DESIGN OF A MAGNETIC PARTICLE BRAKE ABOVE-KNEE PROSTHESIS

by

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ABSTRACT

Innumerable above-knee prostheses have been designed and tested over the past several hundred years, ranging in complexity from the simplest peg-leg to elaborate hydraulically damped polycentric devices. This thesis is a progress report of a new prosthesis which has been designed for initial implementation of control schemes derived from work done on a man-interactive A/K prosthesis simulation system at M.I.T. This also uses a magnetic particle brake for passive torque during swing phase. It has been designed so that knee torque is isolated throughout the entire walking cycle. Initially the prosthesis will use a simple damping scheme with damping proportional to angular velocity, but work is continuing to develop an on-board microprocessor based controller. By using a torque feedback scheme, the prosthesis will function at multiple cadences.

In this thesis a fundamental description of the parameters of normal gait is given. It is shown that the net power required in the knee joint over a cycle is negative. A brief survey of previous A/K prostheses is presented along with some of the parameters necessary for a successful prosthesis. The new prosthesis design is described in detail, along with a description of work to be done.

Thesis Supervisor: Woodie C. Flowers
Title: Associate Professor of Mechanical Engineering

ACKNOWLEDGEMENTS

My involvement with this project was due to a fortunate series of events starting at the M.I.T. Innovation Center, and so I'd like to begin by thanking everyone concerned with them for this and many things. Also, to the following people, more than just thanks are due:

To Woodie, for his design ideas and inspirations, for his higher than high standards, and for rescuing me from a previous project.

To everyone in the M.I.T. Knee Group for their valuable input at meetings and other times, and especially those people past and present upon whose previous work I so heavily depended.

To Hal Robinson, Walt Haimberger, and T. Wally Williams for advice, materials, equipment and open ears.

And, to both Debby and my mother, for patience, understanding and pretending that loud noises I made in the middle of the night didn't bother them.

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1. INTRODUCTION

Undoubtedly one of the most traumatic experiences for a person is losing a leg. In addition to the physical liabilities that result, such as loss or impairment of mobility and pain, are the psychological effects resulting from a changed physical appearance, possible loss of the ability to work for a living at a mormal occupation, as well as other social problems.

A prosthetic device should restore, as much as possible, the functions of the original limb. For above-knee (A/K) amputees, it is essential that the prosthesis allows level walking with a minimum of effort, a maximum of safety, and with as close to a mormal appearance as possible.

Countless attempts have been made over a time span of hundreds of years to construct better leg prostheses, but it was not until recently that any really systematic studies of human gait and optimal prosthetic design were undertaken.

This thesis is a preliminary report on the design of an A/K prosthesis which will be used to apply knowledge gained from two A/K prosthesis simulator systems now being used at M.I.T. (6),(7),(10),(11). Unlike these systems, it is to be totally self-contained, permitting testing in day-to-day use, a more natural environment than a laboratory.

2. BACKGROUND

2.1 The Walking Cycle

Walking is a far more complicated action than it initally may appear to be, as the number of extensive studies undertaken in the last twenty-five years or so helps indicate. Researchers have attempted to measure the parameters of level walking in order to produce a more thorough description of normal gait, to be able to determine when and why gait is abnormal and thus, of special interest to this project, carefully define parameters necessary for an above knee prosthesis that helps restore, as much as possible, normal gait.

During walking, each leg goes through a period when it is in contact with the ground, called stance phase, and a period that is essentially a follow-through, called swing phase. The stance phase accounts for approximately 60% of the walking cycle, and the swing phase the remaining 40%. Figure 1 shows one full walking cycle, with time normalized to percent of cycle, in order to permit easier correlations of events at different walking speeds, or cadences.

If we begin with right heel strike, and call hip, knee, and ankle angles zero, we see that the right leg is extended

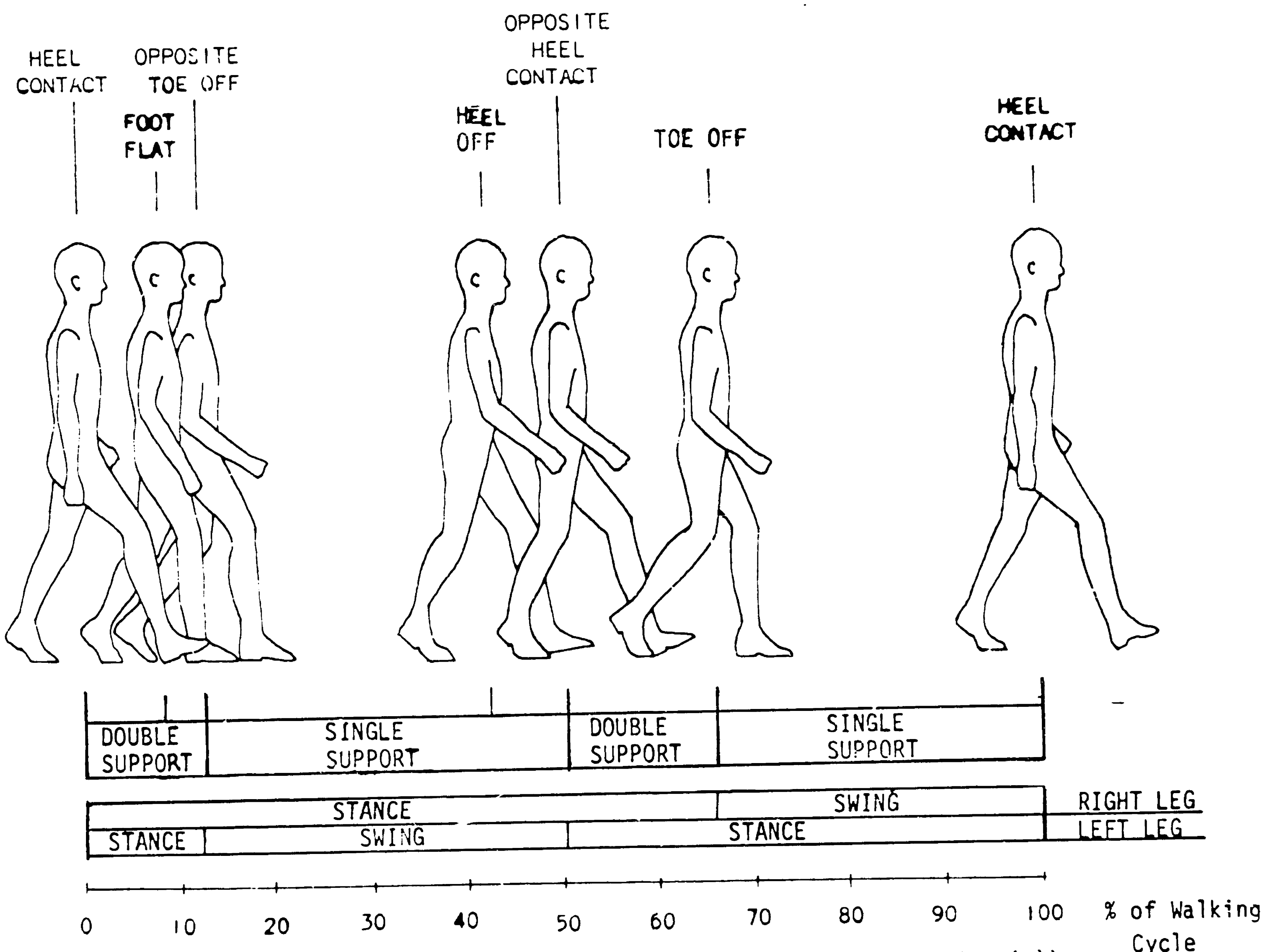


Figure 1 Sequential Gait Parameters (Donath(5), as modified by Grimes(8))

forward. This has been accomplished by moments at the hip and knee. Immediately the ankle flexes, allowing full foot contact as weight is transferred from the other foot. The knee flexes slightly under this load, then extends as the hip continues to flex. Now the knee is pushing against the load, and requires positive power. This flexion-extension process serves to prevent the body center of gravity from rising and falling too much during walking, which helps minimize energy expenditure. The knee then flexes rapidly as the right heel leaves the ground, and it is at this point that the maximum power is exerted by the hip and ankle to propel the body forward. The beginning of swing phase is signalled by the toe leaving the ground. The knee continues to swing up to its maximum flexion, with the muscles serving to brake the lower leg. Here power is dissipative. The hip extends, which, along with gravity, propels the swinging leg forward again to another heel contact. Figure 2 is a diagram of knee moment, angle and power for one stride of four normal subjects. If knee power is integrated over a stride, it can be seen that the net energy is negative. This has some implications which will be considered later.

It is interesting to consider that this motion happens very smoothly with, of course, corresponding movements in the upper body, and with minimal effort and conscious coordination. Few features of the human body can be imitated by

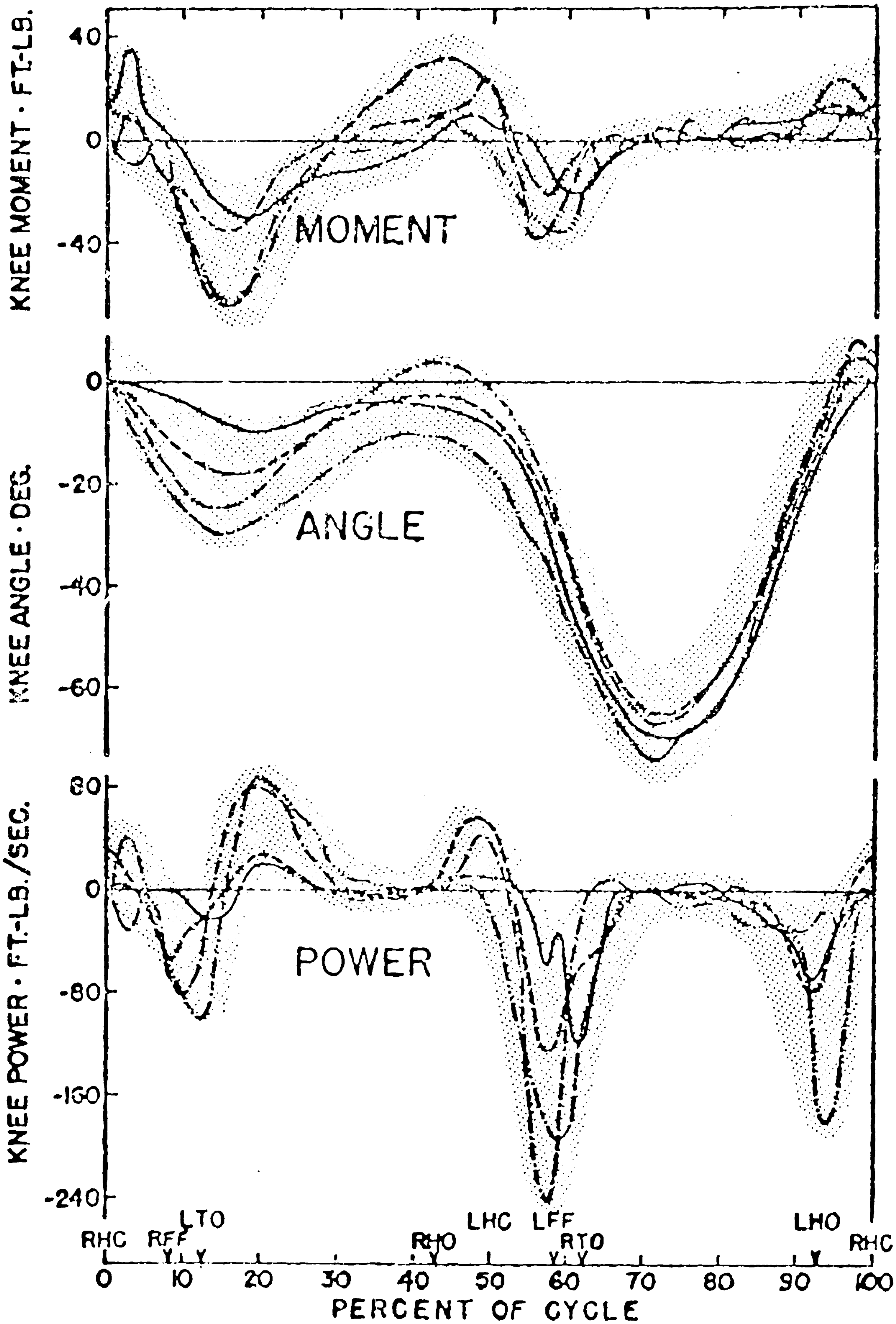


Figure 2 Knee Moment, Angle and Power - Four Normal Subjects (Bresler and Berry (1), as modified by Grimes (8))

designers with the degree of skill that nature uses so casually. It can be seen that duplicating exactly the function of a leg is a formidable task.

2.2 Prosthetic Devices

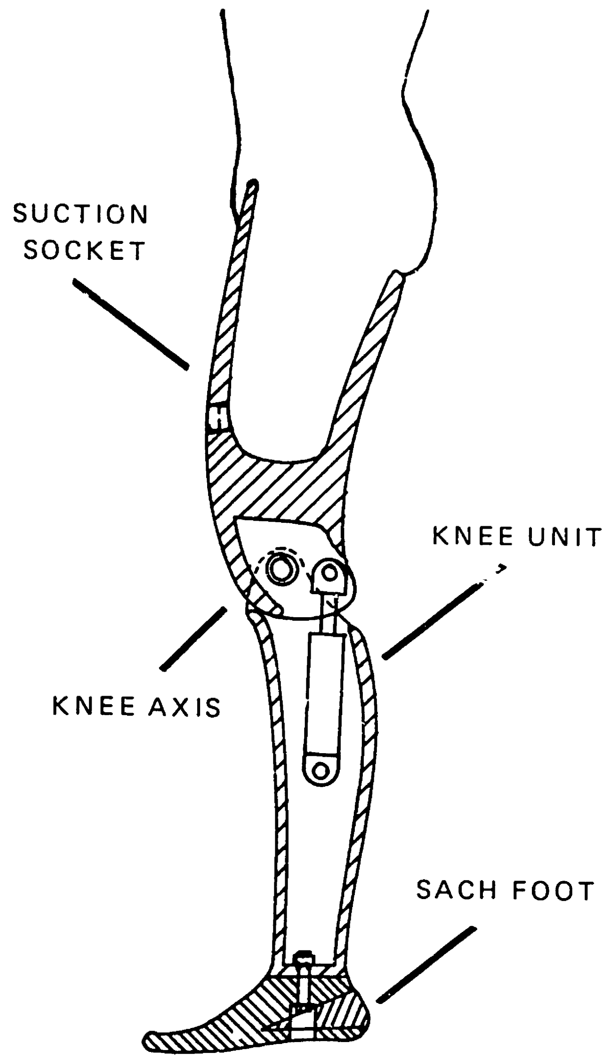
2.2.1 Commercial A/K Prostheses

Currently used A/K prostheses have met with only marginal success, though in most cases they at least restore limited mobility. However, an A/K amputee expends up to 65% more energy during walking than normal humans (2), has a lower walking velocity, and takes wider and shorter strides. Most A/K prosthetics lock during stance phase, so that the flex-extend-flex motion that helps keep the body center of gravity relatively level doesn't occur, and the amputee tends to "vault" over his prosthesis. This gives amputees a characteristic unnatural appearing gait. Figure 3 shows a typical A/K prosthesis.

Standard prostheses can be divided into three main categories of varying complexity as follows (15):

1. Free Knee
2. Swing Phase Control
3. Swing and Stance Phase Control

The simplest prostheses are of type 1, free swinging. These are basic, unadjustable, free swinging prostheses that are essentially cosmetically shaped wood, with any damping caused solely by friction inherent in the knee joint alone. These,



(a) Cross-section

(b) Knee unit

Figure 3 The Conventional Prosthesis (Donath (5), Lampe (10)) as well as those of type 2, hyperextend during stance for stability. Figure 4 is an illustration of hyperextension, showing that the load is forward of the knee axis, and that moments act to "lock" it in place against end stops.

Type 2 prostheses use any of a number of devices to control resistive torque at the knee axis such as rubbing discs, dashpots or pneumatic dampers. These can be controlled so that damping is constant, proportional to angular position, velocity or powers of velocity. These devices generally must be adjusted to change the cadence at which an amputee will feel most comfortable. Type 3 prostheses also use similar swing phase controls, but aid stability during stance phase through either increasing resistive torque during weight bearing or having polycentric linkages which kinematically assist stability. More complete information can be obtained from references (15),(16),(17) in the bibliography.

2.2.2 Prosthesis Research

Perhaps one of the key reasons that prosthesis research and development has proceeded so slowly is the great expense in time and money necessary to implement each new idea. Typically a time period of several months elapses before an idea can be translated into hardware that can be tested. This "build it and try it" philosophy can give useful results, but it is slow, as are most trial and error procedures. Another technique is computer modelling, but there are inherent difficulties in trying to model any man-machine interface, and

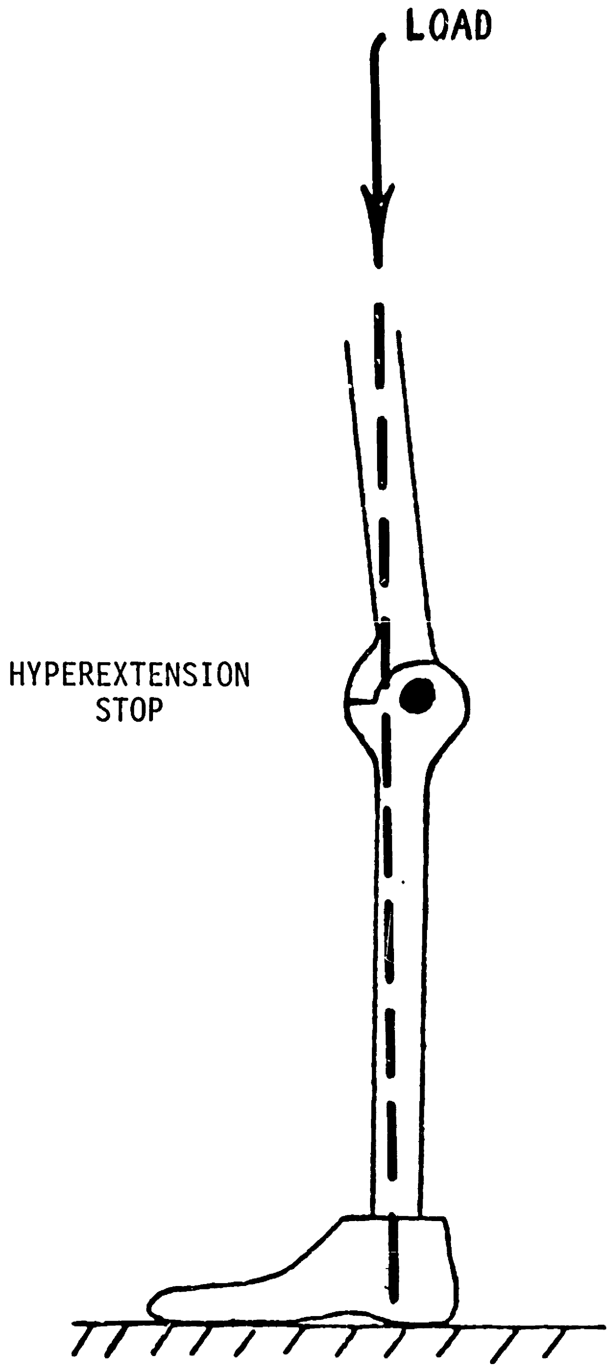


Figure 4 Hyperextension Gives Stability During Stance

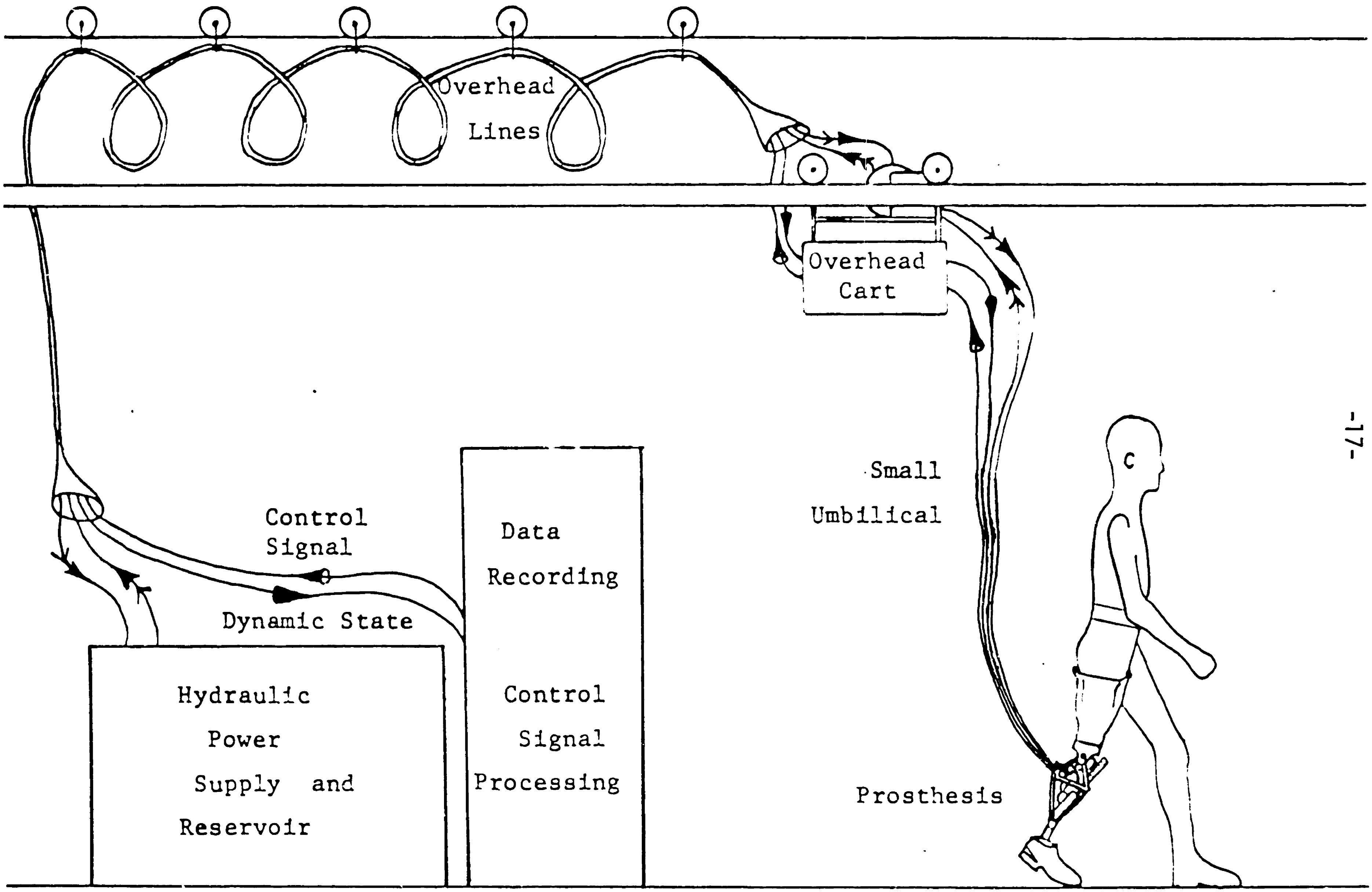
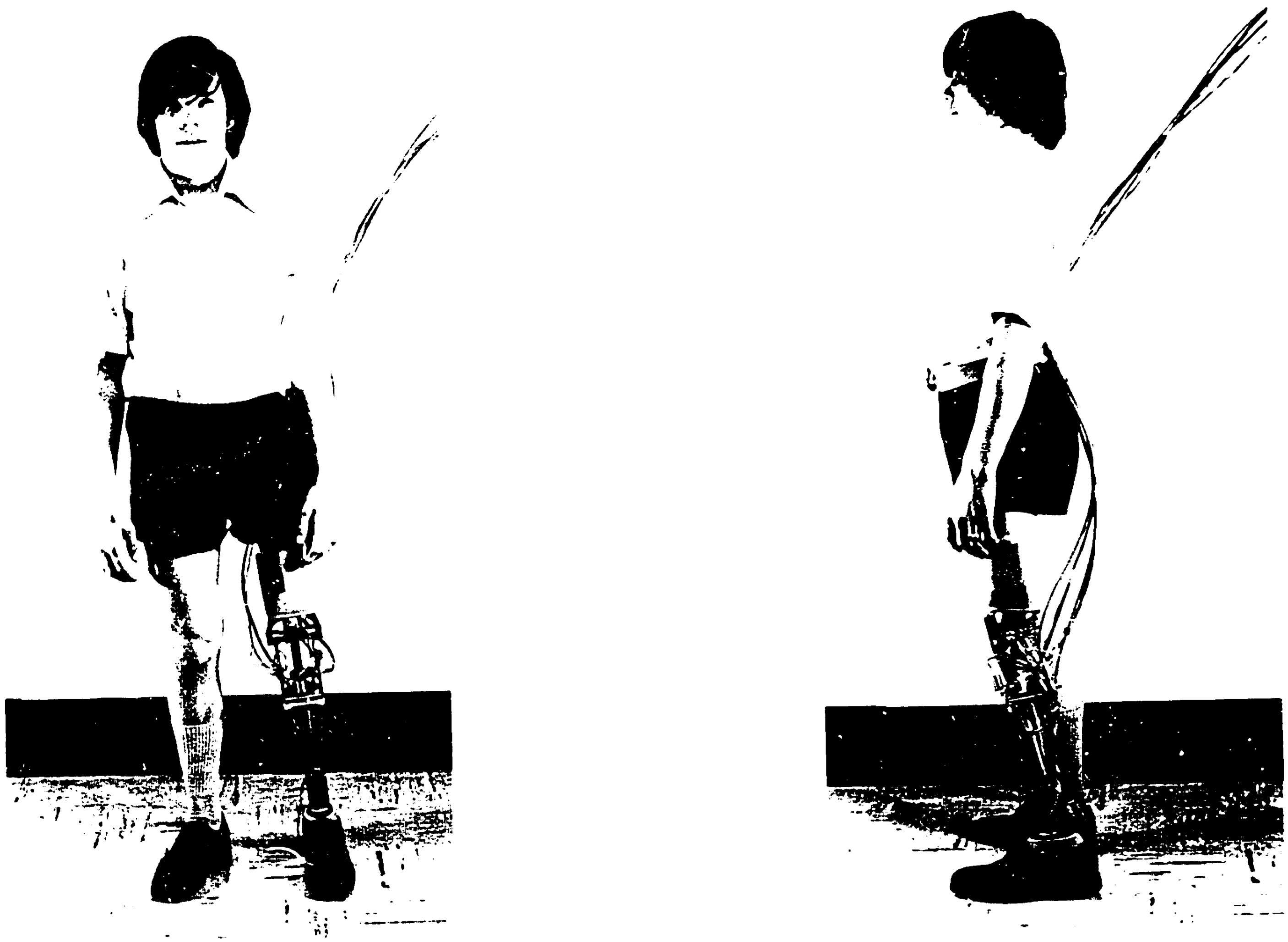


Figure 5 The Electrohydraulic Prosthesis Simulator System (Grimes (8))

Front View

Side View

Figure 6 Amputee Wearing Hydraulic Prosthesis Simulator (Grimes (8))



Front View

Side View

Figure 6 Amputee Wearing Hydraulic Prosthesis Simulator
(Grimes (8))

results must be tested with an actual prosthesis anyway.

Since 1972, two man-interactive prosthesis simulator systems have been developed at M.I.T. (6),(7),(9),(10). Figure 5 is a system diagram of the first of these. Because it is interfaced with a digital computer, this hydraulically-actuated prosthesis simulator can give any resistive torque profile desired during the entire stride, and can be modified on line in response to the amputee test subjects' comments. Results can be obtained much more rapidly this way. The amputee is free to move around in a large, unobstructed walkway while tests are being performed. Figure 6 shows the prosthesis being worn by a test subject. The prosthesis is powered and uses precise feedback techniques for control. The fact that it does have positive power has enabled it to be used for proportional EMG control research (5), and to duplicate the flex-extend-flex action occurring in normals during stance phase (8), helping to reduce energy expenditure and aid in the gait appearance. Results to date with the system have been excellent.

A more recent man-interactive prosthesis simulator has been developed for passive torque studies (9),(10), employing a magnetic particle brake for damping. The magnetic particle brake is capable of more accurate control than the hydraulic system, and is more easily interfaced with a digital computer. Figure 7 is a diagram of this system; figure 8 is a photograph of a test subject wearing the prosthesis. Some of the

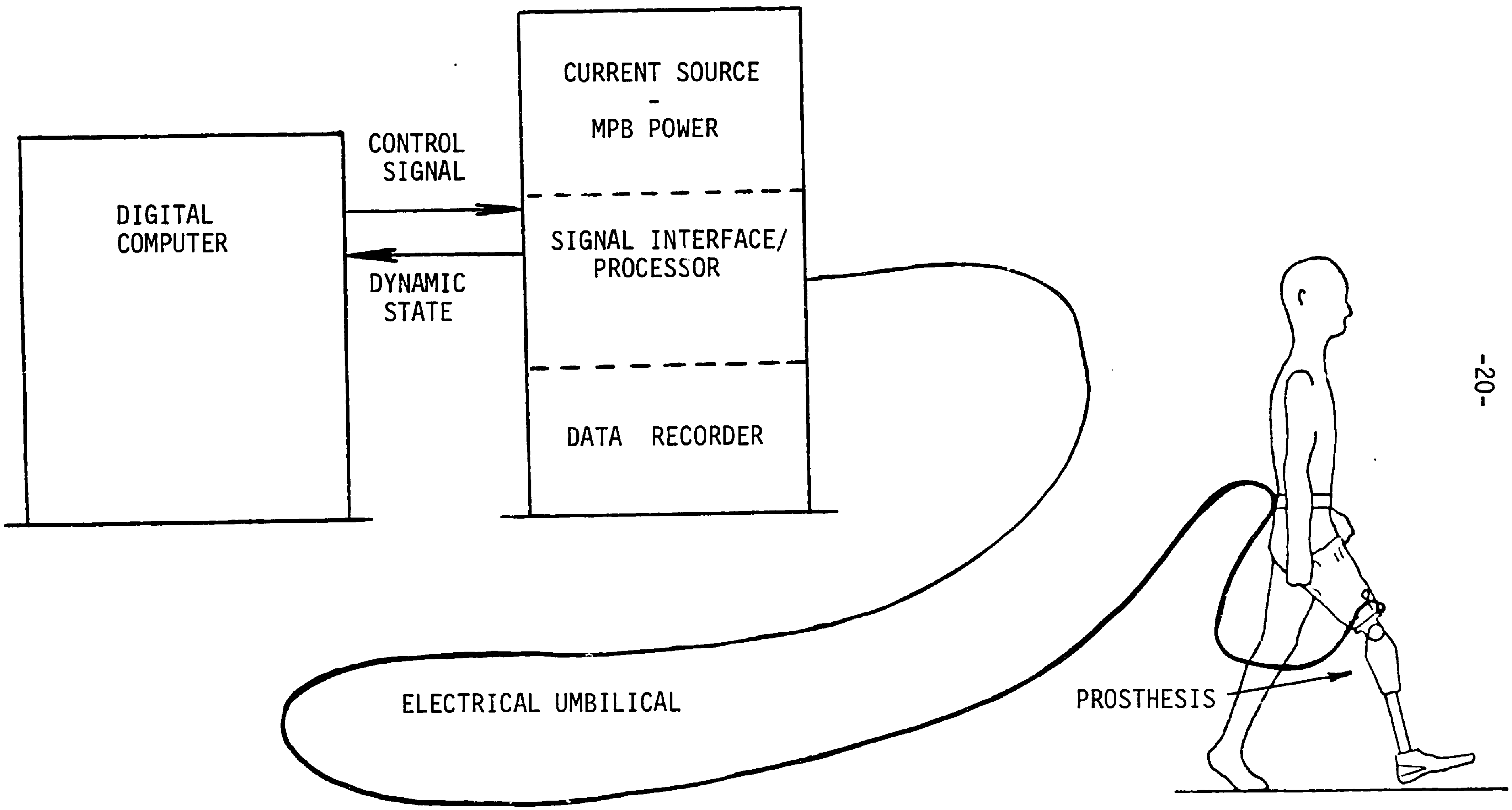


Figure 7 The Magnetic Particle Brake Prosthesis Simulator System (Lampe (10))

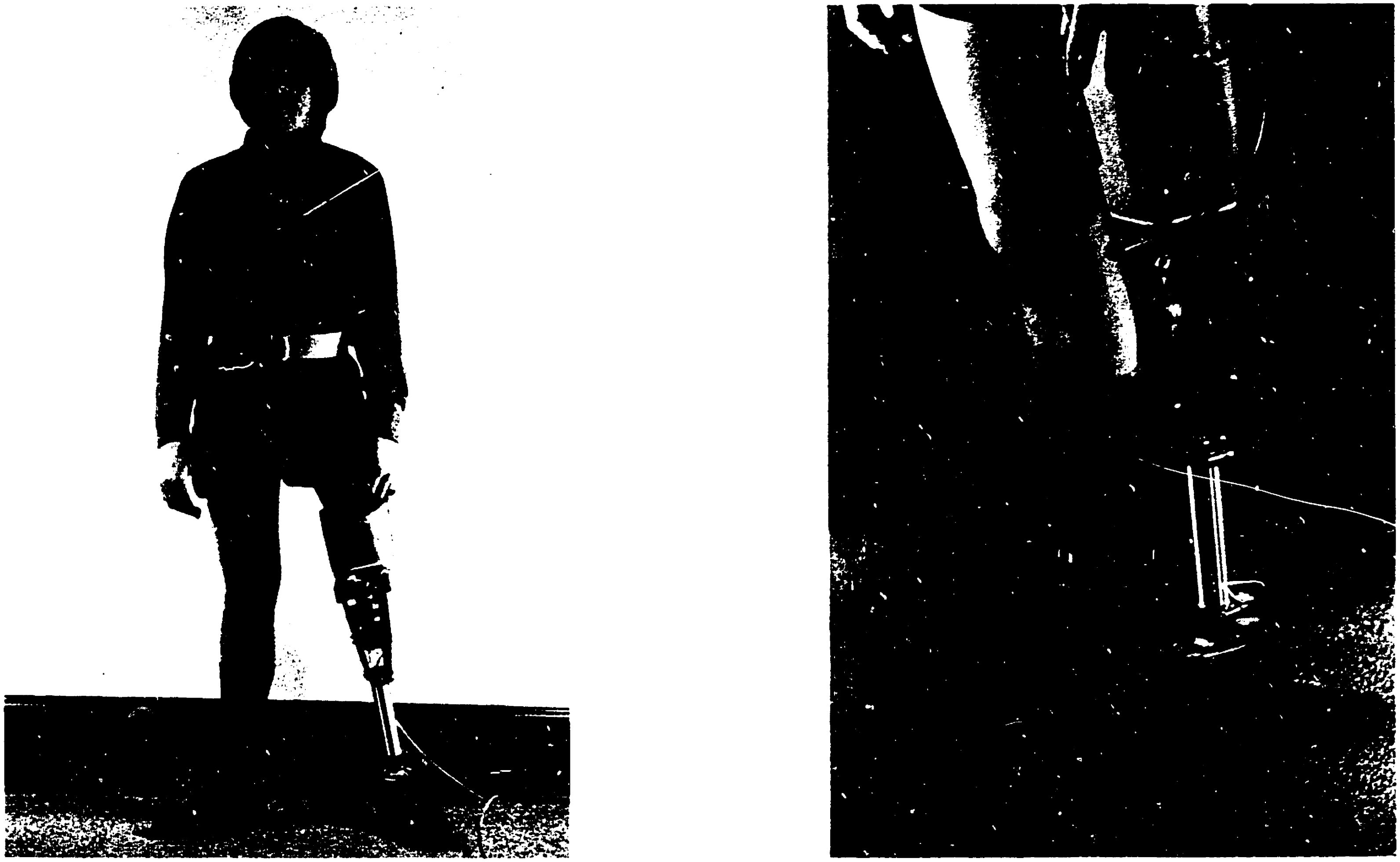
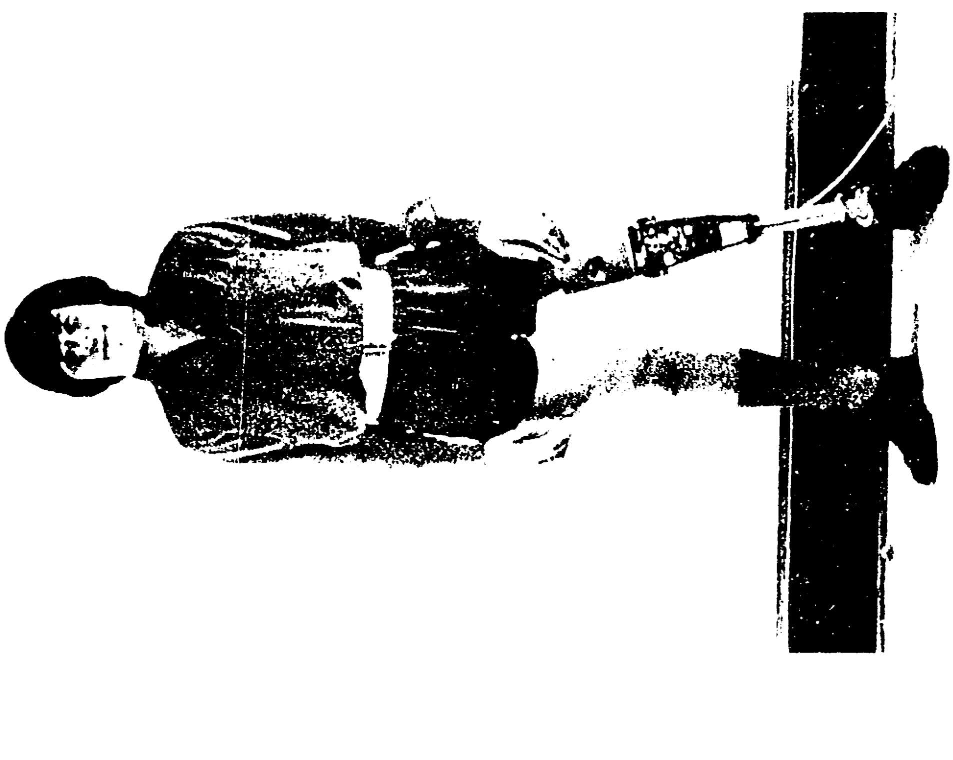


Figure 8 The Amputee Test Subject Wearing the Instrumented Prosthesis (Lampe (10))



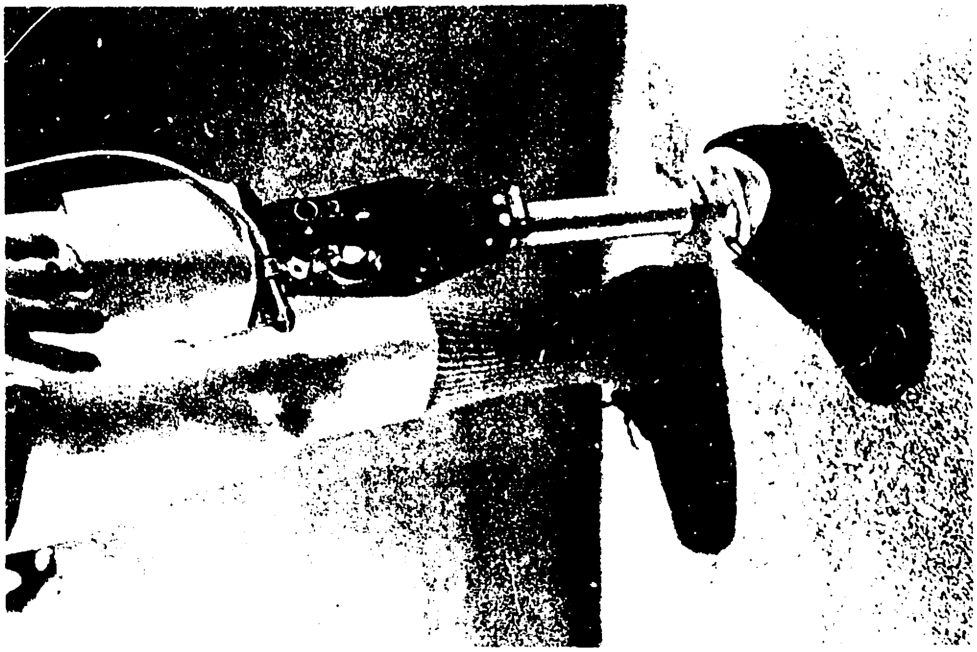


Figure 8 The Amputee Test Subject Wearing the Instrumented Prosthesis (Lampe (10))

preliminary work done on this system used ideal torque profiles simulated on the digital computer (10). Both simulator systems are well instrumented, with knee torque, position and velocity variables easily obtained in multi-channel strip chart recordings along with stride timing information such as heel contact (HC), foot flat (FF), heel off (HO), and toe off (TO). Timing information is obtained from foot switches located on the bottom of the prosthetic SACH (Solid Ankle Cushioned Heel) foot.

2.3 General A/K Prosthesis Requirements

The following is a very brief listing of some of the parameters necessary for a functional A/K prosthesis plus additional requirements specific to the research applications in which this particular prosthesis will be used.

- Conventional prostheses weigh between six and eight pounds, excluding shoe, but much weight can be saved through careful design and choice of materials. A lighter prosthesis conceivably would have many advantages in a research situation, since this allows an adjustable center of gravity, for example, through the use of small internal weights. Damping power requirements are lower, and the weight budget can include more battery and controller weight if necessary.

- Many prosthetics have limited height adjustment, but it is essential that a research oriented prosthesis be easily adjustable. In particular, this prosthesis must interface with 1.5" I.D. shank tubes, since this is standard at the M.I.T. laboratory. In addition, the suction socket (for the stump) should be interchangeable.

- To be truly versatile and thus capable of sophisticated control, a magnetic particle brake will be used, since it puts out a resistive torque reasonably proportional to current applied. A means of measuring torque at the knee axis will allow a torque feedback loop to take care of MPB non-linearities. Variables such as angular position, velocity and foot timing information are also essential in the laboratory. Studies have shown (2),(14) that a torque range capability of 0 - 25 ft-lbs is more than sufficient.

- The prosthesis should be similar to regular prostheses in overall diameter, and should have a good overall appearance. It should rotate a minimum of 90°, and have approximately 3° of hypertension for stance phase stability. Since it won't always be used in a laboratory environment, a reasonable amount of durability is a must.

3. DESIGN DESCRIPTION

3.1 Overview

This thesis details the design and construction of a lightweight, self contained, magnetic particle brake damped above knee prosthesis capable of precise control. Figure 9 is a photograph of the section of the most interest to this thesis- the knee joint. A magnetic particle brake was chosen for damping because it gives a resistive torque proportional to applied current, and is easily interfaced with a digital computer. In addition the M.I.T. MPB prosthesis simulator(10) uses a similar brake- it is important that the new prosthesis be dynamically similar to it so that new control schemes can be implemented more easily. The particle brake is geared up at a 7.5:1 ratio by a zero backlash multiple-cable drive which is shown in figure 10. Figure 11 shows the knee joint without the outer shell, and illustrates the dual, inner-outer structure used for torque isolation. The shell attaches between the top bracket and the bottom bracket (labeled "To Shank and Foot"), and thus is the load bearing portion of the device. The central torque tube-pulley bracket structure allows all of the torque to be measured independent of the

Figure 9 Assembled Prosthesis Without Shank and Socket

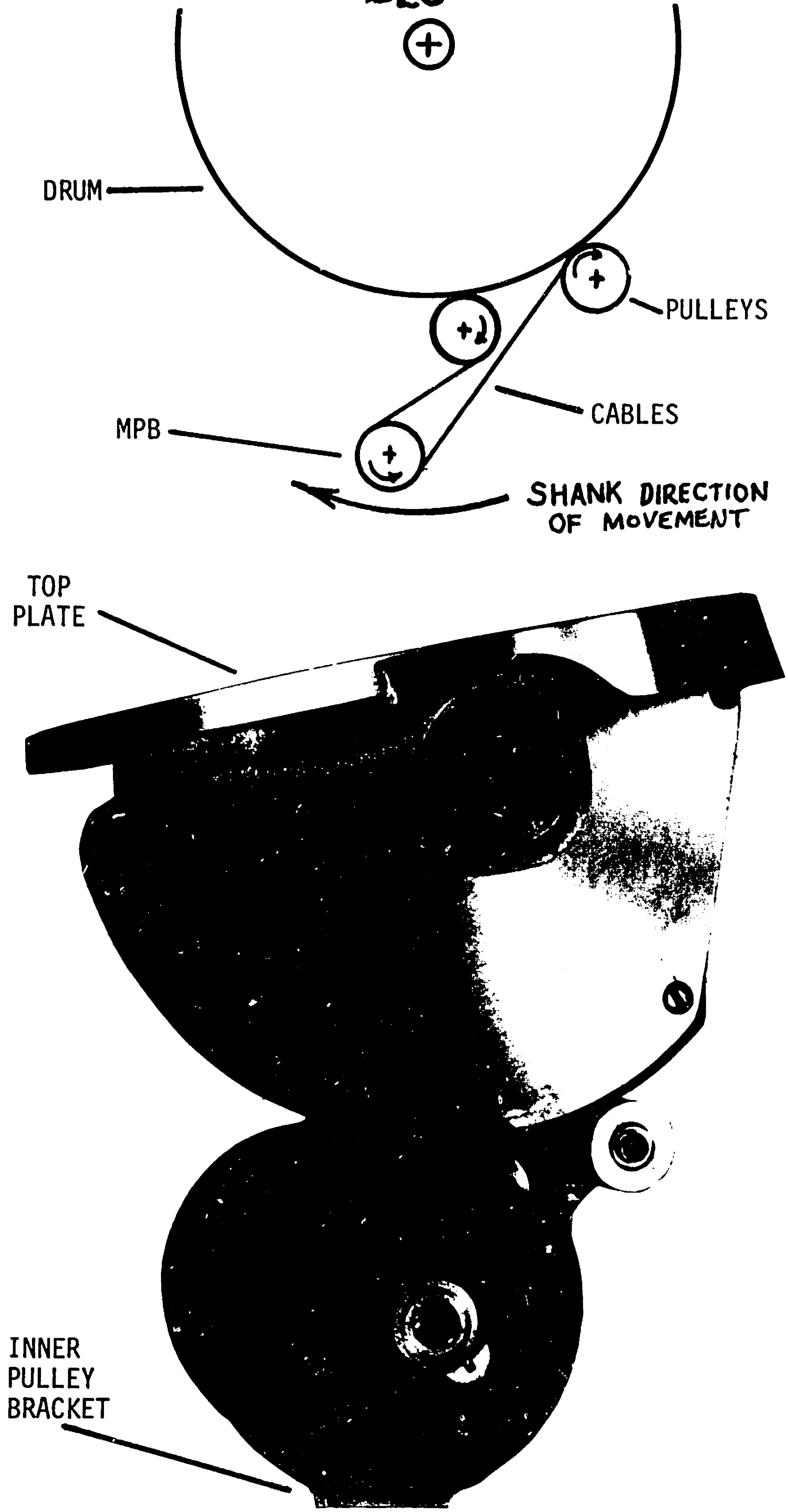


Figure 10 Side View of Cable Transmission

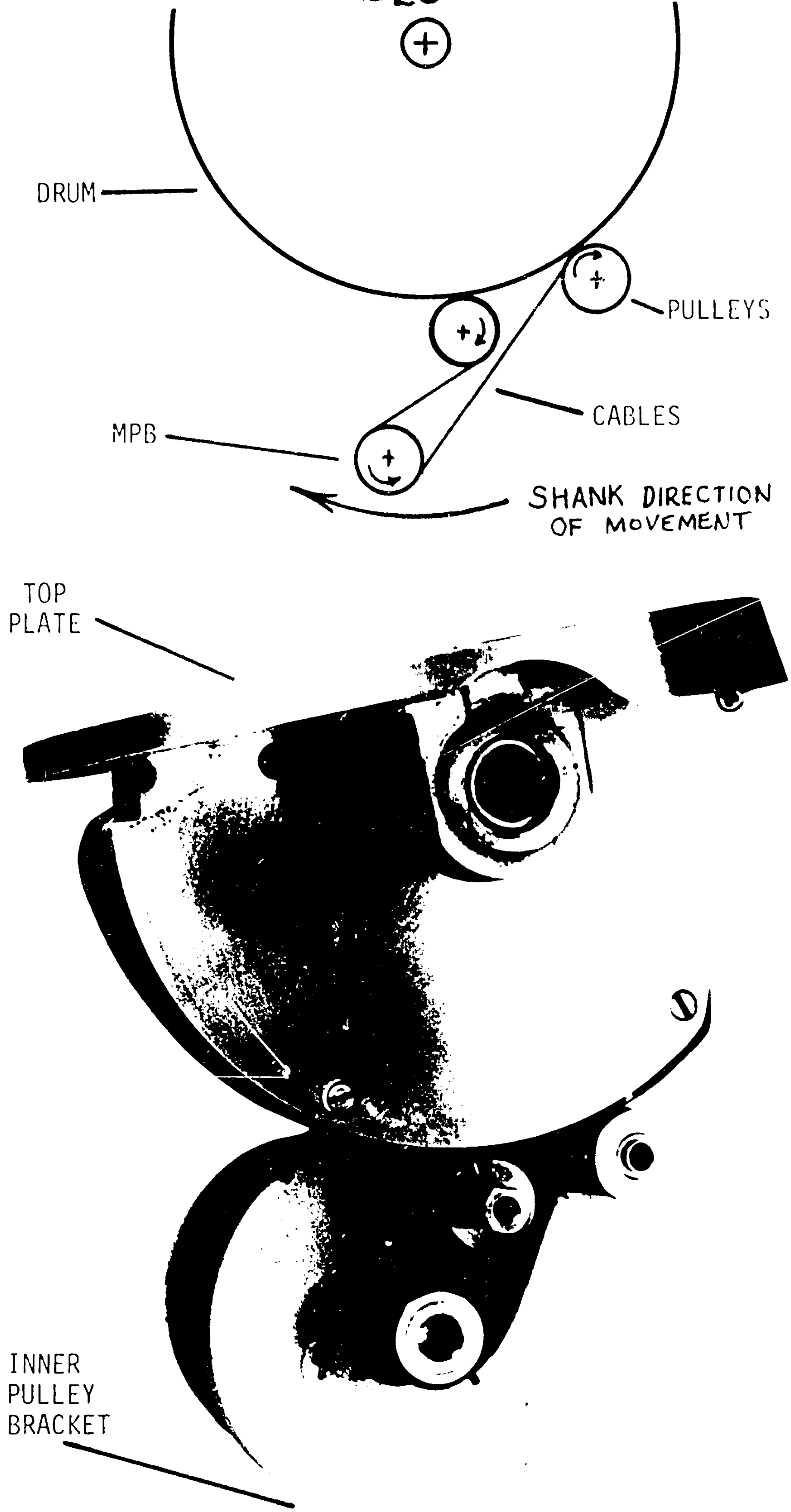


Figure 10 Side View of Cable Transmission

load on the outer shell. Figure 12 has two side views of the assembled knee joint, showing full flexion and extension, and figure 13 is simply a rear view. Figure 14 shows a tubular shank with a standard SACH foot, that attaches to the bottom of the knee joint with an automotive hose clamp. A more elegant clamp is being made by a member of the M.I.T. Knee Group, and it is hoped that it will be ready for use soon.

A few specifications and features of the present project are given below:

Height- Knee axis to top of mounting plate = .75"

Height- Knee axis to top of shank tube = 8 3/16"

Maximum Diameter (at knee axis) = 4.5"

Minimum Diameter = 2.5"

Overall Weight (excluding shank and foot) 2.8 lbs.
(This weight is without onboard batteries)

Weight (also excluding MPB weight) 1.5 lbs.

Power Required- .6 lb. nicad battery @ 24 volts
(Est. of 8 hr walking - does not include any controller power requirements)

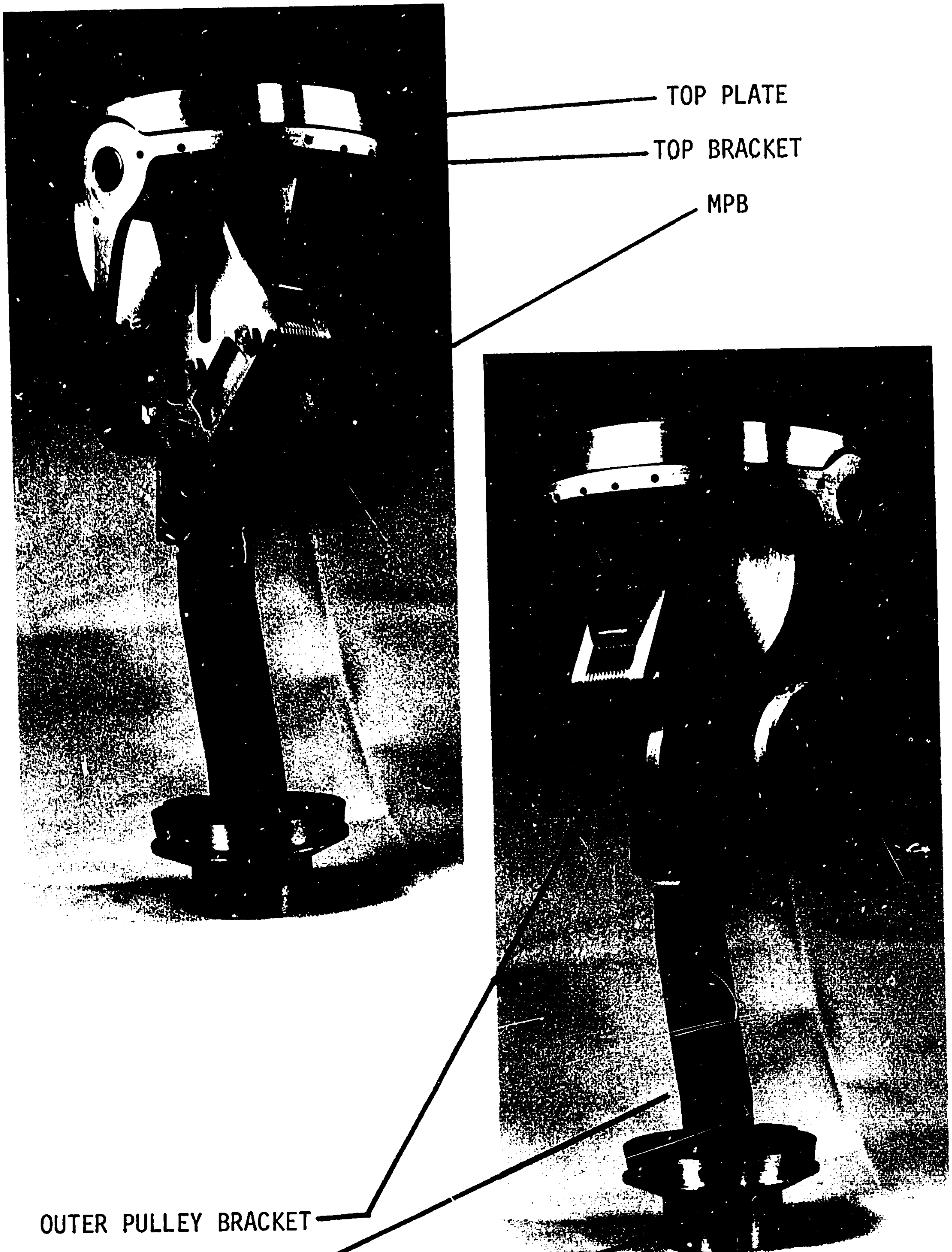


Figure 11 Internal Structure of Prosthesis

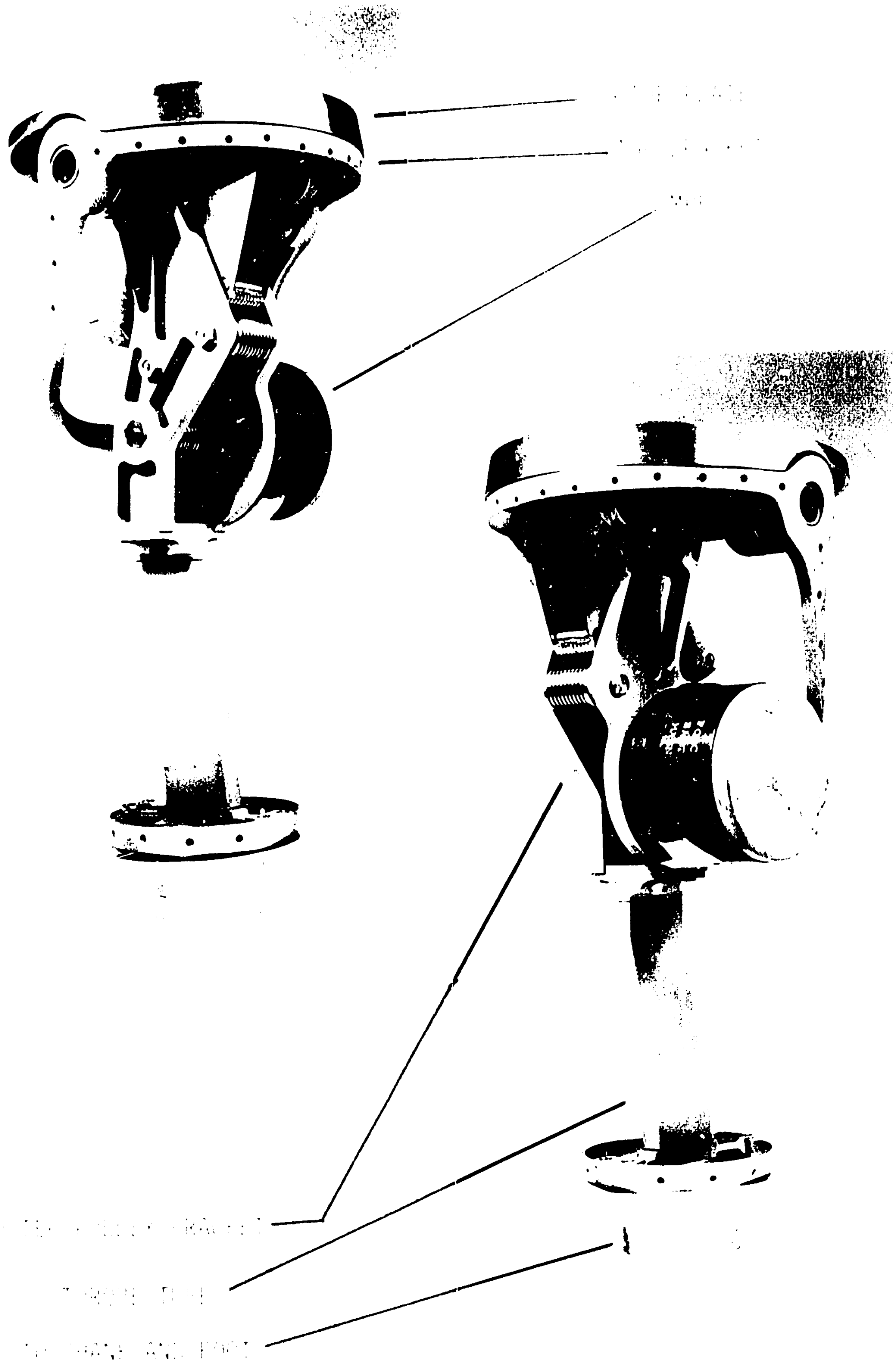
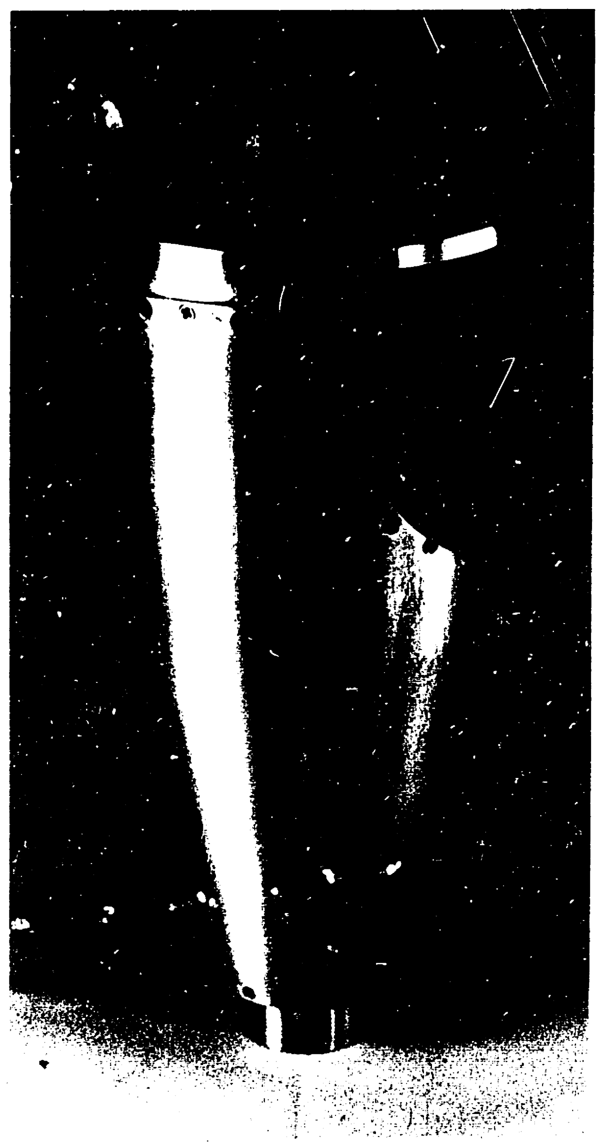


Figure 11 Internal Structure of Prosthesis

(b) Flexion



(a) Extension

Figure 12 Side Views of Knee Joint

Figure 12 Side Views of Knee Joint

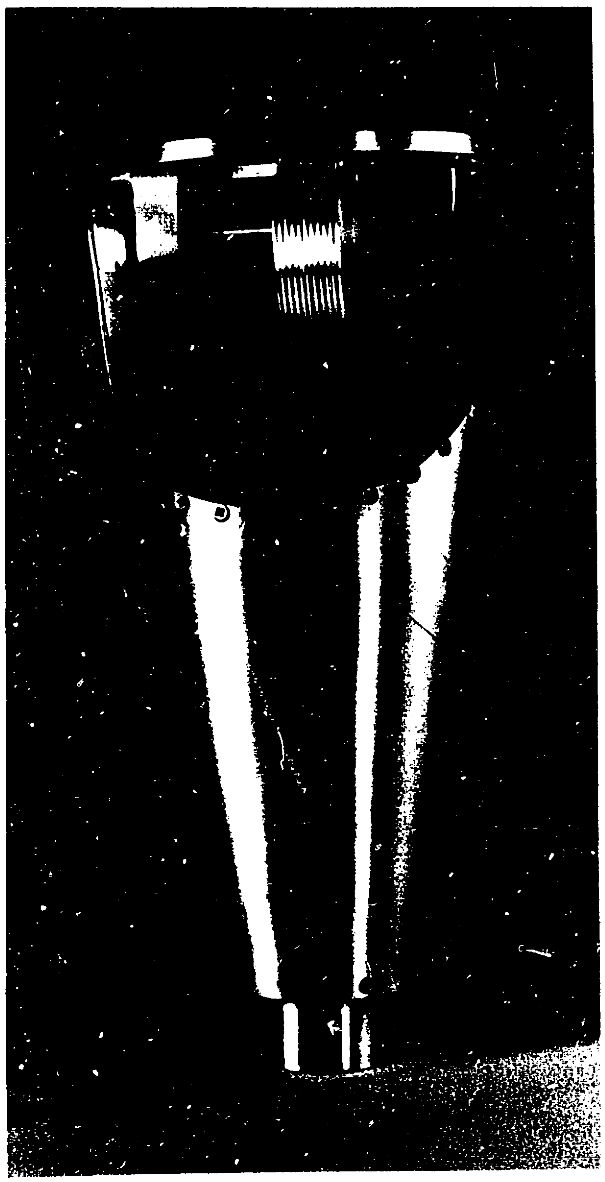


Figure 13 Rear View of Knee Joint

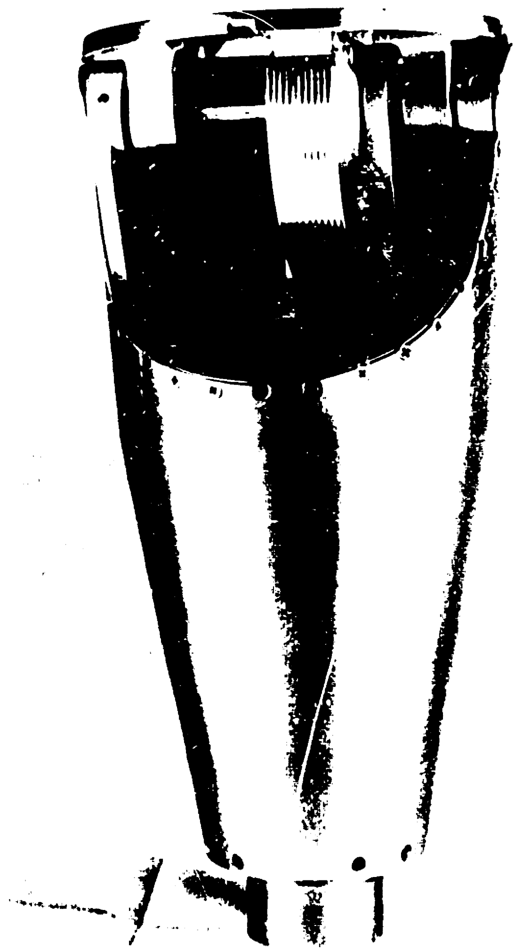


Figure 15. Rear View of Knee Joint

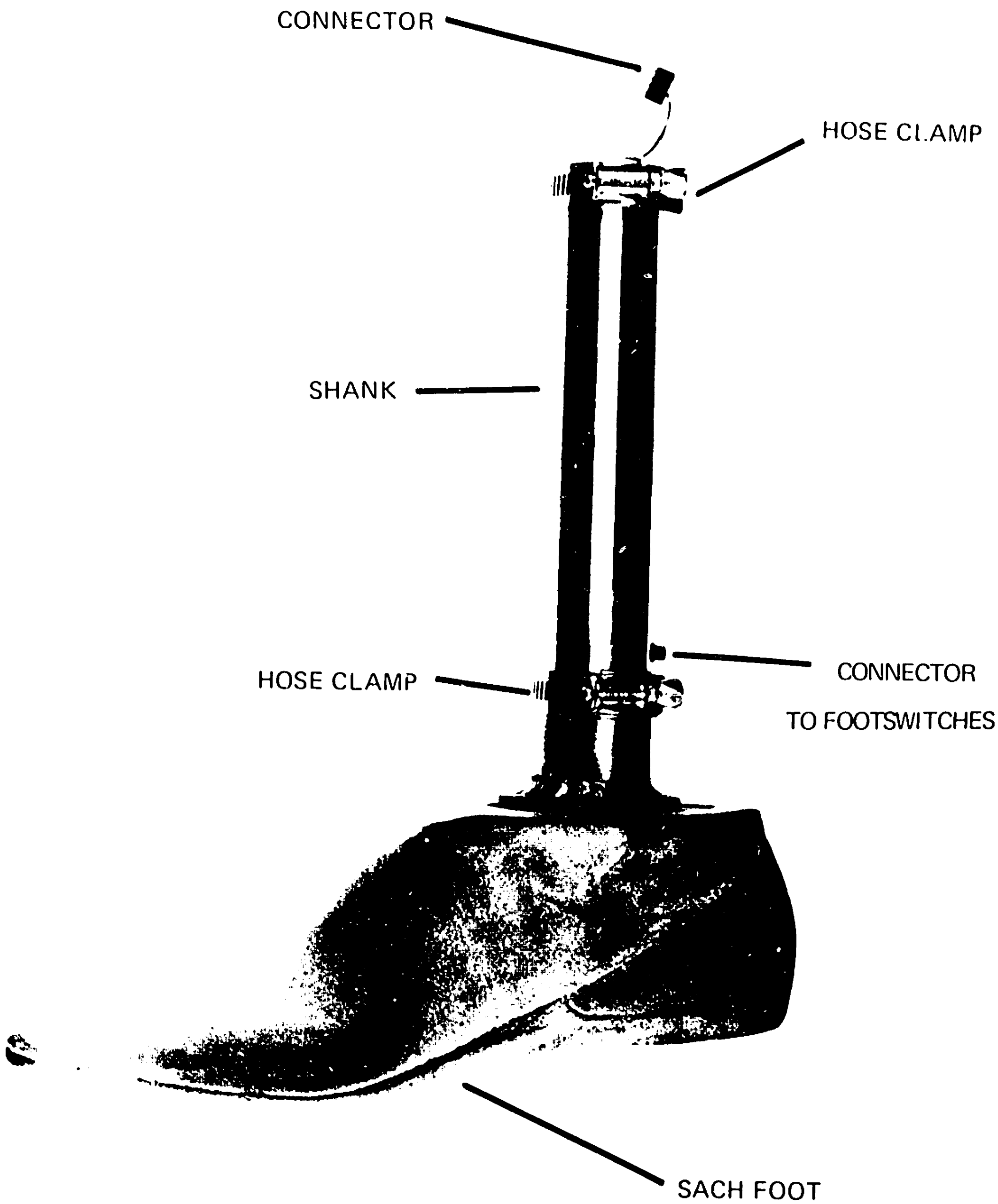


Figure 14 The Shank and Foot
(Lampe (10))

3.2 Detailed Design

This section will explore in greater depth a number of details in the design, with part numbers referred to in brackets [ ]. Figure 15 is an exploded diagram of the knee (excluding the outer shell).

<u>Torque Source</u> The prosthesis uses the model B21SF5 magnetic particle brake manufactured by Force Limited of Santa Monica, California. A magnetic particle brake, as mentioned previously, produces resistive torque as a function of current. This is done through the internal rubbing of radially aligned magnetic particles. Some of the specifications of this particular device are given below:

Torque Range, Total: 0 - 40 in-lb

Linear Torque Range: 0 - 25 in-lb

Response Time: 0 to 25 in-lb in 10 ms

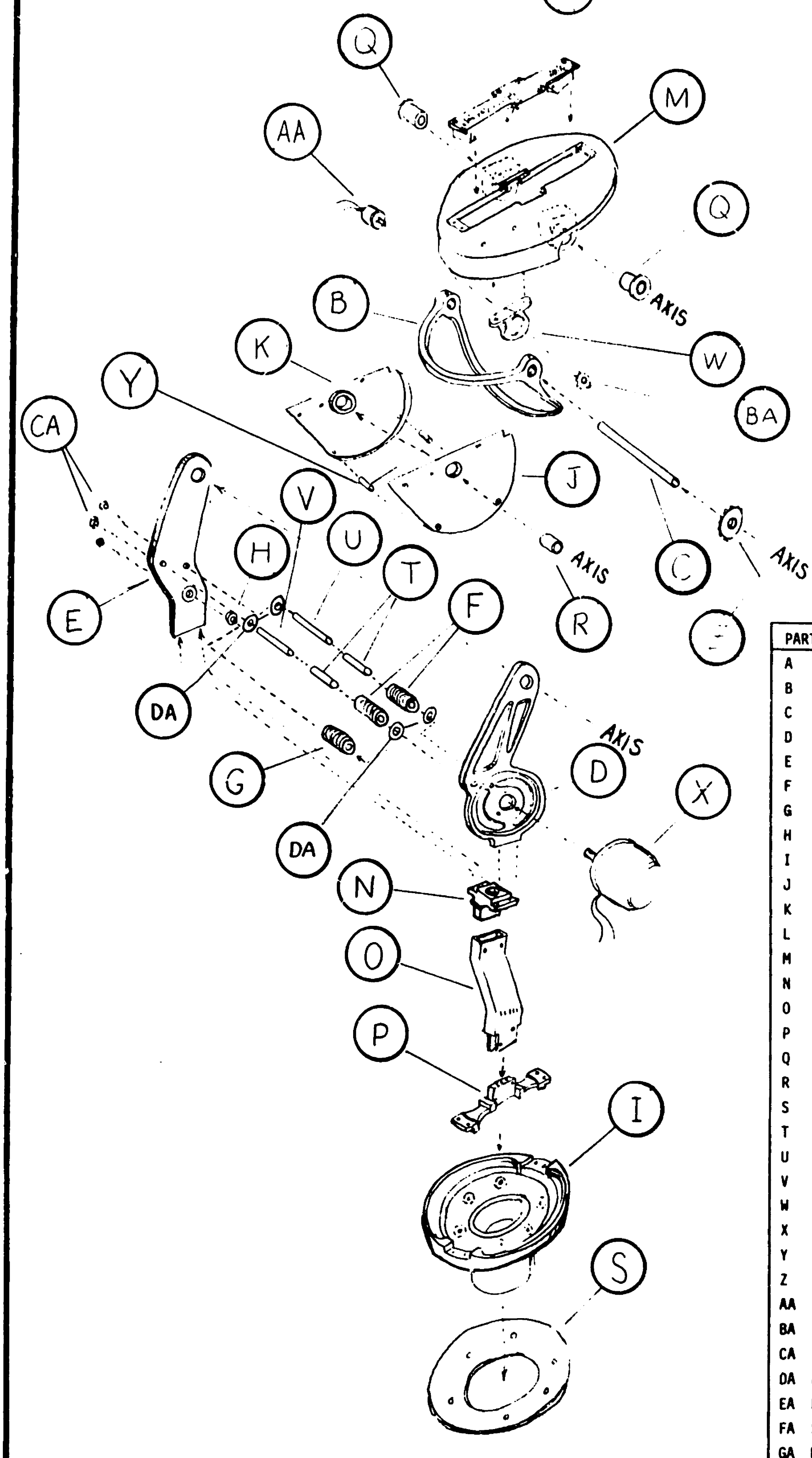
Power Required: 0 to 400 ma @ 24 VDC

Dimensions: 2.1" Dia x 1.75" High- .250 Shaft

Weight: 1.3 lb

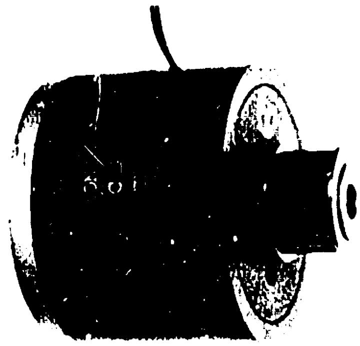
In addition, see figure 16 for a graph of torque vs current, as well as a picture of the brake with a different pulley than the one actually used. The nonlinearity of the device above 25 in-lb can be compensated through a torque feedback system.

<u>Torque Coupling</u> This is essentially a step-up transmission which permits using an MPB capable of generating a torque of



PARTS	
A	OUTER SHELL
B	TOP BRACKET
C	KNEE SHAFT
D	INNER PULLEY BRACKET
E	OUTER PULLEY BRACKET
F	INTERMEDIATE PULLEY (2)
G	MPB PULLEY
H	OUTER MPB SUPPORT
I	BOTTOM BRACKET
J	DRUM HALF "A"
K	DRUM HALF "B"
L	CABLE END PLATE
M	TOP PLATE
N	TOP TUBE HOLDER
O	TORQUE TUBE
P	BOTTOM TUBE HOLDER
Q	KNEE SHAFT BUSHING (2)
R	DRUM BUSHING
S	BOTTOM PLATE
T	BRACKET SPACER (2)
U	LONG LOCATING ROD
V	SHORT LOCATING ROD
W	POTENTIOMETER HOLDER
X	PARTICLE BRAKE
Y	DRUM SPACER
Z	KNEE SHAFT SPROCKET
AA	POTENTIOMETER
BA	POTENTIOMETER SPROCKET
CA	PULLEY BRACKET INSERTS
DA	PULLEY SPACER (4)
EA	HYPEREXTENSION STOP
FA	SHANK EXTENSION
GA	MPB OUTER SUPPORT SCREW

Figure 15 Exploded Diagram



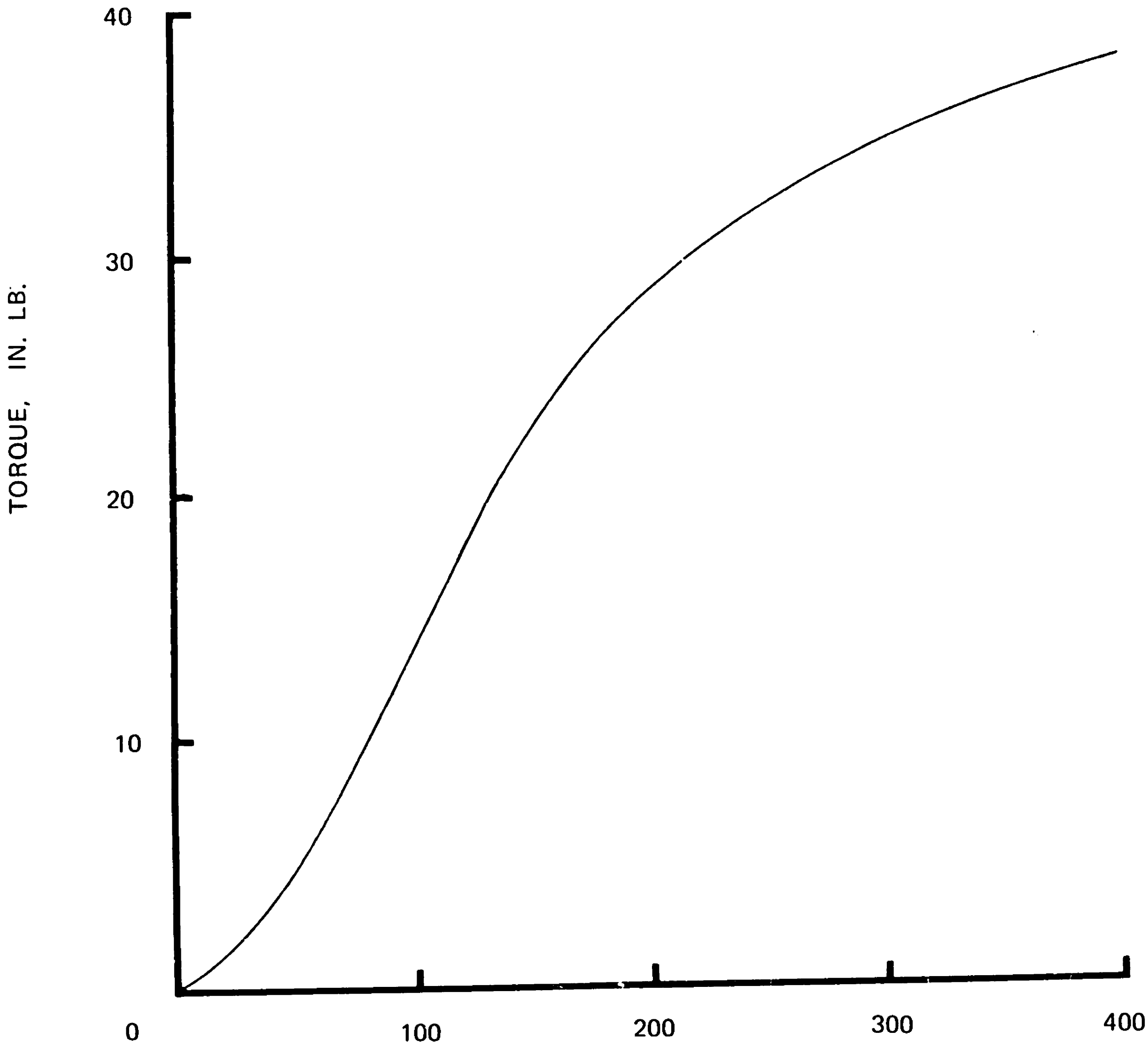
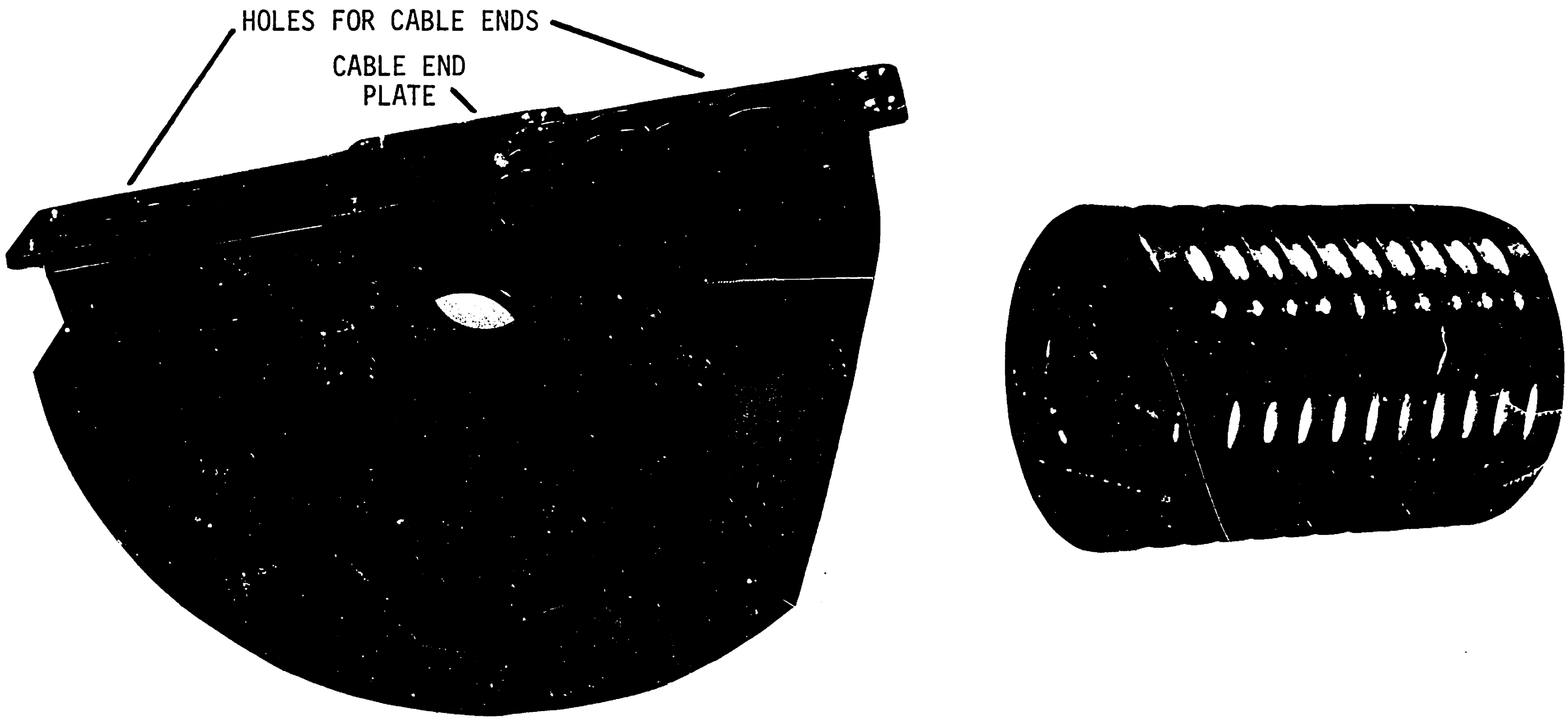


Figure 16 The Magnetic Particle Brake
(Lampe (10))

40 in-lb to generate a torque of 300 in-lb at the knee axis. (Resistive torques) The upper portion of figure 10 is a diagram of the system. Ten highly flexible (343 strand, .052 diameter) nylon coated cables are attached at the cable end plate [L], come out and around the drum [J,K] via intermediate pullies [F] and the particle brake pulley [G]. The ten cables are used for infinite fatigue design life and for a high degree of redundancy. This is a constant path length system with no backlash. Figure 17 has photographs of the drum and a typical pulley. The drum was assembled from two interlocking halves, followed by machining of the grooves with the drum running on a mandrel for maximum concentricity. All grooves were cut to precisely the same depth with a tungsten carbide form tool. The extra care taken here was essential to the success of the project. The intermediate pullies run on stainless steel spacer rods [T] positioned by locater rods [U,V] on the pulley brackets. The MPB pulley is supported at its outer end by part H, which is shown in a close-up photo in figure 22.

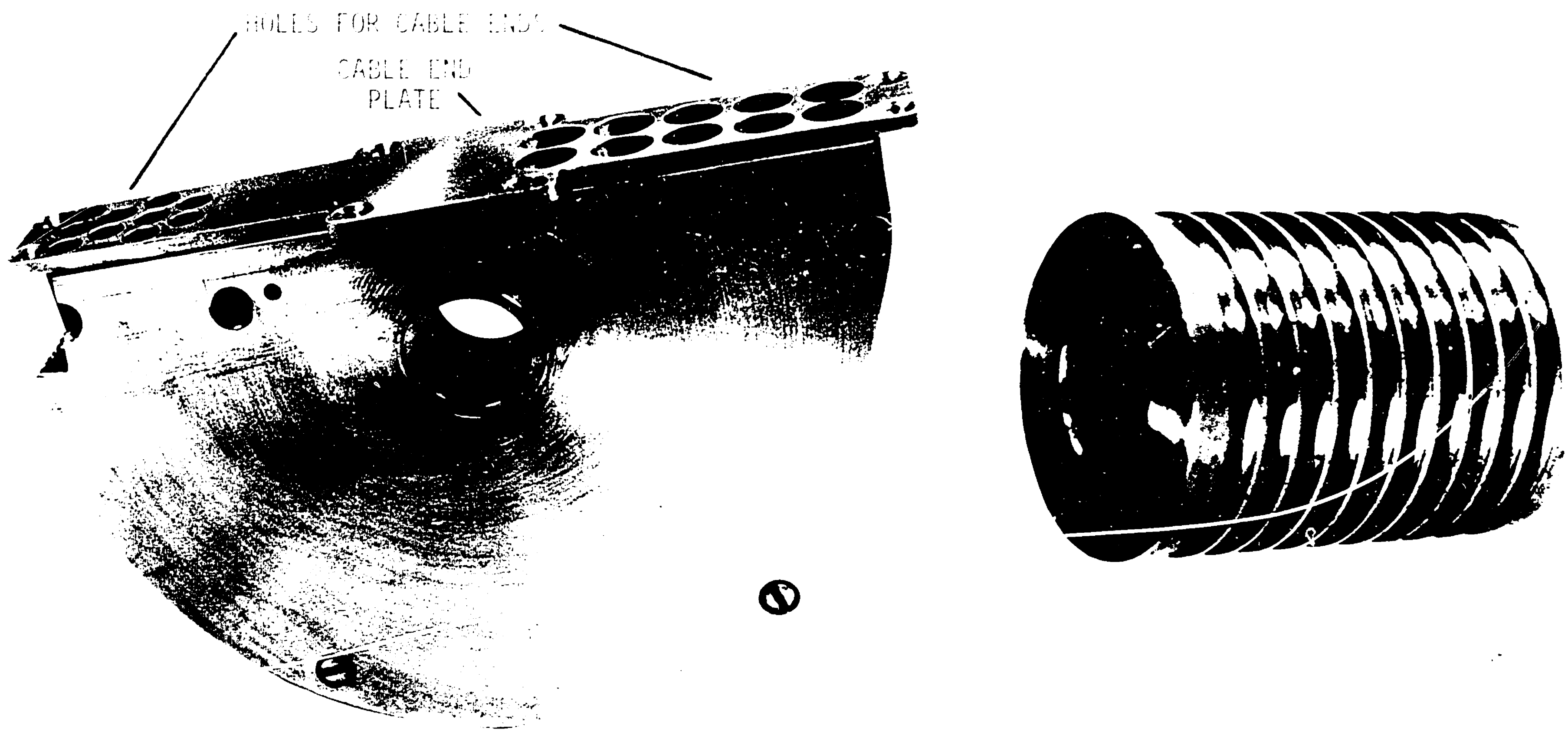
Torque Isolation All of the knee axis torque is borne by a central structure which was shown in figure 11 on page 28. The structure requires extreme stiffness for resisting the front-back moments occurring within the prosthesis during walking, yet needs to be relatively flexible to vertical loads. This dual requirement accounts for the design of tne



(a) - The Drum

(b) - Enlarged Pulley

Figure 17 Torque Transmitting Components



(a) - The Drum (b) - Enlarged Pulley
Figure [illegible] Torque Transmitting Components

bottom tube holder [P]. This was thought to be necessary to prevent the central column from acting like a spring, resisting load during stance phase, and upsetting the proper "feel" of the prosthetis. The torque tube [O] was fabricated from a 1.00 x .300 cross-section extruded aluminum box tube which was ideal from the standpoint of rigidity. Figure 18 shows the two pulley brackets which go between the knee axis and the top tube holder [N]. They are extremely rigid, yet very light due to their ribbed construction. They are also designed to rest against the hyperextension stop and the flexion stop, allowing torque measurement during stance phase and at full flexion. The hyperextension and flexion stops have not been built yet, but one design called for small 2024 rectangular blocks with 4-40 heli-coil inserts (for adjusting screws) to be welded onto the drum.

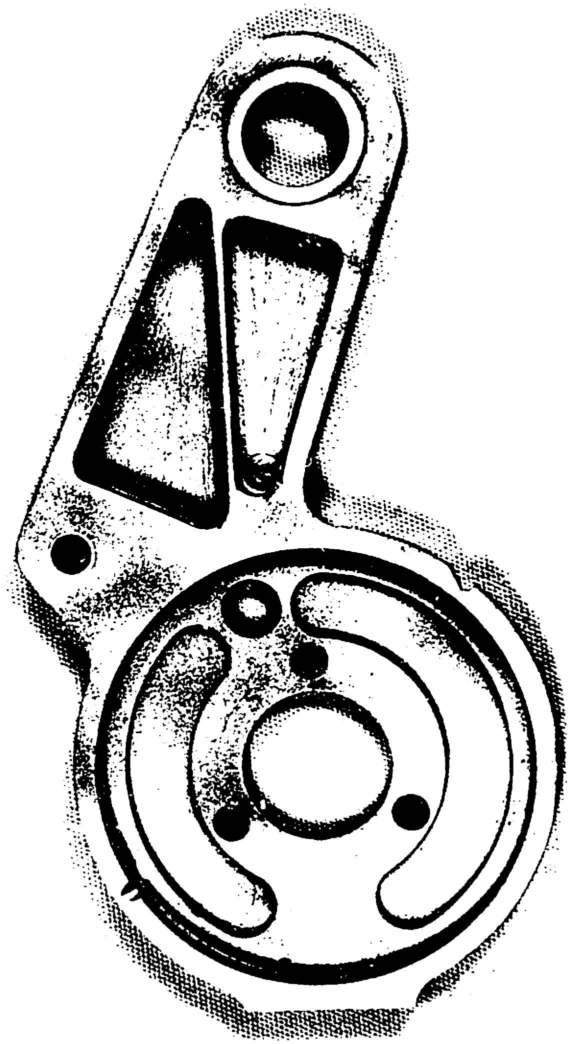
Because the knee shaft is fixed relative to the top bracket [B], there should be no torque measurable there, hopefully eliminating a hysteresis problem which occasionally plagued the MPB prosthesis simulator. Knee shaft torque is to be measured by means of strain gages (not yet installed) on the torque tube, which will see tension and compression, and a slight amount of torque due to the slight offset ot the piece.

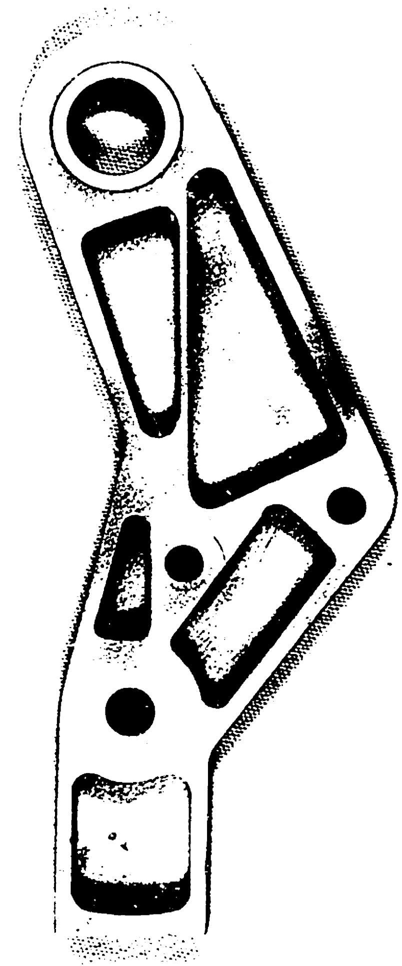
Load Bearing Figure 19 is a photograph of the top plate [M] and top bracket [B]. The suction socket will bolt onto the

(a) - Inner Pulley Bracket

(b) - Outer Pulley Bracket

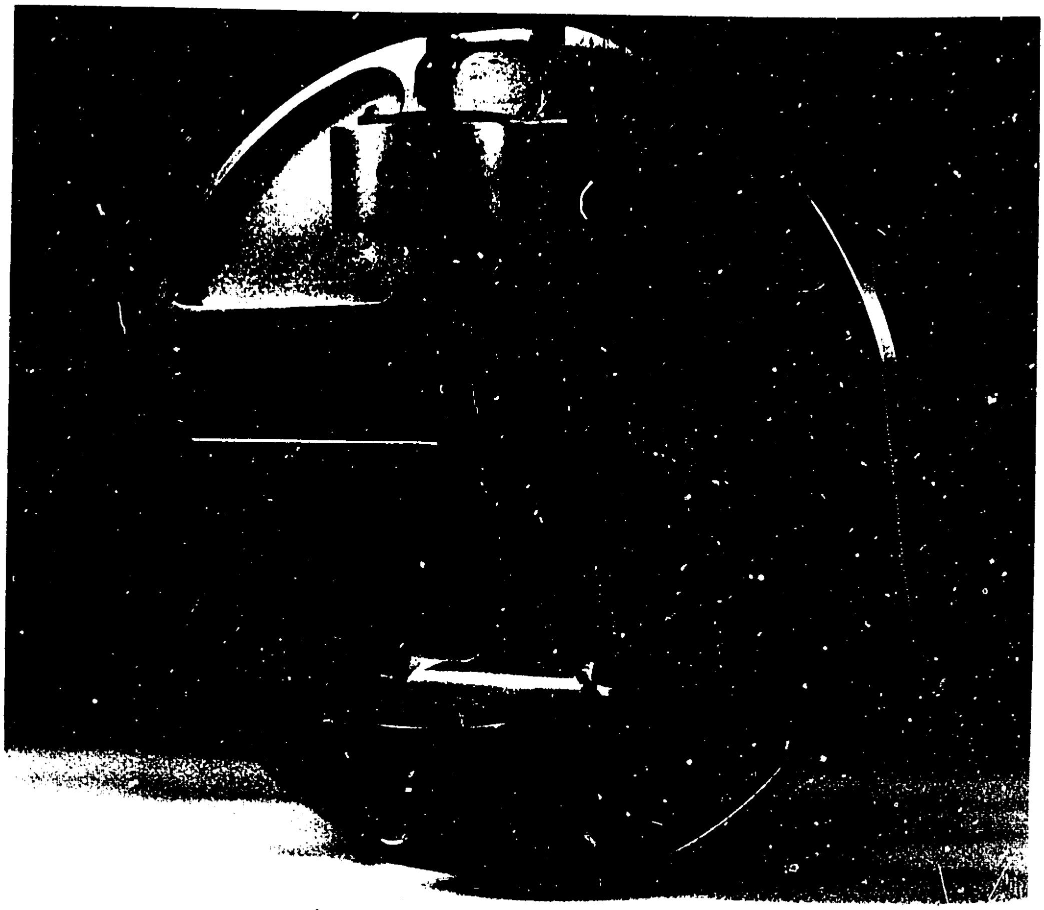
Figure 18 Inner and Outer Pulley Frames





(a) - Inner Pulley Bracket (b) - Outer Pulley Bracket

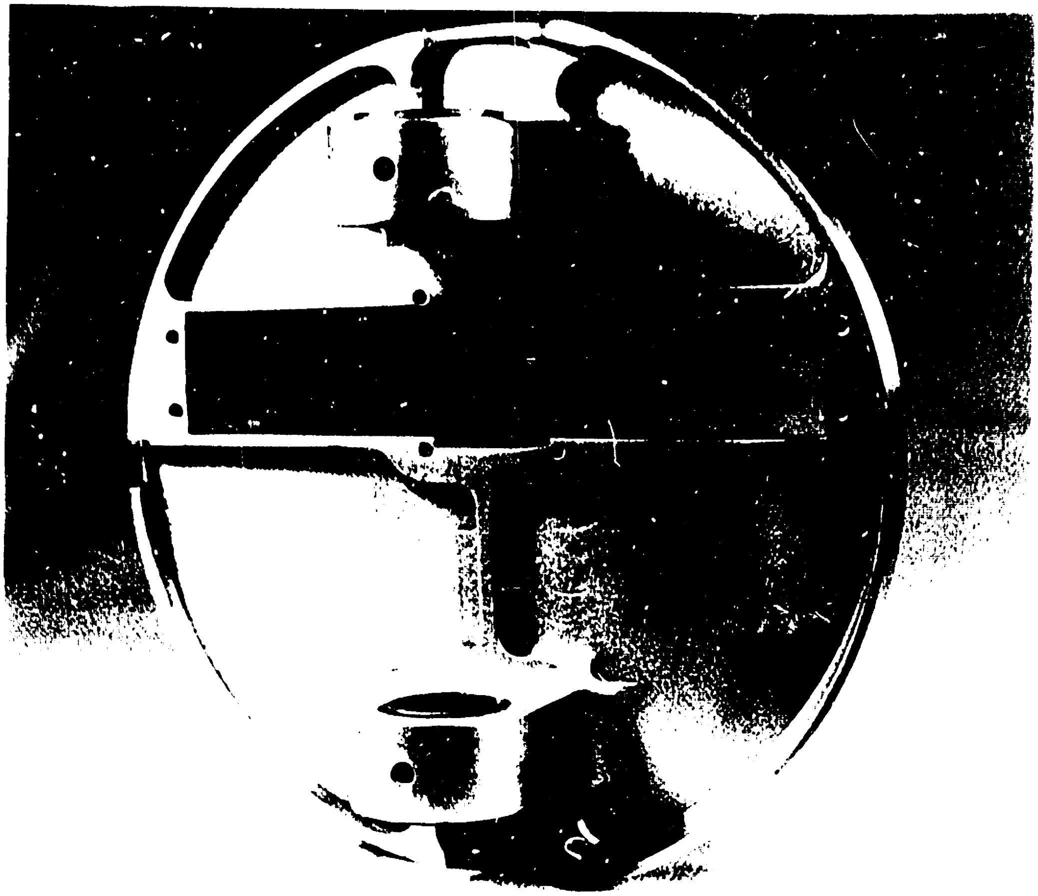
Figure 1: Inner and Outer Pulley Frames



(a) - Top Plate

(b) - Top Bracket

Figure 19-Load Transmitting Members



(a) - Top Plate

(b) - Top Bracket

Figure 19 - Load Transmitting Members

top plate, and vertical load will be transferred through the knee shaft to the top bracket. This is attached inside the top of the outer shell. The bottom bracket [I] bolts to the shell, and the bottom plate [S] attaches underneath the bracket to bear the weight on the shell. The shank tube is then attached with a hose clamp. Except at the top bracket (where there are 30 2-56 screws) load is carried by the parts themselves, and not by bolts in shear.

Control Variables Shank angular position is obtained from a single turn potentiometer mounted under the top plate. Figure 21 is a picture of the potenticmeter in its holder. A small sprocket on the shaft is to be connected by a fine pitch chain to a larger sprocket fixed to the knee shaft. The number of sprocket teeth is to be arranged so that the full rotation of the shank from hyperextension to full flexion is slightly less than one turn of the pot. Angular velocity will be obtained by differentiating the velocity signal, using a circuit given by Lampe (10). This is possible because of the clean signal provided by the high quality (NEI 78ESC502) potentiometer. Torque will be measured as described earlier. Timing of gait cycle events can be obtained through footswitches mounted on SACH feet already in use at M.I.T.

The Shell The shell is a .045 thick spinning of 3003-TO aluminum. This was made by machining a rock maple mandrel

Figure 20 Spun Aluminum Outer Shell

Figure 20 Spun Aluminum Outer Shell

to the shape desired for the shell and having a tapered aluminum cylinder spun over it. Original specifications called for .020 - .025 6061-T6 aluminum, but time constraints and supplier error resulted in the present shell. The mounting holes were drilled through into the brackets while the device was assembled.

A great deal of room is provided inside the knee unit for potential controller and battery needs. It is possible that clips may be used around the torque tube or on the shell for carrying four 6 volt nicad stacks.

NEI 78ESC502

Figure 21 Potentiometer and Holder Close-Up

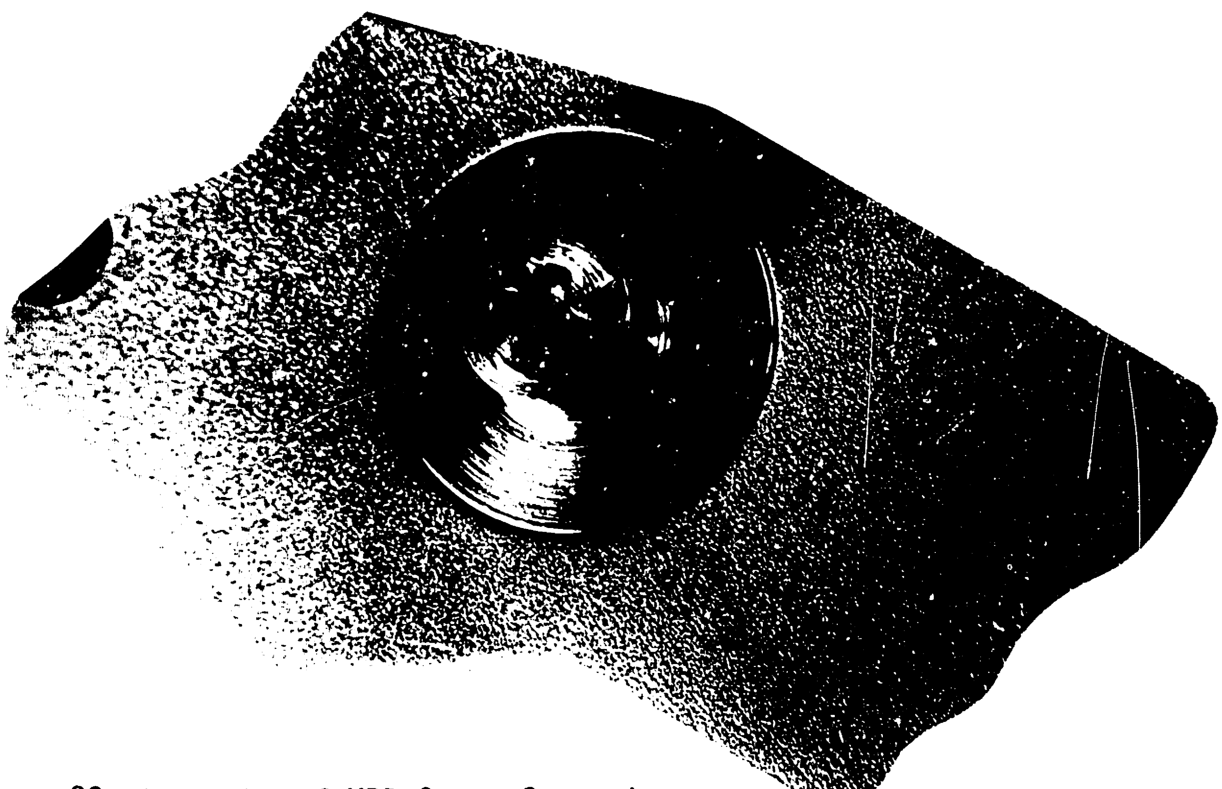


Figure 22 Close-Up of MPB Outer Support

Figure [illegible] Potentiometer and Holder [illegible]

Figure [illegible] Close-Up of MFE Outer Support

4. CONCLUSIONS AND RECOMMENDATIONS

4.1 Progress to Date

At the time of writing, the prosthesis was still incomplete. Minor details remained such as inserting presserts, putting setscrews into the MPB pulley, installing the potentiometer holder, sprockets, and chain, etc. Of a somewhat less-than-minor nature is the fact that none of the aluminum parts have been anodized, and that neither the hyperextension stop nor the flexion stop had been built. In addition, strain gages need to be applied on the torque tube, and allied circuitry installed. The prosthesis swings very freely when the cables are not installed, but it is as yet somewhat stiff when they are. It is hoped that the even tightening of all ten cables (something that hasn't been tried carefully yet), replacement of two badly damaged cables, and installation of the pulley spacers (part number DA) will solve that problem. A 1 1/8" spacer must be made to interface the top plate with the suction sockets used in the laboratory, since the new prosthesis can accomodate that much longer stumps than the simulator systems.

4.2 Conclusions

The prosthesis which has been described will be used to complement the present Magnetic Particle Brake Prosthesis Simulator, and to test control schemes that result from research there and elsewhere. Because it will be totally self-contained, it can also be used outside the laboratory. This will permit the testing of concepts under real-world conditions. Also, since the prosthesis will thus be worn for longer periods of time than the simulator, a more accurate portrayal of various effects can be obtained. Of course, this can't happen until extensive testing within the laboratory has proven the safety of the device.

It is anticipated that initially the prosthesis will be controlled during swing phase by a simple scheme in which damping is proportional to velocity, and will hyperextend and lock during stance phase, unless a passive stance phase control cheme can be developed. Recent developmemts in microprocessor technology have made a single chip controller a distinct possibility, so work is continuing on a controller that will use torque, position and velocity information available within the prosthesis to permit more sophisticated torque control. This will be programmed with data such as that obtained by Radcliffe (14), which is shown in figures 23 and 24 . Lampe (10) has derived cadence responsive torque profiles from these curves, using the MPB prosthesis simulator facility. These are shown as a surface of torque profiles in

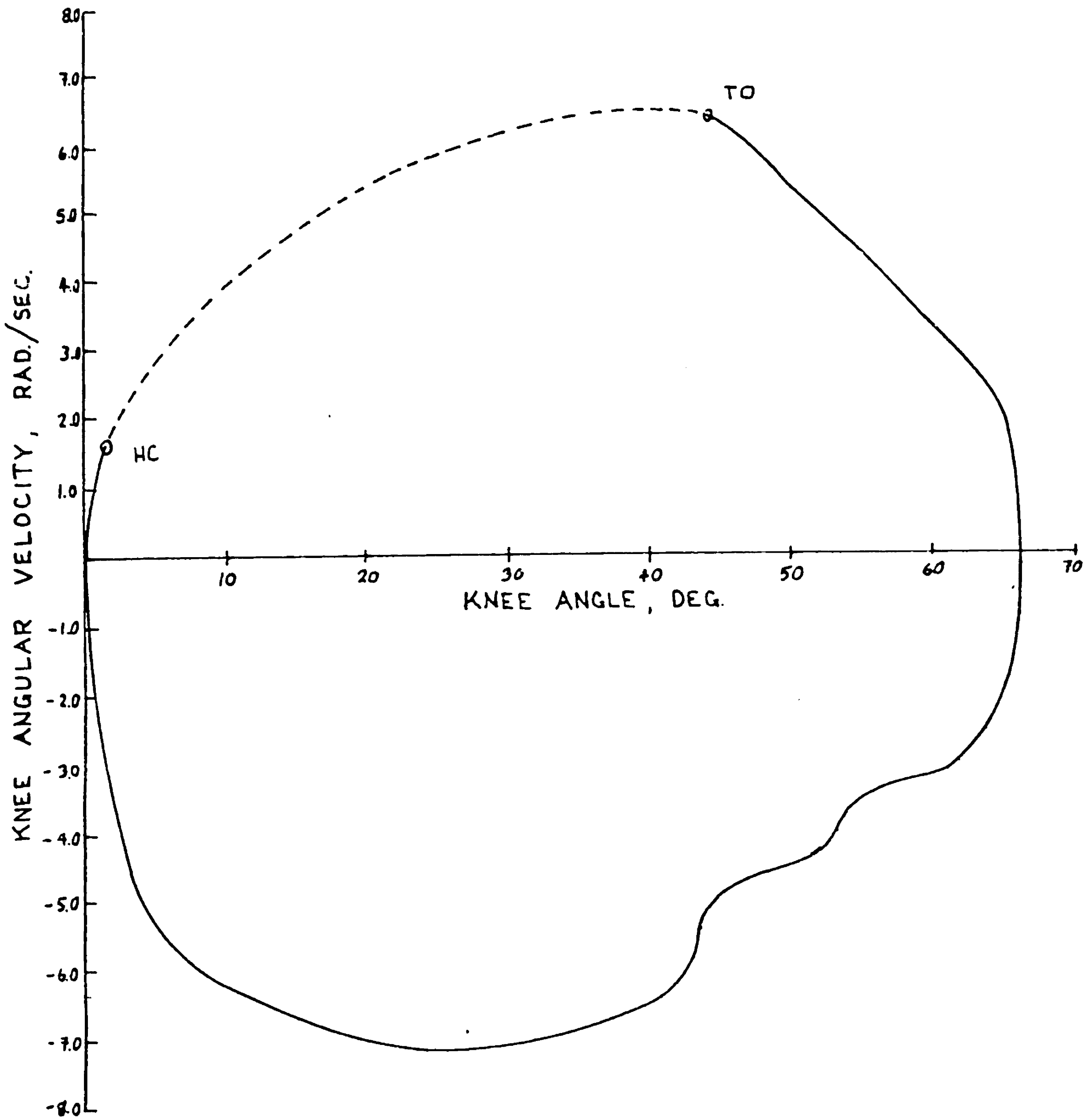


Figure 23 Ideal Velocity Curve (Lampe (10), from Radcliffe (14))

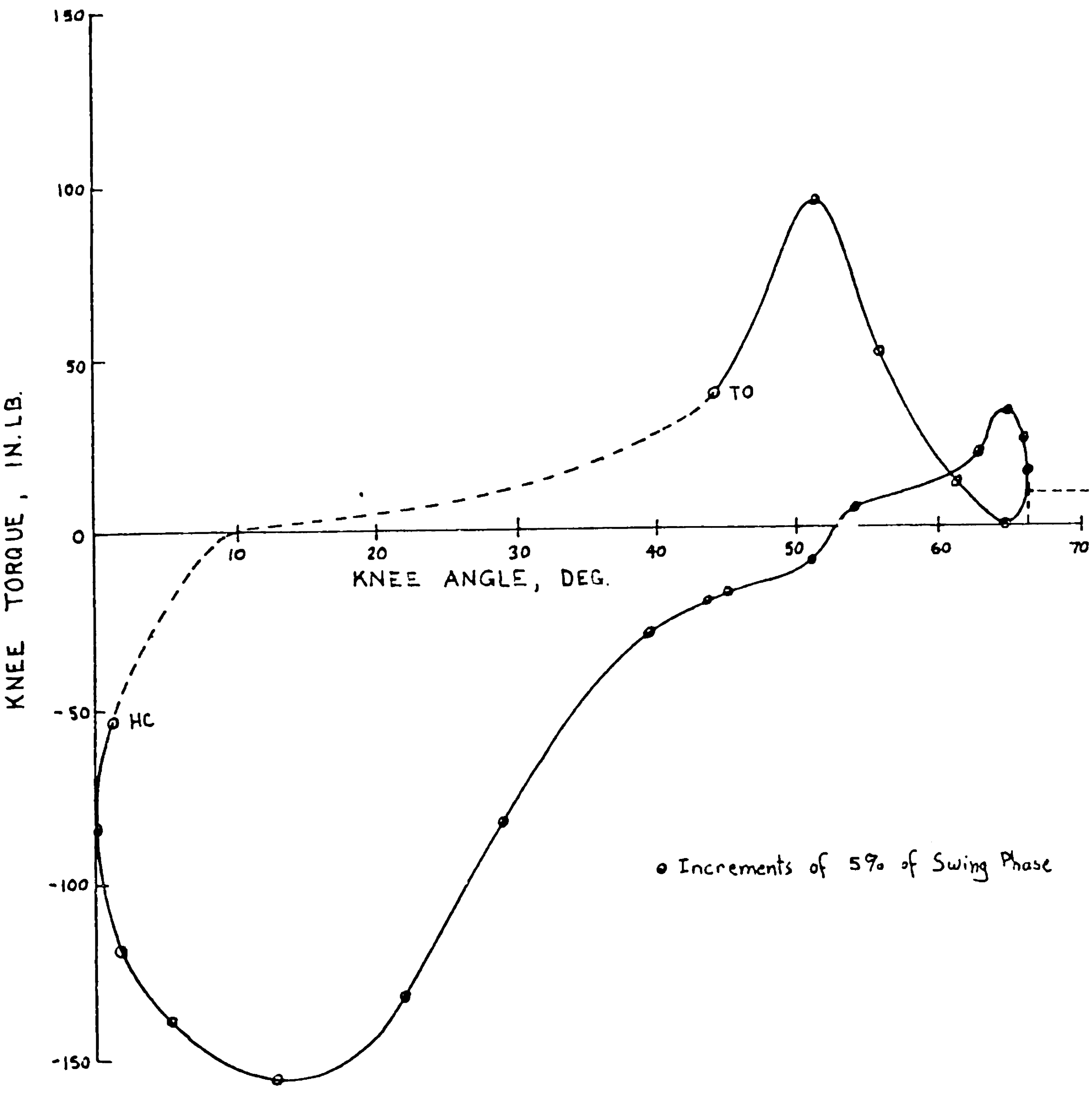


Figure 24 Ideal Torque Curve (Lampe (10), from Radcliffe (14))

in figure 25. Depending on the versatility and capacity of the on-board controller, it may be possible to program even very complicated knee torque schemes such as these.

4.3 Recommendations

Some of the requirements for future work should be apparent after reading previous sections. First the prosthesis must be completed and debugged as originally designed, amd then modified if necessary, as indicated by initial tests. The actual parameters of the prosthesis should be determined, such as center of gravity, equivalent damping in a free swinging mode (essentially the "stiffness" of the knee), and the various frequency responses of knee output torque from inputs to the particle brake. Given that a control interface is carefully designed, the leg should be capable of many of the functions now performed by the present MPB prosthesis simulator. When an on-board controller has been built, the knee will hopefully widen the horizons of the Knee Research Group by taking another step toward real-world applications.

In some ways this project can be viewed as a first-pass attempt at making a mass producible, highly controllable, saleable A/K prosthesis. For that sort of application it would be convenient to have a sliding, spherical shield at the joint that would help prevent clothing from getting snarled when changing from a seated to a standing position, for example. Also a shaped foam cover would serve the dual

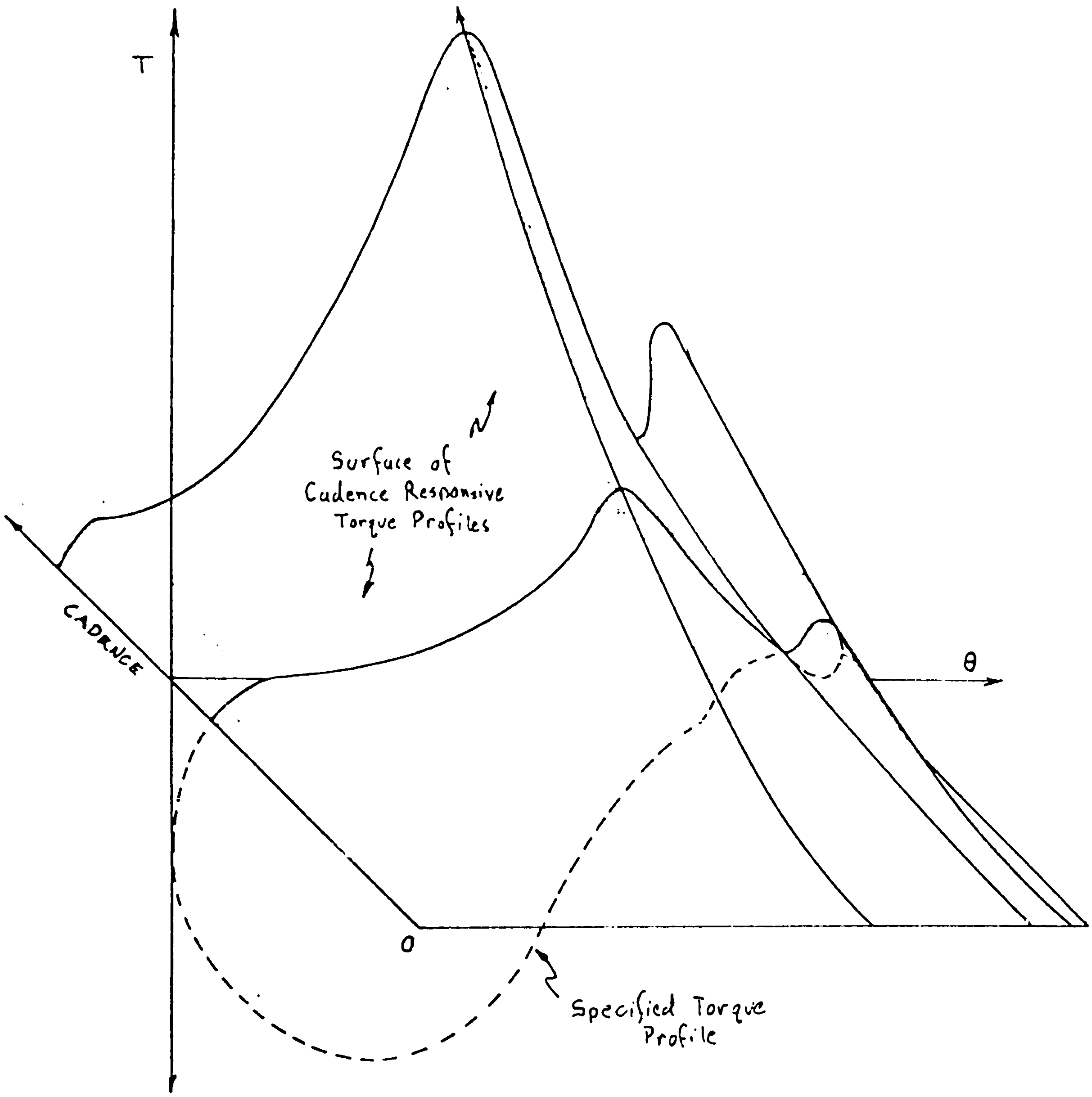


Figure 25 Surface of Dynamics Simulated at MPB Prosthesis Simulator (Lampe (10))

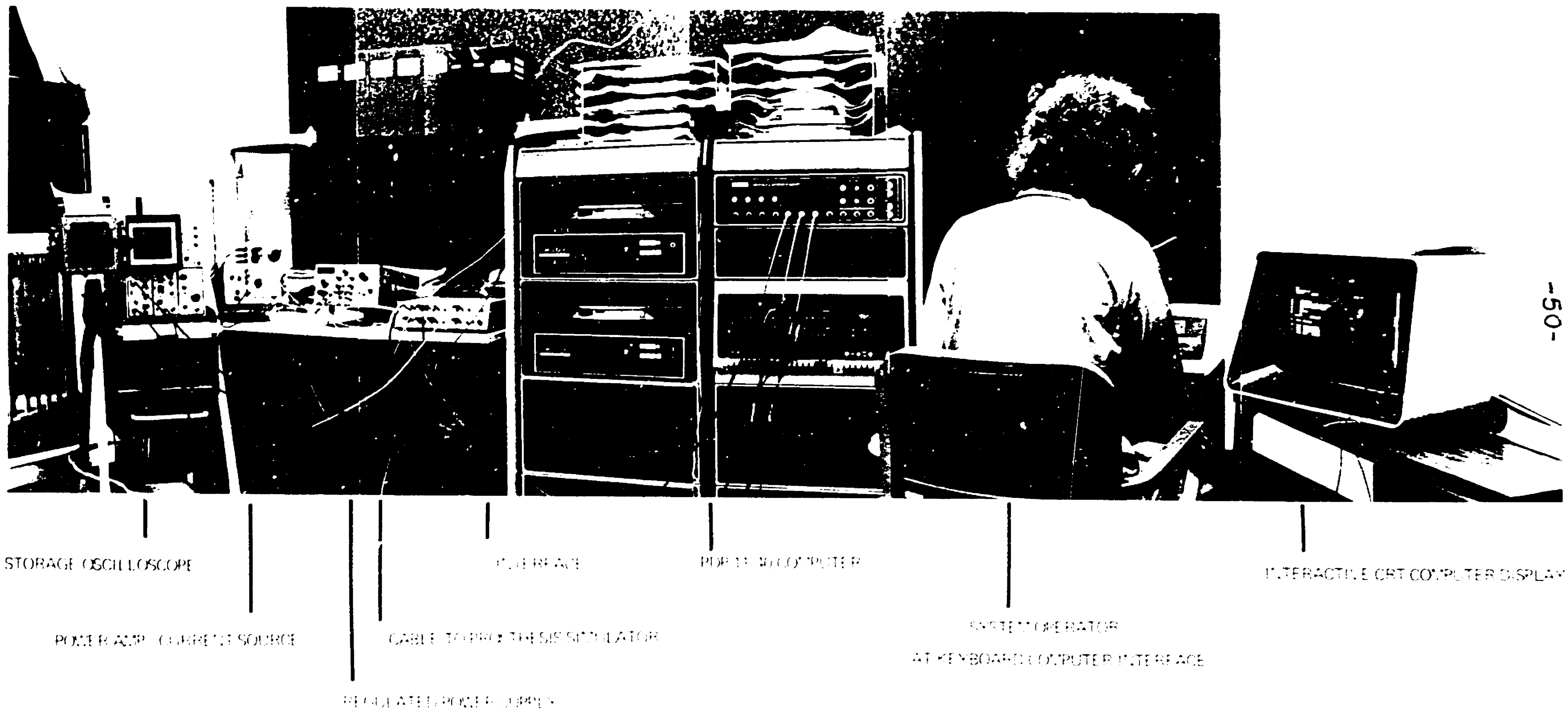


Figure 26 Magnetic Particle Brake Prosthesis Control Facility (Lampe (10))

role of dent protection and improved cosmesis. While many of the parts appear highly complex in this prototype form, it should be remembered that in quantity there are many easier methods of fabricating parts than machining alone. Designs change when production quantities are considered also.

Should a decision be made at some point in the future to actually produce a magnetic particle brake damped prosthesis, it should not be thought that a cable drive system is the only viable solution. An extremely accurate bevel or spiral gear pair, for example, could potentially have small enough backlash to be acceptable, yet still give a high enough gear ratio. A particularly intriging concept proposed recently by Professor Flowers involves using a Saginaw Recirculating Ball Screw to translate linear motion of a pivot point into rotational motion of an MPB. It is feasible to get a 40:1 gear-up at 95% efficiency. This would permit using a much smaller brake than was used in this thesis, saving weight and power. The most exciting feature of this concept is that it could be designed so that it was dimensionally similar to present swing phase control dampers, like that shown in figure 3, so that it could be installed in existing prostheses.

Reports from the many research projects on A/K amputee mobility have been promising, so it appears that it may just be a matter of time before nearly ideal prostheses restore full functionality.

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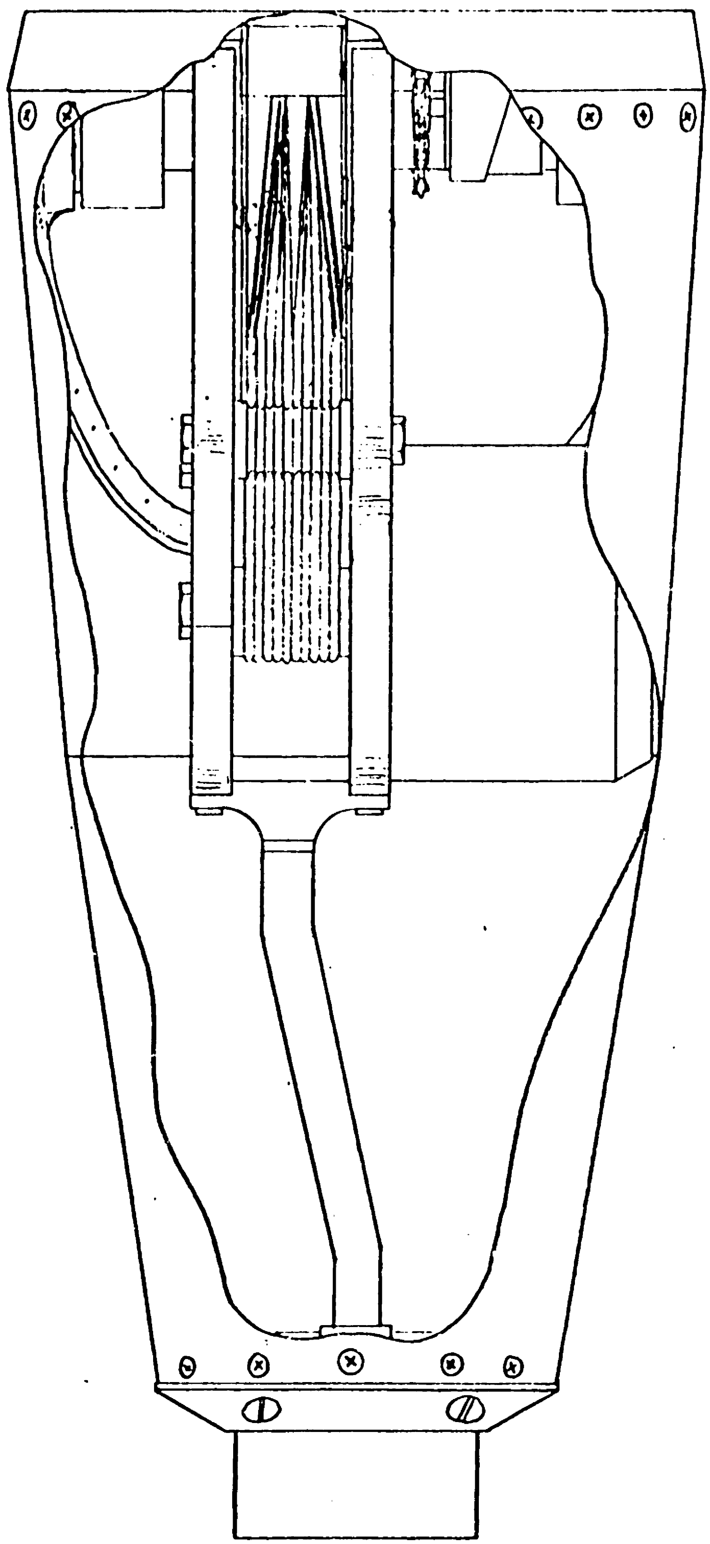
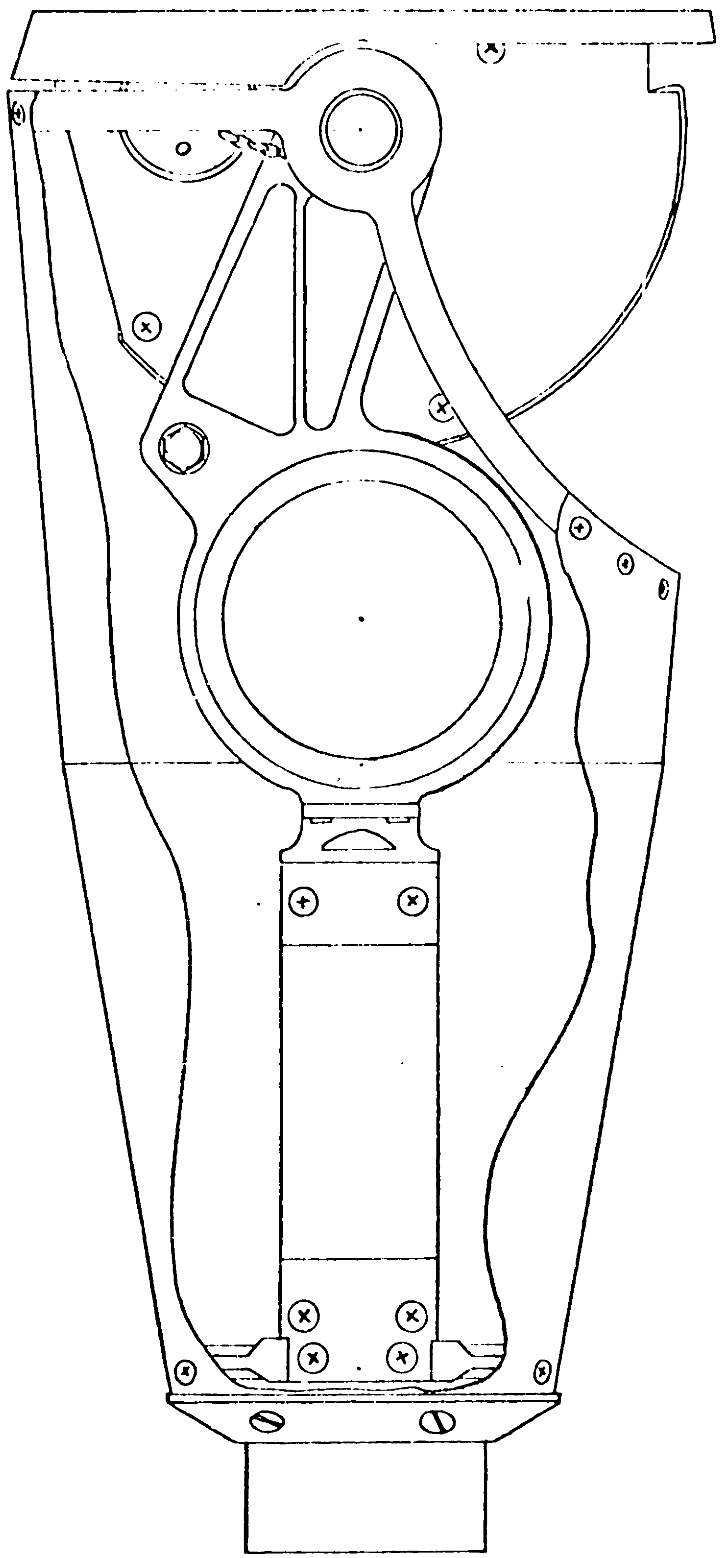
5. APPENDIX

Appendix A -- Assembly Drawings



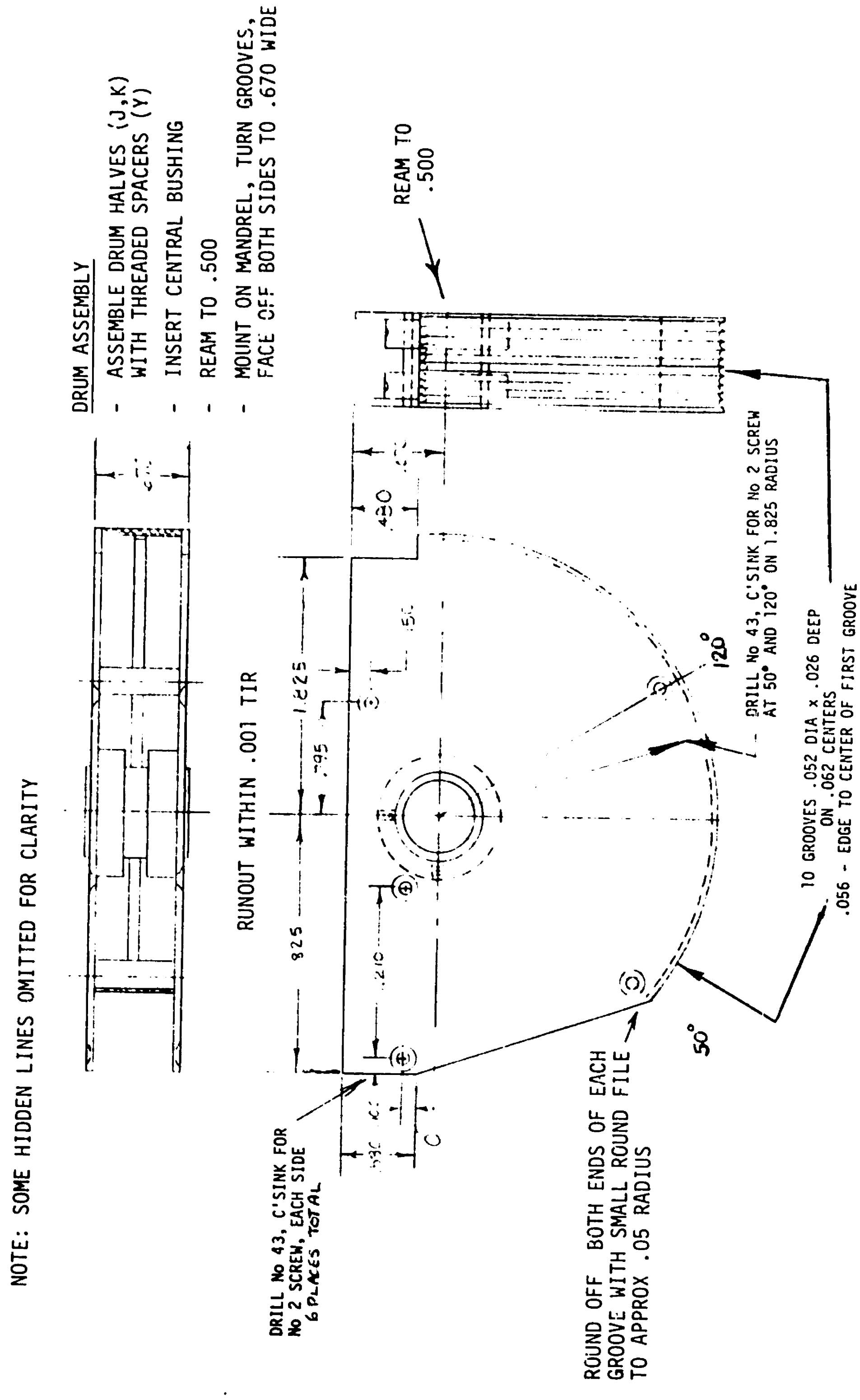
FRONT VIEW

MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
CUTAWAY FRONT VIEW		ASSEMBLY

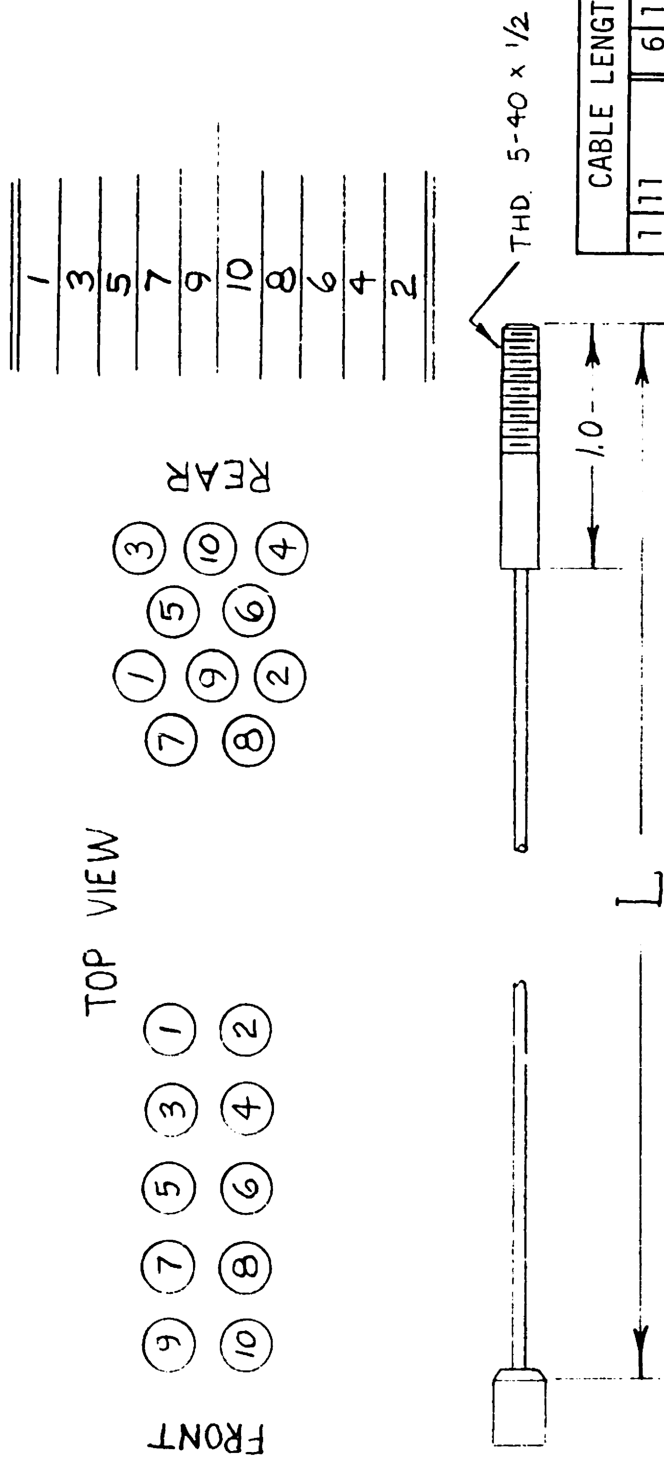


SIDE VIEW

MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
CUTAWAY SIDE VIEW		ASSEMBLY



MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
DRUM ASSEMBLY		



CABLE LENGTHS, L			
1	11	6	10 3/4
2	11 1/8	7	11 5/16
3	10 1/2	8	11 5/16
4	10 1/2	9	11
5	10 3/4	10	10 1/2

LENGTHS IN INCHES

MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
CABLE LAYOUT		10 CABLES

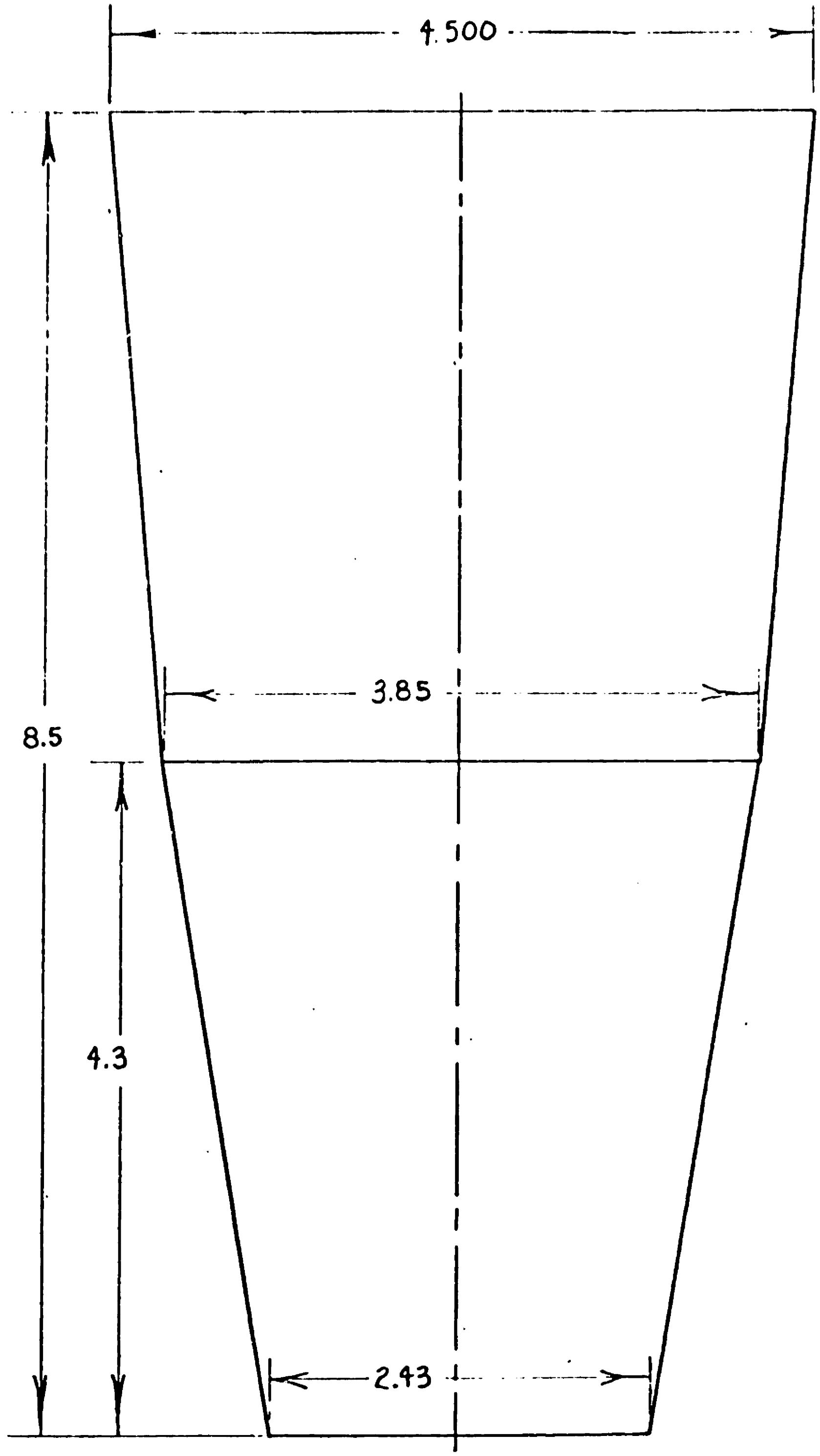
APPENDIX B - HARDWARE

Quantity	Item
76	2-56 x 3/16 F.H. Phillips Screws
4	2-56 x .150 F.H. Phillips Screws
4	2-56 x 1/4 Pan Head Phillips Screws
2	4-40 Nuts
4	4-40 x 1/8 Setscrews
4	4-40 x 1/4 Setscrews
2	4-40 x 3/32 Setscrews
20	5-40 x 3/16 (across flats) Nuts
2	6-32 x 3/16 Setscrews
1	6-32 x 5/8 F.H. Phillips Screw
1	6-32 x 1 3/16 F.H. Phillips Screw
1	6-32 x 1 5/16 Hex Head Screw
3	8-32 x 3/8 F.H. Phillips Screws
10	2-56 Presserts 32N(.040)FL
6	4-40 Presserts 440NKS(.056)
2	4-40 x 5/16 Press-in Studs 34S(.062)-ST-6
3	6-32 Presserts 46N(.040)-EXH-4
	Pressert Part Numbers are from Precision Metal Products, Stoneham, Ma.
10'	SN 2047 (7 x 49) Cable
10	5-40 Threaded Crimp on Cable Ends
10	3/16 Crimp on Ball Ends for Cable
	Cable Supplies from SAVA Industries, New Jersey

Note: All Hardware (Except Cable End Fittings) is to be Stainless Steel

Appendix C -- Part Drawings

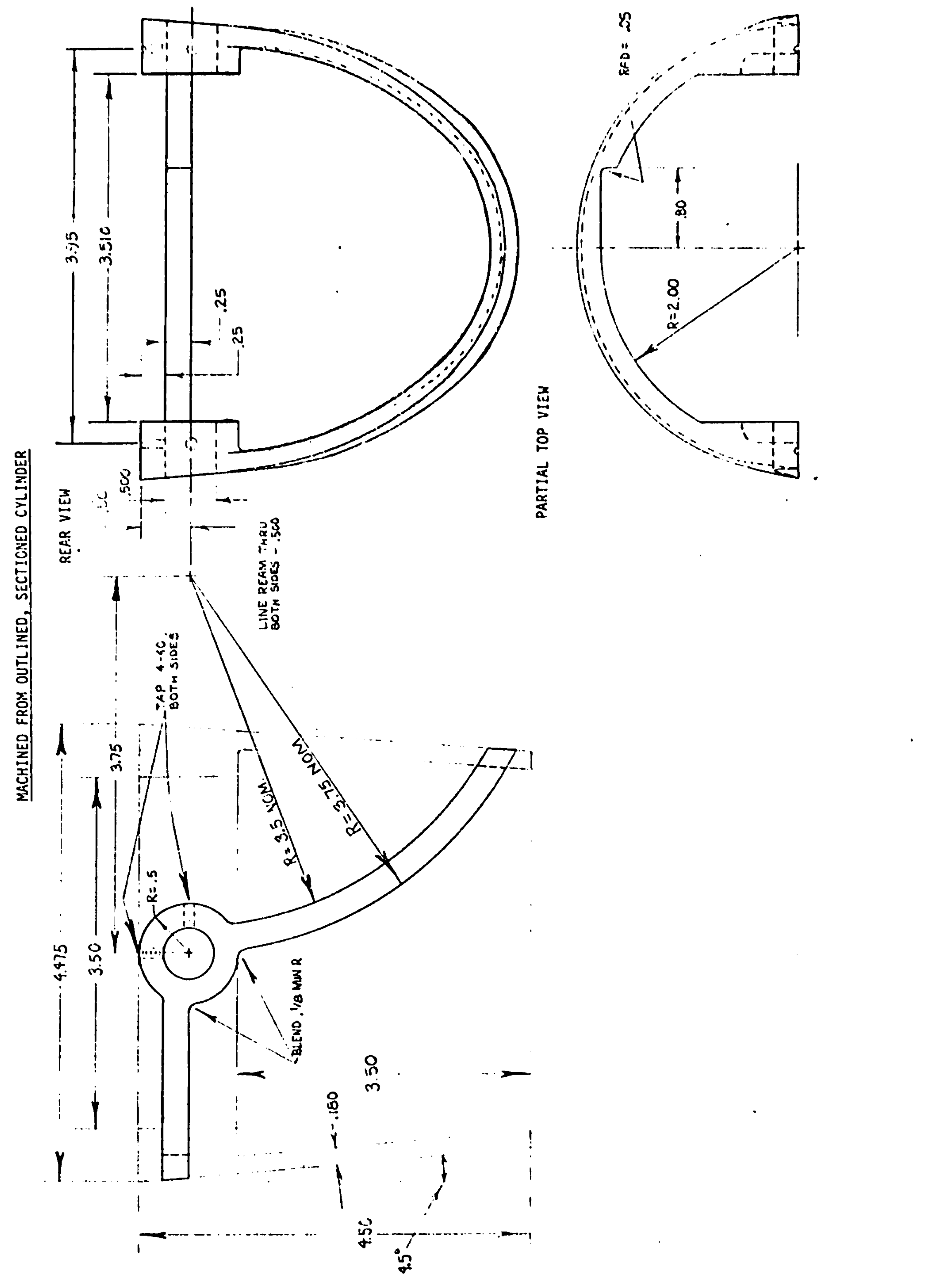
SPINNING FORM FOR OUTER SHELL



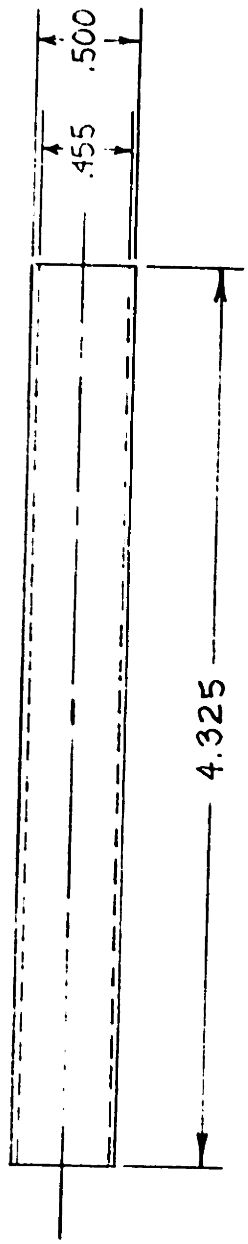
NOTE: DIMENSIONS GIVEN ARE INNER DIMENSIONS OF SHELL

MAT'L- .025 3003-T2 ALUMINUM, FOR SHELL.

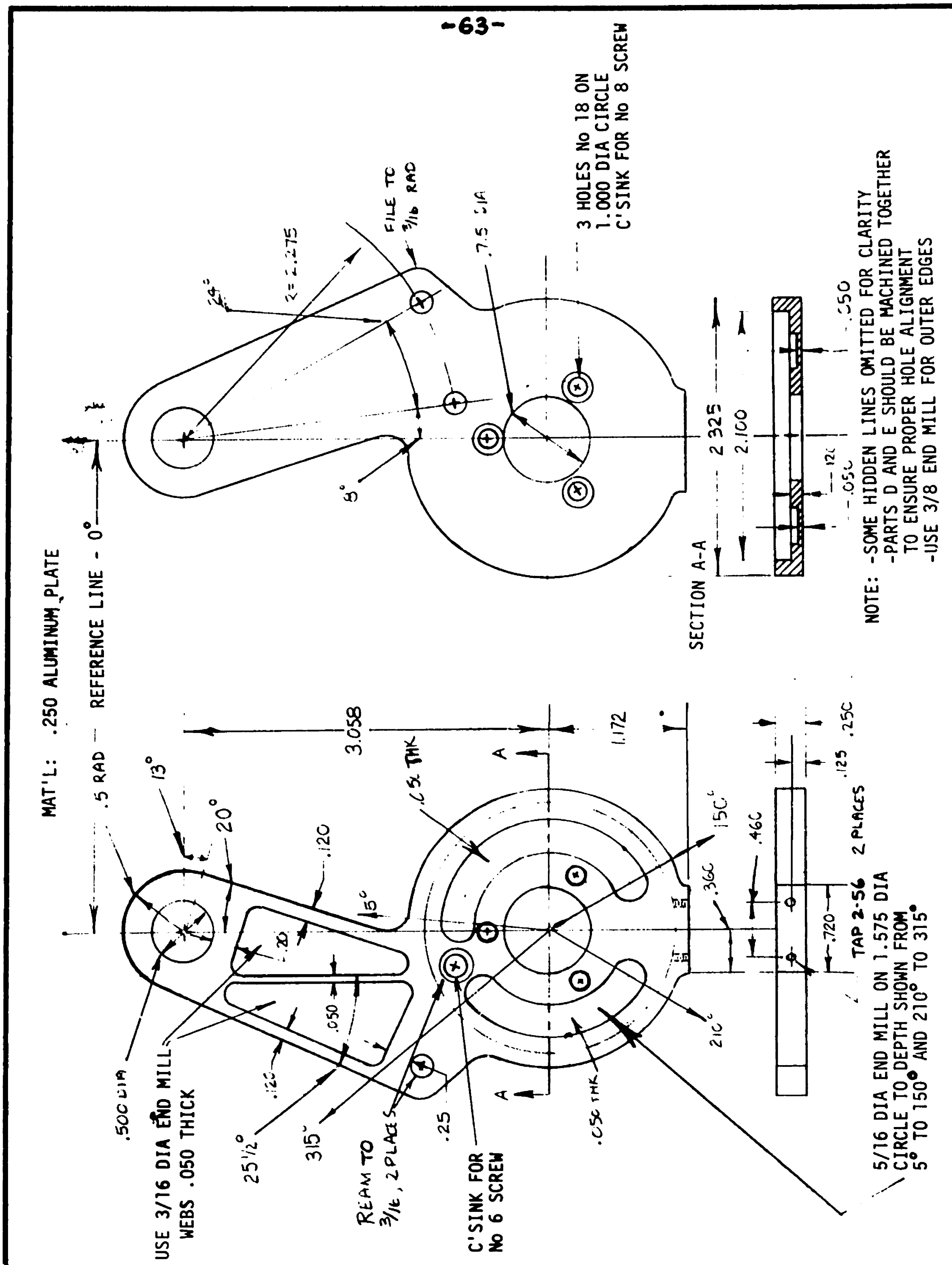
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
OUTER SHELL MANDREL	A	ROCK MAPLE



MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
TOP BRACKET	B	2024-T4 ALUMINUM

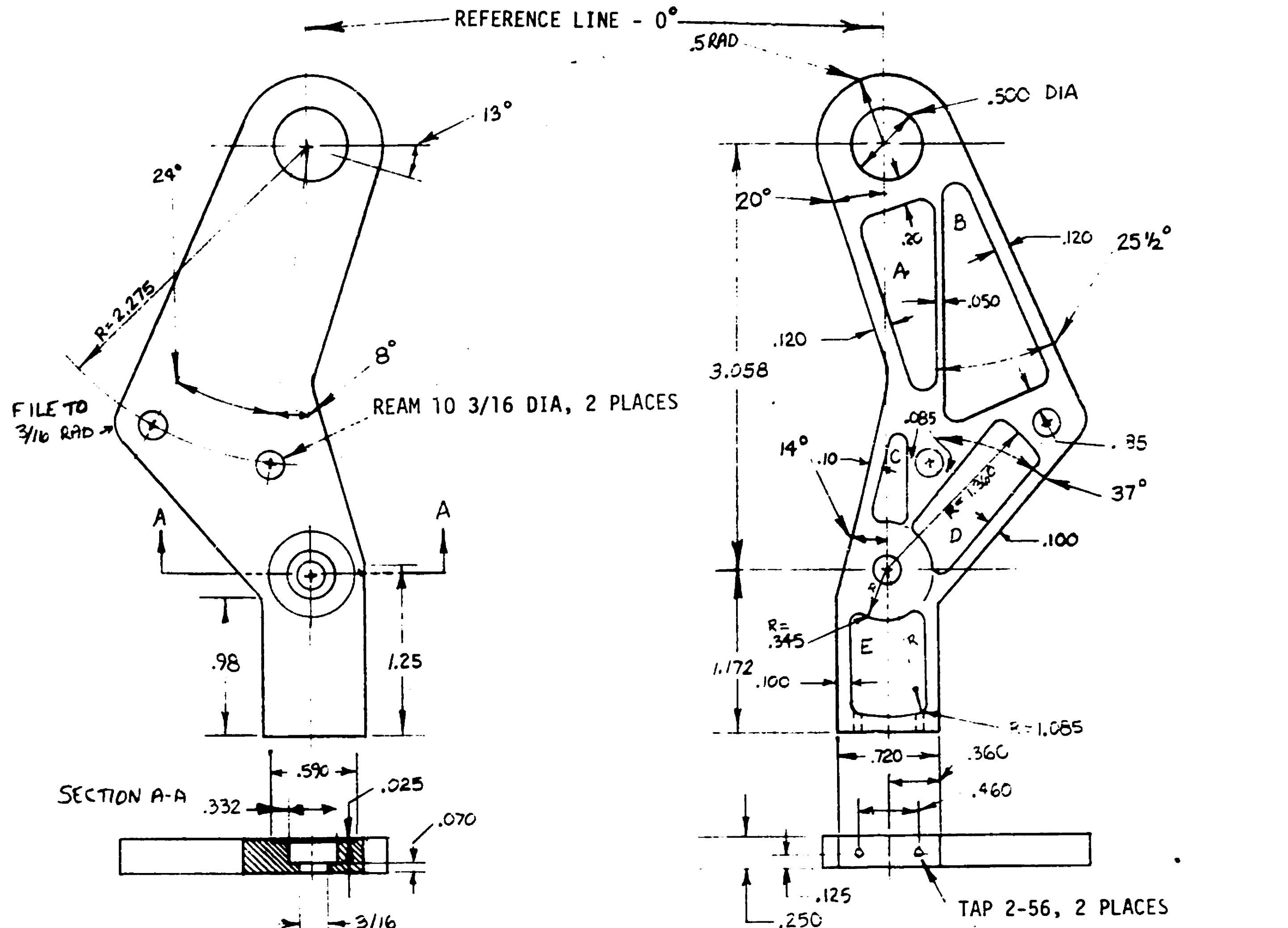


MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
KNEE SHAFT	C	STAINLESS STEEL



MIT KNEE PROJECT SPRING 1977		KENNETH LAUNIE
INNER PULLEY BRACKET	D	6061-T6 ALUMINUM

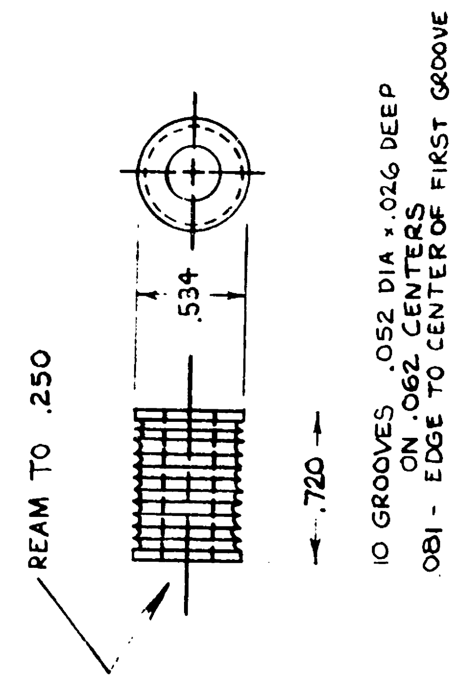
MAT'L: .250 ALUMINUM PLATE



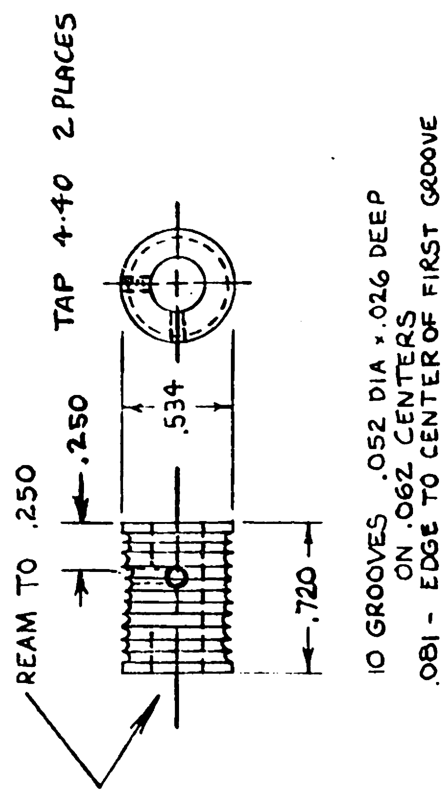
USE 3/16 DIA END MILL x .200 DEEP, WEBS A, B, E
USE 1/8 DIA END MILL x .200 DEEP, WEBS C, D

NOTE:
- SOME HIDDEN LINES OMITTED FOR CLARITY
- PARTS D AND E SHOULD BE MACHINED TOGETHER TO ENSURE PROPER HOLE ALIGNMENT
- USE 3/8 END MILL FOR OUTER EDGES

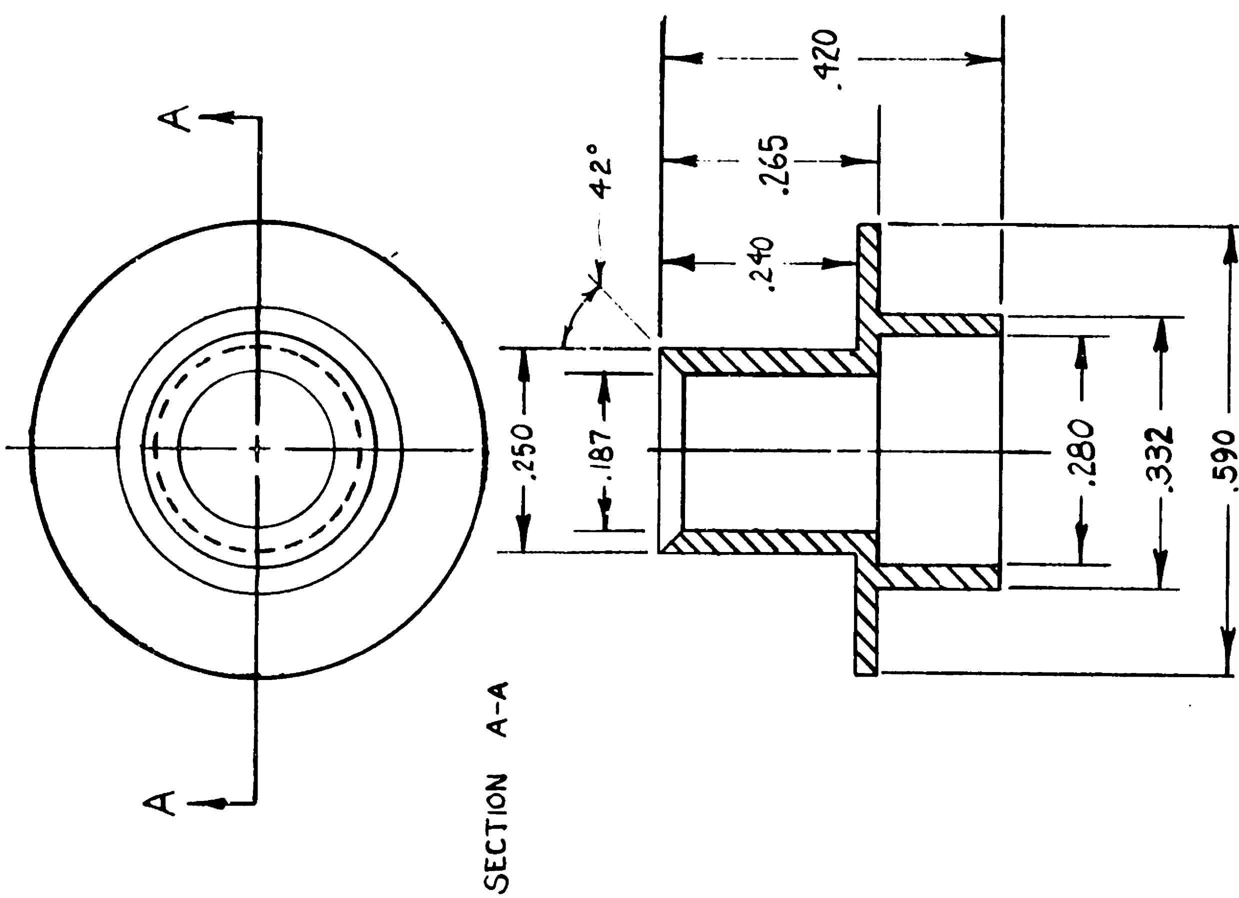
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
OUTER PULLEY BRACKET	E	6061-T6 ALUMINUM



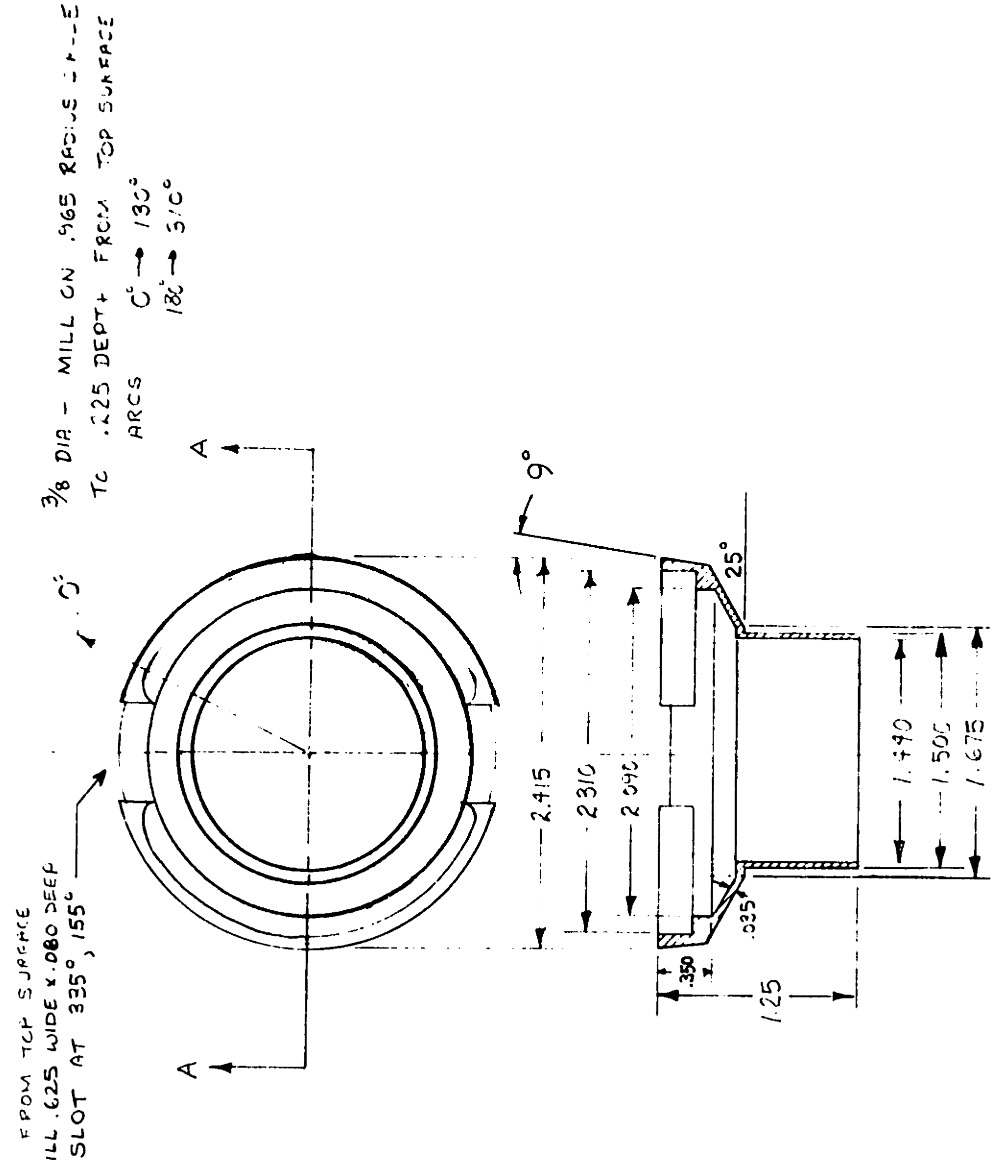
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
INTERMEDIATE PULLEY	F	OILITE BRONZE



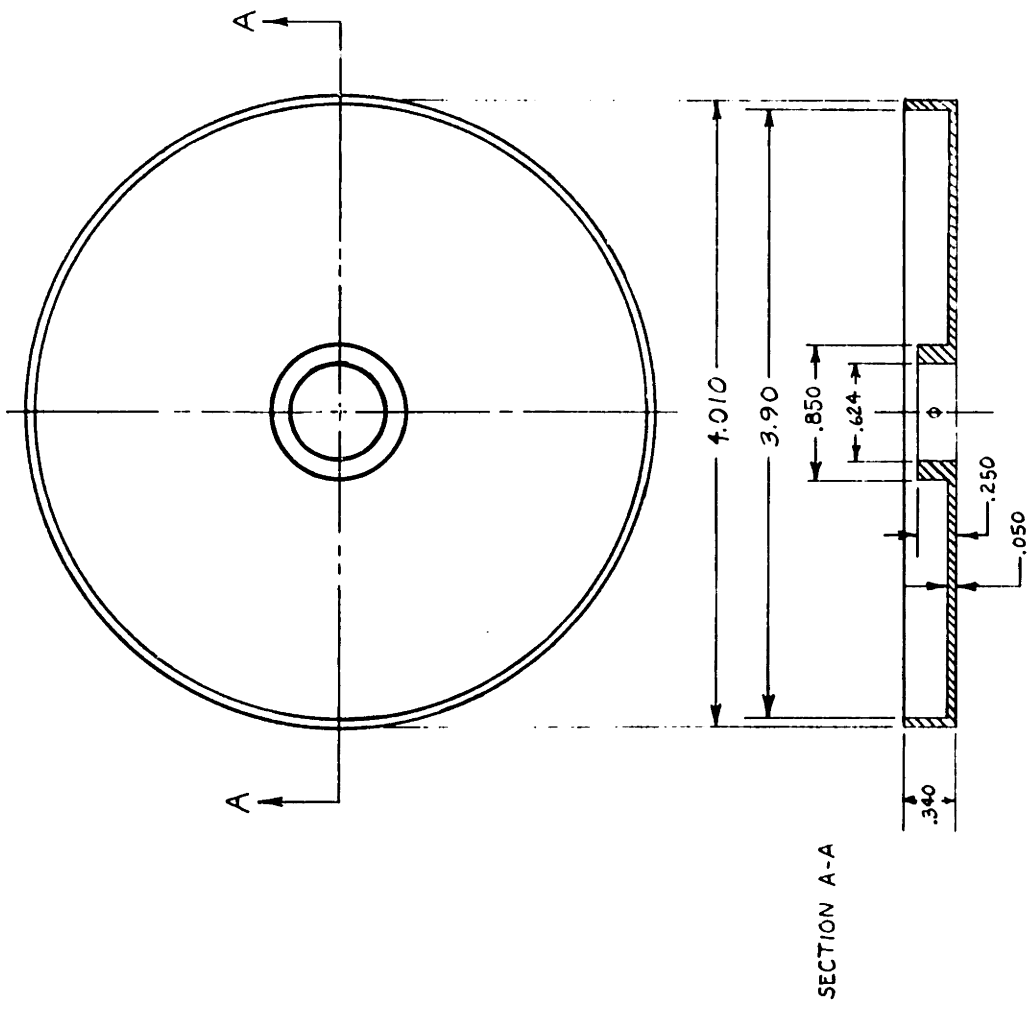
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
MPB PULLEY	G	OILITE BRONZE



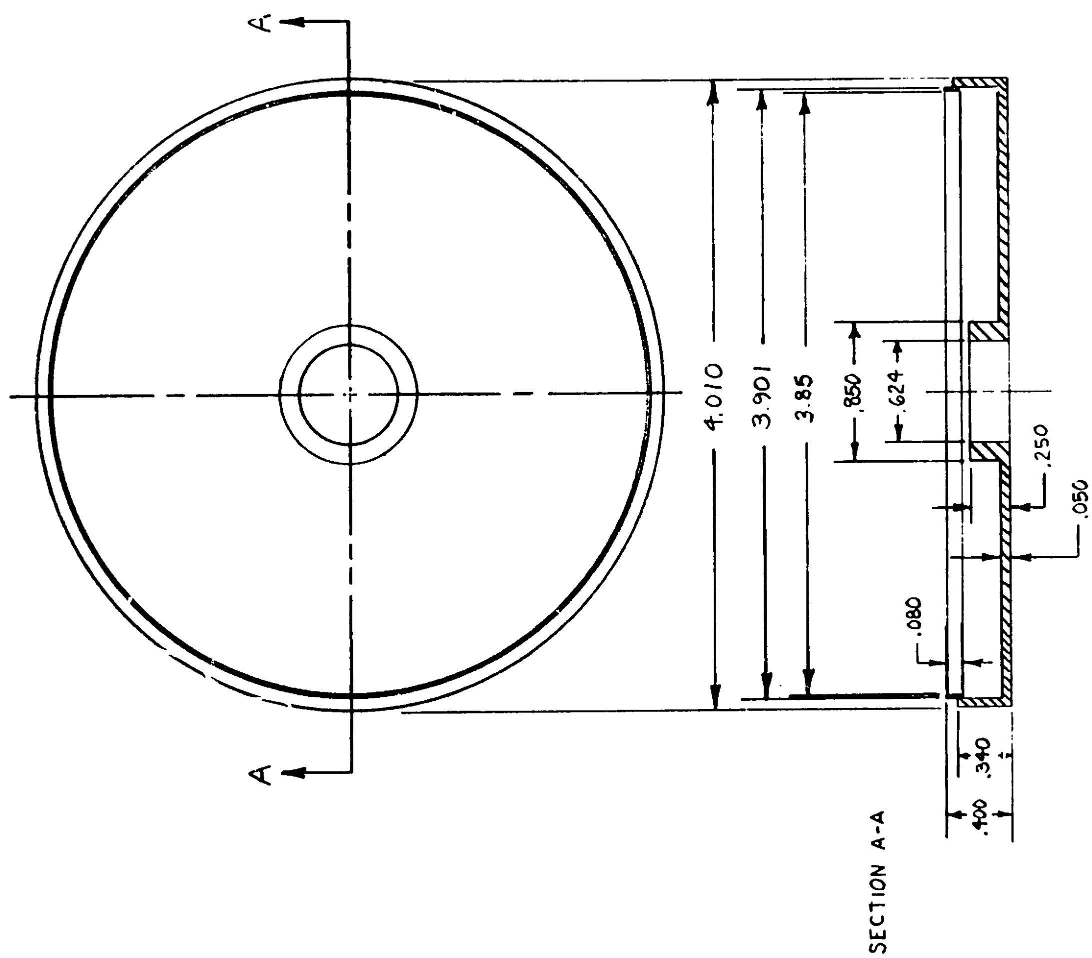
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
OUTER MPB SUPPORT	H	STAINLESS STEEL



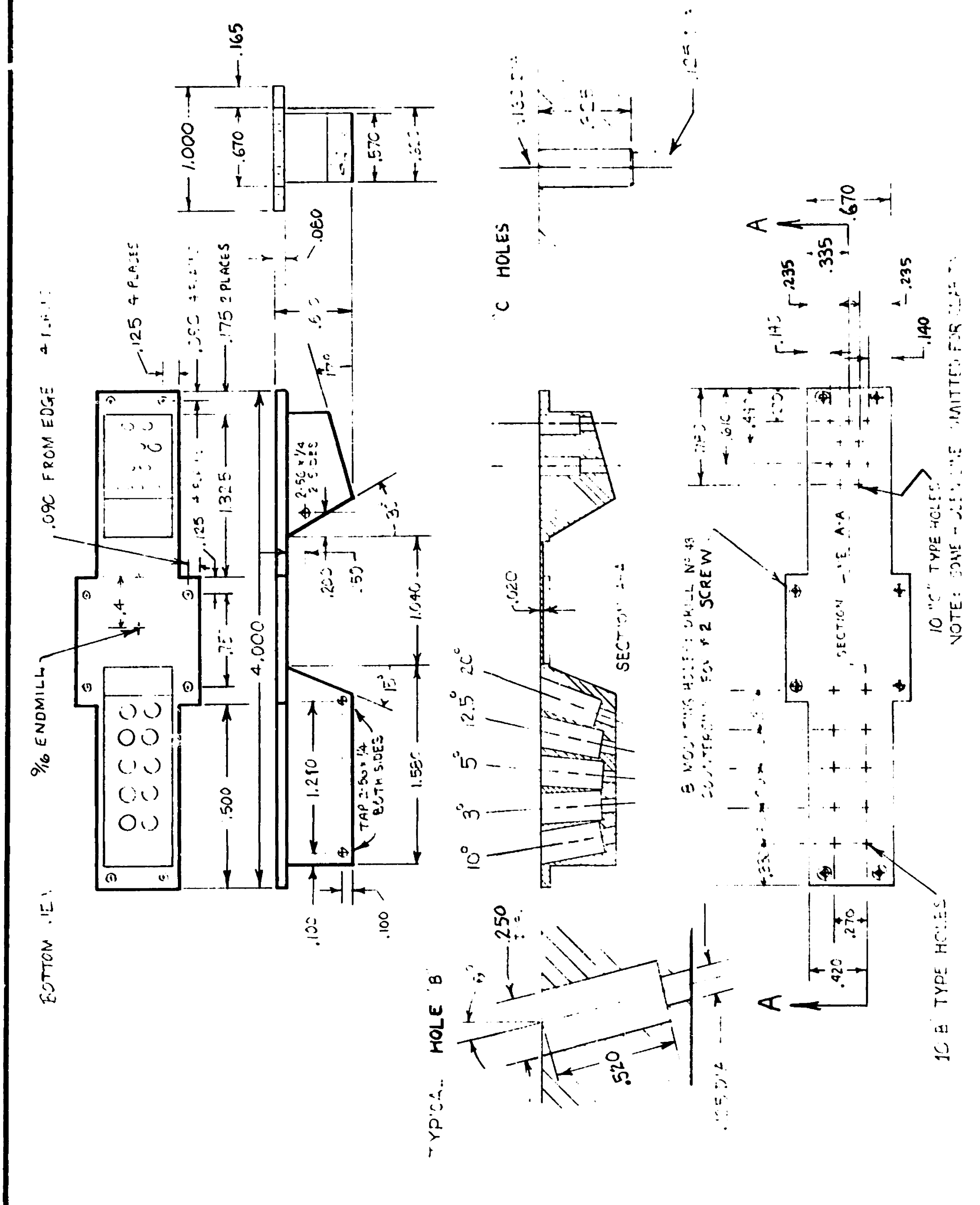
MIT KNEE PROJECT SPRING 1977		KENNETH LAUNIE
BOTTOM BRACKET	I	2024-T4 ALUMINUM



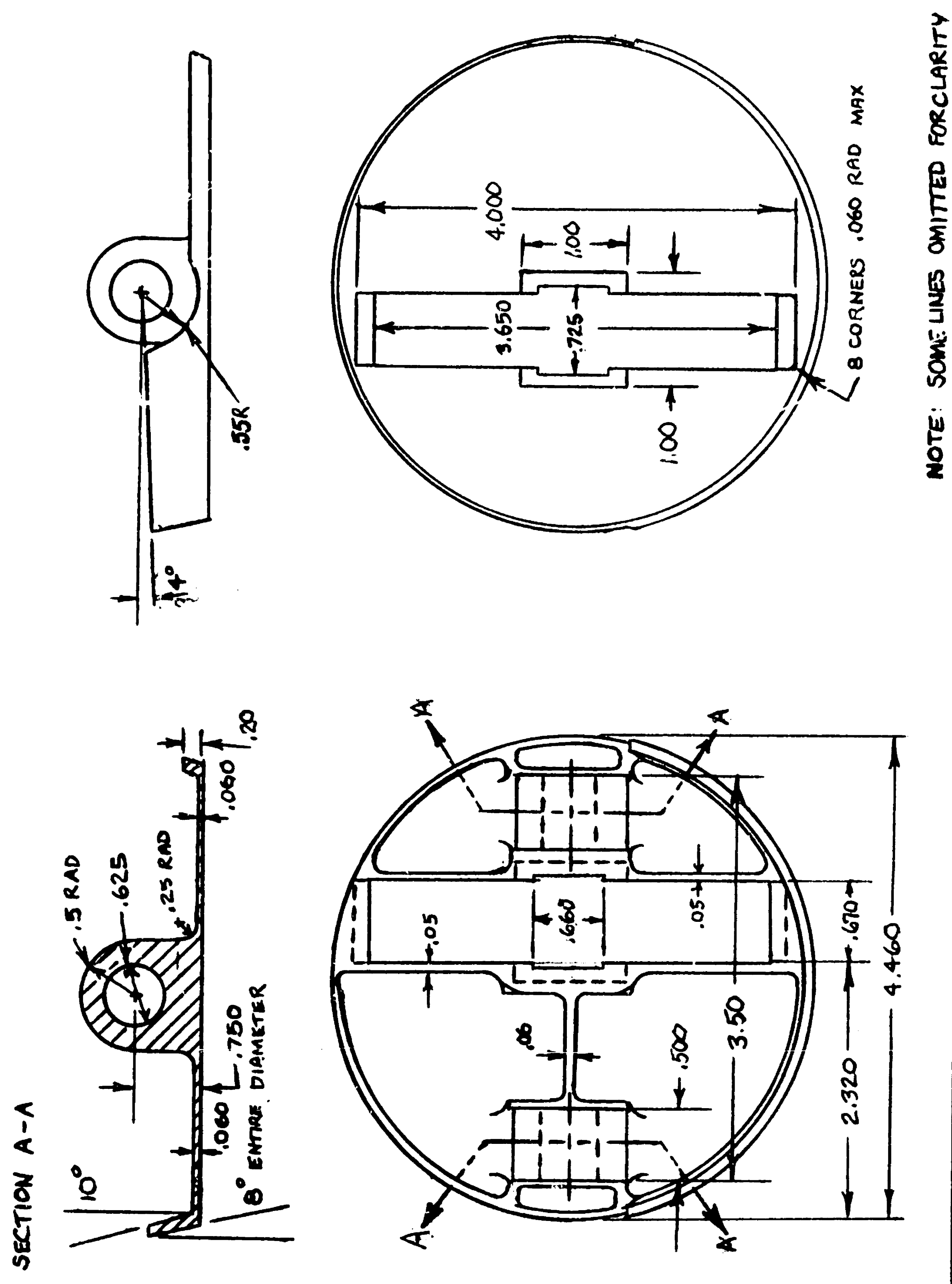
MIT KNEE PROJECT SPRING 1977		KENNETH LAUNIE
DRUM HALF "A"	J	2024-T4 ALUMINUM



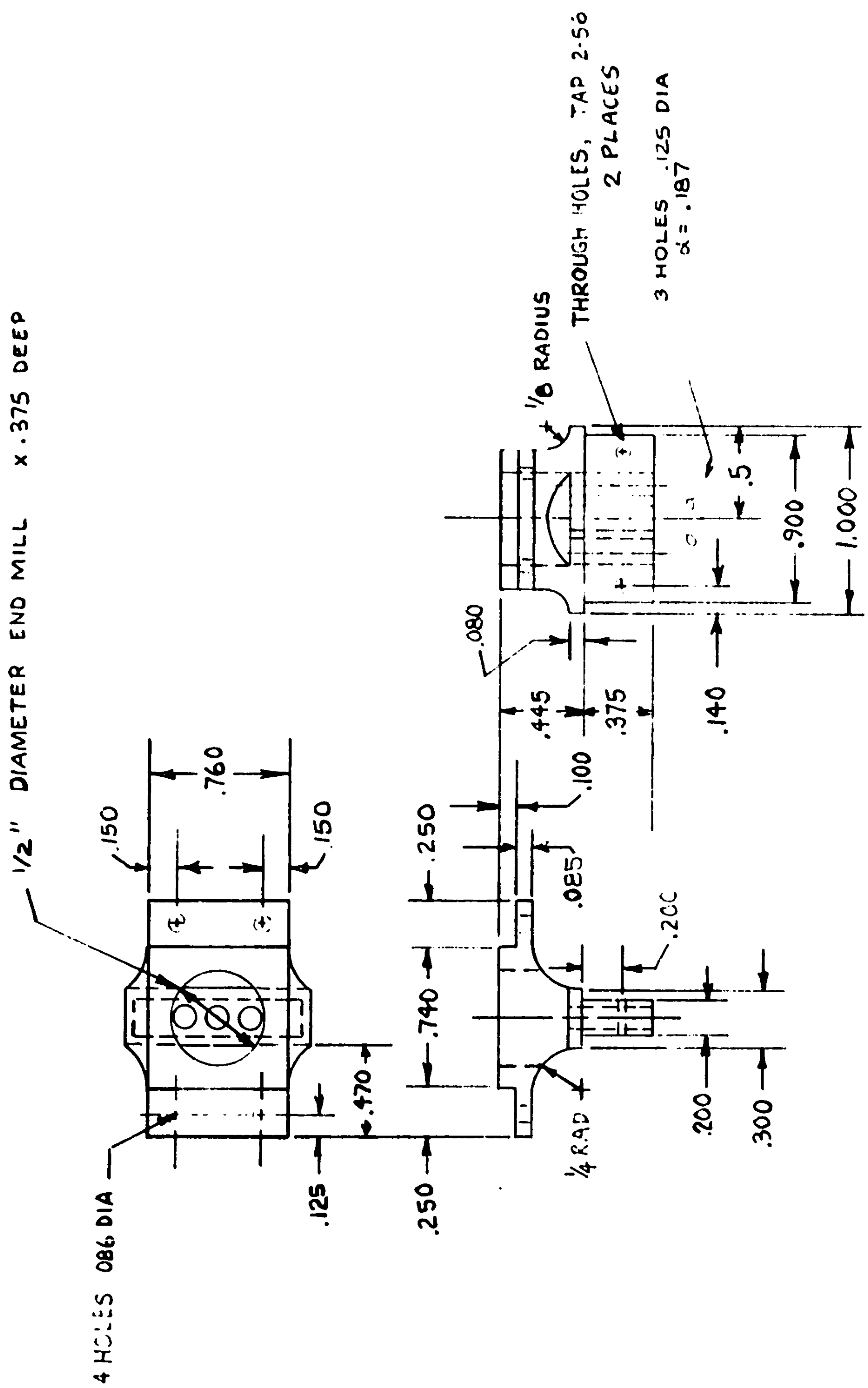
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
DRUM HALF "B"	K	2024-T4 ALUMINUM



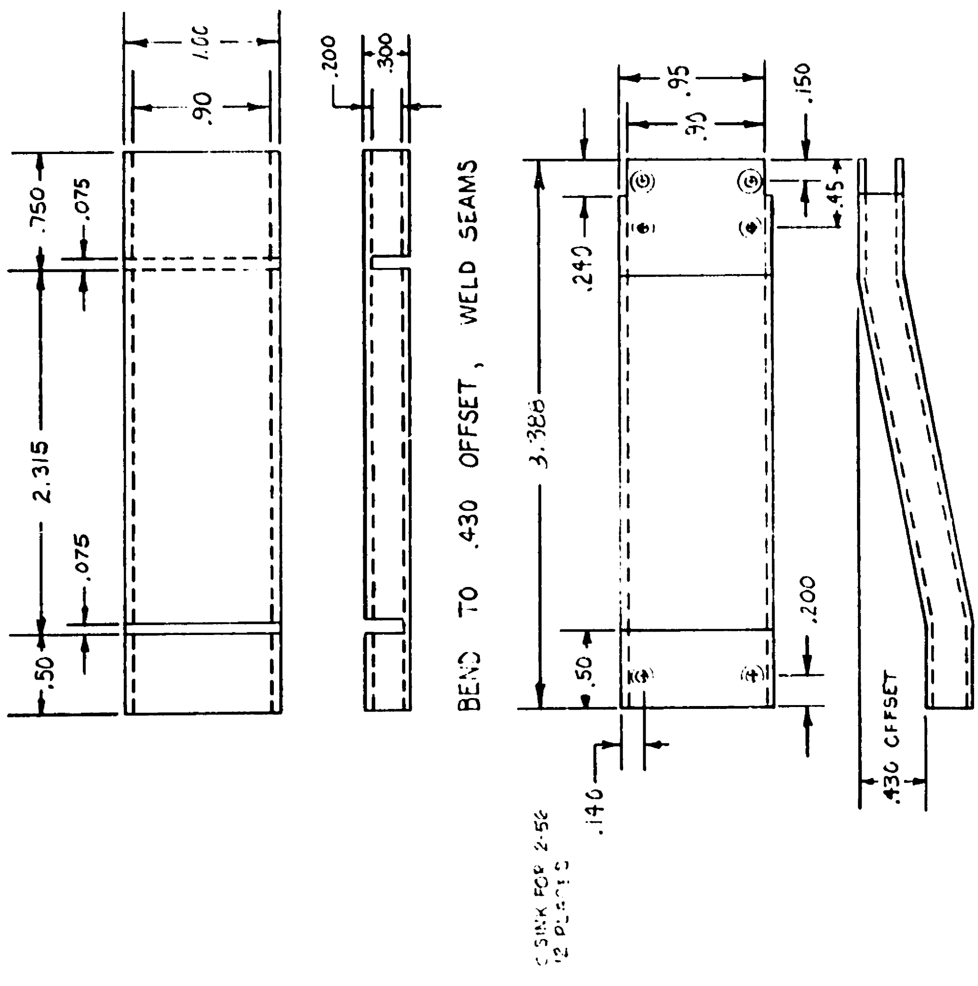
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
CABLE END PLATE	L	2024-T4 ALUMINUM



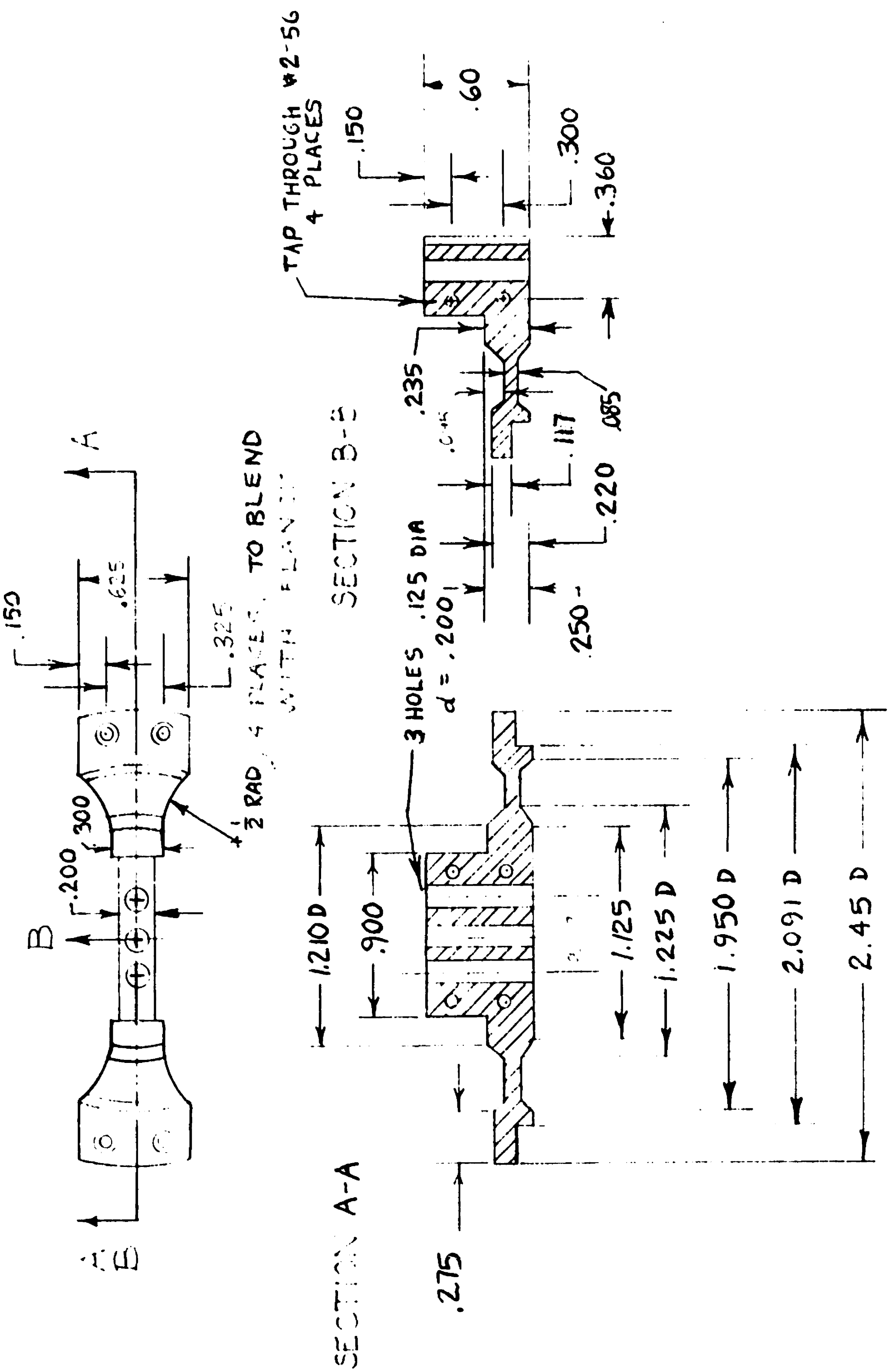
TOP PLATE	M	2024-T4 ALUMINUM



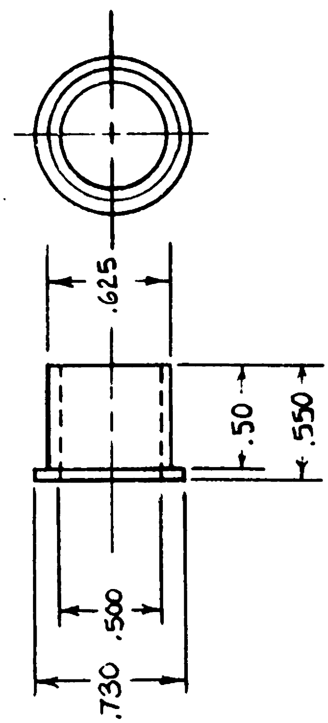
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
TOP TUBE HOLDER	N	2024-T4 ALUMINUM



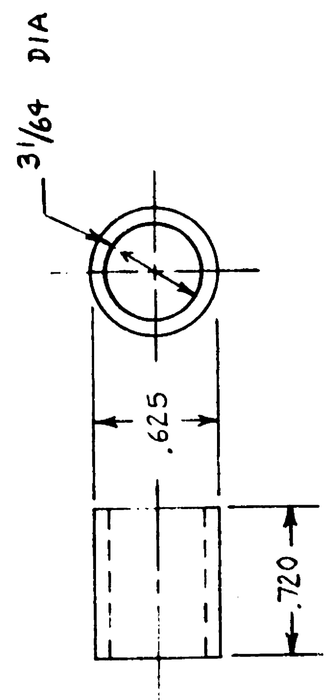
MIT KNEE PROJECT SPRING 1977		KENNETH LAUNIE
TORQUE TUBE	0	EXTRUDED ALUMINUM TUBE



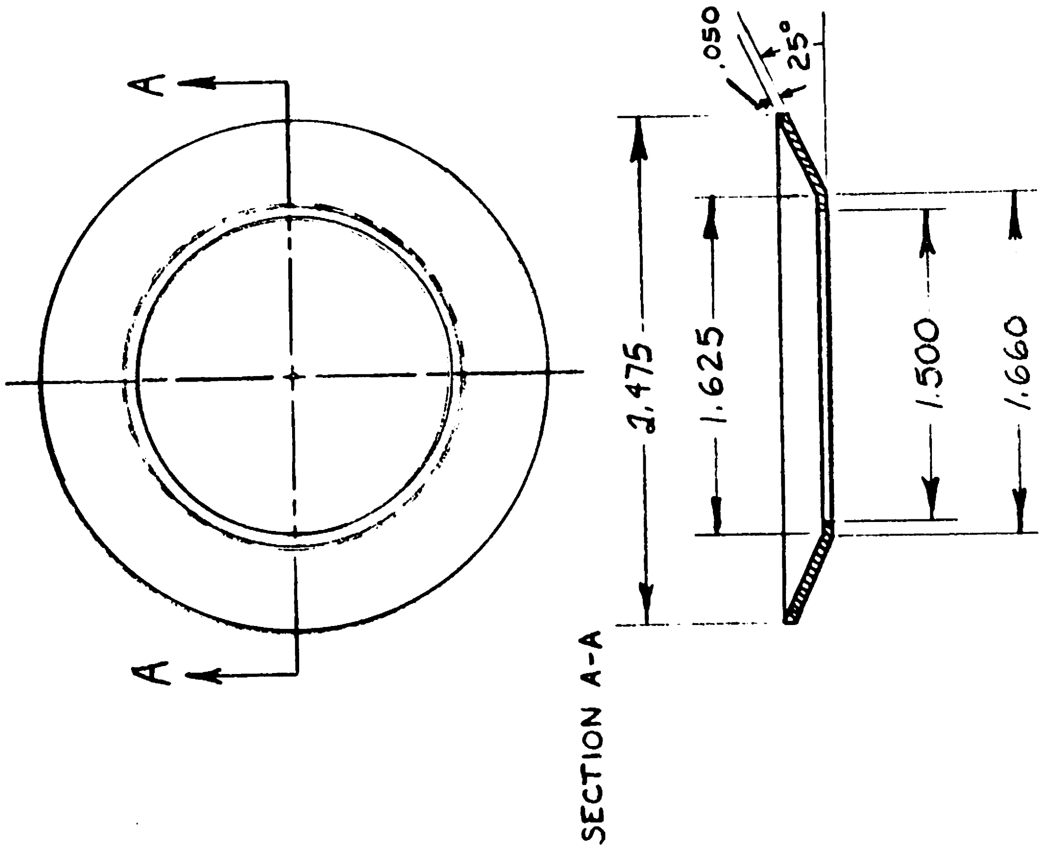
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
BOTTOM TUBE HOLDER	P	2024-T4 ALUMINUM



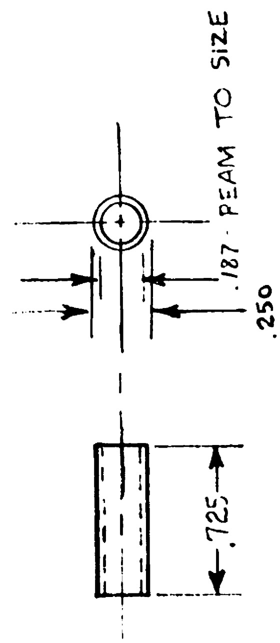
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
KNEE SHAFT BUSHING (2)	Q	OILITE BRONZE



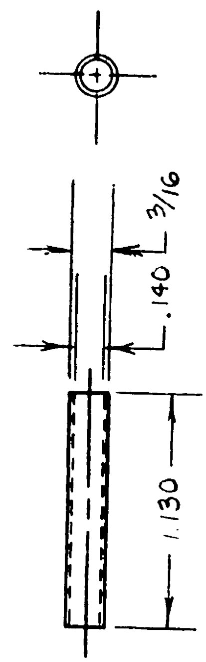
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
DRUM BUSHING	R	OILITE BRONZE



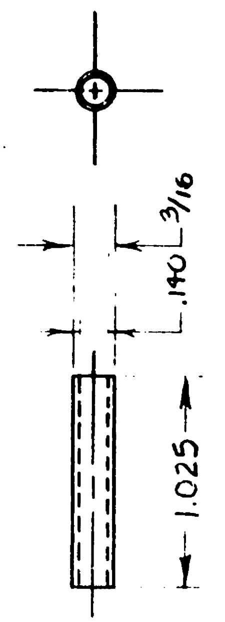
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
BOTTOM PLATE	S	2024-T4 ALUMINUM



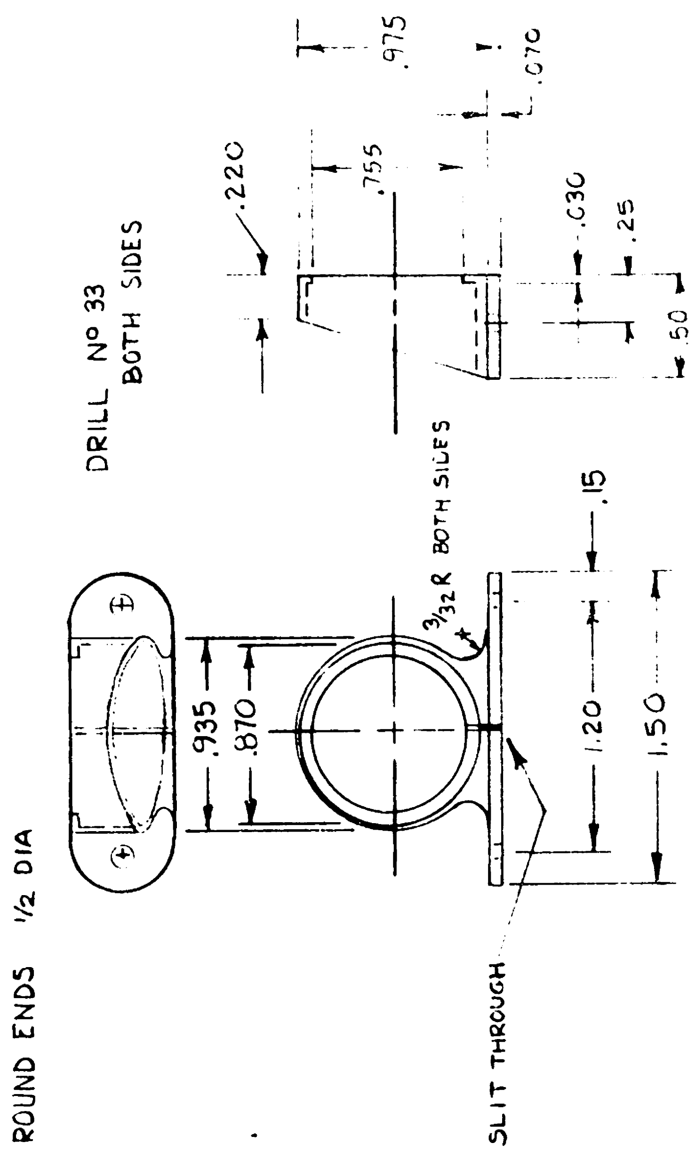
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
BRACKET SPACER (2)	T	STAINLESS STEEL



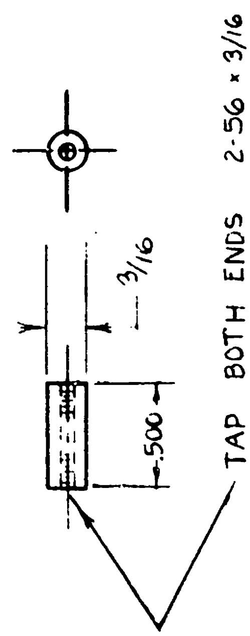
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
LONG LOCATING ROD	U	STAINLESS STEEL



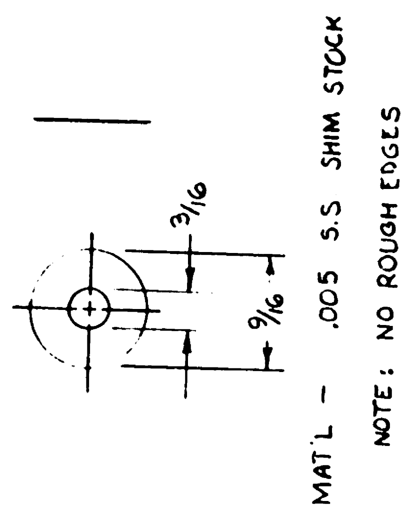
MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
SHORT LOCATING ROD	V	STAINLESS STEEL



MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
POTENTIOMETER HOLDER	W	2024-T4 ALUMINUM



MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
DRUM SPACER	Y	2024-T4 ALUMINUM



MIT KNEE PROJECT	SPRING 1977	KENNETH LAUNIE
PULLEY SPACER	DA	.005 STAINLESS SHIM

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