SEMIANNUAL REPORT OF THE VETERANS ADMINISTRATION PROSTHETICS CENTER

(July-December 1965)

INTRODUCTION

The VA Prosthetics Center continues to emphasize development of pylon structures for "permanent" as well as "temporary" lower-extremity prostheses. The adjustable below-knee pylon has been engineered to the point where it is now ready for limited testing by others. Furthermore, with the cooperation of commercial production sources, development of suitable cosmetic covers is being achieved. With these efforts and with the planned development of the multifunctional above-knee pylon and the several methods of forming lower-extremity stump sockets, we look forward to a new artificial limb era in which the present wood, crustacean designs may be replaced by pylons with cosmetic covers and in which sockets will be achieved more directly, hopefully without plaster-of-Paris impressions. Perhaps later, sockets for the below-knee and above-knee levels may be made from prefabricated, pre-shaped blanks to receive final contouring, especially for a required distal "contact," during or after simple trials on the amputee.

We have long felt that the above-knee, swing-phase control problem had been solved. There certainly is no shortage of mechanisms on the market (ranging from the simplest constant mechanical friction device to the most functional hydraulic). However, there are still new items that require scrutiny. Such a device is the UCB Pneumatic AK Swing Control Knee, a system perhaps not as functional as the more complicated hydraulic devices but one that provides, in a simple way, a swing-phase control of benefit to a large segment of the amputee population.

Our research touches also on more fundamental areas. We seek some rationale on the employment of elastic hosiery and its effect on venous return. The relation between externally applied pressures and the physiological yield deserves attention.

We are now also urging the accumulation of more fundamental data on the early stages of ambulation of a new amputee. Our instrumented pylon was designed for use by those researchers who have observed clinical success in immediate post-surgical application of prostheses. But, we now call for some supplementary, objective data on changing patterns of weight-bearing and on the improved prosthetic control which we estimate accrues from either early ambulation or surgery employing tension myodesis or both.

We continue to review the developments of others in upper-extremity components, particularly those using external power. Our interest in stimulating developments in orthotics through our own still limited efforts never wanes. Shoes, shoe modifications, and plantar supports, by virtue of the huge clinical responsibility we have through our Orthopedic Shoe Program, continue to be orthotic "devices" requiring research in depth we have not yet achieved.

The wheelchair lately has become one of our primary concerns. Recent new developments made by a progressive industry have required us to look carefully at our own VA specifications which not only are outdated but suffer from specificity. We have adopted the "functional standard" in place of the limiting "specification" since we feel it imperative not to limit industrial improvements by requiring, for example, an axle to be X inches in diameter of Y material. Rather we shall ask that the axle provide a certain strength under standardized laboratory conditions replicating actual service requirements as nearly as possible. Thus, the manufacturer can chose his dimensions and materials as he wishes as long as the "functional output" meets our clinical need. Hopefully, we may now open our VA market to progressive improvements in wheelchairs not only for those whom we serve but for all disabled.

Meanwhile, we are still fascinated with the progress of the materials technology especially in relation to prosthetic-orthotic applications. Even though we now foresee the day when plastic laminates may be displaced by prefabricated socket blanks of other synthetic materials and when wood and its necessary plastic laminate reinforcement may no longer be required, we have been urged by Mr. A. B. Wilson of CPRD to check new fiber materials for lamination with the standard resins. It appears now that polypropylene cloth laminates may have a place in prosthetics and orthotics thus requiring the tests we describe.

The staff of our Center especially Henry F. Gardner and those people under the leadership of Dr. Edward Peizer deserve a great deal of credit for their continued productivity in research and development. Dr. Peizer's personal efforts in putting together the research report and those of Frank A. Witteck in preparing the operational data are sincerely appreciated and deserve recognition.

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I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

Effects of Compression of the Lower Extremity. Elastic hosiery is generally prescribed on the basis of girth measurements taken at various levels of the limb. Stockings may fit the leg properly and comfortably, but very little is known about the effects of compressive forces. Knit geometry and the ratio of elastic to non-elastic elements vary among the products of several manufacturers, and the magnitudes and gradients of the pressures they apply are unknown. Elastic hosiery should be prescribed on the basis of physiological indications of the optimal limits of compression. Valid physiological criteria will provide functional standards for evaluating currently prescribed elastic hosiery. Specifications can then be established to guide the fabrication of hosiery with regard to dimensions and the ratio of elastic to non-elastic elements.

This may be possible in terms of the basic relationship between circumferential tension (F) and compressive pressure (P_0). If we know: 1. the relaxed circumference of the stocking (C_2), 2. the circumference of the leg with the stocking on (C_1), and the elastic gradient (k) for the material and knit configuration of the stocking, we can calculate the compressive pressure (P_0). The standard formula for a thin wall pressure container is:

$$F=k(C_1-C_2),\ P_od=2F,\ d=\frac{C_1}{\pi}$$

$$P_od=2k(C_1-C_2),\ P_o=2k\ \frac{(C_1-C_2)}{\frac{C_1}{\pi}}$$

$$P_o=2k\pi\left(\frac{C_1-C_2}{C_1}\right)$$

Because the cross section of the leg is not a circle, these pressures will be modified: somewhat higher pressures will be applied to areas with a smaller radius of curvature, and slightly lower pressures to those with a larger radius of curvature.

A study has been initiated to develop simple and easily recorded physiological measures of the effects of compressive forces applied to the upper and lower segments of the leg. Measurable hemodynamic alterations induced by

raising the pressure applied to the venous tree of the leg may enable us to relate the pressures applied by elastic hosiery to changes in cardiovascular parameters.

The basic aspect of this study involves a search for an appropriate measure of compression effects. The underlying assumption is that substantial forces applied to the lower limb affect the venous return and hence, measurable effects might well be found in pulse rate, arterial pressure, and cardiac output. In the current phase of this study, pulse rate and pedis dorsalis, popliteal, and brachial arterial pressures as well as EKG signals are monitored on a series of normal subjects while the ambient forces on the lower extremities are altered: external pressures ranging from 1.0 to 1.2 atmospheres are applied to the leg which is simultaneously subjected to axial forces ranging from 0.0 g. to 1.0 g. Gravitational effects are altered by means of a tilt table, and atmospheric pressure is adjusted by means of a pneumatic pressure sleeve.

An identical study will be undertaken with wearers of elastic hosiery.

B. Development (Components)

1. Adjustable Below-Knee Pylon for Temporary and Permanent Prostheses. With the widespread use of temporary prostheses, clinicians are permitting patients to use these devices on an out-patient basis with increasing frequency. The temporary prosthesis should therefore be readily convertible to a cosmetically adequate permanent prosthesis.

Based on the early assumption that the prosthesis was "temporary"; i.e., it was very soon to be replaced by a "permanent" prosthesis, the requirements for these devices were simple. Low-weight and simple-to-use alignment adjustability were desired. In addition, for the immediate post-operative prosthesis, a convenient disconnect feature was required to permit the shank and foot section to be detached from the socket during bed rest and replaced without the loss of the original alignment.

However, the tenability of the original assumption is questionable. The line between "temporary" and "permanent" use is not well defined; the curve of increasing patient mobility during training is smooth and there are no lines of demarcation in *performance* to indicate a specific time of readiness for permanent, as opposed to temporary use. Generally, stump condition indicates the readiness of the patient for a permanent socket; when shrinkage is complete and the stump volume and contour are stabilized, little additional socket modification will be required.

Devices for use as both temporary and permanent prostheses must be designed for:

a. greater structural strength and durability,
b. simple alignment adjustability,
c. independence of alignment adjustments,

d. provision for incorporating the alignment device in a permanent prosthesis while permitting subsequent alignment adjustments,

e. cosmetic adequacy.

This evolution in thought has led to the development of a below-knee adjustable pylon designed to meet the needs of patients in the period immediately after fitting in surgery and for "permanent" use as well.

The below-knee system (Fig. 1) is designed for convenient attachment to any type of socket. If socket replacement is anticipated, the straps should be used for atachment while for permanent use, the socket may be glued or laminated to the block on the alignment device. The unit provides *independent*, self-locking linear adjustments in the anterior-posterior and the medial-lateral directions. It also permits separately controlled angular adjustments in the anterior-posterior and medial-lateral planes, up to 10 deg. The pylon and SACH foot are attached to the alignment section by an extremely rigid connection consisting of mating dovetailed sections which are easily disconnected by means of a single screw.

The foot attachment, permanently bonded to the pylon, provides for additional anterior-posterior linear adjustment. The pylon is also designed to receive and retain a cosmetic cover.

The pylon, the foot attachment, and the basic alignment system are also used in the above-knee device described below.

2. *Multiplex AK Pylon.* The design of a versatile pylon for above-knee prostheses has been refined by simplifying the method of installing various swing-control units, development of a specially adapted mechanical friction swing-control unit, and the design of a suitable cosmetic cover.

The upper attachment has been redesigned to receive the knee mechanisms. To preserve the varying geometrical relationships among the knee center and the upper and lower attachments of each hydraulic knee mechanism, a single adapter for each unit has been designed to position the lower attachment properly. The interchange of knee-control mechanisms simply requires the installation of the new unit at the same upper attachment point as the old, and the replacement of the old lower attachment adapter by a new one.

The design changes made to date have improved the ease of interchanging the several hydraulic units but have increased the difficulty of installing the Northwestern Disk Friction Unit. It was found more expedient to design a mechanical, adjustable variable friction knee unit which can be more readily accommodated in the pylon. This unit is still experimental.

A new cosmetic cover has been developed for both below-knee and above-knee use. Fabricated of lightweight flexible vinyl, the new cover is neutral in shape for use on either right or left side. This cover, which fits all SACH feet, is maintained in position by means of a distal sleeve through which the foot attachment passes (Fig. 2).

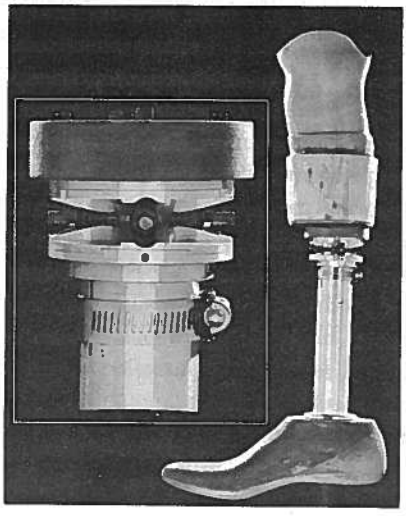

FIGURE 1. Experimental BK adjustable pylon in setup with socket, shank block, and SACH Foot. Inset shows multiple socket attachment screws and three of the four angular alignment adjustment screws.

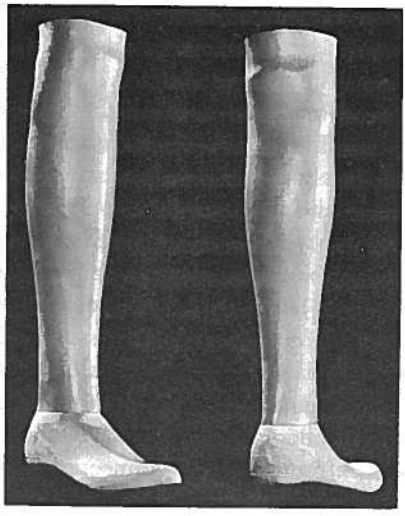

FIGURE 2. New vinyl cosmetic cover for the experimental Multiplex AK pylon. Neuter shape permits use on either right or left side, and the attachment to the foot has been improved.

3. *Instrumented Pylon.* Gait parameters and the effects on patient performance of prosthetic fitting, alignment, and functional components are most accurately and objectively studied by means of photometric and electronic instruments. For many years, force plates and interrupted light cameras have been used for this purpose since they provide data on many of the forces interacting between the foot and the ground in stance phase and on the position of the major lower-extremity joints in both swing and stance phase. Despite the extremely valuable data provided by these systems, a great deal of work is involved in: a. preparing the equipment, b. targeting patients, and c. making multiple runs to assure that the entire foot is on the force plate during stance phase. There are also difficulties attendant on correlating photographic and electronic data. While these problems are inescapable in obtaining these data, we rarely require the full range of information. For example, only axial loads and knee moments may be required for studies involving the use of above-knee swing-control devices or perhaps, torque data alone may be required as a basis for the development of a torque-absorbing component. To provide greater selectivity, to simplify the collection and reduction procedures, and to reduce our dependence on the fixed walkway and force plate, we have developed an instrumented pylon which provides many of the outputs previously

obtained by the present systems. It is basically similar to a device developed at the University of California some years ago.

This unit (Fig. 3), designed for installation in the shank of either a below-knee or above-knee prosthesis, can be used on any walking surface. Thus, it is unnecessary to require the subject to strike a fixed plate properly, permitting studies to be performed in different locations. The pylon system can be used with all standard amplifying and recording equipment.

Twenty-four strain gages are arranged in circuits to respond to axial load, torque, and both anterior-posterior and medial-lateral shear. Knee moment is determined from the differences in the magnitudes of gage outputs at three levels on the pylon, yielding the slope of the moment curve along the axis of the pylon. By electronic extrapolation of the slope, knee moment is attained.

This device has been especially designed for use in a series of studies to determine the effects on gait of SACH foot heel wedges of varying density, anterior-posterior and medial-lateral placement of the foot with respect to the shank, knee joint alignment, and socket weight-bearing characteristics. The instrumented pylon will also be used in the Seattle [a] studies to record the daily change in weight-bearing pattern of patients fitted immediately after surgery.

4. Torque Absorber. In a series of tests the previously designed torque absorbing device (BPR 10–3, Spring 1965, p. 109) performed reasonably well, but its structural strength was inadequate and it was too bulky. The unit has been redesigned to employ the principle of the "Lamiflex" bearing (Fig. 4), a commercial product of the Marlin-Rockwell Co.

Twelve Lamiflex torque absorbers were constructed to the following specifications: a. each unit must develop a resistance torque around the axis of the assembly of 50 in.-lb.$\pm$10 percent at an angular displacement of 15 deg. either side of center with approximate linearity in this property as a function of the angle; b. the complete assembly must withstand a pressure of 1100 p.s.i. at its outer edge without failure of the bonding at any point.

These specifications were drawn from biomechanical data obtained on both normal and pathological gait.

After mechanical tests are completed to determine their compliance with the specified physical properties, these units will be installed in prostheses worn by several patients.

C. Development (Techniques)

1. Pneumatic Casting. Producing a negative form for a socket mold by external-pressure-casting techniques has been the object of several studies in

[a] VA Prosthetics Research Study, Seattle, Washington.

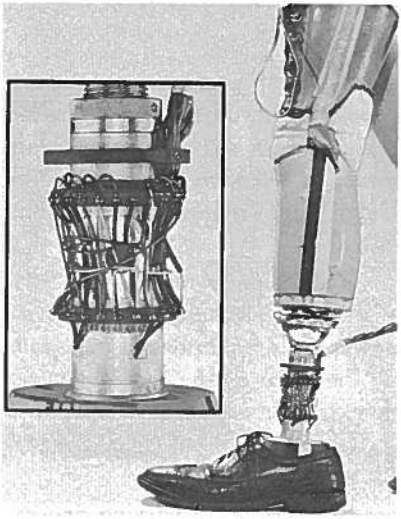

FIGURE 3. Instrumented pylon is installed as shank of a BK prosthesis and aligned by means of VAPC coupling. Inset shows wiring details: black disk in center is a rosette strain gage arrangement which is sensitive to torque; on the lower part of the pylon are shown two temperature compensating gages.

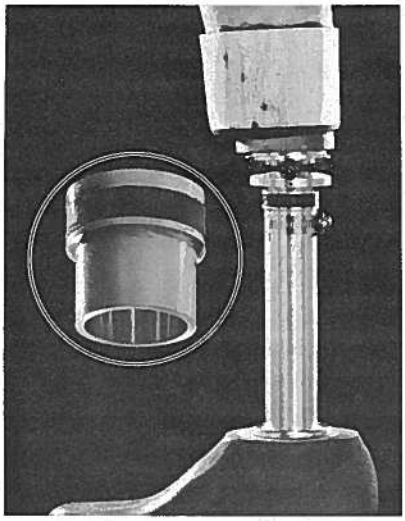

FIGURE 4. Torque absorber installed in the shank of a BK prosthesis below the BK alignment device. In the inset, the bearing material "Lamiflex," consisting of alternate layers or resilient rubber sheets and resistant metal sheets, is shown sandwiched between an upper and lower plate.

the past. Among others, Mr. Paul Leimkuehler of Cleveland, Ohio, devised a vacuum system for this purpose; hydraulic methods for pressure casting were developed by Mr. Colin McLaurin while with the VA project at Northwestern University, and by Mr. Theodore C. Meyer of Detroit, Michigan.

For various reasons these techniques were never fully developed. In searching for a method of controlling stump casting pressures, VAPC reviewed these experiences and developed a pneumatic casting procedure (BPR 10–3 Spring 1965).

During the past year, 70 male below-knee amputees have been fitted with prostheses made from casts obtained by the pneumatic pressure system. In all, 77 casts have been made for these 70 patients, including 7 bilaterals, a fairly typical sample of veteran amputees.

The subjects ranged from 20 to 85 years of age, with 55 of them between 41 and 70 years of age. Height ranged from 5 ft. 2 in. to 6 ft. 2 in. with 62 of the 70 between 5 ft. 7 in. and 5 ft. 11 in. tall. Fifty-nine subjects weighed between 161 and 200 lb. while ten weighed between 120 and 160 lb. and one weighed 245 lb.

Thirty-three patients had amputations due to accidents or gunshot wounds, 33 due to vascular pathology, and 4 to osteomyelitis.

Occupations ranged from professionals to unskilled laborers. However, more than half the sample were found to be clerks (13), skilled workers (10), or retired (16).

The 44 previous prostheses wearers, primarily World War II veterans, included 2 World War I veterans. Of the 70 subjects, 38 became amputees between 1961 and 1965.

Fifty-five stumps were between 5 and 7 in. long, 9 were between 3 and 4 in. long, and 13 were longer than 7 in. The sample is described in Table 1.

A description of the amputees' previous prostheses is given in Table 2. Thirty-four limbs had PTB sockets, 12 of which used cuff suspension with the remaining having lacer suspensions. Twenty-six cases, representing 31 stumps, had no prostheses previously.

TABLE 1.—*Summary Description of Patients*

(Seven Bilateral Amputees Included in Sample)
N=70

Range data			Etiology			Stump length, in. N=77		
Age, yrs.	Height, in.	Weight, lb.	Wounds or accidents	Vascular	Osteomyelitis	3–4	5–7	8–10
20–85	62–74	120–245	33	33	4	9	55	13

TABLE 2.—*Description of Previous Prostheses*
N=77

No Previous Prosthesis	31
Cuff Suspension, PTB, Closed End, Liner	8
Cuff Suspension, PTB, Hard, Soft End	3
Cuff Suspension, Hard, Open End	1
Lacer, Side Joints, Hard, Soft End, PTB	1
Lacer, Side Joints, Hard, Open End (Non PTB wood)	12
Lacer, Side Joints, Soft, Closed End, PTB	3
Lacer, Side Joints, Soft, Open End, PTB	18
Total	77

Conventional alignment techniques were employed in fitting these patients. Either an adjustable coupling or one of the currently available below-knee temporary pylons was used during alignment and fitting trials. The pressure casting apparatus, described in BPR 10–3 Spring 1965 was employed and

stumps were cast over three below-knee cast socks with the outer one forming part of the cast. A casting pressure of 2 p.s.i. was maintained for ten minutes. Forty-two of the sockets were used with typical PTB (cuff) suspension and 35 with conventional below-knee (lacer) suspension, as shown in Table 3.

It was quite clear that the pneumatic pressure casting technique produced highly satisfactory casts and sockets. They were well fitted and comfortable. The preponderant majority of sockets required little or no significant modifications.

As shown in Table 4, all the plaster socket molds required buildups ranging from ¼ to ⅜ in. on the distal end. However, of 77 sockets, 62 required no further modification, 12 were modified by additional relief for the end of the tibia, and 3 required other corrections.

TABLE 3.—*Description of New Protheses*

(All PTB Sockets)
N=77

Cuff Suspension, Closed End, Liner	3	
Cuff Suspension, Hard, Soft End	7	
Cuff Suspension, Hard, Open End	28	
Cuff Suspension, Soft, Open End	4	
Total Cuff Suspension		42
Lacer, Side Joints, Hard, Soft End	0	
Lacer, Side Joints, Hard, Open End	4	
Lacer, Side Joints, Soft, Open End	29	
Lacer, Side Joints, Soft, Closed End, Liner	2	
Total Lacer Suspension		35
Grand Total		77

TABLE 4.—*Mold Modifications and Socket Corrections*

Plaster Mold Modifications	Socket Corrections
17 Molds: ¼ in. build-up end of tibia. 60 Molds: ⅜ in. build-up end of tibia.	12 Sockets: Relief for end of tibia. 2 Sockets: Relief for medial condyle. 1 Socket: Relief for medial hamstring. 62 Sockets: Required no corrections.
Total 77	Total 77

Evaluating these experiences, we find the pneumatic pressure casting method offers several advantages over more conventional methods. This technique permits the selective application of higher pressures to pressure-tolerant stump areas and lower pressures to the soft fleshy areas. More accurate replication is possible through reduction of sizing errors; different prosthetists will produce almost identical casts of the same stump after they have been familiarized with the technique. The casting method can be applied to fabricate a variety of sockets: all kinds of sockets including hard or soft, open end or closed end, and temporary plaster sockets can be fabricated over molds from stump impressions made by the pneumatic system. Relatively inexperienced technicians readily learn the procedures, and extensive mold modifications (plaster removal, plaster build-ups) are not required.

We intend to use the pneumatic method routinely for casting below-knee stumps while continuing our efforts to improve the technique and the equipment.

2. *Direct Forming of Below-Knee Sockets.* Experimental work has continued in an attempt to identify a dimensionally stable material which can be formed into sockets directly on the stump. Eight materials have been evaluated for this purpose, including several polyesters, acrylics, and polyethylenes. They were found to be unsuitable in that either excessively high molding temperatures were required or their characteristic "memories" did not permit them to retain the required contour.

A new material, the most promising to date, is under study. "Orthoplast," manufactured by the Johnson and Johnson Company, is a thermoplastic vinyl. Heated in an oven at 200 deg. F., the material is extremely pliable and capable of being molded very intimately over irregular surfaces. Heat is given off very rapidly and the material can be applied to the stockinet-covered stump within a minute after removal from the oven without causing undue discomfort to the patient. However, working time is relatively short, about two or three minutes.

Although "Orthoplast" is not yet available in cones for stump casting, experimental applications are being carried on with cones fabricated by hand from flat sheets.

D. Evaluation (Components)

UCB Pneumatic AK Swing Control Knee. A revised model (Fig. 5) of the pneumatic swing-control unit developed by the VA project at the University of California at Berkeley Biomechanics Laboratory is currently under evaluation. The unit is designed to provide smooth control of swing phase.

The operative element is an air piston/cylinder unit similar in function to a pneumatic door closer. Flexion of the prosthetic knee causes the pivot

located at the rear of the knee joint to move distally in the shank forcing the piston into the cylinder. This downward motion of the piston results in compression of the air below the piston and the creation of a partial vacuum above it. The pressure difference thus formed provides resistance to knee flexion. An adjustable by-pass flow of air from the pressure side of the piston to the vacuum side also modifies the resistance pattern. Extension of the knee from a flexed position results in similar resistance to extension as the pressure/vacuum relationship on the two faces of the piston and the controlled air flow are reversed.

The seven units received for evaluation will be subjected to a series of mechanical tests to determine resistance characteristics and durability. Also to be studied are the basic alignment and fitting requirements and the adequacy of installation instructions. Biomechanical analyses of performance will be undertaken on three subjects to evaluate the following claims made by the developer:

a. The setup allows more than 120 deg. of flexion which is considered optimum.
b. The resistance provided by the pneumatic knee is proportional to the lower ranges of walking speed, but due to its nonlinearity, effectiveness at higher speeds of walking is decreased.
c. The pneumatic unit offers, in addition to its damping effects, a positive kicker action. In resisting flexion, air is compressed and some of the energy thus stored is then spent in accelerating the shank forward into extension.
d. Less maintenance is required.

E. Evaluation (Techniques)

Air-Cushion Sockets. The air-cushion patellar-tendon-bearing socket developed by E. Lyquist et al.[b] at the University of California, San Francisco-Berkeley, consists of an external, rigid, polyester-laminate shell and an internal, elastic, silicon-rubber-laminate sleeve joined at the level of the tibial tubercle. The rigid socket shell provides direct support for the stump area proximal to the level of the tibial tubercle and extends distally on the outside of the elastic sleeve to about 1 in. above the distal end of the tibia. The elastic sleeve encloses the distal part of the stump beginning at the level of the tibial tubercle. A polyester-laminate cap, bonded to the outside of the distal portion of the rigid socket shell, provides a sealed air chamber distal to the elastic sleeve.

At the recommendaton of the Committee on Prosthetics Research and Development's Fourth Workshop Panel on Lower Extremity Fitting, held

[b] Lyquist, E., Wilson, L. A., and Radcliffe, C. W.: Air-Cushion Socket for Patellar-Tendon-Bearing Below-Knee Prosthesis. Biomechanics Laboratory, University of California at San Francisco-Berkeley Technical Memorandum, April 1965.

July 1965, several air-cushion sockets were fitted at VAPC to obtain information on the fabrication requirements and application indications for the prosthesis.

To date, a total of eight air-cushion sockets have been fitted. Three were fabricated over molds by the method recommended by UCSF and five were fabricated over molds obtained by the VAPC pneumatic pressure technique.

Preliminary results indicate that the air-cushion socket for the patellar-tendon-bearing below-knee prosthesis requires considerably more time to fabricate than does the conventional PTB socket. Moreover, socket modifications required after an initial period of wear, can only be completed at considerable cost in time and materials.

To warrant the increased complexity of fabrication and the additional cost in labor, the air-cushion socket should provide a level of comfort of function which is superior to conventional prostheses. Our experience to date provides no evidence to support the superiority of the air-cushion socket, at least on the type of stumps fitted at VAPC.

Of the eight patients fitted with the air-cushion socket, two subjects rejected it completely because of an inability to tolerate end bearing and three subjects are wearing their legs on a part-time basis for the same reason. The remaining three subjects continue to wear the air-cushion sockets full-time. We shall provide continued assistance to the developer by additional fittings as indicated.

II. UPPER-EXTREMITY PROSTHETICS

A. Development

None.

B. Evaluation (Components)

1. AIPR Externally Powered Components. The program designed to evaluate the CO_2-powered components developed by the American Institute for Prosthetic Research has proceeded through step g of the evaluation program described in the previous issue (BPR 10–4 Fall 1965). In the previous step (f), the subject's right prosthesis (humeral neck) had been completely powered by the installation of a pneumatically operated terminal device and wrist rotator and a powered elbow. In step g, a powered terminal device and wrist rotator and a passive shoulder abduction joint with power lock were also installed in his left prosthesis. Thus, the subject was bilaterally fitted with the full complement of powered components.

After an initial period of controls training, the subject returned to his home station for additional training in the performance of practical daily activities.

After a three month period of wear, the subject's performance was re-evaluated in the Laboratory. The test results (Table 5) indicated that little functional improvement had taken place in the interim with respect to time required to perform a standard task, number of errors committed, or in energy expenditure. The number and type of self-care activities that the subject was capable of successfully accomplishing had not changed appreciably. The performance of the patient had apparently stabilized at the level he previously achieved when the more mobile right side was fitted with the powered components and the left side with conventional components (step f).

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

Tests	Complete power on right side only (step f)	Complete power bilaterally (step g)
Prehension		
Total time, min.	4.8	4.7
No. of errors	14	6
Appearance	Poor	Average
Positioning		
Total time, min.	7.5	6.5
No. of errors	2	3
Appearance	Good	Good
Practical activities (7 tasks)	6 completed 1 failure	6 completed 1 failure
Net energy cost of 2nd prehension test	.955 liters O_2	1.685 liters O_2
Total time, min.	4.4	4.1
No. of errors	6	10
Appearance	Average	Poor

Since the subject has reached a plateau in the training/achievement curve, we have terminated the experimental portion of this study even though we recognize that some additional improvement might be expected from re-fitting the subject with new sockets.

The patient has been scheduled for "final" clinical evaluation to determine the best possible admixture of powered and conventional components. On the basis of a thorough analysis of his experience and performance patterns, he will be refitted with new prostheses. Thereupon, a final report will be presented on this case. Some tentative conclusions on the use of AIPR powered components as well as external power generally will be delineated.

2. *Northwestern University Center Control Hook.* "Center control" hooks, designed principally for operation by external power, are aimed at improving prosthetic cosmesis by eliminating the protruding operating lever and the external control cable. The hook fingers are operated by means of a control cable which runs through the hook stud and wrist unit into the forearm of an above-elbow prosthesis or the socket of a below-elbow prosthesis for a relatively short stump. The Northwestern University Center Control Hook is simple if somewhat unconventional in design (Fig. 6). The hook employs the same pivot as the Dorrance aluminum hooks, but it is offset 1⅝ in. from the stud centerline to allow the cable to pass through the stud center. The fingers are set in approximately 25 deg. of radial deviation. Two prototypes, inexpensively made from a casting alloy, are being studied to evaluate installation and requirements, function, patient reaction to appearance, and utility.

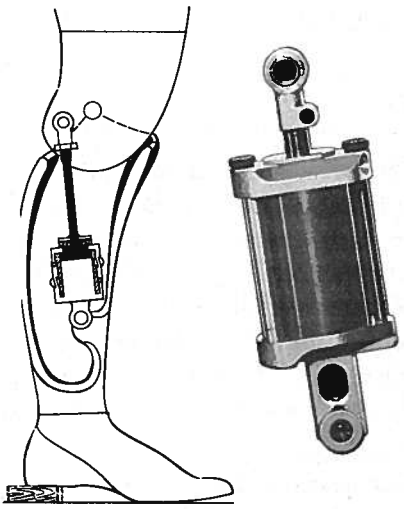

FIGURE 5. Drawing of the UCB Pneumatic AK Swing Control Knee showing the geometric relationship between knee center and points of attachment for the device. As the piston rod (solid block) moves distally when the knee flexes, air in the cylinder is compressed and driven upwards to the partially evacuated chamber above. Flow rate and thus resistance to flexion are adjusted by means of the valves. Inset shows simplicity of design.

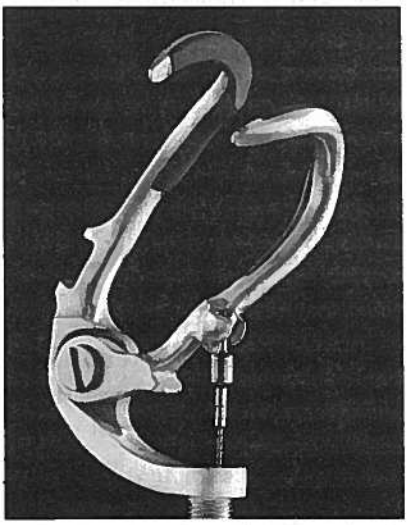

FIGURE 6. Northwestern University Center Control Hook fabricated in aluminum for test purposes. Principal features are angulation of the hook fingers, control cable attachment, and retaining slots for rubber bands.

3. Low-Friction Cable Transmission Systems. Two low-friction transmission systems, one developed by Northwestern University Prosthetic Research Center and the other by the Army Medical Biomechanical Research Laboratory, were evaluated.

The NUPRC system consists of a 1/16 in. steel cable in Teflon "spaghetti" tubing (No. 13 AWG). The cable housing and other components are classed as "heavy duty." The Teflon tubing containing the cable is inserted into the cable housing, and the ends are flared.

The AMBRL system consists of Dacron cord, Teflon housing, and nylon ball and harness connectors. Two types of nylon retainers (Specimens A and B) are available; both were tested.

A comparative analysis was undertaken to compare both AMBRL systems and the NUPRC system with a "conventional" Bowden cable transmission system. Samples of the NUPRC system, both of the AMBRL systems, and a conventional transmission system were fabricated in 18-in. lengths with 12-in. lengths of cable housing, and with harness hangers and large ball terminals.

TESTS

Three tests of tensile strength were employed. In Test No. 1, designed to identify the "weak link" in the entire system, the steel harness connector end of the cable system was attached to the lower test grip, and the large steel ball end (with ball receiver) was attached to the upper test grip. Test No. 2 was designed to measure the tensile strength of the connection between cable and ball fitting. It was performed with the ball receiver omitted and the ball attached directly to the upper test grip. In Test No. 3, used to measure the tensile strength of the cable alone, the cable was attached directly to both upper and lower test grips.

All of these tests were performed at a cross head speed of 0.2 in. per minute, the same speed employed in previous tests conducted at AMBRL.

The AMBRL, NUPRC, and conventional systems were cycled 25,000 times on the fixture shown in Figure 7 to determine durability and frictional characteristics. Durability was determined on the basis of cable wear, and frictional characteristics were determined on the basis of efficiency of force transmission.

Efficiencies of the transmission system were determined by measuring the force required to lift a 4-lb. load. The test was performed once with the cable flexed 135 deg. to simulate maximum elbow flexion and again, without curving the cable, to simulate full elbow extension. Efficiencies, calculated by dividing the load by the lift forces, were expressed in percents.

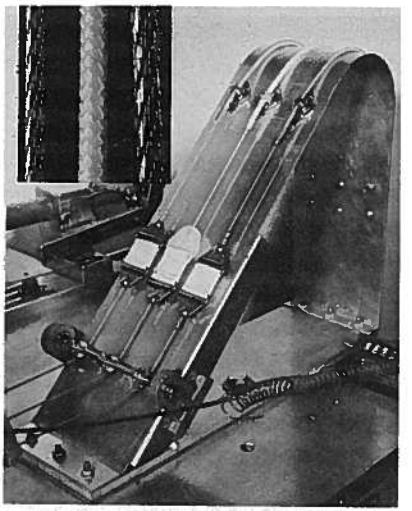

FIGURE 7. Test fixture employed to cycle the AMBRL, NUPRC, and a conventional cable control system to test durability and frictional characteristics.

TEST RESULTS

a. AMBRL Systems

1. Tensile Strength. In Test No. 1 of Specimen A, employing the entire system, the nylon ball pulled out of the steel receiver under a load of 48 lb. In Test No. 2, omitting the ball receiver, the Dacron cord cable broke at the knot within the nylon harness connector under a load of 56 lb. In Test No. 3, employing only the Dacron cord, failure occurred at the point of attachment (knot) at 63 lb.

Elongation under these tensile loads is shown below:

Test No.	Load in pounds	Elongation, percent
1	48	9.4
2	56	10.6
3	63	13.2

The efficiency of force transmission was 81 percent in a flexed position and 95 percent in the extended position.

Identical tests conducted with Specimen B gave essentially similar results, the chief difference being that failure occurred at somewhat higher loads in Tests No. 2 and No. 3 (2 percent higher in Test No. 2 and 17 percent higher in Test No. 3). Tension tests conducted by AMBRL resulted in higher strength values, 97 lb. versus 63 lb. for Specimen A and 116 lb. versus 78 lb. for Specimen B. This discrepancy can only be accounted for on the basis of variations in the test samples.

Elongation and efficiency of force transmission on Specimen B were identical with Specimen A.

2. Durability. Both AMBRL systems withstood the effect of 25,000 cycles without failure. However, as shown in Figures 8 and 9, the Teflon

FIGURE 8. Photograph of Dacron cord cable used in AMBRL system before cycling.

FIGURE 9. Photograph of Dacron cord cable at completion of 25,000 cycles.

housing showed moderate wear with consequent adherence of Teflon particles to the Dacron cord.

b. NUPRC System

1. Tensile Strength. In Test No. 1 employing the entire system, including the terminal connectors, the steel ball pulled out of the receiver under a load of 95 lb.

In Test No. 2, omitting the receiver, the cable pulled out of the swaged harness connector under a load of 310 lb. In Test No. 3, employing only the cable, failure occurred at the point of attachment to the test grip under a load of 490 lb.

Elongation under these loads was minimal, approximately 0 to 1.5 percent.

Efficiency of force transmission ranged between 78 percent in a flexed position to 93 percent in an extended position.

2. Durability. After 25,000 cycles under a 4-lb. load, the NUPRC cable caused excessive wear of the Teflon liner. Although no measurable wear of the cable occurred, the excessive wear on the liner may decrease the efficiency or the life of the system.

SUMMARY

Shown below in Table 6 is a comparison of all test results for the four configurations.

TABLE 6.—*Elongation, Tensile Strength, and Efficiency of Four Upper-Extremity Cable Transmission Systems*

System	Elongation, percent (approx.)	Tensile Strength, lb.			Efficiency, percent	
		Test No. 1, Total system	Test No. 2, Without ball receiver	Test No. 3, Cable only	Flex.	Ext.
Conventional (Standard) 3/64 in. steel cable, standard housing, heavy duty steel components.	0–1.5	95 Steel ball pulled out of receiver.	255 Cable pulled out of swaged harness connector ball fitting.	340 Cable broke at point of attachment to lower test grip.	71	90
NUPRC 1/16 in. steel cable, Teflon spaghetti tubing No. 13 AWG, heavy duty steel components.	0–1.5	95 Steel ball pulled out of receiver.	310 Cable pulled out of swaged harness connector ball fitting.	490 Cable broke at point of attachment to lower test grip.	78	93
AMBRL: Specimen A Dacron cord (sand and green colored), Teflon housing, nylon components.	9–13	48 Nylon ball pulled out of steel receiver.	56 Dacron cord broke at knot within nylon harness connector.	63 Cord broke at point of attachment (knot).	81	95
AMBRL: Specimen B Dacron cord (sand colored), Teflon housing, nylon components.	9–13	48 Nylon ball pulled out of steel receiver.	60 Dacron cord broke at knot within nylon harness connector.	78 Cord broke at point of attachment (knot).	81	95

In cyclic tests of wear, all four systems withstood 25,000 cycles under 4-lb. loads without failure.

The AMBRL system appears to be of inadequate strength for normal adult amputee use but it may be adequate for the child amputee. Although the efficiency of the Dacron cord in the Teflon housing was somewhat higher than steel in Teflon housings, the chief deficiency in the AMBRL system is the excessive elongation under load. The ease of assembly by simply knotting the cord may be advantageous but it does not overcome the deficiencies of low tensile strength and excessive elongation.

The NUPRC system is quite comparable to the conventional transmission system providing greater resistance to tension force primarily as a consequence of the greater cable diameter, $\frac{1}{16}$ in. versus $\frac{3}{64}$ in. Although it is approximately 10 percent more efficient than the conventional system, excessive wear of the Teflon liner may affect its life.

C. Evaluation (Techniques)

None.

III. LOWER-EXTREMITY ORTHOTICS

A. Development

Friction Ankle for Drop-Foot Brace. Designed to prevent the foot from dropping during swing phase, this device (Fig. 10), described in BPR 10–4 Fall 1965, employs friction disks at the ankle joints. The friction ankle joints are mounted distally in a contoured polyester laminate shoe insert designed to improve the cosmetic appearance of the brace by eliminating external stirrups. The proximal sections of the joints are mounted in polyester uprights. The insert can be shaped to influence foot balance.

In an experimental unit worn by a subject for four weeks, the laminate fractured at a point below the ankle joint as a result of the high resistance requirements. The friction setting loosened continually with wear and it required frequent readjustment. However, the patient considered the resistance superior to several other appliances he had previously worn.

We have, therefore, redesigned the unit to improve the friction resistance and the resistance of the laminate to fatigue.

B. Evaluation

Foot Supports. In continuation of our studies of the effects on both comfort and foot balance of various types of arch and foot supports, three subjects are each being fitted with three different appliances: a. leather Schaefer plates, b. Silverman heel stabilizer which is a shoe insert designed to prevent foot imbalance, and c. a dynamically contoured (weight bearing) plantar mold. Patient reactions are periodically recorded and objective data on foot balance and weight application to the plantar surface of the foot are being obtained by means of the barograph and motion pictures.

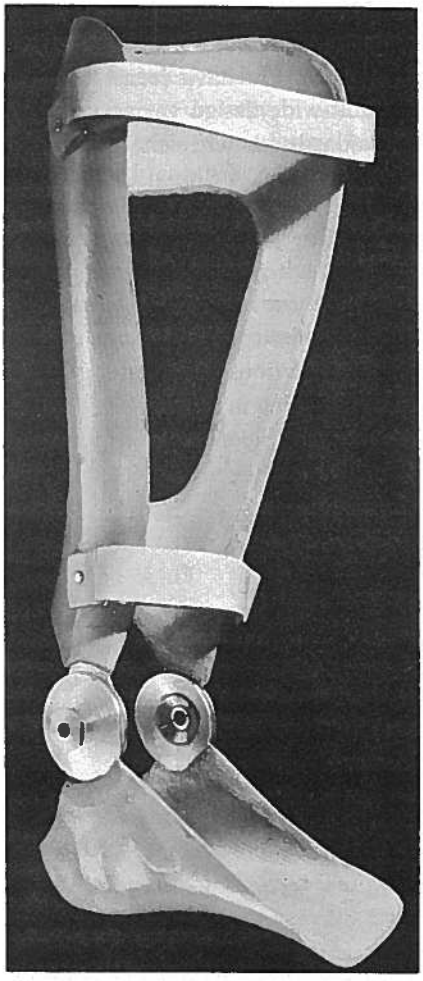

FIGURE 10. An earlier model of the friction ankle for drop-foot braces showing details of friction disks.

IV. ORTHOPEDIC AIDS

A. Development

Functional Standards for Wheelchairs. A great deal of interest has recently been generated in the design and development of wheelchairs, particularly as in the use of new materials such as lighter metals. As a consequence, we have seen during the last few years several new models featuring lightweight aluminum or stainless steel frames, solid polycarbonate casters, and a variety of synthetic upholstery materials. Existing VA Specification (7043400b) have become increasingly outdated and their utility in evaluating new wheelchair designs has decreased correspondingly. Based on our experiences in testing and evaluating approximately 12 new wheel-

chair designs, we are recommending revision of the current VA specifications in order to recognize the increasing use of new materials and to facilitate the evaluation of function and durability. We have recast them in the form of *functional requirements* to permit wide design variations while meeting VA standards. The tentative standards for wheelchairs are currently being evaluated by the chiefs of a number of VA Spinal Cord Injury Services and other interested individuals.

B. Evaluation (Components)

1. Hydro-Crutch Model FP 2000–1. This device is currently being redesigned by the developer to eliminate the deficiencies revealed by the evaluation program (BPR 10–4 Fall 1965). Negotiations are under way to purchase the improved devices for clinical field testing in several VA stations.

Following reevaluation of the new models at the Bioengineering Research Service, they will be sent to several stations where their utility for a variety of patients will be observed.

2. American Wheelchair. An evaluation program was completed on three modernized models of the American Wheelchair. These new chairs were distinguished by four unconventional features: a. 8-in. Lexan polycarbonate casters, b. plastic skirtguards, c. plastic leg rest panels, and d. readily detachable padded plastic upholstery.

Despite its departure from existing VA specifications, particularly as regards materials and the dimensions of certain components, these chairs met the functional requirements implicit in our proposed specifications. The Lexan wheels withstood the durability and impact tests but observation of long-term use may be required to be certain of their adequacy. Certain apparent deficiencies noted in this evaluation have been brought to the attention of the developer. Moreover, we are now planning to participate (with the Research and Development Division of PSAS) in a controlled field evaluation of a number of new and old wheelchair designs. This clinical study will assist us in the further development of the functional standards (described above) as well as in determining field reaction to the several designs.

3. Gendron Lightweight Wheelchair (Model #8515–15). Two lightweight wheelchairs were reevaluated to determine the effectiveness of modifications made by the manufacturer to strengthen the chair as a result of earlier testing which had revealed structural weaknesses. The revised models met all the specifications relating to strength and durability. However, several other apparent deficiencies were noted during clinical tests in which the chairs were used by patients in two VA stations. These matters have been brought to the attention of the manufacturer together with recommendations for improvement.

4. Everest and Jennings Mono-Drive Wheelchair. One model of a motorized wheelchair (Fig. 11) manufactured by the E & J Corporation of Los Angeles, California, was evaluated. The units are powered by a 12-volt storage battery. The battery drives a Parvalux DC motor with a speed reducer. The motor which drives the single powered wheel by means of a chain is controlled by a handle containing switches in series to alter both motor polarity and voltage. The handle is mounted (left or right) on a steering column which provides directional control. Axial twist of the handgrip mounted on the steering column provides two forward and one reverse speeds; turns are made by rotating the handle in the horizontal plane toward the right for right turns and to the left for left turns.

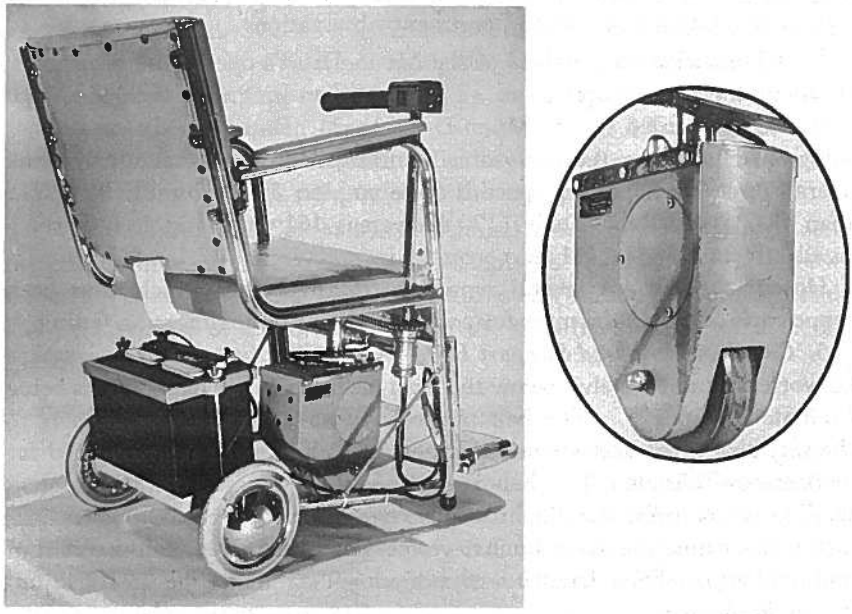

FIGURE 11. The principal features of Everest & Jennings Mono-Drive Wheelchair are the side-mounted control, chain drive, single forward drive wheel, and 12-volt storage battery.

The chair is of cantilever construction with a solid foam padded seat and back. It can be disassembled for storage. A battery charger is supplied with the unit.

Evaluation of this chair consisted of the following:

a. A comparison between the Mono-Drive and the Power Drive chair, another E & J wheelchair powered by two 6-volt batteries which drive two DC Parvalux motors linked to the wheel by belts. This comparison was drawn in terms of the criteria employed for Automotive Wheelchairs, VAPC, July 1, 1957, listed as follows:

(1) Speed—Must be adjustable to a low of 1 m.p.h.

(2) Obstacle Climb—Must be able to mount and cross a 1-in. obstacle.

(3) Ramp Climb—Must ascend and descend 10 percent slope at safe speeds.

(4) Dimensions—Must conform to VA Specification No. 7043400b for Wheelchair, Folding.

(5) Maneuverable—Must be easily maneuvered.

(6) Design—Motor units and battery carriers should not be permanently mounted to the chair.

(7) Power Supply—Power supply must be sufficient to operate for a normal day's use and be rechargeable overnight.

(8) Other—Any other pertinent observations.

b. A biomechanical analysis of the Mono–Drive's operation with respect to safety and ease of operation. The evaluation program revealed several advantages afforded by the Mono-Drive chair. Entry or exit from either side is facilitated by the removable armrests. Despite its relatively small overall dimensions, seating space is quite ample. It is substantially lighter than the Power Drive chair (120 lb. versus 161 lb.), and it requires a smaller turning space (41 in. versus 54 in.).

However, there are several significant disadvantages which have been reported to the manufacturer for correction prior to qualification testing.

5. Cetrone Contoured Support Belt. A sacroiliac belt that is designed to be worn around the pelvis below the level of the anterior iliac spines is being evaluated (Fig. 12). The belt purports to afford maximum support to the sacroiliac joint and according to the developer, eliminates the need for cumbersome corsets. The belt is designed to apply strong lateral forces tending to compress the ilia. In so doing, tension is reduced on the ligaments supporting the lower lumbar vertebrae and those between the sacrum and the ilium. These ligaments, the developer states, are the sites of injury in low back pain.

The device features two anti-slipping pads which prevent the belt from riding up and reducing its effectiveness. The belt, constructed of top grain cowhide leather with a two-pronged buckle, comes in three sizes. At present, two subjects have been fitted with the Contoured Support Belt.

C. Evaluation (Techniques)

None.

V. TESTING

A. Specifications Compliance

1. SACH Feet. During this period, the annual test program for molded SACH feet was initiated. To date, tests have been completed on SACH feet submitted by the Otto Bock and Kingsley Companies.

The products of both these manufacturers met the minimum standards set forth in current VA specifications. Although the Kingsley SACH Foot met the functional standards to the fullest extent, X-ray examination revealed that the belting is attached to the keel by means of a profusion of badly placed staples. While having little immediate consequence on the results of the functional tests, this condition may reduce the life of the product.

The Bock SACH Foot met the specifications requirements despite several marginal minor discrepancies. Flexure resistance was very close to the high resistance limits in one size 8 foot, and in another size 8 foot, heel compression was minimally adequate.

2. *Stump Socks.* Five samples of wool leg stump socks were tested for compliance with specifications in accordance with our interim check program scheduled midway between our annual compliance tests. All samples of stump socks submitted by the Bennington, Accurate, McCann, Ohio Willow Wood, and Knit Rite companies were tested in accordance with procedures listed in Invitation No. 66–1 dated April 14, 1965. All samples conformed to current requirements indicating an improvement in the general quality of stump sock production in this country.

3. *Upper-Extremity Components.* Annual compliance tests were conducted on three upper-extremity components: a. Sierra Internal Elbow, b. Sierra Voluntary Closing Hook, and c. Sierra Voluntary Opening Hand. The performance of all three units met the requirements set forth in the specifications.

B. Materials

1. *Luperco Catalysts.* The catalyst most commonly used in polyester lamination techniques is Luperco ATC, a product of Wallace & Tiernan, Inc. ATC is a benzoyl peroxide paste compounded with tricresyl phosphate. Two difficulties attendant on the use of ATC are that it separates (Fig. 13) and that it cannot be shipped by mail under existing fire hazard regulations.

More recently, Wallace & Tiernan produced two benzoyl peroxides with somewhat different formulations. The first, Luperco ANS–50, is a non-separating form of benzoyl peroxide dispersed in butylbenzoyl phthalate and the second, Luperco AFR, is a fire resistant and non-separating paste form of benzoyl peroxide dispersed in an inert organic plasticizer. This special feature of AFR allows small quantities to be sent through the mail without an I.C.C. yellow label.

The three Luperco pastes, products of Lucidol Division, Wallace & Tiernan, Inc., 1740 Military Road, Buffalo, New York 14240, were evaluated to determine curing patterns, skin reactivity, and general utility.

To determine the effects of the catalysts uninfluenced by factors related to reinforcing materials, pigments, and laminating techniques, a two-phase study was based on comparisons of samples with these factors controlled. In

Phase A, gel times were recorded for potted samples of resins with varying concentrations of catalyst and promoter. In Phase B, polyester laminates were made over a tapered wood mandrel, 5 in. in diameter at one end, tapering to 3 in. at the other end, and 12 in. long with 4 nylon stockinets, and gel time was recorded.

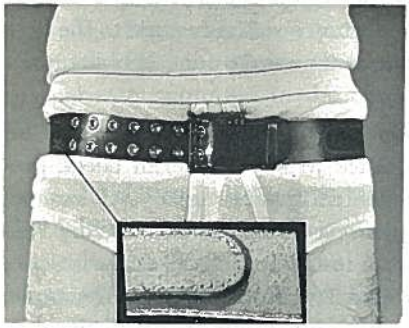

FIGURE 12. The Cetrone Contoured Support Belt, designed for relief of low back pain, worn by patient with the internal contoured sections applying lateral force to the depression below the iliac crest. Inset shows details of the "contour," a section of leather sewn on the inside of the belt.

FIGURE 13. A jar of Luperco ATC showing the typical separation of constituents which takes place during storage.

The controlling factors included:

a. The use of Laminac No. 4110 in all samples.

b. Only concentrations of 1, 2, and 3 percent of each test catalyst (ATC, ANS–50, AFR) were used.

c. Each set of tests in Phase A and B was done once with 8 drops of promoter No. 3 and once again with 16 drops of promoter No. 3.

d. The same experimenter performed all tests.

The results of these analyses are summarized below:

a. *Gel Time.* Potted samples (Phase A) of 1 percent ATC, ANS–50, and AFR gelled respectively in 99 minutes, 126 minutes, and 172 minutes with 8 drops of promoter. However, when the concentration of catalyst was increased to 2 percent, both the ATC and ANS–50 gelled in 76 minutes, while AFR required only 52 minutes under the same conditions. At a 3 percent concentration, gel times were 50 minutes for ATC, 30 minutes for ANS–50, and 35 minutes for AFR. In all potted samples, doubling the amount of promoter to 16 drops halved the gel time.

In Phase B, employing polyester laminates and 8 drops of promoter, similar results were obtained. At a 1 percent concentration of catalyst, gel times were 70 minutes for ATC, 111 minutes for ANS–50, and 149 min-

utes for AFR. At a 2 percent concentration, the ATC required 38 minutes, the ANS–50 required 47 minutes, and AFR required 63 minutes to gel. These differences were reduced to insignificance when a 3 percent concentration of catalyst was used; gel times were 24 minutes for ATC, 33 minutes for ANS–50, and 39 minutes for AFR. Doubling the amount of promoter to 16 drops cut all three gel times by approximately 60 percent.

b. *Skin Reactions.* Ten normal subjects participated in patch tests to observe skin reaction to intimate contact with polyester laminates made with nylon stockinet, 2 percent concentration of each of the three types of catalyst, and 8 drops of promoter No. 3. Three disks, each made with different catalysts, were taped to the volar aspect of the forearm. After 24 hours the patches were removed; all reactions were negative.

The following summarizes the relative advantages and disadvantages of each catalyst in laminates employing 3 percent concentrations of catalyst, 8 drops of promoter No. 3, and nylon stockinet:

Catalyst	Gel time, minutes	Homogeneity	Relative shipping costs
ATC	24 Good	Poor, separating	High.
ANS–50	33 Fair	Good, non-separating	High.
AFR	39 Fair	Good, non-separating	Low.

The sole advantage of ATC is its relatively rapid gel time under normal working conditions. However, the fact that it separates may reduce its advantage. ANS–50 does not separate but it requires more time to gel than ATC. Neither of these two catalysts meets the I.C.C. requirements for mail shipment and therefore procurement costs are somewhat higher. AFR on the other hand does not separate, and it can be shipped through the mails.

Although it requires proportionately more time to gel than the other two products, used in concentrations of 2 or 3 percent gel time differences are negligible.

If gel time is not a factor, AFR seems to be the catalyst of choice. The selection of the other catalysts on the basis of gel time must be balanced by shipping costs and time delay in working. Attempts to reduce gel time of AFR by the simple expedient of increasing the number of drops of promoter is not recommended because of the possibility of discoloration.

2. *Polypropylene Reinforcing Fabric For Lamination.* In general prosthetics-orthotics practice, nylon stockinet is replacing Fiberglas as the reinforcing material for polyester lamination. While the laminates obtained with nylon stockinet and polyester resin are adequate as regards strength-

for-weight characteristics, lighter laminates would be preferred. Fiberglas epoxy laminations are now considered to have the highest strength-weight ratios, but their use in prosthetics and orthotics has been limited by some objectionable characteristics of the constituent materials.

Polypropylene is a thermoplastic polymer obtained from propylene gas, a hydrocarbon by-product of the alkylate refining process of gasoline. Polypropylene fiber is produced by The Vectra Company, Odenton, Maryland.

This material is being evaluated by comparisons with nylon and Fiberglas as reinforcing materials. Although polypropylene fabric has not previously been used in prosthetic applications, it appears to be superior to Fiberglas as the reinforcing material in molded boats.

Sample polyester laminates will be prepared under as nearly identical conditions as possible. One-third of them will employ nylon stockinet, one-third Fiberglas, and one-third polypropylene. Another group of samples will be laminated with epoxy resins, again one-third nylon reinforced, one-third Fiberglas, and one-third polypropylene. The six types of specimens will be compared in terms of process requirements, weight, workability, and resistances to tension, bending, impact, and fatigue.

VI. OPERATIONS REPORT FOR FIRST HALF, FISCAL YEAR 1966

The data in this section reflect the prosthetic and orthotic services provided for veteran beneficiaries during the first half of Fiscal Year (F.Y.) 1966 by the VAPC's Orthopedic Shoe Service and Prosthetics-Orthotics Service.

A. The Orthopedic Shoe Service

As shown in Table 7, an increase of 136 in the number of beneficiaries on the rolls of the Orthopedic Shoe Service had no appreciable effect on the total workload.

However, the issuance of protective footwear (rubbers or overshoes) during the first half of F.Y. 1966 increased markedly. Fifty-eight pairs of rubbers and 103 pairs of overshoes were issued at a cost of $3,360.

In the first half of F.Y. 1966, eighty-eight worn clinic lasts were duplicated at a cost of $836.

B. The Prosthetics-Orthotics Service

The distribution of surgical supports and elastic hosiery is the responsibility of the Orthotic Components Unit of the Prefabricated Appliances Section, Prosthetics-Orthotics Service. Tables 8 and 9 show that the issuance of these items remains relatively constant.

TABLE 7.—*Numbers and Costs of Shoes in VAPC National Orthopedic Shoe Program*

	Fiscal Year				
	1962	1963	1964	1965	First Half F.Y. 1966
Benef. on rolls	9, 344	9, 681	10, 021	10, 322	10, 458
New shoes, prs.	7, 465	7, 249	7, 317	7, 336	3, 622
Prs. of new shoes issued per benef. on rolls per yr.	. 80	. 75	. 73	. 71	[a] . 69
Cost, $	394, 222	387, 757	440, 536	442, 641	198, 892
Repaired shoes, prs.	8, 660	9, 716	9, 487	9, 422	5, 286
Prs. of shoes repaired per benef. on rolls per yr.	. 93	1. 00+	. 95	. 91	[a] 1. 01
Cost, $	92, 514	97, 778	93, 520	130, 916	55, 143
Total cost, $	486, 736	485, 535	534, 056	573, 557	254, 035

[a] Projected for full fiscal year.

As reported in a previous bulletin (BPR 10–3 Spring 1965), the average cost fluctuation is due primarily to changes in the internal accounting procedures.

During the first half of F.Y. 1966, the Prosthetics Components Unit of the Prefabricated Appliances Section distributed 47 additional Hydra-Cadence Above-Knee Systems. The number of beneficiaries receiving this unit since the inception of the distribution program is 1,302.

During the report period, 460 Hydra-Cadence (H-C) units were replaced. In addition, 354 H-C cosmetic covers and 75 H-C feet were replaced. The manufacturer repaired 154 units at no cost; 273 units were repaired and converted at a cost of $40,373. VAPC technicians did minor repairs to 108 units at a cost of $1,060.

The issuance of 68 Mauch HYDRAULIK swing-control units during the report period has brought the total number of wearers of this device to 152. In the same period 150 DuPaCo units were issued increasing the total number of wearers to 256. Twenty-seven Mauch and 27 DuPaCo units were replaced during the report period; many of these were units that had been issued during the clinical application study. DuPaCo repaired 10 units at a cost of $309. Nineteen Mauch units were repaired at a cost of $1,183.

VA Orthopedic Shops received 16 above-knee and 12 below-knee temporary prosthesis units. Seven above-knee and 9 below-knee temporary prosthesis units were sent to commercial limb facilities. Forty-seven SACH

TABLE 8.—*Numbers and Costs in the VAPC Surgical Support Distribution Program*

Issuance Channel	F.Y. 1963		F.Y. 1964		F.Y. 1965		First Half F.Y. 1966	
	No.	Cost ($)	No.	Cost ($)	No.	Cost ($)	No.	Cost ($)
Individuals	2, 162	11, 568	2, 434	12, 901	2, 482	16, 691	1, 234	6, 127
VA Orthopedic Shops	3, 098	14, 239	3, 995	18, 091	3, 722	18, 033	3, 815	9, 118
Total	5, 260	25, 807	6, 429	30, 992	6, 204	34, 724	5, 049	15, 245
Average Cost, $		4. 89		4. 82		5. 60		3. 03

TABLE 9.—*Numbers and Costs in the VAPC Elastic Hosiery Distribution Program*

Issuance Channel	F.Y. 1963		F.Y. 1964		F.Y. 1965		First Half F.Y. 1966	
	No.	Cost ($)	No.	Cost ($)	No.	Cost ($)	No.	Cost ($)
Individuals	5, 806	33, 385	9, 989	47, 460	9, 554	62, 672	4, 212	20, 851
VA Orthopedic Shops	6, 598	37, 938	12, 258	55, 130	11, 049	56, 950	6, 929	30, 277
Total	12, 404	71, 323	22, 247	102, 590	20, 603	119, 622	11, 141	51, 128
Average cost, $		5. 75		4. 61		5. 80		4. 59

feet were issued to VA Orthopedic Shops and 14 to commercial limb facilities for use in temporary prostheses.

During the first half of F.Y. 1966, the Limb and Brace Section delivered 36 above-knee prostheses; twenty-one of these utilized total contact sockets. Only 19 of 68 below-knee prostheses delivered were of the patellar-tendon-bearing (PTB) type. In this same period, 13 above-knee and 16 below-knee temporary prostheses were delivered. Twenty-five upper-extremity protheses were fabricated and delivered to beneficiaries.

Orthotists of the section deliverd 33 leg-thigh and 58 leg braces in addition to 28 spinal braces and 815 arch supports.

The VAPC Clinic Team held 29 meetings during the first half of F.Y. 1966. The 68 beneficiaries seen were referred by 13 field stations.

C. Special Service for Vietnamese Wounded

In the Fall of 1965, 56 paralyzed Vietnamese veterans were brought to the Veterans Administration Hospital at Castle Point, New York for medical treatment. Shortly thereafter the VA Prosthetics Center was authorized to provide the orthotic services necessary to rehabilitate these men.

On November 23, 1965 selected staff members attended a special clinic meeting held at the VAH, Castle Point, N.Y. The group consisted of Anthony Staros, Director, VAPC; Dr. Edward Peizer, Chief, Bioengineering Research Service; William J. McIlmurray, Chief, Prosthetics-Orthotics Service; Dominick Bonarrigo, Acting Chief, Orthopedic Shoe Service; and Werner Greenbaum, Chief, Limb and Brace Section, P-O Service. The clinic was conducted by Dr. Emilio Ejercito, Chief, PM&R Service of VAH, Castle Point with consultation by Dr. Alfred Ebel, Chief, PM&R Service of the VAH, Bronx, N.Y.

At this session, it was decided to concentrate first on the relatively simple cases and expedite their rehabilitation. Ten cases were selected on the criteria that minimal bracing would make early ambulation possible. Prescriptions were developed for these men at this time. Of the 10, 2 would wear braces unilaterally; the other 8 would be bilateral brace wearers. Most of the components used in the braces were prefabricated; the bars were of aluminum. These braces were delivered before the end of December 1965. Typical braces are shown in the accompanying photographs (Fig. 14, 15, and 16). General characteristics and diagnoses as taken from the charts of these 10 men and the braces made are given in Tables 10, 11, and 12.

FIGURE 14. Case No. 8 wearing lateral bar leg braces.

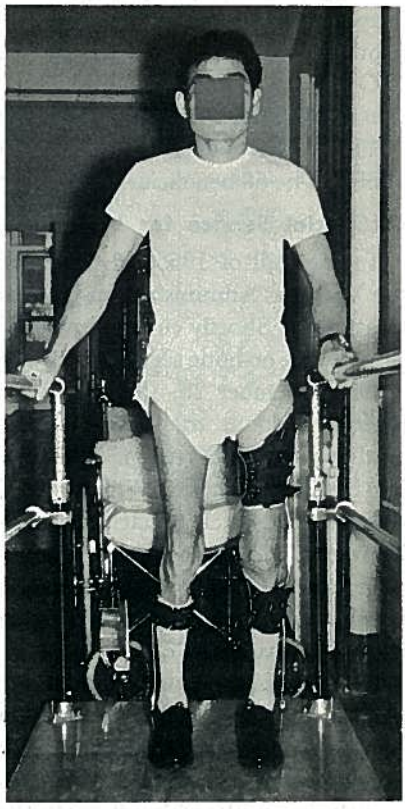

FIGURE 15. Case No. 34 wearing a right lateral bar leg brace and a left lateral bar leg-thigh brace.

FIGURE 16. Case No. 43 wearing conventional medial-lateral bar leg-thigh braces.

TABLE 10.—*General Characteristics of Paraplegic Vietnamese Subjects*

Subj. No.	Age	Date of injury	Marital status	Height	Weight, lb.	Occupation	Service rank
2	34	8/26/63	M	5′0″	95	Merchant	Corporal
4	35	6/11/64	M	5′1″	119	Farmer	Corporal
6	29	12/26/64	S	4′10″	88	Student	Sergeant
8	22	9/23/65	M	5′0″	95	Laundryman	Pvt., 2/c
34	28	12/15/64	M	5′1″	100	Bus Driver	Private
36	29	11/ 7/64	S	5′5″	90	Student	2nd Lt.
39	38	11/11/64	M	5′0″	126½	Farmer	Sergeant
43	24	5/18/64	M	5′0″	103½	Farmer	Private
45	21	4/19/64	S	5′7″	88	Painter	Private
49	33	1/31/65	M	5′4″	119	Farmer	Corporal

TABLE 11.—*Diagnoses of Paralyzed Vietnamese Subjects*

Subj. No.	Diagnosis
2	Myeloneuropathy, chronic, traumatic, probably due to injury of L–4, clinically complete motor wise below L–3 and sensory wise below L–3, and the posterior columns in T–12 for touch and pain. Has loss of function of both lower extremities and of sphincters of bowel and bladder.
4	Myeloneuropathy, chronic, traumatic, due to fracture of C–6, clinically incomplete, with signs of a Brown Sequard's lesion and radicolopathy of the left upper extremity, manifested by good motor function of the left lower extremity by about 50% and loss of function of the left upper extremity, 100%.
6	Myeloneuropathy, chronic, traumatic, due to fracture of L–1–2, clinically complete at this level with sacral sparing manifested by loss of function of left lower extremities estimated about 50% loss of function of the right lower extremities, 100% below the knee, and loss of function of sphincters of bowel and bladder.
8	Myeloneuropathy, chronic, traumatic, due presumably to cord concussion approximately around L–3 without obvious bony injuries and manifested by partial cauda equina lesion between L–3 and S–2 as demonstrated by lower motor neuron signs, by the upper motor neuron lesion below S–2, loss of function of both lower extremities, 100% at ankles and below the ankles and 100% loss of function of sphincter of bowel and bladder.
34	Myeloneuropathy, chronic, traumatic, with probable injury to L–1 clinically below L–1 on the left and L–2 on the right and loss of function of both lower extremities, 100% for the left, 100% for the right at and below the ankle and approximately 50% for the right knee and hip and 100% loss of sphincters of bowel and bladder.
36	Myeloneuropathy, chronic, traumatic, due to injury of L–1 and L–2, clinically complete at L–2 with loss of sensation of both lower extremities and of sphincters of bowel and bladder.
39	1. Cranial-Cerebral injury, due to bullet wound, right temporal region with subsequent left hemiparesis. 2. Corneal opacification, O.D. 3. Bilateral pterygium.

TABLE 11.—*Diagnoses of Paralyzed Vietnamese Subjects*—Continued

Subj. No.	Diagnosis
43	Myelopathy, chronic, traumatic, due to fracture of L–1. Clinically incomplete. Loss of function in the lower extremities, changes in segments below L–3 on the left and less so on the right. Motor impairment extends probably to S–1 but seems to be absent below that level: in brief, this is a partial cauda equina lesion.
45	1. Myeloneuropathy, chronic, traumatic, due to fracture of L–2, clinically incomplete at that level motor wise, but complete sensory wise with loss of function of both lower extremities and of sphincters of bowel and bladder. 2. Penile-scrotal fistula. 3. Bladder stones.
49	Myeloneuropathy, chronic, traumatic, due to bullet wound in L–3 and L–4, clinically complete below L–2, on the left and T–12–L–1 on the right with loss of function of the lower extremities and of sphincters of bowel and bladder.

TABLE 12.—*Orthopedic Appliance Prescriptions*

Subj. No.	Prescription
2	Left leg brace with single lateral bar and spring-loaded ankle joint. Right leg brace with single lateral bar and spring-loaded ankle joint.
4	Left leg brace with medial and lateral bars and positive equinus stop.
6	Left leg brace with single lateral bar and spring-loaded ankle joint. Right leg brace with single lateral bar and spring-loaded ankle joint.
8	Left leg brace with single lateral bar and spring-loaded ankle joint. Right leg brace with single lateral bar and spring-loaded ankle joint.
34	Left leg-thigh brace with single lateral bar with ring lock and solid stop ankle joint. Right leg brace with single lateral bar and solid stop ankle joint.
36	Left leg-thigh brace with medial and lateral bars with ring lock and positive equinus stop. Right leg-thigh brace with medial and lateral bars with ring lock and positive equinus stop.
39	Left leg brace with single medial bar and positive equinus stop.
43	Left leg-thigh brace with medial and lateral bars with ring lock and spring-loaded ankle. Right leg-thigh brace with medial and lateral bars with ring lock and free-motion ankle.
45	Left leg-thigh brace with medial and lateral bars with ring lock and spring-loaded ankle. Right leg-thigh brace with medial and lateral bars with ring lock and spring-loaded ankle.
49	Left leg brace with single medial bar, lateral "T" strap and spring-loaded ankle. Right leg brace with single medial bar, lateral "T" strap and spring-loaded ankle.

On several subsequent visits made to the hospital before the end of the year, prescriptions were developed for braces for an additional 14 men. The prescriptions were prepared by Dr. Ejercito by consultation with Mr. McIlmurray or Mr. Greenbaum.