the object then slips between the thumb and fingers while being lifted, the motion is sensed by the phono cartridge which commands the system to start the hand-closing motor, the hand automatically increasing its prehension force (to a maximum of 12 lb.) until the object no longer slips.

To release an object the wearer gives a second command (control motion) and the hand opens. Hand motion is stopped automatically at the extreme positions by limit switches. The finger motion may be stopped at any intermediate point in the opening or closing range by voluntary interruption of the control signal by the patient.

If finger motion is stopped short of full opening and then started again, the fingers will close and vice versa. To continue motion in the original direction after stopping, the system must be recycled electrically by two control motions to reverse motor polarity twice.

In addition to the conventional controls based on visual and auditory feedback, the hand also features automatic control based on electronic feedback from the phono-cartridge in the thumb to the motor. Slippage of an object in the hand signals the motor to close the hand tighter until slippage stops without decision or effort on the part of the patient. Although visual and auditory cues are available dependence upon them is reduced by the *automatic* hand function.

h. *Discussion.* The USAMBRL hand is unique in effectively controlling undergrasp. It regulates prehension force according to demand, simulating to a degree the sensorium of the human hand. When a normal person picks up an object, he feels in his fingers the prehension force he is applying; if the object slips (undergrasp) he automatically increases prehension force. Once a closing command is given to the USAMBRL hand, it automatically closes to a preset prehension level (2 lb.). If this is insufficient force (undergrasp), the object slips as it would in a normal hand and the force is automatically increased until slippage stops (maximum 12 lb.). This reduction in decision-making responsibility may make the unit particularly valuable for above-elbow and shoulder-disarticulation amputees.

Despite these excellent features, the hand retains one control characteristic which was considered a drawback in the APRL mechanical voluntary closing hand. To stop motion of the fingers and then continue in the same direction requires a "double shuffle," i.e., two control motions. The first reverses motor polarity causing the hand to move in the opposite direction; the second switches the mode back to the original direction. Use of EMG for primary control could eliminate this drawback.

The hand is presently designed for control by a pull cable attached to a conventional harness. It may also be controlled by EMG signals, but at the present time installing any of the available EMG electrode-amplifier systems would add another electronic section and perhaps require the service of personnel other than prosthetists for installation in a prosthesis.

A reduction in weight in the final design would improve the general utility of the hand. While its function may be desirable for above-elbow and shoulder-disarticulation amputees, it is approximately 25 percent heavier than the APRL hand which is generally considered too heavy for higher levels of amputation.

The location of the motor in the wrist may make the device impractical for below-elbow amputees with long stumps. A list of the positive and negative features follows:

POSITIVE	NEGATIVE
1. Automatic control of prehension force.	1. The unit is heavy.
2. Reduction of decision-making requirements.	2. Small opening range.
	3. Double shuffle control.

7. American Institute for Prosthetic Research

C–80100 Pneumatic Hand (Fig. 21)

a. *Source and Availability.* The AIPR hand is not presently available. The models produced to date were for experimental purposes only.

b. *Shape and Size.* The hand consists of the Sierra APRL adult hand shell with the internal mechanism modified to receive a pneumatic piston and cylinder assembly.

c. *Type of Grasp.* The hand provides palmar prehension with the index and middle fingers moving against the stationary thumb.

d. *Prehension Force and Opening Range.* Prehension force is controllable in a range from ½ to 13 lb. Two opening ranges are available by means of the manually controlled two-position thumb; 1⅝ in. and 2⅝ in.

e. *Weight.* The unit weighs approximately 10 oz. without the cosmetic glove which adds 2 oz. This is approximately 2 oz. lighter than the conventional APRL hand and glove. The gas storage tank weighs approximately 25 oz.

f. *Power Source and Pneumatic-Mechanical System.* Power is supplied from a tank containing compressed CO_2. A two-function valve directs the flow of CO_2 to either side of a double-acting piston to open or close the hand. The piston is connected to the modified linkage of the Sierra #44 hand to actuate the index and middle fingers.

g. *Type of Control.* The control valve may be located in the standard figure-eight shoulder harness. Hand prehension force is controlled by harness tension. A maximum of 13 lb. of palmar prehension force can be applied. Control of prehension force is achieved by switching the valve on and off.

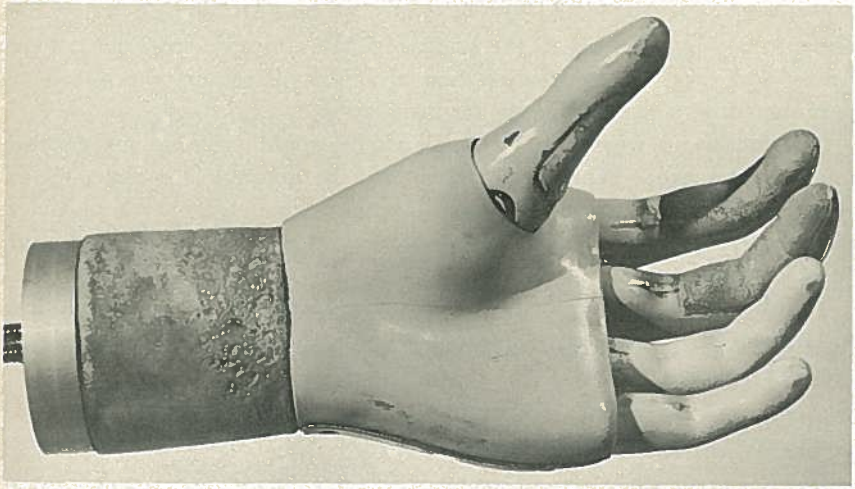

FIGURE 21.—The AIPR C–80100 pneumatic hand.

However, by fine control motions which do not open the valve completely, a roughly proportional control over the rate of change of prehension force is possible. The resistance of the valve as sensed by the patient is roughly proportional to the applied force. However, visual and auditory cues provide primary control information.

h. *Discussion.* The hand is very light, it provides high prehension forces, and (covered by a glove) it is cosmetically adequate. One key advantage of the hand is that it is part of a total system which includes a pneumatically actuated wrist rotation unit and an elbow with automatic lock. The positive and negative features are listed below.

POSITIVE	NEGATIVE
1. The unit is light.	1. Provides primarily visual and auditory feedback.
2. Intended for more high level patient amputees.	

8. *British Myoelectric Hand*

a. *Source and Availability.* The source of information for this analysis is the British Ministry of Health, Biomechanical Research and Development Unit, Roehampton S.W. 15, London, England. The hand is still in a developmental stage and is not available for purchase. A second prototype series is presently being designed.

b. *Shape and Size.* The original prototype was built in the shell of a Sierra APRL adult size hand. The redesigned models will use the Hosmer adult hand shell. A cosmetic glove fits over either hand.

c. *Type of Prehension.* The hand closes in palmar prehension with the index and middle finger moving to the stationary thumb. The ring and little finger are "floaters" moved by the cosmetic glove.

d. *Weight.* The electronic components and the power pack weigh approximately 20 oz. and 24 oz. respectively. Final hand weight will depend on the final design.

e. *Power Source.* The prototype power supply consists of a series of silver-zinc rechargeable cells.

f. *Prehension Force and Opening Range.* The hand provides a maximum prehension force of 6½ lb. and an opening range of 3¼ in.

g. *Type of Control.* Both the rate at which the hand opens (closes) and the force it applies to an object are controlled by EMG signals whose character depends on tension in the flexors and extensors of the wrist. Certain parameters of the electrical output of the muscles are roughly proportional to the tension they develop. The EMG output of *both* sets of muscles is constantly monitored by skin electrodes placed on the dorsal and volar surfaces of the forearm.

The EMG signals from the flexors and the extensors of the wrist are picked up by electrodes, amplified, and independently rectified and smoothed. These signals are fed through a "differencing" circuit whose

output determines the direction and magnitude of motion—opening if the output of one muscle group (extensors) is higher than the other, and closing if the reverse occurs. The dominant signal is fed through two components on its way to the hand motor (Fig. 22). In passing through the first component it is conditioned by a feedback signal from the hand which indicates force. The output from the first component then passes through another component where it is conditioned by another signal from the hand which indicates velocity. As the hand is closing on an object, the velocity feedback signal is compared with the primary EMG input so that the resultant signal controls the velocity of closure. In closing on an object, no significant force is exerted between the fingers and the thumb; thus the force feedback signal has little effect on function. However, when the hand grasps an object, velocity vanishes but force begins to increase rapidly. At this stage the force feedback signal conditions the primary EMG input to control prehension force between the fingers and the thumb. The device therefore features a form of automatic control based on feedback from the fingers to the motor control amplifiers which make the velocity of closure and the force applied to an object directly proportional to the EMG output of the muscles. Since, the feedback serves to modify the primary EMG input in controlling hand function, the *primary control* is the tension which the patient voluntarily produces in the muscles.

h. *Discussion.* The British myoelectric hand is a unique device which uses EMG signals as the control source. This hand features proportional control of closure speed and of prehension force. The stronger the EMG input signal, the faster or more forcefully the hand closes. Proportional control of this type appears desirable although the actual value has not been demonstrated. The hand speed and/or force application will automatically correspond to the primary EMG control signal. But in the performance of a task, control over speed of closure depends upon visual cues which the amputee senses and interprets as a need for greater or lesser closure speed. Then he adjusts the state of muscle contraction to meet the

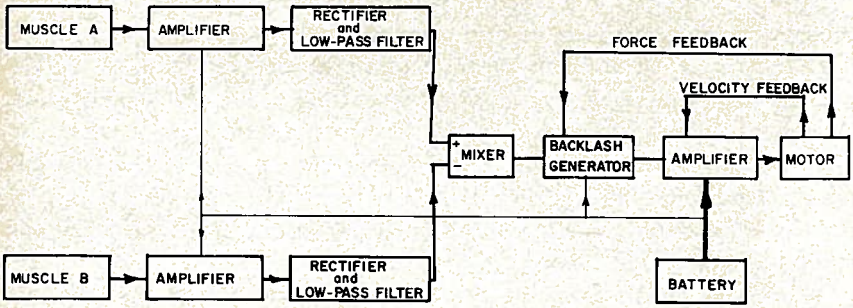

FIGURE 22.—Schematic diagram of British myoelectric control system.

need. Once the object is grasped, he depends again on visual cues to indicate the need for greater or less force, and adjusts muscle tension accordingly. Thus, the system depends completely on visual cues just as conventional mechanical systems. Although it features automatic proportional control at one stage of its operational cycle, it does not, as does, for example, the USAMBRL hand, provide automatic proportional control over the complete performance of a grasping task since the control over muscle tension comes from the CNS. The real question therefore in evaluating this and other EMG controlled devices is how speed and force are best controlled: by muscle contraction at the shoulder translated into motion of the humerus and motion of the hand, or by muscle contraction which is translated into electrical signals to control the hand. Nevertheless, the proportionality to input signals of hand speed and force will make it easier for patients to control this hand than to control other EMG controlled hands without proportional control.

The present development plans call for the design of an easily interchangeable hook. This is desirable since hands by reason of their size and bulkiness tend to be less useful for delicate work principally because they obstruct the view of the user. For such work hooks are usually preferred. Any system that provides both a powered hook and an interchangeable powered hand is desirable.

Although both velocity and force feedback as well as visual and auditory cues are used to control the hand, no tactile feedback is available to the wearer. The amputee still depends on being able to see what control he wishes to exert over the hand. At present the device is designed for below-elbow patients in whom appropriate muscle signals are readily available but whose need for such a device is minimal.

The principal positive and negative features are listed below:

POSITIVE	NEGATIVE
1. Velocity and force are proportional to EMG control signal.	1. Requires conventional feedback (visual and auditory).
2. An interchangeable hook will be available.	2. Due to its weight it may be limited to BE patients where the need is minimal. Use by AE or SD amputees requires an additional control site which may also limit its use.

9. *French Electric Hand* (*Vaduz*) (*Fig. 23 and 24*)

a. *Source and Availability.* The hand, developed by Dr. Edmund Wilms at Vaduz, Liechtenstein, may be purchased from F. Guillot Institute Chirurgie Orthopedie of Paris. As a result of evaluation in this country several modifications were recommended but it is not known which model is now available.

b. *Shape and Size.* The hand is reasonably anthropomorphic when covered with a cosmetic glove. It appears somewhat bulkier and larger than ideal,

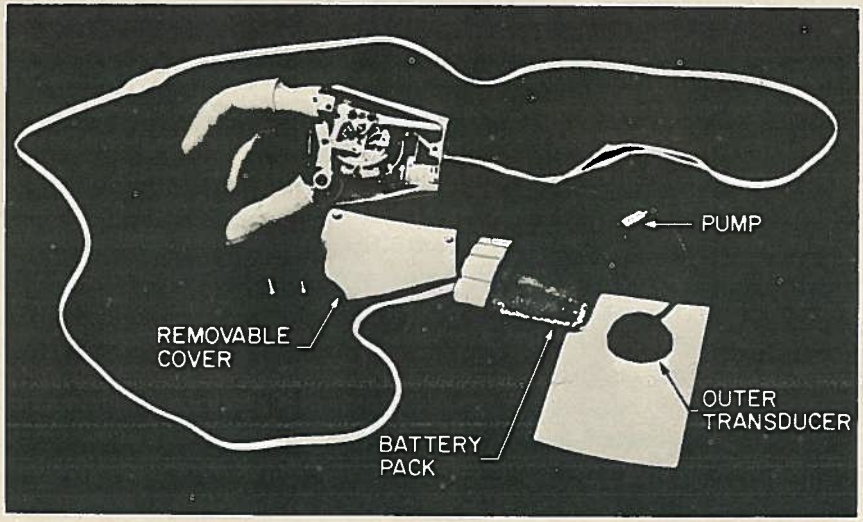

FIGURE 23.—The French electric system (Vaduz).

but nevertheless it is comparable to several currently used conventional American hands.

c. *Type of Prehension.* This hand provides palmar prehension only. The index and middle fingers, moving on a common axis meet the thumb in a three-jaw-chuck grip. The ring and small fingers move on the same axis as the index and middle fingers and approach the palm, but their effect is considered more cosmetic than functional since they do not move after the first two fingers are in contact with the thumb.

d. *Opening Range and Prehension Force.* The hand opens to approximately 2.6 in. between the thumb and index and middle fingers. Maximum prehension force can be graded through a spring adjustment between 2.5 lb. and 7.0 lb.

e. *Weight.* The hand is relatively light at 13 oz. and 15 oz. with cosmetic glove.

f. *Power Source.* Although the actuating transducers are pneumatic, the hand operates on power supplied by a 5-volt, 500 milliampere hours nickel-cadmium battery which is not contained in the hand.

g. *Type of Control.* This unit is controlled by means of a pneumatic transducer which transfers force and fine motion from the muscles (bulge) of the forearm to switches in the hand (Fig. 24). When the electric motor is switched on (by the pneumatic control system), power is fed to the fingers through a gear train which initially provides a 6:1 reduction, the high speed-low force prehension mode. If the fingers meet resistance during closure, or if the motor reaches maximum speed (no resistance during closure) the gears shift automatically to provide a gear reduction ratio of

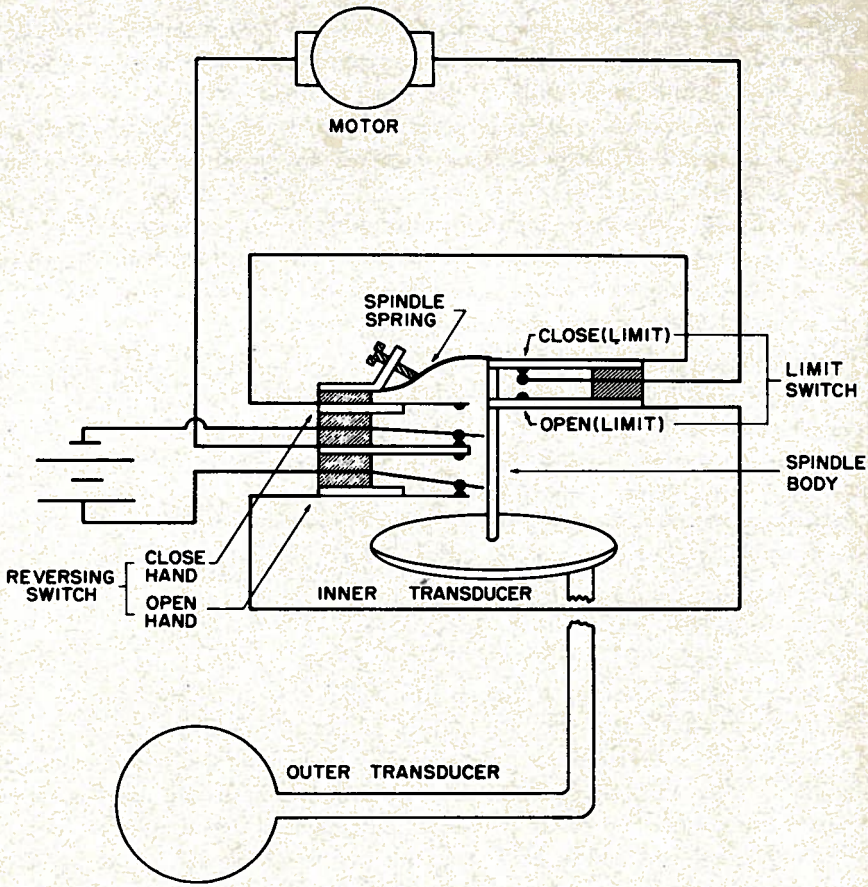

FIGURE 24.—Schematic diagram of French electric hand.

125:1, the low speed-high force prehension mode. Force from the forearm must be continually applied throughout the closing portion of the prehension cycle. *Lightly applied* initial force must be increased to effect full closure.

At some point during closure, a position linkage moving proximally as the fingers close, breaks the closing circuit and shuts off the motor. To make the fingers continue closing, additional force must be applied by the amputee. *High forces* applied to the transducer are not affected by the *position* linkage and result in continuous closure.

To maintain grasp of an object between the fingers, the patient must maintain an appropriate pressure on the pneumatic transducer.

The typical operational cycle is:

a. With hand closed the amputee reduces force applied to the pneumatic transducer to open hand.

b. When increasing increments of force are applied to the pneumatic transducers, the hand closes in small motion increments or applies small force increments.

c. Decreasing increments of force applied to the transducer open the hand in small increments (or reduce force) under control of the muscle.

d. Hand closes at maximum speed under strong forces applied by the patient. At maximum speed of closure the gears shift automatically to the low speed-high force prehension mode. Stronger muscle contraction produces maximum prehension force.

In addition to the conventional visual and auditory feedbacks to the user, this hand also provides some sensory feedback in the form of tactile sensations of pressure between the pneumatic transducers and the patient's muscle. This tactile feedback may be roughly proportional to both the prehension force between the fingers and thumb of the hand, and to finger position.

h. *Discussion.* The Vaduz hand has the advantage of providing some direct sensory feedback through its pneumatic transducer. The patient can sense, from the pressure of the transducer against the skin of his arm, the extent to which the hand is opened or closed. He can also sense the prehension force being applied. Although this seems to be a highly desirable and much-sought-after feature, its practical utility in actual operation is still questionable. If the amputee grasps an object which he sees, *visual* feedback is likely to be the principal source of information he would use to control finger positioning and force. It is absolutely essential that the target (object to be grasped) be seen in order to effect the "rendezvous," i.e., to position the terminal device for grasp of the object. It is not likely that an amputee will thereafter grasp and transport an object without maintaining visual contact and without depending principally on the visual feedback for information (about finger position and prehension force). Thus, in a real sense, proportional tactile feedback is nothing more than an elaboration of (perhaps an improvement on) the tactile feedback (also proportional) that is available in spring-loaded hands activated through a conventional harness and cable system. Moreover, the proportional feedback system of the Vaduz hand may suffer from a drawback not found in conventional systems with automatic locks. Force must be continuously applied in the Vaduz system to maintain prehension force. Conventional voluntary opening hands automatically maintain prehension force although it is limited by the nature of the springs. However, conventional voluntary closing hands feature automatic locks, which, once locked, do not require further force applications to maintain their grip.

An advantage of the Vaduz system is that the control motions (muscle bulge) are very small (fine body motion) compared with the humeral

flexion (gross body motion) required to operate a conventional system. The advantage lies in the fact that the terminal device need not be displaced or, less compensatory body motions may be required.

At 2⅝ in. the opening range of this hand would be considered marginally adequate by American standards. Many conventional hands offer opening ranges of 3 to 4 in.

The limited experience with this hand in this country (5) suggests a history of problems of low durability and need for repeated adjustment. Listed below are the significant features of the Vaduz hand classified according to positive and negative values.

POSITIVE	NEGATIVE
1. Proportional tactile feedback.	1. Continuous application of force necessary to maintain grip.
	2. Durability questionable.
	3. Small opening range.

C. Summary

The more important characteristics of the nine externally powered hands are presented below in tabular form. Four of these devices, the Soviet-Canadian, Italian, Viennatone, and AIPR hands were tested in our laboratory. Although not tested, the USAMBRL hand has been closely examined during several demonstrations. The data on the Tomovic, Heidelberg, British Myoelectric, and Vaduz hands were obtained from the literature published by their developers and in some cases (Vaduz, Heidelberg, and Tomovic) by institutions which have formally evaluated them. This table has been constructed in an analytic rather than descriptive form in an attempt to compare and evaluate the significant features of each unit. The hands have been categorized in two classes. The first group includes devices whose development has reached the stage where they are currently available for clinical use. The second group includes hands which have been developed to the point where they are being or will be applied to patients for *limited* clinical evaluation, but whose availability for general clinic use would require perhaps one or two years of development work. Also included in the table as a reference is the "conventional" Sierra APRL VC Hand, Model No. 44.

Each column contains data relating to *currently used standards* for the evaluation of prosthetic hands. Two sets of standards are available (6) (7). One set is the Tentative Specifications for Voluntary Closing Hands (November 1, 1957) which have been used for many years to evaluate conventional hands. The other, not yet in general use, represents the only known set of functional standards for externally powered hands. Although incomplete and not yet formally accepted by authoritative agencies, the Tentative Design and Functional Standards for Electromechanical Hand and Control

Systems (June 1, 1967), prepared by USAMBRL have also been used for this analysis.

Both sets of standards contain a large number of specifications to cover almost all the requirements for prosthetic hands. For this analysis, we have selected seven specific elements from the standards. These items represent functional requirements for *any* prosthetic hand regardless of the source of power, type of control, or other specific characteristics. They relate to patient requirements and the result to be expected from the use of any artificial hand.

The seven items are listed below. Each is followed by a reference to its source.

1. The mechanical hand shall resemble the human hand to the extent that it conforms to the shape, size and other features of a standard hand designed by APRL (now USAMBRL) (III.A.1 1957).
2. Prehension shall occur between stationary thumb and active second and third digits (index and middle) (II.B.6. 1957).
3. The weight of the hand assembled shall not exceed 0.75 lb. (12 oz.) (III.E.6. 1957).
4. The hand must be capable of receiving and holding objects 2¾ in. thick between the fingertips (III.B.1. 1957).
5. At all locking positions the mechanical hand shall be able to maintain a 35 lb. pinching force between the prehensile fingers without slipping, jamming, or permanent distortion of any part (III.C.3. 1957).
6. Control should not actively involve the sound body side (III.B.5. 1967).
7. Control should be compatible with standard harnessing techniques (III.B.6. 1957).

The standard referred to in Row 1 (Table 2) rests on the degree of resemblance to the appearance and configuration of the normal human (male) adult hand. In general, these externally powered hands have been designed to simulate the appearance of the normal adult hand. However, neither the Heidelberg Hand nor the Tomovic Hand would meet even these vague specifications which nevertheless are in use today for evaluating hand appearance. The Heidelberg Hand has a generally unacceptable claw-like appearance. The Tomovic Hand is not considered adequate as a result of the protruding thumb position and its large size.

With respect to the type of prehension (Table 2, Row 2), it is generally acknowledged that the three-jaw-chuck palmar type of prehension is the most desirable single compromise type of hand closure because, statistically at least, it is used most frequently. Objects are more stable in a three jaw chuck, than in a 2-point pinch type of prehension. With one exception, all of the externally powered hands discussed provide three-jaw-chuck palmar prehension. The Heidelberg Hand provides only fingertip prehension in

TABLE 2.—*Summary of Hand Features*

Row	Standard for hands	Class I hands				Class II hands					Reference hand
		Soviet-Canadian	Italian	Heidel-berg	Vienna-tone	Tomovic	USAMBRL	AIPR	British	Paris-Vaduz	APRL
1	Cosmetically acceptable	Yes	Yes	No	Yes	No	Yes	Yes	Yes	Yes	Yes
2	Prehension										
	1. Three jaw, chuck	Yes	Yes	—	Yes	Yes	Stationary thumb	Stationary thumb	Yes	Yes	Stationary thumb
	2. Fingertip	—	—	Stationary index finger	—	—	—	—	—	—	—
	3. Grasp	—	—	—	—	Yes	—	—	—	—	—

3	Maximum opening range (in.)	3¼[a]	2¾	2	2½[a]	—	2½	2⅝	3¼	2⅝	2¾
4	Thumb prepositioning necessary	No	No	No	No	No	Yes	Yes	No	No	Yes
5	Weight (oz.)	12½	18⅜	—	16⅞	35	15½	12	20	14½	14
6	Prehension force (lb.)	4	17	22	18	4	12	12	6	7	55
7	Type of control										
	Primary										
	1. Gross body motion	—	—	Yes	—	Yes	Yes	Yes	—	—	Yes
	2. Fine body motion	—	—	—	—	—	—	—	—	Yes	—
	3. EMG	Yes	Yes	—	Yes	—	—	—	Yes	—	—
	Secondary										
	1. Audio-visual	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes
	2. Tactile	—	—	—	—	—	—	—	—	Yes	Yes
	3. Automatic within the hand	—	—	—	—	Yes	Yes	—	Yes	Yes	—

[a]A second model measured 3¼ in.

Generally the prehension forces available in the externally powered hands (Table 2, Row 5) lie between those found in conventional voluntary closing hands and voluntary opening hands. The Italian, Heidelberg, and Viennatone hands offer prehension forces approaching those of the normal human hand. The forces provided by the USAMBRL and the AIPR are adequate at approximately 12 lb. The Tomovic, Soviet-Canadian, British, and Vaduz hands fail to provide significantly higher forces than are available in almost any conventional hand which except for the voluntary closing type are generally considered inadequate.

The primary means by which these hands are controlled (Table 2, Row 6) varies considerably. The nine hands include four which are controlled by EMG signals generated in the flexors and extensors of the wrist. Three of the four, the Canadian, Italian, and Viennatone, utilize the EMG signals to turn the hand motor on or off. The magnitude of the EMG signal above a preset threshold is not significant. The *duration* of the signal at a strength above the threshold controls the position of the fingers and the prehension force applied. Thus, the longer the EMG signal persists, the longer the motor runs to close or open the hand. Closed on an object, the continued signal causes the prehension force to increase.

In the British hand, the *magnitude* of the EMG signal is more significant than the duration. Higher EMG magnitudes cause the hand to close or open more rapidly or to exert more force when closed on an object.

Four of the units are controlled by gross body motions—the Heidelberg, AIPR, Tomovic, and USAMBRL. The first two are controlled by overt movement of a body segment (stump, scapula, etc.) to actuate push or pull valves. These on-off valves control the passage of compressed CO_2 gas into actuators which open and close the prosthetic hand. The Tomovic hand utilizes a pressure switch mounted in the harness to control opening and closing. This device has "pressure sensors" in the fingers of the hand that control the *type* of prehension—palmar or grasp. In its present form the USAMBRL hand uses a conventional control cable arrangement to operate the hand.

The last hand, the Vaduz, is controlled by a fine body motion—bulging the muscles of the forearm initiates closure of the hand. Additional displacement, in the form of a larger bulge, must be applied for the hand to continue to close. Otherwise, the circuit automatically breaks and stops the hand from closing further. The fine displacement provided by the bulging of muscles is roughly proportional to displacement or force of the fingers of the terminal device.

In addition to the primary controls which actuate these hands, secondary controls are provided to adjust and regulate position, velocity, and/or force. Secondary control in the form of auditory and visual feedback signals is available in all nine hands as it is in all conventional devices. Tactile feed-

back, another form of secondary control, is available in conventional hands in that information from the hand (resistance to further opening or closing) is fed back through the socket and/or harness/cable system to the skin where it is converted to tactile sensation. Among these nine hands tactile feedback is furnished only by the Vaduz hand. The skin receives tactile sensation from the outer transducer which is directly linked to the inner transducer, a component reflecting the position and force applied by the hand. This form of tactile feedback should be clearly distinguished from sensations of muscle tension, as for example, the sense a patient has as he tenses his muscle to modulate an EMG signal to control a hand. In this instance, there is no feedback from the hand to the patient; only the sensation of tension within the muscle is fed back to the central nervous system which must then predict or see the resultant hand function.

In addition to conventional secondary control, three of the nine hands provide other forms of feedback. In the British hand velocity of closure and force applied are automatically adjusted to correspond with the magnitude of a given EMG signal. However, to complete the task of grasping and transporting an object, the patient must adjust the primary EMG control signal to appropriate levels. A special secondary control in the USAMBRL hand is the automatic adjustment of force required to prevent slippage. A sensor in the thumb feeds back information to the control amplifier which automatically increases prehension force to the required magnitude. An additional form of secondary control is used in the Vaduz hand to sense the position of and force applied by the fingers of the hand. This information is reflected back to the inner pneumatic transducer which transfers pressure to the outer pneumatic transducer lying against the skin of the forearm. The pressure is sensed by the amputee.

D. Conclusions

This study is a deliberately critical analysis of the design concepts of nine hands in relation to patient use. It is not intended and certainly should not be interpreted as criticism of the talent, creativity or productivity of the designers and developers of these hands. To the contrary, in our opinion almost all of these hands are truly engineering or electronic "marvels" revealing ingenuity and great engineering skill. Each is a very respectable technical product. The criticism has been focused on the last and weakest link in the chain between design concept and everyday use by amputees. The result reveals that despite the technical brilliance of many designs, the last link—utilization—had not been adequately considered *before* a great deal of development had been undertaken.

The following conclusions relate to the meaning of externally powered hands for patients. Their special features and their present stages of development are considered from the viewpoints of clinicians who contemplate

prescription of these devices and of administrators of programs which provide funds for the development of externally powered hands.

1. *Current Availability of Externally Powered Hands.* Of the nine hands considered, the following four are available for general clinic use: Viennatone, Canadian, Italian, and Heidelberg. At the present time the Canadian and Italian components are only available at the source. This means that in order to fit a patient today, he must be sent to Montreal or Bologna. These hands therefore are not practically available at the present time for American patients. It is conceivable, however, that the components could be made available but their integration into a prosthesis would require special training of prosthetists, additional facilities, or additional personnel. The Viennatone hand is available in this country through the Fidelity Electronics Company, 5245 Diversey Avenue, Chicago, Illinois. This is a fortunate circumstance since of the four hands in question, the Viennatone, although generally similar to the Canadian and Italian, is superior to the other three with respect to prehension force, durability, and ease of installation. The other five of the nine hands described are not yet generally available for clinical use.

2. *Prescription Indications.* For practical purposes, six of the nine hands, regardless of their present availability, are limited to below-elbow applications. The Heidelberg and the AIPR hands are designed as components of complete systems for above-elbow and shoulder-disarticulation patients. Their control systems are integrated with the overall control system for the entire prosthetic system. However, as a hand, the Heidelberg would appear to be the least effective of the entire series studied. Although the AIPR hand meets or comes close to meeting most of the currently applied specifications, it is not available. According to the developer, further work on this unit has been terminated because the high level unilateral and bilateral amputees (for whom the entire system was designed) were found to perform better with a hook. The USAMBRL hand depends for primary control on a cable pull to start and stop the motor. It is therefore readily adaptable for use with conventional cable control systems. Five of the other six hands (Viennatone, Canadian, Italian, British, and Vaduz) would require the utilization of a new control site in addition to those already used in above-elbow and shoulder-disarticulation prostheses. Current opinion indicates that above-elbow and shoulder-disarticulation patients with conventional harnesses and components are already operating the maximum number of components and are utilizing the maximum number of motions of which they are capable. Although the sixth (Tomovic) hand is also designed for primary control by a pull cable, its present weight at 35 oz. must be regarded as excessive for any patient.

3. *Advantages.* One clear advantage of hands controlled by EMG or muscle bulges is the fact that they can be conveniently operated in many

positions where it is difficult to operate conventional hands. Due to harness restriction and cable excursion requirements, conventional hands are difficult to operate when the shoulder is abducted or flexed near or above 90 deg., or when the hand is positioned well over toward the nonamputated side. The operation of EMG controlled hands is not affected at these positions. However, there is no evidence to indicate that performance, i.e., quality of grasp, facility in handling objects, etc. would be improved without visual feedback. Indeed, lack of visual feedback in positions where EMG *should* be most useful would be a limitation.

4. *Automatic Control.* There has been a great overemphasis on the value of automatic control of certain hand functions. The utility for the patient of automatic proportional control of finger position, velocity, and/or force is questionable where conventional feedback in the form of visual, auditory, and tactile cue is also available. With the exception of the USAMBRL hand, the automatic controls which are emphasized in the British, Tomovic, and Vaduz hands, are really quite secondary control factors. After all, none of these hands relieves the patient of primary decision making with respect to grasping, regrasping, transporting, and releasing objects. With all the internal feedback available, the patient still depends on seeing the relationship between finger position, speed, and/or force and the object in order to operate the hand successfully. The USAMBRL hand, however, goes a half step farther. Although the patient must maintain an input signal, he does not have to make any decisions about the force to apply where non-crushable objects are to be grasped. He must decide voluntarily how to control prehension force where crushable objects are to be grasped.

E. Recommendations

1. Since a great deal of excitement has been generated about complex but as yet unevaluated hand functions, we recommend a study be undertaken to reevaluate hand design requirements with a view toward reviewing and perhaps recasting current hand specifications. Current specifications for mechanical hands were originally designed on the basis of the mechanical APRL voluntary closing hand based on UCLA studies, Northrop experiments, and other data. The proposed specifications for electromechanical hands include the original specifications for mechanical hands. Thus, questions of thumb function, minimum prehension force, types of prehension, and the like remain moot.

2. We recommend that a continuing program be initiated to evaluate all hands whether externally or muscle powered. This program should insure the application of similar, reliable, and valid procedures in all the hand evaluations. Both laboratory and objectively measured clinical studies are essential.

3. We recommend that strong consideration be given to adapting available externally powered devices for use with conventional prosthetic systems for above-elbow and shoulder-disarticulation patients. Any of the several EMG controlled externally powered devices might be readily redesigned for control by conventional cable systems. Hybridization might well provide above-elbow and shoulder-disarticulation patients with very high prehension forces at very low input forces and excursions, a generally agreed upon need for this type of patient.

4. We recommend that administrators consider potential utility for patients *before* granting funds for design and development of external hands. Applicants for funds should be required to include in their proposals an objective discussion of each hand with respect to types of patients it will serve, improvements to be expected in prehension and other patient activities, and its compliance with current principles and standards for amputee management.

3. *Perlon Cable Transmission System.* The "conventional" Bowden cable transmission system used in this country for powering upper-extremity prostheses consists of a length of stainless steel cable (with appropriate steel fittings swaged to each end) which slides inside a tightly coiled steel housing. This system has generally provided efficiencies of 75 percent or greater, during routine wear. However, fraying of the steel cable is not uncommon. It usually occurs at the site of the terminal fittings and leads to decreased efficiencies, and ultimately, to cable breakage.

An upper-extremity transmission system which employs a nylon filament, called Perlon, as the control cable has been used increasingly in Europe. Recently this material, which is manufactured in West Germany, was made available to the VA Prosthetics Center, through Otto Bock Orthopedic Industry, Inc., Minneapolis, Minn.

Substituting Perlon cable for steel cable in otherwise conventional upper-extremity transmission systems has proved highly satisfactory in terms of amputee acceptance and durability. Seven subjects have worn the Perlon cable with conventional Hosmer standard steel cable fittings and Teflon-lined steel cable housing for periods of 4 months to 1 year with no problems.

Further studies have been undertaken in which the steel housing has been replaced with a length of nylon tubing (Nylaflow). This material is commercially available and used in high pressure, hydra-pneumatic systems.

Since June 1967, thirty-two upper-extremity prostheses have been equipped with cable transmission systems of Perlon cable and Nylaflow housing. The use of heavy duty housing retainers was required due to the thickness of the Nylaflow tubing (O.D. 3/16 in.).

The results to date are highly satisfactory; only two negative experiences were noted. In one case the steel ball terminal pulled off the Perlon cable during the 50 lb. axial loading test at the initial checkout of the prosthesis. In

the other case, after two months wear, a right above-elbow subject reported "binding" in the transmission system when his prosthesis was flexed 130 deg. Lubricating the Perlon cable with "Slip Spray," a dry lubricant, corrected the problem.

Simultaneously, a comparative evaluation of the Perlon cable-Nylaflow housing system with both conventional and heavy duty systems has been undertaken in the laboratory. The durability, tensile strength, and efficiency of all three systems have been compared. The actual test procedures are outlined in BPR 10–5 (p. 138). The results appear in Table 3.

Reference to Table 3 indicates that the Perlon cable in the Nylaflow tubing system withstands significantly less tension than conventional steel cables. The Perlon cable system failed under 70 lb. of tension when the cable pulled out of the swaged ball fitting. The cable itself failed at 125 lb. In general, the steel cable systems withstand two to three times higher tensile loads than the Perlon system. The Perlon system elongated approximately 11 percent of its original length under 70 lb. of tension and as much as 16 percent under 125 lb. of tension. This would appear to be excessive "creep" which might possibly be overcome by prestressing. However, the recovery of Perlon, i.e., its memory, is such that it recovers from its creep under load and reattains its original length when unloaded. Surprisingly, in view of subjective reactions received to date, efficiency of the Perlon-Nylaflow tubing system is somewhat lower than Bowden systems. When the Nylaflow tubing is flared with either a cold tool or a drill, the efficiency is higher than when it is flared with a hot tool. This finding is related to the apparent deformation of the heated Nylaflow material on the surfaces of both inside and outside walls. In cycling tests to determine durability, neither the Perlon cable nor the Nylaflow tubing showed measurable wear after 25,000 cycles, a point when the Teflon tubing used in a Bowden conventional system showed considerable wear.

C. Evaluation (Techniques)

Direct Forming of Below-Elbow Sockets. As reported in previous issues a technique has been developed for forming below-elbow sockets directly on a stump using Polysar. In contrast to the fabrication procedure for below-knee sockets described above (see Section I.C.), the VAPC pneumatic casting system is not used. Because Polysar conforms faithfully to surfaces over which it is drawn, it replicates the ribs of stockinet used to cover the stump and may produce a relatively rough inner socket surface. To avoid this possibility, a rubber sleeve (Fig. 25) is drawn directly over the stump. The rubber sleeve compresses the tissues, producing a firm stump for precise fit, making it easier to pull the Polysar tube onto the stump. The rubber sleeve is reusable.

To date, six subjects have been fitted with below-elbow Polysar sockets. Results have been uniformly satisfactory in that the sockets remained dimen-

TABLE 3.—*Elongation, Efficiency, and Tensile Strength of Three Upper-Extremity Cable Transmission Systems*

System	Elongation of cable, percent (approx.)	Efficiency of system percent Flex.	Efficiency of system percent Ext.	Tensile strength of cable, lb. 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
Bowden, 1/16 in. steel cable, standard steel housing, heavy duty steel components	0–1.5	72	91	92 Steel ball pulled out of receiver	275 Cable pulled out of swaged harness connector fitting	445 Cable broke at point of attachment to lower test grip
Perlon cable 1/16 in., Nylaflow tubing 7/64 in. I.D. by 3/16 in. O.D., 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			

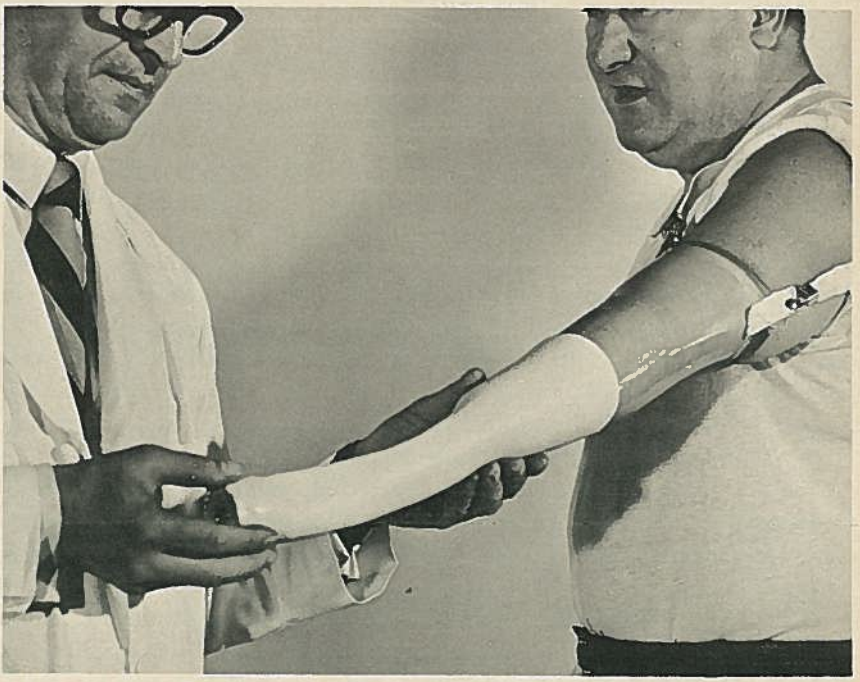

FIGURE 25.—Forming below-elbow sockets directly on a stump utilizing a latex rubber sleeve (between stockinet and Polysar tube) to compress tissue and provide a smoothly finished inner socket surface.

sionally stable and fittings were retained. The only negative indication concerned excessive perspiration experienced by one patient, a matter now under investigation.

A manual has been prepared detailing the current fabrication procedure. It is designed to guide New York University in a test undertaken to evaluate the fabrication procedure and the general utility of below-elbow Polysar sockets.

III. LOWER-EXTREMITY ORTHOTICS

A. Development

Development of Specifications for Orthotic Components. The field of orthotics has not received as much of the attention of developers and researchers as the field of prosthetics. Including appliances for the upper and lower extremity and the spine, orthotics presents a very wide variety of problems which are often extremely complex. It is sometimes far simpler to replace the significant functions lost through amputation than to restrict unwanted motion and to permit desired motion at the same time. Many of the problems are due to a multiplicity of disabilities which must be overcome simultaneously.

There are literally hundreds of orthotic components in the catalogs of manufacturers and many others which are normally custom-made in shop facilities to meet these needs. It is often difficult to select from this bewildering array the best item for a given case. Contributing to this difficulty is the lack of an adequate terminology which describes orthotic components in relation to the specific functions they perform. Recently, the American Academy of Orthopaedic Surgeons has undertaken, under the direction of Dr. George T. Aitken, Grand Rapids, Michigan, the difficult task of developing an adequate functional nomenclature for orthotic components. The VA Prosthetics Center is among the group of individuals and agencies whom Dr. Aitken has called upon to participate in this monumental task. Our work with this group has not only confirmed the great need to categorize the numerous lower-extremity orthotic components *but to develop for VA use, a set of functional standards for these devices.*

The development of standards for brace components depends on a clear understanding of the specific functions of all the items as a basis for establishing specifications. As a first step we have compiled a list of orthotic items currently in use, and we have roughly categorized them in terms of their purported functions. Brace components for the lower-extremity segments including foot, ankle, shank, knee, femur, and hip have been described in terms of the motions they prevent, limit, resist, or permit. A ring lock on a knee joint, for example, *prevents* motion. A 90-deg. stop on an ankle joint *limits* motion. A spring-loaded ankle joint *resists* motion. A free motion knee joint *permits* motion.

The items listed below are arranged in categories related to the segment of the lower extremity to which they are applied. Also included are the sources, materials, and the typical applications for each component.

FUNCTION, SOURCE, AND MATERIALS OF LOWER-EXTREMITY ORTHOTIC COMPONENTS

COMPONENTS	BIOMECHANICAL FUNCTION	SOURCE*	MATERIAL
A. Hip Control			
1. Hip twister with pelvic band	Resists excessive internal or external hip rotation	Becker	Coil spring housed within plastic tube
2. Hip joint with free motion and pelvic band	Prevents hip abduction/adduction	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
3. Hip joint assembly with positive stop and pelvic band	Prevents hip hyperextension, and hip abduction/adduction	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
4. Hip joint assembly with drop ring lock and pelvic band	Prevents hip flexion/extension and hip abduction/adduction	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
5. Hip joint assembly with lever lock and pelvic band	Prevents hip flexion/extension and hip abduction/adduction	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
6. Hip joint assembly with plunger lock, abduction and/or adduction hinge, and pelvic band	Prevents hip flexion/extension and limits hip abduction/adduction	Pope	Carbon steel
7. Hip joint assembly with hyper-extension control, abduction and/or adduction control and pelvic band	Prevents hip hyperextension and limits hip abduction/adduction	Pope	Carbon steel
8. Silesian Belt	Resists excessive hip abduction and external hip rotation	Usually custom made	Metal and fabric
B. Weight Bearing Control (Thigh)			
1. Quadri-lateral socket	Prevents weight application to lower extremity	Usually custom made	Plastic laminate or wood
2. Ischial weight-bearing band with ischial plateau	Prevents weight application to lower extremity	Fillauer	Padding with leather or plastic sheeting cover
3. Ischial ring (half) with ischial plateau	Limits weight application to lower extremity	Usually custom made	Padding with leather or plastic sheeting cover
4. Ischial ring (full) with ischial plateau	Prevents weight application to lower extremity	Usually custom made	Padding with leather or plastic sheeting cover

See footnotes at end of table.

COMPONENTS	BIOMECHANICAL FUNCTION	SOURCE*	MATERIAL
C. Knee Control (Single or Bilateral Bar)			
1. UCLA Functional Long Leg Brace	Prevents knee flexion (stance phase)	U.S. Mfg.	
AK Socket (non-weight bearing)	Reaction point for side bars	Custom	Polyester laminate
Knee Joint, offset	Prevents knee flexion (stance phase)	U.S. Mfg.	Stainless steel
Pre-tibial Cuff	Reaction point for side bars	Custom	Plastic laminate
Hydraulic Ankle joint assembly	Permits plantar flexion, prevents dorsiflexion	U.S. Mfg.	Aluminum, stainless steel, carbon steel
2. Knee Joint, Polycentric	Prevents knee flexion (stance phase)	Becker	Aluminum
3. Knee Joint, Single Axis, Free Motion	Permits normal knee motion in the AP plane and resists motion in the ML plane	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
4. Knee Joint with hyperextension control	Prevents knee hyperextension and flexion/extension	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
D. Knee Component Controls			
1. Knee joint with drop ring lock	Prevents knee flexion/extension in AP plane and resists in ML plane	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
2. Knee joint with plunger lock	Prevents knee flexion/extension in AP plane and resists in ML plane	Pope	Chrome molybdenum
3. Knee joint with lever lock	Prevents knee flexion/extension in AP plane and resists in ML plane	Becker, Pope, U.S. Mfg.	Aluminum, stainless steel, carbon steel, chrome molybdenum
4. Knee joint with turn buckle	Limits flexion/extension in AP plane and resists in ML plane	Otto Bock, Becker, Fillauer	Carbon steel, stainless steel
5. Single-bar joint with hyperextension stop	Prevents hyperextension of the knee	VAPC	Stainless steel
6. Single-bar joint with lever lock	Prevents knee flexion/extension	VAPC	Stainless steel
7. Knee joint, hyperextension control (knee cage)	Prevents knee flexion/extension	Atco, Hastings-McIntosh, Horn, Truform	Stainless steel, aluminum, uprights and leather or fabric covering
8. Knee joint, drop ring lock (knee cage)	Prevents knee flexion/extension	Horn, Truform	Stainless steel, aluminum, uprights and leather or fabric covering

E. Patellar-Condylar Weight Bearing Control			
1. PTB socket	Prevents weight application to lower leg, foot, ankle	Usually custom made	Plastic laminate posterior opening
F. Ankle Controls and Uprights (Single or Bilateral Design)			
1. Leg uprights, spring-loaded ankle joint assembly with one piece or split stirrup	Limits ankle plantar flexion and resists dorsiflexion	Pope, U.S. Mfg., Becker, Fillauer	*Uprights:* aluminum, stainless steel, carbon steel, chrome molybdenum *Stirrup:* stainless steel, carbon steel
2. Leg uprights, positive stop ankle joint assembly with one piece or split stirrup	Limits ankle plantar flexion, resists dorsiflexion and resists inversion/eversion	Pope, U.S. Mfg., Becker, Fillauer	*Uprights:* aluminum, stainless steel, carbon steel, chrome molybdenum *Stirrup:* stainless steel, carbon steel
3. Leg uprights, positive stop ankle joint assembly with caliper	Limits ankle plantar flexion and permits dorsiflexion	U.S. Mfg., Pope, Becker, Fillauer	*Uprights:* carbon steel, stainless steel, chrome molybdenum *Caliper Pins and Box:* carbon steel, stainless steel
4. Stirrup-upright comb. (no joint at ankle)	Limits ankle dorsiflexion/plantar flexion—resists inversion/eversion	Usually custom made	Carbon steel or stainless steel
5. Single leg upright, spring-loaded ankle joint assembly	Limits ankle plantar flexion. Permits axial rotation and displacement	VAPC	Stainless steel, aluminum
6. Single leg upright, single joint, positive stop	Limits plantar/dorsiflexion	VAPC	Stainless steel, aluminum

See footnotes at end of table.

COMPONENTS	BIOMECHANICAL FUNCTION	SOURCE *	MATERIAL
G. Leg Uprights and Ankle Joint Assemblies (Complete Braces)			
1. Posterior upright with foot plate	Resists ankle dorsiflexion	Usually custom made	Blue lock spring material
2. Posterior upright and caliper tube	Resists ankle dorsiflexion	American Brace	Carbon steel
3. Bilateral (wire) uprights with Caliper pin or machine screw and nut for shoe attachment	Resists ankle dorsiflexion	Custom made	Lyman Smith-Music Wire
4. Molded Leg-Ankle-Foot Support (fits within shoe)	Prevents ankle dorsiflexion/plantar flexion, subtalar eversion/inversion and forefoot motion	Custom made	Polyester laminate or molded leather over plaster-of-paris cast reinforced with stainless steel or carbon steel uprights and foot plate
5. Dennis Browne splint	Club foot brace	U.S. Mfr., Pope, Becker, Fillauer	Aluminum, stainless steel, carbon steel
6. UC-BL coaxial mechanical joint	Prevents ankle plantar flexion and provides dorsiflexion		*Upright:* Carbon steel, aluminum *Stirrup:* Carbon steel, stainless steel
7. Biaxial spring-loaded ankle joint		UC–BL	
8. Biaxial positive stop ankle joint		UC–BL	
H. Component Accessories			
1. Knee cap	Prevents knee flexion	Custom made or purchased from R & G	Leather or plastic sheeting, and/or fabric
2. Under patellar strap	Prevents knee flexion	Custom made or purchased from R & G	Leather, or plastic sheeting, and/or fabric
3. Knee cap for knock knee	Prevents knock knee	Custom made	Leather, or plastic sheeting, and/or fabric
4. Knee cap for bow knee	Prevents bow knee	Custom made	Leather, or plastic sheeting, and/or fabric

5. Pressure Pad for Knock Knee	Prevents knock knee	Custom made	Sponge rubber or felt covered with leather or plastic sheeting
6. Pressure Pad for Bow Knee	Prevents bow knee	Custom made	Sponge rubber or felt covered with leather or plastic sheeting
7. "T" or "Y" strap (med.)	Prevents valgus	Custom made or purchased from R & G	Leather, or plastic sheeting, and/or fabric
8. "T" or "Y" strap (lat.)	Prevents varus	Custom made or purchased from R & G	Leather, or plastic sheeting, and/or fabric
9. Ankle Pressure Pad (med.)	Prevents valgus	Custom made	Sponge rubber or felt covered with leather or plastic sheeting
10. Ankle Pressure Pad (lat.)	Prevents varus	Custom made	Sponge rubber or felt covered with leather or plastic sheeting
11. Heel Wedge (med.)	Prevents valgus	Custom made	Leather or neoprene
12. Heel Wedge (lat.)	Prevents varus	Custom made	Leather or neoprene
13. Sole Wedge (med.)	Prevents eversion of forefoot	Custom made	Leather or neoprene
14. Sole Wedge (lat.)	Prevents inversion of forefoot	Custom made	Leather or neoprene
15. Rocker Bar	Prevents application of weight to 1st and 5th mets. Permits rollover from midstance to toe-off.	Custom made	Leather or neoprene
16. SACH Type Heel Wedge	Prevents application of weight to heel and permits plantar flexion.	Custom made	Leather
17. Thomas Heel	Prevents valgus	O'Sullivan or Bilt-Rite	Rubber or neoprene
18. Reversed Thomas Heel	Prevents varus	O'Sullivan or Bilt-Rite	Rubber or neoprene
19. Spreader Bar	Prevents hip adduction	Usually custom made	Carbon steel or stainless steel

See footnotes at end of table.

*Complete names and address of commercial sources follow:

American Biltrite Rubber Co., Inc.
Chelsea, Massachusetts 02150

American Braces
997 Vermont Street
Oakland, California 94610

Atco Surgical Supplies, Inc.
450 Portage Trail
Cuyahoga Falls, Ohio 44222

Becker Orthopedic Appliance Co.
1776 South Woodward
Birmingham, Michigan 48009

Otto Bock Orthopedic Industry, Inc.
219–14th Avenue North
Minneapolis, Minnesota 55411

Hastings-McIntosh Truss Co.
912 Walnut Street
Philadelphia, Pennsylvania 19107

William H. Horn & Bro., Inc.
451 North 3rd Street
Philadelphia, Pennsylvania 19123

The Pope Brace Company
197 South West Avenue
Kankakee, Illinois 60901

R &G Orthopedic Appliances
2037 K Street, N.W.
Washington, D.C. 20006

Truform Anatomical Supports
3960 Rosslyn Drive
Cincinnati, Ohio 45209

United States Manufacturing Co.
623 South Central Avenue
P.O. Box 110
Glendale, California 91209

O'Sullivan Rubber Corp.
P.O. Box 603
Winchester, Virginia 22601

Several surveys of the field are being planned to determine the extent to which these components are used in these and other applications.

B. Evaluation (Components)

VAPC Single-Bar Ankle Brace. In the previous issue (BPR 10-7) we reported the evaluation of a subject fitted bilaterally with the VAPC Single-Bar Ankle Brace and custom orthopedic shoes. The results indicated that the Single-Bar Ankle Brace improved patient comfort as a result of: 1. lighter weight, 2. absence of a medial sidebar, and 3. axial displacement and transverse rotation inherent in the brace configuration. The subject's performance with the brace was characterized by a narrower walking base, longer stride length, and smoother weight application in stance phase. However, these improvements could not be causally related solely to the VAPC brace since new custom orthopedic shoes with sponge innersoles, soft heels, and an external metatarsal bar were also worn with it.

To discriminate the specific effects due to the VAPC ankle brace and the new orthopedic shoes, the subject was refitted with the single-bar braces and with the stock shoes previously worn with his conventional braces.

After 60 days, the evaluation procedure was repeated. An analysis of the data obtained indicated that the single-bar brace was the principal factor in improvement of *performance,* independently of the shoes. The subjective reactions of the patient, however, indicated that the new orthopedic shoes contributed to improved *comfort.*

In general, the subject's performance with the VAPC braces was better than with the conventional as indicated by narrower walking base and longer stride length (Table 4). These effects were found regardless of the type of shoe worn with the VAPC brace. Our previous findings of improved comfort with the VAPC brace were due principally to the use of the new orthopedic shoes worn with them. This conclusion is drawn from the fact that when the VAPC brace was worn with the old shoes, the patient indicated that he had pain in the arch area and that his walking range was significantly reduced, i.e., he tired more quickly.

TABLE 4.—*Walking Base and Stride Length*

Braces	Walking base width, in.	Stride length, in.	
		Left	Right
Conventional with stock shoes	3⅝	41⅓	41¾
VAPC with stock shoes	2½	44¾	45
VAPC with custom shoes	2⅝	46½	46

Additional comments from the patient indicated that the VAPC brace was preferred over the conventional because he "walks easier," and it "gives him a lift."

To date, 12 patients (17 braces) have been fitted with the VAPC Single-Bar Brace with satisfactory results. Components are being manufactured commercially for 50 braces for field evaluation.

IV. ORTHOPEDIC AIDS

A. Development

None

B. Evaluation (Components)

1. Power-Aid for Wheelchair. Developed by California Medical Aids, Montrose, California, this unit is designed to replace one of the casters on a conventional wheelchair converting it from a hand-propelled chair to a power-driven chair. Either the right or left caster is removed and the power unit is attached to the front vertical bar of the wheelchair frame by means of a specially designed bracket (Fig. 26). The complete power unit consists of an electrical motor housed directly above the drive wheel to which it transmits power by means of a chain drive. The motor is driven by a 12-volt battery. The battery is encased in a plastic container and mounted on brackets resting on the horizontal bars beneath the seat. A 12-volt storage battery charger is furnished with the unit. Attached to the unit is a vertical tubular steering column housing a control cable which is attached to the speed control lever. The topmost portion of the vertical steering column is bent to form a horizontal tiller with a vertical handle to rotate the tiller. Squeezing the speed control lever furnishes power to the drive wheel in proportion to the force exerted in squeezing. By rotating the tiller 180 deg. and squeezing the speed control lever, the chair moves in "reverse."

The wheelchair can be folded for storage after the battery and its mounting bracket are removed, a process in which the battery and its bracket are simply lifted off the lower horizontal bars of the wheelchair frame (Fig. 27).

The device, mounted on a conventional International Hospital Supply Wheelchair, was evaluated on the basis of a biomechanical analysis of operating requirements and analyses of its physical, electrical, and mechanical characteristics. It was also evaluated against criteria established for Automotive Wheelchairs, VAPC, July 1, 1957; a limited clinical study was also conducted.

The entire unit including the chair, battery, and the power aid weighs 101.25 lb. The chair weighs 44.5 lb., the battery and bracket 39.0 lb., and the power aid 17.75 lb.

FIGURE 26.—The power aid converts any conventional wheelchair to a motorized wheelchair by replacing one caster wheel.

The vertical projection of the center of gravity with and without passenger falls 8.6 in. and 9.3 in. respectively in front of the axles of the conventional drive wheels. These figures are respectively 1.9 in. and 2.7 in. farther forward than in the standard I.H.S. chair without the power unit. With the center of gravity placed so far forward, the chair is exceptionally stable on level ground.

With certain improvements the power aid could be a useful device due principally to its versatility, ease of installation, and relatively light weight. Any standard body-powered wheelchair may be readily converted to an electrically powered chair. This may reduce the need for two chairs among those patients who ordinarily use a powered chair in the home or hospital and who also need a portable one for use in different environments. With a means to uncouple the drive wheel from the motor, it would also offer other patients the choice of two operational modes: power drive to offset fatigue or during early periods of convalescence when debility may be common; hand propulsion for short periods or as a therapeutic measure.

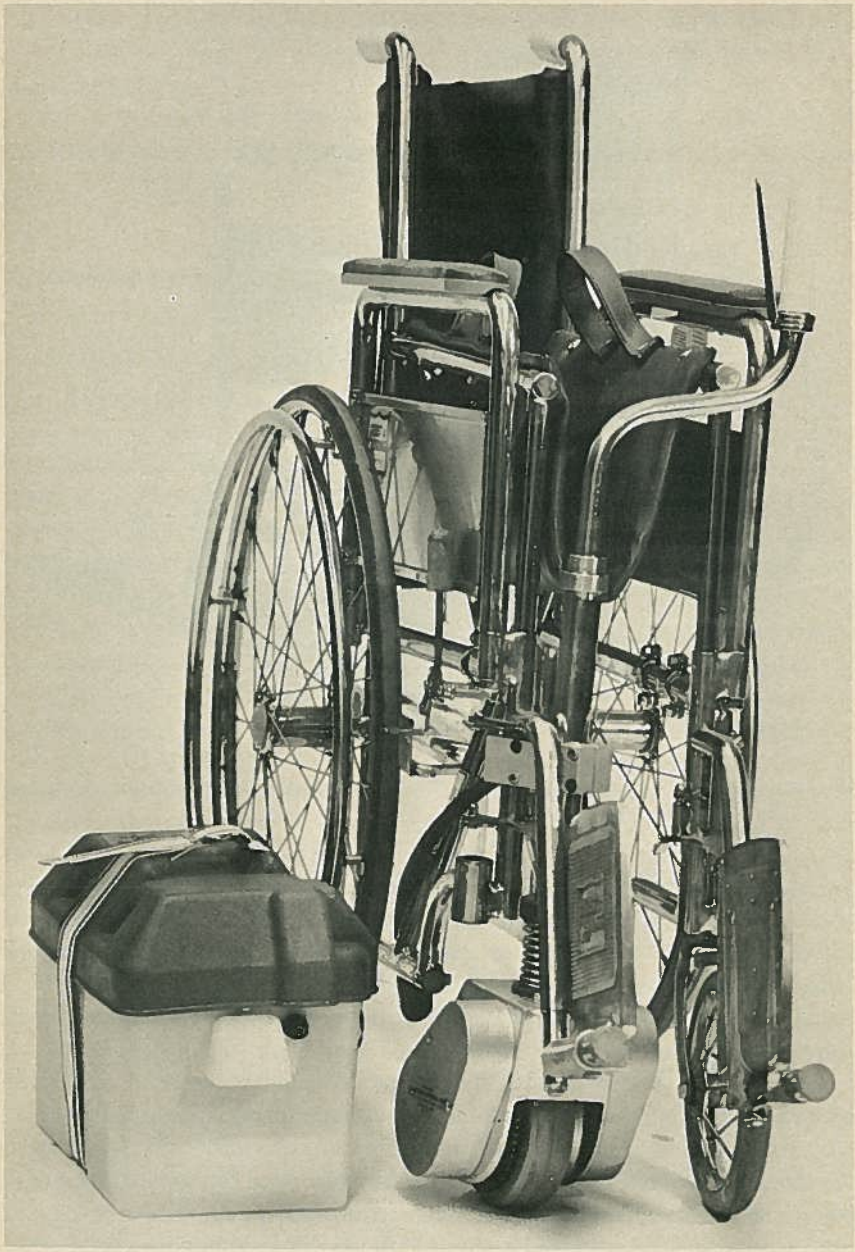

FIGURE 27.—By removal of the battery the unit can be folded for storage or transportation.

With a properly modified unit, hospitals and similar institutions could provide wheelchair patients with power-driven chairs early in their convalescence period; as strength increases, the unit can be removed and the chair, specifically prescribed for the patient, may be continued in use.

2. *Clinical Study of Stryker Floatation Pad.* In BPR 10–7 we reported our findings on the use of the Stryker Floatation Pad by five paraplegic patients who used the device for periods of from 1 to 5 months. Also included were findings from several other field stations and clinicians on their experiences in using the pad for approximately 20 additional patients. The reactions of these clinicians varied extensively with several registering very strong approval, others moderate approval, and still others with highly limited approval.

An analysis of all the data available at that time indicated that use of the Stryker Floatation Pad in hospital facilities over a period of 1 to 5 months resulted in the marked reduction of ulcers. Patients who derived the most benefit were those with histories of susceptibility to ulceration, generally impoverished skin, and those unable to sit up with other types of pads.

Additional data are available after 6 more months of use by four of the five patients previously reported. These four patients had been using the device at the time of this report for slightly more than 6 months. The fifth patient had only used it for 90 days and was therefore not included.

All four patients, previously reported as significantly improved after use of the Stryker Floatation Pad, have developed pressure sores. Three subjects had been discharged from the hospital; the other subject remained an inpatient.

The inpatient had terminated use of the Stryker Floatation Pad after 5 months when pressure sores developed. The other three patients are now unable to sit up due to pressure sores which developed in areas of the coccyx, ischium, and trochanters respectively. These three patients are all now confined to bed and gurney stretchers. They are, however, continuing use of the Stryker Floatation Pad in bed.

Our experience to date indicates that although the Stryker Floatation Pad seems to retard skin breakdown among patients confined to chairs and beds, its value as a preventive depends on hospital supervision. The pad permits patients to sit up sooner than might be expected and to increase the sitting time. When used by outpatients, the pad seems to have less therapeutic benefit, perhaps as a result of reduced professional supervision. Therefore, the best utilization of the Stryker Floatation Pad seems to be for hospital patients with severe ulcerations which prevent sitting up with other types of cushions. Outpatient use may not provide the professional supervision required to obtain the most benefit.

Additional comments indicated that the Stryker Floatation Pad is rather heavy and inconvenient during transfer activities. Otherwise, it was found easy to keep clean.

V. TESTING

A. Standards Development Program

The standards development program has produced tentative functional standards for above-knee knee mechanisms and for prosthetic feet. Standards and specifications for both these components are currently being evaluated (BPR 10–7). We are shifting the focus of this program to the development of standards for stump socks and elastic hosiery. The work on stump socks standards is in a preliminary stage in which we are defining fundamental factors relating to patient preference—utility, comfort, and durability. In this connection, we are collaborating with representatives of the U.S. Testing Company of Hoboken, New Jersey, in studies to validate selected factors such as softness, weight, moisture absorption, abrasion, and stretch-shrink qualities. The work on standards for elastic hosiery is described below.

Elastic Hosiery. The development of standards for elastic hosiery has been impeded by the lack of adequate methods for objectively measuring their effects. It is common experience that elastic hosiery prescribed for such conditions as phlebitis, lymphedema, varicose veins, and the like, afford patients significant relief of such symptoms as pain, edema, and inability to stand or walk for normal periods of time. The most widely accepted theory describes the function of the hosiery as that of reducing hemostasis in the lower extremities occasioned by vascular disorders in which the walls of the veins are distended with consequent loss of valve function and/or seepage of blood fluids into interstitial spaces. By the application of external compressive forces, elastic hosiery presumably aids venous return. Just how this is accomplished is not altogether clear. Do these compressive forces, for example, retard seepage from the veins? Do they reestablish the structural integrity of the veins and consequently reestablish valve function? Do they furnish a form of "milking action" by constantly applied forces?

Elastic hosiery is prescribed on the basis of length and circumferential dimensions of the affected limb. Knit geometry and hose dimensions are presumably designed to apply the required compression. One manufacturer, for example, supplies elastic hosiery which he describes as producing controlled pressures of 25 mm. Hg., 30 mm. Hg., 40 mm. Hg., and 50 mm. Hg. In addition, these stockings are described as providing negative pressure gradients, that is, pressures which are higher in the ankle area and decrease gradually in the more proximal areas. However, when these stockings were tested, neither the described pressures nor the gradients could be demonstrated.

In ordering these stockings for two subjects the recommended procedure was carefully followed. The samples were tested by the method previously described in BPR 10–6. The results are shown in Figure 28. The curves show the pressures exerted by the elastic hosiery designed to produce com-

pression forces of 40 mm. Hg. at least in the ankle area. Sample A exerted a pressure of approximately 40 mm. Hg. at one point only—9 in. above the floor (6 in. above the malleolli). Sample B exerted a maximum pressure of approximately 30 mm. Hg. over an area between 8 and 10 in. from the floor (between 5 and 7 in. above the malleolli). Both samples produced the *lowest* pressures in the ankle region. In fact, both produced a *positive* gradient between the ankle and an area approximately 6 in. above the malleolli. Thereafter, the gradient was generally neutral or slightly *negative*. When the samples were crossed, that is, when the stocking ordered for subject A was stretched to the measurements of subject B, the compression picture improved noticeably. In one case (Fig. 29) compression forces averaged around 25 mm. Hg. with a fairly neutral gradient.

Our experience to date indicates extreme variability in both magnitude and pressure gradients applied by stock samples of commercial elastic hosiery. Secondly, "custom-fitted" elastic hosiery provided neither the negative gradients nor the pressure magnitudes advertised.

We recognize the possibility that these data may not be completely reliable

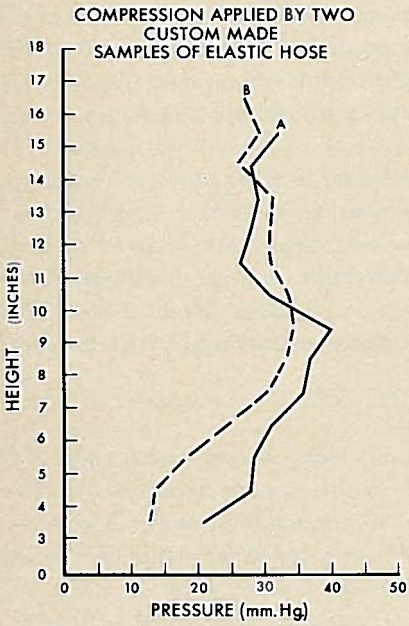

FIGURE 28.—Compression forces exerted by two "custom made" elastic hose (A and B) apply an undesirable positive gradient between the ankle and a point midway up the leg.

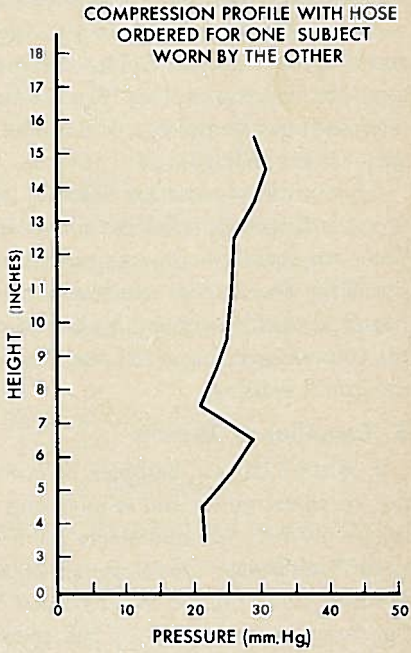

FIGURE 29.—When "custom made" elastic hose ordered for one patient was worn by another thereby "decustomizing," the compression forces were generally neutral except for a small positive gradient noted just above the ankle.

because no specific test procedure has been standardized. In addition, the test employed may not actually simulate the true conditions of wear as faithfully as we might hope. For this reason, these studies are being pursued with a more sensitive instrument to replicate as nearly as possible conditions of actual use.

Despite these areas of doubt, there is general agreement on certain fundamental requirements for elastic hosiery. Based upon these significant areas of agreement, we have drawn up a set of tentative standards for the procurement of elastic support hose. In scope, these standards cover all below-knee elastic support hose used by adult males in the treatment of venous return insufficiencies, varicose veins, or lymphedema.

Two functional standards are specified. One relates to the magnitude and the other to the gradient of pressures applied by elastic hose.

The following terms are used to describe pressures and gradients. A *negative* pressure gradient is one in which the pressure decreases as one moves proximally. A *neutral* gradient is one in which pressure neither increases nor decreases as one moves from proximal to distal. A *positive* gradient is one in which the pressure increases as one moves proximally along the limb.

All qualified hose should provide a standard pattern of radial pressure and a specific gradient. The maximum pressure should not be less than 20 mm. Hg., nor more than 80 mm. Hg. All pressure measurements should be made at elevations from 4 in. above the floor to the top of the hose at the garter line (BPR 10–5).

The elastic support hose should be fabricated so as to provide a *negative* or neutral pressure gradient when used by a patient whose actual limb dimensions are maximum for a particular hose size. Seams and changes in knit geometry should not concentrate uncomfortable or high local pressures. Material that is irritating to the skin should not be used. After 20 washings, the compressive characteristics should not change more than 10 percent from the initial readings.

B. Compliance Testing

1. Stump Socks. Samples of wool stump socks, cotton stump socks for the lower extremity and wool stump socks for the upper extremity were submitted by five manufacturers for testing. In continuation of our Specifications Compliance Test Program all samples were tested for compliance with specifications detailed in Invitation Number 68–4, dated May 10, 1967. All the samples, submitted by the following companies, conformed to the requirements: Bennington Stump Sock Corp., Accurate Knitting Mills, Inc., John McCann Co., Ohio Willow Wood Co., and Knit-Rite Co.

2. Upper-Extremity Components. Annual compliance tests were conducted on the Sierra Voluntary Closing Hand and the Hosmer Internal Elbow during this period. Both met all the requirements of currently applied specifications.

VI. OPERATIONS REPORT FOR FISCAL YEAR 1967

The VA Prosthetics Center provided 58,147 services as compared with 55,753 services provided in Fiscal Year (F.Y.) 1966. We increased the quality of services by a significant additional use of temporary prostheses and the introduction of prefabricated braces and inlay shoes. We were thus able to reduce the waiting period between the receipt of a request for an appliance and delivery. Consequently, this resulted in savings of thousands of dollars, both by the early release of hospital beds and by the lower initial cost of the ortho-inlay shoes and prefabricated brace parts, and the use of temporary prostheses.

The data in this section reflect the services provided for veteran beneficiaries during F.Y. 1967 by the VA Prosthetics Center's Orthopedic Shoe Service, the Prosthetics-Orthotics Service, and the Restorations Service. Fiscal Year comparisons of the various related services are also included where available and meaningful.

A. The Orthopedic Shoe Service

Table 5 reflects full Fiscal Year comparisons of our national Orthopedic Shoe Program from 1963 through 1967. We experienced an increase of 264 beneficiaries for F.Y. 1967. In addition we provided 227 with ortho-inlay shoes. In the past we fitted these beneficiaries with custom orthopedic shoes but since their needs could be met by the fitting of ortho-inlay shoes we experienced an immediate savings of over $13,000. In addition to the figures reflected in Table 5, we issued 221 pairs of overshoes and rubbers.

It is interesting to note that while the total number of beneficiaries serviced by the Orthopedic Shoe Service has increased, the total numbers of shoes and repairs have decreased as compared with the previous Fiscal Year. We believe this can be attributed to the better quality of materials and workmanship and to instructions issued to our shoe wearers regarding care of their

TABLE 5.—*VAPC National Orthopedic Shoe Program*

	Fiscal year				
	1963	1964	1965	1966	1967
Beneficiaries on rolls	9,681	10,021	10,322	10,587	10,861
New shoes, prs.	7,249	7,317	7,336	7,688	7,181
Prs. of new shoes issued per benef. on rolls per yr.	.75	.73	.71	.73	.66
Repaired shoes, prs.	9,716	9,484	9,422	10,432	8,932
Prs. of shoes repaired per benef. on rolls per yr.	1.00+	.95	.91	.99	.82

shoes. Furthermore, a signficant number of shoe wearers have had their regular dress shoes replaced by work shoes which not only afforded them greater protection on the job, but are also longer wearing.

B. The Prosthetics-Orthotics Service

Table 6 reflects the activity in the Orthotic Components Unit of the Prefabricated Appliances Section relating to the distribution of surgical supports and elastic hosiery.

Tables 7 and 8 summarize distribution by the Prosthetic Components Unit of prosthetic and orthotic devices.

The Limb and Brace Section of the Prosthetics-Orthotics Service fabricated and delivered lower-extremity prostheses and braces as indicated in Tables 9, 10, and 11. Other items fitted were as follows: Arches, 1,547; Artificial Arms, 35.

C. The Restorations Service

Effective January 1, 1967 (when jurisdiction of the Regional Office-Outpatient Clinic was transferred to the VA Hospital, New York) the former Plastic Eye and Restorations Clinic of the VA Regional Office was integrated into the VA Prosthetics Center as the Restorations Service.

Table 12 reflects the activity of the Restorations Service in its major prosthetic responsibilities from July 1, 1966.

D. Special Clinic Team

The Special Clinic Team met 47 times during this Fiscal Year. We saw 109 veterans referred by 17 VA field stations. The foregoing are typical examples of cases referred to us.

CASE A

This patient, a World War II veteran, was referred by the VA Outpatient Clinic, Boston, Massachusetts, because he could not be satisfactorily fitted with a right Chopart-orthopedic shoe combination. He had good motion at the ankle and foot stump; dorsiflexion about 35 deg., eversion 30 deg., inversion 35 deg., plantar flexion 40 deg. His general complaint was that the incidence of breakdowns of his prosthesis was excessive. Our solution to this problem was to provide him with a molded, plastic insert with soft plastic shell with posterior opening to accommodate the foot stump allowing for free ankle motion; a hinge at the metatarsal head to relieve pressure at toe-off; an elastic strap on the prosthesis from the Achilles tendon area to the heel, extending forward to mid-foot, to prevent the stump from sliding out of the mold. Custom orthopedic shoes were also fabricated to accommodate the prosthesis. At the final follow-up, 60 days later, the patient indicated that he was quite satisfied with the prosthesis and shoe setup. A spare Chopart and custom orthopedic shoe setup, identical with the one previously delivered, was then prescribed.

TABLE 6.—*VAPC Surgical Support and Elastic Hosiery Program*

Issuance channel	F.Y. 1963		F.Y. 1964		F.Y. 1965		F.Y. 1966		F.Y. 1967	
	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, 162	5, 806	2, 434	9, 989	2, 482	9, 554	2, 634	9, 801	2, 697	10, 866
VA Orthopedic Shops	3, 098	6, 598	3, 995	12, 258	3, 722	11, 049	5, 885	14, 517	5, 056	15, 924
Totals	5, 260	12, 404	6, 429	22, 247	6, 204	20, 603	8, 519	24, 318	7, 753	26, 790

TABLE 7.—*Distribution of Prosthetic Components by VAPC, F.Y.* 1967

Issuance channel	Temporary prostheses [a]		SACH feet	Hydraulic systems [b]		
	AK	BK		Initial issue	Replacement issue	Active wearers
For specific beneficiaries	70	84	115	457	1073	
VA Orthopedic Shops	81	128	382	—	—	
Totals	151	212	497	457	1073	2081

[a] In F.Y. 1966 we distributed 80 above-knee temporary prostheses and 92 below-knee prostheses respectfully.

[b] In F.Y. 1966 we had 1,719 active wearers of hydraulic systems. Ninety-five wearers changed from one hydraulic system to another hydraulic system, or were issued spares.

TABLE 8.—*Distribution of Prefabricated Braces by VAPC, F.Y. 1967*

Items	Number
Cervical Collars	10
Knight Spinal	40
McAusland	60
Total	110

TABLE 9.—*Complete Below-Knee Artificial Limbs Fitted by VAPC, F.Y. 1967*

Type	Permanent	Temporary	Totals
PTB			
Cuff	29	20	49
Lacer	7	3	10
Carved wood	11	—	11
Molded socket non-PTB	58	10	68
Syme	9	—	9
Chopart	3	—	3
Totals	117	33	150

TABLE 10.—*Complete Above-Knee Artificial Limbs Fitted by VAPC, F.Y. 1967*

Type	Permanent	Temporary	Totals
Molded socket, non-total contact	23	50	73
Molded socket, total contact	51	4	55
Carved wood socket	10	2	12
Hip disarticulation	3	1	4
Knee bearing	9	0	9
Totals	96	57	153

TABLE 11.—*Braces Fitted by VAPC, F.Y. 1967*

Item	Number
Below knee	159
Above knee	96
Arm	19
Spinal, custom	100
Spinal, prefabricated but custom fitted	71
Cervical collars	58
All other	36
Total	539

CASE B

This patient was referred by the VA Hospital, Pittsburgh, Pennsylvania. He was 54 years old, about 6 ft. 1 in. and weighed about 190 lb. The patient had a crush injury of the left ankle in 1943 with ensuing pain and disability. He had been walking with crutches for the previous 12 months. An attempt at fusion of the subtalar joint, using two cortical plugs of bone across the subtalar was made about a year ago, prior to the referral. The ankle joint fused. Physical examination revealed a thickened ankle with discoloration and coolness of the skin about the ankle and the foot. However, a dorsalis pedis pulse was palpable and of good quality. The patient's subtalar joint was mobile. Motion was also present in the ankle joint although quite restricted. *Diagnosis:* Osteo-arthritis of the subtalar joint and some osteoarthritis of the ankle.

We fitted this patient with a patellar-tendon-bearing brace. A dramatic improvement in gait was effected and the patient was able to ambulate without the use of crutches. At a subsequent follow-up visit, the patient com-

TABLE 12.—*VAPC Production of Restoration Items, F.Y. 1967*

Item	Number
Artificial eyes	341
Body restorations	5
Cosmetic gloves	85
Ear inserts, hard [a]	1036
Ear inserts, soft [b]	256
Facial restorations, ear	18
Facial restorations, nose	13
Facial restorations, orbital	25
Facial restorations, other	2
Plastic hands, full	44
Plastic hands, partial	9
Tinting of gloves	4
Other items for patients	3
Repairs to appliances (all)	62
Other items or services	216

[a] The standard conventional type of ear insert for an average ear canal is generally made of hard acrylic plastic (methyl methacrylate). The material has the advantages of relative permanence, easy processability and good tolerance by the ear tissues. Its main disadvantage is its rigidity, in that it cannot adapt well to moving ear canal tissues during jaw movements and swallowing reflexes.

[b] Better adaptability to changing ear canal contours and improvement in "feedback" problems can usually be obtained by the use of one of many types of soft plastic materials, either on the mold's canal portion only, combined with hard acrylicplastic for the external part of the insert, or as an all-soft (solid or hollow-shell) ear insert for patients with very severe hearing losses. Thus the use of soft materials for such inserts will generally increase patient comfort and improve the efficiency of the hearing aid.

plained of swelling and pain in the ankle after wearing the brace for a few days. Further surgery was recommended in the form of an orthodox-type subtalar fusion, if the extent of infection would not contra-indicate it. High custom orthopedic shoes were also prescribed. At the time of his next follow-up, about 6 months later, the patient advised us that surgery was performed. X-rays showed a triple arthrodesis with complete bony fusion. The patient subsequently reported that he was able to walk without pain and without swelling and no longer needed crutches. At the time of his final checkout, 2 months later, he continued to express complete satisfaction with the new brace-shoe setup. A spare identical setup was then prescribed and delivered shortly thereafter.

CASE C

This patient was referred by the VA Hospital, Wilkes-Barre, Pennsylvania. He was injured by a combination of land mine and bullet wounds in the right thigh, damaging the sciatic and femoral nerves. As a result he devel-

oped a flail knee and ankle. The knee was fused at about 150 deg. and gave him no pain. The right ankle was the big problem at this time. The patient fell frequently and had on occasion fractured both hands and his nose. He has spastic motion of the dorsi and plantar flexor muscles. He has restriction of dorsiflexion to 80 deg. Plantar flexion is complete to approximately 130 deg. The patient stands in equinus due to the shortening and to the fixed flexion of the knee. All motions of the foot produce spasms and spastic activity of the muscles. We also detected wide areas of decreased sensitivity in the outer aspects of the leg, and also in the dorsal lateral aspect of the foot. The shoes the patient was wearing did not provide enough equinus on walking. The patient was also thrown into varus at the heel and did not experience sufficient stability of the ankle to prevent his turning into a varus position which resulted in his falling.

We prescribed and issued new high custom orthopedic shoes with higher equinus and with a higher lift in the heel so that when he stands the heel will strike the floor. The lateral side of the right shoe was also built up. The right shoe was a high lace type with a "T" strap and ½ in. heel buildup to accommodate a 2 in. shortening. A single-bar brace with a spring-loaded ankle joint was also prescribed to further stabilize the ankle. The patient is now able to stand with the crest of the ilium on an equal plane and walk quite comfortably, without falling.

E. Special Service for Vietnamese Wounded

Beginning with a report in BPR 10–5 Spring 1966, we have been presenting data about the VA Prosthetics Center's participation in the rehabilitation of 57 South Vietnamese soldiers who suffered injuries resulting in paraplegia, hemiplegia, and related disabilities. This is the final report related to this mission.

In late 1965, the VA Prosthetics Center was authorized to assist the VA Hospital, Castle Point, New York, where the Vietnamese were hospitalized, in a rehabilitation program for these men. The VA Prosthetics Center thereupon provided consultations regarding disability evaluation for orthotics prescriptions. Braces and/or other orthotic devices were fabricated or procured by the VA Prosthetics Center, then fitted at the hospital.

The project was originally launched on November 23, 1965, when a group of VAPC specialists, headed by Mr. Anthony Staros, VAPC Director, attended a clinic at the hospital. This constituted the first of a series of weekly visits by VAPC staff. By June 30, 1967, all except three of the Vietnamese were classed as fully rehabilitated as possible here in the United States. Appropriate orthotic devices were fitted to those cases who could benefit. These 54 men then returned home. The other three remained in the United States for vocational or educational rehabilitation.

Table 13 summarizes the diagnoses and brace fittings for all 57 subjects.

TABLE 13.—*The Vietnamese Wounded*

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
1	352.2 785.7 786.2 958.5	F	Orthopedic shoes
2	352.2 785.7 786.2 958.5	S	Left leg brace, single bar Right leg brace, single bar
3	352.2 785.7 786.2 958.3	F	Bilateral leg-thigh braces, double bar Spinal brace Orthopedic shoes
4	352.3 357.5 958.1	S	Left leg brace, double bar
5	352.2 785.7 786.2 958.3	F	Bilateral leg braces, double bar
6	352.2 785.7 786.2 958.5	F	Right leg brace, single bar Left leg brace, single bar
7	352.3 785.7 786.2 958.2	S	Orthopedic shoes
8	352.2 357.5 785.7 786.2 958.5	F	Left leg brace, single bar Right leg brace, single bar
9	352.2 785.7 786.2 806.3	F	Bilateral leg-thigh braces, double bar

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
10	352. 3 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar Orthopedic shoes
11	352. 2 368. 9 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
12	352. 2 368. 9 785. 7 786. 2	F	Bilateral leg-thigh braces, double bar
13	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
14	352. 2 785. 7 786. 2 958. 3	F	None
15	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
16	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
17	352. 1 781. 6 853. 1 951. 1	S	Right leg brace, double bar
18	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar, with pelvic joint Orthopedic shoes

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
19	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
20	352. 2 785. 7 786. 2 958. 5	F	Right leg brace, single bar Left leg brace, single bar
21	352. 3 357. 5 785. 7 786. 2 958. 1	S	Bilateral leg-thigh braces, double bar Orthopedic shoes
22	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar, with corset Orthopedic shoes
23	352. 1 853. 1	S	Left leg brace, double bar
24	352. 2 785. 7 786. 2 958 . 5	F	Left leg-thigh brace, double bar Right leg brace, single bar Orthopedic shoes
25	352. 2 357. 5 617. 9 785. 7 786. 2	F	Bilateral lig-thigh braces. double bar
26	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
27	352. 2 958. 5	F	Left leg-thigh brace, double bar Right leg brace, single bar Orthopedic shoes

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
28	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar, with spinal brace Orthopedic shoes
29	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
30	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar Orthopedic shoes
31	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar, with pelvic bands L.S. corset Orthopedic shoes
32	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar
33	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar, with pelvic joint Orthopedic shoes
34	352. 2 958. 5	F	Left leg-thigh brace, single bar Right leg brace, single bar
35	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar Orthopedic shoes
36	352. 2 357. 5 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
37	352. 2 785. 7 786. 2 958. 3	S	L.S. corset
38	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar
39	352. 1 853. 1 950. 1 950. 9 951. 1	S	Left leg brace, single bar
40	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar
41	010 352. 2 785. 7 786. 2 958. 2	S	Bilateral leg-thigh braces, double bar L.S. corset Orthopedic shoes
42	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
43	352. 2 357. 5 806. 5	F	Bilateral leg-thigh braces, double bar
44	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar
45	352. 2 357. 5 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
46	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar
47	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar L.S. corset
48	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
49	352. 2 357. 5 785. 7 786. 2	F	Left leg brace, single bar Right leg brace, single bar
50	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
51	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
52	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar Ischial
53	352. 2 785. 7 786. 2 958. 5	F	Orthopedic shoes
54	352. 2 785. 7 786. 2 958. 5	F	Bilateral leg-thigh braces, double bar

See footnotes at end of table.

TABLE 13.—*The Vietnamese Wounded*—Continued

Subject no.	Major ICDA codes [a]	S(spastic) F(flaccid)	Orthotic devices originally fitted
55	352. 2 785. 7 786. 2 958. 3	F	Bilateral leg-thigh braces, double bar
56	352. 2 785. 7 786. 2 958. 3	S	Bilateral leg-thigh braces, double bar
57	352. 3 785. 7 786. 2 958. 1	S	None

[a] International Classification of Diseases, Adapted (ICDA):

010 Tuberculosis of meninges and central nervous system
352 Other cerebral and spinal paralysis
- 352.1 Hemiplegia
- 352.2 Paraplegia
- 352.3 Quadriplegia (tetraplegia)

357 Other diseases of spinal cord
- 357.5 Compression, old or not specified as traumatic

368 Other diseases of peripheral nerves except autonomic
- 368.9 Other

617 Other diseases of male genital organs
- 617.9 Other

781 Other symptoms referable to nervous system and special senses
- 781.6 Other disturbance of speech and learning

785 Symptoms referable to abdomen and lower gastrointestinal tract
- 785.7 Incontinence of feces

786 Symptoms referable to genitourinary system
- 786.2 Incontinence of urine

806 Fracture and fracture dislocation of vertebral column with spinal cord lesion
- 806.3 Dorsal or thoracic, open
- 806.5 Lumbar, open

853 Cerebral laceration and contusion
- 853.1 With open intracranial wound

950 Injury to optic nerve(s)
- 950.1 With open wound
- 950.9 Late effect

951 Injury to other cranial nerve(s)
- 951.1 With open wound

958 Spinal cord lesion without evidence of spinal bone injury
- 958.1 Cervical with open wound
- 958.2 Dorsal or thoracic without mention of open wound
- 958.3 Dorsal or thoracic with open wound
- 958.5. Lumbar with open wound

Of the 57 subjects treated 49 had paraplegia (352.2) and except for three (subjects #27, 34, 43) this condition was accompanied by incontinence of feces (785.7) and urine (786.2).

Three (#17, 23, 39) had hemiplegia (352.1). One of these (#17) also had an accompanying speech and learning disorder (781.6).

Four (#4, 7, 10, 21) had tetraplegia (352.3) with incontinence of feces and urine, except for one (#4). One (#57) had quadriplegia (352.3) with incontinence of feces and urine.

Thirty-four of the paraplegics had flaccid paralysis. The remaining 23 subjects had spastic paralysis.

Of the 49 paraplegics, 36 were fitted with double-bar bilateral leg-thigh braces. Two of these (#3, 28) required spinal braces as well; three others (#18, 31, 33) needed pelvic joints or bands, and four (#22, 31, 41, 47) required lumbo-sacral belts or corsets.

Six of the remaining 13 paraplegics were fitted with bilateral leg braces; four (#8, 20, 34, 49) with bilateral single-bar braces, and two (#5, 6) with bilateral double-bar braces. Three others (#2, 24, 27) were fitted with double-bar leg-thigh braces on the left and single-bar braces on the right. Two (#1, 53) of the 13 required custom orthopedic shoes only, and another (#37) only a corset. One (#14) could not benefit from any type of orthosis.

Of the four who had tetraplegia, two (#10, 21) were fitted with bilateral leg-thigh braces, double bar, one (#4) was fitted with a left leg brace, double bar, and one (#7) was issued orthopedic shoes only.

Two of the three hemiplegics were fitted with unilateral leg braces, double bar, and the other, with a left single-bar leg brace.

The only quadriplegic in the group of 57 (#57) could not benefit from any type of orthosis.

Many minor modifications were made to the Vietnamese patients' braces and/or shoes subsequent to original fitting. The patients having flaccid paralysis were somewhat easier to fit and maintain. The spastic cases, especially those having severe contractures, were quite difficult. In some cases, as the intensity of the spasms subsided, appropriate corrections or modifications were made to the braces. It was necessary, for example, during the weekly visits to the Castle Point VA Hospital to realign braces, reshape bands, add "T" straps or modify shoes. Also, components such as spinal braces, hip joints and pelvic bands, and knee caps were removed as progress was made.

VAPC orthotists found it necessary after original prescriptions and fittings to make major modifications to 10 patients' braces as follows:

> *Subject #2:* Due to the severe spasticity of the ankle, it was necessary to convert the single-bar leg braces to bilateral-bar leg braces. It was also necessary to change the patient's stock shoes with orthopedic features to convalescent-type shoes because of the spasticity.

Subject #3: Drop-ring lock hip joints and a Knight spinal brace[b] were added to this patient's bilateral, double-bar, leg-thigh braces.

Subject #12: An anterior control belt was added to this patient's bilateral, double-bar, leg-thigh braces to counteract external hip rotation.

Subject #27: Because of the severe plantar foot deformity (pes cavus and callosities), it was necessary to change the patient's stock shoes with orthopedic features to custom orthopedic shoes.

Subject #30: Drop-ring lock hip joints and pelvic bands[b] were added to this patient's bilateral, double-bar, leg-thigh braces.

Subject #34: Due to severe spasticity of the knee and ankle, it was necessary to convert the left, single-bar, leg-thigh brace to a bilateral-bar, leg-thigh brace. However, the patient continued to use his right, single-bar, leg-thigh brace.

Subject #45: Original bilateral, double-bar, leg-thigh braces were modified to leg braces. However, bilateral hip joint twisters and pelvic bands were added to the leg braces to counteract severe external hip rotation.

Subject #47: A spreader bar was added to this patient's bilateral, double-bar, leg-thigh braces because of severe adduction (scissoring) contracture.

Subject #48: A spreader bar was added to this patient's bilateral, double-bar, leg-thigh braces because of severe adduction (scissoring) contracture.

Subject #54: Drop-ring lock hip joints and pelvic bands[b] were added to this patient's bilateral, double-bar, leg-thigh braces. A spreader bar was also added because of severe adduction (scissoring) contracture.

The 57 wounded soldiers were accompanied by a Vietnamese Army medical team comprised of two physicians and 12 paramedical personnel who were assigned to care for the wounded men under supervision of the Castle Point VA medical team. Thus the Vietnamese could learn the very latest techniques of rehabilitation medicine for possible application in their homeland.

VAPC staff conducted a series of lectures at the VA hospital for the Vietnamese Army medical group. This lecture series covered a variety of subjects including biomechanics, basic principles of prosthetics and orthotics, prescription and check-out of prostheses and orthoses, measurement and casting of stumps, construction and repair of limbs and braces, alignment techniques, performance evaluations, check-out and training techniques, construction of shoes and shoe modifications, and care and maintenance of braces and prostheses. This series of lectures and demonstrations culminated

[b] After Physical Therapy, crutch walking and parallel bar training, the patient's conditions improved and these components were removed.

in an all-day visit to the VAPC for inspection of facilities and a comprehensive review. Two medical corpsmen, Sgt. Tram Huu Loc and Sgt. Nguyen Dinh Lam were then assigned to the VAPC Limb and Brace Section on a full-time basis for a year's training, which ended on June 30, 1967. These men have since returned to Vietnam and are assisting in the operation of a military orthopedic shop.

Particularly difficult was the language problem, but in overcoming some of the communications barriers, we were able to establish Vietnamese terms for some of our brace components. Identified in Tables 14 and 15 are English and Vietnamese expressions for some common orthotic components. Credit is due Dr. Lam Van Thach of the Vietnamese Army medical team for the translations.

TABLE 14.—*Lower-Extremity Braces*

English	Vietnamese
Ischial weight-bearing leg-thigh brace	Giay tat chi duoi chiu nang xuong mong
Leg-thigh braces	Giay tat chi duoi
Knee cage	Khuon goi
Below-knee weight-bearing leg brace (PTB type socket)	Giay chiu sue nang duoi goi
Leg brace	Giay tat ong chan

TABLE 15.—*Lower-Extremity Brace Components*

English	Vietnamese
Pelvic band	Dai xuong chau
Pelvic belt	Nich xuong chau
Hip joint	Khop hang
Ischial ring	Khoen xuong mong
Ischial bearing quadrilateral socket	Nien tu dien chiu xuong mong
Thigh band	Thanh dui, dai dui
Upper thigh band	Dai dui tren
Thigh bar	Thanh dui
Ischial ring extension bar	Khoen xuong mong, thanh dui thang
Thigh cuff	Vien dui
Thigh cuff, Velcro closure	Vien dui, khoa Velcro
Thigh cuff, lacer type	Vien dui, loai mong
Lower thigh cuff and band	Vien va dai dui duoi
Lower thigh band	Dai dui duoi
Connecting bail	Vong noi
Lower thigh bar	Thanh dui duoi
Lever lock	Khoa khop

TABLE 15.—*Lower-Extremity Brace Components*—Continued

English	Vietnamese
Lever lock knee joint	Khoa khop goi
Hyperextension control knee joint	Phan kiem soat khop dui
Ring lock knee joint	Khoen khoa khop goi
Free motion knee joint	Khop goi tha long
Leg bar	Thanh chan
Music or spring wire leg bar	Thanh chan bang lo xo hoac giay cuoc
Below-knee weight-bearing socket	Nien chiu nang duoi goi
Upper, leg cuff	Vien chan tren
Buckle and billet	Khoen khoa va dinh vao
Calf cuff and band	Vien va dai bap chuoi
Molded leg cuff	Vien mau chan
Molded leg cuff, lacer type	Vien khuon chan, loai mong
Calf, cuff, Velcro closure	Dai bap chuoi va khoa Velcro
Flexible calf cuff	Vien bap chuoi co gian
Calf band	Thanh bap chuoi or dai bap chuoi
Molded leg-ankle foot support (plastic laminate)	Khuon nang chan—co chan—Ban chan (lop nhua)
Lower leg cuff	Dai chan duoi
Ankle joint	Khop goi
Stirrup	Vong chan yen or dai mong ngua
Stirrup w/detachable rectangular caliper	Dai mong ngua, voi bong chu nhat go duoc
Stirrup type	Kien vong mong ngua
Round caliper joint	Khop vong tron
Foot plate	Nem chan gia
Caliper plate	Vong phang
Sandal	Giep
Sandal, lacer type	Giep, loai mong
Spring load type ankle joint	Khop co chan loai co lo xo
Blucher type shoe	Kieu giay Blucher or Giay loai Blucher
Bal type shoe	Giay loai khieu vu
Spring wire coil	Cuon lo xo
Pin for spring wire coil	Chot cuon lo xo

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