

Northwestern University Intermittent Mechanical Friction System (Disk-Type)¹

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The Northwestern University intermittent mechanical friction system was recently approved by the Veterans Administration for use by its above-knee amputee beneficiaries when appropriately prescribed. This device, developed at Northwestern University Prosthetics Research Center under Veterans Administration contract, is now in production. An extensive period of testing preceded its approval.

In December 1963 the first production prototypes were received by the Veterans Administration Prosthetics Center in New York City. These models, based on a Northwestern design previously tested in a limited laboratory setting, were then distributed to a number of research facilities in the Artificial Limb Program. Clinical trials followed, disclosing a number of defects which required the redesign of several parts of the system.

The manufacturer of the production prototypes made design changes to solve the problems noted in the clinical trials and in the laboratory studies conducted concurrently at VAPC. By March 1965 sufficient progress had been made to warrant placing the Northwestern University intermittent mechanical friction system in the "approved" category.

DESCRIPTION OF THE UNIT

Figure 1 shows the present system placed in a cutaway of a wood above-knee setup. Basi-

cally, the Northwestern University unit is a simple, multiple-disk friction brake installed in a conventional wood knee block and a shank of an above-knee prosthesis. Figure 2 shows the assembly of the unit to the setup.

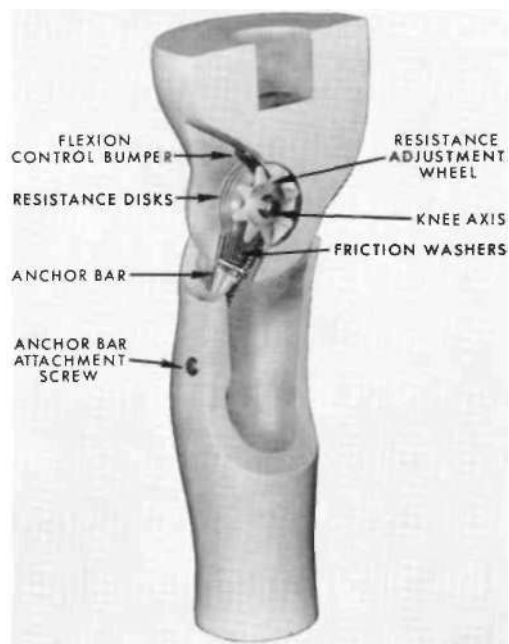


Fig. 1. Cutaway of knee block and shank assembly showing most recent (January 1965) version of Northwestern University disk-friction system.

The several disks have different angular sizes and thus provide a stepped alteration of the resistance pattern during the swing phase (Fig. 3). A particular disk contributes to the overall resistance to flexion and extension depending on its angular size: the larger the

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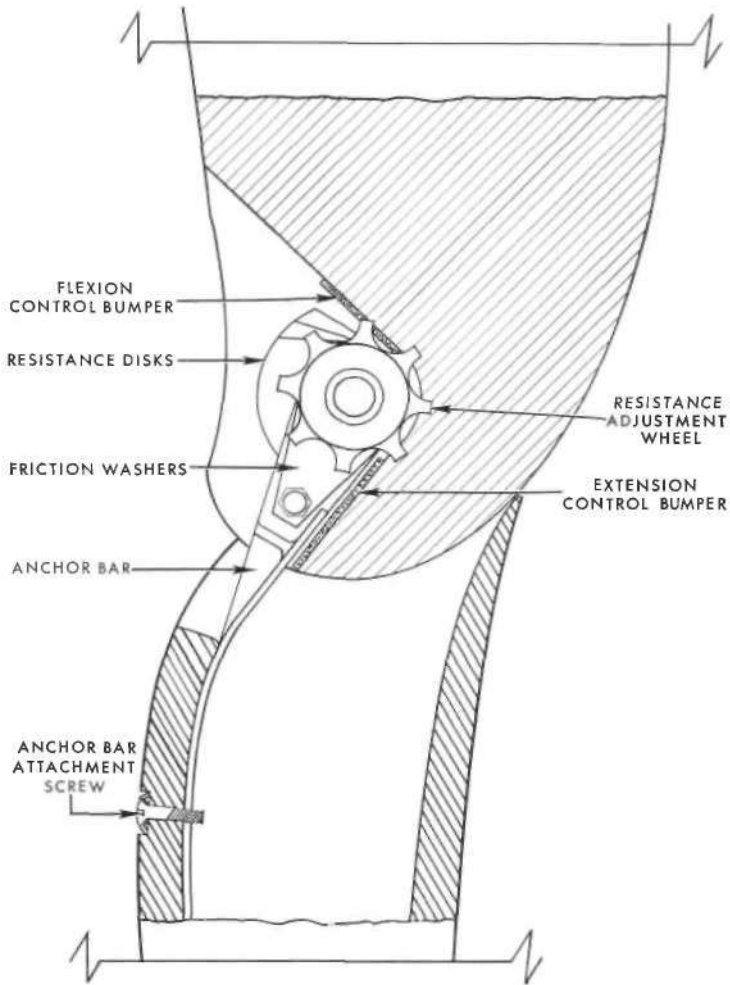


Fig. 2. Drawing showing position of Northwestern University disk-friction system in setup.

disk angle, the longer the disk will provide resistance during the swing phase. The edges of the disks anteriorly and posteriorly are driven by mating surfaces in the knee block, one surface rotating the disks in succession during flexion and the other during extension. The anchor bar is attached to the shank with the disk-friction unit placed over the knee bolt.

The disks⁴ are concentric with the axis of the knee bolt and are driven over friction surfaces also mounted concentric with the knee axis.

⁴ In the original models, three disks numbered 1, 2, and 3 were provided in the unit with a fourth (No. 4) as a separate component. The numbers represented angular size, No. 1 being the smallest and No. 4, the largest.

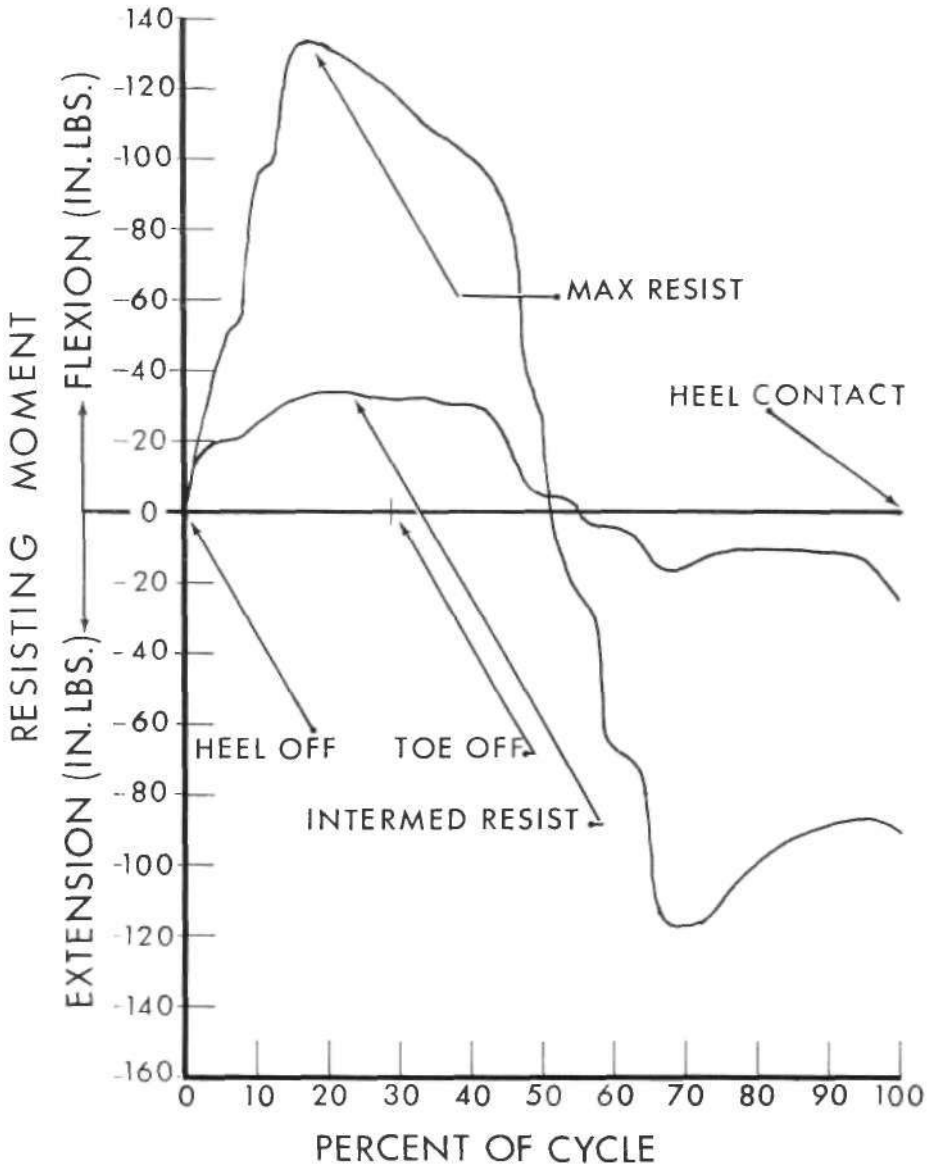


Fig. 3. Graphic record of stepped resistance pattern of Northwestern University disk-friction system during swing phase.

The total factional resistance can also be adjusted by turning a star wheel mounted concentric with the knee axis and thereby increasing or decreasing the force between the disks and the friction surfaces.

CLINICAL TRIALS

In the early clinical evaluations performed at New York University, at the Navy Prosthetics Research Laboratory, at Northwestern University, and at the Veterans Administra-

tion Prosthetics Center, uniform results were experienced. A total of nine units had been clinically tested among these groups, with structural failures occurring after approximately two weeks of use. But the frictional resistance pattern provided by the unit was found to be very desirable.

During the clinical trial period, seven of the units showed elongation of the leather friction disks and an eventual loss of friction (Fig. *i*). Four of the units experienced bending or frac-

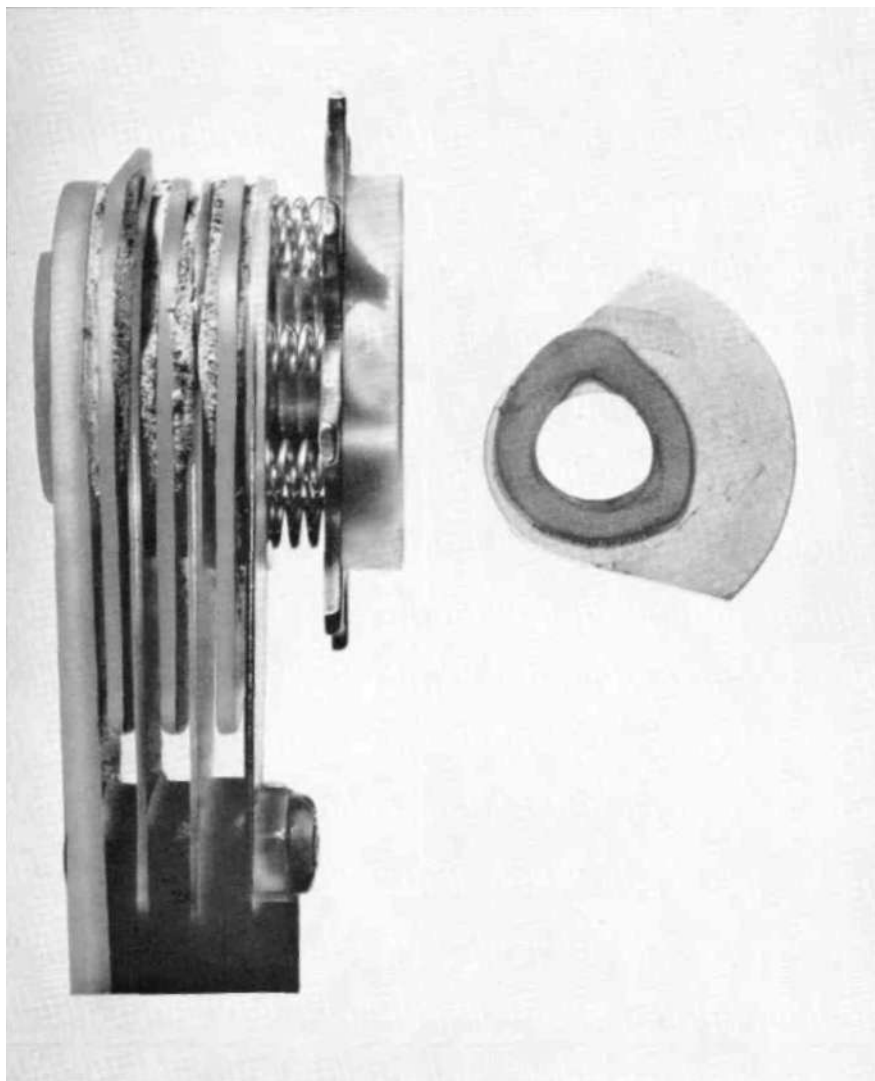


Fig. 4. Typical extrusion and elongation of leather washers in early clinical study.



Fig. 5. Typical failure of anchor bar in early clinical study models of the Northwestern University unit.

ture of the anchor bar (Fig. 5). Four showed deterioration of the control surfaces produced by penetration of the disk edges into the rubber bumpers. In one unit, the posterior aspect of the shank failed because the screw hole in the anchor bar was drilled too close to the posterior proximal brim of the shank. Objectionable noises resulted from these failures.

Recommendations made on the basis of the clinical trials focused on each of these problems.

It was suggested that redesign provide a reinforcing flange on the anchor bar, and a more gradual curve in the whole anchor-bar bend was also recommended. The sharp curve originally provided caused microscopic fractures in fabrication, which were instrumental in the ultimate failure. To help further reinforce the anchor bar, it was suggested that the posterior portion of the wood shank come

up higher than on the earlier unit-to-shank assemblies. A larger portion of this posterior section of the shank, especially if a metal plate were inserted internally, then could support the posterior aspect of the anchor bar and significantly reinforce it.

The leather friction washers were totally inadequate. Therefore, it was suggested that the manufacturer employ the findings of the Navy Prosthetics Research Laboratory on a brake-lining material which would not show the extrusion prevalent with the leather. As another possibility, Celastic, with which the Veterans Administration Prosthetics Center had some success, was suggested.

It was noted that the rubber bumpers used in the clinical trial samples were installed in a setup improvised by the prosthetist prior to fitting. It was agreed generally that the setup should not be developed by the local limb shop to accommodate the Northwestern University unit. Rather, setups should be constructed by the manufacturer with placement of the control surfaces properly standardized and dimensionally controlled to accept the unit. In the clinical sample errors were made in the angulation of the control surfaces against which the disks operate, with the result that the disks did not contact these surfaces properly, with disk edge parallel to the control surface. Penetration of the rubber bumpers resulted. Coincidence of the disk angle (when contacting the stop) with the stop angle at the point of contact is not always achieved in the shop-modified setup. The solution to the problem is achieved by having properly angled stops with rubber bumpers (preferably of about 95 durometer) in a preshaped, mass-produced setup. Increasing the width of the disks—but only at the outer edges of the disk segments where they come in contact with the rubber bumpers—would have also helped considerably.

The planes of the disks, during adjustment of the frictional resistance, did not stay parallel at all times. The problem seemed to be partly due to the lower machine screw, around which there were rubber spacers. As the friction adjustment wheel was turned, each of the disks tended to turn, and one or more of the disks engaged threads of the machine screw and bound, preventing parallel displacement of the

disks. It was suggested that this machine screw be replaced with a smooth shaft, with the thread cut only where needed for the machine nut. It was also suggested that the clearance holes in the disk plates be enlarged slightly and that the rubber spacers be of much lower durometer. In addition, it was believed that the multiple spring arrangement on the friction-adjustment wheel could be replaced with a Belleville spring.

Shank failures, of course, can be prevented if the friction units are provided installed in a shank, with appropriate control in locating the anchor-bar screw hole far enough away from the posterior proximal brim of the shank.

It was suggested that a fifth disk (No. 5)—representing the next larger angle, with an

angular increment the same as between No. 3 and No. 4—be provided. A wider range of control of the frictional pattern would thus be available to the prosthetist; for example, the combinations 1,2,3; 2,3,4; 3,4,5; or even 1,2,4; 1,2,5, etc.

LABORATORY TESTS

While the manufacturer was attempting to make some of the changes indicated, additional testing was performed by the Bioengineering Research Service at the Veterans Administration Prosthetics Center. These studies focused on two of the major considerations for the redesign: the loading on the anchor bar and the adequacy of various friction materials.

Studies to determine the probable causes of

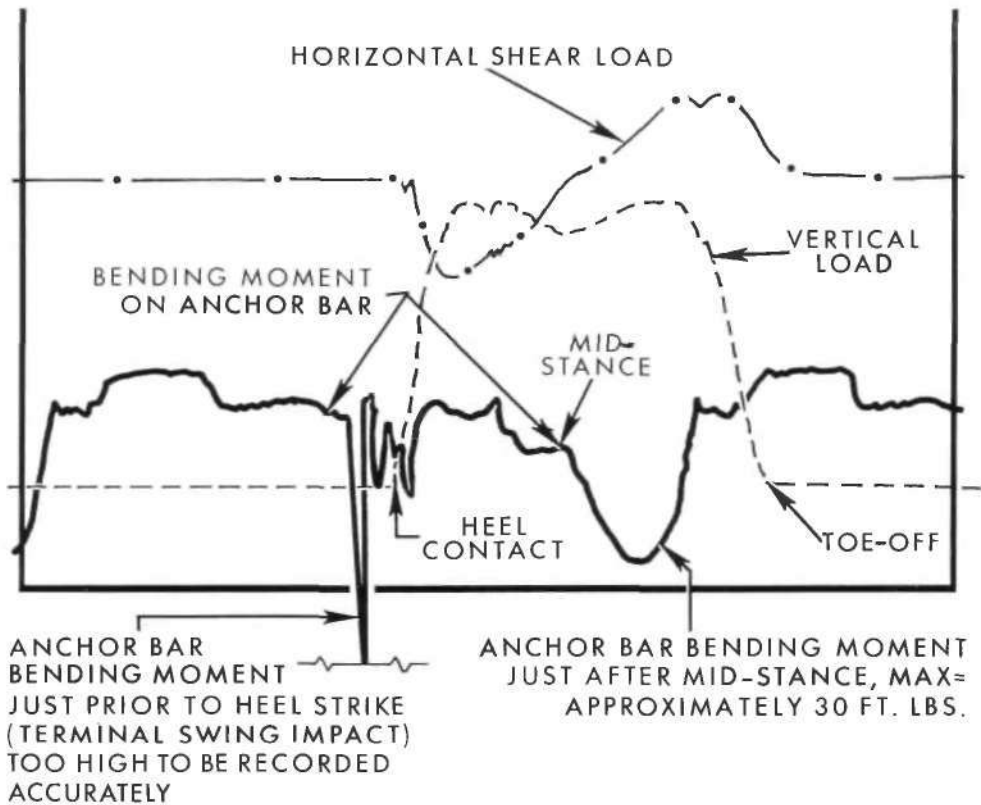


Fig. 6. Graphic record of bending moment on anchor bar of Northwestern University intermittent disk-friction knee.

failure in the anchor bar indicated that substantial forces were applied just after mid-stance and again at terminal impact in the swing phase (Fig. 6). The maximum bending moment in stance phase was approximately 35 ft. lb., a value which is equivalent to a tension of 5,600 psi at the surface of the bar. If the material of the original model of the anchor bar (unreinforced) approximates that of 4130 steel alloy with a working stress limit of approximately 20,000 psi, it would appear that the stance-phase loading falls well within the structural limits of the material. However, in swing phase, much higher loads may be imposed. Figure 6 shows a barely recorded shock load at terminal impact. Obtaining a true value for this load was impossible with the equipment at VAPC. The faint trace indicates that the galvanometer deflection lagged far behind the very rapid input. Therefore, the recorded spike may be of far lower magnitude than the actual loading. If the terminal shock loading was three or four times as great as the stance-phase loading, as it well may be, the working stress limit of the original material might easily have been exceeded.

To evaluate the utility of the resistance mechanism, a subject was fitted with one unit, and an optimum resistance setting was determined by adjustment of the friction disks. From an initial minimal setting, resistance was increased in increments of approximately one-eighth turn of the adjustment wheel. The subject's gait was observed after each change through the entire range. The optimum was found by backing off from maximum resistance to a point where the subject's best gait was achieved. This setting produced an obvious reduction in the terminal impact and heel rise as compared to that observed with minimum resistance. However, the adjustment range was very narrow, being limited to approximately one-quarter turn.

The optimum resistance setting was indexed, and the unit was removed from the setup and installed in the Veterans Administration Prosthetics Center's testing machine. Drop tests were performed at:

1. The optimum resistance setting.

2. A minimum resistance setting determined by adjusting the friction to the lowest increment above a free-falling knee.

3. A maximum-resistance setting determined by adjusting the friction to the highest increment that permitted motion under the 50 in. lb. of applied torque.

The following times in seconds were recorded:

Minimum		Optimum		Maximum	
Flexion	Extension	Flexion	Extension	Flexion	Extension
.22	.24	.25	.27	.27	.30

These data indicate that there is very little resistance-adjustment range between the minimum and maximum settings. Although not directly comparable but as a matter of perspective, these values fall just below the order of minimum-resistance values provided by several hydraulic units. Among five hydraulic systems, minimum drop times in flexion ranged from 0.20 sec. to 0.32 sec. and in extension, from 0.19 sec. to 0.31 sec.

At the same resistance settings used in the drop tester, the following maximum knee moments in inch-pounds were recorded at 43 cycles per min.⁵ on the UCB testing machine.

Minimum		Optimum		Maximum	
Flexion	Extension	Flexion	Extension	Flexion	Extension
4.0	12.5	32.0	25.0	50.0	37.0

The values obtained for minimum-resistance settings of four hydraulic units fell between the optimum and maximum values for the Northwestern University device.

In an effort to determine how a given resistance setting is maintained in use, the unit was cycled 10,000 times at the optimum-resistance setting. After every 500 cycles, drop-test times and knee moments were determined. The data in Table 1 were recorded.

The drop-test results indicate that no significant changes occurred in the resistance values after cycling.

After 10,000 cycles, knee moments at the minimum-resistance setting for flexion and extension were respectively 4.0 and 12.5 in. lb.; at the maximum-resistance setting the values were 50.0 and 37.0 in. lb. These data also indicate the durability of the resistance setting with use.

These results were obtained with the hard-anodized friction disks and a flat, woven, brake-lining material. Other materials, including Nylatron, Celastic, and a laminated brake

⁵ Equivalent to a cadence of 86 steps per min.

TABLE 1

No. of Cycles	Time (Sec.)						Knee moment (in. lb.) at 43 cycles per min.	
	Minimum		Optimum		Maximum		Optimum	
	Flexion	Extension	Flexion	Extension	Flexion	Extension	Flexion	Extension
500	.22	.24	.25	.27	.27	.30	32.0	25.0
1,000	.22	.24	.25	.27	.27	.30	32.0	25.0
5,000	.22	.24	.25	.27	.27	.30	32.0	25.0
10,000	.22	.24	.25	.27	.27	.30	32.0	25.0

material, were tested in the same way and found to have serious deficiencies. The woven brake lining provided a higher coefficient of friction and more resistance to bending and tension than all other materials tested.

After the entire series of machine tests, the unit was replaced in the subject's prosthesis at the originally determined optimum-resistance setting. He used the unit in and around the laboratory at the Veterans Administration Prosthetics Center for approximately six hours. Observations indicated that the subject walked as before and that no further resistance adjustments were required. However, the subject commented to the effect that the knee swing was smoother; "high spots" previously noted during the swing phase were absent. This observation was attributed to "wear in" of friction materials and to the earlier misalignment of individual friction components. A poor class of thread tends to "cock" one bearing segment against another; after a certain amount of wear, plastic deformation increases the area of surface in contact and "smooths" performance.

These tests performed late in 1964 substantiated the problem with the anchor bar. The manufacturer was notified that greater strength should be incorporated in the anchor bar; and he provided the flange on the anchor bar shown in Figures 1 and 2.

The magnitude of the frictional resistances available was comparable with other mechanical frictional units and also with a lower range of resistance available in hydraulic units. Resistance settings seemed to be maintained over a reasonably long cycling period with the

woven brake-lining material. But since the resistance range was still found to be inadequate, a heavier compression spring was recommended to increase adjustability.

Because of the very high terminal load on the anchor bar, which can affect the durability of the bushings in the setup, a more resilient extension bumper was suggested. In addition, it was strongly recommended that impact load conditions could be improved by placing a resilient material between the anchor bar and the shank at the point of attachment.

As a result of the manufacturer's adoption of these recommendations and the extensive laboratory testing performed on the system, it was concluded that the Northwestern University disk-friction unit would provide a very desirable variable swing-phase control which is simple and, therefore, should be inexpensive compared to other types of units providing similar function. The resistance is easily adjustable, not only in terms of overall magnitude, but also by allowing the prosthetist to vary the number of steps in the resistance pattern. There still may be minor maintenance problems, which are common to mechanical friction systems. Hopefully, the manufacturer can solve these problems if they do indeed occur.

The Veterans Administration has drawings suitable for production purposes which will be made available to recognized component manufacturers upon written request to the Director, Prosthetic and Sensory Aids Service, Veterans Administration Central Office, Washington, D. C.