VETERANS ADMINISTRATION PROSTHETICS CENTER
RESEARCH REPORT

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I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

*Preformed Thermoplastic Above-Knee Sockets.* A procedure is currently under study for fitting adjustable temporary above-knee sockets. We usually think of the socket for a brace or a limb as a custom-made component requiring careful design and construction to fit the stump shape and meet the functional requirements of the patient. However, detailed studies of above-knee stump dimensions and contours suggest that above-knee sockets
may be partially prefabricated as semi-standardized components in a modular system.

As a result of a study in which 500 stumps and sockets were measured, dimensional relationships were established for prefabricated above-knee sockets. As shown in Table 1, a fairly reliable ratio between proximal stump circumference and the critical anterior-posterior and medial-lateral dimensions was noted. Anterior-posterior dimensions are approximately one-fifth the circumferential measurement while medial-lateral dimensions are approximately one-third the circumference.

<table>
<thead>
<tr>
<th>Proximal peripheral circumference (inches)</th>
<th>AP dimensions (inches)</th>
<th>ML dimensions (inches)</th>
</tr>
</thead>
<tbody>
<tr>
<td>17</td>
<td>3½</td>
<td>5¾</td>
</tr>
<tr>
<td>18</td>
<td>3¾</td>
<td>6</td>
</tr>
<tr>
<td>19</td>
<td>3½</td>
<td>6½</td>
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<tr>
<td>20</td>
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<tr>
<td>21</td>
<td>4½</td>
<td>7</td>
</tr>
<tr>
<td>22</td>
<td>4½</td>
<td>7½</td>
</tr>
</tbody>
</table>

Note: These dimensions do not apply for very thin stumps. AP and ML dimensions must be measured between landmarks. Due to the absence of soft tissue, an almost triangular area exists between the trochanter, ischial tuberosity, and adductor longus tendon.

The four most frequently required sockets sizes were 18 in. through 21 in. in circumference at ischial level, accommodating over 75 percent of the sample group. On a descending frequency scale, socket sizes 17 in. and 22 in. in circumference accounted for approximately 8 percent of the sample group.

A fairly reliable relationship was also established between length of stump and taper or conical slope. This was established at \( \frac{3}{4} \) in. average taper for each 1 in. of stump length distal to the trochanter.

As regards stump length, over 60 percent of the stumps measured were between 8 in. and 10 in. long. Ten percent were 10 in. and 12 in. or more in length. Thirty percent were in the 6 in. to 8 in. length category.
A series of six left and right sockets in circumferences ranging from 22 in. to 16 in. in 1 in. increments measured at the ischial level seems to meet requirements for most adult males (Fig. 1).

To achieve total contact in a prefabricated socket is, of course, the principal problem once the dimensional standards are established. We also need the capability to make fine adjustments in socket fit to achieve total contact. Polysar X-414, a thermoplastic material previously described (BPR 10–10 Fall 1968, p. 259), permits adjustments to be made in order to improve the fit of the socket. Local socket modifications may be made by the use of a heat gun. Large areas of the socket may be modified by immersion in hot water.

Approximately 12 patients have been fitted by this method. Initial fitting trials are performed by pulling the patient's stump into a prefabricated socket of a size determined by stump measurements. We found it necessary to try several sockets to find the closest fit. Final adjustments to achieve total contact are made by heating and molding specific areas of the socket. More extensive modifications are required with excessively thin stumps. How-

Figure 1.—VAPC preformed thermoplastic above-knee sockets in both rights (upper row) and lefts (lower row) accommodate most requirements of adult male amputees.

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ever, in providing temporary sockets for new amputations this is seldom necessary since new stumps are rarely atrophied.

B. Development (Components)

*SACH Foot Modification.* During the past year we have been actively engaged in revising the standards for prosthetic feet in response to a number of questions which have arisen in relation to the size and shape of currently acceptable SACH feet. In BPR 10-10 (p. 256) we reported our work in cooperation with Professor Charles Radcliffe of the University of California at Berkeley aimed at improving both the function and utility of the SACH foot. This collaboration resulted in standardizing the cross section of the SACH foot through the ankle in three sizes, reshaping the heel to produce a narrower proximal section to permit compression of the heel wedge through the full range, modification of the radius of curvature in the arch area, rescaling of lengths and sizes, and specifying the height of the malleolus in relation to foot length.

Wooden models of appropriately modified feet were distributed to the manufacturers of SACH feet and other interested parties for examination and comment. A number of suggestions for minor variations were accepted and several manufacturers were asked to produce sample feet using temporary molds. These samples (Fig. 2) returned to us for evaluation, were in substantial compliance with functional specifications (Fig. 3 and 4). After testing, they were installed in prostheses which are being routinely worn by patients.

The new model features a convex sole in contrast to the flat sole of the older foot. Initial reactions from the field to this curvature were generally negative because the foot seemed to be unstable; it rocked when placed on a flat surface. Actually, however, this feature was designed to improve the stability of the foot when it is placed in the shoe. The soles of shoe lasts are also convex. The inner surface of the inner sole of shoes lasted upon these forms appears concave when viewed from above. The concavity mates with the convexity of the shoe last and, of course, with that of the new model SACH foot. Preliminary studies indicate that this shape provides a better

![Figure 2.—The redesigned SACH foot (top) compared with the standard designed SACH foot (bottom).](image-url)
distribution of weight on the sole of the foot. It may also increase foot life by reducing the deformation of the sole of the foot where the end of the heel bears on it.

A meeting of manufacturers and others has been scheduled to consider all changes prior to ordering permanent molds and beginning routine commercial manufacturing.
C. Development (Techniques)

None.

D. Evaluation (Components)

1. Blatchford Stabilized Knee. This device, a slightly different model of one tested earlier (BPR 10-4, p. 136) is a product of Charles A. Blatchford
& Sons, Ltd. of Lister Road, Basingstoke, Hants, England. It is designed to provide stance-phase stability by mechanical rather than by hydraulic resistance (Fig. 5). In the developer's view, mechanical systems are more easily manufactured and require less maintenance than hydraulic systems. Moreover, providing stance-phase stability may also permit the use of a softer dorsiflexion foot bumper and require less alignment stability. A softer dorsiflexion bumper, the developer points out, tends to reduce the "vault" or the magnitude of the vertical elevation of the body center of gravity, and to reduce fatigue.

The Blatchford Stabilized Knee relies on the tensions produced in a brake band drawn taut under the influence of the wearer's body weight (Fig. 6). Springs serve the function of releasing the stabilizing effect of the brake band as the applied body weight is reduced to permit knee flexion in swing phase. The body weight applied to the limb is shared between the release springs and the friction activating lever so that knee flexion is possible under very light loads, but as the load increases the effect of the springs is overcome and the knee becomes increasingly stabilized.

As shown in Figure 7, the knee is capable of resisting significant eccentric loads. With the knee flexed 5 deg., 7.5 deg., and 10 deg., the unit was loaded
with 200 lb. in 20 lb. increments. At 5 deg. of flexion the knee withstood a load of 40 lb. with no rotation, but as the load increased from 40 to 200 lb. the knee yielded in small increments to a maximum of approximately 3 deg. The pattern was quite similar when the knee was loaded in 7.5 deg. of initial flexion and in 10 deg. of initial flexion. This resistance pattern may make it possible for patients to apply significant loads to the bent knee as might occur in stumbling with less risk of collapse.

In addition to its stance-phase functions, the unit also controls swing phase by means of mechanical friction. Shown in Figure 8 are the swing-phase resistance characteristics of the unit as indicated by drop times.

As with most mechanical friction knee units, the useful adjustment range is limited to less than 3/16 of one turn of the friction adjustment wheel. The location of the range depends on whether or not the stability mechanism is engaged.

**Figure 6.**—Internal mechanism of the Blatchford Stabilized Knee (BSK).

**Figure 7.**—VAPC stability test results of the Blatchford BSK Knee.
With both release springs tightened all the way in, the stabilizing mechanism is inactive and the useful swing-resistance range is located between the 8/16 and the 10/16 turn. With both springs backed all the way out, the stabilizing mechanism is engaged locating the useful range at the beginning of the scale between 2/16 and 4/16 turn.

Two patients have been wearing this device for periods of 3 to 6 months. The reactions of both are quite positive with indications that they perceive increased security in walking up and down stairs and inclines.

Laboratory studies are in progress to compare the stance-phase stability characteristics of this unit with other stance-control units in walking on inclines.

2. Hosmer Pneumatic Swing Control Knee Unit. The Hosmer Corporation of Campbell, California, submitted one model of a pneumatic above-knee swing-control knee designed along the lines of the UC–BL Pneumatic Knee (BPR 10-6, p. 234), but which utilizes a modified Dupaco cylinder (Fig. 9). The unit is adjusted by means of two valves—the flexion valve and the extension valve. The extension valve corresponds to the level valve of the UC–BL Pneumatic Unit in that it alone is capable of adjusting both flexion and extension resistance. The function of the flexion valve is similar to that of the ratio valve of the UC–BL unit (Fig. 10). When resistance to knee extension is properly adjusted by means of the extension valve, higher resistances to knee flexion may be required to control heel rise. To make this adjustment, the resistance to both extension and flexion is increased by closing the extension valve. Then the resistance to extension only is reduced to the previously achieved desirable level by opening the flexion valve.
FIGURE 9.—The Hosmer Pneumatic Swing Control Unit is designed after the UC-BL Pneumatic Knee Unit.

The pneumatic unit will resist flexion (or cause extension) as long as high pressure air is below piston.

FIGURE 10.—A schematic diagram showing the similarity between the Hosmer and the UC-BL Pneumatic Knee Units.

Both valves are slotted for operation by means of a screw driver or similar tool.

The resistance characteristics of this unit are also quite similar to those of the UC-BL as shown in Figure 11. Both units are adjustable to a peak resistance of approximately 40 in.-lb. The Hosmer Pneumatic Unit exhibits the same low order resistance characteristics as other pneumatic units, including the UC-BL. Limited service testing is in progress to determine clinical applications.
3. Blatchford Pneumatic Swing Control Knee. A product of Charles A. Blatchford & Sons, Ltd., Hants, England, this unit is similar in several general respects to the UC-BL and the Hosmer Pneumatic Unit mechanisms (Fig. 12). It differs however in that it is designed for easy adjustment of resistance by the patient himself by means of two knurled knobs which control the valves governing the resistance to both extension and flexion. A special feature is the ratchet arrangement which produces a "click" when the knurled knobs are turned approximately 60 deg., providing six adjustment reference points per turn of the control knobs. The unit is well designed, featuring
highly effective needle valves with high quality seals. No internal or external leakage occurred during our testing program. The Blatchford Pneumatic Knee is capable of furnishing somewhat higher flexion and extension resistance values than either the UC-BL or the Hosmer units providing up to a peak resistance of 60 in.-lb. (Fig. 13).

Testing indicated that the ratchet arrangement loosens and tends to allow the ratchet to slip from a desired setting. This is a relatively minor problem which can easily be overcome by the manufacturer. As a European product, the pin diameters and threads are sized in the metric system. On a subjective basis, the unit is slightly noisier than the UC-BL system. Several models have been fitted for limited clinical evaluation.

4. Yugoslavian Hydraulic AK Knee Mechanism. Designed on the pattern of the Hydra-Knee, a product of the U.S. Manufacturing Co. of Glendale, Calif., this device (Fig. 14) is produced in Belgrade, Yugoslavia, by the manufacturing enterprise “Rudo.” Although not intended for distribution in this country, this unit exhibits somewhat better adjustment characteristics than the Hydra-Knee in addition to several similar deficiencies. The useful range of resistance adjustment is quite limited; it occurs between two and three full turns. Swing resistance curves (Fig. 15) reflect a pattern similar to that of the Hydra-Knee with respect to the great increases in resistance to flexion and extension between intermediate and maximum resistance settings.

5. Lyquist Through-Knee Prosthesis. Above-knee prostheses are usually fitted with the axis of knee rotation (knee-bolt) positioned at the approximate location of the anatomical knee center. The patient with the longer stump resulting from through-knee, transcondylar amputation cannot readily
be fitted with the more sophisticated fluid mechanisms. The long stump precludes the use of the knee controls which would be of special advantage to these patients. A partial solution to this prosthetic problem has been advanced by Erik Lyquist of Copenhagen, Denmark. He has designed a knee mechanism which is basically a polycentric four-bar linkage (Fig. 16). Since the center of rotation moves in a path well above the knee mechanism and in the area of the anatomical knee center, the shank rotates more normally in swing phase. Although the polycentric knee-control system permits flexion and extension, it does not now contribute effectively to the control
of the swing phase by reason of a relatively inadequate friction resistance system. The developer indicates that he is planning to introduce some kind of a pneumatic control system.

6. Wagner No. 320 Conventional Knee-Shin Assembly. One model of this mechanical friction knee unit was tested to determine functional adequacy, installation requirements, and general appearance (Fig. 17). The device features a built-in extension aid and a single-axis knee bolt without the usual opening on the anterior knee surface. Constructed of poplar wood, it is available in rights and lefts and in calf circumferences from 32 to 40 centimeters.
in 2-centimeter increments. It is readily used with most currently available foot/ankle assemblies. Above-knee amputations at least 1½ in. above knee center can be accommodated. All the available alignment devices can be used.

The friction system consists of a split nylon brake positioned about the knee bolt which is adjusted by a single anterior screw. It provides the classic, rela-
tively low resistance pattern characteristic of mechanical friction units (Fig. 18). Maximum extension is limited by a plated steel bar fastened to the posterior wall of the shank by two machine screws. Extension bias is provided by an adjustable spring housed within the shank. The spring is connected to a cross pin within the slotted posterior section of the knee by means of two nylon tipped metal straps. Tests show that the usable range of friction adjustment is only \( \frac{3}{8} \) of one turn. Other tests indicated that the unit provides only 83 deg. of knee flexion, a deficiency since minimum standards require 110 deg. of knee flexion.
Conventional methods of socket and foot attachment, alignment, shaping, and finishing are entirely adequate, and no problems were encountered in the installation of these units.

7. Wagner No. 209 Stereo Safety Knee. Designed as a "safety knee" along lines similar to that of the well known Bock 3P24, this device (Fig. 19) provides control of swing phase by means of mechanical friction. It does not include an internal extension aid. Made of poplar wood it is available in rights and lefts and in calf sizes between 32 and 34 centimeters inclusive. It is compatible with most foot/ankle combinations and accommodates above-knee stumps which are at least 1 1/2 in. above knee center.

Extension bias is furnished by an externally mounted control. Swing phase is controlled by means of a split nylon floating-type brake mounted on the knee bolt and adjusted by means of a single posterior screw. A control disk mounted behind the knee permits a patient to adjust the resistance by changing the position of the floating brake.

Stance-phase control is accomplished by means of an eccentrically contoured knee surface which produces variable resistance of the low order characteristic of mechanical friction devices (Fig. 20). Under a load the entire knee assembly descends on a lever forcing the leather lined, double grooved distal knee surface to mate with the hardwood section of the proximal shank. The unit features an internal extension stop consisting of a plated steel bar fastened to the posterior wall of the shank. Two felt bumpers set within the double grooved channels also limit extension when they come in contact with hardwood stops at the proximal end of the shank.

This unit only provides 85 deg. of knee flexion which is significantly below the minimum requirement of 110 deg. It may be installed, aligned, shaped, and finished by conventional methods.

Figure 19.—Wagner 209 Stereo Safety Knee features knee stability during stance phase.
Kentucky, the Orange foot is designed as a laminated neuter blank, of crepe rubber (Fig. 21). The designer provides blanks with high heel wedges for above-knee amputees and low heel wedges for below knee’s. An annealed spring steel shank is attached to, and extends forward of, the hardwood keel (Fig. 22). Between the end of the keel and the spring steel extension is a layer of nylon webbing and a layer of Teflon to reduce friction.

Two models of the Orange foot are currently being tested against the existing SACH foot specifications. Clinical trials are being conducted.

8. Orange SACH Feet. Developed by John L. Orange of Russellville, Kentucky, the Orange foot is designed as a laminated neuter blank, of crepe rubber (Fig. 21). The designer provides blanks with high heel wedges for above-knee amputees and low heel wedges for below knee’s. An annealed spring steel shank is attached to, and extends forward of, the hardwood keel (Fig. 22). Between the end of the keel and the spring steel extension is a layer of nylon webbing and a layer of Teflon to reduce friction.

Two models of the Orange foot are currently being tested against the existing SACH foot specifications. Clinical trials are being conducted.
E. Evaluation (Techniques)

None.

II. UPPER-EXTREMITY PROSTHETICS

A. Development

*VAPE External-Power Program*. During the past 5 years a great deal of effort has been applied to the design and development of externally powered components for upper-extremity protheses. A relatively large number of individual researchers and laboratories in this country and abroad are engaged in developing powered hands, elbows, and to a lesser extent, wrists.
and feeder devices. During the past year, the Committee on Prosthetics Research and Development has convened several workshops and seminars for the purpose of determining the current state of development in this field. As a result, it is now clear that most of the work done to date has been the product of individuals who have approached these problems in a narrow, nonsystematic way—they have concentrated their efforts on individual components such as a hand, a wrist, or an elbow rather than on an integrated prosthetic system. Notable exceptions are, of course, the AIPR system and some of the feeder devices of MacLaurin at Ontario Crippled Children's Centre and Allen and Karchak of Rancho Los Amigos.

Despite design differences among the eight or 10 externally powered elbows and hands currently under development, they all seem to have one overriding similarity. Most are built on the physical model (size, shape, appearance) of the conventional hand or elbow they are designed to replace. The mechanical cable and lever/spring-control system has been replaced by either an electrical or pneumatic system and, in some instances, EMG control is featured.

Conventional artificial hands are generally considered less functional but more cosmetic than conventional hooks. Apart from the obvious advantage of appearance, conventional hands are conceded to be too heavy, too bulky to insert in such confined spaces as pockets, and of marginal value in grasping small, smooth, or irregularly shaped objects. Except for the voluntary closing types they generally lack sufficient pinch force. The appearance of powered hands is similar to that of the conventional hand but with few exceptions they provide sufficient pinch force. The other deficiencies in conventional hands, e.g., bulk, weight, and difficulty in manipulating certain objects, also characterize the powered hand. Furthermore, the question of patient utilization of powered components has not been adequately considered.

It is difficult to particularize strong and objective prescription indications for fitting the unilateral below-elbow amputee with a powered hand. A stronger case can be made for prescribing powered hands for the above-elbow and shoulder-disarticulation patients. In these instances, however, the question of the complete prostheses, i.e., the entire system, must be considered in a new light. To get the best results it may be necessary to consider the harness, control system, and the other components.

The above-elbow or shoulder-disarticulation patient may of course operate a powered terminal device by means of a switch connected to his conventional system. Controlling the hand in such a system by myoelectric signals (EMG) may be difficult since available muscle control sites are relatively remote from the hand in terms of both distance and function.

\*Report of Seventh Panel Meeting on Design and Development of Upper Extremity Components of the Committee on Prosthetics Research and Development.
Still unresolved is the optimal method of controlling two powered components in a single system. With more powered components available, it is essential to consider the problems of integrating one or more powered components in prosthetic systems. We have recognized this state of affairs by altering our externally powered component development program to include simultaneous efforts applied to a. elbows, b. terminal devices, c. wrists, and d. control systems.

a. The VAPC electric elbow (Fig. 23) described in BPR 10-10, p. 264, is essentially the same size and approximately the same weight as the Hosermer elbow. This is an advantage which simplifies the direct replacement of a conventional unit by the electric elbow. It fits both the standard forearm and elbow turntable. The device is powered by a small permanent magnet electric motor. The motor shaft is coupled directly to a planetary roller reduction harmonic drive wave generator. This wave generator forces the flexible spine of the harmonic drive to engage the rigid spine. The flexible spine is fixed to the square output tang on the outside of the elbow. The rigid spine, which is also the elbow housing, acts as a reaction point in the gear reduction process. The high motor speed, 980 r.p.m. is geared down by approximately 12 to 1. The harmonic drive achieves a ratio of exactly 80 to 1. The limits of flexion and extension, in the elbow are controlled by two microswitches and four diodes. Attempts to extend or flex the elbow past these limits shut down the power. Two patients have been wearing these units for periods of 2 to 3 months (Fig. 24). Their reactions to date are quite positive.

Figure 23.—a. VAPC Electric Elbow, b. internal mechanism of the VAPC Electric Elbow.
b. *The VAPC hand* (Fig. 25) is the same size and shape as the Viennatone hand. As are the Viennatone, INAIL, and the Bock units, it is constructed on a skeletal framework with a PVC inner shell and a cosmetic glove over the outside. The special features of the VAPC hand are its safety breakaway that permits the hand to open mechanically when sub-
jected to a load of greater than 40 lb., as for example, if a man grasped a handle on a moving vehicle. In addition, the small and efficient motor used insures compatibility with the VAFC elbow.

c. The VAFC wrist is still in the prototype stage. It is being designed for either independent use as a powered wrist, or for use in a coordinated system with the VAFC elbow or any other commercially available elbow.

d. The VAFC control system previously reported in BPR 10-9, p. 138; 10-10, p. 262; and 10-11, p. 278 consists essentially of one or more multi-position microswitches (Fig. 26) that may be inserted into the control attachment strap (CAS) of the below-elbow or above-elbow Figure 8 harness. They may be attached in series with any section of the harness normally used to transmit forces and other information. For example, in a typical above-elbow harness the VAFC elbow control switch would be located in line with the elbow control strap (ECS). The same motion formerly used to lock and unlock the conventional elbow provides full control of the position of the VAFC (or other) powered elbow. Cable excurs and force to operate the elbow are thereby reduced facilitating terminal device control for the above-elbow patient. If a patient were fitted with a powered terminal device in addition to the elbow, a second identical switch is installed in the CAS permitting the patient to operate the terminal

![Figure 26.—VAFC upper-extremity control switch showing two microswitches (upper left), complete unit (upper right), and the unit installed in a conventional AE harness.](image)
device by the same motion he previously used to operate the terminal device, but with less force and excursion.

It is possible to locate two control switches in series with one another. By means of rubberbands, which increase the force required to operate, one switch can be discriminated from the other by the patient on the basis of the different force requirements. These switches are fairly simple to fabricate and have proved quite durable.

e. The VAPC hand/hook, still in a design stage, is an attempt to utilize the same power and transmission system to drive either a functional hand or hook interchangeably. The hook now under consideration is a three-fingered variety similar in size to the skeletal structure used in the Bock series of hands. This approach is an attempt to optimize the best features of hooks and hands with the advantages of external power.

B. Evaluation (Components)

1. Externally Powered Hands. The clinical phase of our evaluation (BPR 10–8, p. 227; 10–9, p. 142; 10–10, p. 270) of commercially available externally powered hands has been completed. During this study, three different EMG-controlled hands (the Canadian RIM, the Italian INAIL, and the Viennatone) were fitted to four below-elbow amputee subjects who were conventional hand wearers. Our purpose was to obtain clinical experience on the application of externally powered devices and to compare their performance with that of conventional terminal devices. No attempt was made to compare the performances of subjects with different powered terminal devices.

As an extension of this limited program, externally powered hands of several types are now being fitted to veteran beneficiaries through our clinics here in the VA Prosthetics Center, New York, and at the VA Center, Sawtelle Blvd., Los Angeles. Clinical feedback from wider experience will be obtained.

2. Steeper Split Hook No. 65. The problem of reducing the force and/or excursion requirements for operating conventional hooks has been considered with varying emphasis for many years. Among the many solutions advanced were pulley systems to increase excursion at the cost of increased force, and shorter operating levers (thumbs) on Dorrance-type hooks for the purpose of reducing excursion at the cost of increased force.

The Steeper Split Hook No. 65 (Fig. 27) is designed to reduce operating forces at very small increases in excursion by means of more advantageous pulling angles due to the shape of the operating lever. It is constructed of aluminum with canted fingers on the model of the Dorrance 5XA hook. Instead of the conventionally used ball bearing race, this device pivots in a relatively large brass bushing on a stainless steel shaft. It has two unique features: the operating lever (thumb) is curved so that the point of cable attachment is distal to that of the Dorrance hook; the neoprene finger linings are bonded
to the inner surfaces of the hook fingers and are designed for convenient replacement at a local shop or, under certain circumstances, by the patient himself.

The unconventionally shaped operating lever provides increasing mechanical advantage as the hook is opened through the first 3 in. of the total opening range. In contrast, the operating lever of the Dorrance-type hook requires slightly higher forces to open through the entire range between ½ in. and 3 in. of opening.

To compare the force/excursion characteristics of the Steeper Hook with a similar conventional hook, both units were tested under identical conditions to determine the force, finger displacement, and cable excursion characteristics.

Shown in Figure 28 are the forces required to open the hooks in relation to the amount of opening obtained. It may be seen that for openings up to approximately 2 in., the Steeper hook requires, on average, approximately 1 lb. less force. Between 2 and 3½ in. of opening, the Steeper device requires approximately 2 lb. less force. Shown in Figure 29 is the relationship between cable travel and hook opening. For openings up to 1½ in. the Steeper hook requires approximately 1/16 in. more travel than the Dorrance hook.
For opening ranges between 1½ and 3½ in., the Steeper device requires up to 3/16 in. more cable travel than the Dorrance. These data demonstrate that the Steeper hook generally requires between 1 and 2 lb. less force for comparable finger openings than the Dorrance hook. This advantage is obtained at the cost of an average increase of approximately 1/8 of an inch in cable excursion.

From a functional standpoint, therefore, this device might be considered advantageous for patients with limited force, particularly those who require high pinch forces. Moreover, it may be particularly advantageous where excursion is not a problem but force is. Further testing of this device is in progress to determine the durability of the replaceable inserts and the resistance to wear of the pivot bushing. A third factor under investigation is the utility of the curved operating lever for handling such implements as knives, forks, and pencils.

3. Direct Forming of Below-Elbow Sockets. The recently developed technique for forming below-elbow sockets of Polysar X-414 (BPR 10-10, p. 270) is being increasingly applied in various centers across the country. An independent evaluation (1) of this technique conducted by the Prosthetics and Orthotics Project, New York University, has, in general, corroborated our own experiences in fitting sockets of this type under laboratory conditions. The NYU study conducted under the general supervision of Sidney Fishman, Ph. D., Project Director, indicated that the instruction manual (BPR 10-11, p. 83, by T. Pirrello and G. Labate) was found to be a comprehensive and clear delineation of the step-by-step procedures. Fabrication time in fitting the first four patients was given as one-half day for each arm including all the procedures beginning with preliminary measurements and ending with final delivery to the patient. This is approxi-

* Registered trademark of the Polymer Corp., Ltd.
mateone-half the time required to complete a fitting with a polyester laminate socket. Halving the fabrication time was possible because of the smaller number of fabrication steps required with the thermoplastic socket.

The soft external covering was also considered a positive feature. The NYU report recommended the use of a denser foam filler covering to avoid collapse under a tight external sleeve and the use of cosmetic covers of thicker, dilaminar material. Although the staff was inclined to believe that the sockets were somewhat too heavy, this matter was never noted by the patients. With improvements recommended by the NYU group we believe that the use of directly formed below-elbow sockets represents a significant improvement in upper-extremity prosthetic technology.

4. Upper-Extremity Control Patterns. An experimental program has been undertaken to compare several different control methods of operating an externally powered above-elbow system (elbow and terminal device). The initial phase was limited to fine body control motions, i.e., motions of small excursions and low force. Two VAPC control switches (BPR 10–9, p. 138; 10–10, p. 262; 10–11, p. 278), one mounted in the control attachment strap (CAS) to operate a terminal device and one mounted in the front support strap (FSS) to operate the elbow, served as one control system. A single four-function switch mounted in the CAS was used as the alternative control system. The first position in the four-function switch opened the hand, the second closed the hand, the third raised the forearm, and the fourth lowered the forearm. The total excursion required to operate this switch was $\frac{1}{8}$ in. The distances between positions ranged between $\frac{3}{8}$ in. and $\frac{1}{2}$ in.

For purposes of comparing the relative ease of learning with each control system, two normal subjects were fitted with conventional Figure 8 above-elbow harnesses. The control attachment strap (CAS) and the front support strap (FSS) were attached to a cuff fastened around the subject's arm (Fig. 30). A visual display, consisting of an externally powered prosthetic elbow and an externally powered hand mounted on a mannequin, was operated by means of the microswitches mounted on the normal subjects' harnesses. These subjects operated the externally powered components mounted on the mannequin by the identical control motions that above-elbow amputees use. Prior to testing, each subject was given a practice session of 30-minutes duration and then tested on his ability to control the two externally powered components. Cards were shown to the subjects in random order with the following commands printed on them: 1. extend elbow fully, 2. open terminal device half-way, 3. flex elbow 30 deg., 4. flex elbow 135 deg., 5. flex elbow 90 deg., 6. open terminal device all the way, and 7. close terminal device. The subjects executed each command as rapidly as possible. An observer recorded the number of errors. Four trials constituted one test session.
As shown in Figure 31, performance with the two-function switch system was superior. The two switches mounted on the front support strap and the other on the control attachment strap were easily controlled by the subject and offered no particular problem except concentration on the part of the subject.

With the four-position switch, the first and second function (opening and closing the hand) as well as the third and fourth function were accomplished fairly well. However, when the subject had to lower the elbow from the 90 deg. or from the 30 deg. position, the elbow would first flex and then extend when passing through the third position to find the fourth switch position.

Other studies of a similar nature are being conducted to determine optimal control methods for various prosthetic configurations employing externally powered components.
III. LOWER-EXTREMITY ORTHOTICS

A. Development

1. Functional Knee Brace. A well-designed functional knee brace should provide a patient with reliable stance-phase stability without undue effort on his part. In addition, it should permit a relatively normal knee flexion pattern during the swing phase of gait. Despite extensive development efforts, there is at the present time no adequate solution to the problem.

Some developers have employed offset knee joints (Fig. 32) in an attempt to maintain significant extension moments during stance phase in order to eliminate the need for a positive lock and to permit the knee to flex and extend in swing phase. These efforts have not been satisfactory because it has been impossible to generate extension moments of sufficient magnitude to permit a reasonably symmetrical stride length without ex-
cessive offsets which are not attractive looking and interfere with the initiation of swing phase.

Other designs have employed offset knee joints in combination with pneumatic or hydraulic cylinders to increase resistance to dorsiflexion or to plantar flexion in an effort to increase the extension moments about the knee in stance (Fig. 33). These devices do not function satisfactorily in early stance phase; to avoid the flexion moment about the knee, the patient is required to shorten his step or flex at the hip in order to produce a countermoment. An even more complex system includes the offset knee joint, the hydraulic dorsiflexion control, and a socket designed to generate an extension moment about the hip and knee (Fig. 34). There is some question as to the effectiveness of the socket in generating the extension moment and in any case, patients generally walk with shorter step lengths on the brace side or with the hip in an attitude of flexion. Another brace design increases the extension moment about the knee by bracing the ankle in a fixed plantar flexion attitude (Fig. 35). Again the patient is required to take short asymmetrical steps.

More exotic designs have utilized myoelectric outputs from the gastrocnemius to lock and unlock a solenoid driven knee lock while others have used electrical contact switches placed in the heel of a shoe for the same purpose.*

* Developed by personnel at Moss Rehabilitation Hospital, Philadelphia, Pa.
All of these devices solve the basic problem to some extent in level walking; at normal speeds several of them permit a patient to flex and extend his knee in swing phase and provide reliable stance control. None of them is adequate when the patient sits and stands, when he walks on uneven terrain, up and down hills, or up and down stairs.

It is possible to control both swing and stance in walking on level ground by any of several methods. Offset knee joints and dorsiflexion limiters act to control the direction of the ground reaction vector to prevent the development of a flexion moment about the knee. Mechanical knee locks can be operated and switches controlled by body weight, EMG signals, or by ankle position. These control methods are of varying orders of complexity,
reliability, and cost. Each can probably be designed to provide dynamic knee control in both swing and stance. The heart of the matter is which if any is the simplest and most likely to provide adequate control not only on level ground but also on uneven terrain, slopes, and stairs.

An analysis of the functions required indicates that control of the knee by ankle position promises to yield the best results. A latching device about the knee controlled by the position of the ankle is a simple mechanical arrangement which need not be expensive, cumbersome, or unattractive. The flexion moment about the knee in stance phase is highly coordinated with ankle position; the onset of swing phase is also correlated with ankle position. Downhill walking and certainly uphill walking require highly coordinated knee moment/ankle position relationships. However, just what these relationships might be in stair ascent and descent is not clear. For this reason we have undertaken a study of the temporal, kinematic, and kinetic aspects of coordinated knee/ankle motion in level walking and under other conditions.

Shown in Figure 36 is a conventional leg-thigh brace modified by the inclusion of a latch to “lock and unlock” the knee. The latch is controlled by means of a cable connected to an outrigger on the stirrup. Angular position of the outrigger locks and unlocks the knee. By varying the position of the outrigger, the position of the angle at which the knee can lock and unlock, can also be varied. Studies are in progress to determine the range of ankle positions required to lock and unlock the knee at different cadences and stride lengths, and in walking on various surfaces. The purpose of these studies is to test the proposition that knee stability and motion requirements are sufficiently correlated with ankle position to warrant the design of a functional knee brace based on these relationships.
2. Foot Appliances. Increasing numbers of patients who cannot be adequately fitted with ordinary stock shoes are turning to Formo-Ped Ortho-Inlay shoes. These shoes are designed to accommodate a heavy innersole or inlay beneath the foot. The shoe and the inlay are supplied together (Fig. 37). The inlay is approximately ½ in. thick and can be modified sufficiently to relieve plantar surface problems and accommodate hammer toes. The shoes are fabricated over stock lasts which reduce their cost to a fraction of the fabrication cost of custom orthopedic shoes. In addition to the cost savings, the usual waiting period of 1 month, which is customarily required for orthopedic shoe fabrication, is reduced to 1 day for those patients who can be fitted with ortho-inlay shoes.

Methods are being investigated to further expand the use of ortho-inlay shoes to include the treatment of more serious plantar problems such as ulcers caused by concentrated loads in plantar areas and other problems commonly found with foot malformations and surgical stumps. By replacing the shoe inlay with a moldable thermoplastic material, we were able to explore the feasibility of “reading” the pressure pattern which the plantar surface of the foot imprinted while walking upon the surface of the softened material (Fig. 38).

The best material for this purpose was found to be a synthetic similar in appearance and physical properties to precipitated balata known by its tradename of Polysar™ X-414. This thermoplastic material in sheet form, is cut to the dimensions of the innersole of the shoe, heated to approximately 175 deg. F., inserted in the shoe, and walked on by a patient. The pressure exerted by his foot forms the Polysar™ into a configuration in which high pressure areas form depressions on its upper surface permitting low pressure areas to assume more of the body weight. The result is a reasonably useful “arch support” whose fabrication time is very short and which does not require extensive shop facilities. However, this technique produces a device which is useful only for cases with mild discomfort such as calluses

* Registered trademark of the Polymer Corporation, Ltd.
and the like. For more severe cases involving painful excrescences and sub-talar imbalance causing excessively high pressures and pain, a capacity for absorbing higher loads is required. This is typically true for the majority of cases we see with ulcers, bone damage due to gunshot wounds, and partial amputations of the foot.

The most efficient load-absorbing material we know, except for human fat, is Spence-gel, an expensive material whose physical properties are quite similar to human tissue particularly as regards its specific gravity of one. For more severe cases, we try to combine the benefits of dynamic molding with those to be obtained with a good load absorbing material.

The two most promising approaches to this plantar relief problem entail the: 1. placement of the “gel” pads in the ortho-inlay innersole (Fig. 39) and 2. the placement of the “gel” pads in a thermoplastic innersole in the pressure areas defined by walking on the softened thermoplastic material. The cushioning pads were made of thin rubber compartments into which Spence-gel was sealed. The size and shape of the compartments matched the cutout spaces in the arch supports or inlays. A thin sheeting bonded over the surface of the inlay and compartments consolidated the plantar surface.
B. Evaluation (Components)

*VAPC Modular Single-Bar Brace.* In previous issues of the Bulletin we have described our efforts in designing and developing a single-bar (unilateral upright) modular brace system whose components are interchangeable and serve to accommodate numerous disabilities of the ankle, knee, and hip (BPR 10–8, p. 273; 10–9, p. 152). Many potential advantages of the single-bar brace construction have been cited:

a. 25 percent lighter in overall weight,
b. reduced bulkiness,
c. increased comfort,
d. closer approximation of anatomical joints, and
e. easier to accommodate foot and ankle deviations.

In limited clinical studies patients who wore the leg and leg-thigh VAPC Single-Bar Braces seemed to walk with improved balance, with less fatigue, and for longer duration.

In addition to the advantages mentioned above, the VAPC Modular Single-Bar Brace system permits versatility in prescription. The ankle joint can be modified by means of a simple telescoping system, consisting of a coil spring and a round upright bar located above the mechanical ankle joint to permit axial displacement and transverse rotation. Similarly, the hip joint can be modified to control internal or external hip rotation by means of a coil spring.

The VAPC Modular Single-Bar Brace system has been manufactured for us by the Pope Brace Co. for use in a more extensive clinical field evaluation.
These components together with a VAPC fitting manual especially prepared for this purpose will be supplied to all VA Orthopedic Shops, and to certain orthopedic facilities of participating members of the CPRD Workshop Panel on Lower Extremity Orthotics. In this evaluation, clinic teams at each station will develop prescriptions for single-bar braces. A certified orthotist will fabricate the brace adhering closely to the details in the VAPC Single-Bar Brace manual.

C. Evaluation (Techniques)
None.

IV. MISCELLANEOUS AIDS FOR THE DISABLED

A. Development
None.

B. Evaluation (Components)
1. Eaton E-Z Bath Lift. An earlier model of the Eaton E-Z Bath Lift was previously evaluated by this Center (BPR 10-11, p. 294) and several deficiencies were brought to the attention of the manufacturer. A second model (Fig. 40) with an improved method for attaching the worm gear to the drive shaft was submitted for reevaluation. In addition a small arrow was stamped on the crank arm to indicate the proper direction for turning up or down.

The test results indicated that the modified Eaton E-Z Bath Lift functions satisfactorily in lowering and raising a patient into and up from a bathtub. In special cases where this service is required and assistance to operate the unit is available, it may prove to be a useful device.

2. Burke Elevating Seat Cushion. The Burke Elevating Seat Cushion (KOSHEN) (Fig. 41) manufactured by Burke Enterprises, 5833 Reeds Road, Mission, Kans., is designed to assist a person from a sitting position to standing position, and from a standing to a sitting position. It can be used with any type of chair or wheelchair.

The complete unit consists of an inflatable seat cushion, a motorized compressor unit, and a control box. Components are connected to one another by means of an air hose and electric lines housed within vinyl tubing. The inflatable seat cushion is 15 in. wide, 16 in. long, and 3½ in. high. Sandwiched between top and bottom plywood sections is an inflatable plastic bellows connected by an air hose to a compressor. A 1-in.-thick sponge rubber cushion is bonded to the surface of the top section. A white vinyl cover, with a zippered side, is fitted over the seat cushion.

The motor-compressor unit is 7½ in. long and 4½ in. in diameter with the compressor head extending 2 in. above the cylinder body. An air hose is connected to one side of the head and a carrying handle extends horizontally from the opposite side.
Mounted on the control box is a manually operated switch button and an air release valve for raising and lowering the seat cushion. Three lines lead from the control box: 1. a 7½-ft.-long electrical 110 volt a.c. power cord, 2. a vinyl clad air hose and electrical line to the motorized compressor unit, and 3. an air hose to the seat cushion. The total weight of the unit is 16 lb. 15 oz. including seat, motor-compressor, and control box.
Figure 41.—The Burke Elevating Seat Cushion (KOSHEN) is a pneumatically operated seat lift to assist a person to rise to a standing position.

To operate the unit the seat cushion is placed on the chair to be used, and the control box is placed on a table nearby to the right side of the cushion. The motor-compressor unit is also placed on the table. To rise, the patient presses the switch button to start the motorized compressor unit which feeds air into the cushion. As the cushion inflates, it raises the patient and "pitches" him forward onto his feet. When he attains a standing position, he
releases the switch. When the patient wishes to seat himself, he leans backwards against the inflated cushion and opens the air release valve slowly allowing air to escape from the bellows. The seat cushion de-inflates until a normal seated position is reached. When the cushion is fully inflated the seat is at an angle of 52 deg. with respect to the floor.

The device is mechanically adequate and performs the function of raising and lowering a patient to and from a seat in a reasonably satisfactory manner. But the basic concept of transforming electrical to mechanical to pneumatic power seems less desirable than a system which converts electrical energy directly to mechanical energy.

3. **Aztec Curb-Climbing Wheelchair Attachment.** The Aztec Curb Climbing Wheelchair Attachment was evaluated by this Center (BPR 10-11, p. 301). The results were generally positive and included three recommendations for modification. The manufacturer is revising the unit to comply with the recommendations.

4. **Theradyne Corporation Wheelchairs.** The Theradyne Corporation submitted three wheelchairs which were examined for compliance with tentative VA Standards for Wheelchairs, Self-Propelled, Folding, Multi-Purpose, dated May 1966.

The "Mountaineer" Model 808 is a heavy-duty type equipped with removable leg rests. Pneumatic tires may be ordered as a separate option (Fig. 42). Model 808 was cycled 100 times in a test to simulate conditions under which a wheelchair occupant would descend curbs approximately 6 in. in height, unaided or with a minimum of assistance. Wheel alignment, cross frame distortion, and axle concentricity were measured at intervals and at the completion of the test. The results were generally satisfactory. Some posterior seat frame deformation was noted under a fully loaded

![Figure 42. Theradyne heavy-duty wheelchair Model 808 (Mountaineer).](image-url)
condition (200 lb.), but the frame subsequently recovered upon removal of load.

The "Sovereign," Model 801, features an extra-strength cross-brace, heavy-duty upholstery, and a flared backrest—there are no removable components (Fig. 43).

The "Voyager," Model 815, is a convertible model which can be used as a conventional chair or for amputees. It is provided with two sets of rear vertical posts, each incorporating an axle housing that will accept the axle of the large drive wheel. The alternate post is 2½ in. to the rear of the conventional vertical post and is intended for use by amputees. Removable arm rests and leg rests are featured on this model (Fig. 44).

The three Theradyne chairs featured several unique and possibly valuable designs, such as the flared backrest and convertibility capability. They are generally of satisfactory construction and workmanship. A number of deviations from the standards on Wheelchairs were brought to the attention of the manufacturer.

5. Steeper Hand Splints. Several lightweight hand splints (Fig. 45) developed by Hugh Steeper, Ltd., of Roehampton, England, are being evaluated. They are designed to assist extension of the fingers in patients suffering partial paralysis, muscle spasms, and dystrophy where difficulty is experienced in extending the digits.

Plastic troughs attached to rubber covered "assist" wire cradle the digits. The plastic troughs are adjustable to individual patient requirements by heating and reshaping. The "assist" wire is bent to position the digits in extension to provide a force opposing the flexor muscles.
The Five Finger (Spider) Splint (Fig. 46) passively extends all five digits by means of the contoured wire over the dorsal portion of the hand. The other splints provide similar functions depending on the extent of involvement and loss. These devices are being fitted to patients through the VARO outpatient clinic, New York.

6. Kennaway Crutch Tips. Several experimental crutch tips (Fig. 47) were submitted for evaluation by Mr. Alexander Kennaway of London, England. They are fabricated of a special rubber developed by Mr. Kenna-
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way for the B. F. Goodrich Co. and were originally designed for nonslip tire treads. The samples were used by a test subject and compared with his standard crutch tips. His comments on the frictional characteristics, durability, and appearance of the crutch tips were recorded after approximately 8 months of total wear. His experience included some exposure to snow, ice, rain, and various types of terrain.

The samples (each of two different designs) were substituted for the standard type tip on the subject's dominant side. The subject found no obvious difference in the friction characteristics or the durability of the samples. However, one sample made "popping" sounds when used on smooth surfaces, such as a tile floor. He objected mildly to the green color of both samples in comparison with the brown wood-like color of tips previously used.

Additional patient trials and physical tests are being planned with Mr. Kennaway's assistance.

7. Typhlocane. As reported in BPR 10-10, p. 280, there has been an increasing incidence of failure in the currently used Typhlocane. Tests of sample canes indicated that they do not conform to specifications regarding the dimensions of the aluminum tubing and the threaded end. The manufacturer, the Precision Grinding Co. of Baltimore, Md., designed a new nylon tip for use with the Typhlocane (Fig. 48) in an effort to overcome the problems. The new tip is not threaded, and at $\frac{5}{8}$ in. in outside diameter, it is slightly larger than the old tip. To install the tip, it is heated for a few minutes in very hot water, after which it is pressed onto the end of the cane. Elimina-
tion of the threads inside the cane shaft may reduce the incidence of breakage in the distal portion of the cane. In addition it might facilitate length adjustment and permit the stocking of one length which could be adjusted to any desired lesser length. The new tip adds 4 grams (0.14 oz.) to the weight of the complete cane, a negligible increase since the tip currently in use weighs 12½ grams (0.44 oz.) and the new tip weight 16½ grams (0.58 oz.). The additional ½ in. in diameter of the new tip might be objectionable because of the 1/16-in. shoulder created at the proximal extremity of the tip.

Eight Typhlocanes were tested: Four with conventional threaded tips and four with the new pressed fit nylon tips. The eight samples of the Typhlocanes were checked against VA Specifications for Long Cane (Typhlocane) dated February 1965. A mechanical analysis of the physical and functional qualities of the canes was performed prior to destructive testing and clinical evaluation.

All eight samples were found to comply with VA specifications regarding the outside and inside diameters and wall thickness of the tubing. To determine the maximum load at failure, the canes were secured by the tips and bending moments were applied to the shaft. The conventional tip failed when a 21 ft.-lb. moment cracked the threaded nylon tip. The cane with the pressed fit tip fractured under a 30 ft.-lb. moment (Fig. 49).

A blind subject who routinely uses a Typhlocane used the new tip for approximately 1 month. Although the slipon tip felt "heavier" and the communication from the floor was different, there was no particular problem in adjusting to these differences. In all other respects, there were no functional differences between the slipon and the conventional threaded tip. The subject anticipated considerable difficulty in removing and replacing a worn tip.

The pressed fit tip offers the advantage of greater strength and easier adjustability of length. However, we have recommended redesign to permit changing tips by users.

8. Bio-Flote Wheelchair Pad. Manufactured by the Bio Clinic Co. of Sausal, Calif. (Fig. 50), this pad features a trapped water and air system to allow even weight distribution for wheelchair patients. The Veterans Administration Clinic, Los Angeles, Calif., provided 10 such units for their hospitalized patients. After use by patients for periods up to 1 year, no negative findings were reported. A number of advantages were noted: The Bio-Flote's weight is reasonable and can be handled easily by personnel. It can be readily adjusted to assure the most desirable distribution of weight. When properly adjusted, patients experience no instability. Bio-Flote pads can be used effectively in wheelchairs or in beds. The vinyl cover can be cleaned easily. No excessive bacterial colonization has been detected in the water or the foam rubber of pads that have been used for long periods of time. There
is no odor when the air is released. Prolonged use of the Bio-Flote pad does not cause excessive sweating. When punctured they can be easily repaired.
In the hospital situation the use of the Bio-Flote wheelchair pad for the seriously disabled patient appears to be a very useful orthopedic aid.

V. TESTING

A. Standards Development Program
None.

B. Compliance Testing
1. Lumbo-Sacral Corset Material. Six manufacturers submitted 12 specimens of fabrics used in lumbo-sacral corsets for the annual Specifications Compliance Test Program. Tests were conducted in accordance with Federal Specification CCC-T-191b, Part 5550.2 (D.O.D.), September 17, 1963, "Shrinkage in Laundering, Cotton, Linen, and Mixed Cotton and Linen Cloth." All 12 specimens were found to comply with specifications.
2. Upper-Extremity Components. Compliance tests were conducted on the Hosmer Internal Elbow, Model E-400, and the Sierra Voluntary Opening Hand Model 223. The Hosmer Internal Elbow met the specification requirements completely. The Sierra VO Hand also complied with specifications, but required frequent lubrication during cycling.
3. SACH Feet. During this period two additional manufacturers submitted samples of SACH feet which were tested for compliance with specifi-
B. The Prosthetics-Orthotics Service

Table 3 reflects the activity in the Orthotic Components Unit of the Prefabricated Appliances Section relating to the distribution of surgical supports and elastic hosiery. Table 4 reflects the distribution of Prosthetic Components.

VI. OPERATIONS REPORT FOR FISCAL YEAR 1969

The VA Prosthetics Center rendered 94,261 services during fiscal year 1969. Of this amount, 16,950 were services rendered directly to veterans reporting to our Center. A total of over 43,561 disabilities were treated, averaging slightly over two services per disability.

A. The Orthopedic Shoe Service

Table 2 reflects full fiscal year distributions from 1965 through 1968 for comparison with the present fiscal year. In the report year (1969), 695 veterans were provided with ortho-inlay shoes at a saving of $37,613. Ordinarily they would have been issued custom orthopedic shoes at far higher unit costs. In addition, we have issued 142 pairs of overshoes and rubbers.

<table>
<thead>
<tr>
<th>TABLE 2.—VA P NATIONAL ORTHOPEDIC SHOE PROGRAM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fiscal year</td>
</tr>
<tr>
<td>------------------------------------------------</td>
</tr>
<tr>
<td>Beneficiaries on rolls</td>
</tr>
<tr>
<td>New shoes, pairs</td>
</tr>
<tr>
<td>Pairs of new shoes issued per beneficiaries on rolls per year</td>
</tr>
<tr>
<td>Repaired shoes, pairs</td>
</tr>
<tr>
<td>Pairs of shoes repaired per beneficiaries on rolls per year</td>
</tr>
</tbody>
</table>

B. The Prosthetics-Orthotics Service

Table 3 reflects the activity in the Orthotic Components Unit of the Prefabricated Appliances Section relating to the distribution of surgical supports and elastic hosiery. Table 4 reflects the distribution of Prosthetic Components.
<table>
<thead>
<tr>
<th>Issuance channel</th>
<th>Surgical supports (each)</th>
<th>Elastic hosiery (each)</th>
<th>Surgical supports (each)</th>
<th>Elastic hosiery (each)</th>
<th>Surgical supports (each)</th>
<th>Elastic hosiery (each)</th>
<th>Surgical supports (each)</th>
<th>Elastic hosiery (each)</th>
<th>Surgical supports (each)</th>
<th>Elastic hosiery (each)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Directly to veteran</td>
<td>2,482</td>
<td>9,554</td>
<td>2,634</td>
<td>9,801</td>
<td>2,697</td>
<td>10,866</td>
<td>2,712</td>
<td>11,201</td>
<td>3,326</td>
<td>11,332</td>
</tr>
<tr>
<td>To Orthopedic Shops</td>
<td>3,722</td>
<td>11,049</td>
<td>5,885</td>
<td>14,517</td>
<td>5,056</td>
<td>15,924</td>
<td>4,630</td>
<td>16,333</td>
<td>4,540</td>
<td>19,104</td>
</tr>
<tr>
<td>Total</td>
<td>6,204</td>
<td>20,603</td>
<td>8,519</td>
<td>24,318</td>
<td>7,753</td>
<td>26,790</td>
<td>7,342</td>
<td>27,534</td>
<td>7,866</td>
<td>30,456</td>
</tr>
</tbody>
</table>
TABLE 4.—Distribution of Prosthetic Components by VAPC, Fiscal Year 1969

<table>
<thead>
<tr>
<th>Issuance channel</th>
<th>Temporary prostheses</th>
<th>SACH feet</th>
<th>Hydraulic systems</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AK</td>
<td>BK</td>
<td>Initial issue</td>
</tr>
<tr>
<td>For specific beneficiaries</td>
<td>77</td>
<td>88</td>
<td>110</td>
</tr>
<tr>
<td>To VA Orthopedic Shops</td>
<td>48</td>
<td>114</td>
<td>151</td>
</tr>
<tr>
<td>Total</td>
<td>125</td>
<td>202</td>
<td>261</td>
</tr>
</tbody>
</table>

* In F.Y. 1968, 82 above-knee temporary prostheses and 251 below-knee temporary prostheses respectively were distributed.

Special Data on Fluid Mechanisms

The total number of wearers of fluid mechanisms is 2,838. The breakdown by type is as follows:

- Hydra-Cadence: 1,316
- Dupaco: 1,134
- UC-BL Pneumatic: 26
- Mauch "A": 54
- Mauch "B": 308

New issue and replacement rate tables are attached for Hydra-Cadence, Dupaco, and Mauch "B" units (Tables 5 to 7).

Figure 51 shows the replacement rates over an 8-year period based on two sample months per year. The Mauch "A" and the UC-BL Pneumatic units were distributed as part of a clinical application study; rates of distri-

TABLE 5.—New Issue Rate (Monthly Average)

<table>
<thead>
<tr>
<th>Year</th>
<th>Hydra-Cadence</th>
<th>Dupaco</th>
<th>Mauch &quot;B&quot;</th>
</tr>
</thead>
<tbody>
<tr>
<td>1962</td>
<td>39</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1963</td>
<td>32</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1964</td>
<td>21</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>1965</td>
<td>9</td>
<td>23</td>
<td>7</td>
</tr>
<tr>
<td>1966</td>
<td>7</td>
<td>19</td>
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<td>1967</td>
<td>8</td>
<td>25</td>
<td>6</td>
</tr>
<tr>
<td>1968</td>
<td>10</td>
<td>23</td>
<td>0</td>
</tr>
</tbody>
</table>
TABLE 6.—Monthly Replacement Rate (Based on Four Sample Months per Year)

<table>
<thead>
<tr>
<th>Year</th>
<th>Month</th>
<th>Hydra-Cadence (percent)</th>
<th>Dupaco (percent)</th>
<th>Mauch “B” (percent)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1962</td>
<td>M</td>
<td>3</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>J</td>
<td>4</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>S</td>
<td>6</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>D</td>
<td>2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1963</td>
<td>M</td>
<td>5</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
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<td>D</td>
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<tr>
<td></td>
<td>D</td>
<td>4</td>
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</tr>
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<td>D</td>
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<td>4</td>
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</tr>
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<td></td>
<td>S</td>
<td>6</td>
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</tr>
<tr>
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<td>D</td>
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<td>1</td>
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<td>4</td>
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<td></td>
<td>S</td>
<td>4</td>
<td>4</td>
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</tr>
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<td></td>
<td>D</td>
<td>4</td>
<td>4</td>
<td>8</td>
</tr>
<tr>
<td>1969</td>
<td>M</td>
<td>6</td>
<td>4</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>J</td>
<td>7</td>
<td>6</td>
<td>4</td>
</tr>
</tbody>
</table>

M—March, J—June, S—September, D—December.

Malfunctions calling for replacement of units are listed below:

- **a.** Hydra-Cadence—tight or loose bushings; oil leaks;
- **b.** Dupaco—oil leaks; air in unit;
- **c.** Mauch “A” or “B”—noise due to side straps or loose flexion-extension adjustment attachment plate in shank.
TABLE 7.—Distribution of Mauch “S–N–S” Units

<table>
<thead>
<tr>
<th>VA location</th>
<th>Patient data</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Era</td>
</tr>
<tr>
<td>Albuquerque, N. Mex.</td>
<td>Viet Nam. 20</td>
</tr>
<tr>
<td>Atlanta, Ga.</td>
<td>do</td>
</tr>
<tr>
<td>Bay Pines, Fla.</td>
<td>do</td>
</tr>
<tr>
<td>Chicago, Ill.</td>
<td>do</td>
</tr>
<tr>
<td></td>
<td>do</td>
</tr>
<tr>
<td>Columbia, S.C.</td>
<td>World War II</td>
</tr>
<tr>
<td>Kansas City, Mo.</td>
<td>Viet Nam. 22</td>
</tr>
<tr>
<td>Kecoughtan, Va.</td>
<td>do</td>
</tr>
<tr>
<td>Minneapolis, Minn.</td>
<td>World War II</td>
</tr>
<tr>
<td>New York, N.Y.</td>
<td>do</td>
</tr>
<tr>
<td></td>
<td>do</td>
</tr>
<tr>
<td></td>
<td>do</td>
</tr>
<tr>
<td>Reno, Nev.</td>
<td>Viet Nam. 20</td>
</tr>
<tr>
<td>Sioux Falls, S. Dak.</td>
<td>World War II</td>
</tr>
<tr>
<td>Syracuse, N.Y.</td>
<td>Viet Nam. 24</td>
</tr>
<tr>
<td>Togus, Maine</td>
<td>do</td>
</tr>
<tr>
<td>Washington, D.C.</td>
<td>World War II</td>
</tr>
<tr>
<td></td>
<td>do</td>
</tr>
<tr>
<td>Wood, Wis.</td>
<td>Viet Nam. 22</td>
</tr>
<tr>
<td></td>
<td>do</td>
</tr>
</tbody>
</table>

The distribution of Mauch “S–N–S” units is given in Table 7. Eight wearers were WW II veterans; 12 were from the Viet Nam era; data were not available for seven patients. Ages of the Viet Nam era patients ranged from 20 to 37 with an average of 24 years. WW II wearers had an average age of 50 years ranging from 45 to 61.

Tables 8 to 10 show the fabrication and delivery of lower extremity prostheses and braces. Also fitted were 68 artificial arms and 1,185 arch supports.

C. The Restorations Service

Table 11 reflects the activities of the Restorations Service.

D. Special Clinic Teams

The VA Prosthetics Center’s Special Clinic Teams met 83 times during Fiscal Year 1969. Twenty VA Field Stations referred 386 veterans to our clinic for consultation. Due to the increased work load, two teams were in operation, each meeting once weekly whenever possible.

356
TABLE 8.—Complete Below-Knee Artificial Limbs Fitted by VAPC, Fiscal Year 1969

<table>
<thead>
<tr>
<th>Type</th>
<th>Permanent</th>
<th>Temporary</th>
<th>Totals</th>
</tr>
</thead>
<tbody>
<tr>
<td>PTB:</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cuff</td>
<td>27</td>
<td>21</td>
<td>48</td>
</tr>
<tr>
<td>Lacer</td>
<td>10</td>
<td>4</td>
<td>14</td>
</tr>
<tr>
<td>Non-PTB:</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Carved wood</td>
<td>44</td>
<td></td>
<td>44</td>
</tr>
<tr>
<td>Molded socket</td>
<td>30</td>
<td>26</td>
<td>56</td>
</tr>
<tr>
<td>Syme</td>
<td>18</td>
<td></td>
<td>18</td>
</tr>
<tr>
<td>Chopart</td>
<td>10</td>
<td></td>
<td>10</td>
</tr>
<tr>
<td>Total</td>
<td>139</td>
<td>51</td>
<td>190</td>
</tr>
</tbody>
</table>

*In F.Y. 1968, 175 below-knee artificial limbs were fitted.

REFERENCES


* Registered trademark of the Polymer Corp., Ltd.
Table 9.—Complete Above-Knee Artificial Limbs Fitted by VAPC, Fiscal Year 1969

<table>
<thead>
<tr>
<th>Type</th>
<th>Permanent</th>
<th>Temporary</th>
<th>Totals</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molded sockets, nontotal contact</td>
<td>43</td>
<td>42</td>
<td>85</td>
</tr>
<tr>
<td>Molded socket total contact</td>
<td>28</td>
<td>21</td>
<td>49</td>
</tr>
<tr>
<td>Carved wood</td>
<td>7</td>
<td></td>
<td>7</td>
</tr>
<tr>
<td>Hip disarticulation</td>
<td>4</td>
<td></td>
<td>4</td>
</tr>
<tr>
<td>Knee bearing</td>
<td>3</td>
<td></td>
<td>3</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>85</strong></td>
<td><strong>63</strong></td>
<td><strong>148</strong></td>
</tr>
</tbody>
</table>

*In F.Y. 1968, 188 above-knee artificial limbs were fitted.

Table 10.—Braces Fitted by VAPC, Fiscal Year 1969

<table>
<thead>
<tr>
<th>Item</th>
<th>Number</th>
<th>Item</th>
<th>Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Below knee</td>
<td>143</td>
<td>Cervical collars, prefabricated but custom fitted</td>
<td>18</td>
</tr>
<tr>
<td>Above knee</td>
<td>149</td>
<td>Spinal, custom</td>
<td>4</td>
</tr>
<tr>
<td>Arm</td>
<td>31</td>
<td>Spinal, prefabricated but custom fitted</td>
<td>79</td>
</tr>
<tr>
<td>Total</td>
<td><strong>424</strong></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* In F.Y. 1968, 463 braces were fitted.

Table 11.—VAPC Production of Prosthetics Restorations, Fiscal Year 1969

<table>
<thead>
<tr>
<th>Item</th>
<th>Number</th>
<th>Item</th>
<th>Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Artificial eyes</td>
<td>391</td>
<td>Plastic hands</td>
<td>73</td>
</tr>
<tr>
<td>Body restorations</td>
<td>5</td>
<td>Skull plates</td>
<td>1</td>
</tr>
<tr>
<td>Cosmetic gloves</td>
<td>149</td>
<td>Ocular conformers</td>
<td>130</td>
</tr>
<tr>
<td>Facial restorations, ear</td>
<td>25</td>
<td>Repairs to appliances (all)</td>
<td>68</td>
</tr>
<tr>
<td>Facial restorations, nose</td>
<td>1</td>
<td>Other items or services</td>
<td>377</td>
</tr>
<tr>
<td>Facial restorations, all</td>
<td>4</td>
<td>Total</td>
<td><strong>1,224</strong></td>
</tr>
</tbody>
</table>

* In F.Y. 1968, 830 items were issued.