V. ORTHOPEDIC AIDS
   A. Development
      None
   B. Evaluation (Components)
      1. Upper-Extremity Appliances
      2. Evaluation of Howmet Wheelchair (New Yorker)
      3. Evaluation of Howmet Lightweight Wheelchair (New Yorker)
      4. Revised Model Power Aid
      5. Motorette Wheelchair Power Unit
      6. Design Analysis of Several Orthopedic Aids
      7. Dierker Experimental Folding Crutch
      8. Wheelchair Pads
   C. Evaluation (Techniques)
      None

VI. TESTING
   A. Standards Development Program
      1. Stump Socks
      2. Specification Check and Preliminary Evaluation of Typhlocane
   B. Compliance Testing
      1. Stump Socks
      2. Upper-Extremity Components
      3. Lumbo-Sacral Corset Material

VII. OPERATIONS REPORT FOR FISCAL YEAR 1968
   A. The Orthopedic Shoe Service
   B. The Prosthetics-Orthotics Service
   C. The Restorations Service
   D. Special Clinic Team

REPORT

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This report details the progress and status of research and development conducted in various laboratories and services of the VA Prosthetics Center. It was compiled with the assistance of Donald W. Wright, Research Physiologist. A change has been made in the format and style of the report. Significant articles will be featured at the beginning of each semiannual report. In general, two kinds of articles will be featured—those of broad, general interest and complete reports or articles on important research or development projects.

The remainder of the semiannual report will continue to carry the details of our work as before.
In this report, two articles are featured in Section I. Although the work described is actually the product of many members of our research staff, the contributions of certain individuals were of special significance. The engineering efforts described in each of the three articles and in particular the design of the electric elbow are the work of Mr. Carl Mason, Staff Engineer. The technical procedures involved in making casts and prosthesis for the studies described were conducted by Mr. Thomas Pirrello, Research Prosthetist.

I. FEATURES

A. Measurement of Pressures Applied to the Lower Extremity

The development and the availability of a reliable pressure gage are significant events in prosthetics and orthotics research and development (BPR 10-9). During the past several years, a great deal of engineering talent and time has been allocated to this problem. The whole question of pressure measurement and currently available instrumentation for that purpose was the subject of a two-day meeting sponsored by the Committee on Prosthetics Research and Development (CPRD) and held at the VA Prosthetics Center, New York. Under the chairmanship of Mr. Colin A. McLaurin, Project Director, Prosthetic Research Training Unit, Ontario Crippled Children’s Centre, Toronto, Canada (Fig. 1), some 30 engineers and clinicians discussed and evaluated the relative merits of five separate systems capable of measuring pressures at stump-socket interfaces and similar sites. As a result of this meeting, it was clearly established that either of the two currently available systems is adequate. Both the Sensotec pressure transducer, a product of Scientific Advances, Inc., of Columbus, Ohio, and a similar device by Micro Systems, Inc., of Pasadena, California and employed at Rancho Los Amigos (Fig. 2) are sufficiently accurate, reliable, rugged, and small for almost every application proposed and known to CPRD. Similar gages whose principal merit is lower cost than the two cited above were also discussed. In these cases, lower cost was accompanied by lower orders of precision and by requirements for additional circuitry, thus reducing the real differential between them and indicating in our opinion the superiority of both the Sensotec and Micro Systems’ gages. There is therefore at the present time no significant reason for clinically oriented laboratories to devote additional time, money, and creative talent toward pressure gage development. Clinically significant problems, employing existing instruments, can now be attacked fruitfully.

In the previous issue of the Bulletin we reported the application of the Sensotec pressure transducer in preliminary studies of pressure at various interfaces—between stump and socket, leg and elastic stocking, and foot and shoe. With the acquisition of 10 Sensotec gages, several
formal studies have been initiated. Appropriate instrumentation was designed and installed at the VA Hospital in Seattle to study the patterns and vicissitudes of pressure between the stump and a plaster-of-paris cast applied immediately after surgery. In cooperation with Dr. Heinz Lippmann of the Albert Einstein Medical College, New York, we are comparing the pressure patterns developed between a stump and a plaster-of-paris cast with those developed by another material, Medicopaste Bandage (used in “Unna’s Boot”), a gauze bandage impregnated with
gelatin and zinc oxide glycerine. Thirdly, we are validating the tentative standards (BPR 10-8) for the compressive forces applied by elastic hosiery.

Under the direction of Dr. Ernest M. Burgess, the Seattle Prosthetics Research Study is monitoring stump-socket pressures, EMG activity of the gastrocnemius and tibialis anterior muscles, and skin temperature adjacent to the site of the amputation wound. To make this study possible laboratory electronic equipment was modified for clinical use. A console was designed with capabilities for recording four EMG channels, four
pressure channels, and two channels of data on forces applied through an instrumented pylon-type shank (vertical load and ankle moment). This equipment (Fig. 3), weighing approximately 1000 lb. (452 kg.), was specially designed for operation by non-engineering personnel. Constituting in effect a “portable laboratory” for the measurement of pressure, temperature, force, and EMG activity of patients fitted with rigid dressings immediately after amputation, the instrumentation was installed by VAPC personnel who also trained clinical personnel to operate the system (Fig. 4).

Pressure gages are now routinely applied by the Seattle group to the stump after wound closure under anesthesia. In one instance, pressure was monitored within 30 minutes after the wound was closed and dressed. The gages remain on the stump until the first cast change, a period of 7-10 days during which pressures are monitored with the patient recumbent,
standing, and walking on the instrumented pylon. EMG electrodes are also being placed on the stumps of newly amputated patients to monitor the electrical activity of the traumatized gastrocnemius and tibialis anterior. Pressure and gait events are being correlated in some patients and EMG activity and gait events are being correlated in other patients. To date, no attempts have been made to correlate EMG activity and pressure in the same patient.

In another study conducted in our laboratory the pressures applied by Medicopaste Bandage to a stump are also being investigated. Plaster of
paris, the current conventional material for immediate postsurgical casting, sets up rigidly. Questions have been raised concerning changes in the pressures applied to the stump particularly in walking when significant changes in stump volume occur. Since the rigid plaster-of-paris cast does not change its volume as the stump does, the pressure applied may vary considerably during the gait cycle. Medicopaste Bandage is similar to “Unna’s Boot,” an inelastic but flexible bandage long used for treating edema and ulceration of the leg. To our knowledge, it has never been previously applied to amputation stumps. Being flexible, it forms a roughly cylindrical container for the stump. As the stump volume changes during certain phases of gait, “Unna’s Boot” is deformed, becoming more oval than circular, thereby reducing its volume and maintaining the pressure applied to the stump. The working hypothesis underlying this study is that “Unna’s Boot” may apply more consistent pressure levels than a plaster-of-paris cast. If “Unna’s Boot” is at least as efficient as plaster of paris for immediate postsurgical treatment it might also represent a useful alternative in those clinical facilities where adequate prosthetics service may not be readily available.

In the first phase of this study, a pressure gage was placed on the mature stump of a below-knee amputee at a point just distal to the bulge of the gastrocnemius. The patient’s stump was fitted with casts of both plaster of paris and “Unna’s Boot” (Fig. 5), and the pressure patterns were recorded as the patient lay recumbent, stood, and walked. As shown in Figure 6, the Medicopaste Bandage seems to apply generally higher pressures to the stump than the plaster-of-paris cast. The baseline pressures after both casts were applied were approximately the same—25 mm. Hg.
and 27 mm. Hg. While the patient lay supine and contracted the residual portion of the gastrocnemius maximally, the pressure developed in the plaster of paris cast was higher (100 mm. Hg) than in the Medicopaste Bandage (80 mm. Hg.). The placement of the gage just below the posterior bulge of the gastrocnemius coupled with the rigidity of the plaster-of-paris wall may have combined to produce higher pressures.
In all other conditions, standing on both feet, standing on the prosthesis alone, and in walking, pressure within the Medicopaste Bandage ranged from 20 percent to 100 percent higher than in the plaster-of-paris cast. These studies are being continued, using four gages instead of one in an effort to record pressures over larger stump areas.

We have previously called attention to the extensive variation in the compressive forces applied by elastic hosiery of the same size and ostensibly for the same patient when they are obtained from different sources (BPR 10-6). Whether these were real differences or artifacts resulting from the imprecision of our measure instruments could not be readily determined. With the availability of the Sensotec gage, the pressures applied by elastic hosiery distributed by several commercial sources are being measured.

B. “Instant Laboratory”

We are attempting to meet the needs of clinicians desiring to conduct significant research projects by making available appropriate instrumentation in a program loosely called “instant laboratory.” It involves the design of special components and instruments, modification, and in some cases mild redesign of conventional electronic equipment for operation by physicians, therapists, and prosthetists. Repacked for easy shipment, they are readily installed in non-laboratory environments.

During the few years, we have received increasing numbers of requests from hospitals and universities for consultation and support in designing and organizing “bioengineering research facilities.” The typical request comes from the director of an orthopedic surgery or rehabilitation service with a highly specialized staff and a large and varied case load. The service chief is either a prominent, highly experienced clinician or a younger, research-oriented physician. Frequently, the service has been engaged in research projects of a clinical nature involving perhaps new surgical procedures, new rehabilitation techniques, or the application of new prosthetic and orthotic materials and methods. Their motives in seeking guidance and help in establishing “bioengineering research facilities” are principally to improve the objectivity of their measurement techniques and the validity of research results.

In the usual course of events such a project will request the financial support of one of the few government agencies which sponsor research programs in this field. The period between the planning of a program and the beginning of operations is often a long and arduous one. Fund-granting agencies must be assured of the competency of the director and his staff and of the adequacy of the facilities and equipment available. A formal statement must be prepared and submitted to an agency. Fund requests are processed through several stages of review and often require revision.
of the proposal before funds are finally granted. After approval, staff vacancies must be filled, and the required facilities and equipment must be procured. Moreover, the program should be put into operation at a rate that permits the preparation and submission of an acceptable report of progress within a year after funds have been granted.

The significance of the proposed research is often in direct proportion to the cost of procuring and operating the required facilities and instruments. A form of Parkinson's Law operates in these situations. A small group, consisting of one or more surgeons, therapists, and prosthetists, may wish to conduct studies requiring objective measurement. Simply to determine what instrumentation will be required, the group must be expanded to include engineering personnel. To operate complex equipment, additional support is frequently required in the way of engineering and electronic technicians. The inclusion of non-medical personnel and the handling of grant funds usually involve the services of additional administrative personnel. There is therefore a tendency for more and more people to be required to conduct programs of this type, including individuals whose primary function is not to collect and analyze research data but to handle organizational and administrative matters. Competent people of any profession are not usually attracted to transient or temporary positions. As a result, research facilities develop needs to perpetuate themselves even though their original interests have long since been satisfied.

The problem is minimized in those clinical facilities which have been operated under these conditions for many years, since their organizations have been structured to conform to the rules of the "grantsmanship" game.

It is far more difficult for otherwise adequate clinical institutions who submit proposals for the first time. Individual clinicians or small groups whose research interests may be highly significant, but perhaps of a one-time or transient nature, recoil in horror from the administration and organization requirements. Creative and sophisticated clinicians, not wishing to become involved in extensive administrative affairs related to the management of a research program, shy away from work which is potentially valuable and professionally interesting to them. An obvious solution is to support researchers with instruments and with technical advice for the duration of a particular project—in short, "instant laboratories."

During the past year several "instant laboratories" were designed for several special projects conducted by clinicians. Mentioned earlier in this report was the instrumentation made available and now operating at the VA Hospital in Seattle in the Prosthetics Research Study where Dr. Ernest M. Burgess is principal investigator. Although employed specifically for measuring pressures in sockets, EMG activity of traumatized muscles,
and gait factors during ambulation soon after surgery, the equipment has far more capability. Substituting other transducers for the pressure gages and EMG electrodes would enable this group to measure almost any other physical or physiological output. For example, acceleration of the total body center of gravity (c.g.) or of any of the limb segments could readily be recorded by simply feeding the output of linear or angular accelerometers directly to an amplifier bank in the console. This would enable the group to undertake studies of the relative efficiency of gait when a patient was

**Figure 7.**—Foot pressure patterns revealed by means of a mirror mounted in the walkway at a 45 deg. angle.
fitted with one prosthetic system or another. Strain gage force transducers applied to the sidebars of a brace or any prosthetic component would enable them to measure the dynamic loads applied to the components under various conditions of use. Pulse, blood pressures, EKG's, body temperature, or respiration rates could be recorded with equal convenience. In fact, any transducer of physiological or physical data whose output is above 10 microvolts can be recorded as can events whose frequency is up to 10,000 cycles per second. Most important is the fact that the entire system has been simplified to the point where non-engineering personnel are capable of operating it with minimal training. A short operating manual left in the hands of the clinicians provides adequate guidance for calibrating and operating the equipment.

The barograph (Fig. 7) is another example of an instrument designed for use by clinicians in diagnosing foot balance problems and in evaluating the effectiveness of various procedures for correcting foot problems. Based on an original concept of Professor Herbert Elftman, Chairman, Committee on Prosthetics Research and Development, NAS—NRC, this unit employs a unique principle which takes advantage of the differences in the refraction indices between a glass-air interface and a glass-polyvinylchloride interface. Our first permanently installed model was redesigned for easy portability and for convenient assembly in almost any clinical environment (Fig. 8). The device is readily positioned between two conventional tables, for example, to provide a walkway across which patients may walk. A conventional 16 mm. camera mounted at an appropriate distance from the mirror placed beneath the barograph enables a clinician to photograph the pressure distributions on the surface of both feet, statically and dynamically. Although primarily qualitative, the photographic data obtained permit easy identification of shifting pressures over the plantar surface of the foot. The effectiveness of arch supports, braces, and other orthopedic aids for the foot can be conveniently studied by comparison of the permanent photographic records made when a patient walks with different devices. Still pictures including Polaroid shots may be made under static conditions.

Previously described (BPR 10–5) was an instrumented pylon (Fig. 9) which represents in effect a portable force plate. The sensitive element of this device is a strain gage instrumented tube designed for easy insertion in the now conventional pylon used in below-knee and above-knee prostheses. The original instrument is capable of sensing most of the significant variables associated with the force interacting between the foot and the ground in standing and walking. These parameters include axial load, fore and aft shears, medial and lateral shears, torques, and most significantly, the direct readout of moments about the knee or ankle. Its advantages over permanently installed and extremely complex force plates
lie in the fact that it can be used on level ground, on stairs, uneven terrain, and in other non-laboratory environments. It is easily calibrated and its output is readily recorded in any strain gage or d.c. amplification apparatus. A simpler version of the instrumented pylon (Fig. 10) is being made available to several clinicians in support of research studies they wish to conduct on the gait patterns of various types of amputees fitted with a variety of prosthetic components.

A portable cyclograph has been designed for use in clinical environments.
Figure 9.—By means of 24 strain gages the VAPC instrumented pylon measures six important parameters of walking.
This device (Fig. 11) consists essentially of a view camera with a rotating shutter driven by a synchronous motor. This unit enables clinicians to record the kinematic factors of their patients’ gait. By placing reflective tapes at strategic points on the patient, displacements of the total body and its segments can be recorded on film. Studies of changes in gait patterns resulting from surgery, training, and the use of orthopedic devices can be readily recorded and compared over a period of time.

This program is yielding significant research results in an efficient and highly economical manner. There is a little or no necessity for administrative or organizational restructuring of clinical groups. The need for engineering and technical personnel is reduced to the vanishing point. The overall costs for instrumentation in these programs is only a tiny fraction of what it would be under conventional circumstances. Multiple use and the return of the equipment to the VA Prosthetics Center at the conclusion of each study combine to reduce costs dramatically. Most important—previously untapped sources of value and reliable research data are now yielding needed information.
FIGURE 11.—Inexpensive, portable cyclograph constructed of an open-shuttered view camera and a power-driven rotating shutter.

II. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

1. Gait Patterns of the Aged. The proportion of older patients in amputation clinic case loads seems to be increasing, perhaps as a result of increased longevity with a corresponding increase in the number of amputations for vascular insufficiency. Expanded social welfare programs through which medical service, including orthotics and prosthetics treatment, has been extended to larger numbers of people may also have an influence. Clinical personnel are therefore applying more time and effort than previously to problems of the aged amputee, commonly classed as "geriatric." In general, prosthetics service to such patients is based on modification of conventional components to reduce weight and energy costs, and to increase stability. Very few prosthetic components have been specially designed for the older patient. In an attempt to determine whether this class of patient requires significantly different prosthetic components, we have undertaken a study to identify gait features which distinguish the aged from the young population (BPR 10-9). This program is designed on the pattern of the earlier studies at the University of
California at Berkeley (1) which produced the major portion of currently
used reference data on locomotion. In the first phase of this program, the
gait of normal, non-amputee patients, 70 years of age and over, is being
evaluated with respect to temporal, kinematic, and kinetic factors. We
plan to evaluate a sample of 25 otherwise healthy older subjects. In the
second phase, a similar sample of amputees will be evaluated.

To date, five subjects ranging in age from 74 to 80 years have been
studied. Preliminary analysis of the force plate curves for the aged group
shows significantly different patterns than those for younger groups. In

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**GROUND REACTION FORCES EXERTED BY AGED SUBJECTS**

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**Figure 12**
particular, the vertical load curve (Fig. 12) shows an uncharacteristic pattern with less-than-expected variation between the peak magnitudes at foot-flat and push-off, and in mid-stance.

These studies are being continued with the cooperation of Dr. Morris Brand, Medical Director, Sidney Hillman Medical Center and Dr. Heinz Lippmann, Albert Einstein Medical College, both in New York.

2. Sources for Control of Prosthetic Knee Function. As previously reported (BPR 10-7), we have been engaged in a study of the phase relationships between the electrical activity of muscle groups of the lower extremity and significant gait "events," that is, such periods in the gait cycle as heel-contact to foot-flat when the flexion moment about the knee is large and the need for knee stability is important. This study is essentially a search for one, or possibly two, muscle groups whose electrical activity, due to its phase and character, may be used to control the resistance of a prosthetic knee mechanism. This problem has been attacked in the past by mechanical systems which react to the application of body weight. An example is the Regnell hydraulic knee whose resistance adjustment valve is closed or opened by means of rods and springs connected to a plate which is depressed and released as body weight is applied or removed from the heel. Vertical forces have been used to control the resistance valve of the Henschke-Mauch Model "A" unit; resistance increases when knee moment increases and angular acceleration of the knee has been of sufficient magnitude to displace a pendulum.

A patient wearing a conventional knee mechanism stabilizes the knee by hip extension, a motion produced by muscle activity and accompanied by increased electrical potentials. Controlling knee resistance by means of EMG signals may represent a closer and perhaps more efficient linkage between prosthesis and patient.

In continuation of this program, we have recorded the activity of both gastrocnemii in a normal person walking across the force plates. Shown in Figure 13 are the vertical loads and shear forces applied by both right and left feet correlated in time with EMG outputs of left and right gastrocnemius muscles.

Ground reaction force curves are, in general, the typical patterns characteristic of normal non-amputee gait. The cyclic activity of each gastrocnemius muscle has two gross phases—a relatively long burst of high magnitude followed by a short burst of low magnitudes. A striking phase relationship can be noted between the EMG activity of one gastrocnemius and the onset of stance phase on the opposite leg. The gastrocnemius is apparently active through most of the stance phase with the exception of the heel-contact to foot-flat period. Its magnitude continually increases until the time of heel-contact (approximately) on the opposite leg when
a sharp reduction in magnitude takes place tapering off shortly thereafter to a "quiet" level.

These time relationships indicate the possibility of controlling the resistance of a prosthetic knee to flexion by means of EMG signals from the gastrocnemius of the opposite leg. At the instant of heel-contact on the prosthetic leg, a sharp diminution of the EMG magnitude may perhaps be used to signal for an increase in resistance to knee flexion. The complete pattern of EMG activity consists of a characteristic burst related to terminal deceleration as the knee angle approaches 180 deg. and flexes slightly just before heel-contact. Just after heel-off, when the knee is almost fully extended, the EMG activity of the gastrocnemius in the same leg is maximal; the entire envelope of increased gastrocnemius activity increases
as the knee extends. It begins to decrease as the knee flexes just after push-off. This activity is related to both the resistance of the gastrocnemius to the high dorsiflexion moments about the ankle at this time and to its active contraction in push-off. The minor burst of gastrocnemius activity occurring just before heel-contact is attributed to the effects of deceleration of the swinging leg causing eccentric loading (stretch) through one head of the gastrocnemius. If further studies corroborate the timeliness and reliability of these signals, steps will be taken to design a device for practical testing.

B. Development (Components)

1. Adjustable Below-Knee Standard Prosthesis (BPR 10-7, 10-8, 10-9). At the present time, more than 150 units are in use in the field as temporary and immediate postsurgical prostheses. One hundred additional units are being manufactured for distribution to VA stations. Reports from the field indicate very little difficulty in their application. Misinterpretation of the instruction sheet accompanying each unit led to technical errors in installation in a few instances. As a result, revised instruction sheets for use of the below-knee standard prosthesis are now included with each unit. A portion of the next shipment of standard below-knee prostheses will be used for a clinical evaluation program conducted through the CPRD.

Customizing the shape and finish of below-knee standard prostheses has been a difficult problem due to the need for accommodating sockets of different sizes, shapes, and alignments with a cosmetic cover. This problem was solved to some extent with the vinyl cover reported in BPR 10-7. As a compromise however, it was not entirely satisfactory due to difficulty of accommodating varying socket diameters.

A plastic foam material (Koroseal) developed by B. F. Goodrich Company is being investigated as an alternative. The material is a closed-cell lightweight flexible foam which can be shaped with the conventional shop tools such as saws and sanders. It is delivered in block form from which a cosmetic shank may be carved. The foam texture is resilient and may be pigmented as desired.

2. Adjustable Standard Above-Knee (Multiplex) Prosthesis (BPR 10-7, 10-8, 10-9). Experience with six prototype models of the above-knee standard prosthesis has provided us with sufficient information to warrant testing on a broader basis. Experience with a large number of units is required to obtain reliable feedback about each of the multi-functions provided by this unit. We have, therefore, contracted for the production of 50 test models with all the accessories for interchanging knee control systems. The contract terms call for the initial production of three models for laboratory testing and inspection with stress analyses and production engineering provided by the manufacturer. Approval of these three units
will lead to production of 50 additional units for distribution to, and evaluation by, several centers.

3. Torque Absorber (Rotation) (BPR 10-7). Structural changes have been made in the design of the torque absorber in an effort to prevent damage to the Lamiflex bearing component. Under axial loads, the torque absorber displaced axially by vertical displacements of individual concentric rings. Under these conditions, bending moments applied by the pylon tended to damage the rubber laminations.

To prevent or at least reduce the transmission of bending moments in the pylon to the torque absorber, a stainless steel sleeve has been installed to receive the lower end of the pylon. Its function is absorption of the bending moments of the pylon without transmitting them to the laminated ring section of the torque absorber.

4. Standardization of Prosthetic Feet. In cooperation with Professor Charles Radcliffe of the University of California at Berkeley, designer of the original SACH foot, changes are being proposed in the shapes and dimensions of currently produced feet. Due perhaps to variation in manufacturing methods, SACH feet vary somewhat in size and shape. Reactions from the field indicate that current specifications do not adequately control the radius of the arch of the foot and the compressibility of the heel of the foot within the shoe.

The arch is generally considered too low due to the large radius of curvature. This sometimes requires prosthetists to reshape the arch in order to fit the foot properly into the shoe, a procedure which introduces variability in a standardized product and destroys the exterior finish of the foot. Often expressed is the view that the heel of the SACH foot is “too hard” for some patients. Several procedures involving drilling of holes in the heel of the foot are undertaken in an effort to “soften” the heel. Professor Radcliffe believes that certain changes due to manufacturing processes have occurred in the shape of the heel portion of the foot, preventing the foot from functioning properly. The posterior contour of the heel has become as wide at the level of the malleoli as it is near the plantar surface. Placed in a shoe, a heel of this shape cannot be compressed properly since its sides bulge against the counter of the shoe, restricting compression. Reshaping the heel to produce a narrower proximal section will permit the material to bulge laterally and to compress through the full range for which it was designed, thus making it “softer.”

The cross section of the foot through the ankle section is also being standardized in three sizes to simplify the problem of fitting and fairing cosmetic shanks.

Nine sets of foot models (Fig. 14) including each conventional size are being fabricated for distribution to manufacturers of SACH feet for
As ultimately agreed upon, the physical dimensions may be standardized for all types of prosthetic feet.

C. Development (Techniques)

Direct Forming of Lower-Extremity Sockets. Techniques for forming below-knee sockets produced from POLYSAR synthetic rubber have advanced to the point where this material is being used in other institutions across the country. Tubes made from POLYSAR have been made available to facilities selected by the Panel on Lower Extremity Prosthetic Fitting of the CPRD in an effort to broaden our experience. In addition, these tubes are available through commercial suppliers. A manual describing the method of socket fabrication, attachment, and the cosmetic finishing of sockets made from POLYSAR has also been distributed. The techniques for molding below-knee sockets made from POLYSAR for temporary prostheses have been introduced to the local commercial facilities in the Los Angeles area. Assistance in molding below-knee sockets produced from POLYSAR was requested due to the high interest in that area and the commercial availability of tubes made from POLYSAR. A complete program is being initiated by CPRD to introduce below-knee sockets made from POLYSAR synthetic rubber to the field and to evaluate their utility.

In the meantime, a method is being developed to produce a total contact socket entirely of material made from POLYSAR. Material made from POLYSAR in tubular form requires that the cavity immediately below the stump be filled with an alginate. The alginate-filled distal end of the tube made from POLYSAR is rigid enough to resist the external pressures produced by the pneumatic sleeve during casting. When the socket has been formed and the alginate removed, the cavity is foamed after dynamic alignment has been completed. To provide total contact, particularly distal contact, under controlled conditions, a closed-end pneumatic pressure
sleeve is used to apply pressure over the entire stump covered with material made from POLYSAR. Reasonably good results are being obtained in our laboratory with below-knee stumps.

The above-knee socket-forming technique is a more difficult task. To obtain the required quadrilateral shape in the proximal area of the socket, a preshaped socket brim (Hosmer type) is being used to deform the proximal brim. The heated tube is formed over the top of the casting brim (Fig. 15). The tube is reheated and replaced inside the casting brim. The patient's stump is pulled into the socket by means of a stockinet. The distal end is formed in another operation by means of
a closed-end pneumatic pressure sleeve. The pressure produced by the closed-end sleeve equalizes the pressure on the stump immediately below the preshaped socket brim.

The results of the initial experimental sockets are sufficiently encouraging to continue this work. In addition, we are investigating the use of preformed above-knee sockets produced from POLYSAR. It may be possible to stock preformed above-knee sockets in several sizes. Socket fitting might be accomplished by selecting the most appropriately sized preformed socket and customizing the fit by heating and molding localized areas of the socket.

D. Evaluation (Components)

*Modified Henschke-Mauch Model “A” HYDRAULIK System.* Since its development, the Mauch Model “A” Swing and Stance Control System has proved to be perhaps the single most advanced design in the world for the control of prosthetic knee motion. Motion about the Mauch knee more closely approximates normal knee motion with respect to velocity and acceleration than any prosthetic knee (BPR 10-2). Increasing use and experience with this unit as a result of the Clinical Application Study by Lewis and Bernstock, reported elsewhere in this issue, revealed certain areas for improvement. Mr. Hans A. Mauch of Mauch Laboratories,

![Figure 16](image)

**Figure 16.** In the original model of the Henschke-Mauch Model “A,” the piston rod attachment (point C) was posterior and superior to the knee axis (point A). In the revised model it is placed posteriorly.
Dayton, Ohio, the developer of this device, has redesigned the unit in several significant ways.

The hydraulic cylinder of the new unit is 3/4 in. (1.27 cm.) shorter than the earlier model and it is approximately 1/2 lb. (.227 kg.) lighter. An eccentric bushing located at the piston rod attachment allows up to 3 deg. of adjustment in the flexion-extension alignment of the knee-shank assembly. The old shank attachment assembly including the knee adjustment screw has been eliminated. Reducing the length of the cylinder has altered the geometrical relationships among the three points of attachment.

The older model was installed with the piston rod pin posterior and proximal to the knee axis. As shown in Figure 16, the piston rod pin was
positioned at an angle of 30 deg. above a horizontal line through the knee axis. In the modified unit, the piston rod pin is positioned in the same transverse plane, or at 90 deg. to the knee center. The effects of these changes are highly significant. The new locations of the attachment points are identical with other hydraulic mechanisms making it more convenient to install in standardized modular-type prostheses (VAPC Above-Knee Standard Prostheses). The new unit is also quite different functionally. Due to the altered moment arm, resistances to motion about the knee are reduced.

Shown in Figure 17, are the functional characteristics of a revised Model “B” unit whose swing phase control is identical with the new Model “A.” Peak resistance to knee flexion at the minimum resistance setting has been reduced approximately 25 percent. At intermediate resistance settings, peak values have been reduced by approximately 20 percent. More important, between 35 deg. and 65 deg., that is, in the second half of the flexion range, resistance values have been halved with a smoother decremental slope.

It is anticipated that these changes will simplify alignment of the knee shank assembly for the prosthetist and broaden the utility of the device for those patients who require relatively low resistance to knee motion, as well as for those with long stumps. In addition, it will standardize the attachment geometry, improve cosmesis, and reduce weight.

E. Evaluation (Techniques)
None.

III. UPPER-EXTREMITY PROSTHETICS

A. Development
VAPC “Hybrid” Above-Elbow Prosthesis. In the previous issue we discussed at length the inefficiency of conventional elbow mechanisms for above-elbow and shoulder-disarticulation prostheses. We referred in particular to the fact that above-elbow and shoulder-disarticulation amputees are required to expend perhaps half of the available excursion to flex the prosthetic elbow and the other half to open fully or close a terminal device. This requires them to control the terminal devices at the extreme limits of available excursion. We also pointed out that locking and unlocking the conventional elbow requires an awkward motion—abduction and extension of the shoulder, which may entail excessive displacement of the terminal device. We advanced the view that an externally powered elbow could easily eliminate the need for separate locking and unlocking motions and reduce half the total excursion required to operate the elbow and terminal device. A “hybrid” system consisting of an electrical elbow and a conventional terminal device might
permit the patient to operate his elbow more efficiently and yet retain the advantage of the valuable linkage between the terminal device and the residual tactile receptors.

The full potential of such a “hybrid” system might be more easily realized if the powered component and its controlling element were operated by the same body motions conventionally employed by trained amputees. Acceptance of this system by prosthetists might be enhanced if no significant alteration were required in conventional fabrication and harnessing procedures.

The VAPC “Hybrid” Above-Elbow Prosthesis was designed to fulfill these requirements. It consists of a conventional above-elbow socket, forearm, wrist unit, and terminal device. The standard above-elbow harness is modified in only one respect—elimination of elbow lock control strap and by the attachment of a control switch mounted in the otherwise conventional control attachment strap (CAS). The principal departure from the conventional prosthesis is the electrically powered elbow which is attached to the socket and the forearm in the conventional manner.

The control switch is an aluminum box, 3/8 in. x 1 1/2 in. x 1 3/8 in. (.95 cm. x 3.81 cm. x 3.49 cm.), weighing 1 oz (33.4 gm.). The switch

Figure 18.—VAPC upper-extremity control switch showing two microswitches (upper left), complete unit (upper right), and the unit installed in a conventional AE harness.
is attached to the conventional harness with a section of elastic sewn under it. Inside the switch, one end of which slides away from the other, are two small microswitches that are activated by the harness which displaces the top of the switch (Fig. 18). Displacing the slide 3/32 in. (.24 cm.) against the adjusted resistance of the elastic, results in the first control signal which produces elbow extension. Displacing the unit another 3/32 in. (.24 cm.) results in the second control signal which produces elbow flexion. Further application of force to the unit overrides both switches and transfers the force applied through the harness to the terminal device. A maximum excursion of approximately ¼ in. (.66 cm.) is required to operate the elbow. Locking is automatic. All the remaining

Figure 19.—VAPC electric elbow.
force and excursion can be used to control the terminal device. The elbow control element can be used with any electric elbow.

The VAPC electric elbow (Fig. 19) is designed for easy installation as it is compatible with conventional harnessing and requires no special fitting skills. The elbow consists of a plastic harmonic drive with a reduction ratio of 80:1. A planetary roller ball-bearing wave-generation assembly with a further reduction ratio of 11.7:1 produces a total reduction ratio of 956:1. The unit is rated at 100 in.-lb. (115 cm. kg.) stall torque, and 35 in.-lb. (40.2 cm. kg.) running torque.

Partially enclosed in the housing of the harmonic drive is a small d.c. motor. The motor mounting also serves to support the plastic housing and to attach the elbow turntable. Two diode block limit switches prevent the unit from being driven into hyperextension or into flexion above 135 deg. The total weight is 2 lb. (.226 kg.) which is approximately the same as the conventional Hosmer E-400 elbow. This system is being installed in a patient's prosthesis to determine the utility of the present design.

B. Evaluation (Components)

1. Gilmatic Electric Elbow Lock. The Gilmatic Company has designed a solenoid-operated electric elbow lock which is installed in a modified Hosmer E-400 elbow (Fig. 20). The device incorporates a power-conserving switch which keeps the solenoid inactive except during the actual locking or unlocking function, a period of approximately 10 milliseconds each. In an earlier evaluation of a demonstration model (BPR 10-6) the problem of controlling the lock was discussed. The operating model now under consideration has been designed for control by means of a “pad” type of switch which is essentially two contact plates held apart by a sheet of plastic foam which can be compressed sufficiently to close the switch. The switch is installed in the socket in a position to be compressed by the bulging of residual biceps or triceps muscles.

This device is a step forward in the attack on one of the most obvious and significant problems of above-elbow and shoulder-disarticulation amputees fitted with conventional components. In addition to flexion of the shoulder joint to flex the prosthetic elbow and to operate the terminal device, these patients must also abduct and extend the shoulder joint to lock and to unlock the elbow mechanism. The Gilmatic electric elbow lock is intended to eliminate the gross and often grotesque motions required for elbow locking and unlocking. This potential improvement is available at the cost of carrying a battery and wires.

Other approaches to this problem have led to fully powered electrical elbows which provide powered elbow flexion and extension and automatic elbow locking and unlocking. Some light on the relative merits of the two systems—fully powered electric elbow lifts versus electric powered locking mechanisms emerged as a by-product of this investigation.
The key point at issue in this matter is the efficiency of locking and unlocking the elbow irrespective of the elbow flexion-extension function. The most valid criteria for estimating elbow locking and unlocking efficiency are the ranges of motion required, the force which must be applied, and the reliability of the system. With this in mind, a comparative analysis was undertaken of the relative efficiencies of three elbow locking systems—the conventional Hosmer E-400, the Gilmatic electric elbow lock, and the VAPC electrical switch controlling an electrically powered elbow (see Section III.A.).

The efficiency of locking and unlocking the elbow was determined by measuring the displacements of the terminal device, elbow, shoulder, and total body action of a patient performing with each of the units. These displacements, which provide the most useful meaning of efficiency, were recorded photographically by the arrangement shown in Figure 21. By the time exposure technique, the origin, course, and termination of the motions of each point were simultaneously recorded.

As shown in Figures 22 and 23, it is obvious that both electrical systems permit the patients to lock and unlock the elbow with far less displacement, overshoot, and “body English” than the conventional system.
requires. The motion pattern obtained with the fully powered system seems more efficient than that obtained with the electric locking system in that overshoot is reduced and the displacements are smaller. This is confirmed by examination of the displacements of points on the sternum, shoulder, elbow, and wrist while the subject flexed his elbow to approximately 90 deg. from an initial position of full extension. The top projection, that is, the view from above the patient's head, shows the relatively large lateral excursions of the point on the wrist with the conventional system. The same operation also required far larger movements of the elbow with the conventional than with either of the other two systems. Similarly, movement of a point on the shoulder was also significantly
Figure 22.—Graphic analysis of the displacement patterns of an AE amputee locking after flexion to 90 deg. a conventional elbow, the Gilmatic electric elbow lock, and the VAPC electric elbow.
Figure 23.—Displacement of key points on prosthesis while patient unlocks and extends a conventional elbow, the Gilmatic electric elbow lock, and the VAPC electric elbow.
greater. The Gilmatic Elbow Lock required the same range of elbow displacement as the VAPC system and both were generally quite similar in this projection.

A similar picture is seen when elbow flexion is viewed from the front. The displacements of a point on the wrist and on the elbow were far greater when the conventional elbow was flexed and locked, than with either of the other two systems. As seen from the side projection, the elbow and wrist of the conventional system were displaced approximately twice as much when flexing and locking the elbow with either of the other two systems. Overshoot, the extension above 90 deg. required to lock the conventional elbow, is clearly shown in the terminal ends of the path of the point on the wrist. Overshoot with the Gilmatic Elbow Lock is of far less magnitude than with the conventional system; elbow flexion and locking are accomplished with even less extraneous movement.

As shown in Figure 23 the differences among the three systems are even greater when a patient unlocks the elbow in a position of 90 deg. of flexion, and allows it to extend fully. Both electrical systems required far less motion of the prosthesis than the conventional. This is particularly apparent in the top projection with respect to the movements on the point of the wrist, and in the side projection with respect to points on the elbow and on the wrist.

The advantages of a completely electrical elbow including lift and lock over a system employing an electrical lock in an otherwise conventional elbow are manifest: the excursion requirements for elbow flexion are reduced; locking function is more efficient; control over the terminal device position and operation is improved. Since the electrical locking system is built into an otherwise conventional elbow, its cost is not likely to be lower than a fully powered elbow, nor is its weight likely to be substantially lower. Power requirements for the locking function alone may be lower than the power required to operate a completely electrical elbow. Nevertheless, in our opinion, the other features far outweigh the power requirement differences, and we believe developers would be well advised to devote their efforts to improve fully powered electrical elbows and to curtail further work on electrical locks until the currently available models can be clinically evaluated.

2. Rimjet Mechanical Elbow Rotator. Developed by Mr. Joseph Ivko, a bilateral above-elbow amputee, this device is designed to provide humeral rotation in the above-elbow prosthesis (BPR 10–9). Mounted between the socket and the elbow turntable it is operated by means of a pull cable attached to the harness. Rotation takes place between two metal plates which are lined with ball bearings and secured by means of a thrust bearing. The locking mechanism is a spring-loaded plunger which engages one of nine locking positions through a range of 110 deg. rotation.
The unit weighs 12 oz. (.34 kg.) and the overall dimensions are 2⅔ in. (7.0 cm.) diameter and 1¾ in. (3.12 cm.) thick. To determine the usefulness of this device, several bilateral amputees will be fitted.

3. Externally Powered Hands. In continuation of our evaluation of commercially available externally powered hands, two additional below-elbow amputees have been fitted with experimental prostheses. Both subjects are undergoing training in control and use prior to the formal evaluation of their prosthetic performance.

One subject (fitted with the Viennatone Hand) previously wore a conventional voluntary closing hand. He has had considerable difficulty in unlearning the "double shuffle" control motion he previously used. The major problems are inadvertent terminal device opening and the dropping of objects. The second subject (fitted with the Italian Hand) has experienced some difficulty in controlling the closure of the terminal device.

4. Robin-Aids Voluntary Opening Hand. One model of a "resilient" voluntary opening hand, size No. 3, was purchased from the Robin-Aids Company for evaluation against the Tentative Standards for Mechanical Hands, established by USAMBRL (1965). This voluntary opening hand is designed as a three-fingered skeletal structure with the thumb and first two fingers opening voluntarily, and closing under tension provided by a wrap spring. The thumb approximates the index finger and the medial side of the middle finger providing a form of three-jaw chuck prehension. A rubber shell covers the hand and two soft foam floater fingers and is designed to be covered with a cosmetic glove.

The Robin-Aids Voluntary Opening Hand is wider, thicker, and heavier than a conventional Sierra Voluntary Opening Hand. The hand is anthropomorphic in appearance, but when it is fully opened (3½ in. [15.9 cm.]) a bulge caused by the wrap spring mechanism appears in the palmar region. The pinch force (approximately 1.2 lb. [.54 kg.]) appears to be minimal in the very small opening ranges.

In general, the Robin-Aids Voluntary Opening Hand conforms with the Tentative Specifications for Hands and is an acceptable VA Contract Item. However, the current standards for prosthetic hands are not completely adequate for evaluating terminal devices and are in need of extensive revision.

C. Evaluation (Techniques)

Direct Forming of Below-Elbow Sockets. As reported in previous issues, a program to evaluate the fabrication procedure for directly forming below-elbow sockets using material made from POLYSAR synthetic rubber has been undertaken by New York University. Several pilot wearers are in the process of being fitted to determine the accuracy of fit, durability of material, and the general utility of the procedure.
VAPC Research

Of the five patients fitted at the VA Prosthetics Center, two have rejected the sockets because of excessive perspiration. One subject previously wore a leather and steel frame socket and expressed a strong desire to return to this type of socket. He had previously rejected polyester-nylon laminated sockets for the same reason.

IV. LOWER-EXTREMIT Y ORTHOTICS

A. Development

1. Formo-Ped Ortho-Inlay Shoes. Custom orthopedic shoes are fabricated at relatively high costs due to the need for special lasts and patterns. In addition, the period between prescription and delivery of custom orthopedic shoes is rarely less than 1 month. Because of the highly specialized nature of these prescriptions little can be done in the present state of the art to reduce either the cost or the time lag between prescription and delivery. For patients with severe deformities, particularly those requiring high cork extensions, no substitute for properly fabricated custom orthopedic shoes is available. However, a number of beneficiaries suffer from milder deformities involving, for example, hammer toes or plantar ex crescences. For these patients, the Formo-Ped Ortho-Inlay shoe is being used. The Ortho-Inlay, it must be emphasized, is no substitute for patients who require custom orthopedic shoes. It does however, meet the needs of many patients who cannot be adequately served by ordinary stock shoes but who may not require the extensive modifications ordinarily integrated in custom orthopedic shoes.

As shown in Figure 24, the inlay is approximately \( \frac{1}{2} \) in. (1.27 cm.) thick, and it is readily modified to relieve areas on the plantar surface of the foot or to accommodate hammer toes. The shoe and inlay are supplied together and the inlay is modified for each patient's needs. Increasing numbers of patients (approximately 10 per week) are being adequately served with Ortho-Inlay shoes. Substantial dollar savings are being realized since the cost of the Ortho-Inlay is less than half that of custom orthopedic shoes. Moreover, patients can now be fitted in one day as against the 1 month waiting period required to provide custom orthopedic shoes. Perhaps even more significant is the fact that hospitalized patients who can be served with Ortho-Inlay molds are no longer retained as inpatients while waiting for their custom orthopedic shoe prescriptions to be filled.

2. AZTRAN. A poromeric material, AZTRAN, produced by the B. F. Goodrich Company is being tested as a substitute for leather in shoe uppers. To be considered in this evaluation are foot comfort, shoemaking efficiency, appearance, and wear qualities. In physical tests conducted by the B. F. Goodrich Company, the material withstood more than three million flexion cycles at temperatures ranging from 70 deg. to \(-40\) deg. F.
Tests of tensile strength, stretch, tear, and water penetration have indicated the adequacy of the material to withstand the hard use involved in shoe wear. Nontoxic and "porous," AZTRAN "breathes" through millions of tiny pores.

The VA Prosthetics Center has fabricated several pairs of shoes for clinical trials, using AZTRAN in the construction of the upper portion of the shoe. Experience to date in the processing of the shoes indicate that workability is comparable with leather in the ease of forming and molding. No difficulties were encountered in fabricating shoes with conventional shoe factory equipment.

B. Evaluation (Components)

Swedish Knee Support. Developed by the Svenska Centralkommitten fur Rehabilitering (Swedish Central Committee for the Disabled), the Swedish knee support is designed to control genu-recurvatum and to support the knee medially and laterally without auxiliary suspension.

The one-piece frame is constructed of aluminum with lateral uprights and a posterior band. The posterior band is designed to fit into the popliteal space, and it is covered with a water-filled pad. Anterior cuffs or straps are attached at the distal and proximal ends of the uprights.
As originally designed, the Swedish knee cage was not generally acceptable to patients. Adult males refused to accept the appliance because the ends of the uprights bulged through the trousers in the knee area when they were seated. In addition, the hydraulic pads leaked after short periods of use. At the suggestion of Mr. Charles Rosenquist, of Columbus, Ohio, the brace was modified by “flexing” the uprights and replacing the water-filled pad with a soft sponge pad.

A limited quantity of the modified Swedish knee support (Fig. 25 and 26) is being distributed to several stations to determine the practicability of this appliance.

C. Evaluation (Techniques)

None.

V. ORTHOPEDIC AIDS

A. Development

None.

B. Evaluation (Components)

1. Upper-Extremity Appliances. The Bio-Tex Devices, Inc., of Freeport, New York, submitted for evaluation an assortment of upper-extremity orthotic appliances for paralytic patients. Included were single and
multiple arm slings, wrist and finger splints, and various eating utensils. These devices are similar in design to other commercially available upper-extremity appliances.

In examining these items, the following observations were made:

a. The Single Strap Arm Sling. Quite similar in design to those fabricated at the VA Prosthetics Center. To improve patient comfort, however, an adjustable pad should be added to the over-the-shoulder strap.

b. The Multiple Strap Arm Slings. Could be refined by eliminating several straps. The Velcro strap that connects the olecranon to the wrist cuff was considered quite sound.

c. The Dorsal Wrist Splint with Pocket. Well designed, but appears to be somewhat flimsy (especially for spastics). It could be improved by reinforcing the pocket in the palm.

d. Combination ADL-Long Opponents Splint. The adjustable feature at the metacarpo-phalangeal joint is well designed. However, some type of positive lock should be provided. To accommodate all adult hands, the appliance is too large and should be shortened.

e. Spring Clip Holder for ADL Splints. Although the basic design of this item has merit, some means should be provided to prevent utensils from telescoping and rotating within the holder.

f. Adapted Eating Utensils. Although the eating utensils seem to be satisfactory, some provision should be made to modify them, or the holder, as mentioned in “e” above.

In general, the Bio-Tex devices examined in this evaluation are adequate and reflect satisfactory workmanship.

2. Evaluation of Howmet Wheelchair (New Yorker). The Howmet Corporation of Ohio, Hospital Division, Archbold, Ohio, resubmitted for evaluation, a folding-type wheelchair, the New Yorker, Model 6488-14-15 (Fig. 27). The chair is constructed conventionally of chrome-plated tubular steel with a double crossbar frame of chrome-plated, flat-stock steel, featuring non-removable desk-type armrests. This model incorporated several changes recommended to the manufacturer as a result of a previous evaluation (March 7, 1968).

The Howmet wheelchair was reevaluated according to current specifications for Standard Wheelchairs, Self-Propelled, Multi-Purpose (May 1966). The results indicated that two of the three previously noted deficiencies have been eliminated. However, the third problem remains: the force required to fold the chair (20 lb. [9.05 kg.]) still exceeds the specifications (15 lb. [6.80 kg.]). Since this is a matter which affects one of the basic features of this type of chair (foldability), the force required to fold should be reduced before the chair is authorized for issue.
3. Evaluation of Howmet Lightweight Wheelchair (New Yorker). One Howmet (New Yorker) Lightweight Wheelchair, Model No. 8515-15, manufactured by the Howmet Corporation, Hospital Division, of Archbold, Ohio, was submitted for evaluation (Fig. 28). The lighter weight of this chair (31½ lb. [14.25 kg.] as against 46 lb. [20.91 kg.] for the Howmet standard model) was achieved for the most part through the substitution of aluminum for steel in the dual cross frame members, drive wheels, drive rims, brake support brackets, and skirtguards. By the elimination of padded armrests and by the use of a lighter gage steel in the remaining structure, additional reductions were effected.

The Howmet lightweight wheelchair was evaluated in accordance with the tentative Veterans Administration Standards for Wheelchairs, Self-Propelled, Folding, Multi-Purpose (May 1966). Destructive testing for the purpose of determining durability of the chair was conducted in a
manner simulating normal conditions where an occupant (weight 175 lb. [79.4 kg.]) might be expected to negotiate a 6-in. curb.

The results of the evaluation pointed up several minor deficiencies which have been communicated to the manufacturer who has taken steps to eliminate them.

4. Revised Model Power Aid. Several deficiencies noted in previous models of this unit manufactured by Medical Aids of California have been overcome. The wheelchair frame has been completely isolated from the electrical system eliminating the potential hazard of electrical shock. As previously noted (BPR 10-9), the unit tended to lose traction when climbing a slope in excess of 4 deg. The manufacturer has redesigned the
unit to carry the battery farther forward by mounting it between the double
crossbar members of the Howmet wheelchair. This forward shift of a
substantial proportion of the total weight of the system also shifted the
vertical projection of the center of gravity several inches anterior to its
previous position improving the traction of the front drive wheel on slopes.
Models have been procured for limited field testing by VA stations in
the New York area.

5. **Motorette Wheelchair Power Unit.** Two models of the Motorette,
an electronically controlled, battery-powered motor unit, were submitted by
the Motorette Corporation of Reseda, California (Fig. 29), for evaluation.
These units are designed to be put on or removed easily and quickly from
any standard wheelchair by turning two self-locking latches. A single
“joy stick” control box is designed to be snapped onto either of the wheel-
chair arms. Two 3/4 horsepower (.253 metric h.p.) motors turning indi-
vidual pinion drive gears which bear on the wheelchair tire surfaces propel
the chair up to 5 m.p.h. (.138 km./min.). These units are currently being
evaluated in comparison with two other commercially available units for
powering wheelchairs.

6. **Design Analysis of Several Orthopedic Aids.** During this period an
analysis of three specific orthopedic devices was completed on the basis

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**Figure 29.**—The Motorette power unit conveniently converts any standard folding
wheelchair into a power chair.
of photographs and brochures submitted to the VA Prosthetics Center. No models were tested.

a. Lectro-Lift. This device is manufactured by Lectro-Lift, Inc., of Roseau, Minnesota. It is a compact, battery-powered vehicle designed to serve as a wheelchair, a powered walker, and a powered lifting device.

b. Universal Out-of-Doors Elevator. Manufactured by Wheelchair Elevators, Inc., of Broussard, Louisiana, the elevator is powered by a ¾ horsepower (.76 metric h.p.) motor from household current. It provides a means of aiding patients in their wheelchairs in and out of their front door.

c. Gates Patient Handler. Designed by Mr. J. T. Gates of Columbus, Ohio, the unit incorporates in one device the functions of a powered wheelchair, a patient lifter, an oscillating tilt board, and a convertible commode seat.

The results of the above analyses have been submitted to the developers together with certain suggestions for improvement.

7. Dierker Experimental Folding Crutch. Mr. Charles E. Dierker, the developer of this device, submitted a single crutch designed for temporary use by leg amputees while traveling. The model unit was easily disassembled into three components, one of which consists of two elements (Fig. 30). It is not actually “foldable” or “collapsible” in the generally used sense of these terms. The crutch is disassembled by depressing spring-loaded pins which fit into appropriate holes in each of the two sidebars. As the longest single section does not exceed 16 in. (40.6 cm.), the crutch can be easily packed in a suitcase or attaché case. It is constructed of lightweight tubular aluminum members and a cast aluminum axillary support. The complete unit weighs only 3 lb. (1.36 kg.).

Figure 30. The Dierker folding crutch can be disassembled into three sections for transportation in a suitcase.
A device of this type may be useful for leg amputees who perhaps, while traveling, would have occasion to move around their hotel rooms without their prostheses. It is difficult to see other realistic applications. For use in and around the home, any of the conventional crutches would suffice. For painful stump conditions, general instability, or prosthetic problems, the “take-apart” feature would seem to be unnecessary since more than temporary use would be required.

8. Wheelchair Pads. Several types of wheelchair pads are currently being evaluated at the Bronx, Manhattan, and East Orange, VA hospitals, the New York VA Regional Office Outpatient Clinic, and at the Georgia Warm Springs Foundation.

Our experience to date with an inflatable cushion (Bye Bye Decubiti) seems to indicate a degree of usefulness for paraplegics. Those with small decubitus ulcers were able to continue sitting without retarding the rate of healing. These patients were all capable of voluntarily shifting their weight. Some difficulties were noted among patients with high level lesions (quadriplegics), those with problems of instability, and those with weakness of the upper extremities. The “bounce” of the cushion as the air shifted tended to “push” these patients into uncomfortable and awkward positions. Frequent repositioning of the patient by ward personnel was required.

Fear of punctures was expressed by some patients. In one case, a pad actually leaked air while the user was riding in his automobile. By the time he arrived at his destination, the pad was completely deflated.

Follow-up of one of the Stryker Floatation Pads, reported previously in BPR 10-8, showed that it was out-of-service; it had “rolled up into a ball” and was no longer useful. Two additional Stryker pads, showing evidence of extreme wear, were brought to our attention. One of the pads was part of an early run in which changes had been made in formulation of the gel as an economy measure. This unit was replaced by the manufacturer. The other worn unit was repaired and returned to the patient.

C. Evaluation (Techniques)

None.

VI. TESTING

A. Standards Development Program

1. Stump Socks. In an effort to develop objective standards and functional specifications for stump socks, samples of currently qualified stump socks were subjected to a program of objective physical tests by the United States Testing Company, Inc., Hoboken, N. J., to discriminate qualities which are assumed to be important to amputees. The tests measured fiber content, wool grade, thickness of the material, softness,
pilling resistance, coefficient of friction, water absorption, air permeability, and compression under load and ability to recover from deformation. All of these tests were performed with standard apparatus employing five sample socks from each of five manufacturers before and after laundering.

An analysis of the results indicated that certain of these tests clearly discriminated differences among the samples. Other tests failed to distinguish differences. The most useful tests were:

1. Wool grade (ASTM Method D 472) and handleometer (Thwing Albert Tester) tests for softness.
3. Pilling (Random Tumble Test ASTM D 1375) or resistance to balling of fibers.
4. Resiliency (Instron Model TTC) or recovery after compression.

To determine the validity of these data as indices of stump sock quality as perceived by amputees, a limited field trial has been initiated. Samples of each type of stump sock will be issued to a small group of highly knowledgeable and experienced amputees. Prior to issue, all the socks will be laundered in the laboratory and all identifying marks recorded. Subjects will wear each of the coded, but otherwise unidentified brands of stump socks in their normal sock wear pattern. After wearing each sock they will record their reactions on a card and mail it to this Center. Comments will be elicited on softness, shock absorption, sweat handling, abrasiveness, and general comfort.

2. Specifications Check and Preliminary Evaluation of Typhlocane.

In response to a report from Mr. Donald Smith, Chief, VA Prosthetics Distribution Center, Denver, Colorado, on bending and breakage of Typhlocanes, eight sample canes were checked against VA Specifications for The Long Cane (1965). Four of these specimens were new, recently manufactured canes, and four were used and damaged (Fig. 31).

Both new and old canes were substantially below specification requirements with respect to wall thickness and inside diameters. In addition, the threaded area of the damaged canes had a wall thickness of approximately .013 in. (.033 cm.). The unused canes had a wall thickness in the threaded section of approximately .023 in. (.059 cm.).

It is quite clear that significant departures from the VA Specification requirements are contributory factors in the failure of these canes. Tube walls are too thin and, in some models, threads are too deep. These problems are further complicated by the length of the threaded section. The thin wall section and the depth of the thread might not make the cane as susceptible to damage if the length of the threaded section were reduced. However, reducing the length of the threaded section would reduce the range of adjustability available in the Typhlocane.
FIGURE 31.—The four damaged Typhlocanes on the right and the new one on the left were checked against specifications to determine possible causes of failure.

As a palliative solution to the immediate problem of breakage, the manufacturers have been informed that the aluminum tubing is significantly below specification requirements, particularly as regards wall thickness and inside diameters. They were also informed of the discrepancy in thread depth of different manufacturing runs.

As a more definitive solution to the overall problem, we have recommended a quality control test program to permit the testing of selected samples for compliance with current specifications with particular reference to the discrepancies cited above. In addition, a complete review of the Typhlocane design with a view toward eliminating the threaded section entirely is being undertaken.

B. Compliance Testing

1. Stump Socks. Samples of a "brushed" stump sock (equivalent to an American size 4, length 22 in. [56.0 cm.]) were submitted for evaluation by Hugh Steeper, Ltd., Roehampton Lane, London. Submission of a
sample in this size, completed the test program on Remploy stump socks previously reported in BPR 10-9. This evaluation (Table 1) and the previous results (BPR 10-9) indicated that the Remploy brushed stump socks substantially meet all the current specifications except for a marginally acceptable number of wales per inch (2.54 cm.).

2. Upper-Extremity Components.

**Hosmer Experimental Internal Elbow, Model E-400-2.** One model was tested for compliance with “Tentative Specifications for Adult Size Elbow, Artificial, Internal, Alternating, for Above Elbow Amputees.” Physical strength tests were performed and the reliability of the unit was checked during 300,000 cycles of elbow locking operation. All of the requirements were satisfactorily met. The use of Delrin instead of Almag 35 on part No. E-402 Yoke seemed quite adequate for sustaining the loads applied during testing.

<table>
<thead>
<tr>
<th>TABLE 1.—Evaluation of H. Steeper Stump Socks—Leg, Wool</th>
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<td><strong>Type</strong></td>
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<td>Size, American Standards</td>
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<tr>
<td>Length, in.</td>
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<tr>
<td>Width:</td>
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<tr>
<td>2 in. from toe (5.08 cm.)</td>
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<tr>
<td>6 in. from toe (15.20 cm.)</td>
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<tr>
<td>Current VA Specifications</td>
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<tr>
<td>Specified tolerances</td>
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<tr>
<td>Yarn size 16-18.5</td>
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<tr>
<td>Ply 3, 4, 5, 6, 7</td>
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<tr>
<td>Stitches/In. 18 ±2 (S/2.54 cm.)</td>
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<tr>
<td>Wailes/In. 12 ±1 (W/2.54 cm.)</td>
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<tr>
<td>Maximum Shrinkage %</td>
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<td>after 5 washings</td>
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<tr>
<td>Length</td>
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<tr>
<td>Width at 2 in. (5.08 cm.) 10%</td>
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<tr>
<td>Width at 6 in. (15.20 cm.) 10%</td>
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<td>* Not within current VA Specifications</td>
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282
3. *Lumbo-Sacral Corset Material.* Annual specifications compliance tests were completed during this period on samples of lumbo-sacral corset material submitted by four manufacturers. Tests were conducted in accordance with Federal Specification CCC–T–191G, Part 5550.2 (DOD), September 17, 1965, Shrinkage in Laundering, Cotton, Linen, and Mixed Cotton and Linen Cloth.

The Specifications were met or exceeded by the Warrior Surgical Supply Company, S. H. Camp Corset Company, Kellogg Corset Company, and Atco Surgical Supply Company.

**VII. OPERATIONS REPORT FOR FISCAL YEAR 1968**

During the past fiscal year, the VA Prosthetics Center rendered 95,645 services compared with 63,821 for the prior fiscal year. Of the overall total of services rendered, 15,141 were rendered directly to veterans reporting at our Center for an average of 60 services daily. The number of disabilities treated was 46,337; thus, on the average, each disability received a little over two services per annum.

Issues of new custom orthopedic shoes and shoe repairs showed significant increases. The issue of surgical supports and elastic hosiery remained relatively constant. Distribution of prosthetic components, including hydraulic systems, also showed increases. As suspected and reported in the previous report (BPR 10–9 Spring 1968), the increases were due primarily to the vast influx of returning wounded veterans from the Viet-Nam era, and the increasing frequency of disabilities associated with age of veterans of prior wars.

**A. The Orthopedic Shoe Service**

Table 2 shows full fiscal year comparisons in our national Orthopedic Shoe Program from 1964 through 1968. In addition to the items reported

<table>
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<th>Table 2.—VAPC National Orthopedic Shoe Program</th>
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<td><strong>Fiscal Year</strong></td>
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<tr>
<td>Beneficiaries on rolls</td>
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<tr>
<td>New shoes, prs.</td>
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<tr>
<td>Prs. of new shoes issued per benef. on rolls per yr.</td>
</tr>
<tr>
<td>Repaired shoes, prs.</td>
</tr>
<tr>
<td>Prs. of shoes repaired per benef. on rolls per yr.</td>
</tr>
</tbody>
</table>
### Table 3. — VAPC Surgical Support and Elastic Hosiery Program

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Directly to veterans</td>
<td>2,434</td>
<td>9,989</td>
<td>2,482</td>
<td>9,554</td>
<td>2,634</td>
</tr>
<tr>
<td>VA Orthopedic Shops</td>
<td>3,995</td>
<td>12,258</td>
<td>3,722</td>
<td>11,049</td>
<td>5,885</td>
</tr>
<tr>
<td>Totals</td>
<td>6,429</td>
<td>22,247</td>
<td>6,204</td>
<td>20,603</td>
<td>8,519</td>
</tr>
</tbody>
</table>
in Table 2, we provided 259 veterans with Ortho-Inlay shoes at a savings of $11,645. Ordinarily we would have provided them with custom orthopedic shoes at a far higher cost. We also issued 106 pairs of overshoes and rubbers.

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 4.—Distribution of Prosthetic Components by VAPC, F.Y. 1968

<table>
<thead>
<tr>
<th>Issuance channel</th>
<th>Temporary prostheses *</th>
<th>SACH feet</th>
<th>Hydraulic systems b</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>128</td>
<td>154</td>
</tr>
<tr>
<td>VA Orthopedic Shops</td>
<td>5</td>
<td>123</td>
<td>241</td>
</tr>
<tr>
<td>Totals</td>
<td>82</td>
<td>251</td>
<td>395</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* In F.Y. 1967 we distributed 151 above-knee temporary prostheses and 212 below-knee prostheses respectively.

b Seventy-seven wearers changed from one hydraulic system to another hydraulic system, and, or were issued spares. In F.Y. 1967 we had 2081 active wearers of hydraulic systems.

TABLE 5.—Complete Below-Knee Artificial Limbs Fitted by VAPC, F.Y. 1968

<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>19</td>
<td>46</td>
</tr>
<tr>
<td>Lacer</td>
<td>18</td>
<td>4</td>
<td>22</td>
</tr>
<tr>
<td>Carved wood</td>
<td>12</td>
<td>—</td>
<td>12</td>
</tr>
<tr>
<td>Molded socket non-PTB</td>
<td>53</td>
<td>13</td>
<td>66</td>
</tr>
<tr>
<td>Syme</td>
<td>21</td>
<td>—</td>
<td>21</td>
</tr>
<tr>
<td>Chopart</td>
<td>8</td>
<td>—</td>
<td>8</td>
</tr>
<tr>
<td>Totals</td>
<td>139</td>
<td>36</td>
<td>*175</td>
</tr>
</tbody>
</table>

* In F.Y. 1967, 150 below-knee artificial limbs were fitted.
Bulletin of Prosthetics Research—Fall 1968

The Limb & Brace Section of The Prosthetics-Orthotics Service fabricated and delivered lower-extremity prostheses and braces as indicated in Tables 5, 6, and 7. They also fitted 42 artificial arms and 1,588 arch supports.

C. The Restoration Service

Table 8 lists the major activity of the Restorations Service.

D. Special Clinic Team

The VA Prosthetics Center's Special Clinic Team met 55 times during fiscal year 1968. One hundred and forty-four veterans were referred to our clinic for consultation by 22 VA field stations.

<table>
<thead>
<tr>
<th>TABLE 6.—Complete Above-Knee Artificial Limbs Fitted by VAPC, F.Y. 1968</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type</td>
</tr>
<tr>
<td>Molded socket, non-total contact</td>
</tr>
<tr>
<td>Molded socket, total contact</td>
</tr>
<tr>
<td>Carved wood socket</td>
</tr>
<tr>
<td>Hip disarticulation</td>
</tr>
<tr>
<td>Knee bearing</td>
</tr>
<tr>
<td>Totals</td>
</tr>
</tbody>
</table>

* In F.Y. 1967, 153 above-knee artificial limbs were fitted.

<table>
<thead>
<tr>
<th>TABLE 7.—Braces Fitted by VAPC, F.Y. 1968</th>
</tr>
</thead>
<tbody>
<tr>
<td>Item</td>
</tr>
<tr>
<td>Below knee</td>
</tr>
<tr>
<td>Above knee</td>
</tr>
<tr>
<td>Arm</td>
</tr>
<tr>
<td>Spinal, custom</td>
</tr>
<tr>
<td>Spinal, prefabricated but custom fitted</td>
</tr>
<tr>
<td>Cervical collars, prefabricated but custom fitted</td>
</tr>
<tr>
<td>Total</td>
</tr>
</tbody>
</table>

* In F.Y. 1967, 539 braces were fitted.

Forty-six prefabricated braces of all kinds were distributed in addition to the above which were either fabricated or custom fitted.
### Table 8.—VAFC Production of Restoration Items, F.Y. 1968

<table>
<thead>
<tr>
<th>Item</th>
<th>Number</th>
</tr>
</thead>
<tbody>
<tr>
<td>Artificial eyes</td>
<td>383</td>
</tr>
<tr>
<td>Body restorations</td>
<td>22</td>
</tr>
<tr>
<td>Cosmetic gloves</td>
<td>65</td>
</tr>
<tr>
<td>Facial restorations, ear</td>
<td>9</td>
</tr>
<tr>
<td>Facial restorations, nose</td>
<td>4</td>
</tr>
<tr>
<td>Facial restorations, orbital</td>
<td>12</td>
</tr>
<tr>
<td>Plastic hands</td>
<td>48</td>
</tr>
<tr>
<td>Repairs to appliances (all)</td>
<td>58</td>
</tr>
<tr>
<td>Other items or services</td>
<td>229</td>
</tr>
</tbody>
</table>

### Reference