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REINFORCED LOWER-LIMB ORTHOSIS-DESIGN PRINCIPLES

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For some time there has been a need for a method to express the relative flexural strength of various orthotic components. Specifically, it is desirable to know which dimensions are critical and what impact on flexural strength each dimension has. A metallic sidebar can be characterized either as having a given flexural strength, or more importantly, a given resistance to flexure.

Consider the beam shown in Figure 1, which is supported on each end and with a deforming force applied from above at the center. The maximum deflection of the beam occurs at the center and can be calculated as

$$y_{max.} = \frac{FL^3}{24EI}$$

y max = maximum deflection (see Figure 1) in inches
F = deforming force in pounds
L = length of sidebar in inches
E = modulus of elasticity (material property) in psi
I = moment of inertia in $(\text{inches})^4$.

The two factors in the denominator of equation (*1*) are E, modulus of elasticity, and I, moment of inertia.

The modulus of elasticity is a material property, and thus can be changed only by changing the type of material. As long as the orthotic component is not stressed beyond its elastic limit (i.e., does not stay bent), the modulus of elasticity is a constant (see Table 1 for typical values). It can be seen in Table 1, that changing material from aluminum to steel will increase the modulus of elasticity by a factor of three. Inversely, the same change will produce a reduction in the maximum deflection, y max, in Figure 1 of the

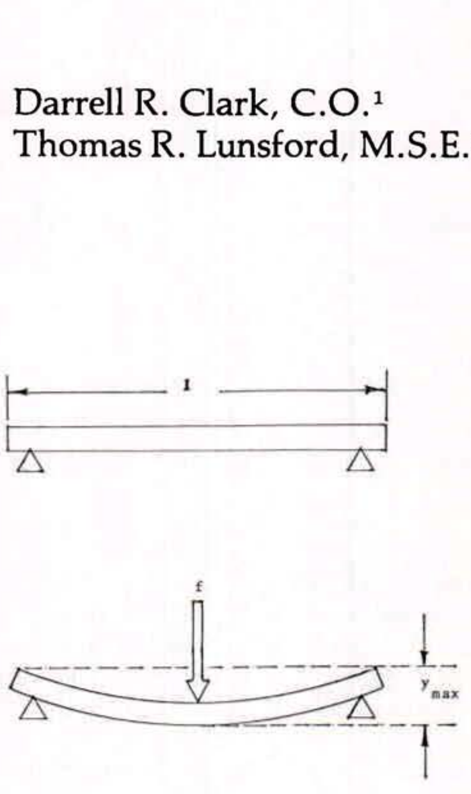

Fig. 1. Basic Beam Deflection

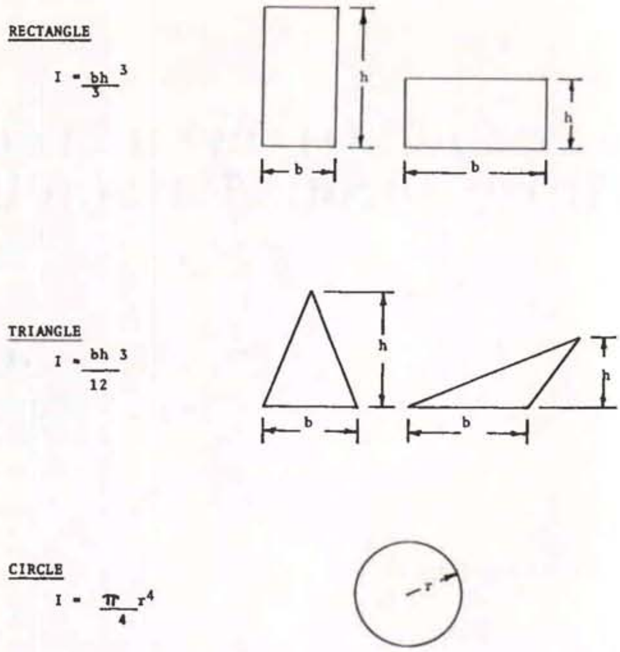

Fig. 2. Moment of Inertia

orthotic component by a factor of three.

The other factor in the demonimator of equation (*1*) is the moment of inertia, I, which relates the cross-sectional shape of the orthotic component to its strength. The higher the moment of inertia, the less will be the deflection for a given deforming force. Several cross-sectional shapes and their corresponding moments of inertia are shown in Figure 2.

Three examples of how equation (*1*), maximum deflection, can be used for design purposes are presented below.

KAFO Genu Valgum

The first case involves a postpolio patient who had bilateral KAFOs and complained that his orthoses flexed medially during weight-bearing (Fig. 3). The flexure resulted in significant instability during stance, and chronic fracture failure of the medial sidebar.

Fig. 3. KAFO with Medial Flexure

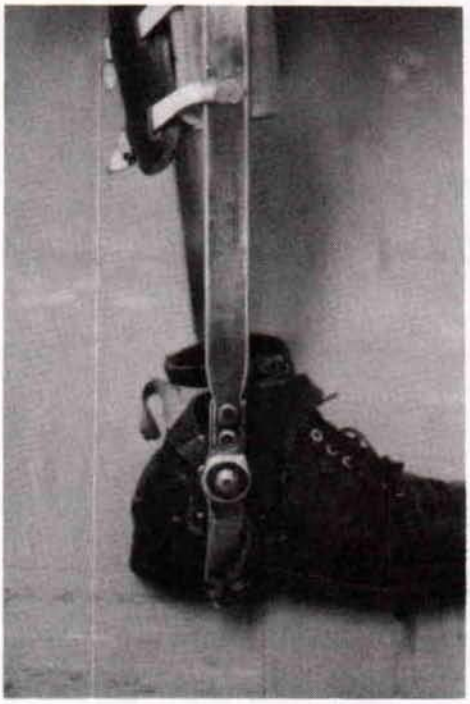

Fig. 4. One Inch Sidebars

In an attempt to correct this condition, the 3/4-inch sidebars were replaced with one-inch wide sidebars, as shown in Figure 4, but without success.

In analyzing the problem, it is necessary to first determine the deforming force producing the knee valgum. That force can be calculated with help of Figure 5 through the following equation:

$$F = \frac{(B.W.) \sin \phi}{2 \cos \theta}$$

F = knee valgum deforming force in pounds

B.W. = body weight in pounds

ϕ = knee valgum angle in degrees

θ = hip abduction angle in degrees

The deforming force calculated by equation (2) will increase if body weight, hip-

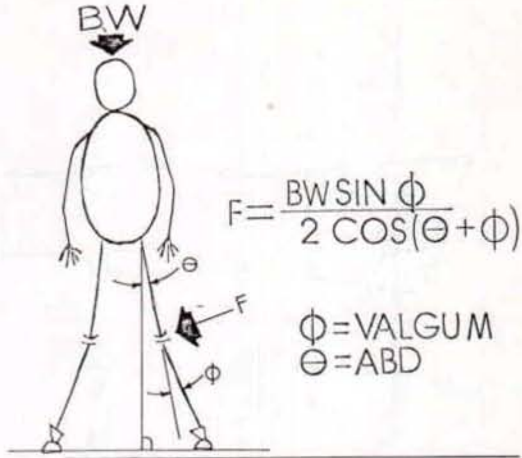

Fig. 5. Force Diagram

abduction angle, or knee-valgum angle are increased. It is necessary to estimate this deforming force as it is directly responsible for the deflection of the knee joint of the KAFO.

The sidebars of a KAFO can be represented as a rectangular bar supported at each end with a deforming force being applied in the middle as shown in Figure 6. The maximum deflection, y max, can be obtained by subsituting equation (*2*) into equation (*1*).

$$y_{max} = \frac{(B.W.)\, L^3 \sin \phi}{24\, EI \cos \theta}$$

In analyzing the terms of equation (*3*), it is apparent that there are certain factors which cannot be controlled; for instance, body weight, the uncorrectable knee valgum, and hip abduction angles are not easily changed. The valgum angle, ϕ, can be controlled if the valgum is correctable, but would be minimal. The abduction angle, θ, would be determined by the patient's gait and corresponding stance stability. The length of the KAFO, L, is fixed. Therefore, the KAFO sidebar material is the remaining element

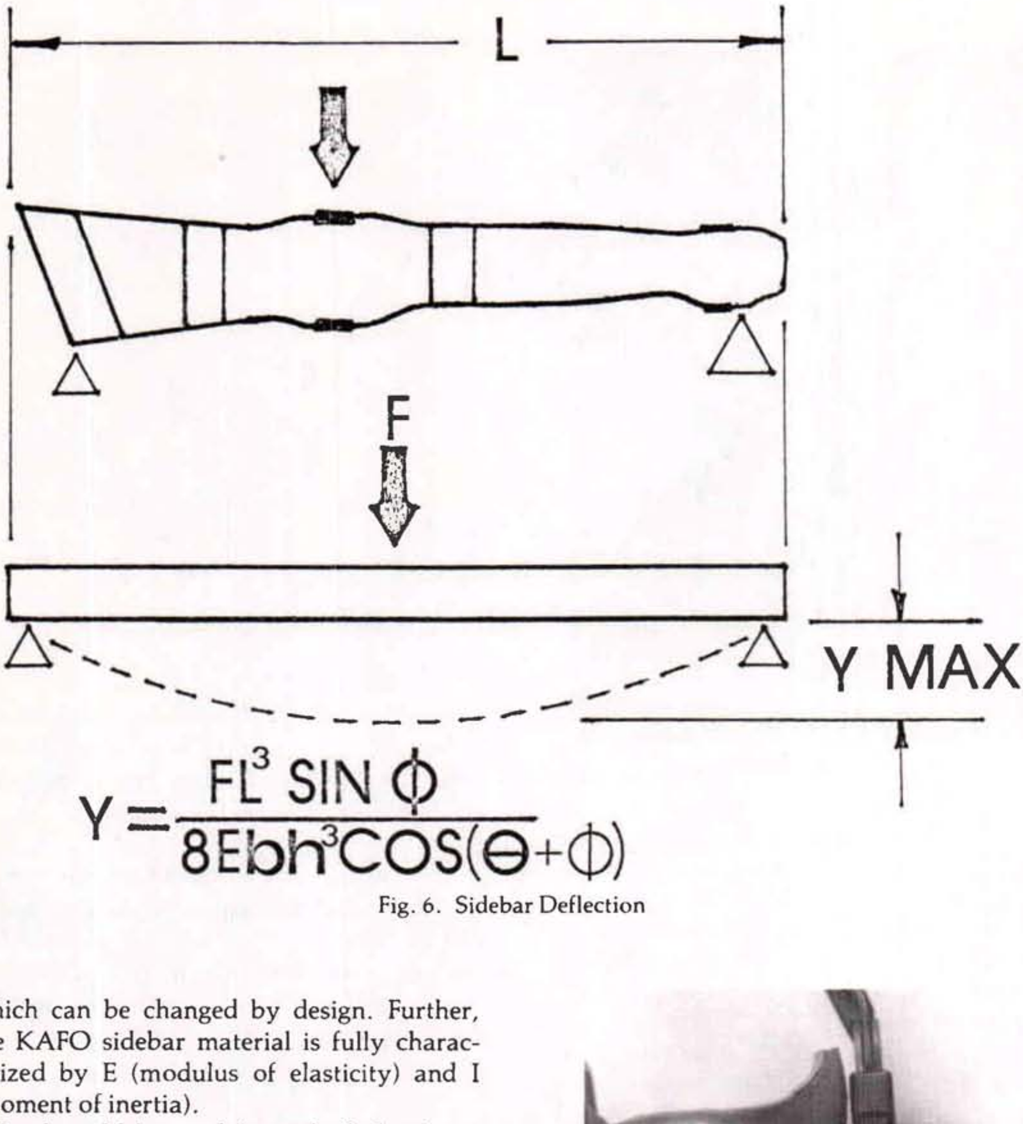

Fig. 6. Sidebar Deflection

which can be changed by design. Further, the KAFO sidebar material is fully characterized by E (modulus of elasticity) and I (moment of inertia).

Steel could be used instead of aluminum for the sidebars since the modulus of elasticity for aluminum and steel is 10,000 psi and 30,000 psi, respectively. This change will cause maximum deflection, y max, to be reduced (improved) by a factor of three.

The remaining ingredient in equation (*3*) is I, the moment of inertia, the indicator of the strength of a geometric shape. This parameter depends solely upon the shape of the cross-sectional area of the sidebars. According to Figure 2, the moment of inertia

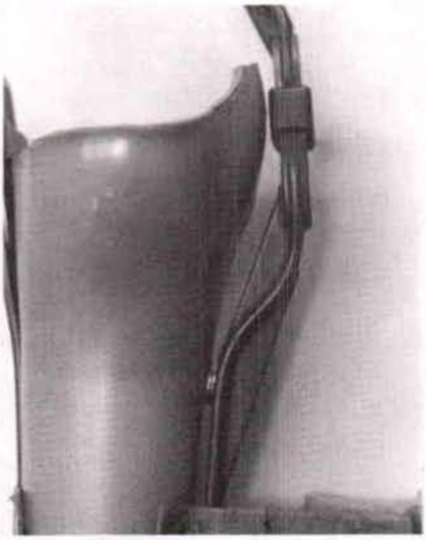

Fig. 7. Reinforced Sidebar

for a rectangular cross-sectioned beam is given by:

$$I = \frac{bh^3}{3}$$

The cross-section of a typical sidebar has a base (b) of 0.75 in. and height (h) of 0.25 in. Traditionally, enlarging the base was attempted as a means of strengthening the KAFO. In the case of this postpolio patient, the width (base) of the side bars was increased by .25 in., from 0.75 to 1 inch. Substituting .75 and 1.0 for the sidebar widths (base) into equation (4) yields, $I = .75h^3/3$ and $I = 1.0h^3/3$. The moment of inertia will increase in the same proportion (25 percent) as the base. Further, the genu valgum deformity of the KAFO, y max, will be reduced (25 percent) directly in proportion to how much (b) is increased.

However, increasing the thickness (h) of the sidebar by the same .25 inch will produce a more dramatic effect on the moment of inertia and hence the amount of genu valgum deformity. For example, if the thickness is increased from .25 to .50 inch, the moment of inertia becomes $I = .015bh^3$ and $I = .13b/3$, a factor-of-8 increase. The genu valgum deformity will accordingly be reduced by a factor of eight.

In the case of the postpolio patient the new KAFO's were modified by welding perpendicular struts to the sidebars in the vicinity of the knee-joint contours as shown in Figure 7. This design resulted in increasing the thickess (height) by a factor of three over its original value. The moment of inertia increased by a factor of 27 (or 3^3). The genu valgum deformity was accordingly reduced by a factor of 1/27. The patient's medial-lateral stability was improved with this modification as shown in Figure 8.

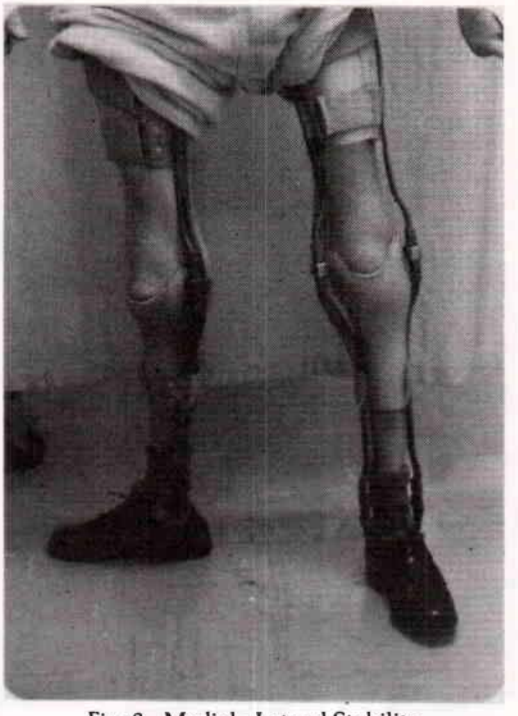
Fig. 8. Medial - Lateral Stability

Stirrup Failure

Because of the severity of involvement of many patients seen at Rancho Los Amigos Hospital, it is often necessary to increase the ankle and knee stability through the use of a locked-ankle AFO. This produces a severe bending moment on the tongue of the stirrup causing transverse fracturing of the tongue as shown in Figure 9.

Commercially available heavy-duty stirrups have been utilized in these instances. There is a percentage of patients who fracture the heavy-duty stirrups (Fig. 10). Historically, the heavy-duty stirrups were reinforced with struts welded from the vertical member of the stirrup to the tongue, across the tongue, and over the other side to the other vertical member (Fig. 11). There are

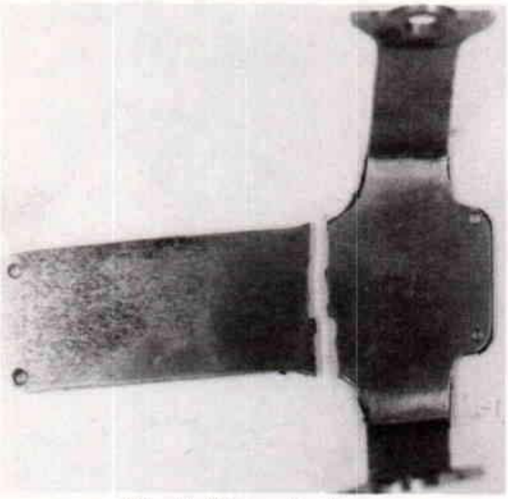

Fig. 9. Stirrup Fractures

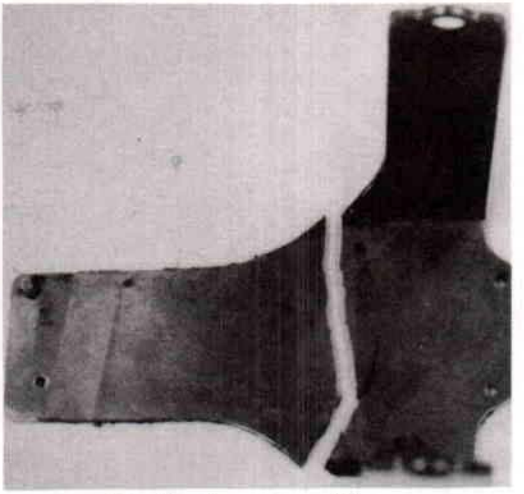

Fig. 10. Heavy Duty Stirrup

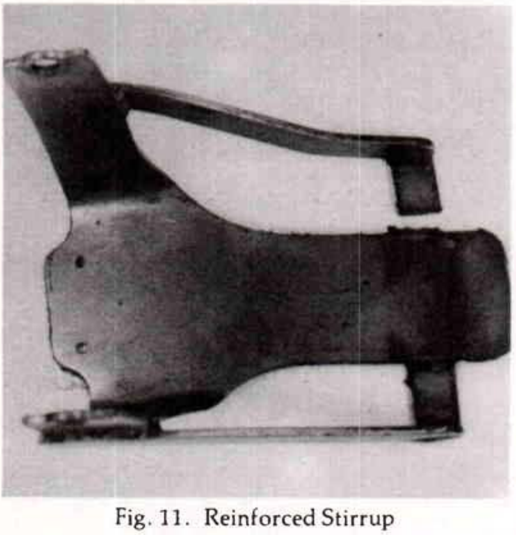

Fig. 11. Reinforced Stirrup

patients who fracture this reinforced stirrup.

In analyzing this problem, the deflection, y max, created by the reactive force of the tibia and the ankle dorsiflexion angle in terminal stance as shown in Figure 12a must be considered. Equation (*4*) for the moment of inertia of a rectangular cross-section where "b" is the base and "h" is the height is applicable here. The cross-section of the tongue of the stirrup is shown in Figure 13. Increasing the height will produce the highest moment of inertia and the most resistance to flexure. However, this would result in a very thick stirrup that would be extremely heavy and very difficult to attach to the bottom of the shoe.

Fortunately, the same strength advantage can be gained by making 90 deg. contours at the edges of the stirrup as shown in Figure 13. A stirrup which extends the full width of the shoe and curves 0.5 inch superiorly on each side, as shown in Figure 14, was fabricated in two lengths, 8 in. and 6 inches (Fig. 15).

During the terminal stance phase of gait the simplified force diagram for a stirrup is shown in Figure 12b. For a patient in the terminal stance phase of gait the forces are as shown. The calf force is F, the ankle joint reaction is P, and the force tending to flex the shank of the stirrup is equal and in opposite direction to the body weight (B.W.). For purposes of analysis the stirrup may be treated as a beam suspended on one end and a deforming force applied at the other end.

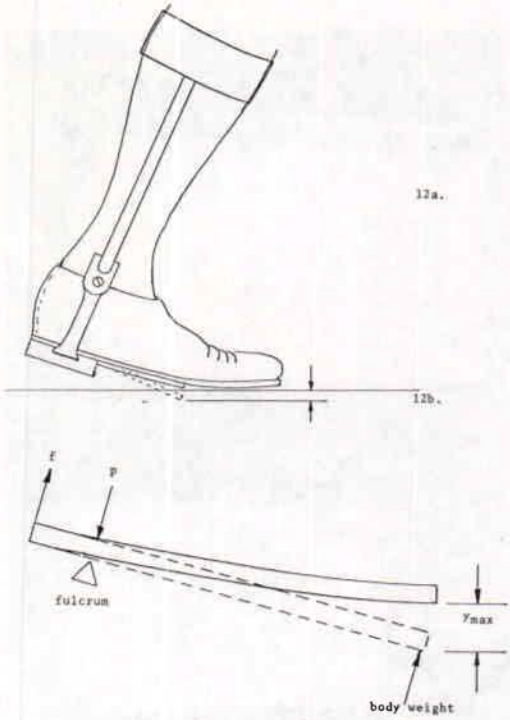

Fig. 12. Force Diagram, Terminal Stance

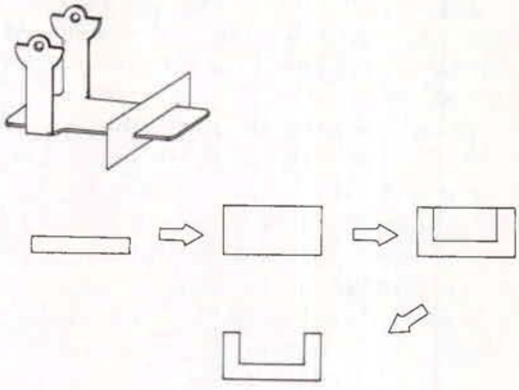

Fig. 13. Stirrup, Transverse Section

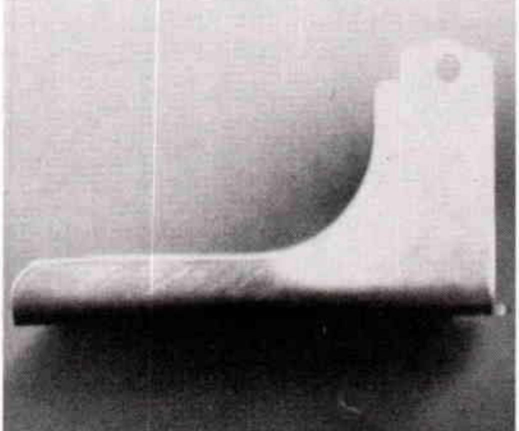

Fig. 14. Photo of Reinforced Super Stirrup

For this type beam the deflection at the end, y max, is given by

$$y_{max} = \frac{WL^3}{3\,EI}$$

where

- W = the deforming force applied at the end (in this case it is equal to the patient's body weight)
- L = length of stirrup undergoing flexure. For an 8 inch stirrup, L = 5 inches
- E = Modulus of elasticity with 30 x 10^6 psi for stainless steel
- I = moment of inertia

The moment of inertia for a conventional stirrup with a 2.0-in. width and .125-in. height is .0013 $in.^4$ The moment of inertia for the contoured stirrup is .044 $in.^4$ Substituting these values into equation (5) yields end-deflections for the standard stirrup and contoured stirrup of 0.16 in. and .0016 in. This represents a deflection of one-one hundredth of that allowed by the conventional design.

A sizable improvement in fatigue life is expected from the contoured stirrups since fatigue life is dependent upon the number of flexures (steps) and the amplitude of each flexure.

A weight comparison between the conventional commercially available heavy-duty stirrup and the reinforced Rancho Los Amigos stirrup shows an increase of 5 to 10 ounces depending on size. The reinforced stirrup installed on the shoe does not interfere with the metatarsal toe break thereby allowing the patient normal gait dynamics.

Polypropylene AFO

In those cases where the orthotic objective is to stabilize the tibia during stance, the conventional polypropylene AFO allows excessive dorsiflexion range. Further, continued flexure from the neutral position into dorsiflexion and back to neutral promotes the common fatigue fracture on the posterior region of the AFO at the talo-crural axis as shown in Figure 16. Therefore, the design goal was to introduce maximum resistance to dorsiflexion without a severe weight penalty or compromise to cosmesis.

Conventional polypropylene AFO's produce an obvious bulge at the ankles in terminal stance (Fig. 17), thereby losing stability.

In an attempt to minimize the dorsiflexion metallic reinforcing struts were added to the lateral and medial sides of the AFO. Again, the design was directed at taking advantage of the height-cubed (h^3) factor, in the moment of inertia equation.

The reinforcement strut used in this case is .5 x .125-in. steel. The .5-in. dimension was "h" and the .125 was "b" as shown in Figure 18. It is contoured to avoid the malleoli. The general location of reinforcement strut is posterior to the malleoli. This allows modification of the plastic over the malleoli if required.

The reinforcement struts should extend into the foot portion of the orthosis and superiorly six inches below the proximal fibular head.

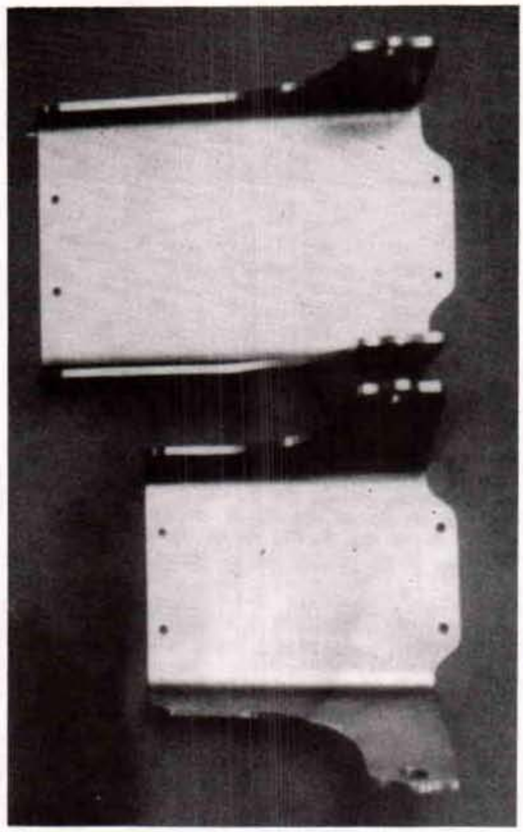

Fig. 15. Two Lengths

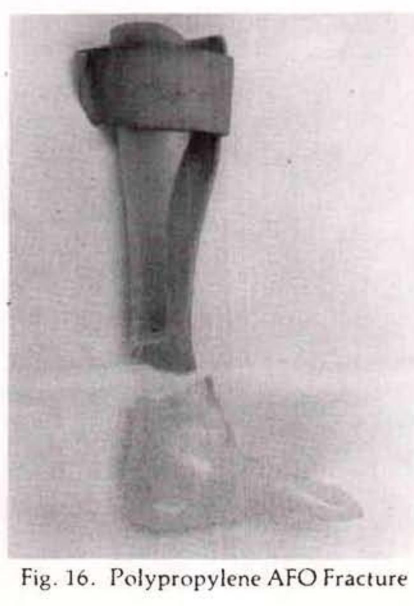

Fig. 16. Polypropylene AFO Fracture

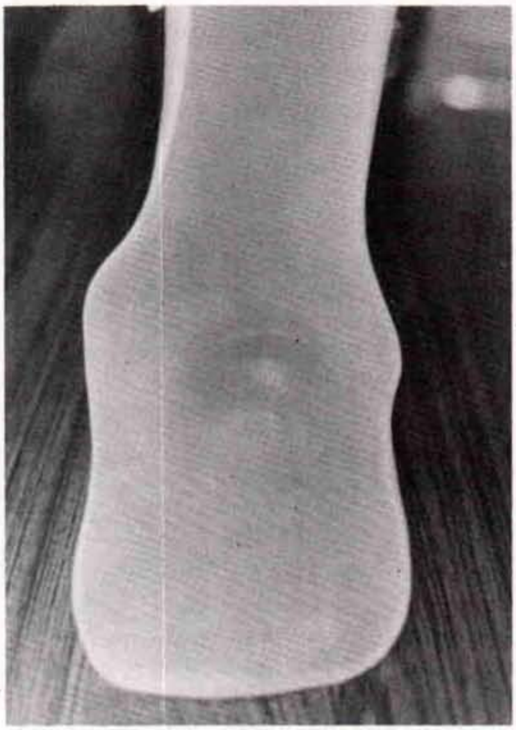

Fig. 17. Characteristic Bulge

With the reinforcement in place it is possible to maintain a narrow medial lateral profile allowing the orthosis to fit into the shoe easily and be acceptable cosmetically.

A testing apparatus was developed using a below-knee prostheses which had been modified to allow dorsiflexion. The testing apparatus is shown in Figure 19.

Polypropylene AFO's with and without the reinforcement were vacuum formed to fit the prosthesis used on the test fixture. The foot portion of the prosthesis was anchored to a stable base and dorsiflexion angle indicator calibrated in degrees was attached. A force gauge was attached to the proximal end of the prosthesis and an anterior tibial force was applied to simulate the stance force of the weightbearing limb. The amount of dorsiflexion deformity of the AFO as well as the force required to produce the deflection was recorded. The results are shown in Figure 20.

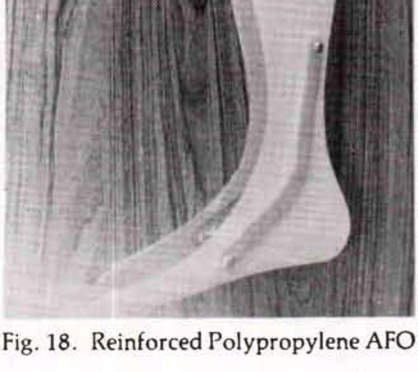

Fig. 18. Reinforced Polypropylene AFO

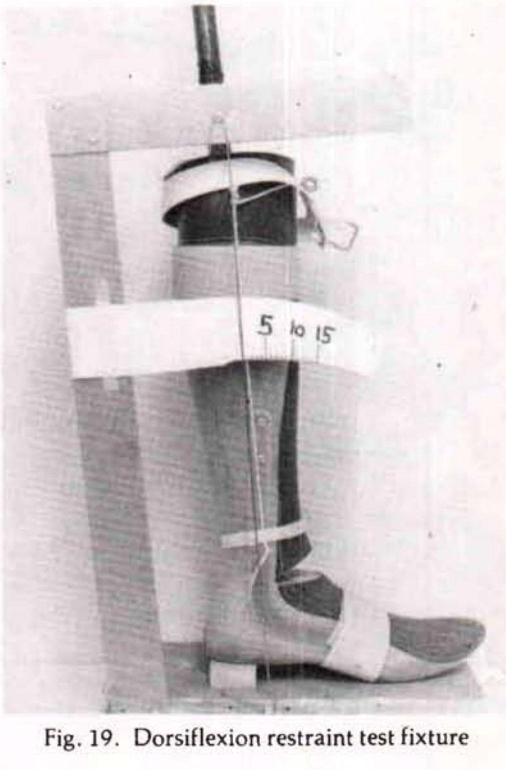

Fig. 19. Dorsiflexion restraint test fixture

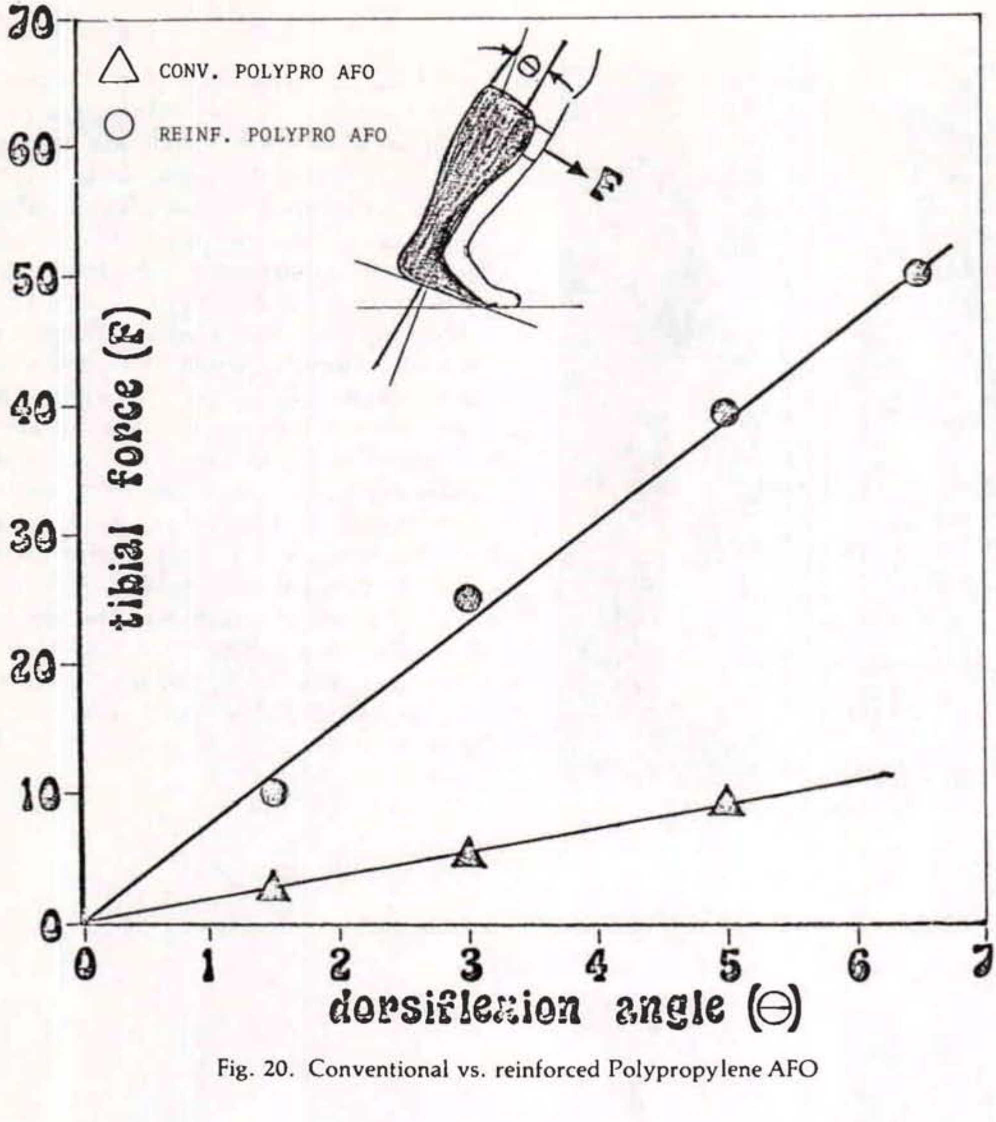

Fig. 20. Conventional vs. reinforced Polypropylene AFO

It was determined that without reinforcement, only eight (8) pounds are required to produce a 5- or 6-degree angle of dorsiflexion. With the strut reinforcements, a minimum of 50 pounds of force was required to produce the same amount of deflection.

Summary

Three lower-limb orthoses have been examined with the intent of improving stability. The principal design concept involved the careful orientation of the reinforcing members to provide an optimum moment of inertia.

These ideas and supporting data show promise. Our plans are to continue to evaluate and fit patients with modifications of these devices in an effort to gain further experience and add credence to our early results.

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Footnotes

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TABLE 1.
Modulus of Elasticity

Material	E
Steel	30×10^6 psi
Cast iron, gray	15×10^6
Cast iron, malleable	25×10^6
Wrought iron	28×10^6
Brass	15×10^6
Bronze	12×10^6
Copper	16×10^6
Aluminum	10.3×10^6
Magnesium	6.5×10^6