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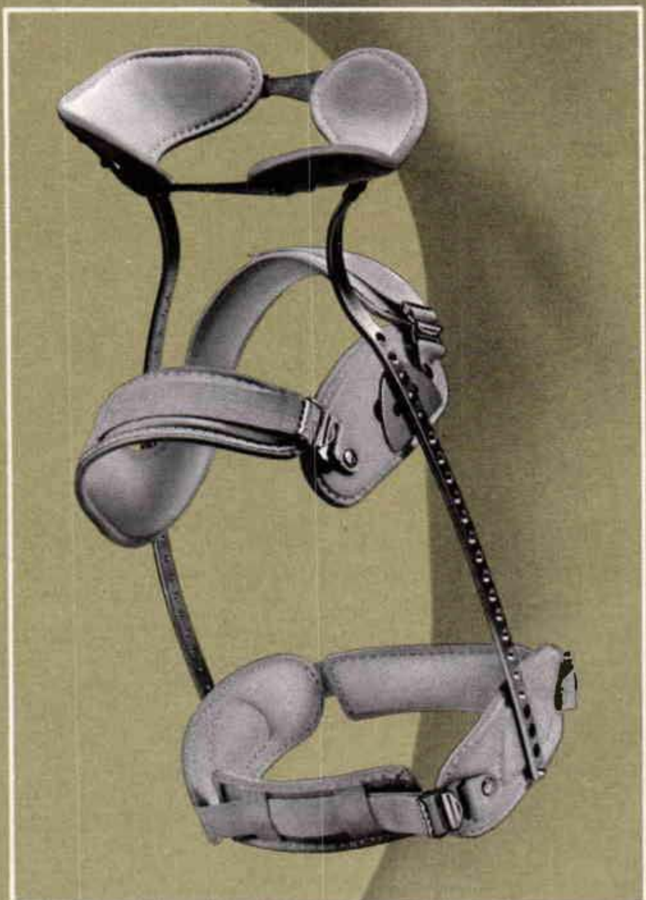

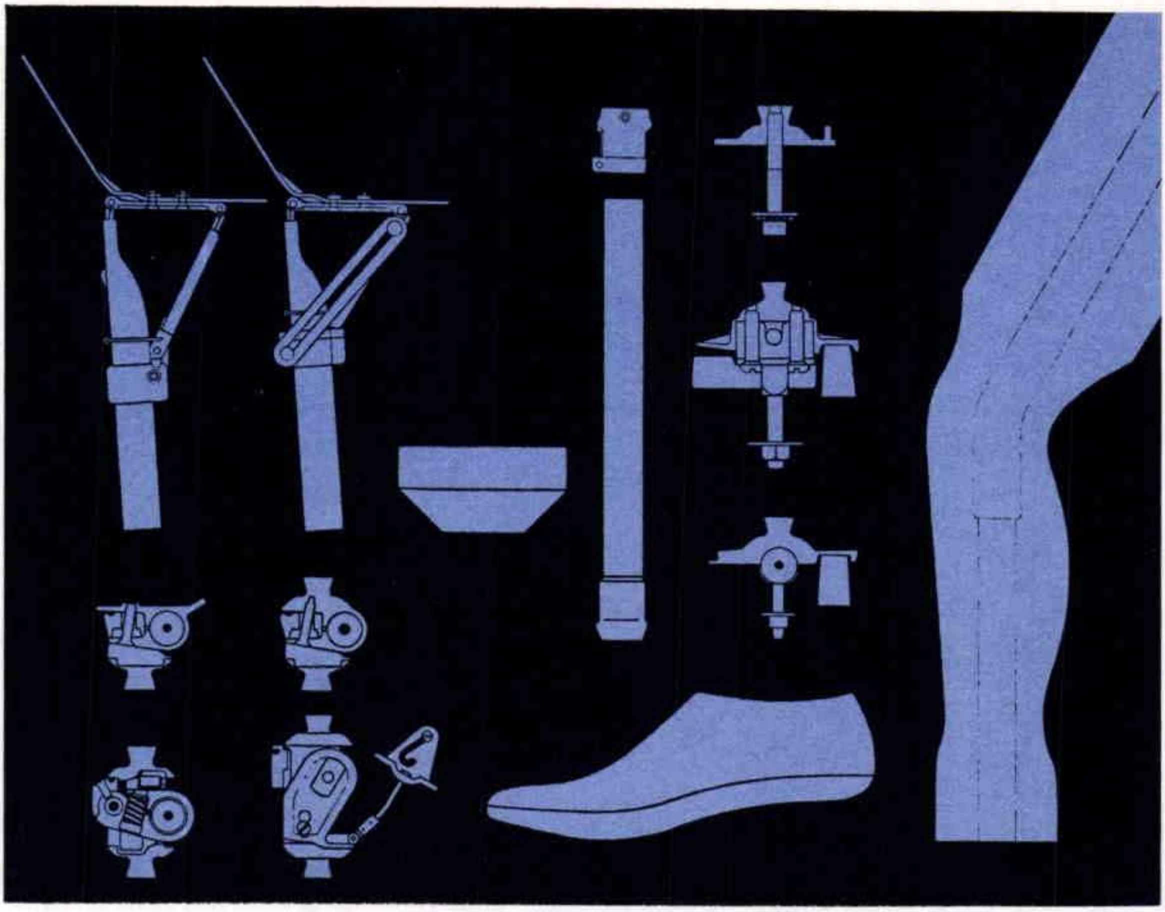

THE MARK OF EXCELLENCE

A few years ago, certification in orthotics and prosthetics was called the "Mark of Merit". Certification today is still a "mark of merit", but the "mark of excellence" is what goes beyond. Excel, according to the dictionary means "be better than; do better than". The commitment necessary to excel in orthotics and prosthetics today is total. That commitment covers many areas such as patient management, new technology, continuing education, and business administration, to mention a few.

To be admired for one's professional efforts, a person must excel. A person who is admired by the medical community he serves did not get that admiration by being status quo. He earned it by excellence and by paying the price. The price comes high, and is not monetary only, but consists of self sacrifice, discipline, and lots of hard work. Everyone has the right to excel, but every right has an equal responsibility that cannot be forgotten or put in a second rate position.

Certification, then, is the beginning of the achievement of the "mark of excellence". We have today two tools to help us achieve excellence. Both are pledged to the betterment of our profession. Both have distinct functions and areas in which they should be used. How we use these tools is up to us for they are ours. I am, of course, referring to AOPA and AAOP. Membership in either one or both organizations does not bestow on one the "mark of excellence", but can be of invaluable help. We have to use the tools to the best advantage possible. We must participate in order to use these tools to their fullest extent.

This "mark of excellence" has no formal diploma or certificate or other form of recognition that can be displayed on the office wall. We all know that it is real and that it can be worth our efforts. To achieve it will take all we can give and then some. It sounds simple: "be better than; do better than". Easier said than done? Yes. Worth the effort? That is up to us!

Dan Rowe, C.P.O.

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Orthotics and Prosthetics, Vol. 32, No. 2, pp. 3-11, June 1978

A PROSTHETIC AND ORTHOTIC MEASURING TABLE

J.A.E. Gleave, FIBST, FISPO, AMBIM[1]

In recent years a number of devices and procedures have been introduced which facilitate the taking of measurements or plaster moulds of various types of deformity. Indeed it is seen that several differing types of apparatus have been developed to take an impression of only one type of deformity.

The fact that many of these designs or procedures leave something to be desired in terms of accuracy or application is perhaps a reflection of an inadequate appreciation of the ergonomics involved, and a seeming preoccupation with the deformity presented, rather than the patient as an entity.

No criticism is implied, but while it is agreed that measuring equipment currently available can produce reasonably accurate dimensions, the end product often falls short of an ideal; and it is suggested that the basic problem which arises is not so much the nature of the design or procedure being used, but to three interrelated factors.

1. The alignment of the body at the time the procedure is being carried out.
2. The relationship of the body to the device being used.
3. The contractile state of the musculature of the body part during the given procedure.

The provision of an appliance is an aspect of treatment, the objective of which is maximum restoration of function, and this is, in part, relative to the alignment of the human and mechanical components. At present this is achieved by the rather uneconomical procedures of static and dynamic alignment. It is suggested that these could be simplified greatly if it were possible to determine relative body alignment at the outset. Equally the relationship of the measuring device to the body is significant in that unless the alignment of the two coincides any measurement or plaster mould being taken will not be accurate.

The part of the appliance in contact with the body must be shaped so as to withstand the forces which will be exerted at the interface without causing discomfort or impeding circulation; thus the disposition of soft tissure at the time of measurement or plaster mould is also significant, since it must affect the amount of modifications to be done to the plaster cast.

Muscle activity will be discussed in a later work. Suffice it to say here that any change in cross-sectional area can be used to exert force, but, if this potential force is to be controlled it is essential that the muscle groups be relaxed at the time the plaster mould or measurement is being taken.

It will be apparent that unless the patient is comfortable and secure, it will be impossible to achieve relaxation and thereby disposition of soft tissue. Under these circumstances it may be proposed that there are given criteria which must be met in order to achieve an accurate representation of the part:

a) There is control over the position of the body and its relationship to the device being used to take the measurement.
b) It should be possible to determine the static alignment of the appliance during the procedure.
c) That any active musculature should be relaxed while the procedure is being undertaken.
d) There is control over the disposition of soft tissue.
e) The procedure for measuring any deformity should not be fatiguing for either the patient or the person taking the cast.

The immediate implication in meeting any of these criteria is stability, and in this state it should be possible to control the position of the body in its relationship to the device being used and that of the person taking the cast. Since these problems are present with all levels of deformity it seemed appropriate to design a basic device which would meet the criteria mentioned, with the possibility of attaching either newly designed measuring or moulding equipment, or modifications of existing ones.

The main problem is one of control of body position and its center of gravity; it is apparent that with the disabled person this cannot be achieved with the patient sitting or standing unsupported, and obviously the position of greatest stability and relaxation of muscle is with the patient lying horizontal on a bench or table.

However, if it is necessary to cater to a variety of deformities and determine static alignment, a horizontal position provided by a table, although meeting some of the criteria, does not meet all; unless it were possible to control the angle of the table top.

It is this concept which led to the development of the equipment described here, which is based upon the assumption that with ade-

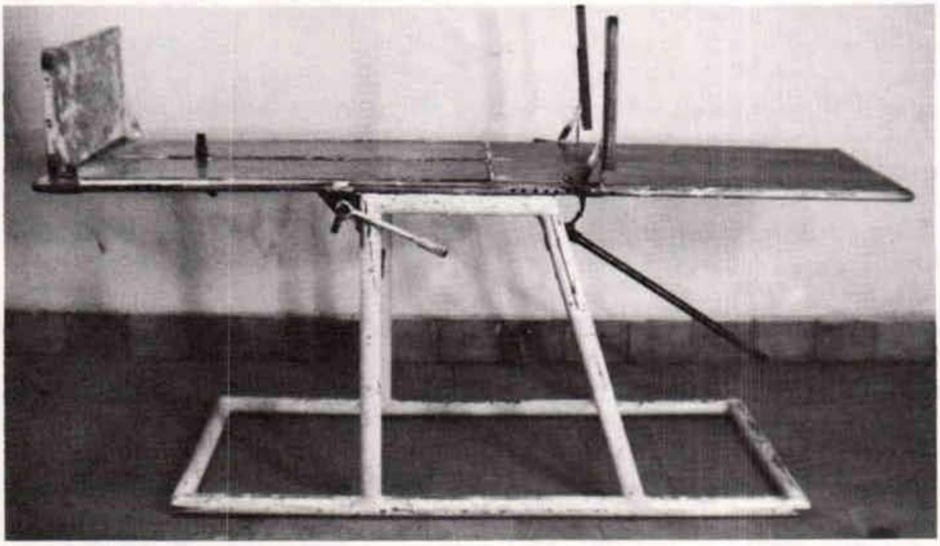

Fig. 1. The basic table shown in the horizontal position.

quate stabilization of the body on the surface of the table, it should be possible to tilt it and the patient through a range of 90 deg., thus meeting the criteria for a variety of deformities.

The Basic Table

The design (Figs. 1 & 2) is similar to that of a tilting table used for treatment in physiotherapy departments; indeed, some of the existing models could be altered for use in the present context except that in order to achieve sufficient rigidity it is necessary to fit locking levers to the table hinges and to redesign the table top.

The design consists of two frames. The lower, or base, frame is designed so that the table will not overturn when the upper frame is brought from one position to another. The uprights have a cross-bar that contributes to the rigidity and provides a support for an additional locking mechanism. A geared mechanism for raising and lowering the table top was considered, but not included in the present design for reasons of economy. Moreover, the lever arm formed by the length of the table from its head to the hinges is long enough to enable the upper frame to be raised or lowered with comparative ease.

The upper frame has a cross-bar to which is fitted a round longitudinal slide that receives the lower-limb adaptor (Fig. 3).

Fitted to the frame are two arm rests, on which the patient may support himself, and a footrest. The table top consists of four boards any of which may be removed for convenient working during a given procedure.

The design of the table lends itself to a variety of attachments which are described here briefly, although it will be seen that there is scope for further design activity and that with minor modification some existing equipment could be used.

The Symes/knee-disarticulation attachment (Figs. 4, 5, and 6) consists of a board 18cm x 16cm the upper surface of which is padded with microcellular rubber and cov-

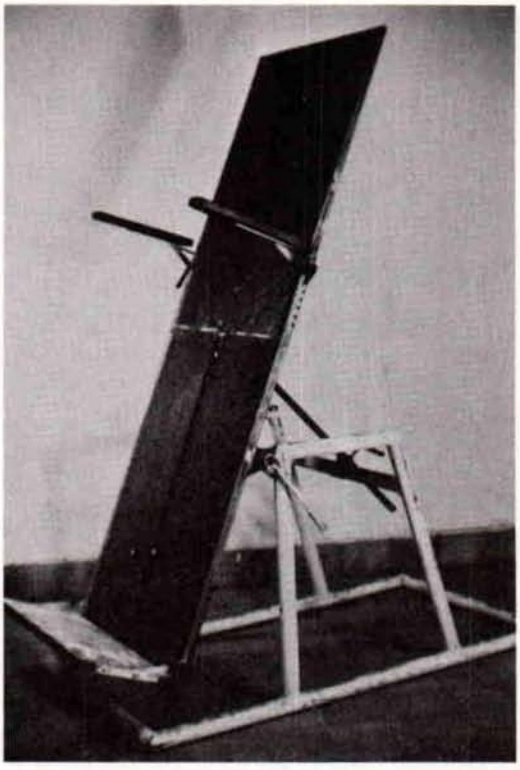

Fig. 2. The basic table shown in the vertical position.

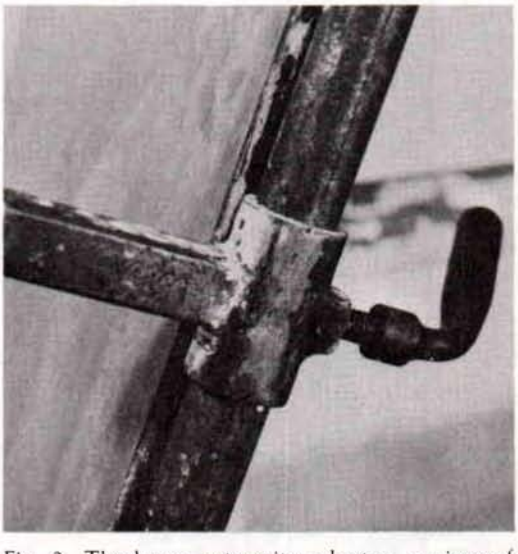

Fig. 3. The lower extremity adaptor, a piece of square bar welded to a slide which allows movement up or down the frame as well as rotating left or right.

Fig. 4. The Symes/knee disarticulation board clamped to the adaptor and set for a right Symes amputation.

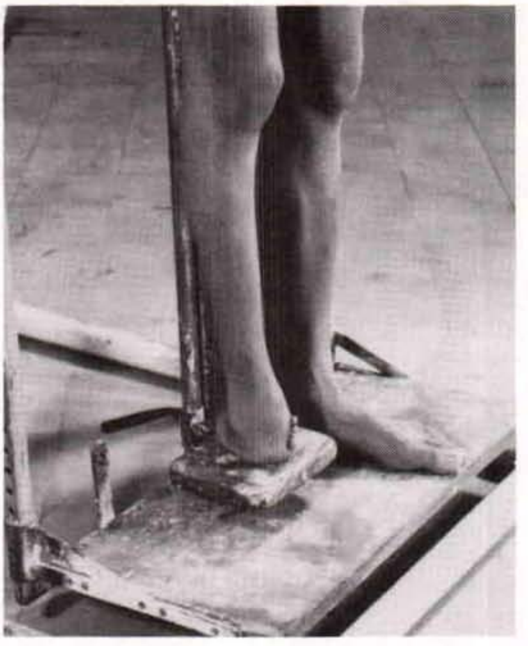

Fig. 5. Patient in position ready for the application of plaster: note the right board has been removed to facilitate working.

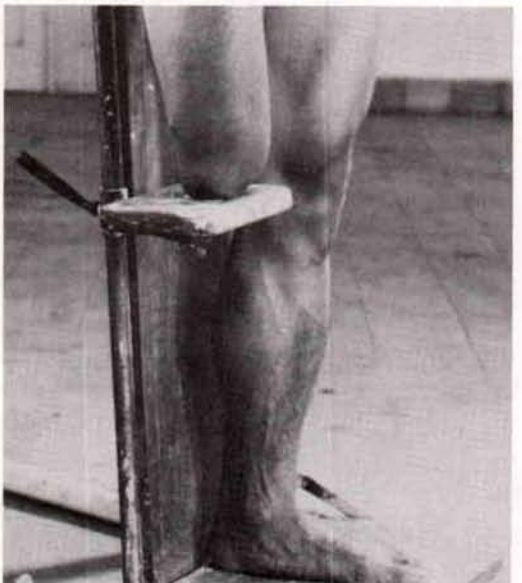

Fig. 6. Knee disarticulation patient positioned ready for the moulding procedure.

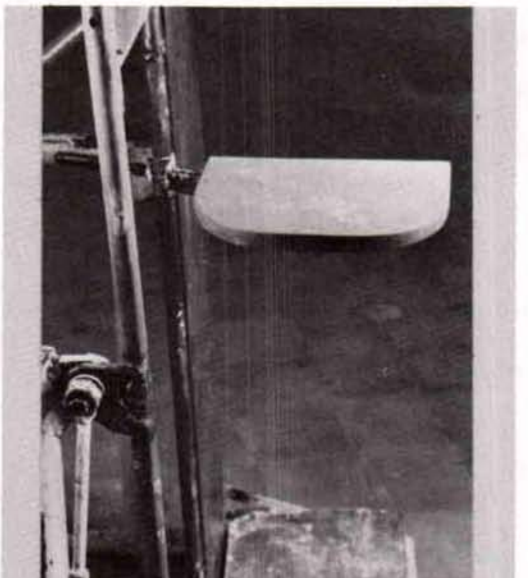

Fig. 7. Hip disarticulation board in position. This is attached to the frame with the same extension arm as that used for the Symes/knee disarticulation board.

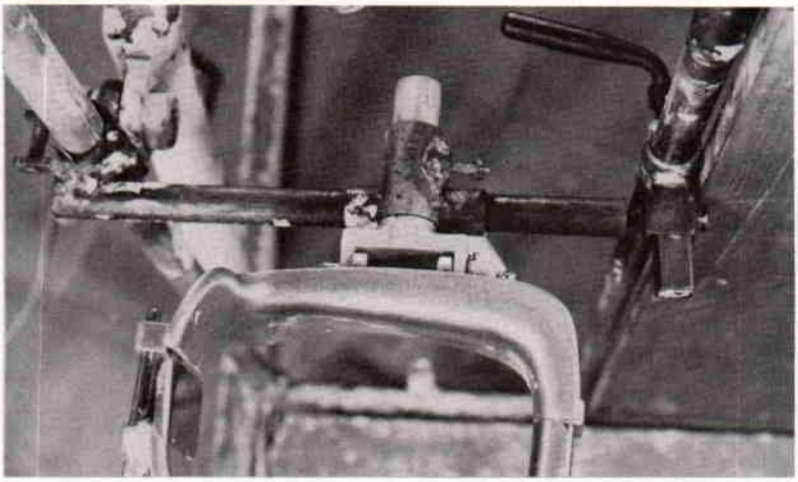

Fig. 8. The above knee attachment with the combination or horizontal vertical and A.P. slides it is possible to adjust the position of the socket-brim as well as control flexion extension.

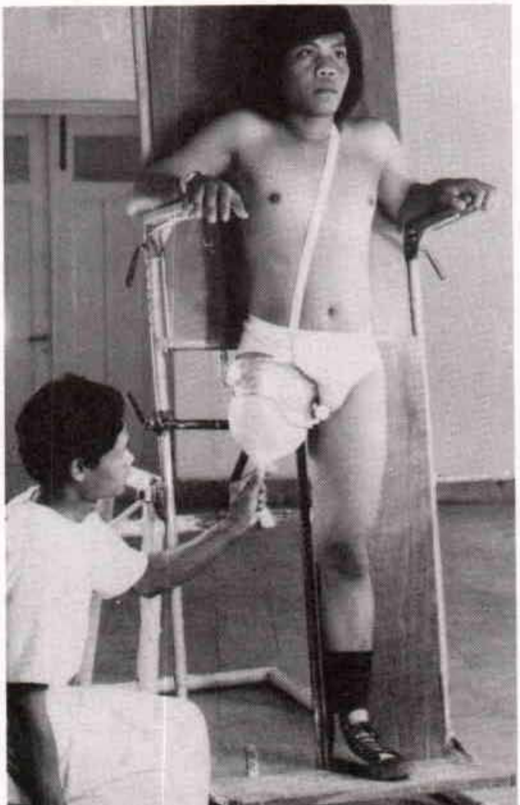

Fig. 9. Patient in position, plaster mould being taken.

ered with vinyl sheet. It is fixed to an extension arm 16cm long which in turn clamps on to the lower-limb adaptor. The point of fixation of the board to the arm is by a pivot and may be locked by a wing nut. The reason for this is that since the lower-limb adaptor rotates about the longitudinal slide the board may be used for either left or right amputations.

The hip-disarticulation attachment (Fig. 7), is also a board with similar padding but different shape. By removing the Symes/ knee-disarticulation board from the extension arm it is possible to use this to support the hip disarticulation board. However, because of the shape and positioning of the board for either left or right it is not possible to have one central pivot, and two are used, one for left, one for right amputations.

The above-knee amputation attachment consists of a horizontal slide which clamps to the lower-limb adaptor and the side of the frame (Figs. 8 and 9). The moulding device, in this instance a Berkeley Brim, is fitted to a locking slide and may be moved medially or laterally; the height may be adjusted by moving the assembly up or down the frame. The horizontal slide may be used for left or

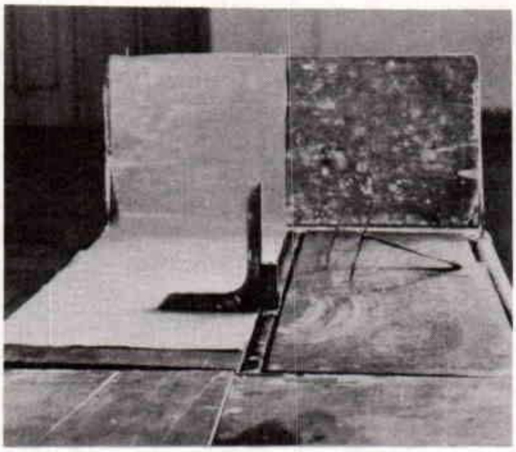

Fig. 10. The right angled pointer used to locate the perineum. Note that the tracing paper extends to the foot board making it possible to determine limb profile tibial torsion, ankle position, and angle of the foot at the one time.

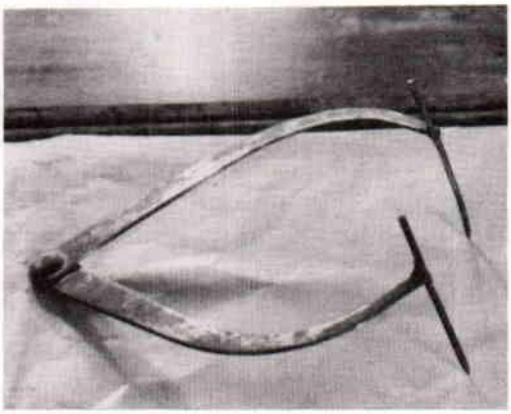

Fig. 11. The dividers used to obtain M.L. dimensions, tibial torsion, and ankle joint position.

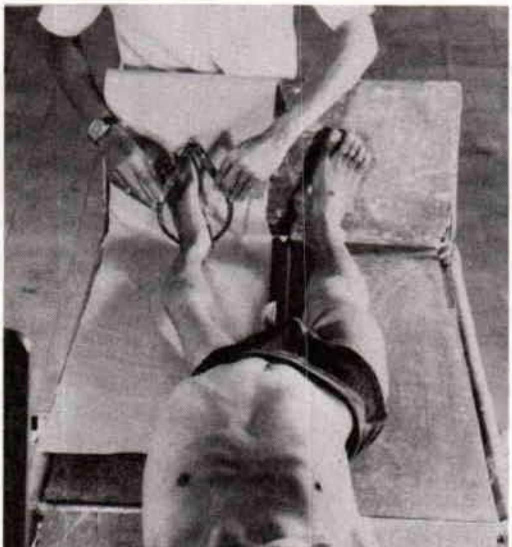

Fig. 12. Patient in position on the table, note the centre line of the table establishing a point of reference for the future alignment of the brace.

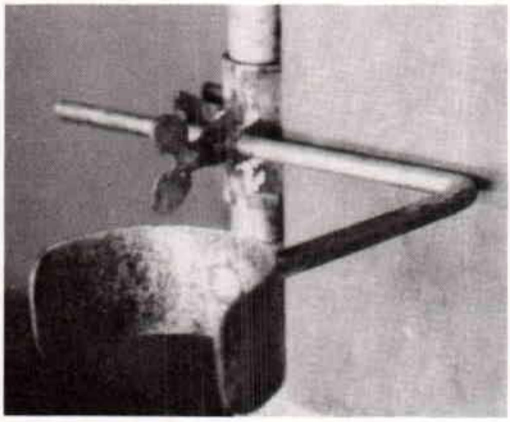

Fig. 13. The below elbow stabilizer to hold the arm in the correct position while moulding.

right amputations by simply removing the central support and reversing the slide.

The lower-limb orthotics measuring device (Figs. 10, 11, and 12), permits measure to be taken either in the horizontal position or at an angle of 45 deg., the foot board and center of the table being the basic reference points.

A right-angle pointer is fitted to the lower-limb adaptor and is used to locate the height of the perineum; and a pair of calipers or dividers are used to measure the medio-lateral dimensions of the knee and ankle. It

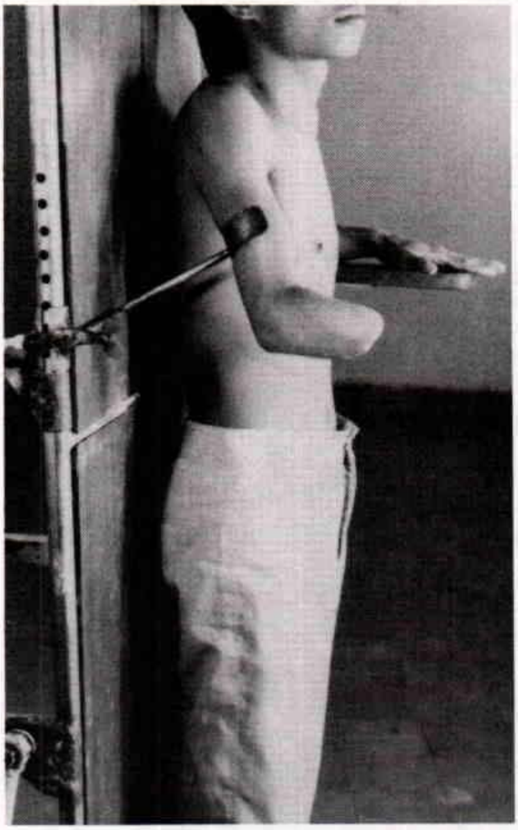

Fig. 14. Patient in position ready for moulding procedure to begin.

should be noted that the measuring paper in this procedure extends under the foot, thus giving the possibility of limb profile, tibial torsion, when present, and the foot angle and position of the malleoli.

The below-elbow arm stabilizer (Figs. 13 and 14) is simply a padded cuff fixed to an adjustable support and is used to steady the upper arm during the application of plaster. Its particular advantage is not so much in the molding procedure but rather in the fact that since the table gives the frame of reference it is easy to ensure that the forearm is held in the correct position while the mould is being taken.

The above-elbow device (Figs. 15 and 16) uses the same clamp as is used in the below-elbow case, but has a former which fits into the axilla allowing the orthotist/prosthetist to use both hands to control the shape of the plaster and the position of the stump.

The spine and trunk unit (Fig. 17) is a suspension apparatus that is clamped to the hood of the frame for use when measuring under traction. It will be apparent, however, that by removing and repositioning some of the boards, moulds and measurements may

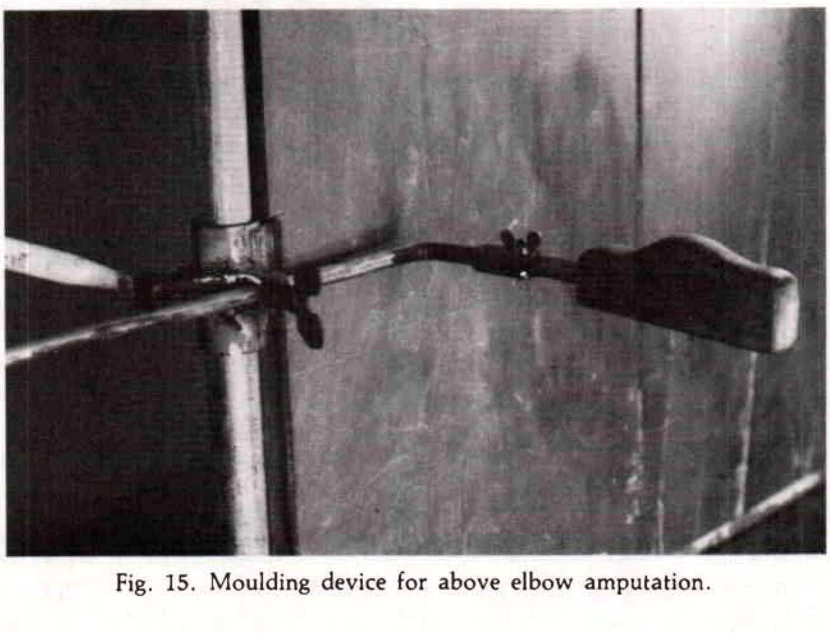

Fig. 15. Moulding device for above elbow amputation.

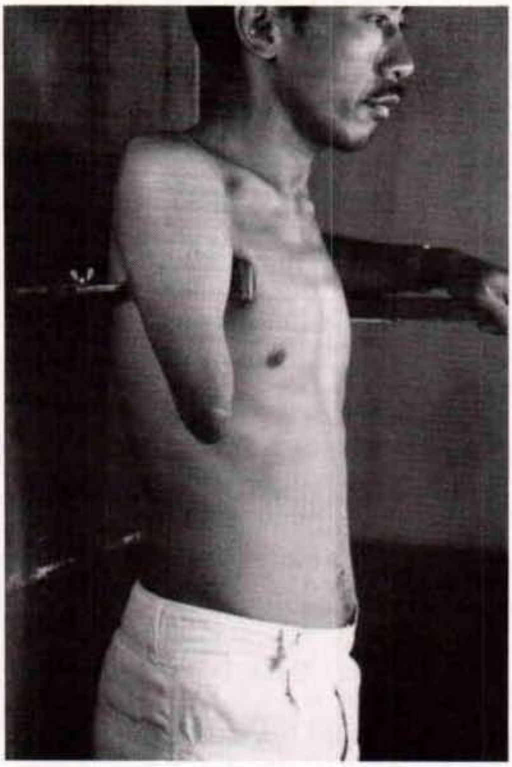

Fig. 16. Patient in position ready for the application of stockinet and plaster.

Fig. 17. Patient in suspension ready for the application of plaster bandages. With the use of supports it is possible to have part or all of the procedure in the horizontal position.

be taken with the table in a horizontal position.

Of the devices discussed above the reader will have noted there is none described for that perennial subject, the below-knee amputee. Efforts have been made to devise a stabilizing bar to hold the stump in the correct position while taking the plaster mould, but this has so far not proven to be more efficient than procedures already available.

Discussion

At first sight the concept of a basic device upon which a range of fitments may be used may seem complex and perhaps costly, and it is as well to examine briefly these factors rather than leave the subject as an apparent technical tour de force.

At the time of writing, three tables as described have been in use for periods of 2-3 years with only minor changes in the original design. The concept of using one basic device to cater for a variety of deformities has been readily accepted and it seems appropriate to make a preliminary assessment of the design.

Patient Acceptance: A series of questions and observations were made of both new and experienced patients; the consensus was that they all felt secure, were able to relax and cooperate fully during the procedure irrespective of the deformity being catered for.

Professional Staff: The immediate feature which became apparent was that after instruction in the various procedures, professional staff became more aware of the relationship of the deformity to the patient. Since there is a frame of reference, body alignment can be readily observed, corrected, or taken into account, for subsequent static alignment.

Of particular interest was the fact that, with new students it was possible to demonstrate spatial relationships which are otherwise abstract until the often fatiguing process of dynamic alignment begins. Finally the design appears as such that all procedures can be carried out with a minimum of fatigue of the clinician.

Technical and Production: Comparisons were made between conventional procedures and those using the table. The space required is less than usually acceptable since the table can double for an examination couch. In most plaster rooms a couch is necessary. Preparation and cleaning times are the same or slightly less since all components are located in one place. Time required per procedure is approximately the same or slightly less; however the rejection rate for given moulds was markedly less, and cast modifications required much less time. Measurements could be taken at given angles, as the table provided reference, and were therefore more accurate. This also applied to static and dynamic alignment.

These factors taken into account indicate that the initial cost of producing the table is soon offset by increased efficiency in the department.

As presented here the table has been deliberately designed to be as simple as possible commensurate with efficiency and it will be apparent that further improvement can be made. This is an aspect of progress.

Footnotes

[1]P.O. Box 5583, Riyad, Saudi Arabia.

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PROCEDURES FOR OBTAINING CASTS FOR ANKLE-FOOT ORTHOSES

J.H. Tyo, C.O.[1]
R.D. Koch, C.O.[2]

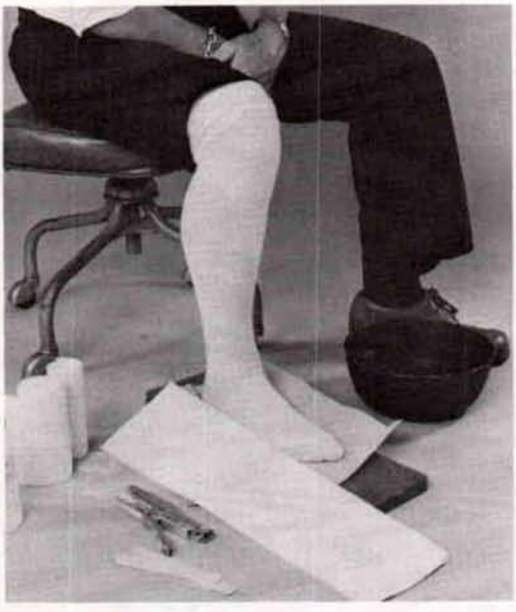
Over the past several years the use of molded thermoplastic ankle-foot orthoses has become an accepted tool for orthotic management. A number of articles and publications have dealt with the types of plastics used and the various molding techniques. Some attention has been paid to trimlines and how they affect the performance of the orthoses-patient combination.

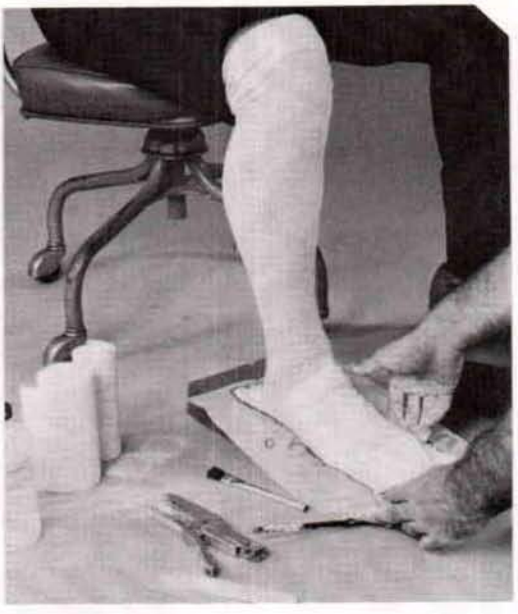
In the majority of articles presented to date, however, little has been done to update casting procedures. While wrap casting can prove adequate if the user is skilled, it is quite easy to distort soft tissue and lose sight of the landmarks of the foot and shank.

The principle problems with the wrap cast technique are the tendency of the soft tissue to assume a cylindrical shape under circumferential pressure and the difficulty of removing the mold from the patient without distortion. A wrap cast also increases pressure over bony prominences and can create hollows or depressions between these prominences.

To eliminate these problems a technique has been developed that allows control of the foot and shank at different intervals of the procedure, allows for an accurate reproduction of the extremity, and is easily removed with a minimum of distortion.

Materials necessary for this procedure are a foot board or shaped foam block, stockinet sewn closed at one end, mineral oil or Vaseline, two rolls of six-inch wide extra fast plaster-of-Paris bandages, tongue depressors, cast pencil, and bandage scissors.



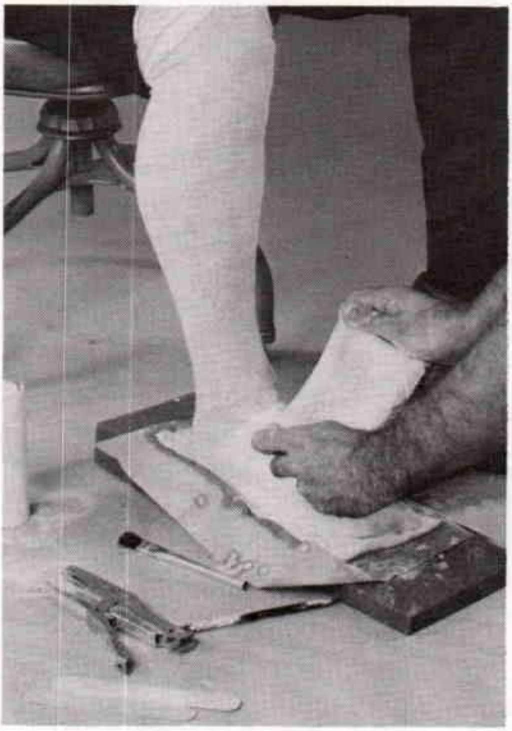


After examination, the patient's limb is covered with stockinet and necessary landmarks are indicated with a cast pencil. To facilitate this portion of the procedure the stockinet may be wetted prior to its application.

Although the stockinet will usually stay in place by itself, the patient is asked to hold the proximal edge of the stockinet since this reduces tendency for the patient to make unwanted movements.

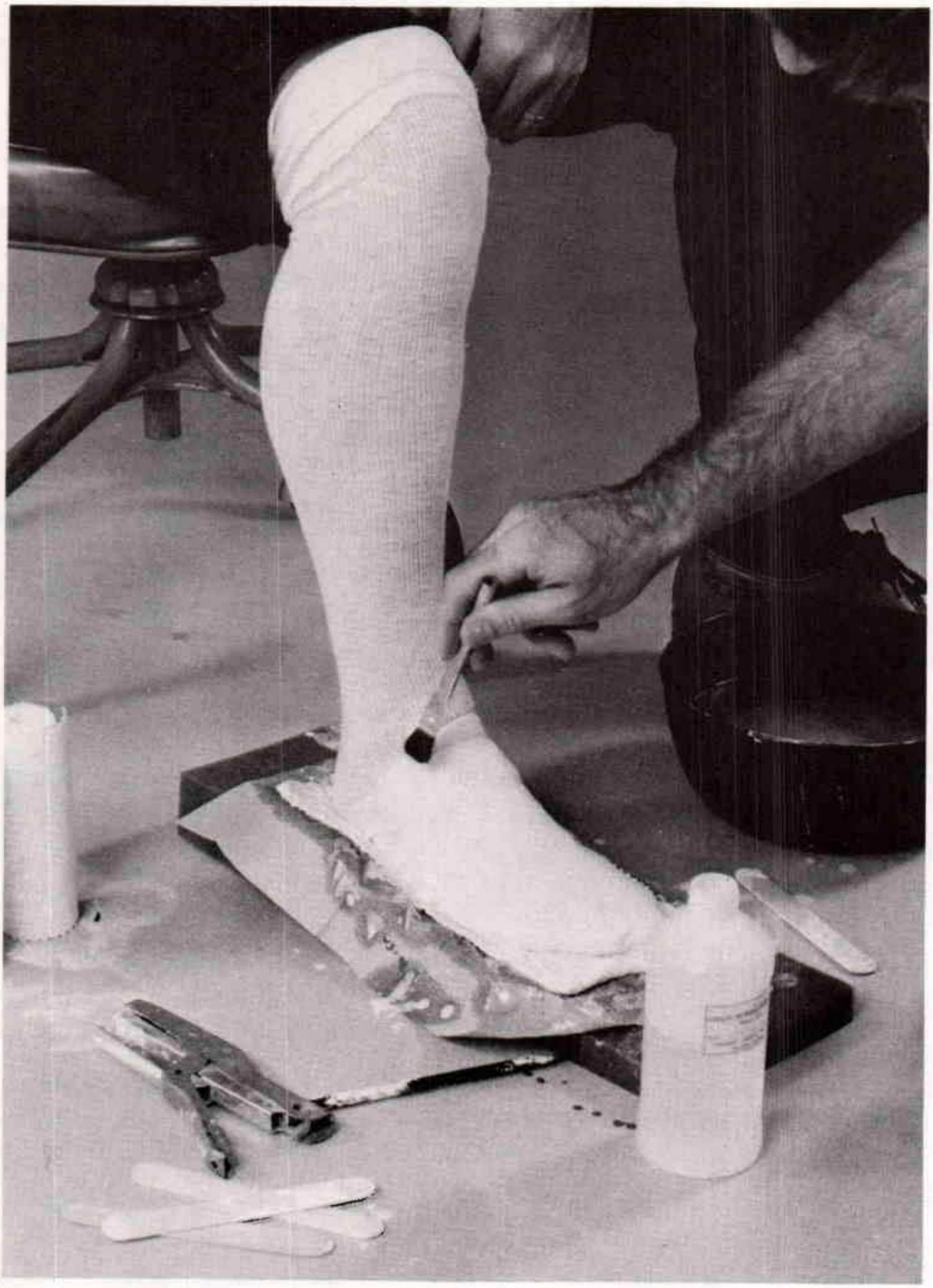
The position of the foot and lower leg is explained to the patient. During this explanation the foot is physically located in the proper position on the foot block. While the patient is in the proper position the first section of the plaster is measured and cut. This is a four layer splint twice the length of the foot from heel to toe.

The patient's foot is removed from the cast block, the splint is soaked and located to the position formerly occupied by the foot. At this time care should be taken to remove any wrinkles. The excess length is allowed to fall anterior to the toes. As soon as the foot is relocated on the cast block the excess plaster bandage is placed over the dorsum of the foot and smoothed out.

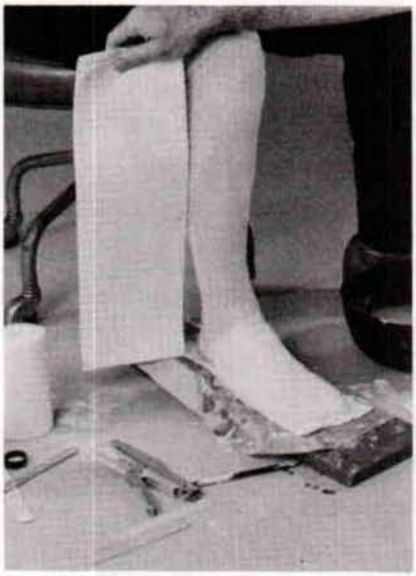


An additional splint is located over the dorsum of the foot to increase the proximal bulk of the cast and to make sure the lateral and medial edges of the foot are covered.

At this time the tongue depressor is used to push the bandage tightly around the foot. While this portion of the mold is drying the foot is held in a corrected position. Whenever possible this is done without putting pressure of the mold itself thus avoiding distortion and subsequent modification problems.

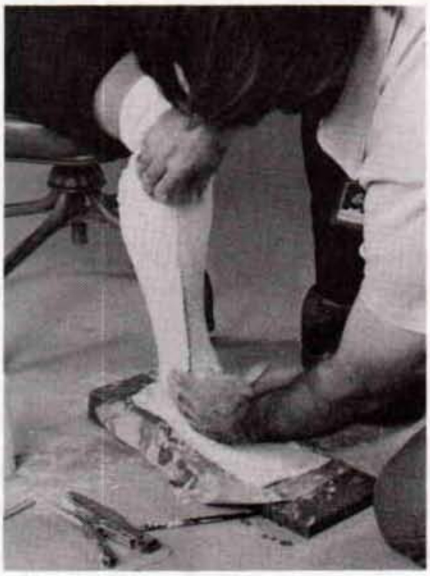


After this portion of the mold is dry it is coated with mineral oil or Vaseline over the dorsum and posterior aspects.



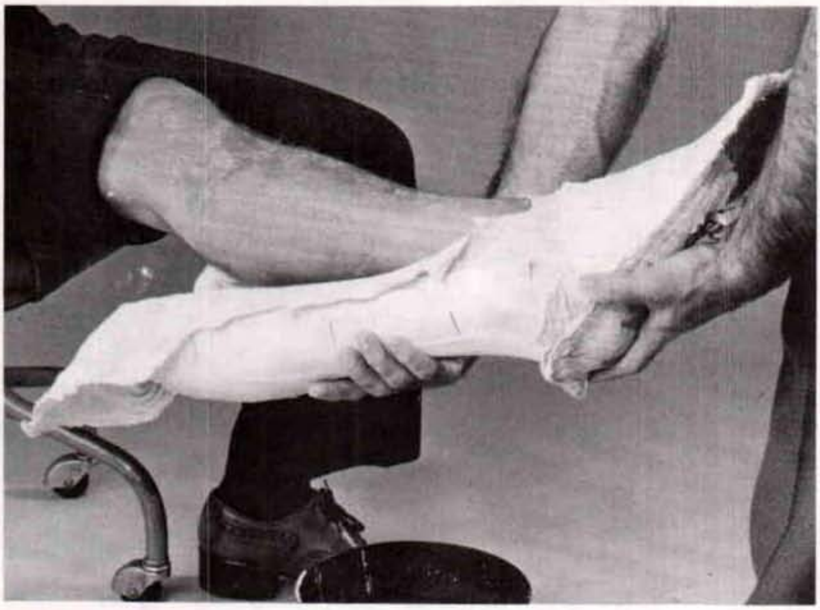
The second splint is located over the posterior shank. The length is determined by adding approximately three inches to the length of the leg from the posterior crease of the knee to the base of the calcaneous. The splint is three layers thick.

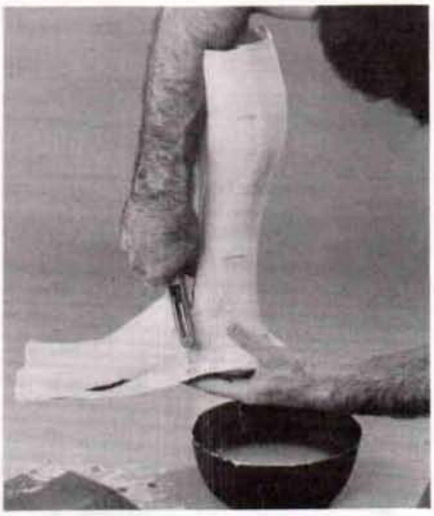
This part of the splint is started at its most proximal aspect and smoothed distally. The bandage should be as wet as possible to insure adhesion to the stockinet.



Additional splints are placed as necessary to cover the area of the extremity that will be covered with the AFO.

After the entire cast has hardened it is removed in sections. The new anterior portion of the stockinet is cut and pulled free of the plaster.

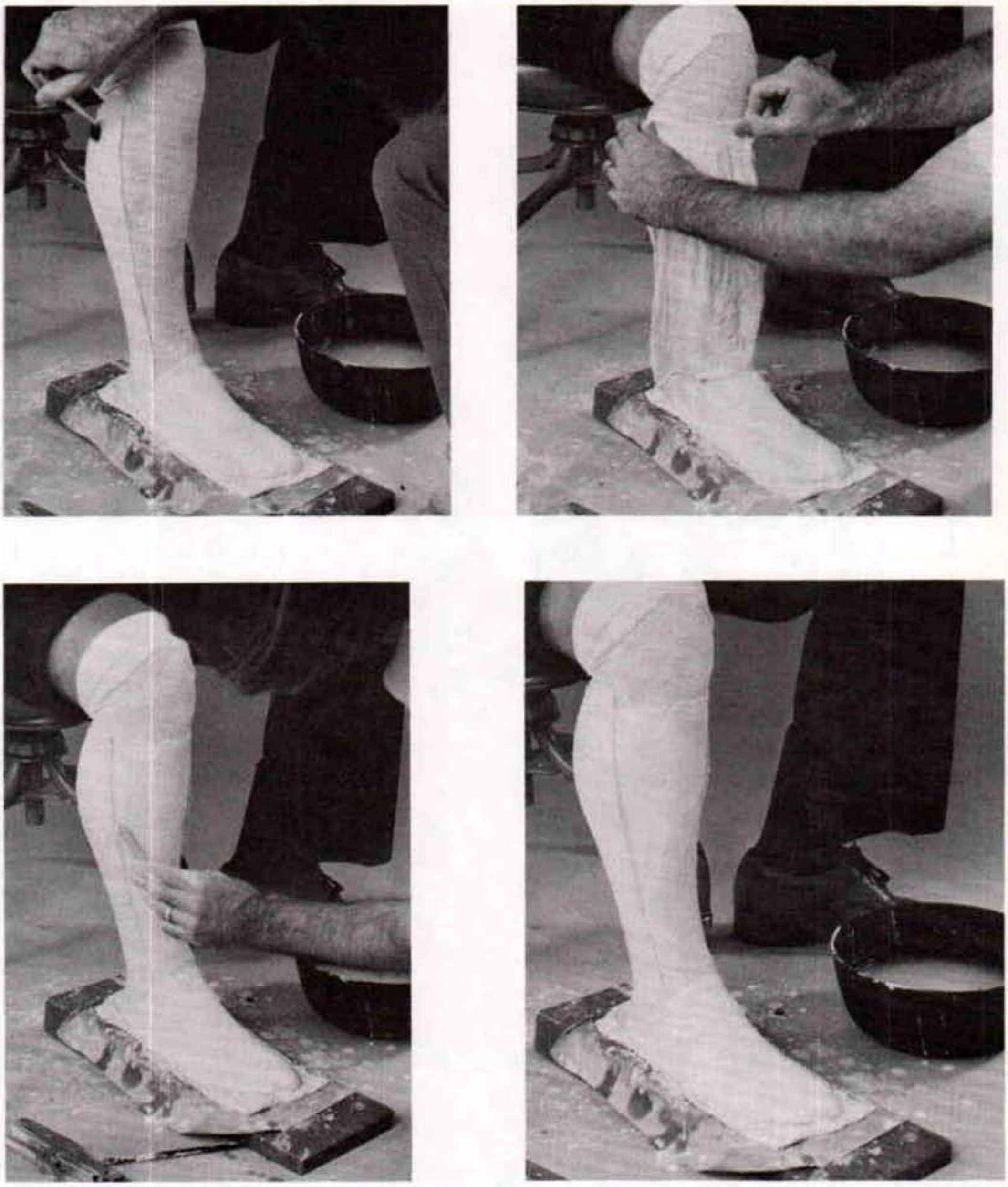




The entire foot is lifted free of the foot block and the knee is extended. The foot is manually plantarflexed and dorsiflexed. This should be done gently to avoid injuries to both the patient and the cast.

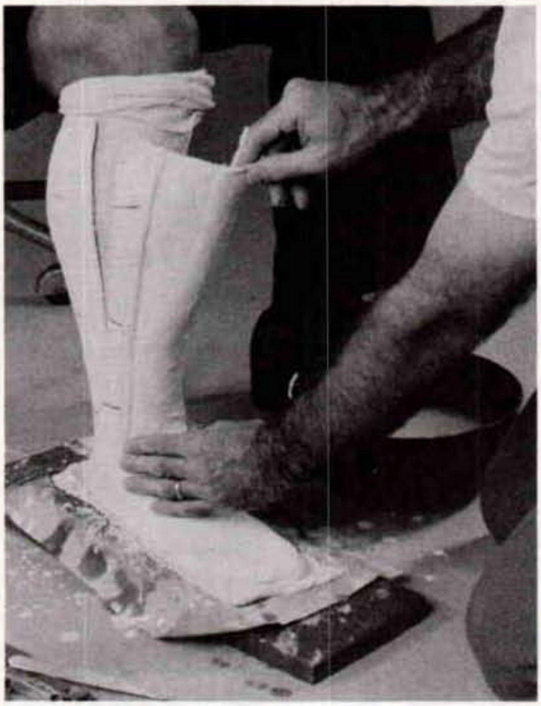
At this time the shank portion of the cast can be pulled away, the stockinet is cut down to the posterior base of the calcaneous and the foot portion can be removed with gentle traction.

The stockinet is pulled free of the plaster, the cast is reassembled, stapled and allowed to dry.



In cases where a mold of the entire circumference of the extremity is necessary, the procedure can be extended as shown above and in the rest of the photographs in this article. As indicated previously the edges of the already dry portion of the mold are lubricated and splints are applied. This last section is allowed to dry and can be removed easily by just pulling upward on the proximal anterior stockinet. The rest of the procedure remains unaltered.

Application of a splint to the anterior portion of the shank.
Establishing total contact.
Completed cast.



Removal of anterior section.

The anterior section is replaced and
ready for pouring the positive model.

Footnotes

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Orthotics and Prosthetics, Vol. 32, No. 2, pp. 21-24, June 1978

THE FUNCTIONAL RATCHET ORTHOTIC SYSTEM

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Because of the anatomical structure of the cervical spine, a C-5 on C-6 vertebral fracture with resulting C-6 neurological deficit is commonly observed. The residual muscle functions for this level, as described by McKenzie (1), are the shoulder flexors and abductors, scapular muscles, and elbow flexors.

Owing to improved medical care, the survival rate of individuals acquiring a high-level cervical lesion has increased over the past few years. For example, at the Rancho Los Amigos Hospital Spinal Injuries Center, a total of 985 quadriplegic patients were admitted from 1964 to 1974. Of these, 32 percent, or 314 patients, had C-5 on C-6 lesions.

Orthotic management of the involved upper-limb patient has historically presented the rehabilitation team with an enigma, especially in the case of traumatic quadriplegia. Level of independence and activity are closely related to level of lesion. Restoration of function involves many factors, including the application, acceptance, and utilization of orthotic systems.

When the lesion is complete, the sensory deficit will be complete. Incomplete lesions will display mixed evidence of both motor and sensory loss or function. Individual patient evaluations must be thorough if full advantage is to be taken of the residual functions.

The involvement of the radial wrist extensors is of great importance, for this muscle group can be utilized through the application of a wrist-driven flexor-hinge orthotic system to provide force for prehension.

The individual suffering a fracture of C-4 on C-5 lacks the residual functional musculature to power a wrist-driven orthosis. At this level of neurological deficit, the radial wrist extensors are absent, and the patient is forced to seek additional mechanical assistance.

Previous attempts to utilize external power systems have resulted in non-acceptance for a number of reasons. Both the electric and compressed gas systems previously described require a substantial amount of familiarity and expertise by the orthotist to achieve a functional application for the patients' needs (2) (3). In addition, the gadgetry of the systems required too much effort by the patient, and many were quickly discarded after the patient left the rehabilitation center.

Beard and Long (4) have conducted a 12½-year follow-up on the use of externally powered orthoses. Their findings indicate an overall usage rate of 33 percent, and that "both the poor quality of performance and the small number of activities which can be accomplished, due to limited range of motion and lack of forceful movement, lead to disuse. Without good proximal arm function, the externally powered hand splint is apparently of little value to these patients. The additional time required for application of the entire system is not justified. Poor quality of performance of activities was the reason most frequently cited by patients for disuse of externally-powered orthoses." The primary reasons for discarding the orthosis are:

1. Time required for application.
2. Poor quality of performance of activities.
3. Slow speed of performance of activities.
4. Hindrance during other activities.

To remedy these shortcomings, the ratchet principle first demonstrated at Warm Springs, Georgia (5) was utilized. In the original design, the ratchet system simply allowed the patient to maintain prehension of an object over any given period of time required to complete a task. However, it left much to be desired with respect to adaptability and adjustment.

Through a sequence of clinical applications, the ratchet (Fig. 1) has undergone several modifications. The first major change of the original design involved conversion of the ratchet principle to the existing wrist driven wrist-hand orthosis. This design utilizes opposition of the thumb and the first two fingers. A friction wrist joint was then applied to maintain stability, yet allow some adjustment of flexion and extension in a clinical setting.

The spring-activated ratchet lock in itself (Fig. 2) prohibits release of an object until the user so desires. At this point, pressure applied to the release lever of the ratchet bar frees the object held in grasp. Finger opening is accomplished through the application of a return spring system. It was noted that in a number of cases, objects had a tendency to slip under firm prehension. To remedy this problem, a prehension compression spring

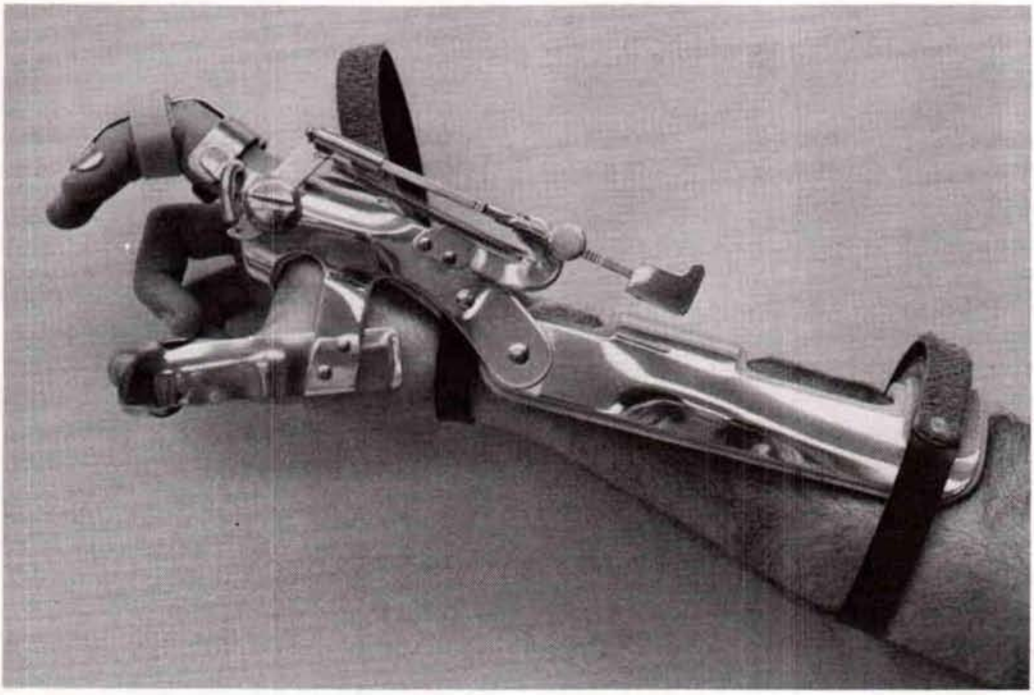

Fig. 1. Ratchet orthosis in MP extension indicating full positional opening of finger pieces.

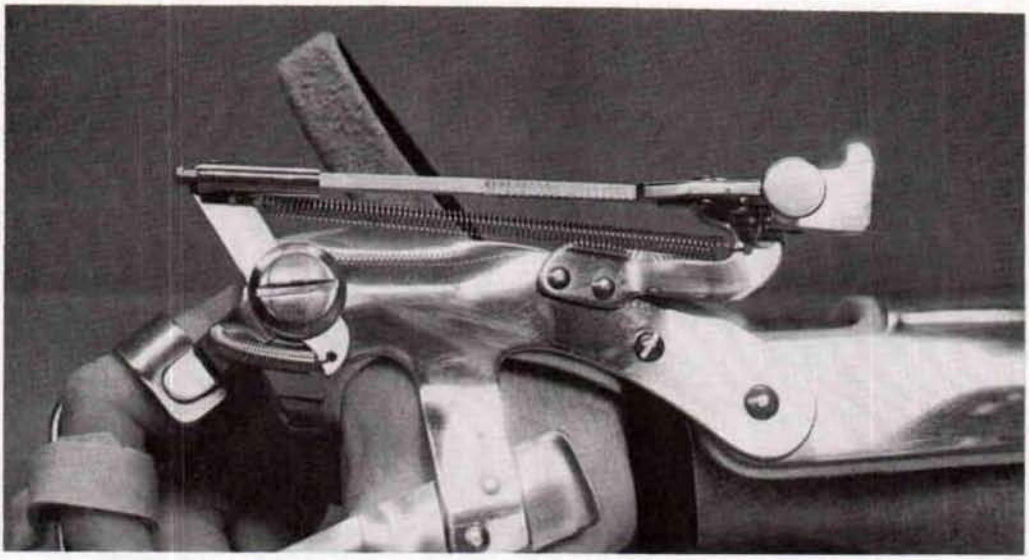

Fig. 2. Ratchet element including compression prehension component, ratchet bar, spring activated lever, and return spring system.

was devised to maintain constant pressure.

That which has been presented to this point provides a general overview to acquaint the orthotist with the ratchet principle. It must also be emphasized that this functional level of quadriplegia requires additional forms of assistive devices to attain proficient levels of activity. The ratchet is one primary component of an entire system which would ordinarily include radial mobile arm supports, a powered wheelchair, lap trays, mouth sticks, special seat cushions, etc. It is beyond the scope of this discussion to detail these items, but the practitioner must be aware if their importance in the care of the quadriplegic. A more detailed discussion of fitting indications for the ratchet is required if the practitioner is to benefit the patient. It must be kept foremost in mind that a high rejection rate of devices is prevalent in this category of patient.

Acceptance depends primarily on a well-defined purpose or rehabilitation goal. Mechanical efficiency and accuracy of fit are demanded of the orthosis. Replacement of lost function can only be achieved by the knowledgeable orthotist with the skill and dexterity to meet the challenge.

The functional ratchet orthotic system is comprised of the following component parts:

1. forearm section with proximal stabilizing strap,
2. palmar section with wrist strap,
3. finger and thumb pieces,
4. the ratchet element,
5. wrist and MP joints.

The purpose of both the forearm and palmar sections is to maintain the hand in a functional position. Careful observation must be made to ensure that the ulnar styloid and MP joints are free of obstruction so as to avoid pressure and restriction of movement.

Mechanical axes of the wrist and MP joints must match the anatomical axes as precisely as possible to avoid unwanted relative motion between the orthosis and hand. Functional opposition can be achieved only if the thumb and first two fingers are aligned properly. Both the finger and thumb pieces must be adjusted to provide direct finger and thumb tip opposition.

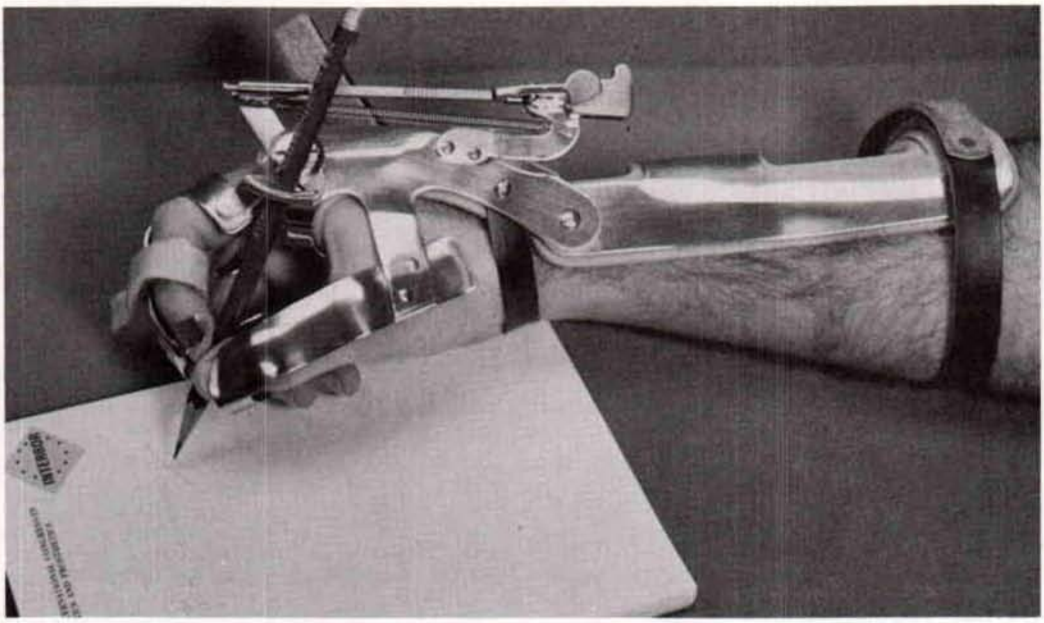

Fig. 3. Adaptive spring allows functional positioning of writing implement.

Fine adjustment on the ratchet bar is accomplished by allowing the spring-activated lever to position itself with the notched increments. To activate the ratchet and close the orthosis, the patient simply presses the finger pieces together against either the chin or the other hand. Opening is achieved by depressing the spring-activated lever and allowing the return spring to facilitate full MP extension.

Tension of the return spring system is easily adjusted by shortening or lengthening the spring. It should be noted that an adaptive spring is provided to permit use of a pen or pencil (Fig. 3). With a certain amount of training and persistence, writing can be accomplished effectively.

Training is an essential segment of the overall rehabilitation for these patients. The authors emphasize that the skills of a qualified therapist be employed to accomplish the training goals.

Although intimate fit of the orthosis is essential, fabrication need not be required of the individual practitioner. Central fabrication of these systems in component parts or completed definitive form have been utilized with a great deal of success, and are now a part of the practitioner's armamentarium.

One point should be reiterated. Because of the precision that is the essence of the hand, the orthotist is presented with his greatest challenge in any attempt toward its restoration.

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TREFOIL ALIGNMENT ADAPTOR FOR THE VACUUM-FORMED BELOW-KNEE SOCKET

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Vacuum-forming techniques have been employed widely in orthotics during the past five years or so, and recently have been applied successfully to the fabrication of ultralight prostheses that consist essentially of an all-polypropylene composition (1).

Since May of 1976, a method of fabrication has been employed at the VAPC which permits inclusion of an alignment device in a below-knee prosthesis that uses a vacuum-

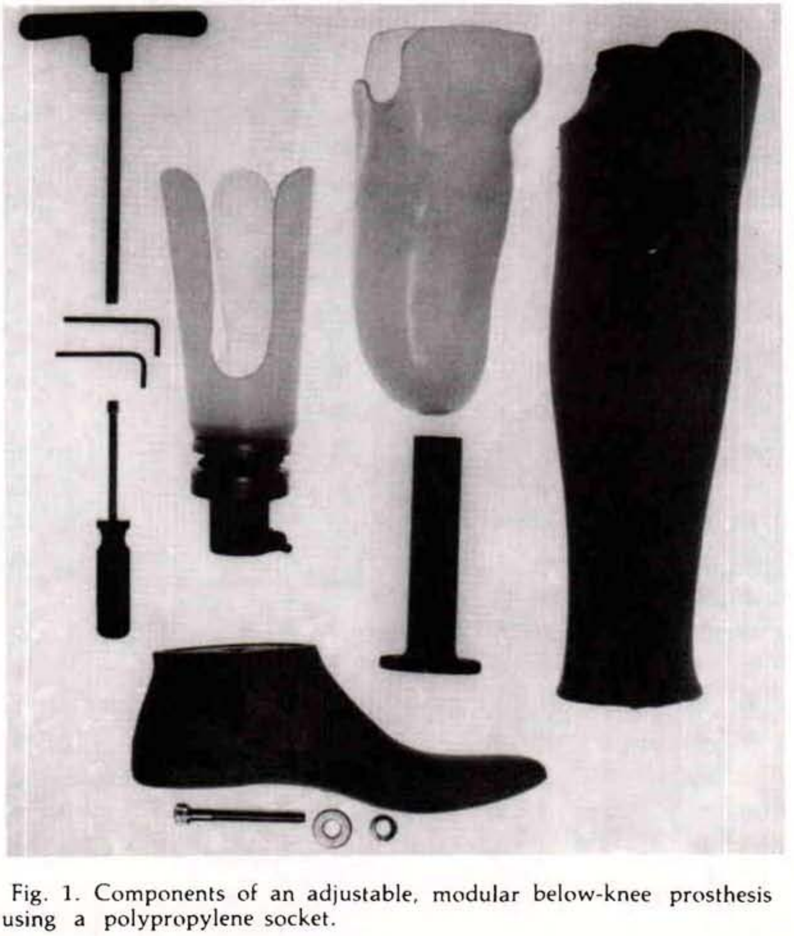

Fig. 1. Components of an adjustable, modular below-knee prosthesis using a polypropylene socket.

formed socket. The essential element in this technique is a trefoil (tulip-shaped) polypropylene adaptor used with a VAPC adjustable shank. The components are illustrated in Figure 1. Up to this time the method described below has been successfully employed for seven patients.

Fabrication Technique

1. A polypropylene socket is vacuum formed over a modified cast. Three-eights of an inch thick polypropylene is used for below-knee stumps up to five inches in length; one-half inch thickness is used for longer stumps.

2. The proper size (small, medium, or large) of a prefabricated trefoil (tulip) polypropylene adaptor is selected.

3. After the trim lines on the socket are identified and the socket is trimmed, it is held in a position of static alignment in the adaptor with tape. Major adjustments can be made in the adaptor and the adjustable shank can be used for minor corrections. When the alignment desired is achieved the socket is welded to the adaptor, the alignment device being retained in the prosthesis. The entire prosthesis is then formed, shaped, provided with a cosmetic cover, and delivered. If subsequent alignment changes should be necessary they can be made readily after removal of the cosmetic cover for access to the alignment device.

Acknowledgement

We wish to thank Dr. Gustav Rubin, Orthopedic Consultant, VAPC, for his cooperation and encouragement in carrying out this project.

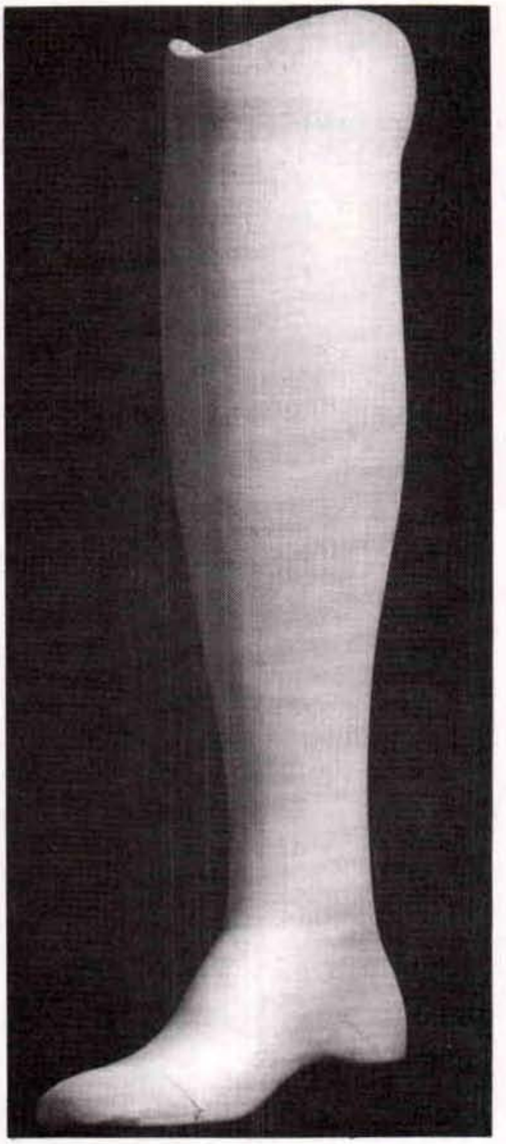

Fig. 2. The finished prosthesis.

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Footnotes

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THE ORTHOTIC MANAGEMENT OF LUMBAR LORDOSIS AND THE RELATIONSHIP TO THE TREATMENT OF THORACO-LUMBAR SCOLIOSIS AND JUVENILE KYPHOSIS

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William Bannell, M.D.[2]

Historically an attempt has been made to reduce the lumbar lordosis somewhat when casting a patient for a Milwaukee CTLSO. This attempt has been made with varying degrees of concern and usually with varying degrees of success.

Recognizing that the orthotic management of the lumbar lordosis is necessary in the treatment of juvenile kyphosis (Scheuermann's disease), and thoraco-lumbar scoliosis with a CTLSO or TLSO, this paper will present the rationale for such management, beginning with the development of the lumbar posture and its relationship to the overall spinal positioning.

Postural Development in the Saggital Plane

Posture can be defined as the relationship of the parts of the body to the line of the center of gravity (4). The sagittal curves of the spine, through development and growth, play an important role in maintaining the proper postural and body relationships to the center of gravity.

Beginning in the uterus the fetus is in the position of flexion with the convex curve of the spine lying against the curve of the uterine wall. Following birth the development of posture is affected by the constant forces exerted by gravity.

The newborn lies either supine or prone and the gravitational force is exerted on a horizontal plane and tends to unroll the primary ventrally convex curve or "coiling" that was assumed in the uterus. (Fig. 1A)

The infant between two and six months begins to lift its head and to sit, causing the development of a cervical compensatory curve. (Fig. 1B)

The nine- to eighteen-month old begins to stand and walk in an upright position and the weight of gravity is exerted in a vertical direction resulting in the development of the lumbar compensatory curve or lordosis. (Fig. 1C)

This stage of development results in four curves in the sagittal plane, a) the thoracic and sacral curves *concave* ventrally (the primary curves because they were present during fetal life), and b) the cervical and lumbar curves *convex* ventrally, the secondary or compensatory curves (developed after birth) (2).

The three curves above the sacrum, the cervical, dorsal, and lumbar, are functional and add to spinal elasticity and strength. The three afford a greater resiliency to the weight forces of head and body, and those exerted by gravity, than would a single curve. The

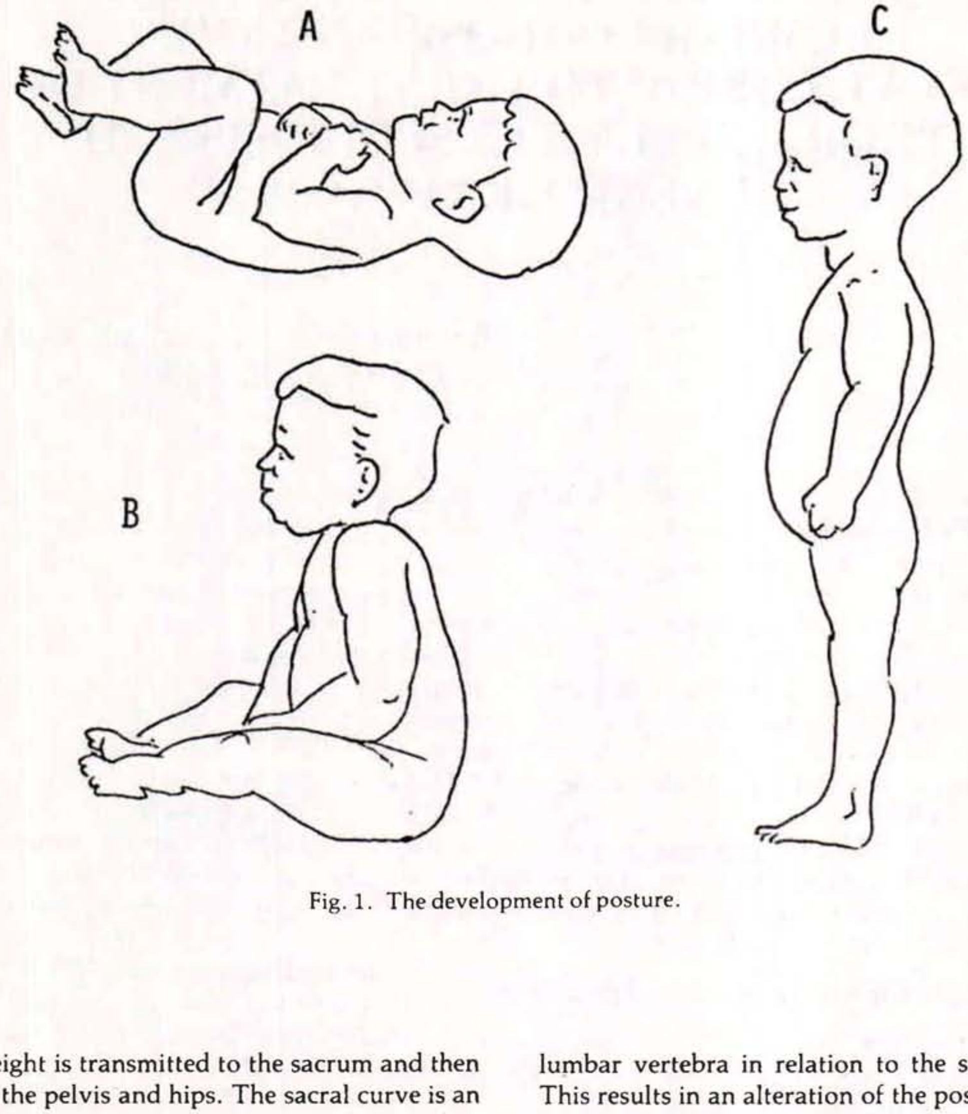

Fig. 1. The development of posture.

weight is transmitted to the sacrum and then to the pelvis and hips. The sacral curve is an adaptation to the inclination of the pelvis and is not a factor of weight transmission.

The Lumbar Lordosis - Pelvic Tilt Relationship

The pelvis is the base upon which the spinal column rests, the lumbo-sacral joint, (lumbar five and sacral one,) and any changes in its inclination will result in a corresponding change in the position of the 5th lumbar vertebra in relation to the sacrum. This results in an alteration of the posture of the entire spine. An increase in the inclination of the pelvis causes any increase in the lumbar curve. A decrease in the inclination causes a decrease in lumbar lordosis.

Inclination of the pelvis is controlled by the muscles of the hip. It is decreased by contraction of the hip extensors, the glutei, hamstrings, and the posterior portion of the hip adductors; and is increased by contraction of the hip flexors, the iliopsoas, rectus femorus, pectinieus, and the anterior position of the hip adductors. The spine is flexed by the iliopsoas and abdominals and is extended by the erector spinae.

The abdominals act synergistically with the glutei, the latter decreasing the pelvic inclination and the former reducing the lumbar lordosis (3) (Fig. 2).

The "Fick" angle of inclination is one measurement used to determine the degree of pelvic tilt. The angle of a line from the foremost portion of the pubic symphysis, to the superior posterior spine of the ilium, measured to the horizontal. The normal male angle of inclination is 50 deg., with a slight increase in the female (Fig. 3).

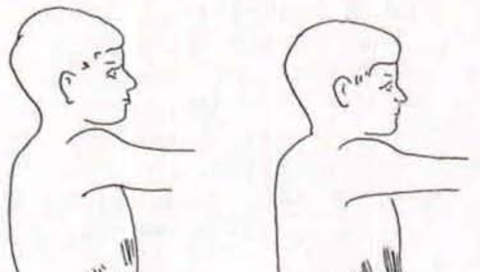

Fig. 2. Pelvic tilt.

The range of pelvic tilt is determined by the tension in the hip joint capsule and reenforcing ligaments, principally the "Y" ligament. If further pelvic tilt is attempted, flexion of the hips is necessary.

In the sitting position the ligaments no longer restrict the pelvic tilt so that the Fick line angle becomes horizontal. This is accompanied by a flattening of the lumbar lordosis.

The Lumbar Lordosis - Thoracic Spine Relationship

The orthotic management of the lumbar lordosis becomes important in the treatment of idiopathic scoliosis and juvenile kyphosis. The righting reflex, the re-alignment of ligaments and muscle leverages, and the neutralization and direction of forces must be considered.

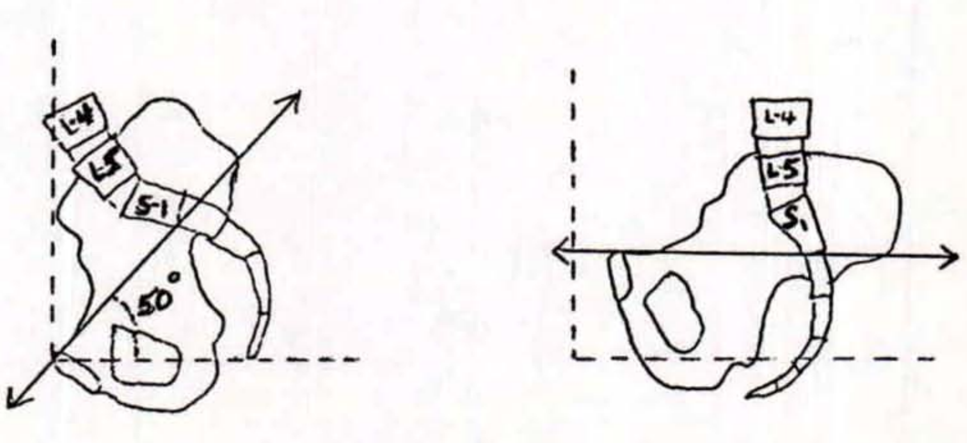

Fig. 3. Fick angle of pelvic inclination.

The Righting Reflex

In juvenile kyphosis, Scheurmanns disease, the anterior wedging of the thoracic vertebrae and the increase of the thoracic curve in the sagittal plane appear as a postural rounding of the back. The increase in the primary thoracic curve results in the development of an increase in the compensatory cervical and lumbar curves, caused by the effort of the body to right itself over the center of gravity, thus creating an even greater "apparent" postural deviation.

Conversely, in the orthotic management of juvenile kyphosis with the Milwaukee CTLSO, the force systems employed, namely 1) the three-point force system of an inferior and superior anterior force countered by a posterior force just below the apex of the kyphus, and 2) the distractive force between the iliac crests and the base of the occiput, are re-enforced and aided by the "righting" reflex created by the correction of the lumbar lordosis. As the lordotic curve is reduced, the shoulder and head are projected more anteriorly, and again the body's mechanism to right itself over the center of gravity results in an anti-kyphotic force and an extension of the thoracic spine (Fig. 4).

The reduction of the lumbar lordosis in the pelvic base of the orthosis results in the righting reflex assisting the anterior posterior forces maintained by the orthosis as well as an immediate "better" posture and the positive beginning of the treatment program.

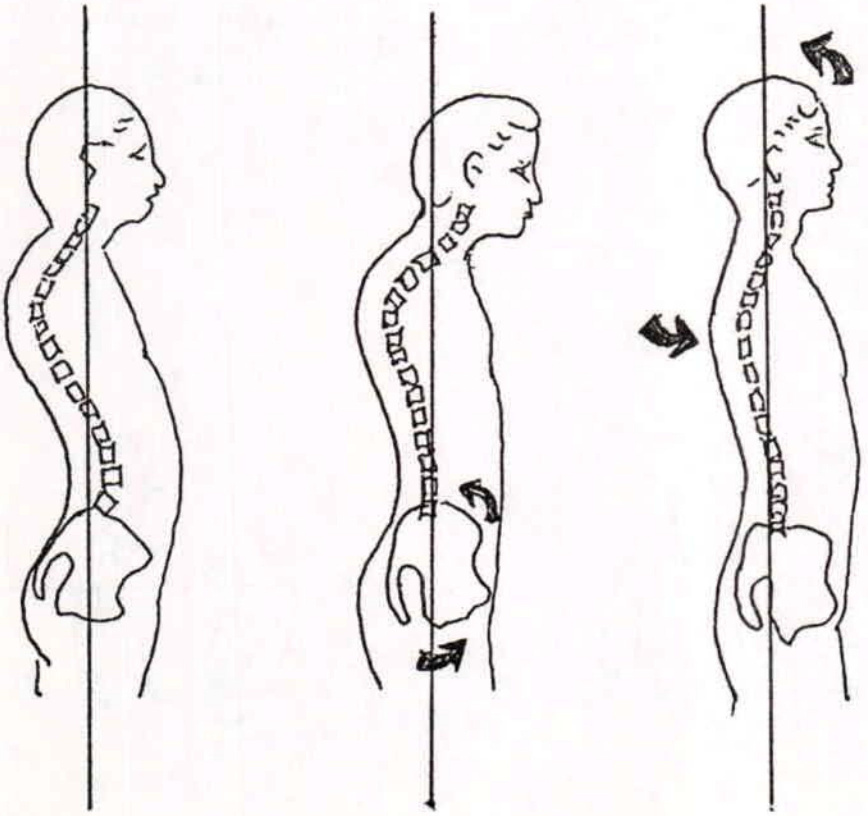

Fig. 4. "Righting" reflex to center of gravity.

Fig. 5. Left, x-ray of spine before treatment; right, x-ray of spine in Milwaukee CTSLO.

An example of the effectiveness of this concept is seen in the comparison x-ray views shown in Fig. 5. On the left is a view of an untreated lordosis, and on the right the same lordosis in the base of a Milwaukee CTLSO, casted in a reduced lordotic position.

For comparison, the angle of lordosis is measured from the inferior border of L-5 to the superior border of L-1 by the "Cobb" method.

The untreated degree of lordosis on the left is 50 deg. The corrected positioning on the right measures 30 deg.

The comparison view of the thoracic spine (Fig. 6) demonstrates the results of the lumbar positioning in the Milwaukee CTLSO. The degree of kyphosis on the left measures 72 deg. The corrected position on the right measures 46 deg.

The reduction of the lordotic and kyphotic curves are readily seen in the clinical comparison slide (Fig. 7) of the patient with and without his Milwaukee orthosis.

The Re-Alignment of Ligament and Muscle Leverages

In the management of thoraco-lumbar scoliosis the control of the lumbar lordosis allows re-development of the mechanical advantages: 1) the re-alignment of ligament and muscle levers, 2) the neutralization and direction of forces, and 3) the stabilization of the pelvis and the lumbar spine.

It is recognized that the scoliotic spine not only deviates laterally but also rotates in the direction of the convexity. This deformity involves the spine and ribs along with the muscle and ligaments attached.

The muscle mass acting on the spine is a multi-layered complex of long and short fibers extending in many directions. Some remain mid-line from vertebra to vertebra, while others project laterally from mid-line to transverse processes and to rib angles. The long fibers divide into parallel columns along the spine spanning from two to ten vertebrae. In scoliosis as the spine deviates

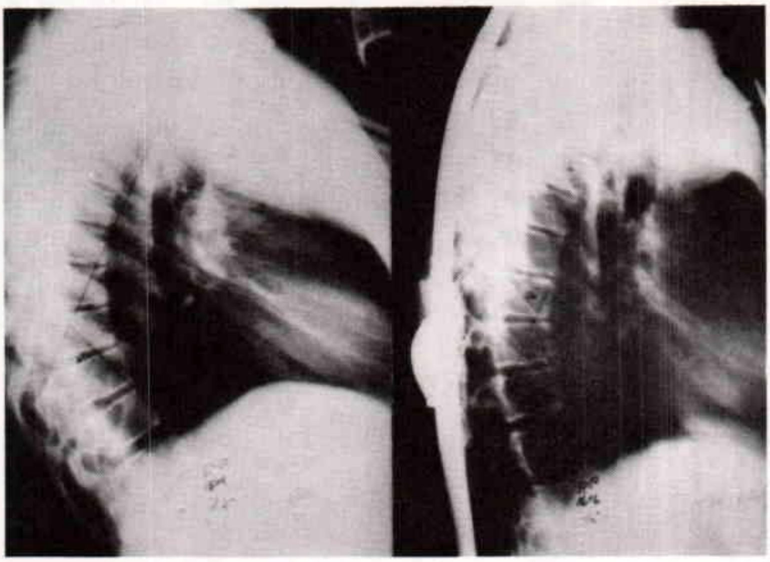

Fig. 6. X-rays of thoracic spine: left, untreated; right, with use of Milwaukee CTSLO.

laterally and the vertebrae rotate, the muscle attachments re-align adversely so they lose their leverages on the spine.

In the thoracic spine the lateral and rotational deviations are seen as a prominence of the rib cage, posterio-laterally (Fig. B). Because of the presence of the rib cage an external force applied to the rib prominence, a postero-lateral to anterior force, has an effect on the vertebrae involved. A lateral force applied to the rib affects the vertebrae at least two levels above where applied. Normal rib angulation is downward so that the lateral rib border is two vertebral levels below its attachment to the spine (1).

In the lumbar spine the absence of rib attachments results in no lateral projections upon which to apply a corrective force. In lumbar scoliosis the lateral and rotational deviations are seen as a muscle prominence and a pelvic obliquity towards the convex side (Fig. 9). Add to this the tendency towards a lumbar lordosis and there develops a three directional deviation.

Re-positioning the vertebrae and thereby re-establishing the muscle and ligament at-

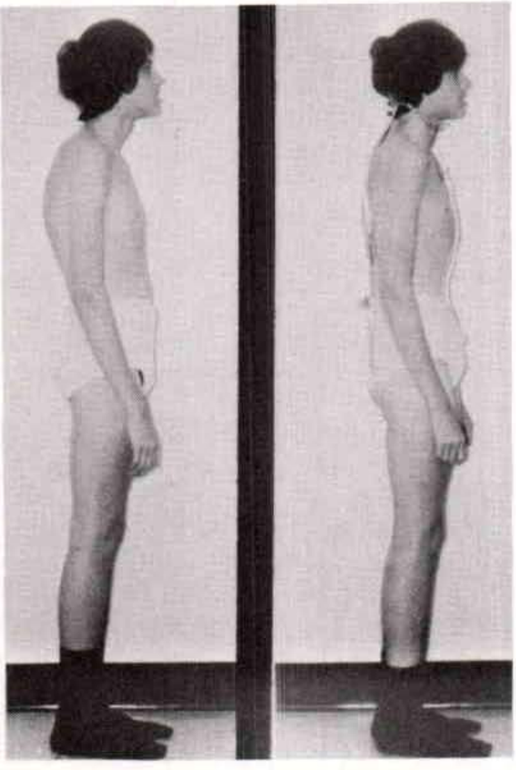

Fig. 7. Patient without and with Milwaukee CTSLO.

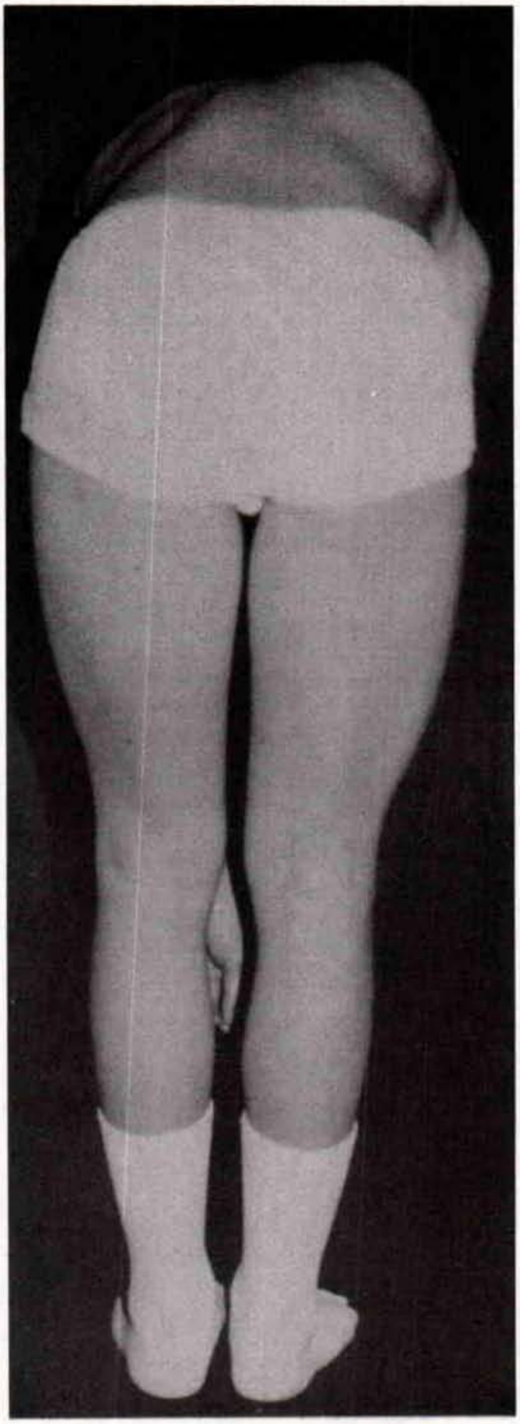

Fig. 8. In the thoracic spine the lateral and rotational deviations are seen as a prominence of the rib cage, postero-laterally.

tachments to their position of leverage becomes a primary concern.

In the management of lumbar scoliosis it is necessary to reduce the lumbar lordosis and to straighten the posterior to anterior "sway", positioning the vertebrae in a mechanically advantageous position before applying a lateral corrective force. Since the corrective force is directed from a posterio-lateral angle, the force lessening the lordotic posture neutralizes the posterior to anterior moment of the corrective force and increases the effect of the lateral moment on the spine. The rotation of the vertebrae and angulation of the transverse processes along with the muscle bulge towards the convexity, are also held in position by the anti-lordotic force to accept the corrective exterior force.

A mechanical advantage is also realized in releasing the stretch of the anterior and posterior longitudinal ligaments caused by the excessive lordosis. The stretch of the anterior ligament and bowing of the posterior ligaments in the lordotic position is critical when the vertebrae are rotated in the scoliosis. The ligaments align more laterally in the scoliosis and when released in the reduction of the lordosis allow more flexibility and less resistance to the corrective lateral forces being applied.

The stabilization of the pelvis upon which to build the orthosis and the increase in intra-abdominal pressure by the encompassing girdle add to the necessity for the reduced lordotic positioning.

Summary

This paper has emphasized the importance of proper orthotic management of lumbar lordosis and its relationship to the treatment of juvenile kyphosis and thoraco-lumbar scoliosis.

In the orthotic management of juvenile kyphosis with a CTLSO the effect of the

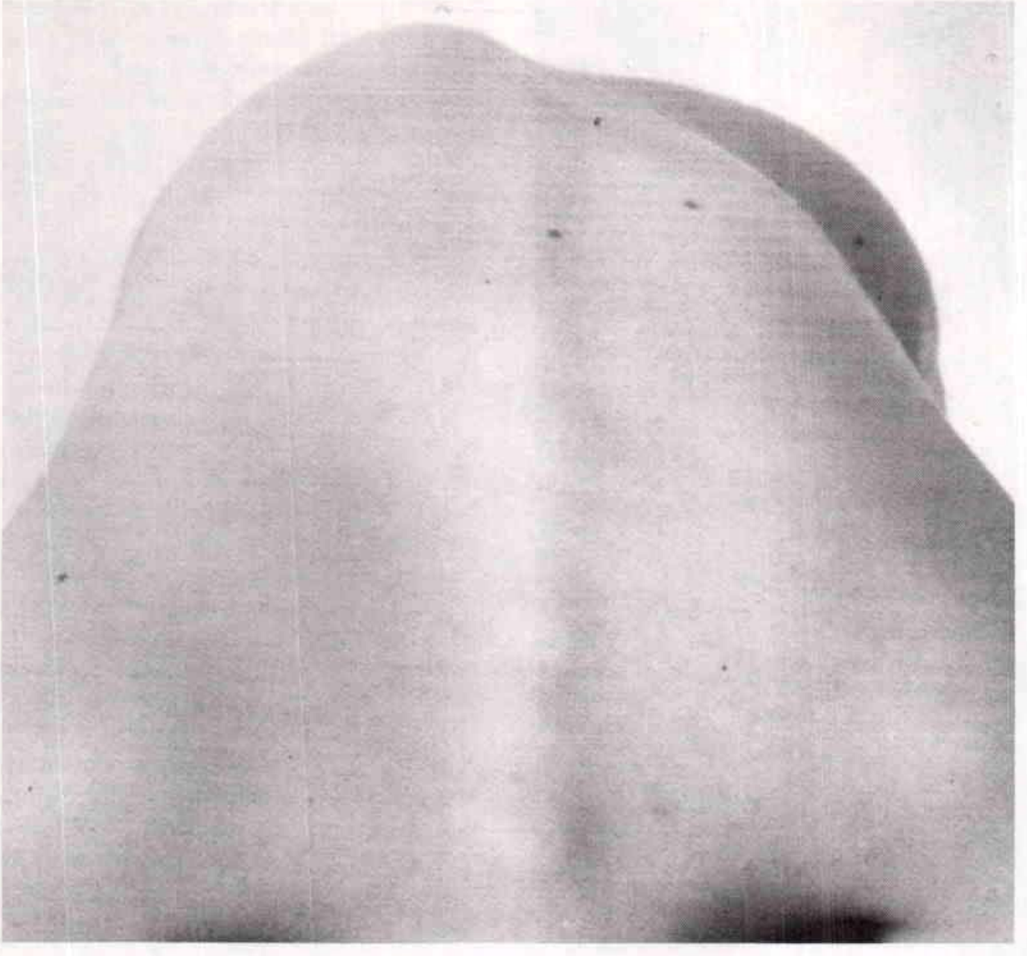

Fig. 9. In lumbar scoliosis the lateral and rotational deviations are seen as a muscle prominence and a pelvic obliquity towards the convex side.

body's righting reflex, as an adjunct to the externally applied corrective forces, has been recognized.

In the orthotic management of the lumbar scoliosis with a TLSO the re-positioning to advantage of the rotated vertebrae and muscle-tendon attachments has been outlined.

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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 (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

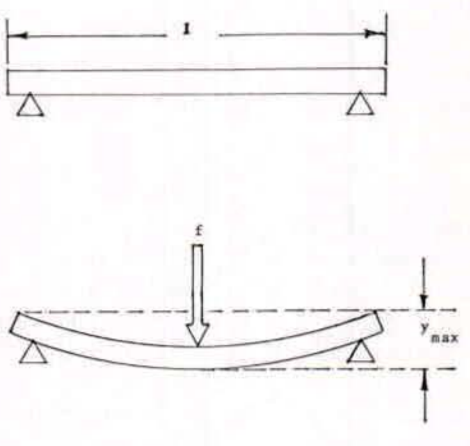

Fig. 1. Basic Beam Deflection

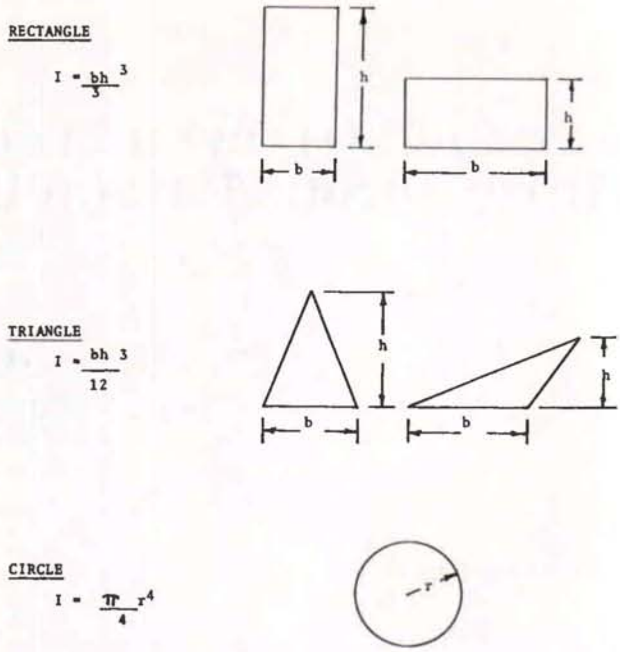

Fig. 2. Moment of Inertia

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

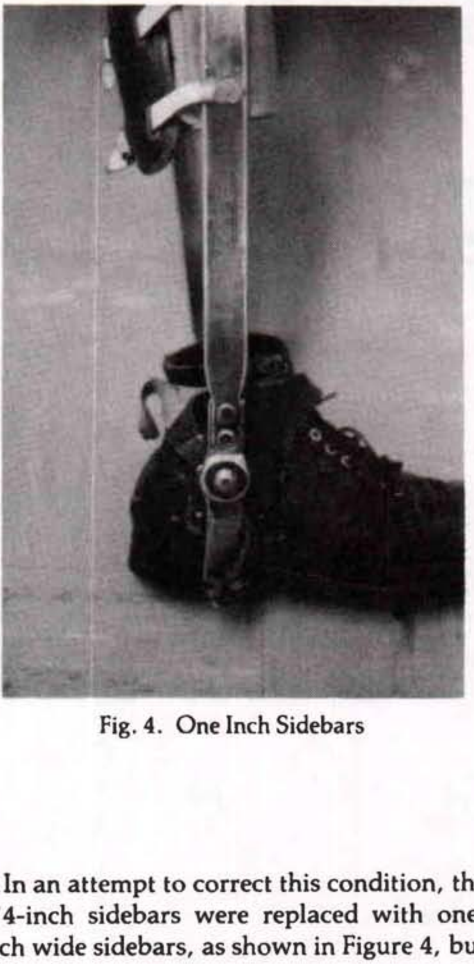

Fig. 4. One Inch Sidebars

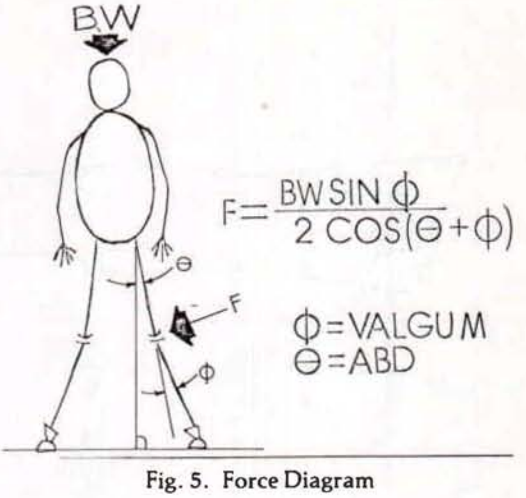

Fig. 5. Force Diagram

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-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

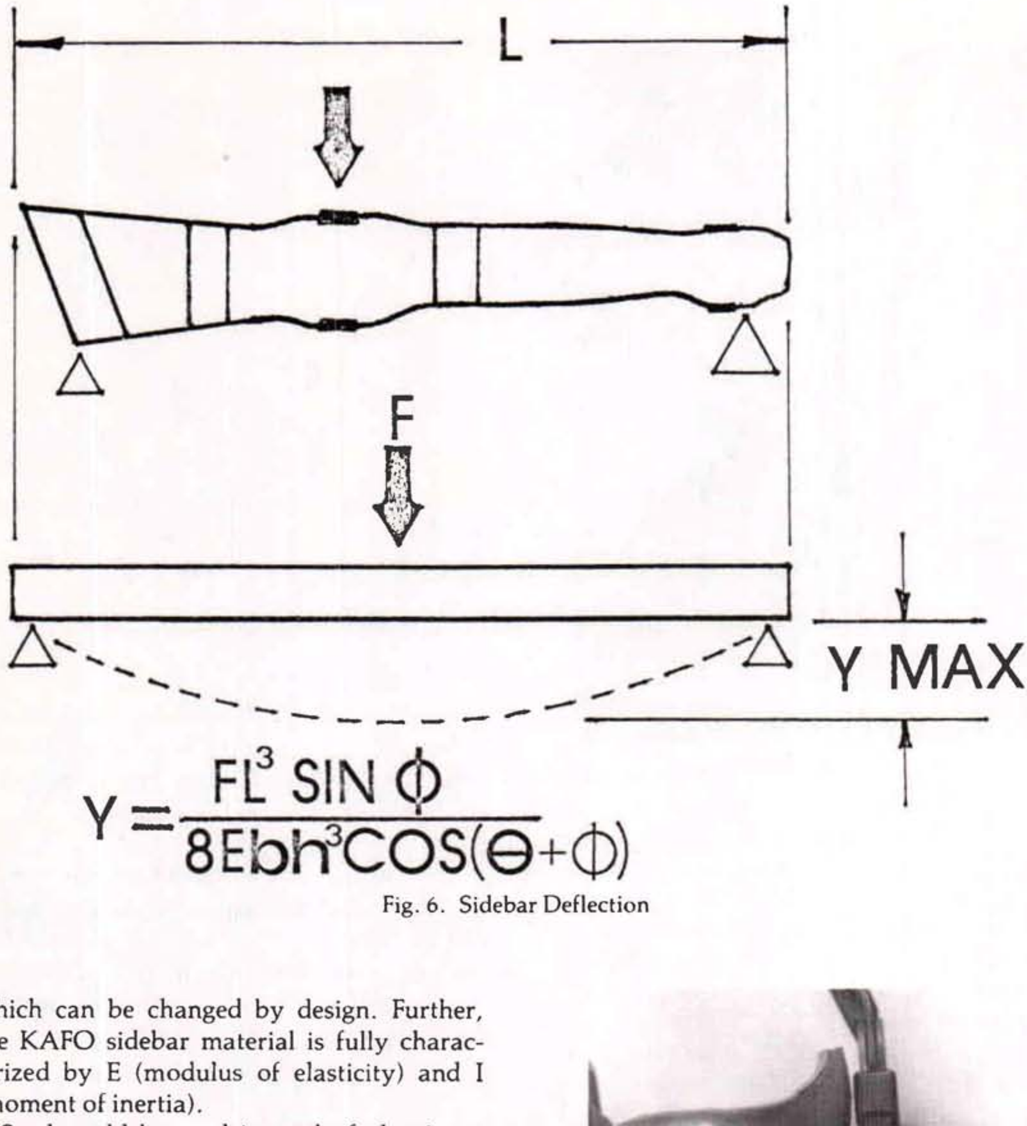

Fig. 6. Sidebar Deflection

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.

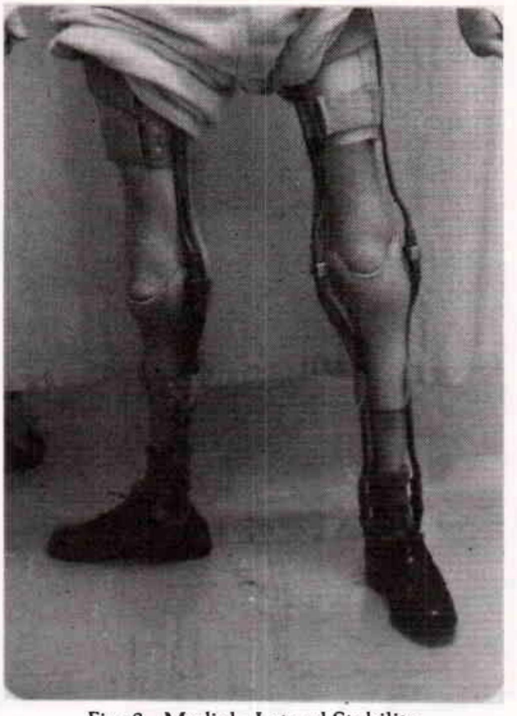

Fig. 8. Medial - Lateral Stability

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

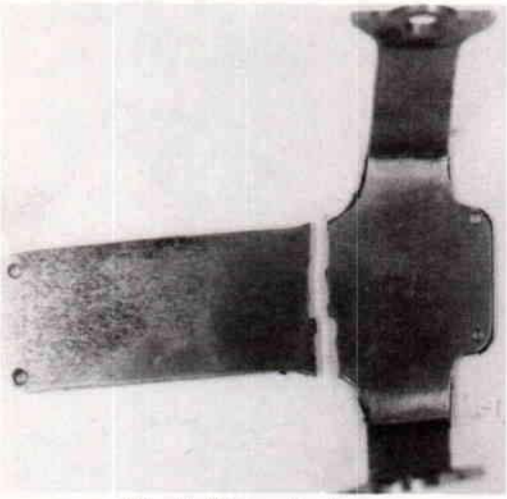

Fig. 9. Stirrup Fractures

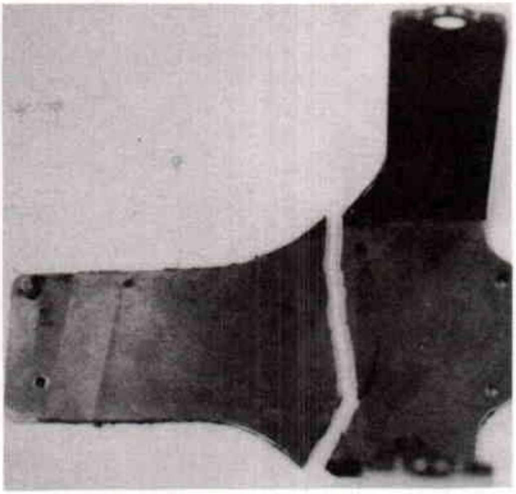

Fig. 10. Heavy Duty Stirrup

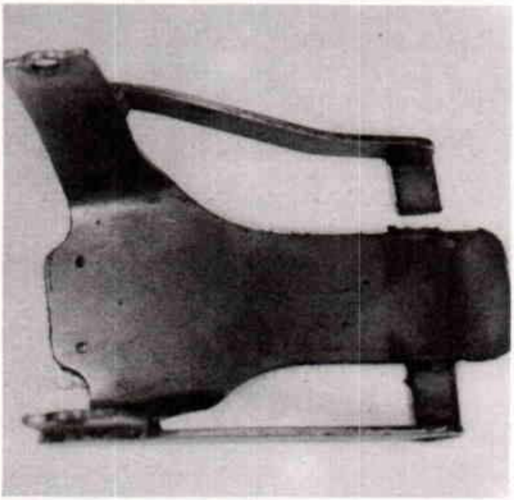

Fig. 11. Reinforced Stirrup

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.

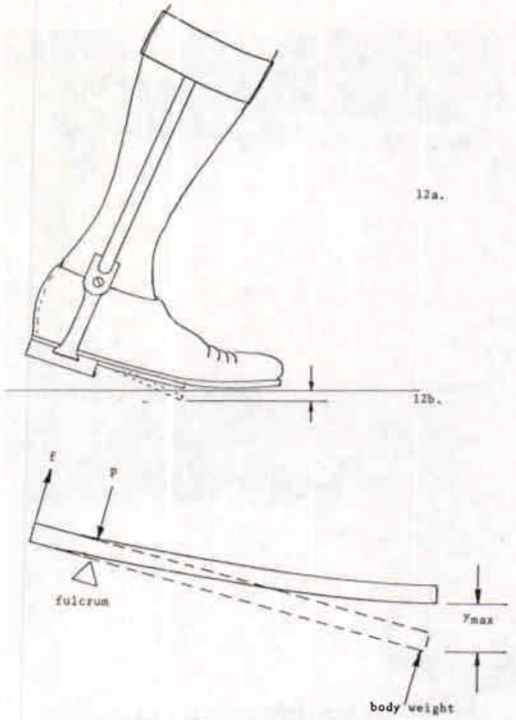

Fig. 12. Force Diagram, Terminal Stance

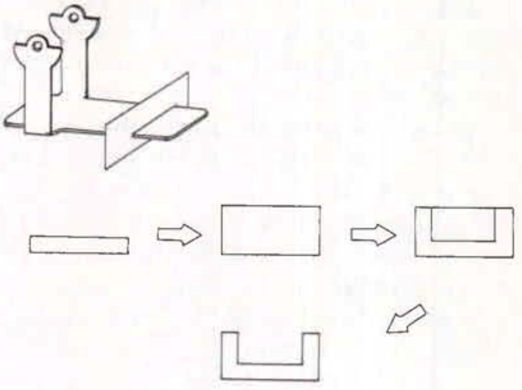

Fig. 13. Stirrup, Transverse Section

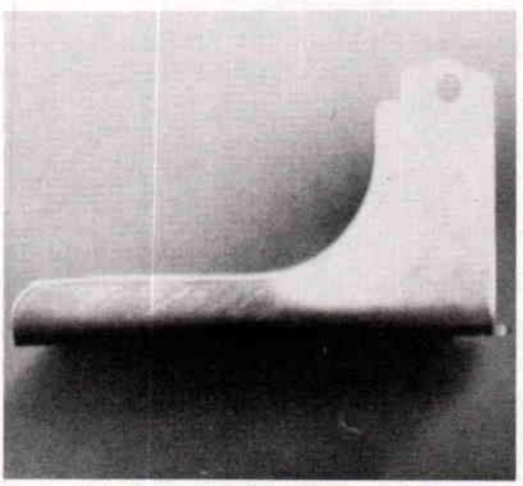

Fig. 14. Photo of Reinforced Super Stirrup

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.

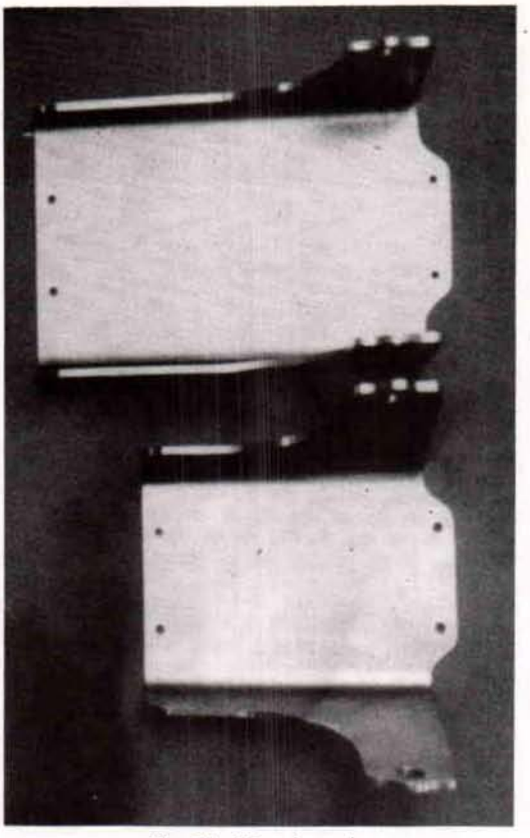

Fig. 15. Two Lengths

Fig. 16. Polypropylene AFO Fracture

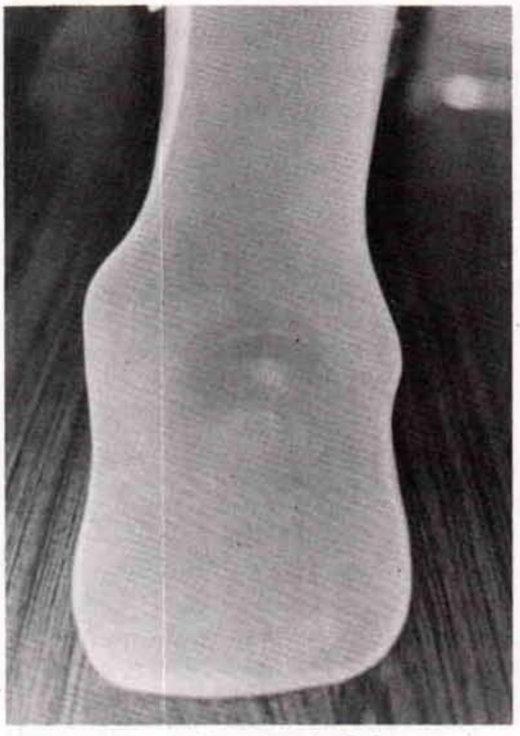

Fig. 17. Characteristic Bulge

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.

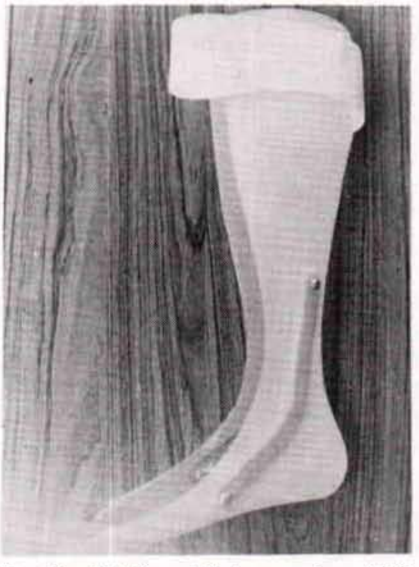

Fig. 18. Reinforced Polypropylene AFO

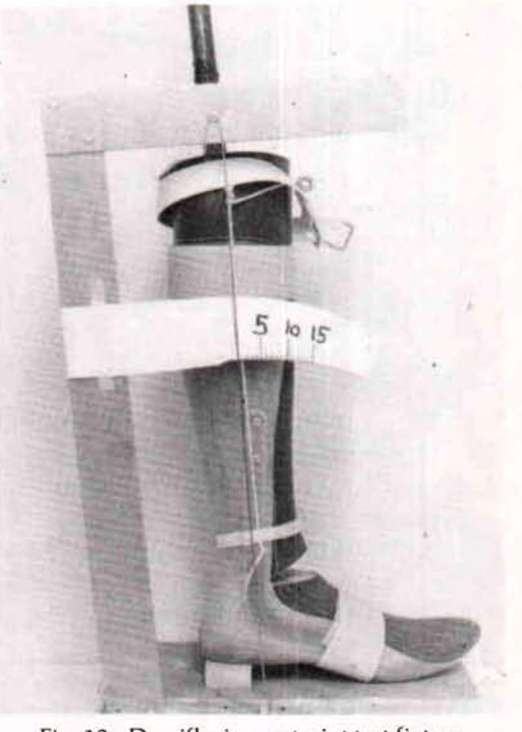

Fig. 19. Dorsiflexion restraint test fixture

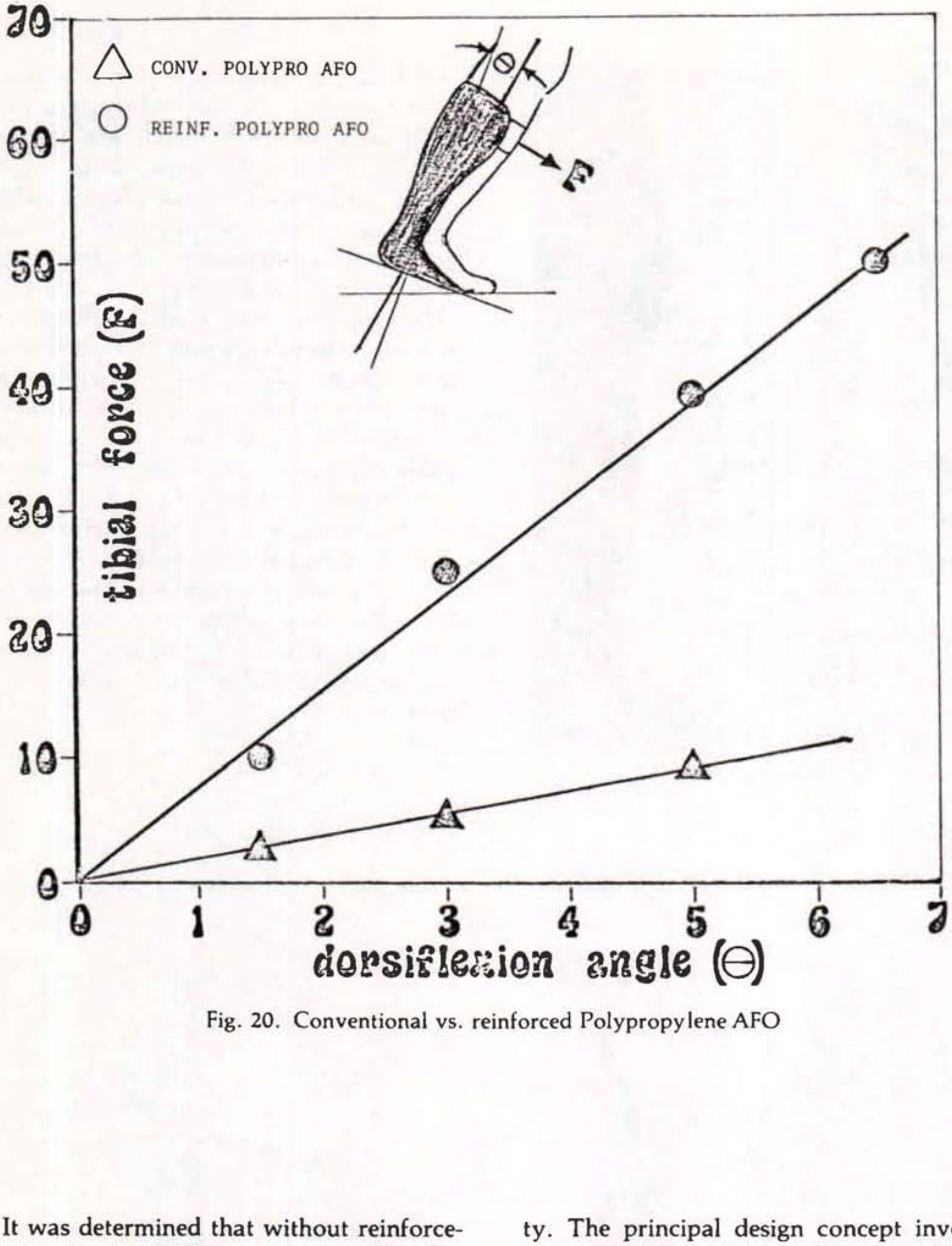

Fig. 20. Conventional vs. reinforced Polypropylene AFO

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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4. Sloane, Alvin: "Mechanics of materials", McMillan Co., 1952; Chapter 4, "Deflection Theory".

Footnotes

1 Supervisor, Orthotic Department, Rancho Los Amigos Hospital, 7450 Leeds Street, Downey, California 90242.

2 Chief, Orthotic Department, Rancho Los Amigos Hospital, 7450 Leeds Street, Downey, California 90242.

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

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LOWER-LIMB ORTHOTICS; A Manual. A. Bennett Wilson, Jr., David Condie, Charles Pritham, Melvin Stills, Publication #M-1-78, Rehabilitation Engineering Center, Moss Rehabilitation Hospital, 12th Street and Tabor Road, Philadelphia, PA 19141; $10.00 pp.

This manual contains new information on material handling techniques that add much to the armamentarium for lower-limb orthotic management of patients. In addition the reader is guided through patient assessment, orthotic nomenclature, and terminology. Matching of required orthotic functions to orthosis designs is covered along with criteria used in formulation of a prescription.

Basic principles of lower-limb orthotics are discussed. The fabrication techniques along with casting and measuring of patient completes this very useful manual. It provides a review guide for physicians and therapists, and, when followed, will ensure that patients benefit from recent research.

This manual is complete and will be a great adjunct to current educational programs.

Bernard C. Simons

ERRATA

The affiliations of Warren A. Carlow, Jr. and Manuel J. Almeida, authors of the article "Plastics in Lower-Limb Orthotics" which appeared on pp. 25-31 in the March 1978 issue of "ORTHOTICS and PROSTHETICS" are incorrect. They should read:

"Owner, Carlow Orthopedic and Prosthetics Inc.,
Warwick, Rhode Island" for Mr. Carlow, and

"Director of Orthotics, Carlow Orthopedic and
Prosthetic Inc." for Mr. Almeida.

We sincerely regret this error.

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RESOLUTION CONCERNING THE METRIC SYSTEM

The following resolution was adopted by the Board of Directors of the American Orthotic and Prosthetic Association at its meeting in San Diego October 3, 1973:

WHEREAS by Act of Congress it has been determined that the United States should proceed towards adoption of the metric system as used almost universally throughout the rest of the world, and

WHEREAS the technological professions and many segments of the health professions have commonly used the metric system over an extended period of time, and

WHEREAS it is important for members of the orthotic/prosthetic professions to interact with their colleagues in the medical and technological communities for optimum patient service be it hereby

RESOLVED that the American Orthotic and Prosthetic Association endorses the use of the metric system by its members and other orthotic and prosthetic practitioners in the United States, and in witness of this endorsement and Association urges the editors of its journal *Orthotics and Prosthetics* to commence the dual reporting of weights and measurements in both the English and metric systems at the earliest possible date with the objective of employing the metric system solely by the time of the 29th Volume in 1975.

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Conversion Factors

LENGTH

Equivalencies

angstrom	= 1 x 10^{-10} meter (0.0 000 000 001 m)
millimicron*	= 1 x 10^{-9} meter (0.000 000 001 m)
micron (micrometer)	= 1 x 10^{-6} meter (0.000 001 m)

To Convert from	To	Multiply by
inches	meters	0.0254†
feet	meters	0.30480†
yards	meters	0.91440†
miles	kilometers	1.6093

AREA

To convert from		
square inches	square meters	0.00063616†
square feet	square meters	.092903

VOLUME

Definition

1 liter = 0.001† cubic meter or one cubic decimeter (dm^3)
(1 milliliter = 1† cubic centimeter)

To convert from	To	Multiply by
cubic inches	cubic centimeters	16.387
ounces (U.S. fluid)	cubic centimeters	29.574
ounces (Brit. fluid)	cubic centimeters	28.413
pints (U.S. fluid)	cubic centimeters	473.18
pints (Brit. fluid)	cubic centimeters	568.26
cubic feet	cubic meters	0.028317

MASS

To convert from	To	Multiply by
pounds (avdp.)	**kilograms**	**0.45359**
slugs‡	kilograms	14.594

FORCE

To convert from	To	Multiply by
ounces-force (ozf)	newtons	0.27802
ounces-force (ozf)	kilogram-force	0.028350
pounds-force (lbf)	newtons	4.4732
pounds-force (lbf)	kilogram-force	0.45359

*This double-prefix usage is not desirable. This unit is actually a nanometer (10^{-9} meter = 10^{-7} centimeter).
†For practical purposes all subsequent digits are zeros.

STRESS (OR PRESSURE)

To convert from	To	Multiply by
pounds-force/square inch (psi)	newton/square meter	6894.8
pounds-force/square inch (psi)	newton/square centimeter	0.68948
pounds-force/square inch (psi)	kilogram-force/square centimeter	0.070307

TORQUE (OR MOMENT)

To convert from	To	Multiply by
pound-force-feet	newton meter	1.3559
pound-force-feet	kilogram-force meters	0.13826

ENERGY (OR WORK)

Definition

One joule (J) is the work done by a one-newton force moving through a displacement of one meter in the direction of the force.

1 cal (gm) = 4.1840 joules

To convert from	To	Multiply by
foot-pounds-force	joules	1.3559
foot-pounds-force	meter-kilogram-force	0.13826
ergs	joules	1×10^{-7}†
b.t.u.	cal (gm)	252.00
foot-pounds-force	cal (gm)	0.32405

TEMPERATURE CONVERSION TABLE

To convert °F to °C $$°C = \frac{°F - 32}{1.8}$$

°F	°C
98.6	37
99	37.2
99.5	37.5
100	37.8
100.5	38.1
101	38.3
101.5	38.6
102	38.9
102.5	39.2
103	39.4
103.5	39.7
104	40.0

*A slug is a unit of mass which if acted on by a force of one pound will have an acceleration of one foot per second per second.

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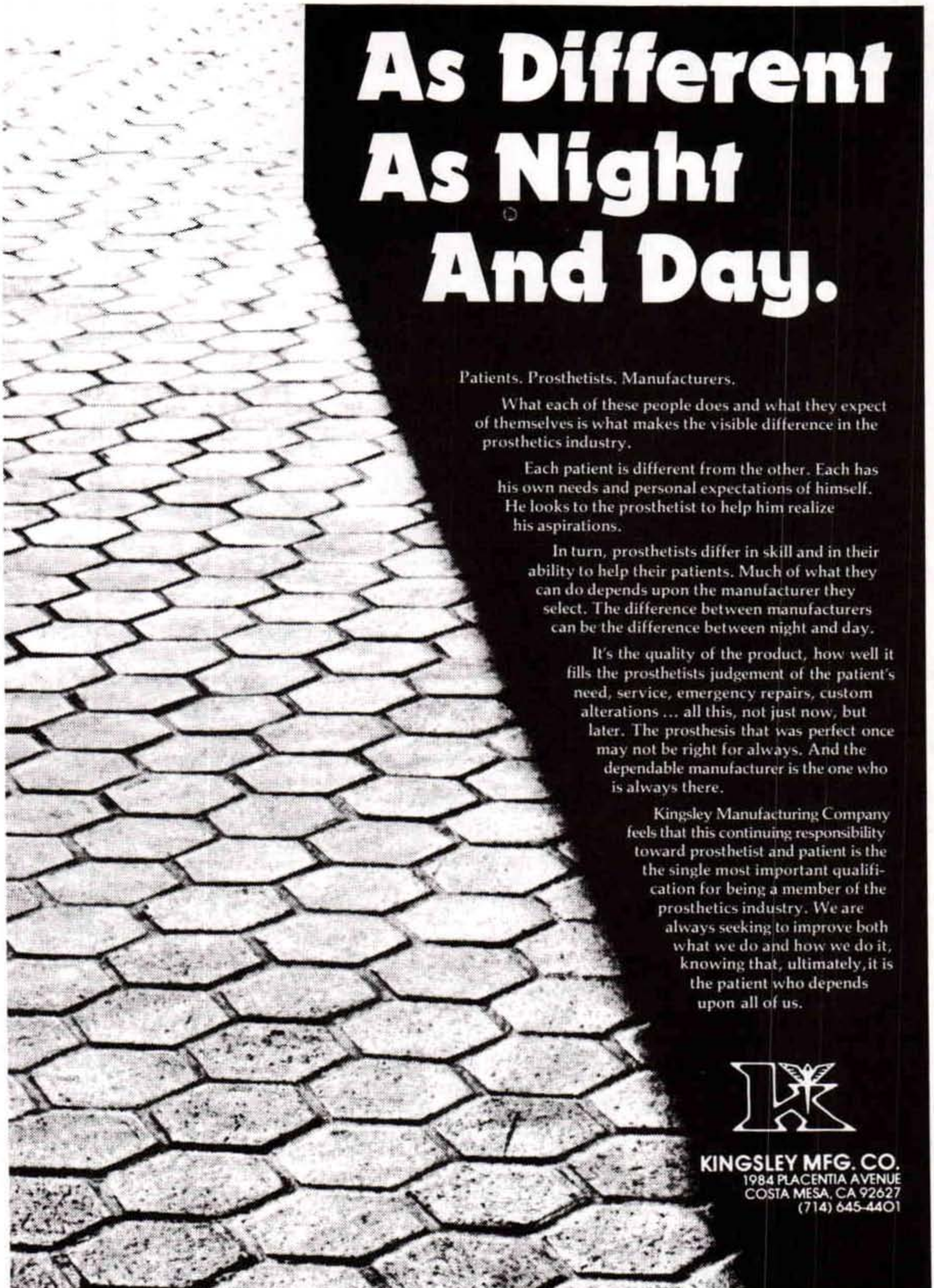
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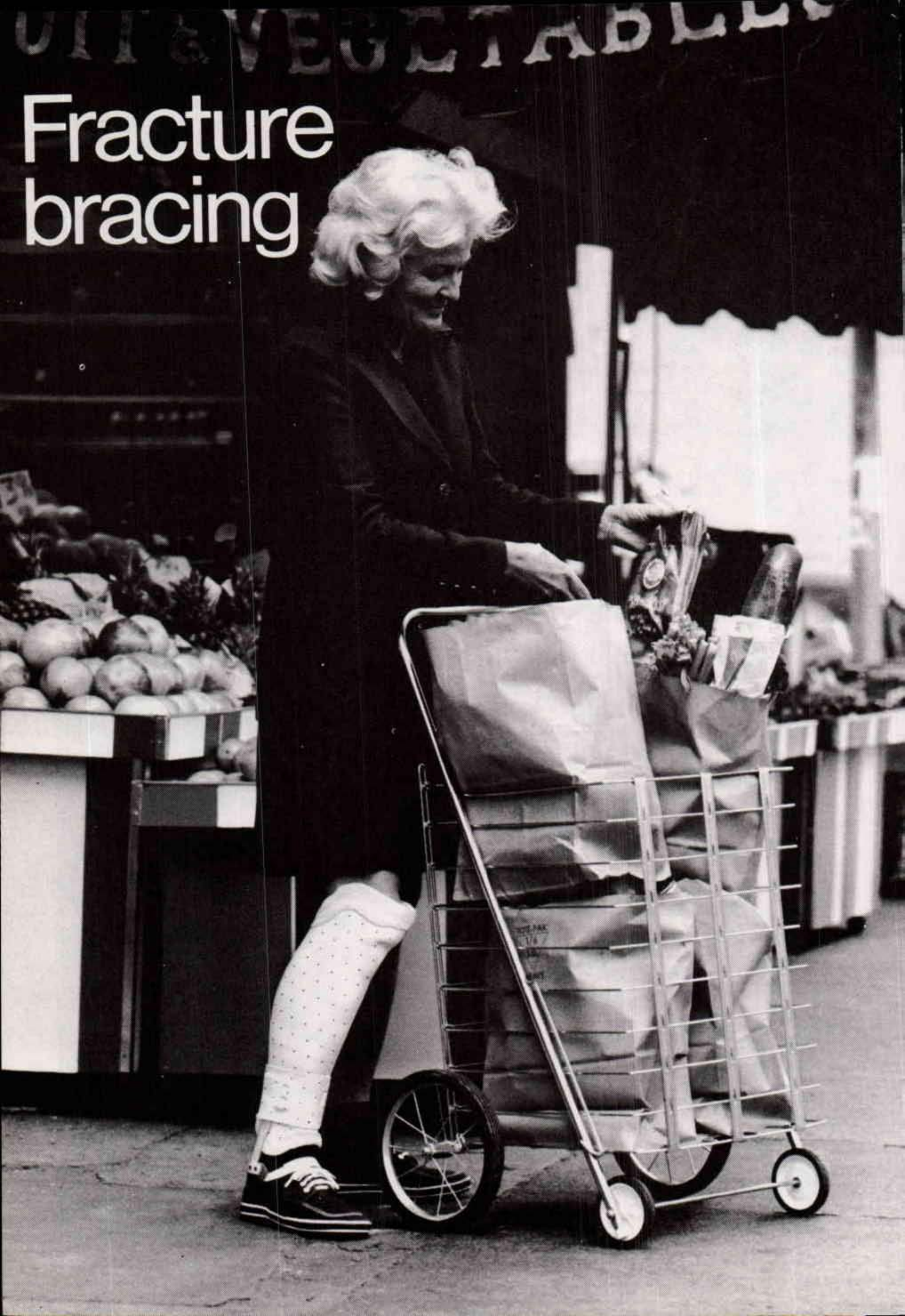

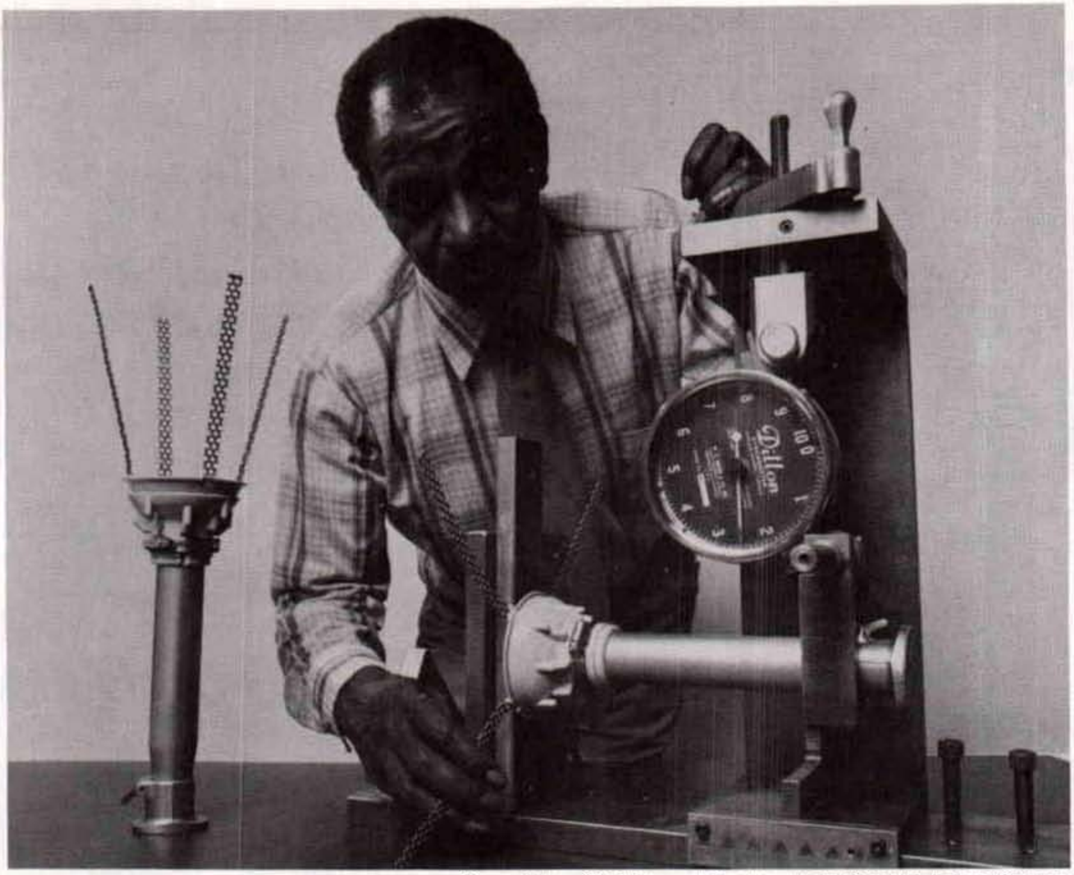

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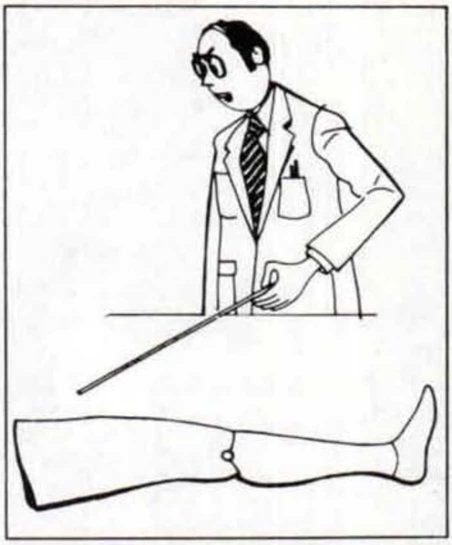

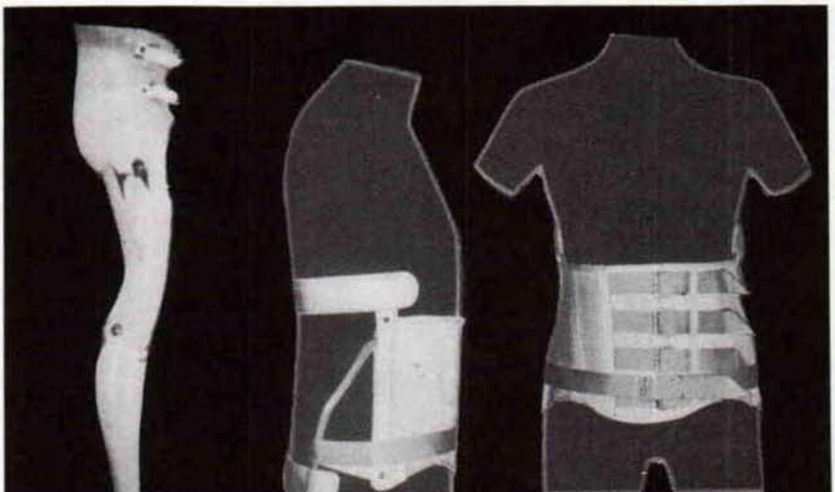

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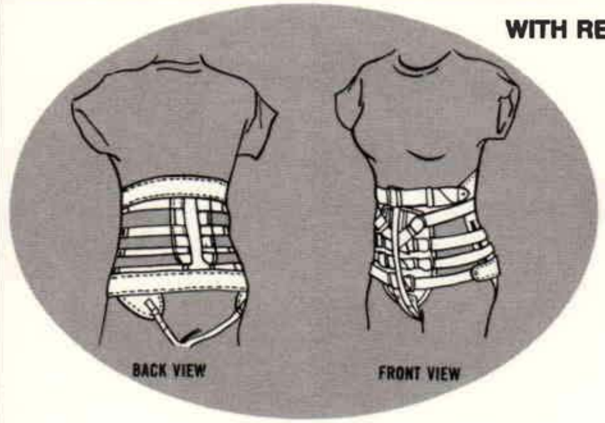

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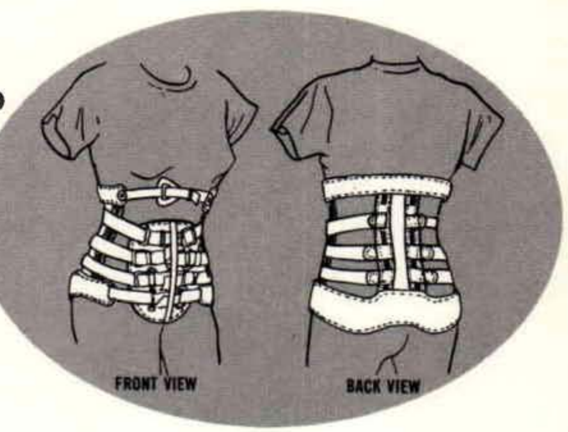
2B-2212-XL	X-Large	40" to 43"	20"	24"	15"



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2B-2213-MD	Medium	32" to 35"	16"	20"	13"
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