4. Cetrone Contoured Support Belt
5. Seat Lifts
6. Proreco Curb-Climbing Wheelchair Assembly
7. Evaluation of Prototype Staircane
8. Preliminary Evaluation of Joe J. Stump Socks
9. Knight Spinal and Taylor Brace with Detachable Coverings and Aprons

C. Evaluation (Techniques)
None.

V. TESTING

A. Specifications Compliance
1. SACH Feet
2. Stump Socks
3. Upper-Extremity Components
4. Lumbo-Sacral Corset Material

VI. OPERATIONS REPORT FOR SECOND HALF, FISCAL YEAR 1966

A. The Orthopedic Shoe Service
B. The Prosthetics-Orthotics Service
C. Special Service for Vietnamese Wounded

REPORT

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I. LOWER-EXTREMITY PROSTHETICS

A. Basic Studies

1. Effects of Compression of the Lower Extremity. A two-part study has been undertaken to develop functional specifications for elastic hosiery in order to provide an objective basis for their design, prescription, and procurement. In one portion of this study, samples of commercially available elastic hosiery were evaluated with respect to the pressures which they apply to the limbs for which they are prescribed. At the present time, neither the magnitudes nor the gradients of the pressures applied by elastic hosiery are known. The rated or assumed pressures are only estimated from the dimensions, knit geometry, and ratio of elastic to non-elastic elements in the hose.

Compressive forces applied by four commercially available samples of elastic hosiery have been measured by means of the standard formula for

a thin-walled container (BPR 10–5). Pressure magnitudes and gradients were calculated for each of four samples prescribed for a single normal subject (Fig. 1). Any one of these four samples might have been prescribed for this subject based on the length and circumferential measurements of his leg.

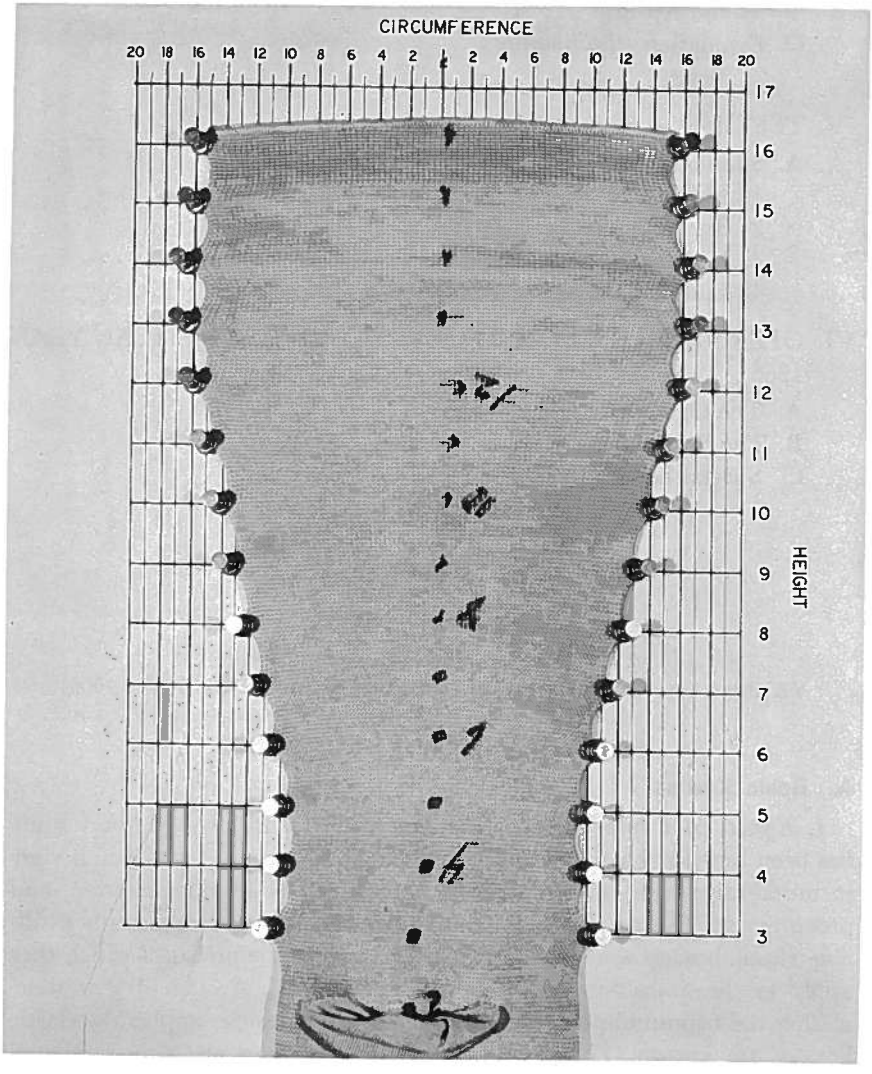

FIGURE 1. — Sample of elastic hosiery stretched on profile board to approximate circumferential distension when worn by patient for whom it was prescribed. Each pin along one side is removed and tension is measured at that level by means of a force gage. These data are used to calculate the pressure gradient applied.

As shown in Figure 2, the most striking result was the extreme variability in both magnitudes and gradients of pressure applied by each of these samples. Sample 1 provides approximately 1 lb. per sq. in. (70 gm./cm.^{2}) of compression in the area between the ankle and a point 3 in. (7.62 cm.) above it. This compression falls off very rapidly to approximately 3 or 4 oz. per sq. in. (13.2 or 17.7 gm./cm.^{2}) at a point 7 or 8 in. (17.78 or 20.32 cm.) above the ankle whereupon the pressure increases again to just under 1 lb. per sq. in. (70 gm./cm.^{2}) at a point between 11 and 12 in. (27.94 and 30.48 cm.) above the ankle.

Sample 2 shows a similar pattern with a substantially lower magnitude. The compression is high in the area just above the ankle, diminishes sharply over the next 2 in. (5.08 cm.), and then slowly begins to increase.

Sample 3 indicates a compression pattern in which moderate pressures are applied above the ankle and at the top 2 or 3 in. (5.08 or 7.62 cm.) of the leg while almost twice this pressure is applied half way up the leg giving the appearance of "choking."

Sample 4 shows a relatively uniform and flat pressure gradient over the length of the leg; however, the pressure almost doubles in the topmost 1 in. (2.54 cm.). The maximum pressure applied by any one of these samples does not exceed 50 mm. Hg. or approximately 1 lb. per sq. in. (70 gm./cm.^{2}).

The second portion of this study, carried on simultaneously, is an attempt to develop simple and easily recorded physiological measures of the *effects* of pressures on the upper and lower segments of the leg. The volume of the venous return to the heart influences the character of systemic blood flow with respect to cardiac output, pulse rate, and blood pressure. The extent to which compression of the venous tree in the lower extremity influences *central* venous return and hence the character of arterial blood flow is not clear. If alterations in the external pressure applied to the leg can be correlated with measures of general circulation, then pulse rate or blood pressure may accurately reflect the compressive effects of elastic hosiery. This in turn would provide a convenient index for objectively evaluating the effectiveness of elastic hosiery and perhaps offer a guide for prescription.

On the assumption that environmental factors such as atmospheric pressure and gravity affect blood flow in the lower extremities, pulse rate and pressure in the dorsalis pedis, popliteal and brachial arteries will be monitored on six normal subjects when these environmental forces and factors are varied. To date, however, only gravitational effects were studied. By means of a tilt table the six patients were placed in a vertical upright position, a horizontal position, and a position in which the head was tilted 20 deg. below the horizontal. Pulse rates and pressures were measured in each position. In the vertical position, the venous return

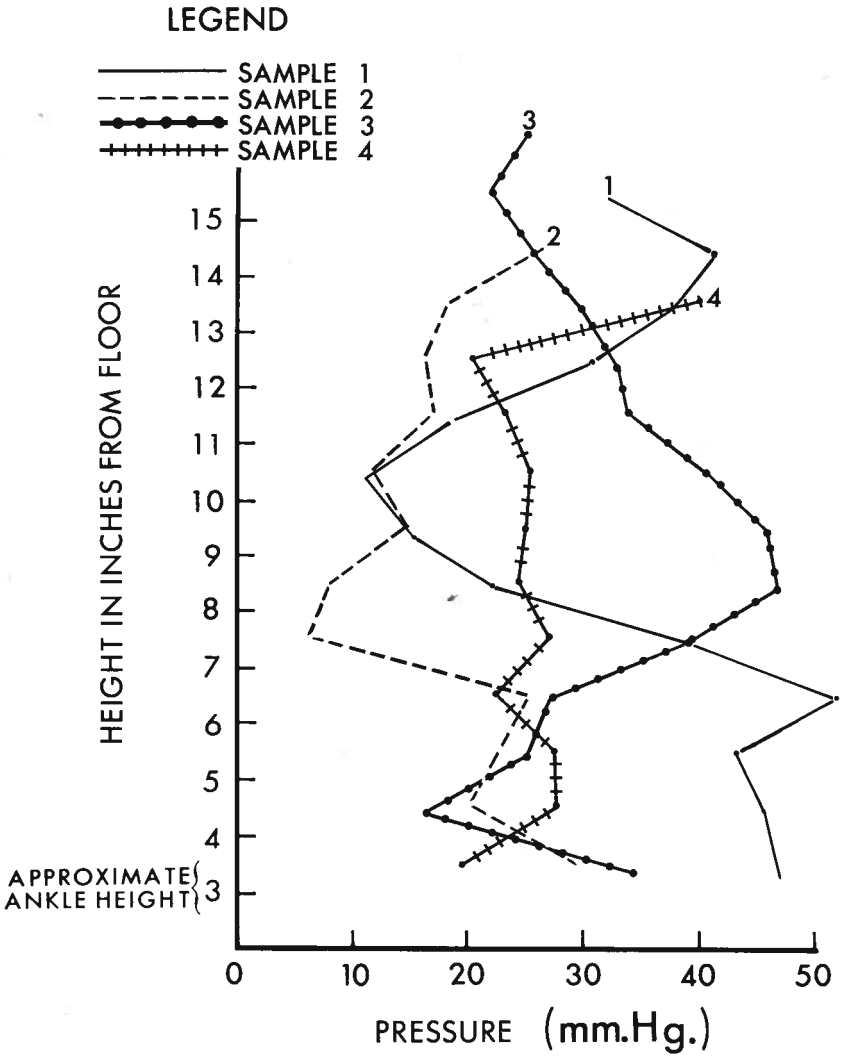

FIGURE 2. — Profiles of pressures applied by samples of elastic hosiery.

flowed against a force of 1 g. which was reduced to zero in the horizontal position. In the head down position, venous flow was *aided* by a force calculated to be 0.34 g.

It was found that altering the gravitational effect from a retarding force of 1 g. to an assistive force of 0.34 g. had no discernible effect on either systolic or diastolic pressure in the brachial arteries of the six subjects. Moderate rises were noted with increasing gravitational loads in the *systolic* pressures of the popliteal and dorsalis-pedis arteries (Fig. 3). Pulse

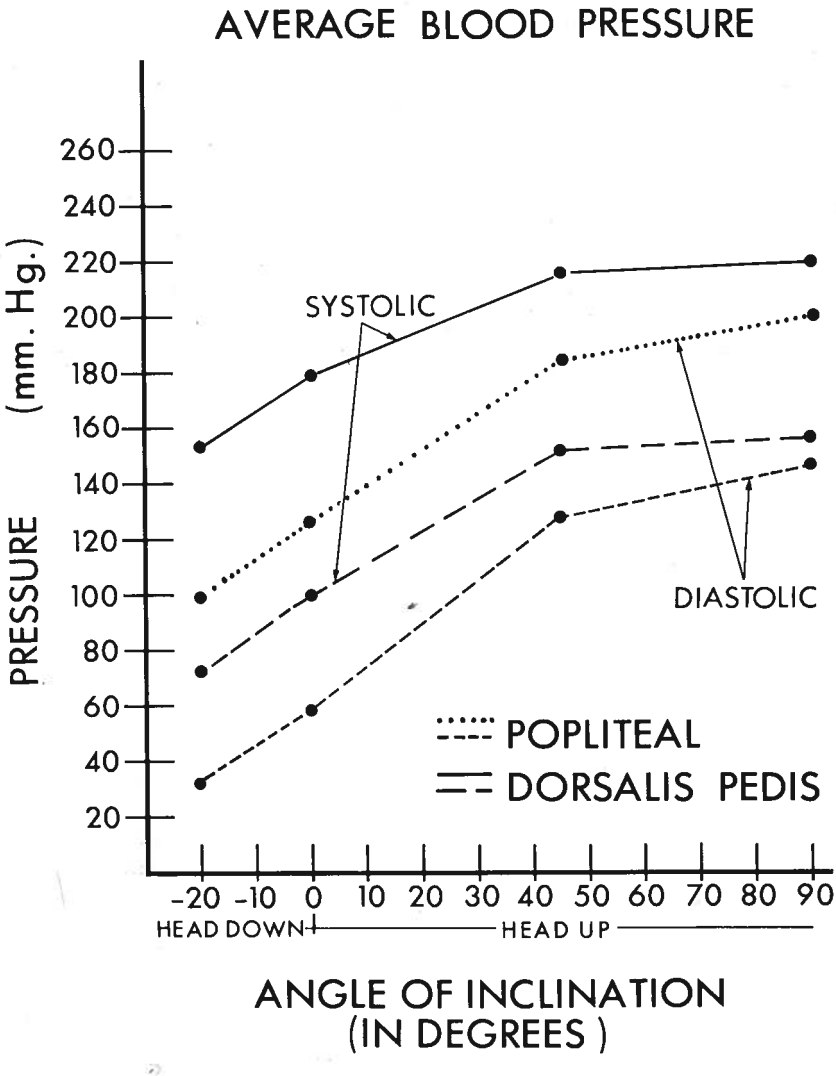

FIGURE 3

rate followed a similar pattern. However, the most significant effects were found in the *diastolic* pressures.

Diastolic pressures in the popliteal and dorsalis-pedis arteries (Fig. 3) rose as the retarding gravitational force increased. Of the two, *diastolic* pressure in the dorsalis pedis appears to be only slightly more related to gravitational changes when the actual pressures are compared. But, the diastolic values are far more sensitive indicators when considered in terms of percent of change.

These studies are being continued in an effort to obtain a valid description of the pressures applied by elastic hose and to develop reliable measures of their effects.

2. *Work and Energy in Walking.* The question of efficiency and energy expenditure in walking has received intermittent although sometimes intensive consideration. Several theories have been advanced over the years to explain various phenomena associated with the distribution of energy in walking. To date, no unified theory has been proposed which satisfactorily explains the efficiency of walking at various speeds or the energy consuming and energy conserving factors in gait. For example, when the net energy cost per meter walked is plotted against increments of velocity, the portion of the curve (Fig. 4) between speeds of 30 and 125 meters per minute (approximately 1 to 4.7 miles per hour) is surprisingly flat and seems to indicate that a range of speeds exists in which no significant increase in energy is expended per meter walked. This phenomenon suggests that somehow we adapt our walking pattern to increases in speed, at an apparent *increase* in efficiency of walking. The exact nature of these adaptive mechanisms is not clear.

It might be theorized that some postural adaptation is involved that places the body center of gravity in increasingly more effective positions for receiving an impetus. That is, the normal subject increases his trunk inclination about the hip (flexion) creating an increased "falling" moment, thus using gravitational forces to better advantage.

For this reason, a study has been undertaken to define in objective terms the relationship between energy expenditure and the mechanics of walking in cooperation with the Subcommittee on Fundamental Activities of the Committee on Prosthetics Research and Development, NAS-NRC, with the participation of Dr. Verne T. Inman and Dr. Herbert Elftman.

3. *Early Postsurgical Weight Bearing.* Among the dramatic results of early postsurgical prosthetic fitting is the fact that patients bear weight on the amputated limb within 24 hours after surgery. In some cases patients have walked between parallel bars for distances of 10 to 15 ft. (3 to 5 m.) within three days after amputation. It has become of increasing interest to determine the nature of these weight-bearing patterns with respect to such factors as axial loads, shears, torques, and moments about the knee.

An instrumented below-knee pylon (described in BPR 10–5 Spring 1966) was fitted to a 28-year-old, 134-lb. (60.91 kg.) man whose right leg was amputated below the knee by Dr. Ernest M. Burgess of Seattle, in accordance with the procedures he has described (1). The rigid dressing was applied to the patient's stump in the prescribed manner and the proximal attachment for the pylon was installed at the same time. Twenty-four hours later the drain was removed through a window in the cast and

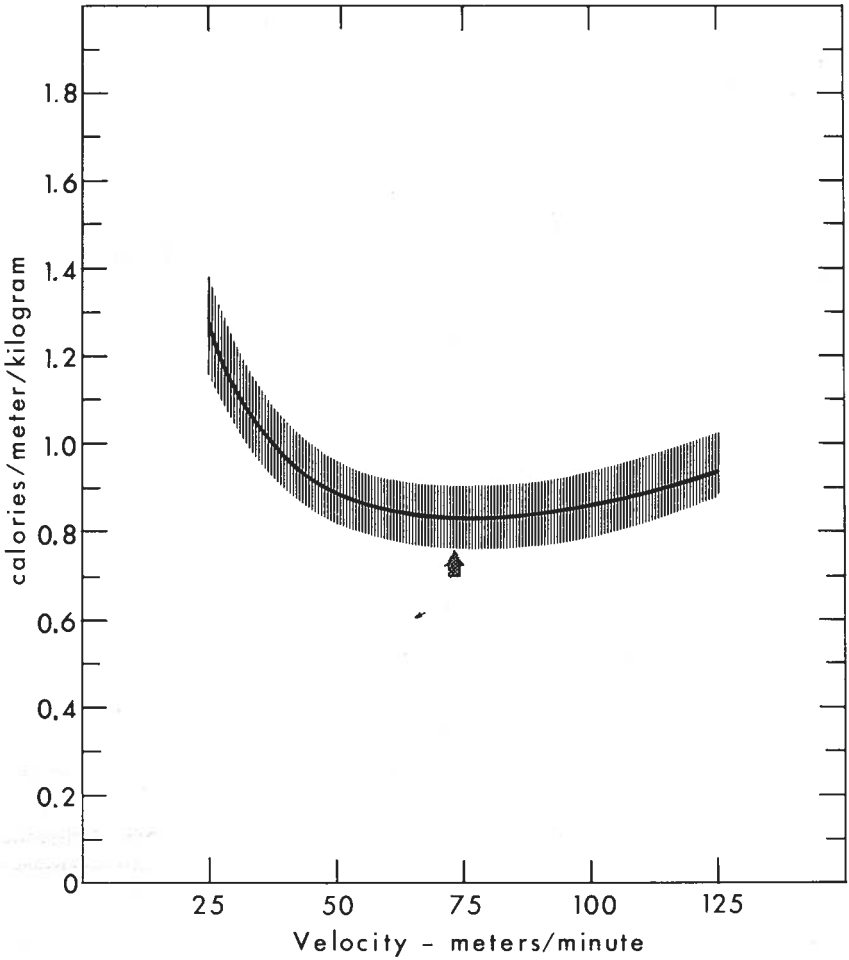

FIGURE 4. — Energy cost of normal subjects walking at different speeds. Energy is expressed in terms of gross calories used per meter per kilogram of body weight (data originally published by Biomechanics Laboratory, UCB).

the subject stood, bearing weight on the amputated limb. Twenty-six hours after surgery, the instrumented pylon was substituted for the conventional pylon in his prosthesis, and recordings were made of axial loads, torque, knee moment, and duration of the loads applied to the limb. These recordings were carried out periodically over a 76-hour postoperative period.

As shown in Table 1, maximum axial loads during this period approximated 25 percent of the body weight, an indication that the remainder was supported by the hands on the parallel bars. Negligible torques were recorded; the maximum was approximately 4 in.-lb. (4.6 cm. kg.). Typical

maximum knee moments were approximately 20 percent of those normally applied by below-knee amputees. It may also be noted that the time duration of the stance phase (from heel contact to toe-off) decreased markedly over this period indicating increased confidence and willingness to apply loads to the amputated leg approached a normal below-knee timing pattern.

TABLE 1. — *Weight Bearing on BK Amputated Limb Immediately After Surgery*

	Post op., hrs.	Activity	Axial load, % of body wt.	Torque, in.-lb.	Knee moment, in.-lb.	Step time, seconds
Patient	26	Step	21	2	40	9.8
	26	Standing	12	2	130	
	48	Step	18	3	10	7.0
	48	Standing	19	2	10	
	53	Standing	16	4	50	
	53¼	Standing	18	1	50	
	71½	1st. step	24	2	20	
	71½	Ave. step	19	2	24	6.2
	75¼	1st. step	19	2	40	
	75¼	Ave. step	25	3	60	3.9
Typical BK		Step	Approximately 100% of body weight	45 —19	322 —77	2.2

These preliminary studies indicate the feasibility of monitoring the patient's progress over a period of perhaps several weeks after surgery for diagnostic and evaluation purposes.

B. Development (Components)

1. Adjustable Below-Knee Standard Prostheses. As discussed in the previous issue (BPR 10–5 Spring 1966), the line between "temporary" or "preparatory" and "permanent" use of a prosthesis is not clearly defined particularly since the advent of early postsurgical fitting. As a consequence of early fitting, stump stabilization is achieved earlier and patient mobility improves steadily to the point where the initially fitted prosthesis is used in the same way as a permanent prosthesis. Therefore, the initially fitted device should have adequate structural strength, durability, alignment adjustability, and provision for incorporating the alignment device directly into the permanent prosthesis. Needed is a standard below-knee prosthesis

which can be fitted immediately after surgery and then conveniently converted to a permanent prosthesis by the addition of a permanent socket and cosmetic cover.

The newly developed VAPC adjustable below-knee standard prosthesis provides independent, self-locking, linear adjustments in the anterior-posterior and medial-lateral directions. It also permits separately controlled angular adjustments in the anterior-posterior and medial-lateral planes. The pylon and SACH foot are attached to the alignment device by a rigid connection consisting of mating sections readily disconnected by means of a single screw. The foot attachment, integral to the pylon, provides additional anterior-posterior linear adjustment, and the pylon is designed to retain the cosmetic cover (Fig. 5).

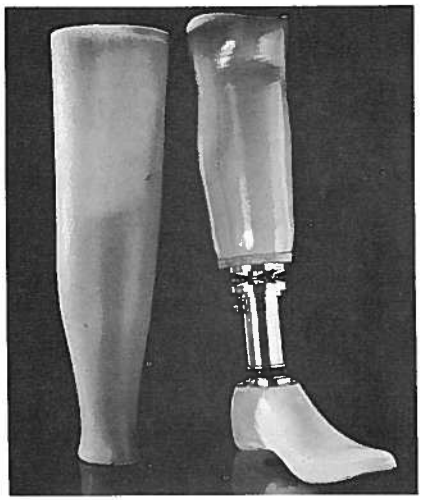

FIGURE 5. — VAPC standard adjustable BK prosthesis with alignment device, pylon, SACH foot, and cosmetic cover.

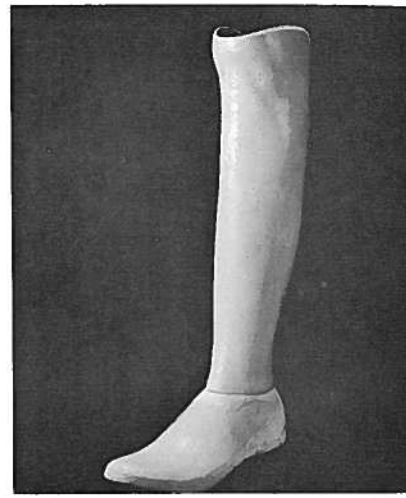

FIGURE 6. — Fitted with vinyl cosmetic cover, the standard BK prosthesis is ready for permanent use.

Vinyl cosmetic covers have been procured in three sizes. They are designed to fit soothly over the entire socket. After it is heated and stretched over the socket, the upper section of the cover maintains firm contact with the upper end of the socket (Fig. 6). The ankle section of the cover is fastened firmly to the foot by means of the foot bolt passing through a wood base plate installed in the distal end of the cover. Tightening the foot bolt compresses both the distal lip of the cover and the wood plate between the foot and the foot attachment flange of the pylon. Control of cover resiliency is achieved by varying the wall thickness. The distal end and the ankle section are thicker and less resilient than the top which is thin and flexible.

Thirty units have been manufactured for clinical evaluation. They will be distributed to cooperating field stations and research *center* facilities to obtain information on adaptability, applicability, durability, and function.

2. *Standard Above-Knee (Multiplex) Prostheses.* The design of this unit, described in the previous issue, has been refined in several ways. It has been reworked to permit the installation of four hydraulic knee mechanisms of different types. In addition, a simple adjustable mechanical friction unit has been designed.

Molds for cosmetic covers have been designed as follows: four sizes for the right and left sides with each in two ankle sizes. Lengths range from 18 to 21 in. (45.72 to 53.34 cm.) in increments of 1 in. (2.54 cm.). The top edge of the cover contains but does not restrict the motion of a specially designed neuter knee cover. The knee cover slips over the knee mechanism without disturbing the knee assembly (Fig. 7).

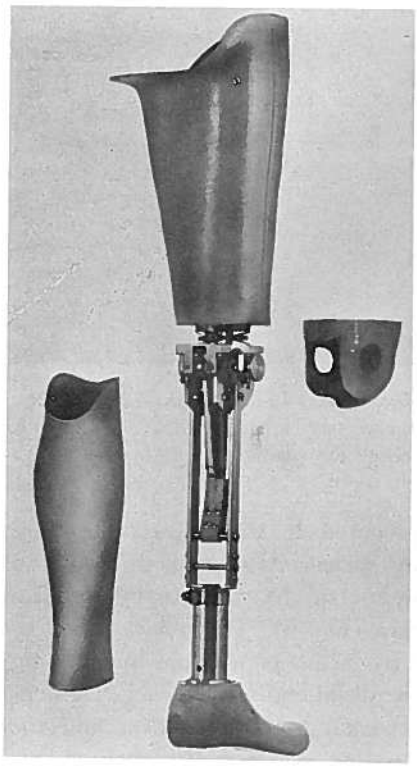

FIGURE 7. — Standard AK prosthesis with molded knee cover and cosmetic shank cover.

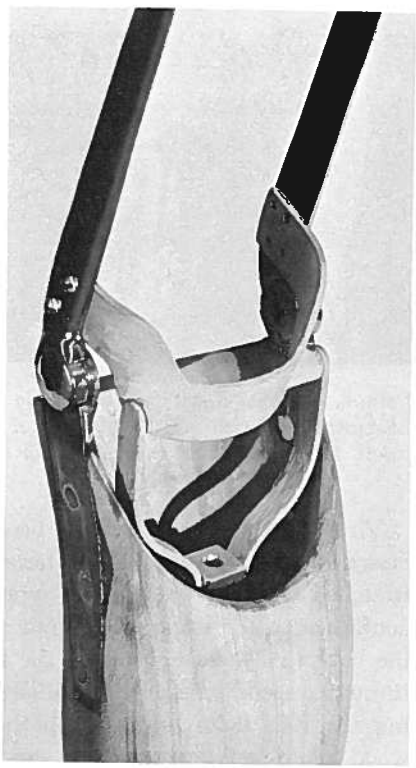

FIGURE 8. — VAPC through-knee prosthesis designed to accept fluid-controlled knee systems.

The alignment section has been redesigned to permit adjustment by means of a wrench. A large threaded plug has been made part of the upper plate of the alignment device; a mating phenolic plate permits rapid attachment of sockets.

3. Fluid Control for Through-Knee Prostheses. A prototype device for using available fluid systems in a through-knee prosthesis is under development (Fig. 8). It is anticipated that manufactured setups in three knee widths should be sufficient to accommodate most adult amputees. The yoke design encircling the posterior of the through-knee socket permits conventional attachment of the socket to the uprights. Both molded leather and plastic laminate sockets may be fitted in the conventional manner.

4. Torque Absorber. The "Lamiflex" torque absorber has been modified to reduce its dimensions. A thrust bearing and a vertical column have been incorporated into the device to improve resistance to the high bending stresses applied to the ankle. Several models are now being evaluated to determine the resistance control patterns most desirable for inclusion in production models.

C. Development (Techniques)

1. Pneumatic Casting. The VAPC pneumatic casting technique, described in BPR 10–3 Spring 1965, has been used effectively in the past year in the fitting of more than 100 below-knee amputees. Similar experience has been reported by the several other centers using this technique.

Thirty new sets of pneumatic casting equipment have been ordered for distribution to VA Orthopedic Shops to fabricate temporary prostheses. An up-to-date instruction manual is currently being prepared and will be sent with the casting kit.

2. Direct Forming of Below-Knee Sockets. We have continued our studies to develop a technique for forming a socket directly on a stump in order to reduce the number of fabrication steps and to improve fit and comfort. In all the materials tested to date, the molding temperatures have been too high, or due to the inherent "memory" characteristics the desired socket contours could not be retained, or it was extremely difficult to procure properly sized cylinders at reasonable cost.

Currently under investigation is a material whose physical properties are identical in many respects with precipitated natural balata, a relatively inelastic rubber. Without the natural esters and acids found in balata, Polysar Cast is equivalent with respect to viscosity, tensile strength, modulus of elasticity, elongation, hardness, and specific gravity. The material becomes highly pliable when immersed in water at 180 deg. F. (82 deg. C.) and preliminary tests indicate that it tends to retain its form when cool. Arrangements are being made to procure cylinders of suitable

size to continue these experiments in direct forming of lower-extremity sockets.

D. Evaluation (Components)

1. UCB Pneumatic AK Swing Control System. Seven units of this revised pneumatic swing-control knee mechanism, submitted by the Biomechanics Laboratory of the University of California Medical Center, San Francisco, California, were evaluated. Designed primarily to provide control of swing phase, the operative element is an air piston/cylinder unit. Flexion of the prosthetic knee causes the pivot located at the rear of the knee joint to move downward into the shank forcing the piston down into the cylinder. This downward motion 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 bypass 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 (Fig. 9).

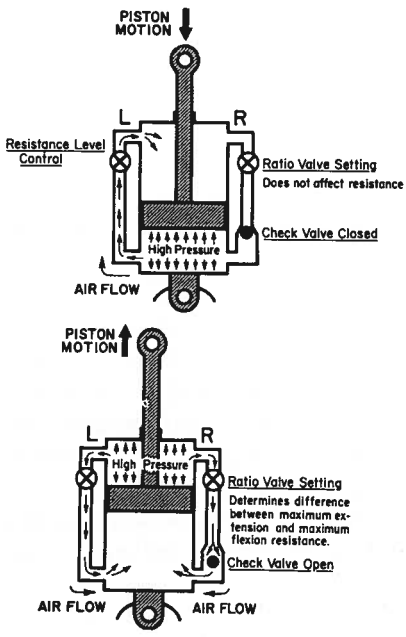

FIGURE 9. — Functional diagram of UCB Pneumatic AK Swing Control System.

Included in this evaluation was a series of mechanical tests to determine resistance characteristics, durability, basic alignment and fitting requirements, and the adequacy of installation instructions.

Mechanical tests indicated that these units permit extensive adjustment of flexion-extension resistance within practical ranges. As shown in Table 2, resistances, indicated by drop times, are quite variable in both flexion and extension. Careful manipulation of the level and ratio valves allows a useful range of independent adjustment of flexion and extension resistances comparable to other fluid mechanisms. As indicated by knee moment values (Table 3) these units offer adequate cadence response in that increases in knee moments were roughly proportional to increases in cadence. Figure 10 shows that at all three resistance settings, in each of the three cycling velocities, the resistance to flexion generally increases linearly until maximum knee flexion occurs (50 percent of cycling program). Thereafter, the resistance offered by the unit to extension increases gradually until the effect of the air "spring" tending to *extend* the knee is reduced (approximately 65 percent of cycle). In the last 10 percent of the cycle, the resistance to extension due to air compression controls terminal impact. In general, as is desired, the configuration of all the curves is similar and the magnitudes differ.

TABLE 2. — *Typical Drop Times (Sec.), UCB Pneumatic AK Swing Control System*

Unit No.	Resistance adjustment settings					
	Minimum		Intermediate		Maximum	
	Flex.	Ext.	Flex.	Ext.	Flex.	Ext.
207	.31	.31	.36	.36	.44	.58
209	.33	.33	.37	.42	.52	.78
216	.34	.34	.38	.38	.47	.67

TABLE 3. — *Maximum Knee Moment, UCB Pneumatic AK Swing Control System*

Speed, c.p.m.	Resistance adjustment settings					
	Minimum		Intermediate		Maximum	
	Flex.	Ext.	Flex.	Ext.	Flex.	Ext.
36	29 (53)	34 (39)	32 (37)	40 (46)	40 (46)	53 (61)
43	49 (56)	44 (51)	52 (60)	49 (56)	56 (64)	58 (67)
60	73 (84)	8 (3)	75 (86)	9 (10)	79 (91)	11 (13)

Note — Readings are given in in.-lb. Parenthetical figures are equivalent in metric system (cm. kg.).

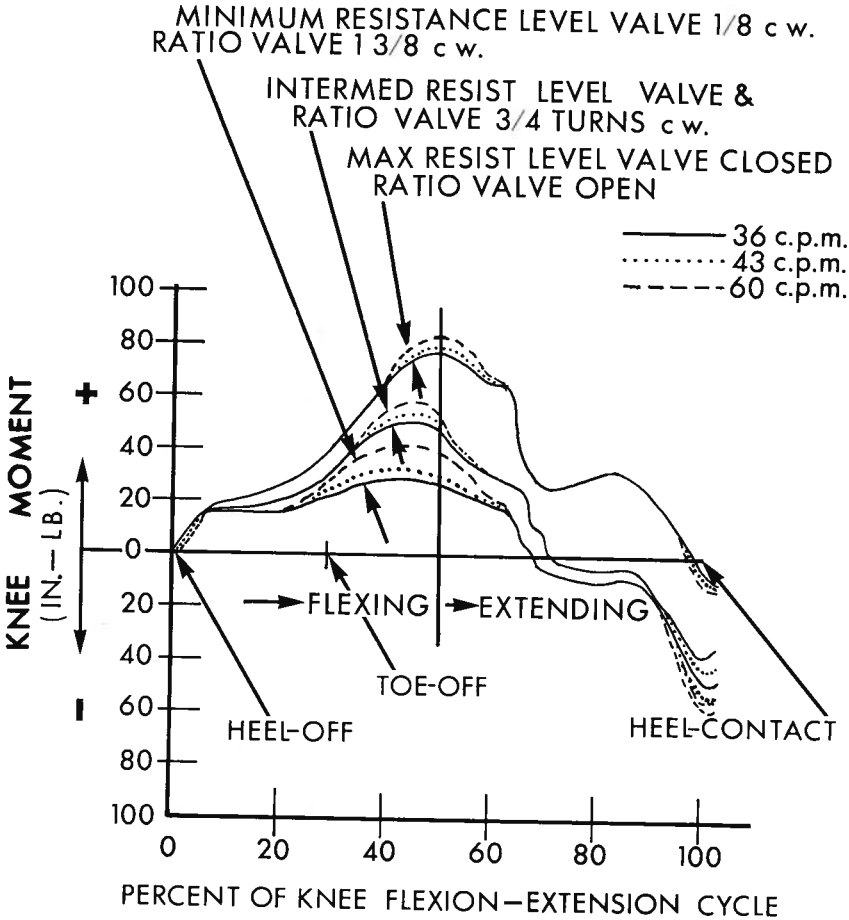

FIGURE 10. — Knee moments of UCB Pneumatic Systems at various resistance settings and cadences.

Due to the mechanical design of the knee mechanisms and the properties of pneumatic cylinders, the length of the piston rod is extremely critical. Improper adjustment of piston rod length will seriously affect these resistance values.

Five subjects have worn this unit for periods ranging from two to eight months. In all these cases amputee experience has been satisfactory. In two cases, the knee bolt loosened as a result of the nut backing off. The exact causes are difficult to define, but one significant possibility is that the knee bolt appears to be short and requires tightening of the nuts until they bear on the shank attachment straps. In one case, the distal attachments for the knee mechanism failed. Although this is the only

experience of its kind to date, it is a problem previously recognized by the developer. He has taken steps to replace the wooden shanks with cast aluminum shanks containing presumably stronger attachment structures. Fitting and alignment of this unit are otherwise quite routine.

2. *Joint for Hip-Disarticulation Prostheses (NHJ–100).* Manufactured by the A. J. Hosmer Corp., the joint (Fig. 11) provides a certain amount of adjustability in alignment by means of a spherical section in the attachment component. The spherical section permits angular displacement of 15 deg. in the plane parallel to the slot and 9 deg. in the plane perpendicular to the slot. The maximum composite angle is approximately 17.5 deg. Units of this type have not been available for fitting and therefore clinical experience could not be applied to this evaluation.

Mechanical tests were undertaken to determine the strength of the bolt in particular and the entire assembly in general as well as the maintenance of adjustment settings. A test fixture was employed to apply a range of torque loads to the unit under conditions simulating as nearly as possible actual use. A "worst" condition was simulated by assuming a 200-lb. (90.90 kg.) patient plus approximately 100 lb. (45.45 kg.) of inertial forces applied along the posterior wall of the socket (as it might at heel contact) and an alignment with the knee center located approximately 6 in. (15.24 cm.) anterior to the weight line. The spherical section was tightened with a torque wrench in increments ranging from 10 to 65 in.-lb. (11.5 to 74.75 cm. kg.). When the spherical section was tightened with an applied torque of 65 in.-lb. (74.75 cm. kg.), the unit resisted a moment of 300 ft.-lb. (41.52 m. kg.) without slipping.

On the basis of these data, the overall strength and resistance to slippage seem adequate, and any adjustment in alignment for the hip disarticulation would be a valuable asset.

3. *Otto Bock "Greissinger" Foot and Ankle Assemblies (Model 1A6) with Compressible and Non-Compressible Heel Wedges.* Both assemblies feature a five-way functional ankle identical in design except for the compressible heel wedge incorporated in one unit. The foot and the ankle assembly consist of a wood ankle section mounted in a molded plastic foot well, a wood foot with a molded plastic sole section, and a rubber ankle block recessed into the plastic foot well (Fig. 12). The foot and shank are joined by means of a "U" bolt and yoke-type assembly, the lower end of which is recessed into the foot well while the upper end is recessed into a shank well.

As there are no applicable specifications for assemblies of this type, the functional characteristics of these units were evaluated by measuring deflections in the plantar-flexion dorsiflexion range and in the eversion-inversion range under loads corresponding to those imposed by normal non-amputees. Deflections in the transverse plane under appropriate

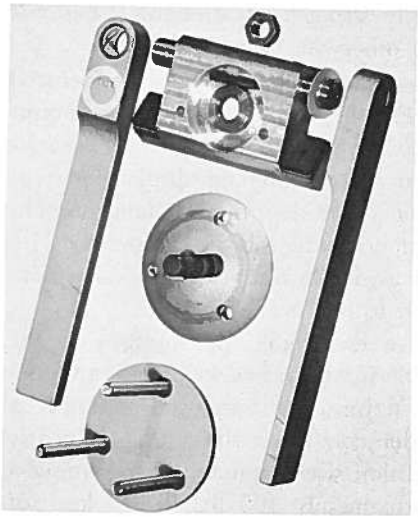

FIGURE 11. — Hosmer Canadian-type joint (NHJ 100) provides a spherical section for making fine angular adjustments in alignment.

FIGURE 12. — Otto Bock "Greissinger" Foot and Angle assembly with compressible heel wedge. Shown is the substitute foot for this assembly with non-compressible heel wedge.

torque loads and the compressibility of the heel wedge were also measured. In addition, the design and certain prosthetic factors were examined.

As shown in Table 4, the "Greissinger" Foot with the compressible heel deflects 15 deg. under the normal maximum plantar-flexion moment of 18 ft.-lb. (2.49 m. kg.). The foot with the non-compressible heel deflects similarly, but with lower magnitudes due to the absence of the influencing compressible heel. The unit produces 12.5 deg. of dorsiflexion under the normal maximum dorsiflexion moment of 80 ft.-lb. (11.07 m. kg.). Both plantar-flexion and dorsiflexion characteristics are quite comparable to normal deflections under similar loads. The foot deflects 0.5 deg. under the normally applied eversion moment of 2 ft.-lb. (.28 m. kg.) and inverts 4.5 deg. under the normally applied inversion moment of 15 ft.-lb. (2.08 m. kg.). Under the normally applied maximum torque of 6 ft.-lb. (83.9 m. kg.) the unit rotates transversely 7.0 deg. internally and 7.0 deg. externally. Generally, all of the values noted above, with the exception of deflection under eversion moments, approximate normal values rather closely. However, the characteristic hysteresis of these designs was noted in their failure to return to initial positions when unloaded.

TABLE 4. — *Comparative Deflection Under Normal Maximum Loadings Applied During Stance Phase*

Motion	Normal maximum moment and torque		Rotation in degrees		
	ft.-lb.	kg.-m.	Normal	Trautman Universal Ankle[a]	Otto Bock Greissinger with compressible heel
Plantar flexion	18.0	2.5	18	8.5	15.0
Dorsiflexion	80.0	11.1	10	4.5	12.5
Eversion	2.0	0.3	6	4.0	0.5
Inversion	15.0	2.1	6	18.0	4.5
Torsion					
Internal	6.0	0.8	9	12.0	7.0
External	6.0	0.8	8	12.0	7.0

[a]See pp. 110–112, BPR 10–4 Fall 1965.

Under load, the foot with the compressible heel deflects linearly when forces are applied to the foot-shank assembly. The compressibility of the heel wedge is roughly comparable to a "medium" SACH foot heel grade. The weight of the foot and ankle joint hardware is 1.35 lb. (.61 kg.), well below the maximum acceptable weight of 1.80 lb. (.82 kg.) for SACH feet.

The following deviations from current standards were noted: The toe break for both feet is 1.25 in. (3.18 cm.) short for their respective sizes. Toe flexure resistance of the foot with the compressible heel falls slightly above the specified resistance range (Fig. 13); the foot with the non-compressible heel falls outside the permissible range. The higher durometer of the toe of the foot with the compressible heel caused the excessive resistance to flexure.

The "universal" motion of the bolt assembly seems to provide adequate five-way function. Four patients, three below-knee and one above-knee, reacted favorably to the unit, particularly as regards use in such activities as golf.

No problems arose in aligning the foot. Although the plantar-flexion dorsiflexion attitude cannot be adjusted, the manufacturer recommends a special alignment to position the foot in the proper attitude for maximum function. The resistance to plantar flexion and dorsiflexion is adjusted by tightening the ankle bolt. This feature may adversely affect the function and stability of the unit. The feet can be readily finished with paint or horsehide, but lamination requires careful application to produce an acceptable product. An adequate cosmetic effect is achieved with foam filler or a leather fairing.

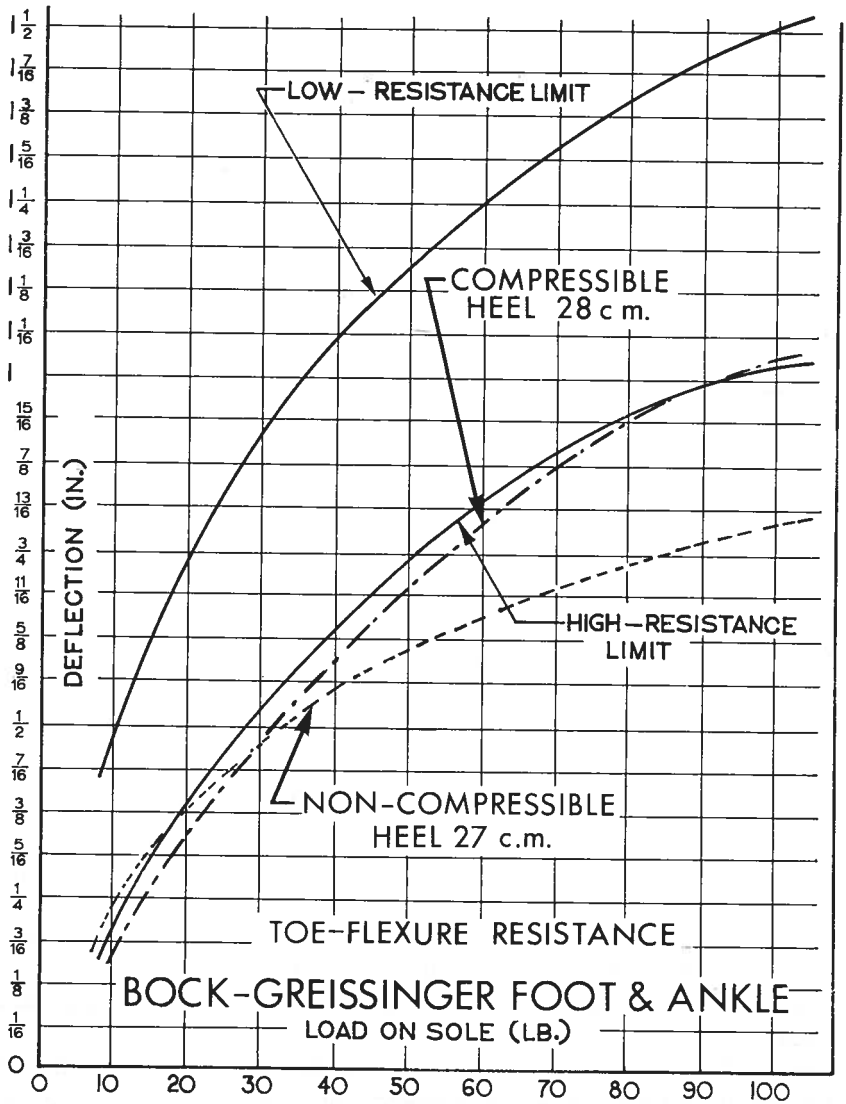

FIGURE 13

4. Teufel SACH-Type Foot and Ankle. This unit is a partially molded SACH-type foot featuring a wood external keel with a felt and sponge toe assembly, a laminated sponge rubber heel wedge, and a sponge plastic sole (Fig. 14). Although it is not a *molded* SACH foot, for evaluation purposes it was compared with current specifications for SACH feet.

FIGURE 14. — Teufel SACH-type Foot and Ankle with external keel and felt and sponge toe.

The results of this analysis indicated that this unit does not meet the standards in several respects. The following deviations were noted:

a. Toe-break is 5/16 in. (.8 cm.) under allowable limits.
b. Toe curl is 1/16 in. (.16 cm.) in excess of tolerance.
c. Toe flexure resistance exceeds the high limit at approximately 55 lb. (Fig. 15.)

While this unit does not meet all the standards for SACH feet, it appears reasonably serviceable in other respects. The manufacturer has been apprised of the deficiencies.

5. *Teufel Telasto Foot and Ankle Assembly.* The unit consists of a wood ankle section and a partially molded SACH-type foot featuring a wood external keel with a felt toe and a plastic foam sole (Fig. 16). Foot and shank are held together by a single axis steel ankle joint mounted on a rubber bushing. A rubber plantar-flexion bumper is set in a recess on the foot behind the ankle joint. A laminated plastic sponge dorsiflexion bumper is located on the foot forward of the ankle joint. The foot features plantar-flexion and dorsiflexion control; a small degree of eversion-inversion control is possible due to the mounting of the ankle bolt in the rubber sleeve. Measures were obtained of plantar flexion and dorsiflexion under load both with and without the plantar-flexion bumper. Applicable standards for SACH-type feet were applied to evaluate toe-break location, degree of toe curl under load, and toe flexure resistance under load.

The unit permits a maximum of 22.5 deg. of plantar flexion. Further plantar flexion is limited by contact between the shank and the foot. Without the bumper, a plantar-flexion moment of 4.5 ft.-lb. (.62 m. kg.) produced the full range of deflection, 22.5 deg. With the plantar-flexion bumper installed, a bending moment of 20 ft.-lb. (2.76 m. kg.) produced the full plantar-flexion range, a response which approximates the normal in which 18 deg. of plantar flexion occur under maximum normal plantar-flexion moments of 18 ft.-lb. (2.49 m. kg.). Except, perhaps, for persons

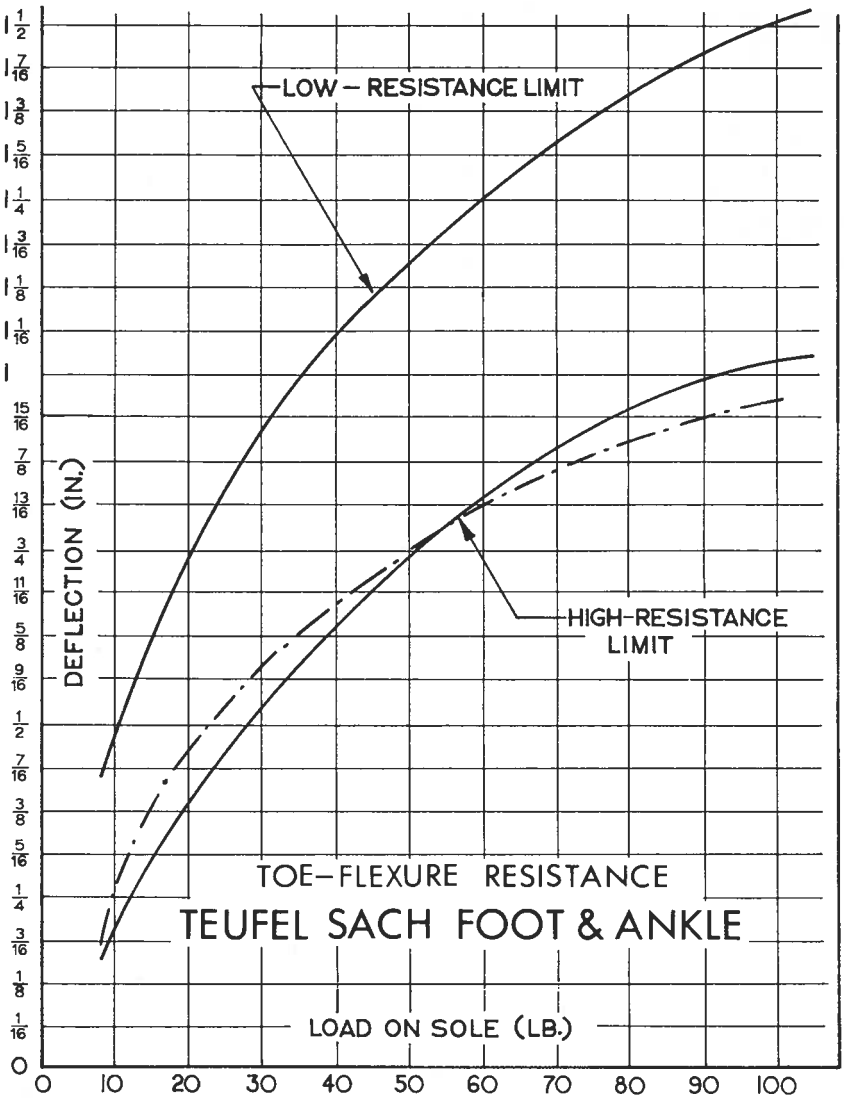

FIGURE 15

whose gait is "flat footed," that is, with little or no transition between heel contact and foot flat, the unit should be used with the bumper.

The unit permits 2.2 deg. of dorsiflexion which is obtained under a 20 ft.-lb. (2.8 m. kg.) dorsiflexion moment.

The toe-break occurs at $5\frac{15}{16}$ in. (15 cm.) from the edge of the heel This is ¾ in. (1.9 cm.) short of the specified $6\frac{11}{16}$ in. (17 cm.)

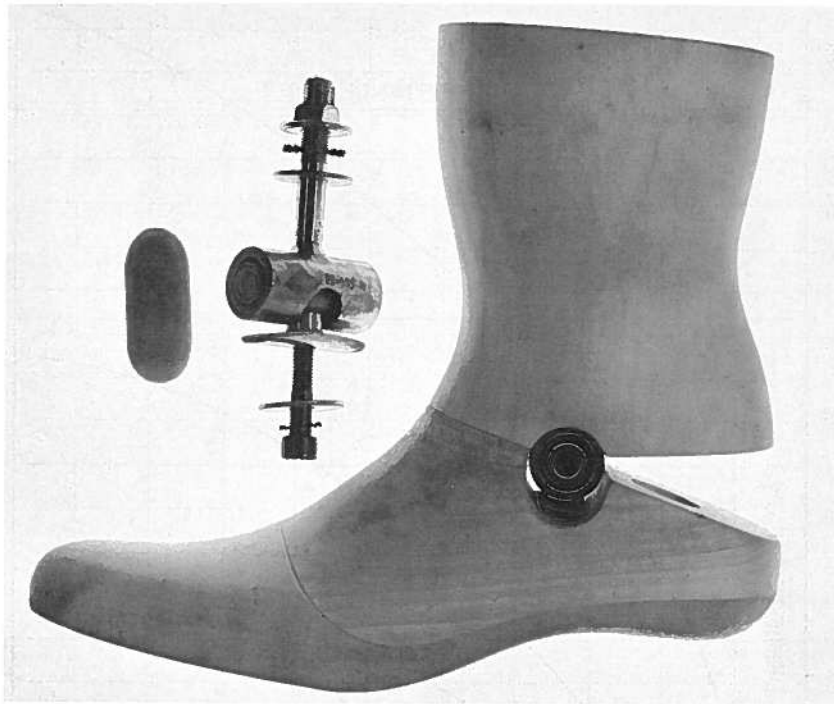

FIGURE 16. — Teufel Telasto Foot and Ankle assembly featuring a rubber mounted single axis ankle and rubber plantar-flexion bumper to be installed in well of heel.

for a foot of this size (9) and 5/8 in. (1.59 cm.) short of acceptable limits. The toe deflected 3/32 in. (.239 cm.) under the standard load, a figure well within the 1/8 in. (.3 cm.) limit the standard for toe curl imposes. The weight of the foot and ankle joint hardware is 1.43 lb. (65 kg.), well below the maximum weight standard of 1.69 lb. (79 kg.) for SACH feet.

As shown in Figure 17, toe flexure resistance exceeded the high resistance limit. The measured flexure resistance would be higher, if the resiliency of the dorsiflexion bumper were eliminated.

From the prosthetist's standpoint this unit can be easily shaped and finished by laminating. However, the ankle-heel separation may make it difficult to achieve adequate cosmesis.

6. *Navy Intermittent Friction Knee.* Ten units submitted for evaluation by the Navy Prosthetic Research Laboratory are being evaluated on the basis of mechanical tests and clinical fitting experience. The current model of this device is similar in design to older models as regards major func-

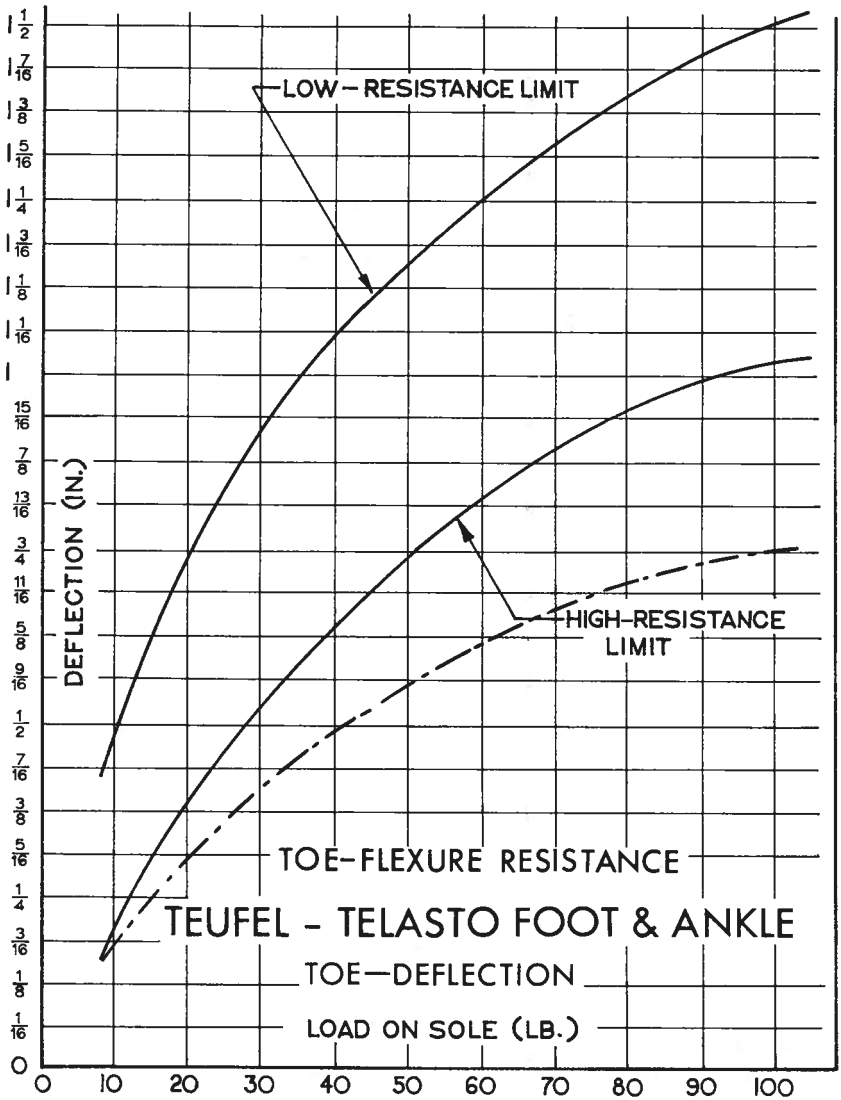

FIGURE 17

tional features; it differs in the use of new materials, fabrication method, and in other minor details.

The knee mechanism is mounted in a urethane-filled prefabricated knee shell made of three layers of polyester laminate (Fig. 18). Available in five sizes, these shells are neuter and may be used on the right or left side. The unit employs a conventional 9/16-in. (1.43 cm.) knee bolt

FIGURE 18. — Navy Intermittent Friction Knee showing urethane-filled knee shell and two friction brakes mounted on knee bolt on either side of extension stop bar.

mounted in split Oilite bronze bearings covered with a thin rubber sleeve. The split bearing permits adjustment for wear on the bolt by tightening a gland nut. The friction thus developed also provides uniform resistance to knee motion throughout the swing phase.

The two friction brakes, mounted on the knee bolt, are lined with metal impregnated brake material. Extensions projecting from each brake oscillate between two bumpers as the knee flexes and extends. Because the bumpers for the right and left brakes are not symmetrically positioned, the oscillation of the brakes is slightly out of phase and their resistances

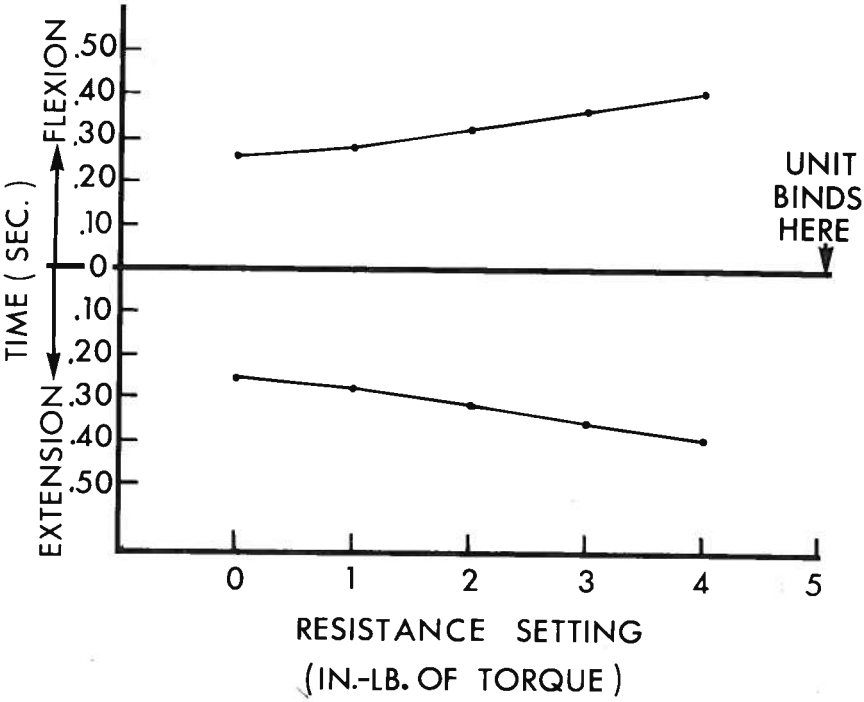

FIGURE 19. — Resistances to flexion and extension of Navy Intermittent Friction Knee as indicated by drop times.

are applied sequentially. The resistance of the brakes is adjusted by means of screws.

A stop bar on the knee bolt extends midway down into the shank and acts to limit extension by contact with the knee block and with the inner rear surface of the shank. In extreme flexion, the stop bar comes in contact with the anterior inner surface of the shank.

Knee flexion and extension are therefore limited by the stop bar and the resistances to flexion and extension are equal to the sum of the frictions in the bearings and in the brake bands. The friction in the bearings is constantly applied throughout swing phase and the friction applied by the brakes is adjustable by tightening or loosening the brake bands.

An earlier model was the subject of an evaluation conducted at New York University (2) in which it was found that: 1. while the unit could be adjusted initially to provide adequate swing phase control, it required frequent adjustment to take up for wear on the brake friction bands, 2. readjustment was not conveniently accomplished by the patient in that he had to remove the leg, 3. some patients objected to the "free swing" range; in taking short steps there was no effective control of knee motion.

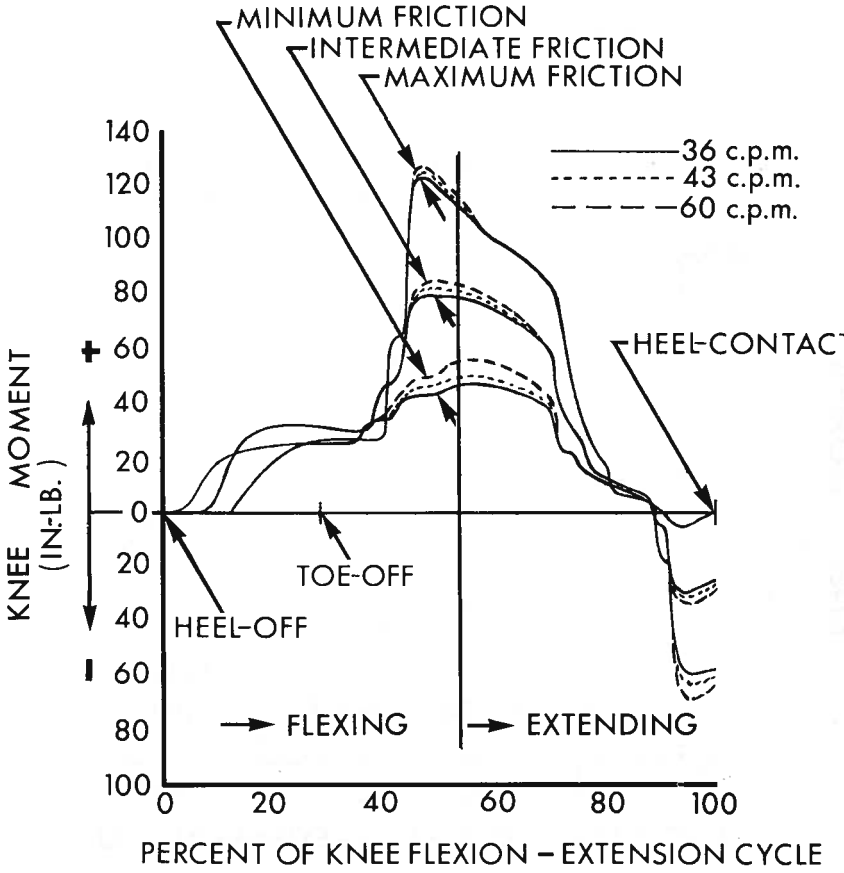

FIGURE 20. — Knee moments at varying cadences of Navy Intermittent Friction Knee.

Mechanical testing indicated that the unit provides a degree of resistance to flexion-extension comparable, if not superior to, other mechanical friction devices. As shown by drop times and knee moments (Fig. 19 and 20), the unit provides resistances to flexion-extension comparable to the lower values of those provided by hydraulic knee mechanisms. Figure 20 also reveals, as expected, that the unit is not significantly cadence-responsive in that resistances were not markedly affected by cadence regardless of friction setting.

When these units were retested after 6000 cycles, there was appreciable change in resistance characteristics as compared with the results after 1000 cycles. As shown in Figure 21, there was a general reduction in knee moment values. However, little significant change was noted in the last 10 percent of swing phase where control of terminal impact is critical.

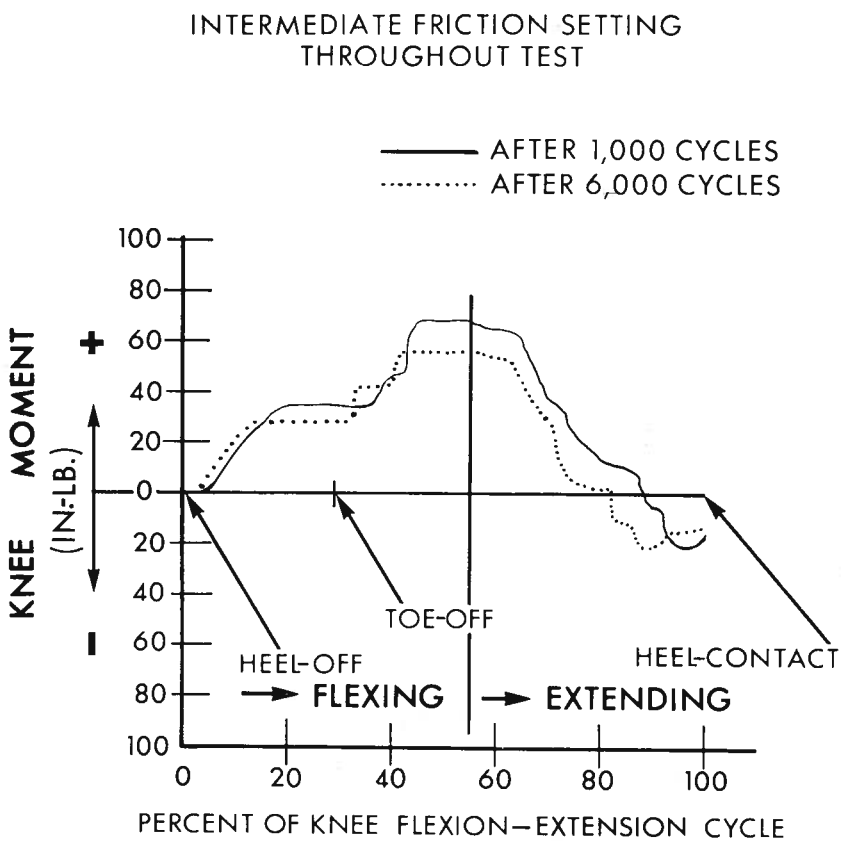

FIGURE 21. — Resistances of Navy Intermittent Friction Knee after 1000 and 6000 cycles at an intermediate resistance setting.

Several patients have been fitted with this device in order to obtain the clinical and biomechanical performance data required to complete this evaluation.

7. *Secura Knee.* The unit submitted by Teufel of Stuttgart, Germany, consists of a knee and shank assembly joined by a link between the knee center of rotation and a secondary center of rotation located on the shank, inferior and posterior to the knee center (Fig. 22). It is primarily designed to provide stance phase stability by means of friction developed between two mating surfaces, one on the knee block and the other on the shank. Swing-phase control is effected by two features in the unit: a compression spring assembly which also acts as an extension bias, and by the mating friction surfaces when they are engaged by means of the adjustable link position stops. The compression in the spring is adjustable. The

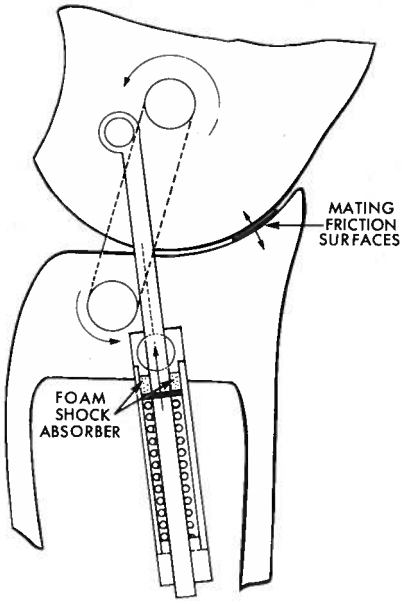

FIGURE 22. — Drawing of Secura Knee showing four-bar linkage system, two mating friction surfaces, and a compression spring assembly.

position relationship between the rotation centers is also minimally adjustable.

By means of the crossed four-bar linkage, axial loads (even those with small flexion moments) cause the upper link in the knee block to rotate forward and downward driving the two friction surfaces together. As swing phase is initiated and the axial loads diminish, the extension bias spring lifts the knee block reducing the friction and permitting easy transition into swing phase. Terminal impact is controlled by means of a plastic foam shock absorber located above the piston of the swing-control extension-bias unit.

The literature provided by the manufacturer asserts that the stance-phase control feature provides substantial resistance to knee flexion in the heel contact to foot flat period, a time when normally applied bending moments are of considerable magnitude (maximum approximately 100 in.-lb. [115 cm. kg.] for above-knee amputees). With the recommended adjustments, the unit can resist torques of 10 in.-lb. (11.5 cm. kg.) under a 60-lb. (27.27 kg.) axial load. As the axial load is increased, of course, the resistance of the unit to rotation increases proportionately. But, as axial loads increase during stance, flexion moments about the knee are also reduced as the ground reaction vector moves anteriorly and the need for high resistance then is also reduced.

The unit resists knee flexion during early stance to an extent which reduces the effort on the part of the patient to stabilize his knee. It does

not "lock" the knee to an extent which makes his effort unnecessary. Between heel-off and toe-off, when the amputee's knee normally begins to flex, the residual axial load and the consequent resistance of the knee to flexion may possibly increase the effort of initiating flexion.

Swing phase in this unit is only partially controlled by the spring-loaded cylinder in which the spring is compressed as the knee flexes. The friction mechanism seems to provide the major control. This finding appears to contradict the literature which, in referring to Figure 22, indicates that friction is minimal during swing.

The spring acts to retard flexion and to limit heel rise. As the knee begins to extend in swing phase, the compressed spring acts as an extension bias. The force of extension diminishes as the knee extends, a fact which may reduce the effectiveness of extension bias when very short steps are taken. Terminal impact in swing phase is controlled by means of a plastic foam ring positioned above the piston, and to some extent by the "braking" action of the friction surfaces which come into contact when the unit is fully extended even when no weight is borne.

Figure 23 shows drop-time values for flexion in the range of 0.34 seconds to 0.42 seconds at the minimum setting, 0.38 seconds to 0.52 seconds in the intermediate setting, and 0.50 seconds to 0.75 seconds

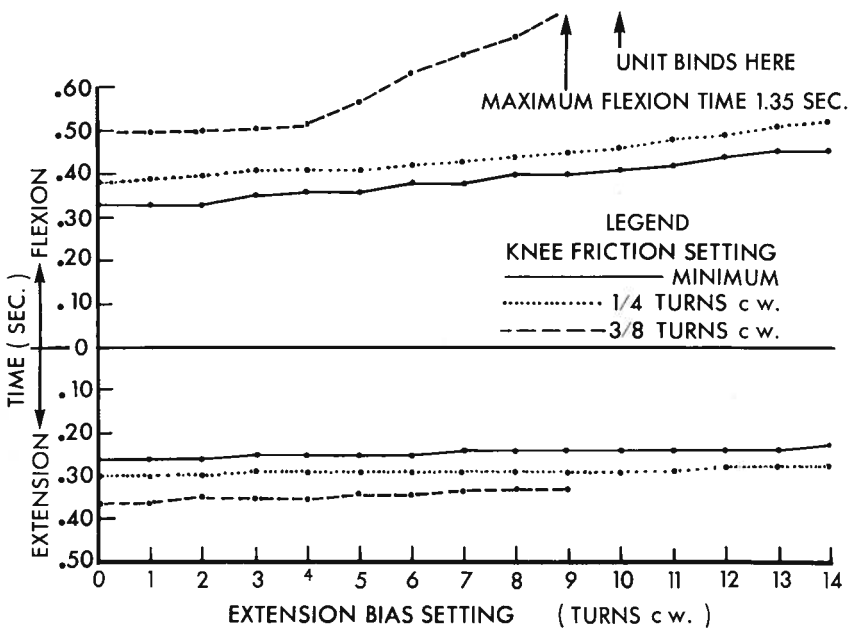

FIGURE 23. — Resistance of Secura Knee at various settings as indicated by drop times.

(approximately) before the unit binds in the maximum resistance setting. At each setting, the extension bias control provides small but proportionately increasing resistances. Extension control, as indicated by drop times, is effectively adjusted by altering the friction setting but it is relatively unaffected by the extension bias control.

Figure 24 indicates a fairly low order of resistance to knee flexion at the minimum setting where the peak swing-phase moments cluster around 60 in.-lb. (69 cm. kg.) at all extension bias settings. At the intermediate setting, the knee moment pattern was similar but of higher magnitude clustering around 75 in.-lb. (86.25 cm. kg.). Knee moments obtained at the maximum setting were only slightly higher. Figure 24 also shows that the friction settings contribute significantly to swing-phase resistance while the extension bias furnishes relatively little resistance. At the minimum setting, the extension moment is 12 in.-lb. (13.80 cm. kg.) during the last 10 percent of the cycle, indicating minimal control over terminal impact. At the intermediate setting, control is somewhat improved in that moments

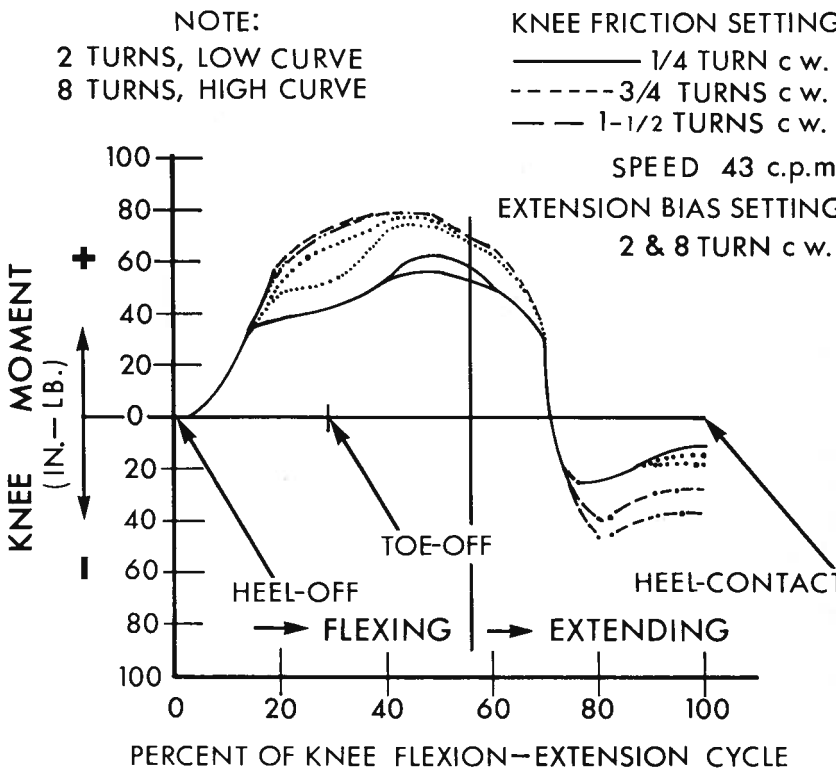

FIGURE 24. — Knee moments provided by Secura Knee at various extension bias settings.

of approximately 16 in.-lb. (18 cm. kg.) were recorded. At maximum settings, terminal impact control is substantially improved with knee moments approximately 35 in.-lb. (40.25 cm. kg.). As expected, at any setting, the unit displays very little cadence response in that cadences of 36, 43, and 60 cycles per minute produced little change in resistance. Knee moment values in Figure 24 for stance phase (heel-off to toe-off) do not reflect the effects of axial loads since the test apparatus cannot simulate weight bearing.

Functionally, this unit provides a measure of swing control comparable to other low-resistance units. Its stance-phase control appears to have merit. Biomechanical data and clinical trials are in progress to complete the evaluation of this unit.

E. Evaluation (Techniques)

None.

II. UPPER-EXTREMITY PROSTHETICS

A. Development

1. Improvement of Force-Excursion Relationships for Shoulder-Disarticulation Amputees. Among the many significant problems identified and discussed at the last meeting of the Workshop Panel on Upper-Extremity Prosthetics Fitting, Harnessing, and Power Transmission of the Subcommittee on Design and Development, CPRD, was the need for "excursion amplification" devices or techniques, particularly for the shoulder-disarticulation amputee. The problem discussed in the CPRD report revolved around the relatively high-control forces available to the shoulder-disarticulation amputee and the generally limited excursions available for elbow flexion and operation of terminal devices. Previous approaches to the problem led to the design of pulley systems to increase excursion at the cost of increased force (Fig. 25). While many of these devices function adequately, as a class, they are difficult to align, they add hardware and weight, and they produce an uneven posterior surface of the shoulder-disarticulation socket.

Input forces required to operate the conventional terminal devices usually prescribed for shoulder-disarticulation amputees generally range from approximately 3 lb. (1.36 kg.) where one rubber band is used to 5 lb. (2.27 kg.) for two rubber bands, and 8 lb. (3.64 kg.) for three rubber bands. The force to initiate elbow flexion in a reasonably well-aligned prosthesis is approximately 10 lb. (4.55 kg.). The normal individual easily develops forces up to 60 lb. (27.27 kg.) in scapular abduction, the primary control motion for the shoulder-disarticulation amputees. While the amputee may have somewhat less force available, it is nevertheless substantially in excess of the requirements.

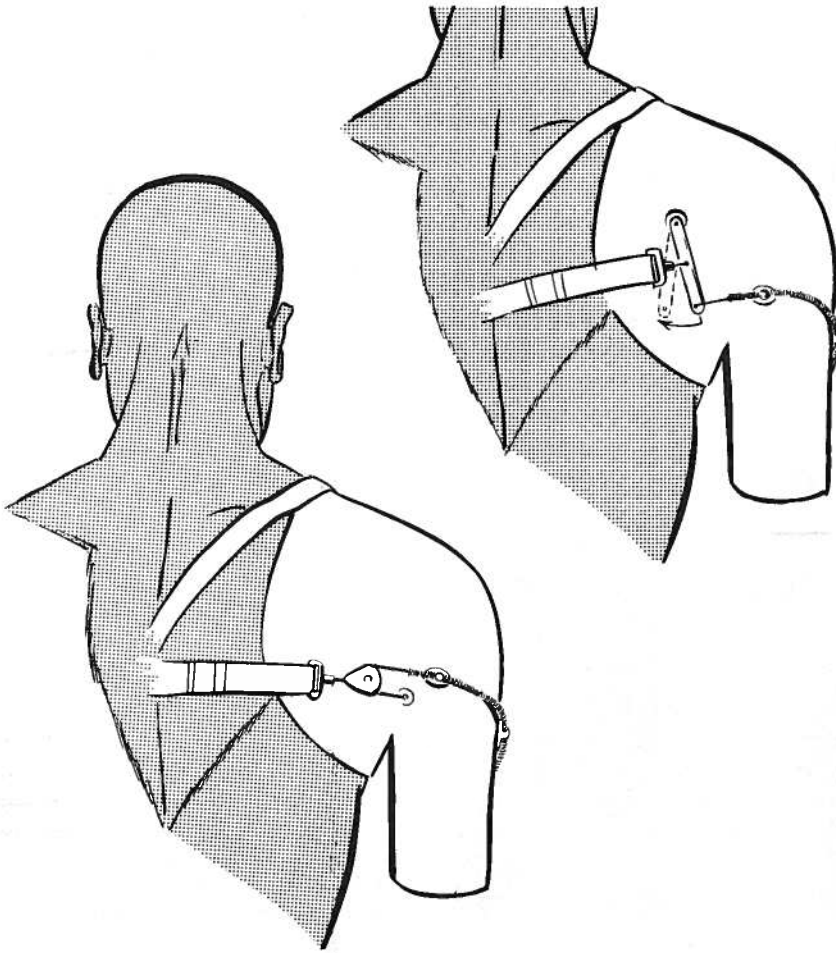

FIGURE 25. — Lever system (top) and pulley arrangement (bottom) designed to increase cable travel.

Approximately 1½ in. (3.81 cm.) of cable travel are required to effect full opening of the conventional terminal device, and approximately 2 in. (5.08 cm.) of cable travel are generally required to effect 100 to 120 deg. of elbow flexion. Shoulder-disarticulation amputees rarely achieve more than 2½ to 3 in. (6.35 to 7.62 cm.) of useful excursion. Full terminal device opening, therefore, is generally difficult or impossible for these patients with the elbow flexed. The force-excursion requirements for operating conventional hooks depend not only on the number of rubber

bands being used and friction in the system, but also on the length of the lever arm which is designed to optimize the force excursion requirements. Similarly, force transmission systems are conventionally aligned to reduce the force requirements for elbow flexion.

It may be possible to reduce the excursion required to operate conventional terminal devices by simply reducing the length of the lever arm. The force requirements would increase proportionately, permitting a "trade" to decrease excursion requirements by increasing force requirements. By departing from conventional control cable alignment technique, it may also be possible to decrease the cable travel required to flex the elbow at the cost of proportionately increased force.

FIGURE 26. — Conventional Dorrance 88X hook modified to permit operation with a shortened operating lever.

To test this hypothesis, the forces and excursion required to operate a conventional hook were recorded and compared to those required to operate a hook which was modified in a manner to reduce the length of the operating lever (Fig. 26). As shown in Table 5, use of the modified hook reduced the excursion requirement to 1.2 in. (3.05 cm.) as against the 2.2 in. (5.58 cm.) required with the conventional. The force requirements were increased proportionately. But the maximum force required to initiate terminal device opening did not exceed 7.0 lb. (3.18 kg.) with the one rubber band used in this test. The use of three rubber bands would probably increase this requirement to approximately 21.0 lb. (9.55 kg.), well within the capabilities of many shoulder-disarticulation amputees.

TABLE 5. — *Force-Excursion Relationships for Operating a Dorrance 88X Hook With Conventional and Shortened Operating Lever*

Terminal device lever arm	Excursion to open fully		Force to open			
			Initial opening		Open fully	
	in.	cm.	lb.	kg.	lb.	kg.
Conventional	2.2	5.6	4.0	1.8	3.3	1.5
Shortened	1.2	3.0	7.0	3.2	6.5	3.0

Similar decreases in excursion with accompanying increases in force were effected by control cable alignments which placed the elbow lift reaction point closer to elbow center than the conventional 1.0 in. (2.54 cm.) superior and 1.0 in. (2.54 cm.) distal. As shown in Table 6, with conventional control cable alignment, 4.5 lb. (2 kg.) and an excursion of 2.1 in. (5.33 cm.) were required to flex the elbow to 90 deg. In this case, conventional alignment placed the lift lever reaction point 1.25 in. (3.18 cm.) distal to the elbow center and 1.0 in. (2.54 cm.) above it. Positioning the elbow lift reaction point 0.75 in. (1.91 cm.) above and 1.0 in. (2.54 cm.) distal increased the force to 6.0 lb. (2.73 kg.) but reduced the excursion to 1.7 in. (4.3 cm.), a reduction of approximately 0.50 in. (1.27 cm.). In the third test position, the lift reaction point was placed 0.75 in. (1.91 cm.) distal to the elbow center and 0.50 in. (1.27 cm.) above it, requiring a force of 11.0 lb. (5 kg.) but reducing the excursion to 1.0 in. (2.54 cm.), a net savings of approximately 1.2 in. (3.05 cm.) as compared to conventional alignments.

TABLE 6. — *Force and Excursion in Relation to Location of Forearm Lift Reaction Point*

Forearm lift reaction point		Force to flex elbow to 90°		Excursion to flex elbow 90°	
in.	cm.	lb.	kg.	in.	cm.
1.25	3.2 distal	4.5	2.0	2.1	5.3
1.0	2.5 superior				
1.0	2.5 distal	6.0	2.7	1.7	4.3
0.75	1.9 superior				
0.75	1.9 distal	11.0	5.0	1.0	2.5
0.50	1.4 superior				

These studies indicate the possibility of reducing the excursion requirements by 1.0 in. (2.54 cm.) and the terminal device requirements by 0.50 in. (1.27 cm.) for a net reduction of approximately 1.50 in. (3.81 cm.). It may therefore also be possible to reduce the operating excursion requirements from 3 to 3.50 in. (7.62 to 8.89 cm.) to approximately 2.0 in. (5.08 cm.).

2. *Direct Forming of Below-Elbow Sockets.* A newly available material, Polysar Cast, a synthetic balata referred to earlier in this report, has been used to form a below-elbow socket directly on a stump. The material was available in extruded cylinders of several diameters and wall thicknesses. The Polysar Cast cylinder was immersed in hot water of 180 deg. F. (82 deg. C.) for several minutes and reformed into a cone over a silicone-coated tapered mandrel. The cone was again heated by immersion in water and pulled over the below-elbow stump covered with silicone impregnated stockinet. The patient forced his stockinet covered stump into the undersized cone, molding it precisely to the contour of the stump (Fig. 27). While still soft, the "socket" was additionally contoured to the correct biomechanical shape and the proximal brim was flared by the prosthetist. Immersed thereafter in ice water, the cooled "socket" became relatively rigid. The finished product, shown in Figure 28 was fabricated by laminating a wrist fitting to the distal end and covering the completed socket with a cosmetic shell.

FIGURE 27. — Heated Polysar cone is formed over patient's stump by the prosthetist who molds it into required biomechanical form.

FIGURE 28. — Finished Polysar prosthesis covered with an experimental cosmetic cover.

Polysar is the most promising material by far for this purpose. Tests are being continued in an effort to develop techniques for attaching a pylon directly to the completed socket to receive the wrist fitting and for finishing the prosthesis with a "soft" cosmetic cover.

B. Evaluation (Components)

1. AIPR (American Institute of Prosthetic Research) Externally Powered Components. In continuation of this study, a clinical evaluation was undertaken of the subject's functional status. The major purpose was to determine the best possible combination of powered and conventional components as a basis for final prescription. In addition to the test and performance data obtained in this study, the evaluation of these components depends on the subject's reactions to the utility of each of the experimental prostheses. These reactions must necessarily be based on a comparison between the externally powered components and the conventional system with which he was initially fitted. Due to the lapse of approximately 18 months since he wore conventional components, comparisons were difficult. The subject, therefore, has been refitted with conventional components to be worn for a period of 3 months to reestablish the original baseline for comparison. At the end of this period, the evaluation will be completed.

In the meantime, a new series of studies has been initiated to "map" the muscular activity patterns involved in the control of the AIPR components. The electrical activity of specific muscles in the shoulder girdle is being studied by conventional EMG techniques to identify the most efficient control sites for AIPR and other externally powered systems.

2. Northwestern University Center Control Hook. This unique device is a single pivot, voluntary opening hook whose closure is effected by rubber bands (Fig. 29). The fingers are offset medially 30 deg. permitting the control cable to pass through the attachment stud and wrist unit. The fingers are lyre-shaped and lined with neoprene. The maximum opening of the hook is 4 in. (10.16 cm.). Six prototypes were made available to several facilities for preliminary testing.

Tests of durability and general utility were not performed as the prototypes were cast in soft aluminum for studies of design feasibility only. The evaluation included analysis of installation requirements, grasp characteristics, and patient reaction to function and appearance. Two subjects, one below-elbow and one above-elbow, participated in the evaluation. Both subjects are highly active hook wearers. The below-elbow subject ordinarily wears a voluntary opening Northrop-Sierra two-load hook, and the above-elbow subject wears a Sierra 4C hook, a voluntary closing device. The experimental hook had the same basic finger configuration as

the terminal devices worn by the subjects and training time therefore was minimal.

The unit was installed by screwing the ½ in. (1.27 cm.) 20 threaded stud into the wrist unit in the conventional manner. However, the conventional cable connector (3/16 in. [.478 cm.] triple ball swivel) butted against the base of the hook, preventing full finger opening. A single ball connector was required for proper clearance.

The offset (30 deg.) fingers were advantageous in positioning objects in space particularly with respect to the midline of the body. For example, a telephone could be brought to the proper attitude with the receiver at the ear and the mouthpiece at the mouth without repositioning. In general, office activities, such as writing and handling papers, were adequately accomplished. However, it was difficult to fold a piece of paper.

It was quite difficult to grasp objects lying on a flat surface unless their long axes were parallel to the frontal plane. When grasping an object from a flat surface, the subject must either change his body position or supinate the hook fingers in order to align the line of action in finger closure perpendicular to the long axis of the object. When the fingers are rotated medially to grasp an object, the offset angle of the hook causes the lower finger to strike the surface first making it difficult to grasp small objects. For the same reason, it is not possible to use a driving ring which is attached to the steering wheel. In eating, the absence of the operating lever (thumb) results in a less stable grasp of several utensils.

The below-elbow subject, who wore the Northrop-Sierra two-load device, was noncommittal on the overall appearance. The above-elbow subject,

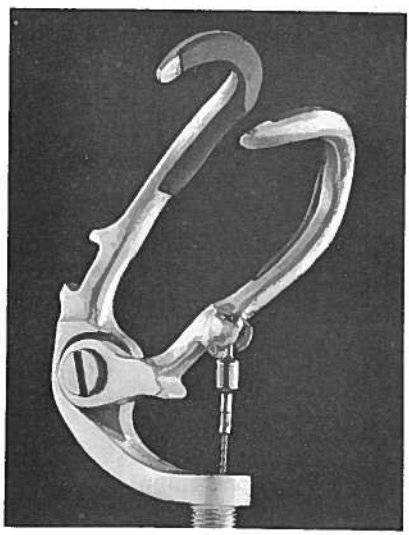

FIGURE 29. — Northwestern University Center Control Hook eliminates protruding operating lever.

who wore the Sierra 4C device, rejected the hook as "too conspicuous." However, laboratory observers agreed that when the amputee is wearing a shirt or coat with long sleeves, the hook is not markedly different in appearance from the Sierra devices. We have concluded that this device offers little or no advantage over conventional hooks, and we have therefore discouraged further development.

3. *CAPP Terminal Device (Sumida Hook).* The adult size Sumida hook features one conventionally shaped finger and one broad finger. The latter finger provides some degree of "palmar prehension." The "palmar" surface contains a resilient insert designed to offer a large contact surface for grasping irregularly shaped objects. It was originally designed to be actuated by an electrically driven push-pull rod which passes through a standard ½ in. (1.27 cm.) 20 threaded mounting stud. This arrangement permitted 360 deg. of hook rotation without adversely affecting control. In the absence of a suitable external power system, the hook was adapted for the conventional (Bowden cable) transmission system (Fig. 30).

FIGURE 30. — CAPP Terminal Device (Sumida Hook) features one broad finger lined with a resilient insert and one conventionally shaped finger. The outrigging was necessary to adapt Sumida Hook for conventional Bowden cable transmission system.

A rubberband-controlled four-post outrigger was fitted to the hook. This temporary arrangement made it possible to fit the terminal device to an amputee's existing prosthesis and to evaluate its functional capacities when used by one unilateral below-elbow wearer in the laboratory. This report is based on the experiences and observations in fitting the Sumida hook to one adult unilateral below-elbow wearer.

The combination of one broad and one conventionally shaped finger provided very secure grasp of large or irregularly shaped objects. Two other functions that are normally provided by conventional hooks—fine fingertip prehension and hooking—were performed with difficulty. Fine

"fingertip" prehension of the type used when sorting file cards or slides or when picking up coins from a table was not possible with this terminal device. Use of the single finger for hooking was possible by opening the fingers, but this was awkward in both appearance and performance.

Although the outrigger arrangement required relatively high input forces to operate, the primary problem in poor fingertip prehension was related to the finger configuration. We have recommended that any further development of this device be focused on applications for children where fingertip prehension is less frequently required than among adults.

4. Gilmatic Elbow Lock. In an effort to develop a simple device to lock the elbow of an upper-extremity prosthesis, the Gilmatic Company designed a solenoid-operated electric elbow lock which was installed in a modified Hosmer E–400 elbow (Fig. 31). By means of an ingeniously designed switch, power is conserved and the solenoid is not active except during the actual locking or unlocking action, periods of approximately 10 msec. each. The elbow lock is controlled by means of a cable connected to the harness which utilizes the same body control motions as the conventional elbow.

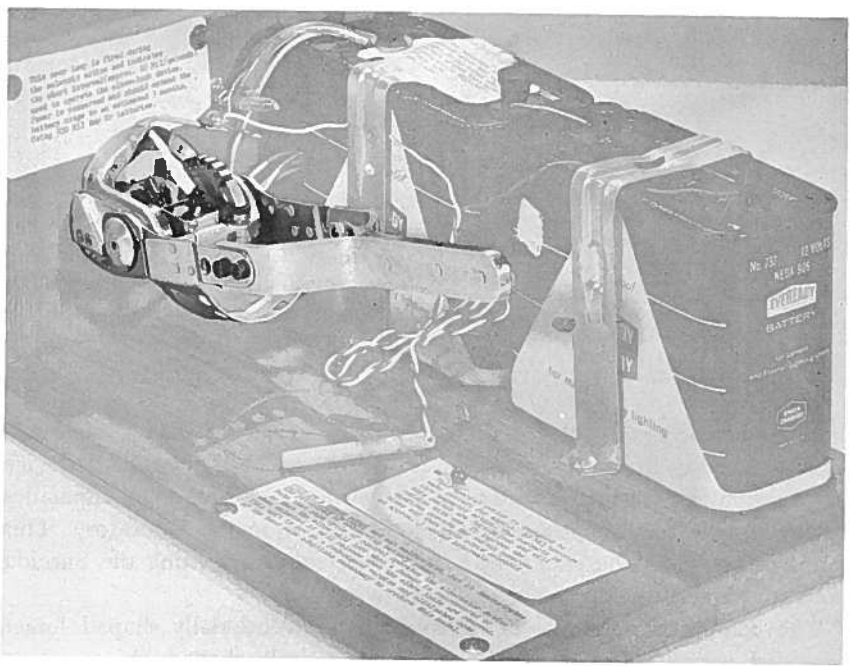

FIGURE 31. — Demonstration model of Gilmatic Elbow Lock installed in commercially available elbow.

Since this model illustrated a design concept and did not represent a prototype, the evaluation was based on an analysis of potential advantages and operational requirements. Mechanical tests and clinical studies were not undertaken.

The problem of control is a critical matter in this and other similar devices. It is difficult to see advantages resulting from the actuation of an elbow lock by the same body motions which ordinarily lock the conventional elbow even though the excursion requirements might be reduced. The unilaterally fitted shoulder-disarticulation case usually needs more control than the basic chest strap harness provides, and it is frequently necessary to elaborate the harness to the point where it approaches the triple control configuration. As a result, almost every available body control motion is brought into play to its fullest excursion and force capability. To operate an electrical elbow by means of a pull would seem a small gain even though the actual force requirement and excursion may be reduced. Elbow lock control remains a basic problem because even under the best circumstances mechanical actuation requires displacement of the terminal device and conspicuous motions.

The power conservation concept is extremely valuable in that it places the battery charge life expectancy of this system in a practical range. This idea should certainly be retained in the evolution of this development. We are cooperating with the developer to determine the utility of actuation by means of EMG signals, muscle bulges, or other fine anatomical motion.

Adequate control of a single isolated function such as provided by this powered elbow lock will probably represent the first clinically useful prosthetics application of such sophisticated sources as EMG.

C. Evaluation (Techniques)

None.

III. LOWER-EXTREMITY ORTHOTICS

A. Development

None.

B. Evaluation (Components)

Becker Polycentric Knee Joint. One model of a polycentric knee joint for use in leg-thigh braces was submitted for evaluation by the Becker Company of Birmingham, Michigan (Fig. 32). It is designed to stabilize the knee during stance phase and to permit flexion of the knee during the swing phase of gait. In the recommended alignment, the instantaneous center of rotation is above and behind the anatomical knee center producing a stabilizing extension moment. For most cases, this will mean that the knee joint tends to remain stable even after the knee is flexed 10 to 15

deg. In theory, this could reduce the forces required of the wearer to stabilize the knee joint in early stance. Proper functioning of this unit depends on coaxial alignment of the four mechanical centers of rotation on the medial side and the four on the lateral side, a difficult procedure or one which will require a special tool. A test subject will be fitted with this unit to evaluate functional utility and fitting problems.

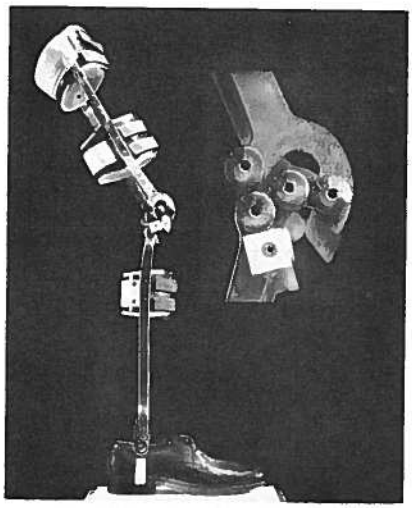

FIGURE 32. — Becker Polycentric knee joint installed in leg-thigh brace. Inset shows four-bar linkage and extension stop.

IV. ORTHOPEDIC AIDS

A. Development

Clinical Evaluation Study on Wheelchairs. A program is being planned to obtain information on the utility of wheelchairs in actual use to supplement the tentative VA wheelchair specifications currently under development. The basic purpose is to determine the relative merits of various chairs with respect to different categories of disabilities.

As presently conceived, sets of nine different chairs will be shipped to each of five stations after inspection by the Bioengineering Research Service. A round-robin type of procedure is being considered in which individual patients would use each chair for periods of not less than 2 weeks each. The participation of stations with specific types of caseloads is being considered to provide the required data. The durability of the wheelchairs under conditions of actual use will be studied by carefully monitoring the repair and alteration histories on each chair. Patient preference for various chairs and the specific reasons therefor will be obtained by means of questionnaires and interviews.

1. Reevaluation of the Hydro-Crutch (Model FP–2100). Two earlier models of the Hydro-Crutch, a battery powered, hydraulic telescoping crutch designed to assist patients in getting up and down from a seated position, were evaluated in January 1965 and again in July 1965. These studies resulted in several recommendations to the manufacturer for improving the electrical and hydraulic systems, reducing the noise level, and generally improving the appearance and quality of the crutches. In July 1966, two redesigned models (Fig. 33) of the Hydro-Crutch (FP–2100) were received for evaluation.

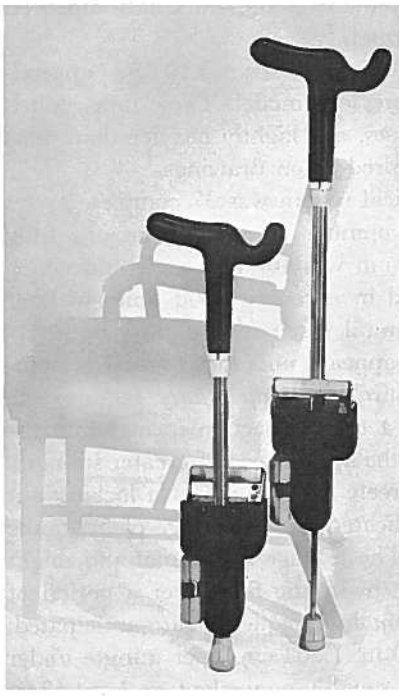

FIGURE 33. — Redesigned model of battery - powered Hydro - Crutch showing screw-type adjustment for patient height.

The test program, designed to evaluate the modified crutches, included an analysis of the electrical and hydraulic systems and a series of cycling tests to determine their functional reliability. Details of test procedures and the test fixtures were previously reported in BPR 10–4 Fall 1965, pp. 170–6.

In general, the function of the Hydro-Crutch has been improved to an extent which makes it acceptable as a mechanical device. The electrical and hydraulic systems operate reliably, and little significant improvement can be recommended within the limits of economy and practical application.

All of the "major" engineering problems, except that of weight reduction, have been solved. The active components have been placed in the handle, the electrical limit switch was replaced with a hydraulic pressure relief valve, a headless piston with superior oil seals has been installed, a diaphragm-type no-leak release valve replaced the older one, and the motor torque-speed-efficiency curve and the pump speed-spressure curve have been optimized to meet the extension-force requirements. Relocating the major components in the handle produced a more favorable center of gravity location, a smaller and thinner auxiliary support, and a neater appearing power package. This and the other changes have improved the reliability of the systems, extended the cycle life of the units, and improved the physical appearance of the Hydro-Crutch.

The unit weight has been reduced to 6 lb. 13 oz. (3.10 kg.) approximately 1 lb. (.454 kg.) less than the previous model. These units, fabricated under limited production techniques, are slightly heavier than mass produced units would be according to production drawings.

The average charge life of the electrical system was 37 complete cycles of extension-retraction. The minimum number of cycles from one fully charged power pack was 30; the maximum was 46. The crutch on which this test has been performed was cycled in excess of 4000 times without failure. It is being cycled continuously until mechanical failure occurs.

Although the lower extension screw appears underdesigned it is actually of adequate strength. As this is purely an *appearance* factor some attention might be given to the use of a tubular section to maintain the weight-strength relationship and to give the appearance of greater strength. The lower extension screw rotates too freely and the use of a locking nut or other device to reduce rotation is indicated. Some leakage of hydraulic fluid occurs at the plastic hose fittings. The leakage is minimal and might result in the very slow depletion of the hydraulic fluid over a period of perhaps two years, but no other detrimental consequences are anticipated.

The unit retracts approximately 0.080 in. (.203 cm.) per minute under a 75-lb. (34.09 kg.) axial load. This is roughly equivalent to 1 in. (2.54 cm.) in 12 minutes. To apply a 75-lb. (34.09 kg.) load to each crutch would require a 150-lb. (68.18 kg.) man to bear full weight on the crutches for this period, or for a 300-lb. (136.36 kg.) man to bear partial weight for a similar period. In either case, readjustment of crutch length is simply a matter of actuating the extension mechanism.

The minor deficiencies cited above do not detract significantly from the function of the Hydro-Crutch. However, the patients who can benefit most from this device and their reaction to comfort, balance, weight, and noise need to be determined.

2. *Sears Wheelchair (Model 8A1516N).* One model (Fig. 34) of this unit was evaluated on the basis of compliance with recently developed

tentative wheelchair specifications. By these standards, the Sears chair is of generally marginal adequacy. While it met the minimum requirements in most respects, several features were below standard. The durability of the chair is questionable since a wobble of 1/16 in. (.160 cm.) measured at the rim was recorded in wheel rotation through 360 deg. after a test simulating 100 curb descents. The space between the hand rim and the drive wheel is only 3/4 in. (1.91 cm.) wide. As a result, the thumb comes in contact with the tire of the drive wheel and the hand rim spacers. When propelled, the caster wheels wobble slightly. Excessive play was noted between the fork stem and its bearings. The durometer of both drive wheel and caster tires is too high in that under load the area of the drive wheel tire in contact with the ground was only 3/4 sq. in. (4.838 cm.^{2}) or 1/4 sq. in. (1.7 sq. cm.) below the requirement. The area of the caster tire in contact with the ground was 3/8 sq. in. (2.419 cm.^{2}) or 1/8 sq. in. (.806 cm.) below specification. The chair is difficult to fold, requiring approximately 25 lb. (11.36 kg.) of force, a figure 10 lb. (4.55 kg.) higher than desired. Several minor deviations from the tentative specifications were also

FIGURE 34. — Sears-Roebuck folding wheelchair with rigid padded seat.

noted. While this chair may be quite useful for many purposes, it does not quite meet the standards inherent in the tentative specifications. The results of clinical trials will be used to complete the evaluation.

3. Colson "Classic" Wheelchair (Model A83F147151–270). One model of this wheelchair was evaluated by comparison with the functional specifications for wheelchairs. In general, the Colson chair is a "deluxe" device incorporating all conventional and several novel features (Fig. 35). New items include reversible and removable armrests with positive locks. This gives the user the option of full height armrests or "desk-type" height.

Weighing 58 lb. (27 kg.), it exceeds the specification by 10 lb. (4.55 kg.). It is difficult to fold requiring 2½ lb. (1.14 kg.) of force over the allowable minimum. Its folded width is 13 in. (33.02 cm.), 1 in. (2.54 cm.) above the desired 12 in. (30.5 cm.). Clinical trials are in progress to complete this study.

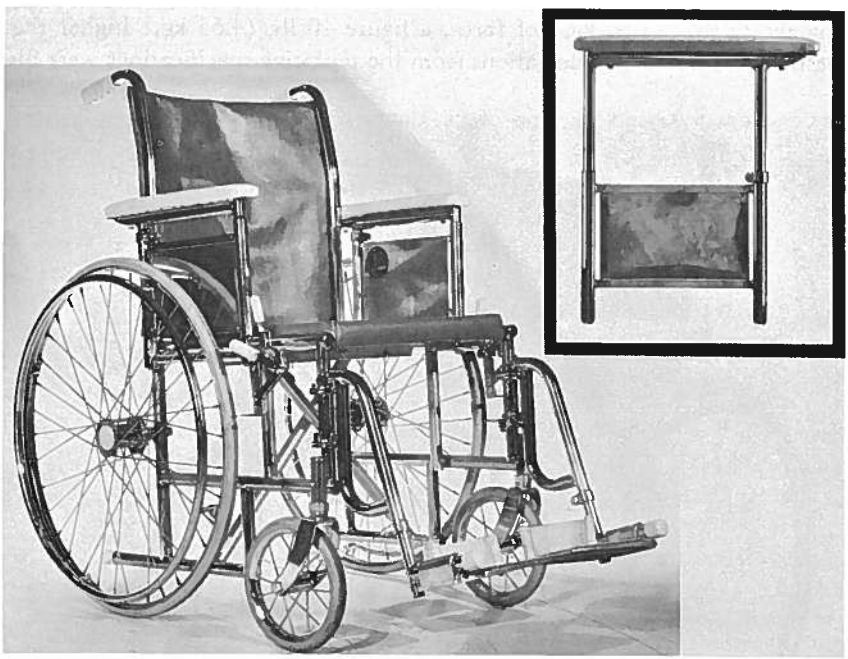

FIGURE 35. — Colson "classic" wheelchair features reversible and removable armrests with positive locks.

4. Cetrone Contoured Support Belt. The Westchester Belt Company of Jefferson Valley, New York, submitted for evaluation a specially designed sacroiliac belt. This appliance, worn beneath the outer clothing below the level of the iliac crests, is a means of relieving discomfort due

to low back pain, and to prevent further insult to the muscle tissue and ligaments in that area (BPR 10–5 Spring 1966).

The device is a simple leather belt, 1/16-in. (.160 cm.) thick and 1 3/4-in. (4.45 cm.) wide, with a two-pronged buckle for length adjustments. It features two leather reinforcing pads sewn on the inner surface of the belt to fit under the iliac crests. By conforming to the shape of the pelvis in the hollow under the iliac crests, these pads serve not only to prevent the stretching of the leather, but also to prevent the belt from "slipping up." Designed to encircle the pelvis, the device lends support to the entire "pelvic arch." By applying strong medially-directed forces which tend to compress the ilia, tension is reduced on the ligaments supporting the low lumbar vertebrae and those between the sacrum and ilium.

The principle of reinforcing the pelvic arch by encircling it with a belt to relieve low back pain is not new (3). In discussing a "hoop" effect for example, Jordan states:

> "The hoop must be firm (non-elastic) and to act at the right site, it must lie between the iliac crest and the greater trochanter. For practical purposes, the desired result is obtained most simply with a one-inch strap of webbing which encircles the pelvis between the iliac crest and the trochanter and is fastened in front with a buckle. The effect of this simple appliance is often quite startling. It may therefore be used as a test before constructing a more complicated belt."

The literature furnished by the manufacturer describes a number of allegedly preventive and therapeutic functions resulting from wearing this device. Five specific claims can be extracted from this literature:

a. Relieves discomfort of backache due to sacroiliac sprains.
b. Aids in the prevention of sacroiliac sprains.
c. Permits greater mobility with more comfort than other similar devices.
d. Helps bring down healing time.
e. Low cost compared to corsets.

It is currently available in three sizes to fit waists up to 34 in. (86.36 cm.), 40 in. (101.60 cm.), and 48 in. (121.92 cm.). Each size is adjustable through a range of 6 to 8 in. (15.24 to 20.32 cm.) by means of a buckle and grommets.

This report is based on experiences in fitting the Cetrone Contour Belt to five subjects, of whom two were previous wearers of orthopedic appliances for the relief of chronic low back pain and one for whom a lumbar corset had just been prescribed.

The specific condition referred to here as "low back pain" is the generally recognized syndrome characterized by acute pain localized in the sacroiliac area, reduced mobility, a recovery time from 1 to 3 weeks for each incident, and periodic recurrence. It does not include pain which is referred to the low back from higher sections of the spine, a condition requiring bracing to immobilize the spine.

No information regarding the prevention of sacroiliac sprains (claim b), was obtained in this study. A clinical field-type evaluation would be required for this purpose.

A total of five male subjects (Table 7) in generally good health wore the Cetrone Contour Support Belt for periods ranging from one week to two months. Subjects one and two suffered periodic "low back pain" characterized by the classic symptoms of pain, reduced mobility, and a recovery time varying from one to three weeks for each incident. To relieve the symptoms during the recovery period and sometimes to offset an incipient attack, one subject wore a lumbar corset, and the other alternated the use of a lumbar corset and a hinged lower back brace. Subject number three was in the acute stage of a second incident of the "low back" syndrome. The previous incident had not been treated by means of any orthopedic appliance. The severity of this attack led him to an orthopedic surgeon who prescribed a 9-in. (22.86 cm.) lumbar corset and bed rest.

TABLE 7. — *Subjects Wearing Cetrone Belt*

Subject	Age	Condition	Orthopedic appliance used previously
1	51	Chronic low back syndrome—acute stage.	Lumbar corset.
2	49	Chronic low back syndrome — acute stage with extreme pain and spasms in sacroiliac area.	Lumbar corset and hinged lower back brace.
3	37	Second attack of low back syndrome.	9-in. lumbar corset prescribed at time of attack.
4	67	Chronic back discomfort generalized — not clearly low back syndrome — intermittent spasms.	None.
5	51	Chronic back discomfort generalized — not clearly low back syndrome — intermittent spasms.	None.

The other two subjects of this study were physicians with chronic back discomfort not properly classified in the same category as the acute "low back pain" syndrome. One had "back discomfort for 6 or 7 months," with "occasional spasms." The other had "backache from time-to-time."

For subjects one, two, and three with the classical low back syndrome, the relief of acute symptoms provided by the Cetrone Belt was at least equal to that afforded by the lumbar corsets or the hinged lower back brace. Each of them was quite positive in his opinion that the Cetrone Belt permitted greater activity during the acute stage. Greater range of hip motion without excessive pain was possible. The Cetrone Belt was worn with much less general discomfort than the corsets and braces. Excessive pressure and sweating and the stiff posture associated with the use of corsets and braces were reduced sharply.

Subject four with generalized back discomfort, indicated that the Cetrone Belt seemed to reduce his "backache" but registered doubt whether the belt "gives proper support." He also found it necessary to readjust the position of the belt quite often in changing from sitting to standing and vice versa. Subject number five, with generalized "backache," found the Cetrone Belt of no use in relieving his discomfort. He implied that it caused sciatic pain.

On the basis of these findings, the Cetrone Belt seems to offer significant advantages over corsets and braces in the treatment of acute symptoms of low back syndrome particularly with pain in the sacroiliac area. While the Cetrone Belt does not seem to offer any clear-cut therapeutic superiority, its potential advantages relate to increased comfort, reduction of inactivity and lost time, and economy.

We have therefore recommended that the ability of the Cetrone Contoured Support Belt be determined in a field study utilizing a minimum of twenty-five patients. We have also recommended that the manufacturer consider using Dacron webbing instead of leather and a Velcro closure in a redesign of the belt.

5. *Seat Lifts.* Two seat lifts, both designed to assist patients in rising to a standing position or lowering to a sitting position, are being evaluated.

The Cushion Lift manufactured by Ortho-Kinetics, Inc. of Waukesha, Wisconsin (Fig. 36) is an electrically powered device which operates on ordinary house current (115 volts). As the cushion rises, it tilts forward at an angle that can be adjusted, in four attitudes, to provide the optimum standing angle.

The E & J Elevating Wheelchair Seat is a hydraulically operated elevating wheelchair seat mechanism (Fig. 37). The seat can be installed in existing E & J chairs and is readily removable for folding the chair. It also facilitates transfer from wheelchair to bed for patients with lower-extremity involvement.

The evaluation program to assess the features of these devices includes laboratory analysis of the lift capabilities as well as clinical evaluation.

6. *Proreco Curb-Climbing Wheelchair Assembly.* Several years ago a great deal of interest was generated in the design and development of stair

FIGURE 36. — Electrically powered Cushion-Lift designed to aid a seated patient to stand.

climbing wheelchairs. Of the many design concepts advanced, only a few were produced. However, a fundamental problem persists in attempting to integrate two primary functions in one vehicle. The design requirements for conventional wheelchairs are often incompatible with the irreducible minimum in hardware for stair climbing. As a result of this impasse, designers have sought an intermediate goal: the design of a vehicle to ascend and descend one step, i.e., a curb climber. Among the several designs recently advanced for this purpose was the Proreco Curb-Climbing Wheelchair Assembly developed by a student engineering group at Catholic University under the direction of Professor Marshall M. Lih. This proto-

FIGURE 37. — E & J elevating wheelchair seat is manually operated hydraulic lift.

type consists of two lifting levers mounted on a modified rear axle and two auxiliary caster assemblies installed on a conventional wheelchair. To ascend curbs, the occupant backs the wheelchair against the curb and lowers the auxiliary caster device which tilts the chair backwards. The rear end of each lifting lever is placed on the curb and the forward ends are

FIGURE 38. — Patient ascending curb using a prototype Proreco Curb Climber. Once in position, patient raises himself to level of curb by pulling the lifting levers.

grasped by the passenger. Employing the lifting levers (second class), he lifts the chair upward and backward until the rear wheels are on the curb. He then rolls the chair back and retracts the auxiliary casters (Fig. 38).

Although the principle of this design seems to have merit, there are very few patients capable of operating the device. A paraplegic patient with a low level lesion (cauda equina) could not mount a 4-in. (10 cm.) curb. In order to reach the handles of the lifting levers, the subject had to flex his trunk at the waist. In this position he was not only incapable of extending his trunk under load, but pulling on the levers simply flexed his trunk further (reversal of action of the muscle groups).

The forces to operate the device are actually somewhat higher than those given by the developer because the occupant cannot apply the force perpendicular to the lifting bars.

This device represents an ingeniously simple approach to a complex problem which unfortunately remains unsolved.

7. Evaluation of Prototype Staircane. One prototype model of the Staircane (Fig. 39), a device designed to reduce effort in ascending stairs, was submitted for evaluation by Special Devices, Inc., 76 Reade Street, New York City, the developer. Intended for cardiacs, the unit consists of an upright "cane," attached foot rest, a cylinder and piston, a compressed nitrogen storage tank, and actuating valves. Held by a patient on a staircase landing, the unit is operated when body weight is applied to the foot rest by one foot placed into the stirrup. Compressed nitrogen drives the foot rest up, raising the patient to the required height (a distance which is adjustable for risers up to 8 in. [20.32 cm.] high). Thereupon, the patient steps *forward* onto the step with his "free" foot. When the stirrup is unloaded, a spring retracts the piston. During this phase, the entire weight of the device is supported by the patient's toe within the stirrup and by the arm with which he grasps the cane portion.

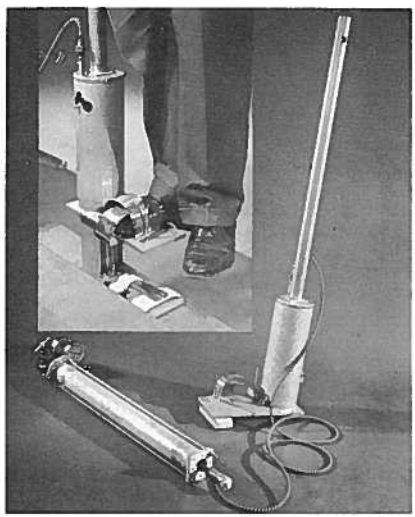

FIGURE 39. — Staircane showing compressed nitrogen storage tank, upright "cane," cylinder housing the piston, and footplate.

As a prototype, this model of the Staircane was evaluated on the basis of functional considerations of safety, maneuverability, and ease of operation. Matters of durability and reliability were not considered.

In general the utility of a device of this nature is questionable. Operated by compressed gas, the unit requires either a portable cylinder to contain an adequate amount of gas for perhaps 6 to 10 stair ascents without refilling, or to be operated from a large non-portable tank by means of an "umbilical" tube. The former adds weight, and the latter appears inconvenient and potentially dangerous in maintaining a large tank at either the foot or the head of the stairs and in the use of a long tube on the stairs. The portable tank designed for this prototype has a capacity of approxi-

mately 3 cu. ft. (84.960 cu. cm.) of gas at the recommended pressures, a volume sufficient for approximately 20 complete extension cycles. Thus, the tank would have to be refilled after each one or perhaps two ascents of a standard 12-step flight of stairs. Descent is also difficult in that the patient must either carry the device down the stairs or "bounce" it down.

We have not been able to determine a definite need for this device. Its potential utility in conserving effort is offset by the fact that the patient substitutes static work at the ankle, knee, and hip for the positive work normally done against gravity in performing the vertical component of the climb. Added to this is the effort of supporting the unit's weight after the piston retracts.

Moreover, in the opinion of the Chief, Cardiology Section, NYRO, the Staircane was not a feasible device for cardiac patients. Severely disabled cardiac patients who are confined to bed have no need for a stair climbing device. Less severely disabled who have the capacity and are permitted to climb stairs merely ascend them at a slower pace. Inside the home it is rarely necessary to climb more than one flight. Others use elevators, a more convenient and less taxing mode. Although a need may exist for a *portable* stair climber among those cardiac patients residing in metropolitan areas where they must climb subway stairs daily, the Staircane with its bottled gas supply and long umbilical cord would not meet this requirement. The results of our evaluation have been made available to the developer.

8. Preliminary Evaluation of Joe J. Stump Socks. One below-knee amputee was fitted with the Joe J. sock according to the procedures specified by the manufacturer. Although the socks are "made to measure," their construction of ⅛ to ¼ in. (.318 to .635 cm.) foam rubber between two layers of jersey cloth, provides little elasticity, and it is difficult to fit a patient adequately. In addition, the tapered end of the sock is joined by a zigzag stitch which reduces the stretch at the distal end.

Initially, the subject found the sock more comfortable than the 5-ply wool sock that he previously wore. Minimum discomfort was experienced during the first day of wear with some "pressure" noted on the stump along the seams of the sock. Definite discomfort was experienced on the second day. Seam pressure was more noticeable; abrasions appeared on sensitive stump areas requiring the patient to discard the sock in favor of his conventional wool sock. He also reported heat and perspiration problems.

The Joe J. stump sock initially seemed to offer definite advantages in cushioning, "bounce," and shock absorption. However, within hours, these effects were lost.

Shrinkage of the socks was negligible as measured after five wash cycles. Although tried on only one case, we believe that clinical usefulness of this sock will be extremely limited. It serves like the formerly popular BK-PTB

socket liner either to compensate for poor prosthetic fitting or to provide extra cushioning for extremely sensitive stumps. But such stumps should be cushioned with a contoured or fitted liner as the original BK manual specified.

9. Knight Spinal and Taylor Brace with Detachable Coverings and Aprons. Early in 1965, the VA Prosthetics Center undertook the evaluation of prefabricated Taylor and Knight spinal braces wtih detachable aprons and coverings. The cover consists of a double thickness of white coutil fabric with foam rubber used as padding for the metal frames and Velcro for attachment of the coverings to the frames.

Both the Taylor (Fig. 40) and Knight (Fig. 41) spinal brace frames with one complete covering and apron are supplied by the manufacturer in four sizes: small, medium, large, and extra-large. To be able to wash the entire brace covering and particularly the covering of the apron is a hygienic advantage. Providing a patient with one brace and two coverings would permit one covering to be washed while the other was in use. This eliminates the need for two complete braces, a potential savings of approximately 40 percent of the total cost of two braces.

To date, we have fitted 51 patients with these braces at the VA Prosthetics Center; 22 other cases have been fitted in five VA Orthopedic Brace Shops. Experience indicates that the time required to measure and

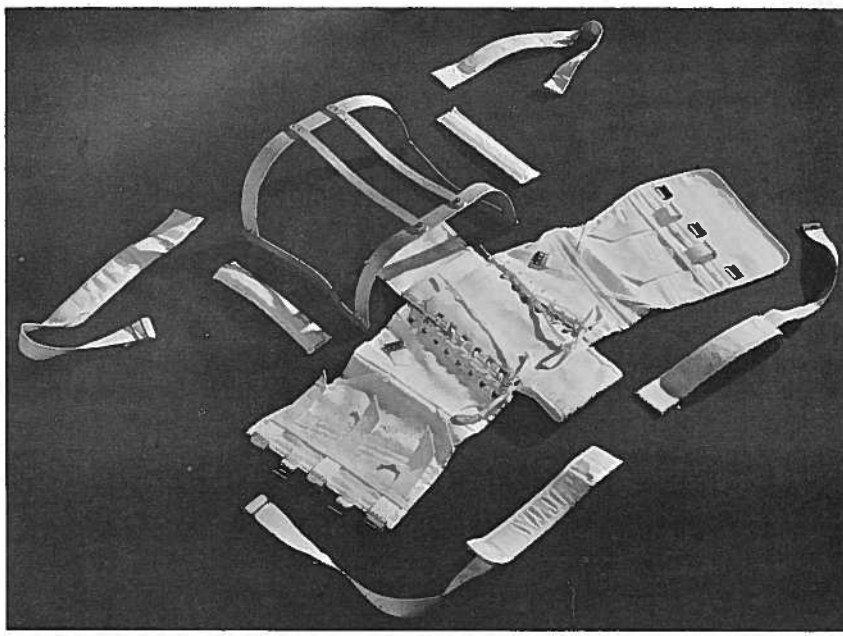

FIGURE 40. — Taylor back brace with detachable covering and apron.

fit the Taylor and Knight spinal braces was equal to or less than the time required for conventional prefabricated braces. The consensus among orthotists was that the adjustability of the covering by means of laces and Velcro closures was a desirable feature; it reduces the time for final adjustment particularly for patients ready for discharge from a hospital. It was also noted that the coverings were easily removed and replaced by patients themselves and that washing the coverings was no different from washing any other fabric.

This study also brought out several deficiencies in the original design. Aluminum screws, used in both Taylor and Knight spinal braces, stripped and failed. They have been replaced by stainless steel screws. The most significant problem in both braces was excessive bulkiness due to the Velcro fasteners, a problem yet to be solved.

We find both the Taylor and Knight spinal back braces with detachable aprons and coverings generally acceptable items particularly for incontinent patients. Until the excessive bulkiness of these braces is reduced, however, we cannot recommend general application.

As an added result of this study, some additional light has been shed on the utility of prefabricated braces of this type as compared to custom-made appliances. Prefabricated Knight spinal and Taylor back braces can probably meet the needs of perhaps 50 percent of our beneficiaries and, with

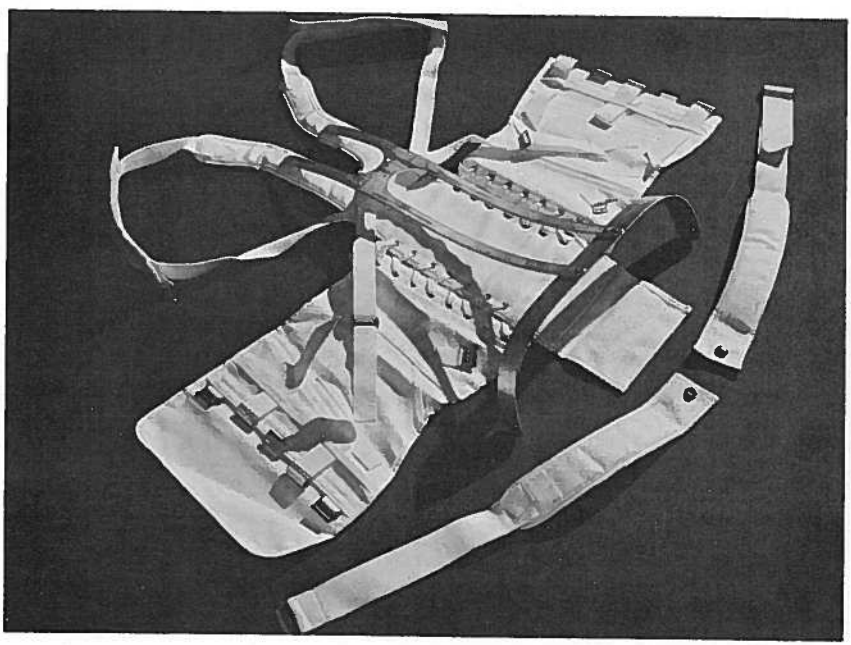

FIGURE 41. — Knight spinal brace with detachable covering and apron.

minor modification, prefabricated appliances could serve even a larger fraction. This particular design permits easy modification when necessary. It would appear that for a large majority, prefabrication would result in uniformity of product, reduction of time lag between prescription and fitting, and easier part replacement.

B. Evaluation (Components)

None.

V. TESTING

A. Specifications Compliance

1. SACH Feet. One prototype of a SACH-type foot was submitted by the U.S. Manufacturing Company for evaluation. The foot is constructed of a laminated wood external keel and a molded plastic sole and toe section (Fig. 42). The unit was tested by comparison with applicable standards for SACH-type feet (VA Specifications No. VAPC-L-5902-3).

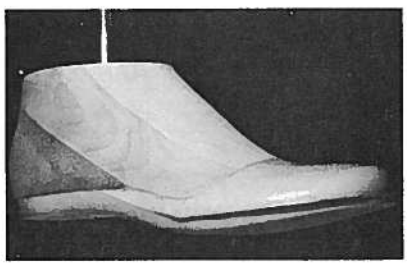

FIGURE 42.—U. S. Manufacturing Corp. SACH-type foot with external keel and molded plastic sole.

With two exceptions the foot met all the functional requirements set forth in the specifications. The 9¾ in. (24.77 cm.) length exceeded the specifications for nominal size 7 foot (Spec.: 9½ ±⅛ in. [24.13 ±.318 cm.]) by ⅛ in. (.318 cm.), placing it closer to the nominal size 8 foot (Spec.: 9⅞ ±⅛ in. [25.08 ±.318 cm.]). Toe flexure resistance was generally well within the prescribed limits, up to loads of approximately 85 lb. (38.64 kg.); in this region, the flexure curve began to exceed the high resistance limits by small increments (Fig. 43). Heel compressibility tests place the heel in the border area between the "firm" and the "regular" heel hardness.

To consider this device for VA acceptance as a contract item several modifications would be required. Adjustment of the length dimension and the heel compressibility would bring this foot within specified functional limits. Proper identification of heel compressibility, date of manufacture, and location of A and B reference lines would upgrade the product to an acceptable level. To be eligible for VA contract the foot must also be available in the required sizes.

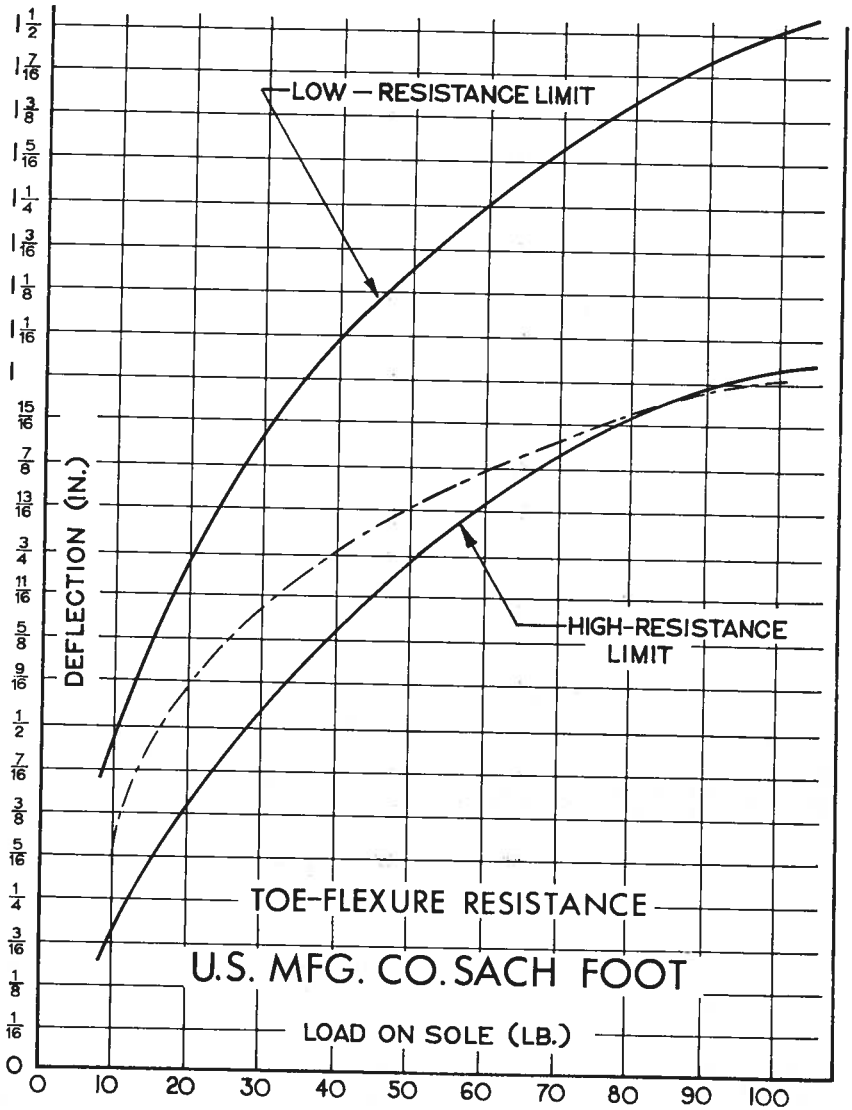

FIGURE 43

During this period, another manufacturer submitted samples of SACH feet for annual compliance testing. The samples could not be rated as acceptable in view of three significant deviations from specification requirements:

a. Toe break too short.
b. Toe curl in excess of allowable limits.

c. Heel durometer below requirements.

2. *Stump Socks.* Samples of wool, leg stump socks submitted by five manufacturers were tested for compliance with specifications detailed in Invitation No. 67–2 in continuation of our annual Specifications Compliance Test Program.

The samples submitted by two companies conformed completely to the specifications requirements with no deviations. Minor deviations from the specifications were noted in the products of the other companies. Each of these companies was informed of the noted deficiencies.

3. *Upper-Extremity Components.* Compliance tests were conducted on four upper-extremity components during this period: a. Hosmer Internal Elbow, Model E–400, b. Miracle Voluntary Closing Hand, c. Becker Imperial Voluntary Opening Hand, d. Dorrance Voluntary Opening Hand. The Hosmer E–400 Internal Elbow met the full specifications requirements completely; the Dorrance Voluntary Opening Hand met them marginally. Testing of the other two units was terminated in each case due to failure of components: a linkage in the Becker Imperial Hand and a pivot pin in the Miracle Voluntary Closing Hand.

4. *Lumbo-Sacral Corset Material.* Seven manufacturers submitted a total of 28 specimens of fabrics used in lumbo-sacral corsets for annual Specifications Compliance Test Program. Tests were conducted in accordance with Federal Specification CCC–T–191b, Part 5550.2 (DOD), September 17, 1963, "Shrinkage in Laundering, Cotton, Linen, and Mixed Cotton and Linen Cloth." The products of five of the seven manufacturers were found to comply with the specifications. One was rejected for excessive shrinkage and another for poor workmanship.

VI. OPERATIONS REPORT FOR SECOND HALF, FISCAL YEAR 1966

The data in this section reflect the prosthetic and orthotic services provided for veteran beneficiaries during the second half of Fiscal Year (F.Y.) 1966 by the VA Prosthetics Center's Orthopedic Shoe Service and the Prosthetics-Orthotics Service. Fiscal year comparisons are also included where available and meaningful.

A. The Orthopedic Shoe Service

Table 8 reflects fiscal year comparisons of our national Orthopedic Shoe Program from 1963 to and including 1966. An increase of 129 beneficiaries for the second half of F.Y. 1966 was experienced.

Issuance of protective footwear (rubbers and overshoes) during the second half of the fiscal year continued to show a marked increase; 166

TABLE 8. — *The VAPC National Orthopedic Shoe Program*

	Fiscal year			
	1963	1964	1965	1966
Beneficiaries on rolls	9,681	10,021	10,322	10,587
New shoes, prs.	7,249	7,317	7,336	7,688
Prs. of new shoes issued per benef. on rolls per yr.	.75	.73	.71	.73
Repaired shoes, prs.	9,716	9,487	9,422	10,432
Prs. of shoes repaired per benef. on rolls per yr.	1.00+	.95	.91	.99

pairs of overshoes and 73 pairs of rubbers were issued, bringing the totals of 269 and 131 respectively for the fiscal year. Also, 423 worn clinic lasts were duplicated for a total of 511 for the fiscal year.

B. The Prosthetics — Orthotics Service

The Orthotics Components Unit of the Prefabricated Appliances Section is responsible for the distribution of surgical supports and elastic hosiery. Table 9 reflects activity from F.Y. 1963 through F.Y. 1966. Please note the upsurge in the issuance of these items for F.Y. 1966: a 37 percent increase in surgical supports and an 18 percent increase in elastic hosiery over the previous year.

The Prosthetics Components Units of the Prefabricated Appliances Section is responsible for the distribution of prosthetic and orthotic devices. Table 10 summarizes the distribution.

The Limb and Brace Section of the Prosthetics-Orthotics Service fabricated and delivered items as indicated in Tables 11, 12, and 13.

The VA Prosthetics Center Special Prosthetics Clinic Team held 31 clinics during the 2nd half of F.Y. 1966. Eighty-three veterans were referred to the Clinic by 14 field stations.

C. Special Service for Vietnamese Wounded

As reported in the previous Bulletin (BPR 10–5 Spring 1966), the VA Prosthetics Center was authorized in late 1965 to provide orthotic services for Vietnamese veterans at the Veterans Administration Hospital at Castle Point, New York. By the end of December 1965, ten cases were fitted with braces. In the six-month period ending June 30, 1966 an additional 34 patients were fitted with braces. As of that date, 44 of the 56 patients had received braces. The braces fitted were similar to the typical braces shown in the above-mentioned Bulletin.

The next issue of the Bulletin will contain a detailed report of all the cases serviced from the onset of the program.

TABLE 9. — *VAPC Surgical Support and Elastic Hosiery Program*

Issuance channel	F.Y. 1963		F.Y. 1964		F.Y. 1965		F.Y. 1966	
	Surgical supports, ea.	Elastic hosiery, ea.	Surgical supports, ea.	Elastic hosiery, ea.	Surgical supports, ea.	Elastic hosiery, ea.	Surgical supports, ea.	Elastic hosiery, ea.
Directly to veterans	2,162	5,806	2,434	9,989	2,482	9,554	2,634	9,801
VA orthopedic shops	3,098	6,598	3,995	12,258	3,722	11,049	5,885	14,517
Total	5,260	12,404	6,429	22,247	6,204	20,603	8,519	24,318

TABLE 10. — *Distribution of Prosthetic Components by the VAPC, F.Y. 1966*

Issuance channel	Temporary prostheses				SACH feet		Hydraulic systems				Active wearers of hydraulic systems, end of F.Y.
	AK		BK				Initial issue		Replacement issue		
	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	
VA orthopedic shops	35	52	27	62	102	149	14	21	46	83	—
Commercial facilities	21	28	21	30	39	53	203	453	527	1012	—
Totals	56	80	48	92	141	202	217	474	573	1095	1719

TABLE 11. — *Complete Below-Knee Artificial Limbs Fitted by the VAPC, F.Y. 1966*

Type	Permanent		Temporary		Totals	
	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total
PTB						
Cuff	14	26	7	13	21	39
Lacer	32	76	11	15	43	91
Carved wood	4	17	0	0	4	17
Molded socket non-PTB	9	11	0	0	9	11
Syme	5	16	0	0	5	16
Chopart	0	2	0	0	0	2
Totals	64	148	18	28	82	176

TABLE 12. — *Complete Above-Knee Artificial Limbs Fitted by the VAPC, F.Y. 1966*

Type	Permanent		Temporary		Totals	
	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total	2nd half F.Y.	F.Y. total
Plastic sockets — non-total contact	11	22	18	28	29	50
Plastic sockets — total contact	31	55	6	7	37	62
Wood sockets	7	14	2	5	9	19
Hip disarticulation	2	4	0	0	2	4
Knee bearing	6	9	0	0	6	9
Totals	57	104	26	40	83	144

TABLE 13. — *Other Prosthetic and Orthotic Items Fitted by the VAPC, F.Y. 1966*

Period	Arch supports	Spinal braces	Leg braces	Leg-thigh braces	Upper-extremity prostheses
2nd half F.Y.	878	116	57	81	19
Total F.Y. 1966	1693	144	129	123	44

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