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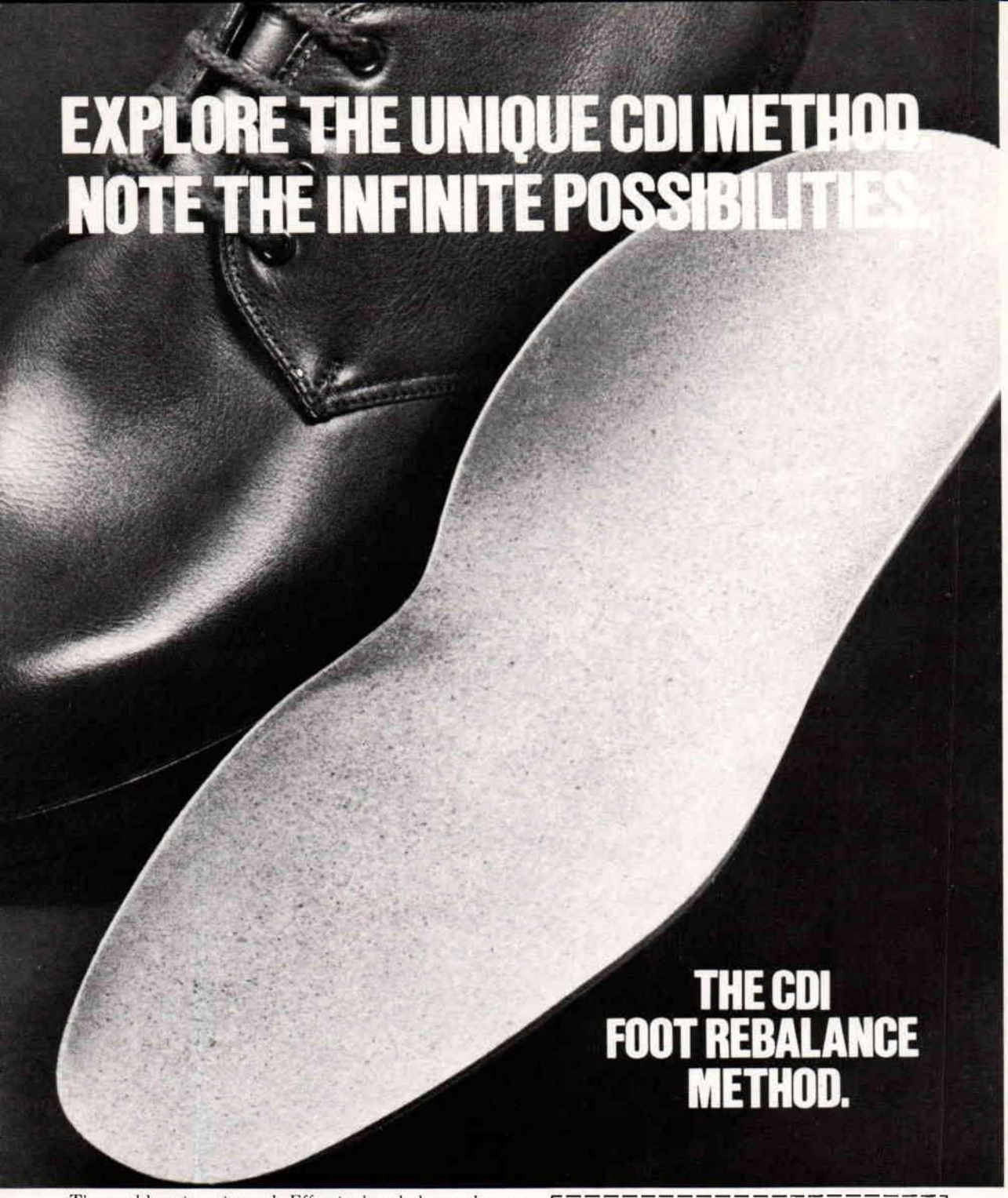

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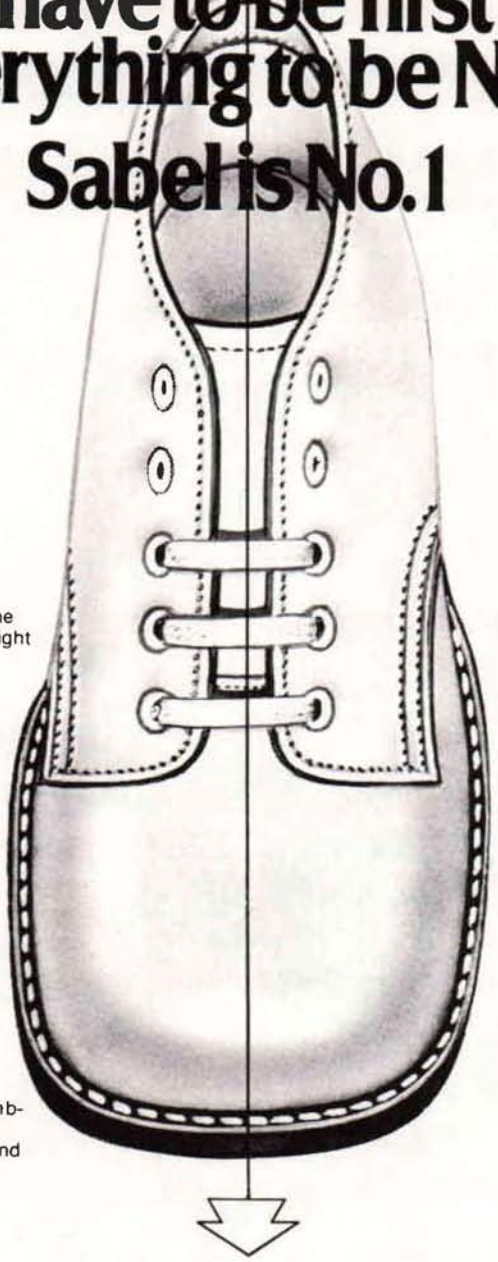

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Orthotics and Prosthetics, Vol. 29, No. 4, pp. 1–5, December 1975

INTER-PROFESSIONAL RELATIONSHIPS[1]

William A. Spencer, M.D.[2]

"Professionalism" and "being a professional" are concepts that deserve consideration in understanding reasons for successes and failures of professional development and professional relationships among the health-related disciplines. This topic is difficult to engage so as to avoid an argumentative style which is not intended. Analysis of trends occurring in professional development of many kinds locates orthotics and prosthetics as an emerging profession and in the light of a new perspective. Traditionally, the hallmarks of a profession are found in the existence of a defined and unique body of transmissible knowledge and technology. A professional helps develop, use, and transmit this knowledge to others. The tendency today is for almost any technical skill to be considered a profession. Obviously, a professional has to do more than practice the techniques of an art. In the last quarter of a century, particularly in orthotics and prosthetics, there is remarkable evidence of professional development. The transition from simply practicing an art to actively developing new knowledge and procedures and to sponsorship of educational efforts such as this seminar suggests that the capability of the orthotic and prosthetic disciplines to serve handicapped people more effectively is increasing.

Willingham[3] has defined some of the operational characteristics of the professional person and his rewards. We should appreciate that there are three principal kinds of professional activities engaged in by professional people. There are practitioners; there are teachers; and there are scholars or scientists. It is interesting that almost every professional body champions the existence of a "compleat"* professional within that discipline—a person who is able to pursue all these activities equally well. The reality is that it is becoming more difficult to accomplish such a balanced and broad professional life; nonetheless, this completeness remains a very worthwhile target. Why is this expectation so commonly held forth for professionals? Willingham points out criteria that many of you have observed during supervision and training of a professional person by the practitioner. Improvement of the skills required by a discipline through observation and demonstration, with a focus on practical problems, is such a criterion. This concept is particularly comfortable to the orthotist/prosthetist. Early involvement in professional affairs such as academies, congresses, and professional associations is sought. There is visible and recognized performance within the profession by certification. Finally, there is evidence of academic competency and of the efforts to improve the quality of the profession.

On the job, the practitioner's standards of success become his leadership characteristics, his certification, his rate of advancement in the area of work he chooses for his practice, and the money he receives for doing it. His patients' gratitude is a major reward. The teacher, on the other hand, depends more on institutional recognition, student and alumni nominations, the judgment of his colleagues, and a leadership role in an academic community to allow him to develop his academic programs as he desires. The scholar/scientist professional is recognized by his tendency to achieve independent accomplishments, that is, accomplishments independent of the current state of the art and knowledge base of his profession. The fact that he does innovative or creative work, makes inventions, and/or uses his creativity in conceptualizing problem areas that need to be solved and finds solutions for those problems are some of his success criteria. The quality and importance of his publications and his academic competency thus define the scholar's

[1]From a presentation made at the First Annual Roundup, American Academy of Orthotists and Prosthetists, Freeport, B.I., January 1975

[2]William A. Spencer, M.D., Professor and Chairman, Department of Rehabilitation, Baylor College of Medicine; and Director, Texas Institute for Rehabilitation and Research, Houston, Texas.

[3]Willingham, W. W., Predicting success in graduate education, *Science*, 183:273–278, January 1974.

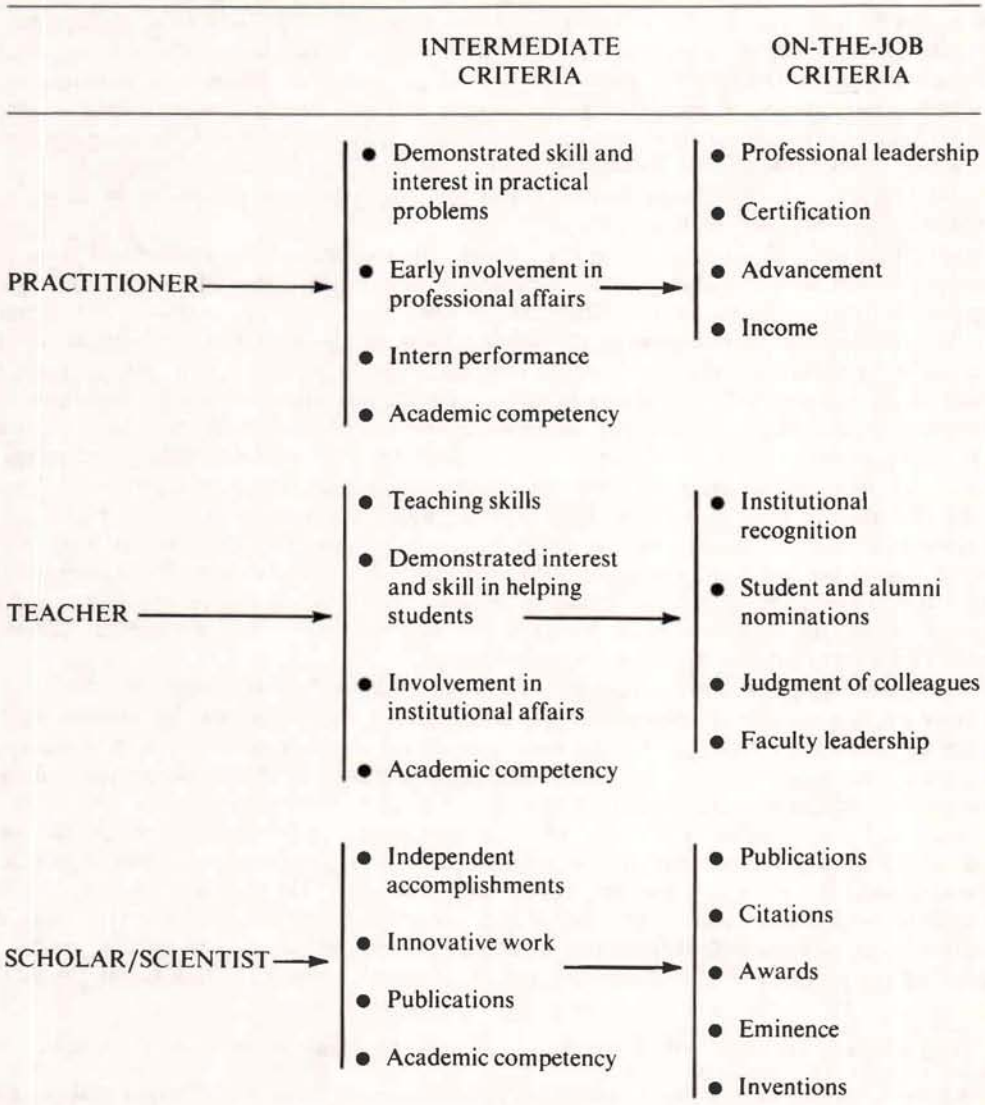
*Apologies to Izaak Walton.

professional status. These publications, citations, and awards reflect his eminence in his field. His problem-solving capabilities and the utility of the inventions or discoveries that he makes are substantial evidence of his success (Fig. 1).

It is obvious that we mean many things by professionalism, the state of being a truly professional person. We are beginning to believe that the appropriateness of different kinds of activities in-

Figure 1.

DESCRIPTIVE CRITERIA OF DIFFERENT PROFESSIONAL ROLES*



*After Willingham

cluding those of the practitioner, the teacher, and the scholar/scientist is proper and necessary. Some rare individuals exist among us who have the attributes of several or all of these functions. The crucial question is whether our educational processes and experiences foster these goals of professionalism or hinder their achievement.

The transition from tradesmanship to professionalism is more than the existence of a unique body of knowledge or skills. It is the development of a field, the transmission to other people of what was learned, and a kind of life-long commitment of the professional person to his own self-development. The requirement of professionalism is competence, both conceptual and technical. How well the person can think out what it is he wants to do and why he wants to do it, as well as how well he does it, is crucial. The simple difference between tradesmanship and professionalism is found in the characteristic of the professional who is interested in *why* as well as how. Trends in the application of scientific methodology indicate that attention to the *why* is essential if we are to devise better *hows*. It is also true that for professionalism to flourish, factors in the physical and social environment of the individual must be reinforced. The professional needs to have personal attributes of competence, willingness, and motivation to develop creativity, and ability to communicate and collaborate with other people. Yet, if he is not in the proper environment with colleagues to model and with helpful physical resources, these attributes may not be expressed. If there are no supportive personal and behavioral relationships which promote opportunities for personal decision-making by the student and encourage sharpening of analytic skills through recognition and problem-solving, the self-growth and developmental process is starved. A developing individual needs not just supervision but consultation and guidance. He especially needs persons to emulate, to respect, and to model, as well as sources of accurate information. Without these intangibles, it is difficult for professionalism to be nourished and to flourish. It is probably in this aspect that the requirements of inter-professional relations can be found. The older professions, such as medicine, have a tremendous responsibility to newly emerging professions, such as orthotics and prosthetics, to provide or stimulate the kind of behavioral climate that the orthotist and prosthetist needs. I do not believe this is widely appreciated. Intra-professional relationships usually are concerned with gaining and transferring knowledge that is appropriate to share among professions. Seminars, including other professionals, special publications, demonstrations, and teaching efforts, do provide examples of your organizational commitment to the development of new people in your field. Inter-professionalism is mostly thought of as teamwork; yet I have a great deal of trouble with that word. The reason is that most of the accomplishments of inter-professional development have not been done by a "football team." It is not a question of a quarterback calling signals and the linemen taking their assigned positions. What we mean is the circumstance where people can, together and simultaneously, perceive a problem or need and solve it with the harmonious and orchestrated interaction of their collective competence. Teamwork, thus, is not putting together a group of incompetent or partially competent people and hoping it will add up to one competent activity. It is putting together different competencies which are equivalent in quality. It includes the sharing of responsibility and even transferring of responsibility back and forth to one another.

Most professions (and I include medicine in this allegation) currently try to differentiate the uniqueness of what they do. This is a kind of territory-defining that means we are not supposed to cross one another's boundaries—as if problems were neatly compartmented into isolated pieces! This kind of behavior absolutely destroys the opportunity for teamwork. It is a dreadful problem, but it is happening in academics, in business, in industry, and in bureaucracies of all kinds. The current vogue is to try to define one profession's body of knowledge so uniquely and so rigidly circumscribe what the professional does that no one else can do it or be involved. Thus, an attitude which is the whole antithesis of teamwork is created. Teamwork is actually an inter-personal attitude which encourages learning from each other, giving credibility to each other's point of view, and recognizing that willingness to use each other's knowledge to solve a common problem more complex than any one person can handle is not piracy but a privilege. Putting together a "martini" of "X" number of proportions of orthotics, prosthetics, medicine, engineering,

psychology, etc., is not the way to go. Multiple professions should be used when the problem solution requires it.

I have never seen professional education or training of any kind which insures or guarantees that a person will know how to be an effective team member or colleague. This quality is much more defined by the attitude and manner of the person, his willingness to learn from other people, his willingness to share responsibility, and his opportunity to have shared responsibility, than it is by professional or technical training. I defined professionalism before I got into the topic of inter-professional relations because, depending upon personal attitude rather than the kind of profession, inter-professional relationships can be developed by learning how to improve and strengthen professionalism. We need to develop our personal and individual insight into those activities we do which are self-gratifying, self-rewarding, and, at the same time, can contribute to our profession and help the growth and development of others. We have to ask ourselves, "Is what I am doing improving myself and improving my profession or only one or the other?" We want to maximize on doing both. There is a personal responsibility to seek an environment which is supportive rather than repressive of professionalism, whether in private practice, institutional practice, or even academic activities. Each of these entirely different functions can be very supportive of professional development or very inhibitory of it. Each professional person has an obligation to teach for more reason than just transmitting knowledge—how much do I know that I can give to someone else. It is the test of usefulness of knowledge in the teaching situation. Teaching tests not only the quality of skills and techniques, but the quality of ideas. Students are particularly adroit at testing this capability. They are increasingly more interested in the "why" as well as the "how" of what they are learning. It is also perfectly proper for the orthotist or the prosthetist to question the physician's clinical decision process and for the physician to question the selection or choice of a device solution if both parties have the correct attitudes.

Independent evaluation of products will foster the reasonableness of questioning the validity of current methods of procedures. The replication by others of what we do (a verification and validation process) has its greatest value in fostering learning from one another. Using the principal analytic rules of science in evaluation is a neutral ground. Understanding the limits of one's competence is another factor which has a great deal to do with inter-professional relations. As a director of an institution, I have great trouble getting people to realize that it is a bigger error to cover up limits of competence for fear of revealing what they do not know than it is to seek advice and consultation. Soliciting advice and assistance at the appropriate time creates a situation to learn and improve what one knows. The orthotics and prosthetics profession has already started by improving the process of certification. I do not know yet that you have approached quality assurance as we in medicine are now being directed to do. This will be increasingly important to your field, as it is in medical care, because both the public and the purchaser of services want to be assured that what they are getting for the investment is provided efficiently and is effective. The personal effort to begin to understand quality assurance and how to carry it out can give your profession a running start on medicine. You can be ready when it becomes mandatory. Usage of any device or procedure that affects physical and emotional well-being of people will undoubtedly be subject to quality surveillance. Certainly, then, you also want to direct quality surveillance toward improving practices, not to policing for failures and errors in practice. The latter approach is a mistake found in many current quality assurance procedures. Most quality control procedures, therefore, are error-detection systems. The focus is on trying to discover poor judgments after the fact, improper actions, and errors of commission and omission. The only way to achieve quality is to increase the level of competence, knowledge, and skills of practitioners in a continuing process. Using quality-control procedures as an educational device can help achieve this quality through a progressive elevation of standards.

Today many of your members are seeking to share and to transfer responsibilities according to needs and particular unique competencies among you. Understanding the difference between consultation and collaboration, and practicing those activities appropriately are important ingredients of this personal prescription for your professional development. The last area which I feel to be sig-

nificant is the need to understand the trends occurring today in health service delivery. Increasingly our citizens expect all health professionals to have an organized, systematic, efficient, and effective structure in which to deliver individualized and personal services. Defining the proper manner in which to achieve such structure comprises a dilemma of world-wide dimensions. The solutions required are not yet clear. Many experiments have been made in the organization and financing of services. There have been many failures because it was not recognized that the ways activities are labeled or services are financed do not seem to have a great deal of effect on how an individual professional practices. He is conditioned mostly by his educational experiences, his postgraduate education, and his professional self-development.

In conclusion, a few popular (or unpopular) caveats are needed. No profession is totally self-sufficient and will be less so in the future. A profession should not be exclusively self-serving or it will simply return to tradesmanship. The best new development comes through creative and innovative professional practices; but development without accompanying widespread availability and distribution is insufficient for thousands of handicapped people if only the developer himself is able to use it. The problem of how to achieve widescale transfer of complex solutions has not yet been solved, especially when the transfer requires drastic changes in practice. Professional boundaries should be overlapping rather than isolated or atrophy of growth and development of each of us will occur. Failure to incorporate related developments and knowledge from others—whether it be in new materials, appropriate application of engineering science and technology, or developments in new engineering knowledge—must be avoided.

Quality of education has to improve so we can learn the method of applied science. We need to learn how to teach each other and to transfer more effectively so we can expand the educational curriculum into the usage of the principles of life sciences and engineering. The orthotic/prosthetic displine needs to develop master's level graduate programs which even include course work in the behavioral sciences and social sciences. Clinical learning experiences should be developed to be supportive of professional growth and development rather than be limited to the provision of technical training under a certified orthotist or prosthetist. The most difficult lesson to learn is to accept failures as knowledge, too. Discovering that something does not work is sometimes the secret to learning what will work. We can be overzealous in thinking that we must find a workable solution every time we try, and overlook the most important value of negative information.

Finally, we have to satisfy the needs of the patient in circumstances of his daily life. We cannot be satisfied only with the efficiency with which we can produce devices or the economy we can gain in shop production. If developments we make in the field, either individually or jointly, do not find practical utility in the daily lives of handicapped people, we have satisfied ourselves but have not solved their problems or met their needs. The orthotic/prosthetic profession is at the frontier of a whole new era of scientific development. We are just beginning to understand the requirements for coupling devices to man to sustain, to improve or to restore personal function, and to limit the disabilities and impairments that occur during growth and development. A greater degree of professionalism is essential in solving such complex problems. More science is going to be necessary, as well as craftsmanship and clinical experience. The profession is penetrating this frontier because more and more I hear you asking "why" as well as telling each other "how."

Orthotics and Prosthetics, Vol. 29, No. 4, pp. 7–13, December 1975

A REFINED CONCEPT IN THE ORTHOTIC MANAGEMENT OF SCOLIOSIS
—A PRELIMINARY REPORT—

John Hall, M.D.[1], M. E. Miller, C.P.O.[2], William Schumann, C.P.[2], and William Stanish, M.D.[1]

Idiopathic scoliosis is a medical enigma. It exists presently as a clinical entity without a known etiology. Several experimental manipulations in animal models have been successful in reproducing spinal curvatures, but to date none of these experiments, it seems, has advanced significantly the remedy for scoliosis.

Valuable contemporary contributions to the surgical amelioration of spinal deformities have been made by Harrington (*6*) and Dwyer (*3,4,5*). These techniques for spinal fusion afford the surgeon new confidence in achieving and maintaining an enhanced degree of correction of the scoliosis.

Although ancient and twentieth century medical literature offered an awesome array of spinal orthoses, it was fundamentally the pioneering of Dr. Walter Blount and Dr. Albert Schmidt (*1*) that has provided the medical profession with the "Milwaukee Brace," an orthosis that can successfully abort the progression of scoliosis. When fabrication and fitting of the Milwaukee Brace are carried out properly, carefully selected individuals suffering from scoliosis can be treated successfully without surgery. These vital prerequisites must be combined with a structured exercise program in order for the treatment to yield optimum results.

The conventional Milwaukee Brace has undergone many modifications since its creation, but the metal vertical superstructure with a pelvic foundation still comprises the basic components.

The Milwaukee design has been well received, in spite of its unconventional approach, but even its innovators will admit the Milwaukee Brace still leaves something to be desired. It requires considerable time to fabricate and fit. With the superstructure its appearance is ungainly—a very real factor in the age group for which treatment is generally needed.

In an effort to overcome some of the well-recognized problems associated with the "Milwaukee Brace," a design that has been designated the "Boston System for Non-Operative Control of Scoliosis" has been developed and applied at the Children's Hospital Medical Center, Harvard University.

The system is based on the Milwaukee Brace concept but the necessity to take a cast, pour, and rectify a positive model is eliminated in an estimated 95 percent of the cases by use of prefabricated plastic pelvic girdles (Fig. 1). The

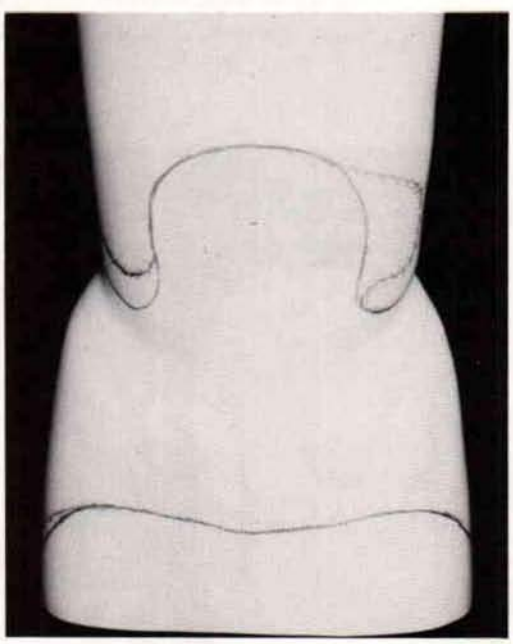


Fig. 1. Preformed polypropylene pelvic girdle. Typical trimlines can be seen. Sixteen sizes are available, but six seem to be all that is needed for 80 percent of the patients.

[1]The Children's Hospital Center, 300 Longwood Avenue, Boston, Mass. 02115.
[2]National Orthotics and Prosthetics Corporation, 300 Longwood Avenue, Boston, Mass. 02115.

girdle is fitted so that its entire inner surface is in contact with the skin (Fig. 2). A superstructure may or may not be used, depending upon the severity and type of case (Figs. 3 and 4).

The pelvic girdles are molded of sheet polypropylene and lined with Ali-Med, a synthetic sponge. Pelvic tilt is incorporated and the lumbar pad is an integral part of the structure.

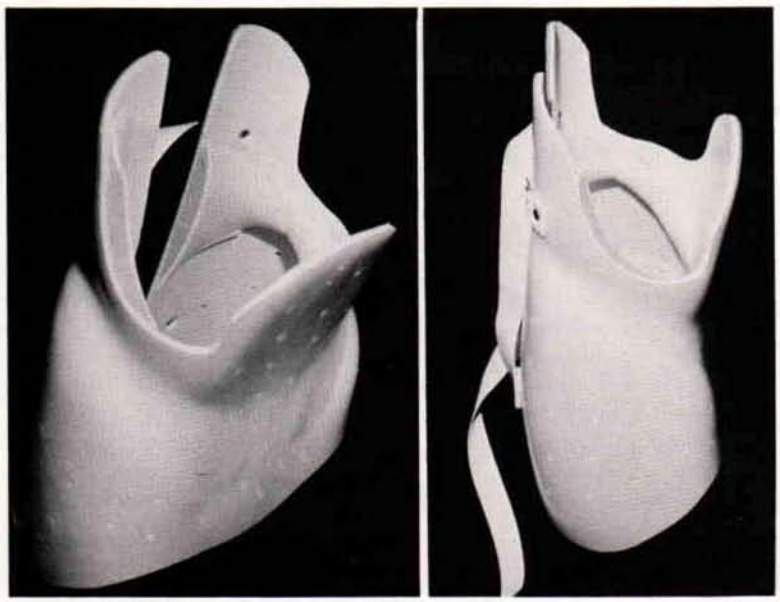


Fig. 2. Two views of polypropylene pelvic girdle lined with synthetic sponge, trimmed, and ready for fitting.

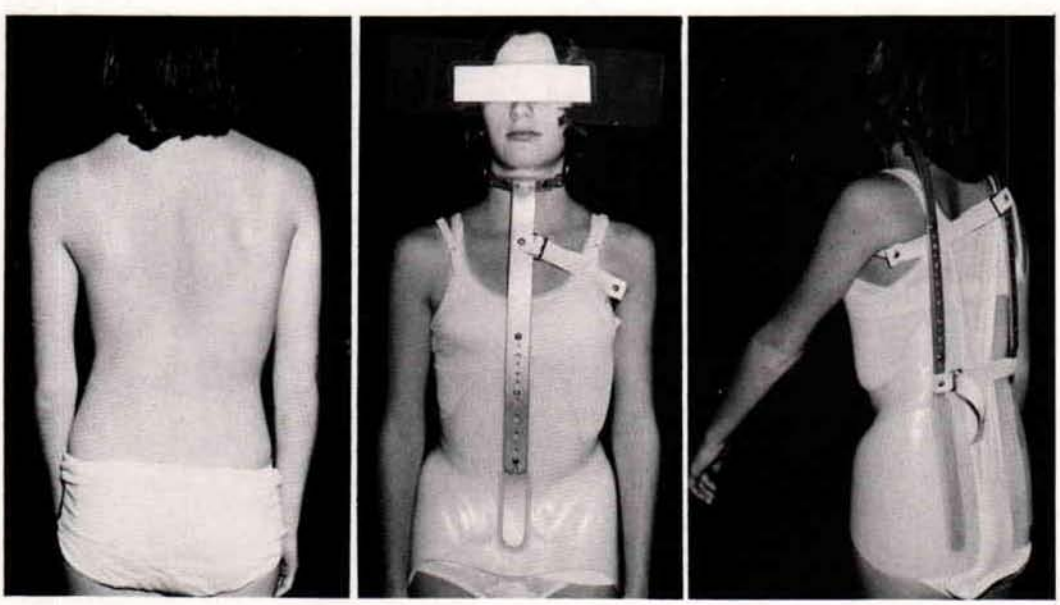


Fig. 3. An example of the Boston System when the superstructure is used.

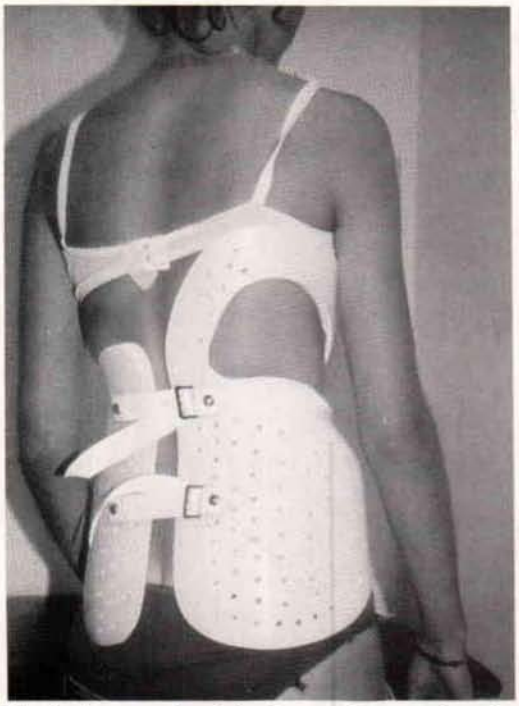


Fig. 4. An example of the Boston System when the superstructure is not used.

"Total contact" is provided, but the pressure is not distributed uniformly, since relief areas are incorporated so as to allow the pathological curve to move to a more ideal position. Static correction of the curve is achieved by direct stress via orthosis, and dynamic correction is accomplished by exercises executed while in the orthosis. These static and dynamic stresses encourage the pathological curve to migrate to the relief area.

The pelvic girdle is available in 16 sizes. Six of these sizes seem to be all that are needed to fit 80 percent of patients for whom an orthosis is prescribed. All 16 sizes make it possible to fit about 95 percent of the cases, and the remainder require a custom-made girdle.

Although fabrication and delivery times are shortened, proper application requires a great deal of knowledge and skill, and therefore should not be attempted until the team has had adequate training.

The advantages of the Boston System are:

1. A reduction in fitting and fabrication times
2. A reduction in delivery time
3. A reduction in skin problems owing to the intimate fit which tends to decrease the degree of shear between orthosis and skin.
4. Improved control of lumbar and thoracolumbar curves owing to the intimate fit of the girdle.
5. A reduction in maintenance problems, owing mainly to the properties of polypropylene.
6. Improved cosmesis (when the superstructure can be eliminated).

EVALUATION

At the Scoliosis Clinic of the Children's Hospital Medical Center approximately 2000 patients with various forms of spinal deformities are followed at any one time. About 500 new patients are seen annually. For approximately three years our team has used the Boston System in an effort to determine its effectiveness.

OBJECTIVES OF STUDY

The study was carried out in an effort to assess two major areas:

1. *Toleration of the Device:* It was an objective to ascertain whether the patient could tolerate the rigid construction of the orthosis designed for total contact. Also, an attempt was made to gain some insight into the greater ease of socialization for the child fitted with this type of orthosis.
2. *Curve Correction and Maintenance of Correction:* It was felt to be imperative that a report be offered on the degree of correction achieved with the Boston System, and also to offer a preliminary report on the maintenance of the correction.

COMPOSITION OF THE GROUP STUDIED

A group of 200 patients was selected on a random basis from the files of the Scoliosis Clinic at the Children's Hospital Medical Center.

Each of these two hundred patients met the following criteria:

1. Had scoliosis that has been diagnosed as idiopathic.
2. Had been fitted with the Boston System, with or without superstructure.
3. Would be a participant in a follow-up program conducted by the orthopaedic staff in con-

junction with the personnel of the National Orthotic and Prosthetic Company who operate the orthotics facility on the premises of the Children's Hospital Medical Center.

SOURCES OF DATA

The data were obtained from chart and radiological review, questionnaire, personal interview, and examination.

The clinic and hospital charts of all patients were reviewed completely. Roentgenograms were also reviewed and measurements recorded according to the Cobb method (*2*). Patients were categorized according to curve patterns as follows:

1. Thoracic
2. Thoracolumbar
3. Lumbar
4. Double curve

The questionnaire was administered to the patient and the parent attending. This was carried out at the time of personal interview and examination of the patient in most instances. Telephone interviews were employed to augment the data when needed for completeness.

RESULTS

Curve Pattern—Thoracic:

Number of patients	62
Sex Incidence	54 females 8 males
Mean Age at Time of Diagnosis	11 8/12 years old
Mean Age at Time of Bracing	12 2/12 years old
Mean Age at Application of the Boston System	12 8/12 years old
Patients fitted with Boston System including the superstructure	50
Patients fitted with Boston System without the superstructure	12
Mean Initial Curve with Boston System including the superstructure	33 degrees
Mean Prime Correction	21 degrees
Percent Correction	36
Mean Initial Curve with Boston System without the superstructure	24 degrees
Mean Prime Correction	10 degrees
Percent Correction	58

Skin Problems:	Nil	32 patients = 52 percent
	Mild	24 patients = 38 percent
	Moderate	3 patients = 4 percent
	Severe	0
Pain:	Nil	36 patients = 58 percent
	Mild	24 patients = 38 percent
	Moderate	0
	Severe	2 patients = 4 percent
Socialization:	Normal	46 patients = 74 percent
	Abnormal	16 patients = 26 percent
Rejection of Orthosis		
Partial		5 patients = 8 percent
Complete		1 patient = 1.6 percent
Patients weaned without progression of curve		2
Patients weaned with progression of curve		1

Comments:

1. Skin problems are deemed to be mild when there is merely hyperemia without blistering; moderate is defined as skin irritation to the point of blistering in one focus; severe skin problem is defined as blistering in more than one focus.

2. Mild pain is that discomfort which does not promote removal of the orthosis for any portion of the prescribed wearing time; severe pain is when the brace cannot be tolerated whatsoever.

3. Socialization is abnormal when either the child or a parent confesses that wearing the Boston System handicaps the child from doing activities that she has a desire to do, i.e., school activities, athletics, and coeducational social activities.

4. One patient weaned with progression of her curve progressed three deg. from 12 deg. to 15 deg., over a two-year period. On her last visit she had arrested at 15 deg.

Curve Pattern—Thoracolumbar:

Number of Patients	54
Sex Incidence	50 females 4 males
Mean Age at Time of Diagnosis	12 8/12 years old
Mean Age at Time of Bracing	13 years old
Mean Age at Application of the Boston System	13 3/12 years old
Patients fitted with Boston System with the "Milwaukee" superstructure	10
Patients fitted with Boston System without the "Milwaukee" superstructure	44
Mean Initial Curve with Boston System with the "Milwaukee" superstructure	24 degrees
Mean Prime Correction	15 degrees
Percent Correction	42
Mean Initial Curve with Boston System without the "Milwaukee" superstructure	25 degrees
Mean Prime Correction	10 degrees
Percent Correction	64
Skin Problems: Nil	38 patients = 70 percent
Mild	12 patients = 22 percent
Moderate	0
Severe	4 patients = 8 percent
Pain: Nil	38 patients = 70 percent
Mild	16 patients = 30 percent
Moderate	0 patients
Severe	0 patients
Socialization: Normal	48 patients = 88 percent
Abnormal	6 patients = 12 percent
Rejection of Orthosis: Partial	2 patients
Patients weaned without progression of curve	1
Patients weaned with progression of curve	2

Comments:

1. One patient who absolutely refused to wear the Boston System with the metal superstructure also refused to wear the modified pelvic girdle alone, and progressed to spinal fusion with Harrington instrumentation.

The second patient was converted to the Boston System without superstructure and has been controlled since that time.

2. Of the two patients who progressed following weaning, one patient has lost three deg. of correction which is considered negligible at the present time. However, one patient who presented with a 30- deg. curve had a prime correction of 12 deg. after an insidious weaning period progressed to 35 deg., and went on to spinal fusion with Harrington instrumentation.

Curve Pattern—Lumbar:

Number of Patients	26
Sex Incidence	22 females 4 males
Mean Age at Time of Diagnosis	13 1/2 years old
Mean Age at Time of Bracing	14 years old
Mean Age at Time of Bracing with the Boston System	14 2/12 years old
Patients fitted with the Boston System with the "Milwaukee" superstructure	0

Patients fitted with the Boston System without the "Milwaukee" superstructure	26
Mean Initial Curve with the Boston System	26 degrees
Mean Prime Correction	12 degrees
Percent Correction	54

Skin Problems:	Nil	20 patients = 76 percent
	Mild	6 patients = 24 percent
	Moderate	0 patients
	Severe	0 patients
Pain:	Nil	20 patients = 76 percent
	Mild	0 patients
	Moderate	6 patients = 24 percent
	Severe	0 patients
Socialization:	Normal	24 patients = 92 percent
	Abnormal	2 patients = 8 percent

Rejection of Orthosis:	
Partial	0 patients
Complete	0 patients
Patients weaned without progression of curve	1 patient
Patients weaned with progression of curve	0 patients

Curve Pattern—Double Curve:

Number of Patients	58
Sex Incidence	56 females 2 males
Mean Age at Time of Diagnosis	11 7/12 years old
Mean Age at Time of Bracing	12 9/12 years old
Mean Age at Time of Bracing with the Boston System	13 years old
Patients fitted with the Boston System with the "Milwaukee" superstructure	30

Patients fitted with the Boston System without the "Milwaukee" superstructure	28
Mean Initial Curve with the Boston System with the "Milwaukee" superstructure	26 degrees
Mean Prime Correction	18 degrees
Percent Correction	30
Mean Initial Curve with the Boston System without the superstructure	27 degrees
Mean Prime Correction	14 degrees
Percent Correction	49

Skin Problems:	Nil	44 patients = 75 percent
	Mild	12 patients = 21 percent
	Moderate	2 patients = 4 percent
	Severe	0 patients
Pain:	Nil	46 patients = 77 percent
	Mild	12 patients = 23 percent
	Moderate	0 patients
	Severe	0 patients
Socialization:	Normal	50 patients = 86 percent
	Abnormal	8 patients = 14 percent

Orthosis Rejection:	
Partial	0 patients
Complete	0 patients
Patients weaned without progression of the curve	2 patients
Patients weaned with propression of the curve	1 patient

Comments:

The one patient who has progressed following weaning deteriorated from an 18–24 deg. curve to a 20–30 deg. curve over a one-year period. This progression occurred after a program using the Boston System without a superstructure.

CONCLUSIONS

The Boston Scoliosis Orthosis is a prefabricated appliance for the nonsurgical treatment of idiopathic scoliosis. Constructed, primarily, from sheet polypropylene, the orthosis is designed to offer a total-contact fit.

In patients selected properly, the need for the metal superstructure is obviated. Preliminary report of 200 patients with idiopathic scoliosis treated with the Boston System reveals, after a mean follow-up period of 17 months, a 28 deg. curve (mean) can be corrected to a 14 deg. curve (mean) yielding a 50 percent correction. While wearing the brace 20 hours per day (mean), 85 percent of the patients enjoyed a completely normal life style.

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ABOVE-KNEE PROSTHESIS SC-75

S. Sandrolini Cortesi[1]

To improve the function of above-knee amputees during walking, we focused on some problems relative to the design of an above-knee prosthesis with particular regard to:

- Conditions of stability during heel-contact, the stance phase, and push-off.
- The kinematics of the prosthesis during swing phase with particular reference to the minimum height of the trajectory of the foot above the floor.

The result is the prosthesis SC-75 shown in Figures 1 and 2.

The mechanism (Fig. 3) is composed of two coaxial tubes "A" and "B" which slide freely upon each other, being guided by bushings of a material with a low coefficient of friction. The external tube "A" is connected to the socket by the two connecting links "C"; the rod "B" is connected to the base of the socket by a hinge; the foot is hinged to member "A" at point "F" and is connected to "B" by the hinge "G" through the ball-and-socket joint "E." A plastic spring "H" provides for adjustment of the stroke of "B" with respect to "A." Knee flexion of 45 deg. produces about 20 deg. of dorsiflexion.

The mechanics of the prosthesis gives the possibility of improved function, including automatic lock of the knee joint and possible adjustment of the foot when it is not on center.

The prosthesis shown in Figure 1 is complete including a device for rotation of the foot about the vertical axis during stance phase. The "rotator" uses a spring of Neoprene installed below the knee joint.

To analyse knee stability during the stance-phase and push-off, we consider a hip-moment, M_H, of zero (Fig. 4), and consider as an index of prosthesis stability the length "E," which is related directly to the ratio between the moment necessary to flex the knee joint and the applied load.

The magnitude of "E" corresponds, in a single-axis-type prosthesis, to the distance between the

Fig. 1. Prosthesis Rizzoli SC-75.

[1]Dr. Ing., Direttore Tecnico of Officine Ortopediche Rizzoli, Bologna, Italy.

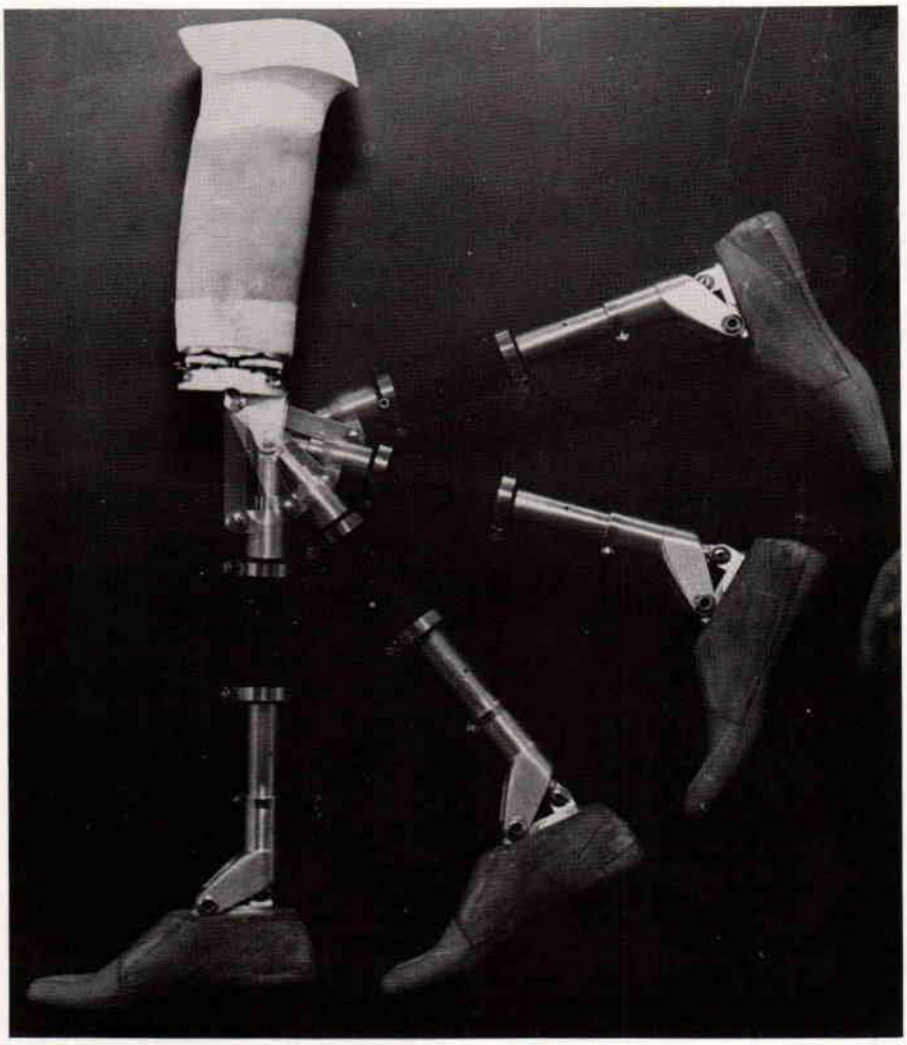

Fig. 2. SC-75 mechanism in different positions of flexion.

weight-bearing line and the center of the knee axis (Fig. 4). In a polycentric-type knee, "E" is measured from the instant center of rotation to the weight-bearing line.

In the SC-75 the moment at the knee joint is obtained by summing up the moment given by the axial load on the joint and the moment transmitted from the ankle joint to the knee joint. The second moment is a function of the distance between the center of joint "F" and the weight-bearing lines (Fig. 5).

The moment is a function of the distance "C" between the center "F" and the resultant floor reaction "R" and thus is dependent on the geometry of the ankle and knee-joint axes. Therefore it is possible to predict the function of the total knee assembly in providing better stability.

In comparison to the traditional prostheses, a further variable has been added which makes it possible to obtain a desired function of "E" to give the proper resisting moment.

The stability of the knee joint in traditional single-axis prostheses when $M_H=0$ increases gradually with the movement of loads from the

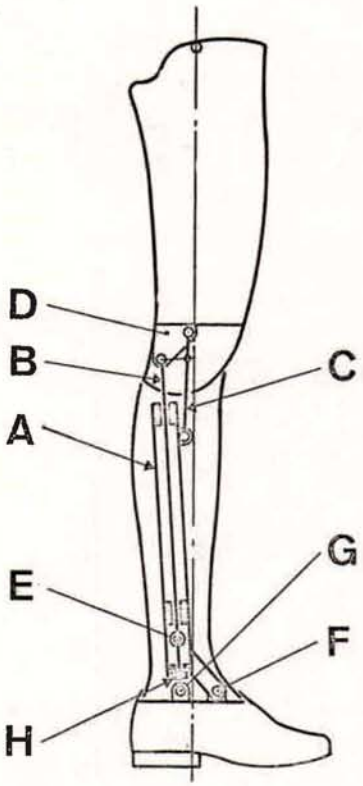

Fig. 3. Schematic diagram of the SC-75 prosthesis.

heel to the ball of the foot. This causes "overstability" and therefore difficulty in initiation of flexion at push-off just before the beginning of swing phase. This undesirable characteristic can be reduced with a higher instant center of rotation, bringing it nearer the center of the hip joint (*1*). Devices which provide this possibility constitute a remarkable contribution to the improvement of walking, but only amputees with stumps able to exert sufficient moments of hip extension can profit from these devices because the stability is controlled by this moment, and the value of "E" is very low or possibly is negative.

With regard to the present SC-75 prosthesis, we have obtained, by a particular geometry of the joints of the mechanical structure, a design which is characterized by a high value of stability at heel strike and during midstance phase, and of which, as the load moves towards the ball of the foot, the stability becomes zero and subsequently negative.

In this way, control by the stump at push-off is possible without compromising stability in other conditions. For a better explanation of this concept, please refer to Figures 4, 5, and 6 which show the values of eccentricity "E" and distances "C" which are determined by the projection of the ankle joint on the floor and the weight-bear-

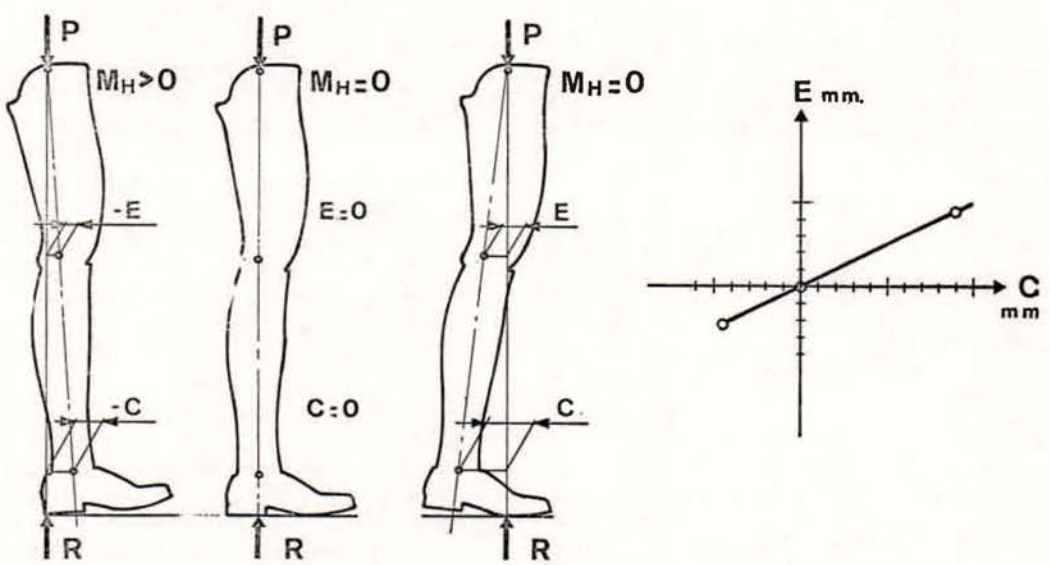

Fig. 4. Diagram of the stability index "E" for a single-axis type of prosthesis. C is the distance of the load axis from the ankle joint; E is the distance of the same axis from the center of rotation of the knee joint.

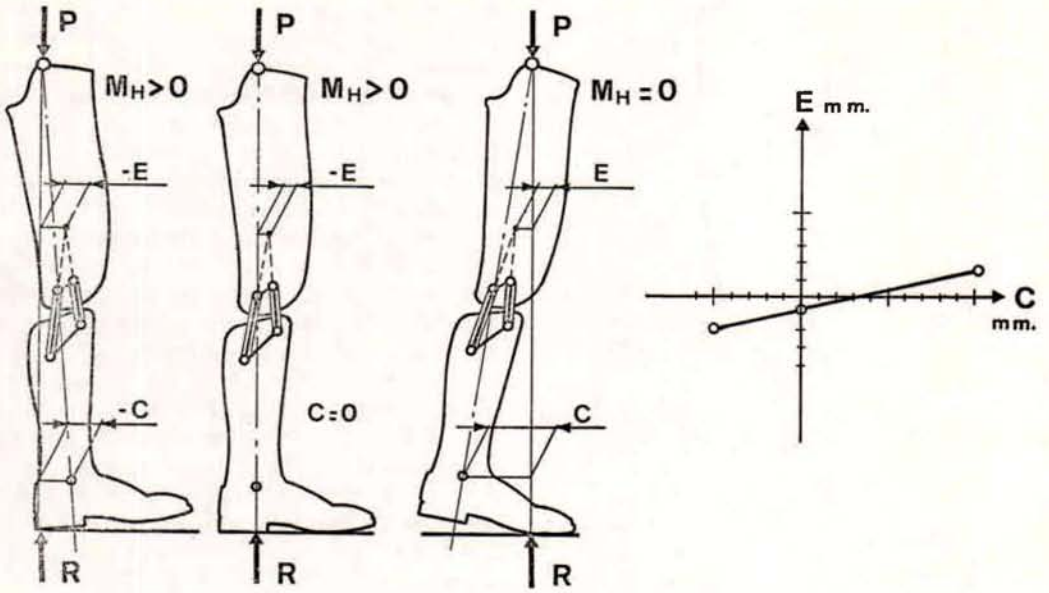

Fig. 5. Diagram of the stability index "E" for a polycentric type of prosthesis. See Fig. 4.

ing line. The diagrams show clearly the characteristic of the prosthesis SC-75 which results in optimum stability and ease of transition from heel contact to swing phase.[2]

Kinematics during the swing phase of the prosthesis has been carefully studied also. The design is so as to have the centers of relative motion between leg and socket (Fig. 7) quite similar to the physiologic centers in the knee in order to give an aesthetic and natural gait (*2*), but above all we obtained a very important characteristic necessary for correct, steady walking; dorsiflexion of the ankle in coordination with knee flexion. In this way the foot maintains a position parallel to the floor during the entire swing phase. The paths of the extreme points of the prosthesis give maximal distances from the articular coxofemoral center, lower than that provided by other types of single-axis prostheses (Fig. 8).

All this is helpful to the amputee during swing phase while negotiating uneven ground, because he can avoid lifting the hip thus acquiring an easier gait. Locking in full extension, at the end of the swing phase, is made possible by a plastic spring that is compressed at the beginning of the weight-bearing phase. Dynamic loads on the heel increase the joint stability. The same spring acts to limit the stroke in flexion to 115 deg.

A complete investigation on the validity of this prosthesis compared with traditional designs, and above all an investigation to define the problems of optimum conditions in the designing and manufacturing of a prosthesis, is going on in cooperation with Dr. Ing. Leo of the Centro di Automatica of C.N.R. in Rome and with Dr. Ing. Cappozzo of the Institute of Human Physiology at the University of Rome. The object of this research is to point out the kinematic and dynamic conditions of the prosthesis in different phases of walking and to try to determine which parameters are to be examined to define better the degree of perfection obtained in the different types of prosthesis.

[2]A different criterion in evaluating our prosthesis SC-75, particularly useful in a graph-design of the prosthesis, consists of determining the instant center of rotation of the foot in reference to the socket. This will be situated on the straight line that joins the instant center of the knee joint and the center of the joint "F." The distance from "F" will be

$$r = \frac{r_1}{1 - \frac{a}{b}}$$

in which r_1 is the distance between the knee instant center and F, a and b are the distances from the axis of tube B, from the knee instant center and from F. In this device the said instant center of rotation will be in a more elevated position and backward with respect to the coxofemoral joint.

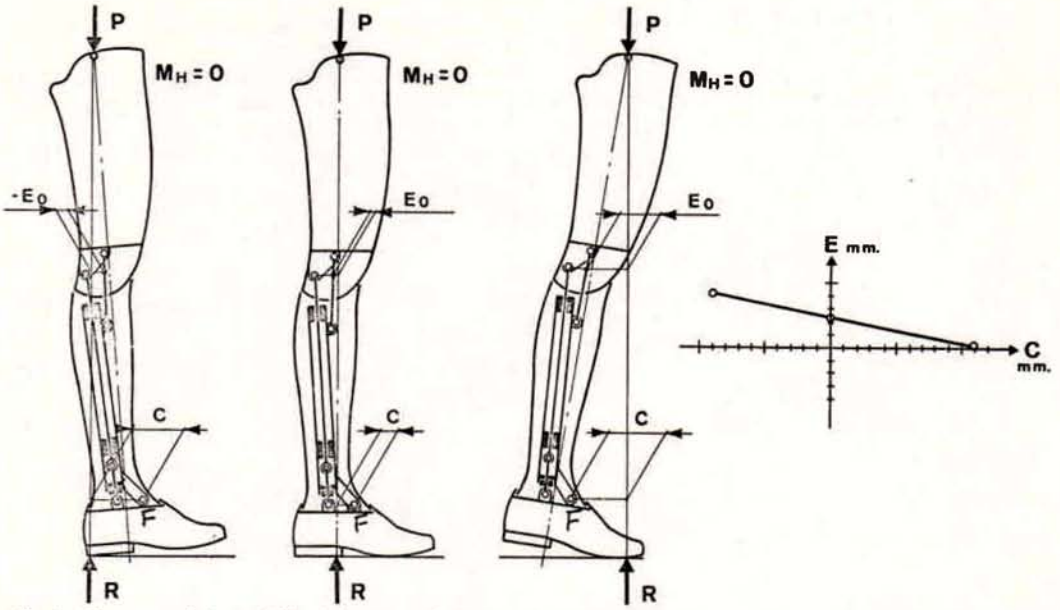

Fig. 6. Diagram of the stability index "E" in SC-75 prosthesis. The values of C are always the distances of the load axis from the ankle joint, whereas the value of E is the result of an expression like: $E = Eo - \Phi C$ in which Eo is the distance from the load axis to the articular center of the knee and the values of ΦC are such that they cancel E when the load axis moves to the ball of the foot.

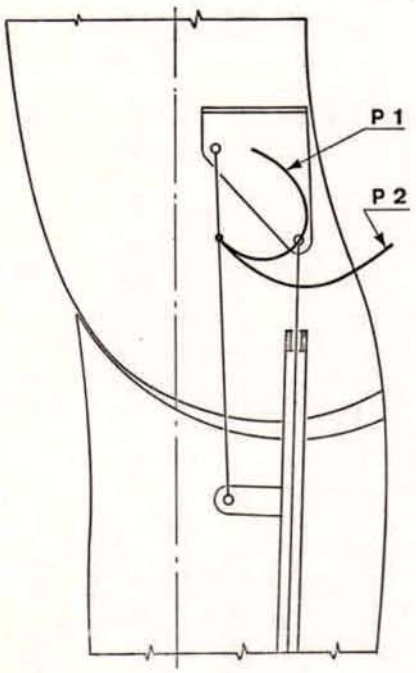

Fig. 7. Centers of relative motion between leg and socket.

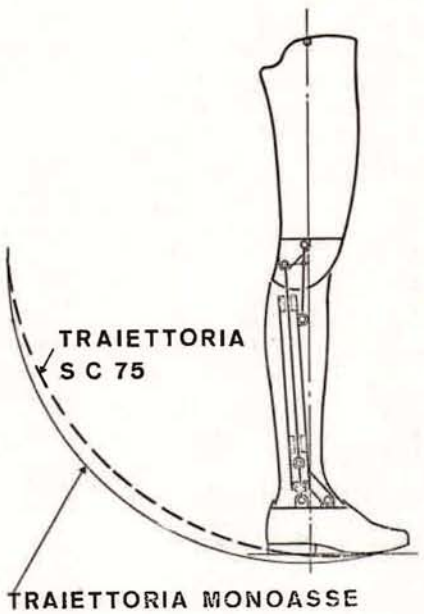

Fig. 8. Comparison of the trajectory of the external point of the foot of a single-axis prosthesis with that of the external point of SC-75. The dorsal flexion of the foot increases the distance from the floor during swing phase.

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AN ORTHOSIS FOR MEDIAL OR LATERAL STABILIZATION OF ARTHRITIC KNEES

S. Cousins[1] and James Foort[1]

THE PROBLEM

An orthosis may offer the only relief possible for patients suffering gross damage of the knee because of certain forms of arthritis. An orthosis is indicated when the knee requires support either for relief of pain or to give stability for preservation of function. These conditions may occur when a surgical approach is not appropriate, and when physiotherapy alone is not sufficient.

Because clinicians at the Canadian Arthritis and Rheumatism Society in Vancouver, Canada, wanted to be able to identify cases that might benefit from use of an orthosis, and then to be able to provide them with an appropriate orthosis, a knee clinic was established which included the authors as the Engineering Component. We were struck by the gross nature of the disabilities from which the patients were suffering, and found it difficult to sift out just what was pertinent in order to identify what was needed. The average age of the patients was about 70. They often had gnarled hands and feet. Sometimes, other joints were affected. The difficulties presented often had been present for a long time. Some cases had seemed to fall apart rapidly, often presenting insurmountable problems for those providing therapeutic assistance. The problems seemed to be in such a mixed-up array, as our untrained eyes saw them, that we were forced to isolate some specific malfunction as a point of entry into the problem of orthosis design.

It was decided that mediolateral knee instability would be the most likely malfunction to tackle for development of an orthotic solution.

THE APPROACH

Initially, we were observing only in order to become more familiar with those elements we could identify as common to the sorts of knee disabilities seen in the clinic. Soon we were willing to enter into a simple program of activities which would allow us to develop in whatever directions it took us. Two approaches became apparent. First, conventional "long leg braces" used on some of the patients included plastic laminate sleeves around the knees. Secondly, objective data were sought, initially through search of the literature, and subsequently by making direct goniometric measurements of motions about the ankle and knee in two planes as each patient walked. These data would be useful for defining the degree of motion occurring in the mediolateral directions, telling us when these motions were occurring in the walking cycle, and indicating how other joints were functioning in relation to the particular malfunction being observed. On the basis of practical application of known orthotic solutions and on the objective studies we had done on joint motion patterns, we outlined some criteria for orthosis design.

DESIGN CRITERIA

As we established our design criteria, we found that plastic laminate "splints" and conventional long leg "braces" were unsatisfactory solutions for the problem of mediolateral instability at the knee. Also, in the review of the literature we found only the "Michigan Brace" (*1*) to be designed specifically to control mediolateral knee instability for arthritics, and it, too, fell short of satisfying the criteria we had outlined. Because our own solution related directly to these criteria, most of them are satisfied by the design.

The criteria that were established and how the new design meets them are indicated in Table 1.

THE SOLUTION

A schematic diagram to show the application of forces supplied by the orthosis during the walking cycle for the patient with "knock-knee" is

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TABLE I.

	Criteria	*Results*
1.	Light weight	Under a pound—.450 Kgms.
2.	Weight evenly distributed	Note symmetry of orthosis in Fig. 2
3.	Minimum area of contact with the body	Approximately 100 sq. in. (645 sq. cms)
4.	Minimum interference with surroundings (clothes, furniture, etc.)	Not as good as with conventional or Michigan orthosis, but better than with the plastic-laminate splint.
5.	Force application optimal for comfort	Improvement in size and shape of knee support is required, other elements are satisfactory.
6.	Wanted joint motions uninhibited as unwanted motions are blocked	Hip and ankle function is normal. Knee flexion is not inhibited. There is some restraint of knee extension.
7.	Acceptable cosmesis	Better than previous orthoses except for the plastic splint.
8.	Easy to fit to the patient	Prefabricated, only strap adjustments that include sewing and riveting are required.
9.	Easy for the patient to don	Easier than other orthoses, but needs improvement.
10.	Adjustable	The adjustment strap which controls the position at which medial or lateral displacement is blocked allows the patient to set this for himself.
11.	Safety for the patient	Pinch-proof. Catastrophic failure unlikely.
12.	Reliability	Will not jam or catch, or disengage in action.
13.	Minimal costs	Less expensive to make and maintain than the other types of knee orthoses used for this purpose.
14.	Minimal skills needed for application	Prefabricated in two types and one size: valgus right side is also varus left side. Sometimes a heat gun is needed for adjustment of the plastic cuffs.

given in Figure 1. The force at the knee is provided to balance the forces required at mid-thigh and shank on the opposite side to keep the knee joint in a more normal mediolateral position. Forces would be applied in the opposite direction for correction of a varus, or bowlegged, deformity.

Fig. 1. Schematic showing application of forces by the experimental orthosis.

The orthosis used routinely is shown in Figures 2 through 6. The subject in this instance is a person without physical problems about the knee. Figure 2 shows the position of the orthosis for a knee which is unstable toward the medial side (valgus). The parts that stabilize the knee are the thigh and shank cuffs, which are made of transparent, rigid PVC; the knee cuff which is made of leather; and the telescoping tube assembly that connects the three cuffs. The knee cuff is triangular in shape, and is anchored to the shank cuff on one apex of its base and to the thigh cuff on the other apex of its base. The apex remaining is anchored to the telescoping tube assembly by a strap that crosses behind the knee. Thus, the needed forces are provided. Figure 3 shows how the orthosis would be used on a person with a later-

Fig. 2. Anterior view of the experimental orthosis applied for medial instability.

Fig. 3. Anterior view of the experimental orthosis applied for lateral instability.

ally displacing knee. Figure 4 shows the action of the telescoping assembly as it shortens to permit knee flexion, and displaces backward to relax the knee cuff. The plastic fixtures at the ends of the telescopic beam allow the plastic thigh and shank cuffs to tilt in any direction on the telescoping assembly, and the tubes of the assembly are free to rotate with respect to each other. Figure 5 shows how the knee cuff fits against the side of the knee to hold it against movement. Figure 6 shows how the strap links the telescopic assembly to the thigh and shank cuffs. A waist-band system secures the orthosis to the limb. Inside the telescoping tube an elastic cord prevents inadvertent separation when the orthosis is not being worn.

The universal hinge at the thigh cuff is shown in Figure 7. An identical unit is used at the shank cuff. The joint consists of a polypropylene molding with a threaded metal insert, a nylon bolt that screws into the insert to attach the joint to the cuff. The bolt is secured to the cuff by a nylon nut in such a way that the head of the bolt is inside

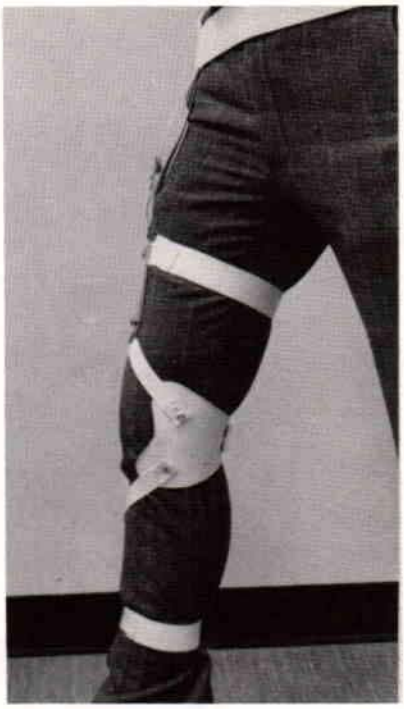

Fig. 5. Medial view showing placement of knee cuff when installed on medial side.

Fig. 4. Lateral view of experimental orthosis to show action of the telescoping assembly.

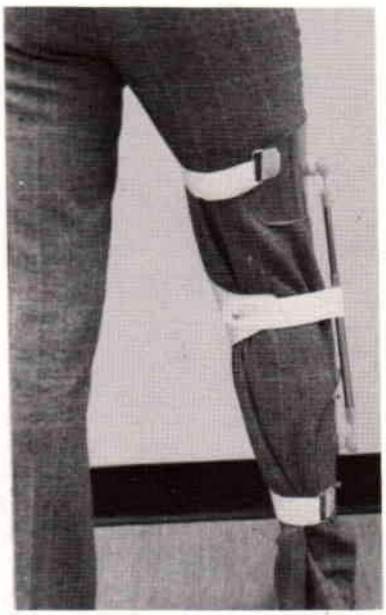

Fig. 6. Posterior view of experimental orthosis to show how the strap links the telescoping assembly to the thigh and shank.

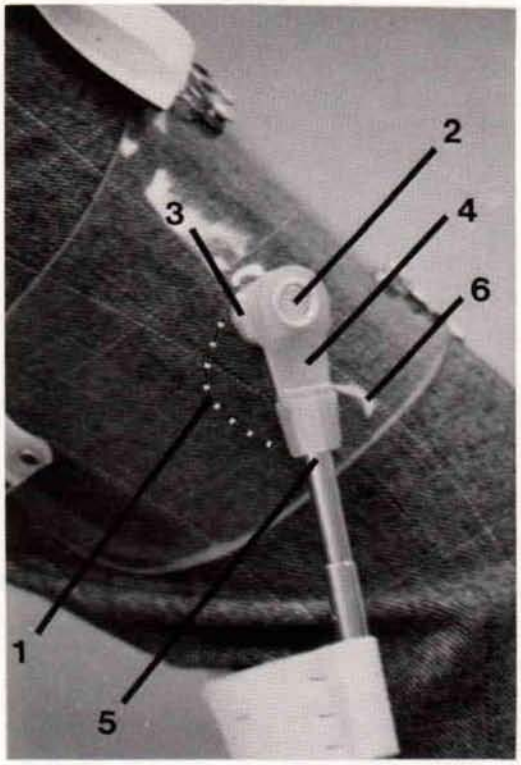

Fig. 7. The universal hinge used to fasten the telescoping assembly to thigh and shank.

the cuff, countersunk, and the lock is outside the cuff. Extra bolt is cut off after the polypropylene pivot is set at the best distance from the cuff by turning it on the threads of the bolt. The thin section of the polypropylene pivot allows abduction-adduction motions to occur (plastic spring effect). The tubing is "press fitted" into the base of the polypropylene pivot. A hole drilled through to the tubing along its axis allows installation of the elastic cord, which is secured by a knot. The tubes can rotate relative to one another. When not on the limb, the orthosis goes limp like an unheld puppet. When properly in place it stiffens to its task.

THE FUNCTION

Joint motion studies taught us that the knees we were trying to brace needed support mainly during the extension part of stance phase. Thus, we felt that what was needed was an orthosis which would exert an intermittent force active during extension, but not active during all other parts of the walking cycle, and completely inactive during sitting. When the knee is flexed, the telescoping beam of the orthosis moves posteriorly so that the tension on the knee cuff is reduced. Conversely, as the knee extends, tension in the strap is increased to give the required support. Telescopic action of the beam, coupled with the pivoting, flexing-joint system at both ends and the relative rotation of the tubes to one another, removes all but the constraining forces that are needed, except for contraint of extension—a factor that can detract from effectiveness of the orthosis for some patients. Quadriceps strengthening has been given for some of these cases, and others have been considered to be unsuitable users. If there is a tendency toward hyperextension at the knee, then this feature can be beneficial.

APPLICATION IN THE CLINIC

The procedure we use is nearly optimal. The patient is seen in the clinic, and a prefabricated orthosis is used as part of the assessment procedures. Various adjustments are made to control straps until it is necessary to adjust only the knee cuff strap so that the time difference can be determined between when the tension is on and when it is off. If, with tension on the strap, the patient indicates improvement, then he is held for completion of the fitting, which is done either during or immediately after the clinic. Improvement is indicated by a positive response from the patient and, more objectively, evidence of increased speed of walking or walking with the cane held up.

The biggest problem in adjusting the orthosis lies in getting the straps of the knee cuff at the required length. The straps are permanently fastened to the shank and thigh cuffs when they are set correctly. The posterior strap to the telescoping beam is left adjustable. After everything is fixed, and the excess part of the nylon bolt is cut off, the patient practices applying the orthosis and adjusting the back strap for a few times. If there are difficulties of remembering, or special tabs are needed, these matters are dealt with at subsequent therapy sessions. The whole process seldom takes an hour per patient, excluding time in the clinic.

RESULTS

Approximately 24 orthoses have been fitted to date. The longest period of use is just over a year (the original prototype). Approximately half the cases have had their orthoses for three months or more. It is the positive response of the patient and the increased speed of walking that give the most important clues to success. The deformity is not corrected necessarily, but comfort and function are improved because the orthosis prevents the knee from sagging into a painful position. Conditions which cannot be handled well using this orthosis include rotational instability and mediolateral instability coupled with knee extensor weakness.

More detailed results will be reported by clinicians responsible for patient care. Similarly, details of construction will be reported through a future report or publication by us and those involved in fabrication.

CONCLUSIONS

We have added not only a new orthosis to the list but a new principle through the use of the telescoping beam and the associated joint system. We already know that we can apply this concept to a number of orthosis types. In mind is a design which incorporates two such beams so organized that the orthosis will offer control of flexion and extension instabilities. We can envision the use of such beam-joint structures for improvement of the "Milwaukee Brace." We are in the early stages of applying it to fracture bracing. Our hope is that the processes we have gone through and the design we have developed will add scope to the work of others similarly involved. Our own plans are to make permanent molds for fabrication of the PVC cuffs, to develop a plastic knee cuff, to improve fastenings, to instrument the orthosis for monitoring forces, and to expand field testing outside the home area until we have results on approximately 200 applications.

We have named the orthosis the C.A.R.S.-U.B.C.[2] Knee Orthosis Valgus-Varus.

ACKNOWLEDGMENTS

Teammates in this development include Dr. L. Truelove, Patricia McBain, Bruce Clark, Richard Hannah, Ronald Wassen, and John Hoare.

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THE "GRANGER" ORTHOSIS FOR RADIAL NERVE PALSY[1]

William Crochetiere, Ph.D.[2], Steven Goldstein[2],
Carl V. Granger, M.D.[3], and
Judith Ireland, OTR[3]

Lesions of the radial nerve result in a characteristic wrist- and finger-drop deformity. Because the synergic extensors of the wrist are paralyzed, an effort to grasp is accompanied by excessive flexion of the wrist joint. As a result, the patient finds it very difficult to grip an object.

To reduce the disability until such time as spontaneous recovery occurs or corrective surgery is performed, the hand may be fitted with an orthosis to fix the wrist in a position of functional extension to allow the active muscles to function, and to keep the muscles at their proper lengths to avoid capsular contraction. Several orthoses have been reported in the literature which incorporate springs or elastics to hold the hand in extension. The orthosis reported here, which was designed by C. V. Granger, makes use of gravity instead of a spring force.

FUNCTION OF THE ORTHOSIS

The splinted hand in positions of extension and flexion is shown in Figure 1. Note that the horizontal support extends from the forearm cuff to the proximal phalanx, then angles upward to provide a fixed center of rotation for the finger stirrups. Because of the constraint imposed by the orthosis, as the finger flexors contract two things happen simultaneously: the wrist extends and the fingers flex, assuring a strong and complete grasp.

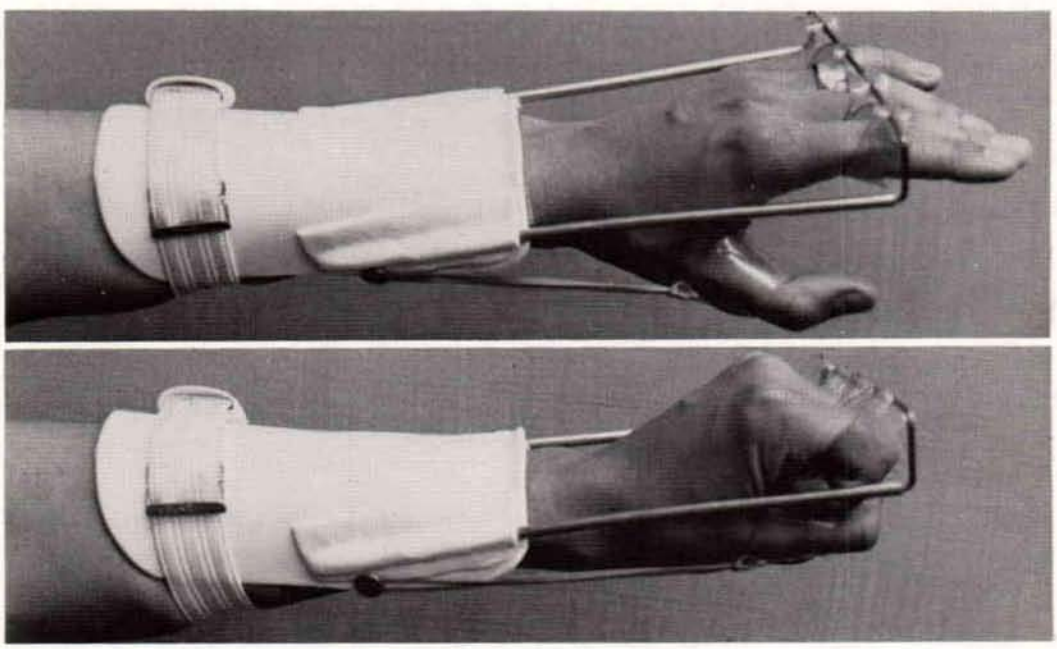

Fig. 1. Two views of the Granger hand orthosis.

[1]Supported in part by Rehabilitation Services Administration Grant No. 16-P-56800/1-10(RT-7).

[2]Engineering Design Department, Tufts University, Boston, Mass.

[3]Rehabilitation Institute, Tufts-New England Medical Center, Boston, Mass.

Conversely, when the finger flexors relax, the wrist flexes under the influence of gravity, and the fingers extend.

DESIGN SPECIFICATIONS FOR THE ORTHOSIS

Except for the lengths of the stirrups, all of the dimensions of the orthosis are determined so as to fit the size and shape of the hand. *It is very important, however, that the stirrup lengths be sized properly or the orthosis will not function properly.* If they are either too long or too short, the orthosis will not allow the hand to close completely. A kinematic analysis of the hand/orthosis combination which derives this principle is in the appendix.

To determine the optimal stirrup lengths, one must first of all measure the lengths A and B on the patient's hand as shown in Fig. 2a. Length A is measured from the center of rotation of the MP joint of the index finger to the point where the hand will be supported by the stirrup. Length B is measured from the center of rotation of the MP joint to the center of rotation of the wrist joint. The value B/A is then computed and entered on the graph in Figure 3. The corresponding value of L/A is then selected. Multiplying this value by the measured value of A then gives the optimal stirrup length for the index finger (L1). This same procedure is also repeated for the fifth finger to yield L2.

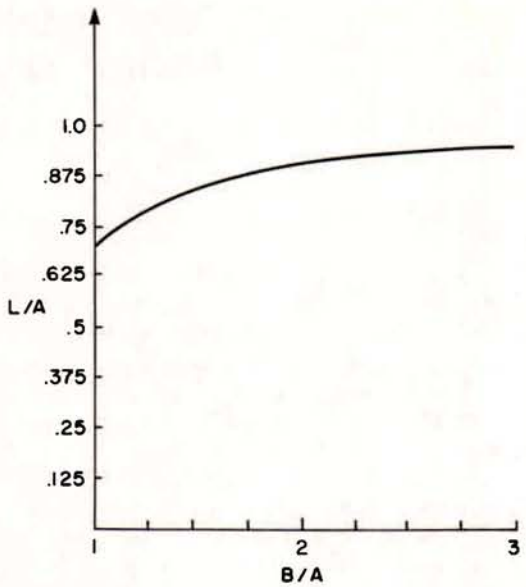

Fig. 3. Graph for determining optimal stirrup lengths (L).

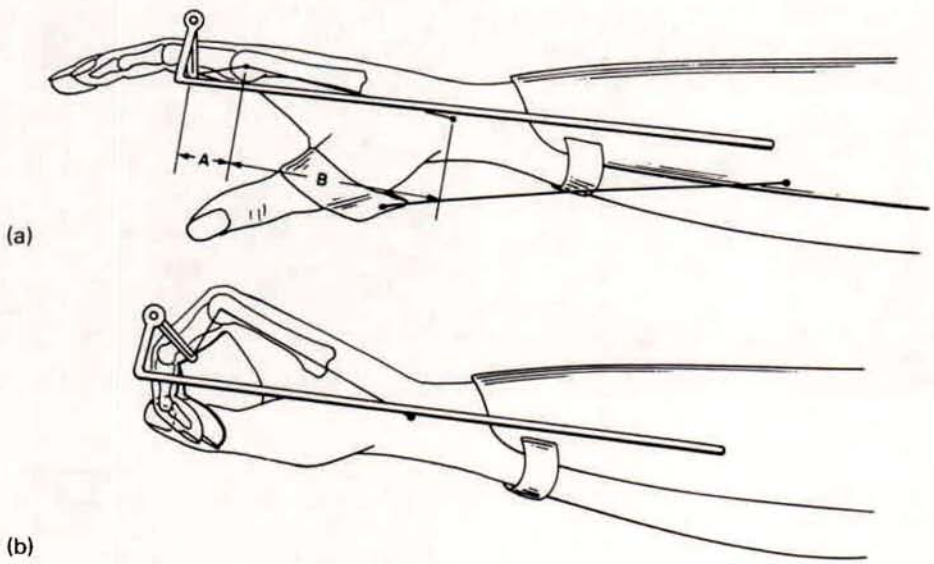

Fig. 2. Diagrams showing operation of the hand orthosis in extension (*top*) and in flexion (*bottom*).

Obtaining the stirrup lengths in this way yields the theoretically optimal values. Because of the difficulty in measuring and maintaining a high degree of accuracy, however, a more practical approach which gives very good results is to make the stirrup lengths about .9A (90% of length A). These values are then used to construct the orthosis by either of two methods as shown in Figure 4. In Figure 4a, individual stirrups are shown, whereas in Figure 4b the individual stirrups are replaced by a padded stirrup bar which supports all four fingers. In either case, the function of the orthosis is the same.

BENEFITS OF THE ORTHOSIS

This orthosis is most beneficial for a patient with radial nerve palsy simply because it allows him to function in a near-normal manner throughout the day. This activity level is an excellent way of maintaining normal range of motion, and good strength of those muscles unaffected by the radial nerve injury. A patient who is *active* while wearing the orthosis is even able to maintain normal thumb range successfully without wearing a special outrigger for abduction/extension. However, when necessary it can be added as an elasticized

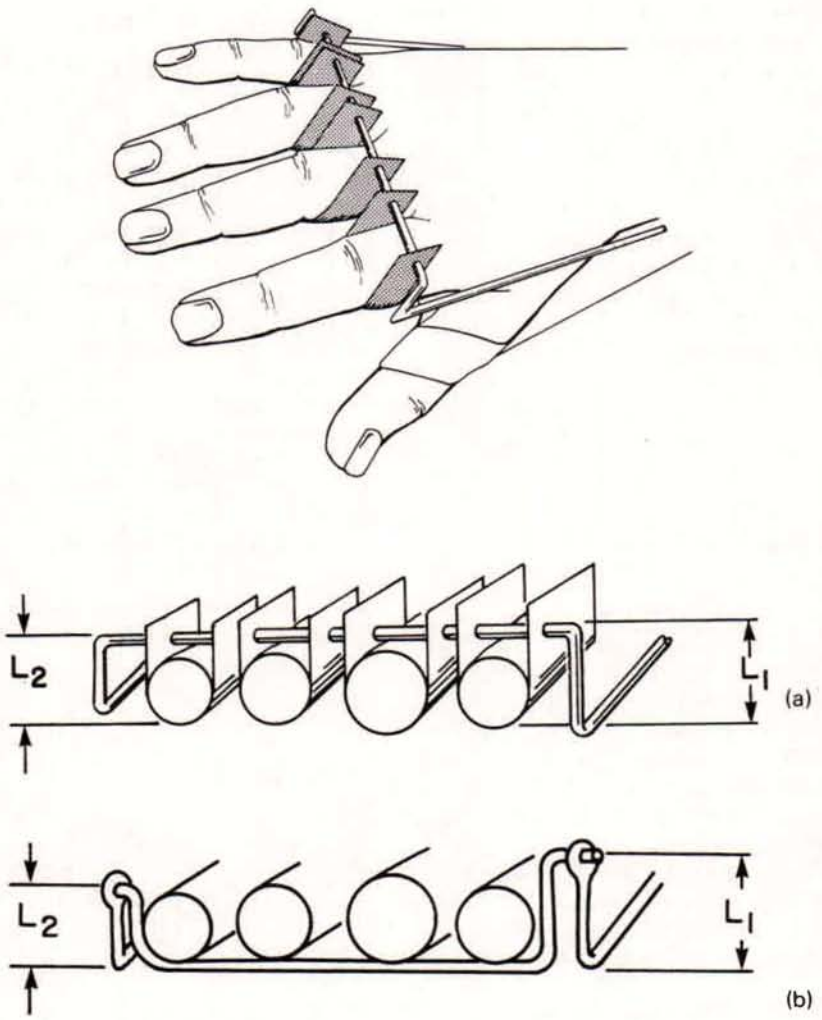

Fig. 4. Two designs for finger supports. Top and middle views show individual stirrups. The stirrup-bar design is shown in the bottom view.

band as shown in Figure 1a, without the need for an outrigger. The patient is able to flex and extend his fingers fully. He can oppose all fingertips to his thumb, and he can accomplish lateral pinch. The volar surface of his hand is uncovered completely and available for maximum sensory feedback. Unlike other dorsal "wrist drop" or dynamic splints, the outrigger on this splint is less bulky and does not present a problem in donning most clothes.

Patients have found that the orthosis allows considerable freedom and function at home or at work. The patient may work with water or other liquids. He is limited in wrist supination/pronation, but this can be compensated for at shoulder and elbow. To date, 11 orthoses of this type have been fitted and used successfully at the Rehabilitation Institute, Tufts-New England Medical Center hospitals.

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2. Thomas, G. B., *Calculus and Analytic Geometry*, Addison-Wesley, 1968.

3. Tyson, H. N., *Kinematics*, John Wiley & Sons, 1966.

APPENDIX

KINEMATIC ANALYSIS

The sketch shown in Figure 4 is a kinematic representation of the four-bar linkage which is formed by the hand-orthosis combination. The linkage is shown in positions of extension and flexion. Note that the approximation is made that the horizontal projections of links A and B are equal to A and B. A coordinate system has been added for convenience.

At full flexion, the angle between links A and B is 90 deg. The problem is to find the optimal link length L such that, when the MP joint is fully flexed, the lift of the hand will be a maximum.

Since link L rotates about a fixed center, its free end describes a circle described by:

$$X^2 + (Y - L)^2 = L^2 \qquad (1)$$

In the flexed position, links A and B form two sides of a right triangle. The hypotenuse has a length $\sqrt{A^2 + B^2}$ and forms an angle (α) with the horizontal. The coordinates of the free end of this member are given by,

$$X_1 = A + B - \sqrt{A^2 + B^2}\cos\alpha \qquad (2)$$

$$Y_1 = \sqrt{A^2 + B^2}\sin\alpha \qquad (3)$$

This point must be coincident with the free end of link L. Solving equations (1,2,3) simultaneously yields,

$$L = \frac{(A + B - \sqrt{A^2 + B^2}\cos\alpha)^2 + (\sqrt{A^2 + B^2}\sin\alpha)^2}{2\sqrt{A^2 + B^2}\sin\alpha}$$

letting $K = B/A$ and reducing,

$$\frac{L}{A} = \frac{1 + K + K^2 - (1 + K)\sqrt{1 + K^2}\cos\alpha}{\sqrt{1 + K^2}\sin\alpha} \qquad (4)$$

This yields a family of curves of L/A versus α for various values of K as shown in Figure 5. Notice that as B/A increases L/A also increases. Note also that, except for the lowest point on each curve, there are two values of α for each value of L/A. Referring back to the linkage diagram on Figure 4, the single value for α corresponds to the configuration where link A is tangent to the circle described by link L. It is, therefore, impossible for the linkage to assume the larger values of α unless the angle between links A and B (the MP joint angle) becomes less than 90 deg., a physical impossibility. The angle which corresponds to the lowest point on the curves of Figure 5, therefore, represents the maximum lift of the hand. This may be determined analytically by differentiating equation (4) with respect to α and setting it equal to zero (horizontal tangent).

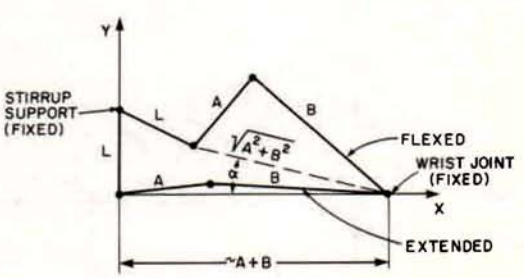


Fig. 5. The hand orthosis combination viewed as a linkage.

$$\frac{d(L/A)}{d\alpha} = \frac{\sqrt{1+K^2}\sin\alpha\,[(1+K)\sqrt{1+K^2}\sin\alpha] - [1+K+K^2-(1+K)\sqrt{1+K^2}\cos\alpha]\sqrt{1+K^2}\cos\alpha}{(1+K^2)\sin^2\alpha} = 0$$

$$\alpha = \cos^{-1}\left[\frac{(1+K)(1+K^2)}{(1+K+K^2)\sqrt{1+K^2}}\right] \qquad (5)$$

This expression was evaluated on the digital computer for values of K ranging from 1 to 3 in increments of 0.1. The results are plotted in Figure 6. This curve of α versus B/A represents the maximum value that α can attain for the given value of B/A. The hand will have this maximum lift only if the proper value of L/A is selected according to equation (4). This expression was also evaluated on the digital computer using the above values of α, and the results are plotted in Figure 7.

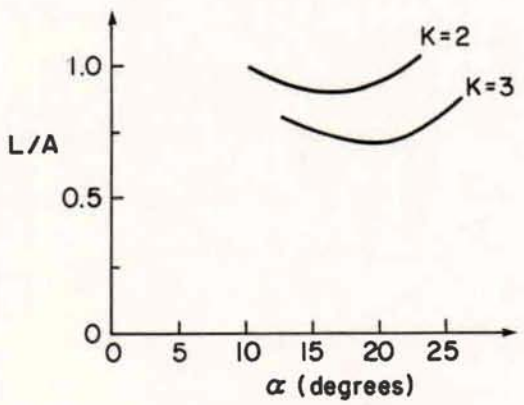


Fig. 6. Graph showing that the lift angle (α) is bivalued except for the minimum value of L/A, with K = a constant.

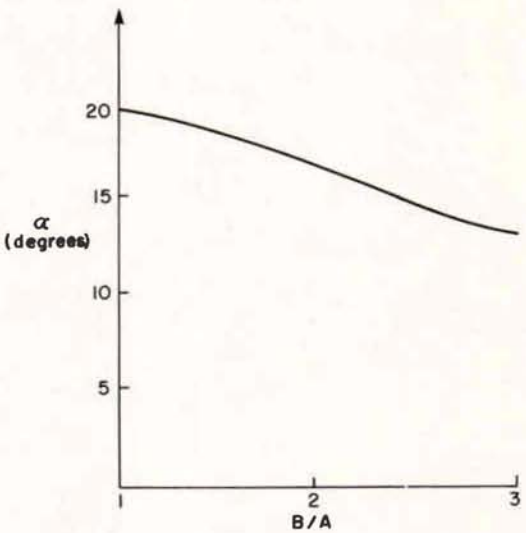


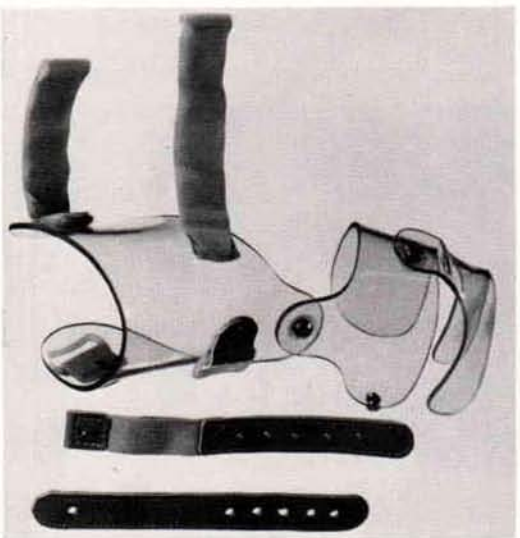
Fig. 7. Maximum possible lift angle (α), for differently proportioned hands (B/A).

Orthotics and Prosthetics, Vol. 29, No. 4, pp. 33-35, December 1975

AN ORTHOSIS FOR CORRECTION OF ULNAR DEVIATION

Michael Danisi, C.O.,[1] and
Anthony Altobelli[2]

It is our purpose to present an orthosis designed not only to apply continuous forces for the correction of ulnar deviation of the hand, but also to include an adjustable feature, so that the desired position, when obtained, can be held statically for maintenance of the correction.



The orthosis (Fig. 1) is designed for either static or dynamic control of ulnar deviation; a change from one type of control to the other can be accomplished quite simply.

The body of the device is fabricated of 1/8 in. thick Nyloplex—a transparent, rigid, thermoplastic with exceptionally high impact strength. The Nyloplex, which can be formed at a temperature of 250 deg. F, is light and inexpensive. It may be formed directly over the patient's hand and forearm, when a protective stockinet is placed over the contact areas of the skin. The material is brought to a malleable temperature (about 250° F), either in an oven or, more simply, with a heat gun, without precise control of temperature. The Nyloplex is shaped directly on the patient's hand and forearm.

The critical areas of fit are over the dorsal aspect of the carpal joint and over the ulnar aspect of the fifth metacarpophalangeal joint (Fig. 2). The Nyloplex sections are joined in these areas by a bushing and screw with a Teflon washer.

The rigid control strap shown in Figures 3 and 4 may be interchanged with the elastic strap shown in Figure 1. The strap, whether rigid or elastic, is attached to the radial aspect of the orthosis with metal truss studs, and is removable and adjustable (Fig. 5).

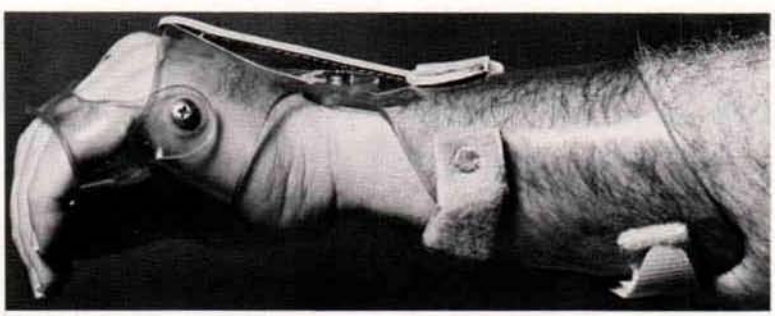


Fig. 2.

The Nyloplex is transparent, allowing pressure

[1]Supervisor, Orthotics Laboratory, Veterans Administration Prosthetics Center, 252 Seventh Avenue, New York, N.Y. 10001.

[2]Supervisor, Shoe Laboratory, VAPC.

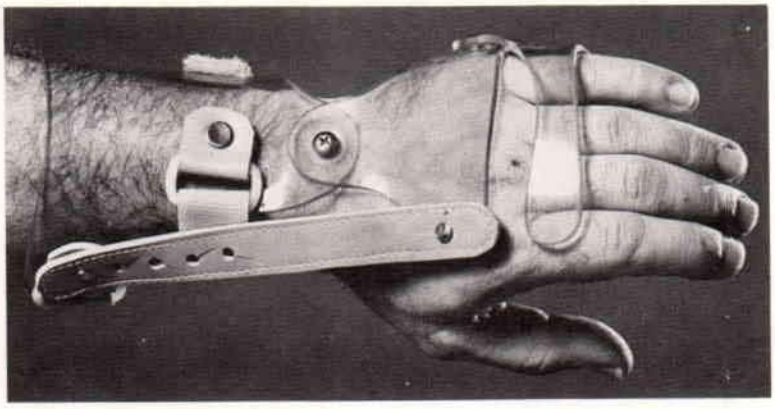


Fig. 3.

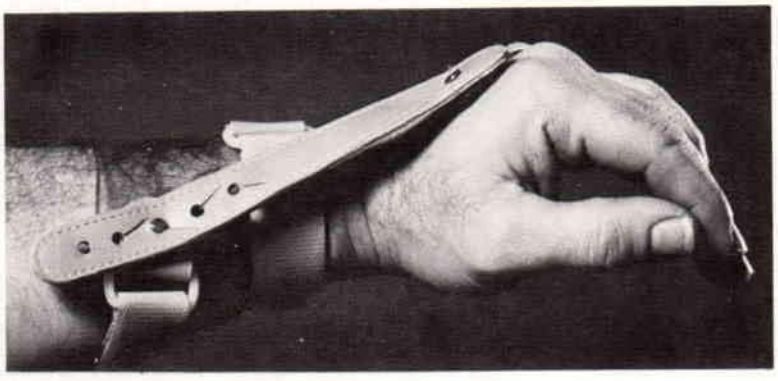


Fig. 4.

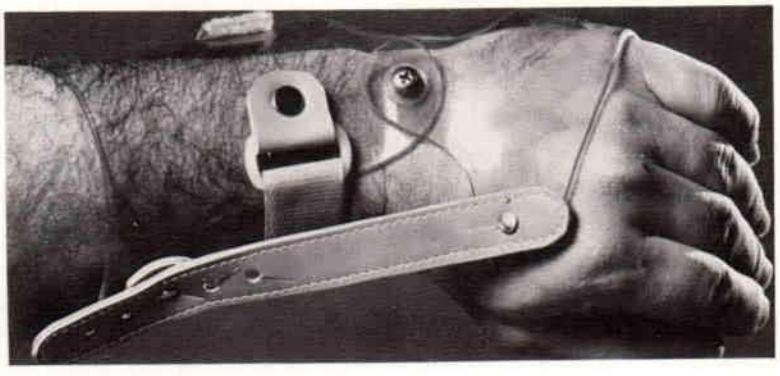


Fig. 5.

areas to be observed, and it is cosmetically acceptable. In its functional position, when elastic control straps are employed, the orthosis permits a sufficient range of joint motion to allow the patient to perform many of his activities of daily living. Finger and thumb motions are permitted (Fig. 6). When the nonelastic straps are used, a fixed supportive position can be obtained. The amount of wrist dorsiflexion is adjustable through a range of 5 to 20 deg., and templates can be prefabricated in small, medium, and large sizes.

SUMMARY

An orthosis has been presented which was designed to provide active correction of ulnar deviation of the hand. The orthosis includes a feature which will permit passive maintenance of the position achieved.

ACKNOWLEDGMENT

The authors wish to thank Dr. Gustav Rubin, Orthopedic Consultant, VAPC, for his advice and assistance is preparing this paper.

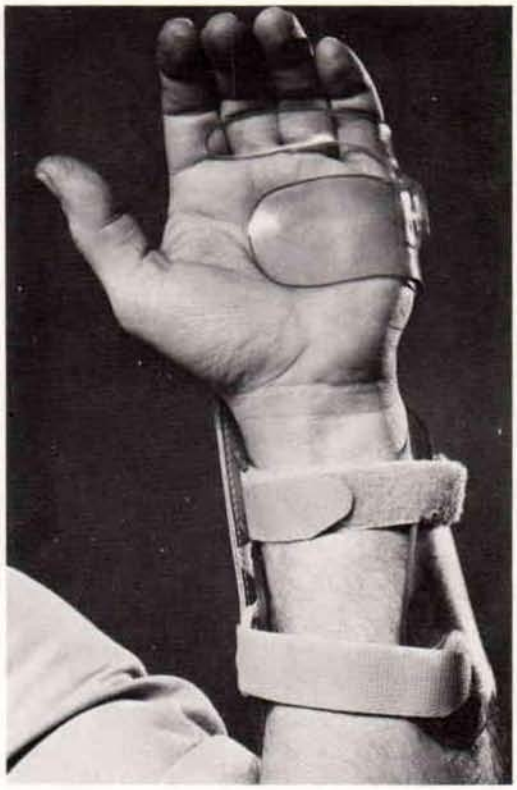


Fig. 6.

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A BELOW-KNEE WEIGHT-BEARING, PRESSURE-FORMED SOCKET TECHNIQUE

Robert F. Hayes, C. P.[1]

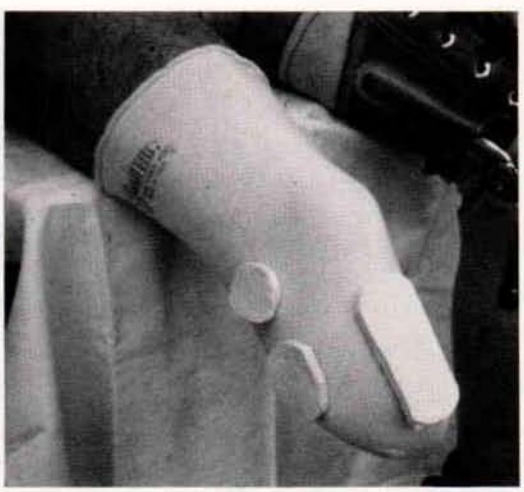
When my son was fitted for ski boots it occurred to me that we might use for fitting amputees some of the techniques used by ski-boot designers. The ski boot had an inflatable inner bladder. With the foot under weight-bearing, a conforming material similar to certain silicone compounds was injected into the bladder to give a perfect form-fitting in the attitude of weight-bearing.

The incentive to apply this technique to limb prosthetics was reinforced while I was watching a dentist take impressions for dentures. It then occurred to me that if this technique is sound for fitting ski boots and dentures the same principles should apply in lower-limb prosthetics.

The current method of fitting a below-knee amputee involves taking a negative cast in a non-weight-bearing condition, forming a positive model modifying it in size to preset dimensions by removing material to create pressure, and applying material to relieve pressure on the stump as appropriate. A socket is then molded over this model with the hope that, with small adjustments, it will fit.

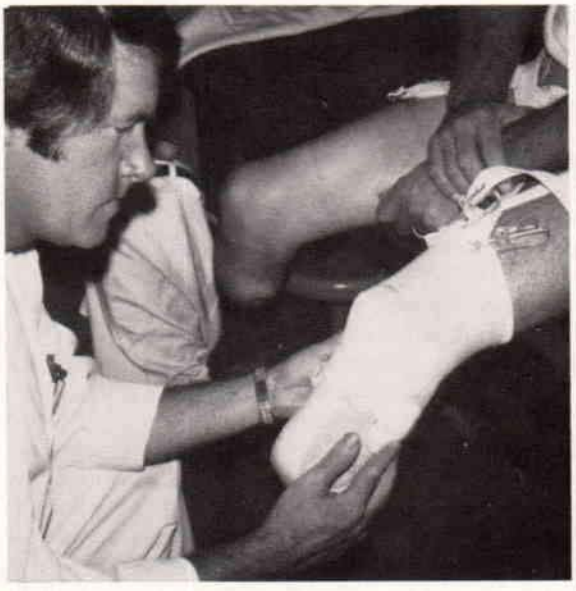
Wouldn't it be wiser to develop a socket that will adjust to and fit the patient, rather than fit the patient to the socket? In trying to answer this question, the procedure described here was developed.

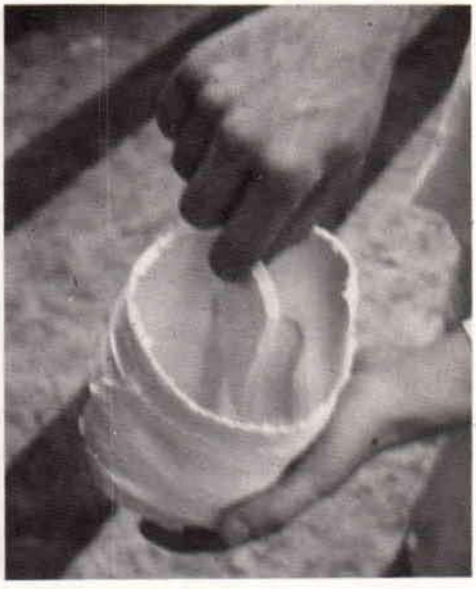
THE PROCEDURE



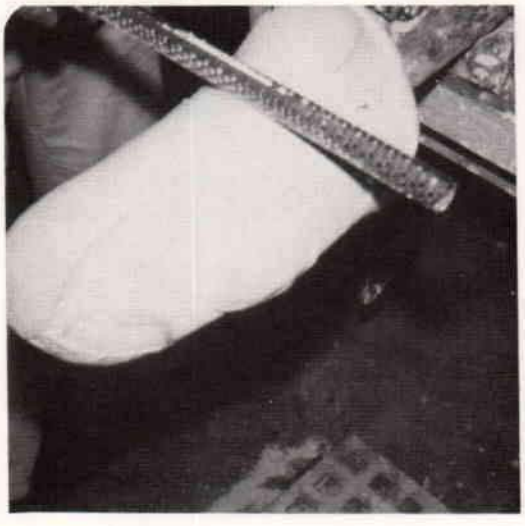
Measure the patient in the usual manner. Place a sheet of plastic wrap material, such as Saran, over the patient's stump to keep it clean of indelible pencil, and to make removal of the cast easy. If a wool sock is to be used, apply it, and then apply the plastic wrap. Apply a cast sock or tube gauze over the stump. Bond 1/4 in. felt over all pressure-sensitive areas—the crest of the tibia and the head of the fibula.

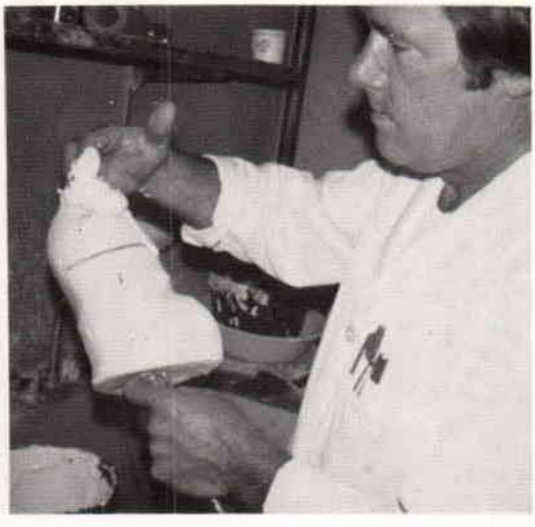
[1]Hayes Prosthetics, Inc., 1309 Riverdale Street, West Springfield, Massachusetts 01089



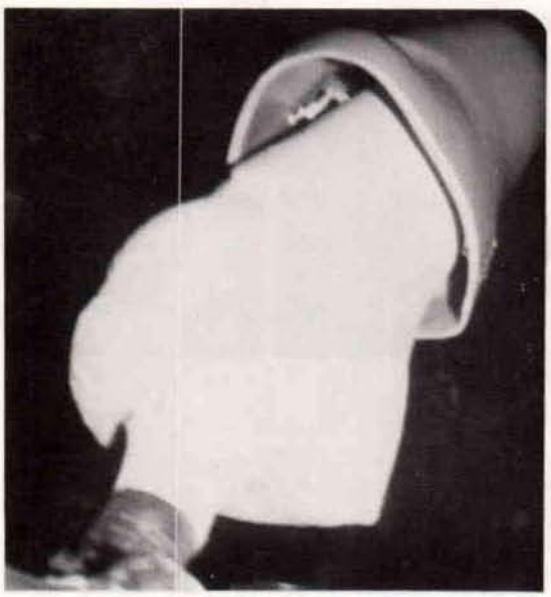


Using elastic plaster-of-Paris bandage 4 in. wide, wrap the stump in the usual manner, and reinforce with 3 in. or 4 in. wide regular plaster bandage. Remove the cast, and remove the tube gauze and felt buildup from negative cast.





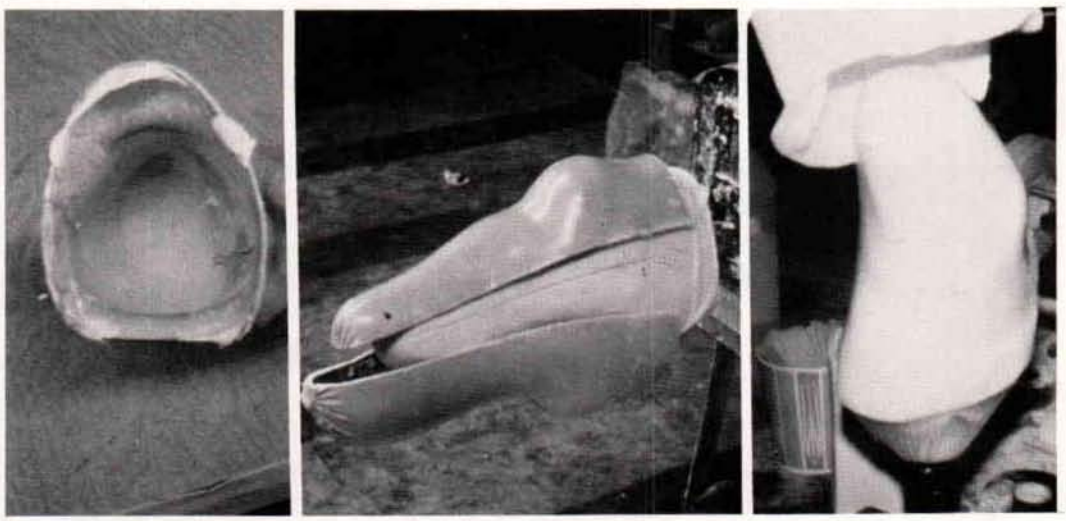
Pour the positive model, remove the negative cast, and modify in the usual manner, but do not touch areas that were covered with felt. Build up the distal end of the positive model at least 1 inch.



Make a check socket. This is a perfect application for vacuum-forming. Plaster bandages or laminates can, of course, be used. Drill two holes 1/4 in. in diameter in the distal end and rough up the inside surface of the socket. For the first fitting, apply the stump sock of choice, and place plastic "wrap" over the stump sock to act as a separator. An invaginated balloon will not work because it adheres to the alginate that is to be used later.

Mix about 1/2 pt. of dental impression cream or alginate which is more economical. Pour about 1/3 of the total amount in the distal part of the socket and, with a spatula, spread the rest around the remaining surface of the socket. It is necessary to work quickly at this point.

Place the socket on a fitting stool adjusted for height. Use some sort of pad to prevent slipping, and cover the drilled holes in the socket with your thumb and forefinger. Have the patient place his stump in the socket. Let the alginate escape through the distal holes until the patient is lowered into the socket to the proper level, at which time the holes are blocked. Alginate will now escape along the proximal brim of the socket.



As soon as the alginate has set up, remove the stump from the socket and immediately fill the socket with plaster. The rigid socket and alginate are removed by using a cast cutter. The mold resulting is a perfectly smooth, pressure-formed, positive model that can be used in any method of fabrication desired.

When this technique is used, patients can be fitted with sockets without soft liners.

Only a minimal amount of additional time is required. I feel that the technique allows better fitting of "problem" stumps, and that it may be used as a routine procedure to advantage, especially in central fabrication systems. Vacuum-forming procedures recently introduced make this approach to fitting even more attractive.

Orthotics and Prosthetics, Vol. 29, No. 4, pp. 41–51, December 1975

THERMOFORMED ANKLE-FOOT ORTHOSES[1]

Melvin Stills, C.O.[2]

Clinical and research groups have for many years felt that the use of plastics in orthotics had great potential for the development of improved designs. Attempts to use polyester and epoxy laminates that were adopted by prosthetists in the fabrication of artificial limbs more than 20 years ago largely met with failure when applied to orthoses that were subjected to repeated bending loads, owing no doubt to the relatively short fatigue life of the polyester and epoxy resins.

However, with the availability of thermoplastics such as polyvinyl chloride, polyethylene, and polypropylene in sheet form, and with good resistance to fatigue, a number of new designs emerged in the 1960s (*1*). Most of the new designs were ankle-foot orthoses for use with "drop foot."

In 1970 the Committee on Prosthetics Research and Development (*1*) identified 11 ankle-foot-orthosis designs for treatment of "drop foot" that seemed to have advantages over the conventional "short-leg brace" which consisted of two metal uprights and two ankle joints, and were attached to a special shoe or to one that was modified extensively. The Committee requested the Rehabilitation Engineering Center at Moss Rehabilitation Hospital (*2*) to conduct laboratory and clinical studies of each of the 11 designs in order to provide clinicians with guidance in prescribing and applying the various devices.

The 11 orthoses (Fig. 1) were studied and the origins of their design are listed on page 42.

Where the need for an orthosis was minimal the VAPC shoe-clasp orthosis was found to be the most acceptable, especially in ease of application and cost (*2*). For more severe cases the TIRR molded polypropylene orthosis seemed to have advantage over the other designs. It is light in weight, almost unnoticeable, requires no shoe modifications, is relatively easy to fabricate and fit, and the amount of control provided about the ankle can be varied (*2*).

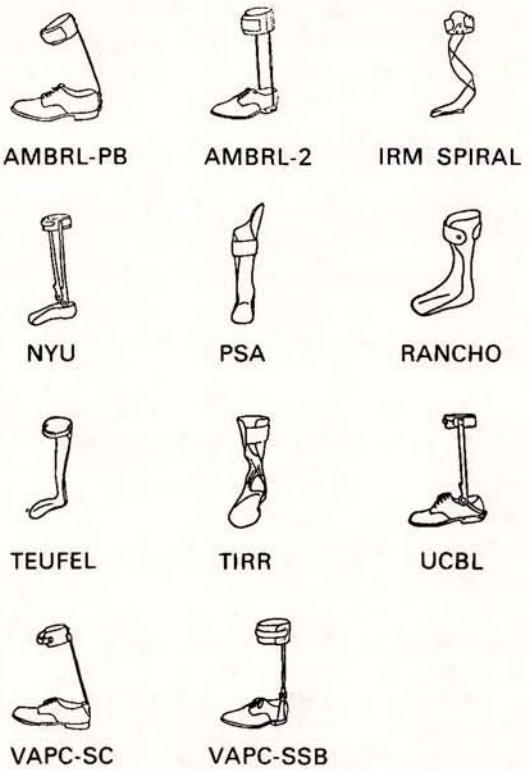


Fig. 1. Sketches of orthoses studied originally by the Research Engineering Center, Moss Rehabilitation Hospital.

While these studies were being carried out at Moss Rehabilitation Hospital, the Rehabilitation Engineering Center at the Ontario Crippled Children's Centre (Toronto, Canada) was developing methods and equipment for forming sheet thermoplastics for orthotic and prosthetic applications. Because of the encouraging results obtained in the orthosis evaluation project with the TIRR AFO and other molded orthoses, the Rehabilitation Engineering Center at Moss Re-

[1]This work was carried out with partial support of the Rehabilitation Services Administration, Department of Health, Education, and Welfare under Grant #23P-55518.

[2]Director of Orthotics, Krusen Center for Research and Engineering, Moss Rehabilitation Hospital, 12th Street and Tabor Road, Philadelphia, Pennsylvania 19141.

AMBRL Posterior-Bar Drop-Foot Orthosis AMBRL Two-Rod Drop-Foot Orthosis	U.S. Army Medical Biomechanical Research Laboratory Walter Reed Army Hospital Washington, D.C.
IRM Spiral Below-Knee Orthosis Posterior Solid-Ankle Orthosis	Institute of Rehabilitation Medicine, New York University, New York, N.Y.
NYU Insert Orthosis	New York University Post-Graduate Medical School, New York, N.Y.
RANCHO Polypropylene Drop-Foot Orthosis	Rancho Los Amigos Hospital Downey, California
TEUFEL Ortholen Drop-Foot Orthosis	Teufel, 7 Neckarstrasse, 189-191 Stuttgart, West Germany
TIRR Below-Knee Orthosis	Texas Institute for Rehabilitation and Research, Houston, Texas
UC-BL Dual-Axis Ankle Orthosis	University of California, San Francisco Medical Center, San Francisco, California
VAPC Shoe-Clasp Drop-Foot Orthosis VAPC Single-Bar Below-Knee Orthosis	Veterans Administration Prosthetics Center, New York, N.Y.

habilitation Hospital decided to extend the clinical study to include vacuum-forming of orthoses.

Because of delays in making commercially available the equipment designed by OCCC, a machine manufactured by Plastic Vac, Inc., but not designed specifically for orthotics, was purchased by Moss in 1973. Since that time molded lower-limb orthoses have become standard at Moss Rehabilitation Hospital. Development work is continuing but sufficient experience has been gained so that others can use some of the new designs with confidence.

MOLDED ANKLE-FOOT ORTHOSES

The molded ankle-foot orthosis (Fig. 2) is formed from a single piece of sheet thermoplastic. Polypropylene is usually the material of choice because of its high resistance to fatigue, low cost, and very light weight. It can be worked easily and can be welded, but will not take adhesives.

Molding is carried out over a positive plaster-of-Paris model of the foot, ankle, and shank, usually with the use of a vacuum system.

Although the molded AFO is a single unit, it has two basic components—the posterior calf enclosure and the foot enclosure (Fig. 3).

Fig. 2. Typical molded Ankle-Foot Orthosis.

The perimeter is referred to as the "trimline," and consists of five parts:

1. Proximal trimline
2. Anterior trimline
3. Ankle trimline
4. Foot trimline
5. Metatarsal trimline

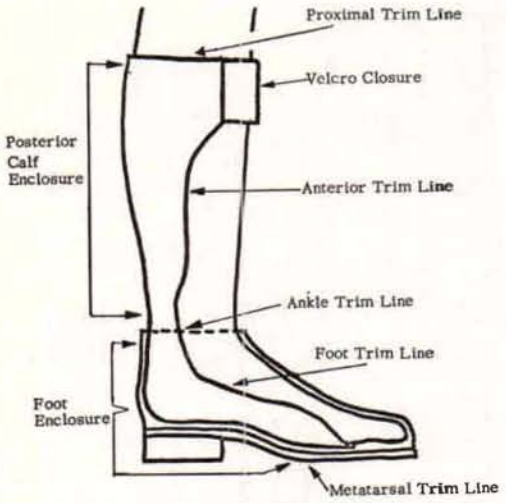


Fig. 3. Lateral view of molded AFO showing the various areas and trimlines.

FUNCTION

The function provided by a molded AFO is largely dependent upon the degree of resistance that is provided to rotation about the ankle joint. The resistance to rotation about the ankle joint, or rigidity of the ankle area, for any given material varies with the amount and distribution of material. Obviously, the thicker the walls the more rigid the orthosis will be for a given cross-section geometry. Not so obvious is the effect of the shape of the transverse cross-section.

Because the entire inner surface of the molded AFO is in contact with the limb, the shape of the foot, ankle, and calf of the patient dictates the basic geometry of the transverse cross-section. Everything else being equal, rigidity in the sagittal plane varies inversely with the ankle between the two sides of the orthosis in the ankle area (Fig. 4).

The rigidity provided by the shape of the anatomy can be modified by: location of the trimline; thickness of the material; and introduction of corrugations.

Thin walls are generally desirable and the orthotist will try to make the walls as thin as possible. They should be no more than 4 mm. thick at the thickest part, tapering down to 2 mm. However, corrugations are seldom needed because the orthotist can usually obtain the desired amount of flexibility by location of the trimlines. The effect of the orthosis on the patient will also depend upon the location of the weight line (Fig. 5, center), which in turn will depend upon the position of the foot part of the orthosis with respect to the shank part of the orthosis for a given heel height.

The location of the weight line must be considered at the time of casting and not left until the time of fitting with the idea that it can be controlled by heel height changes (Fig. 5).

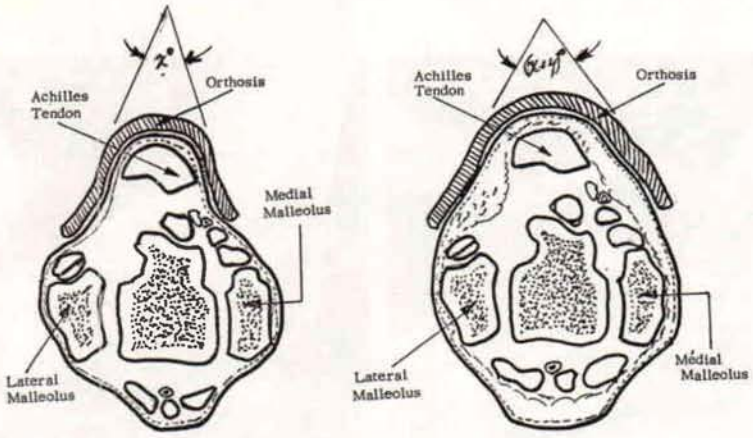


Fig. 4. Transverse sections through the lower shank that demonstrate how the transverse cross-sections of molded AFOs vary. Everything else being equal, rigidity in the sagittal plane varies inversely to the amount of the angle between the two sides.

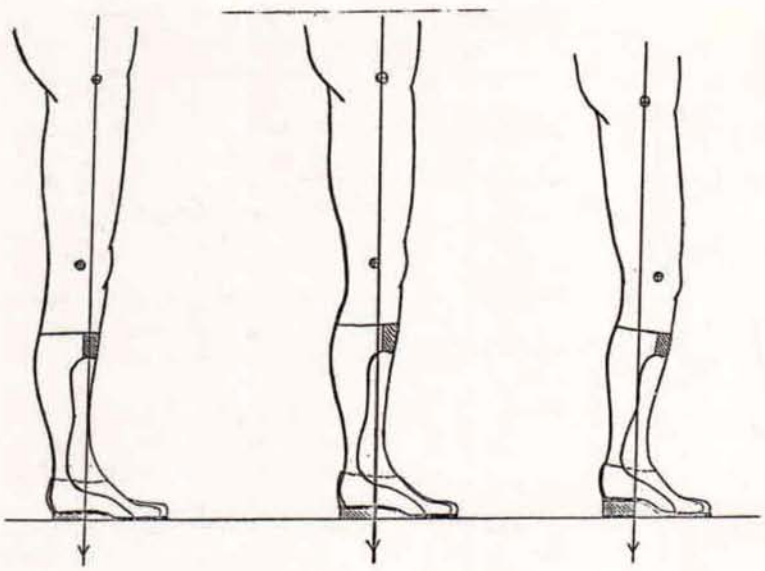
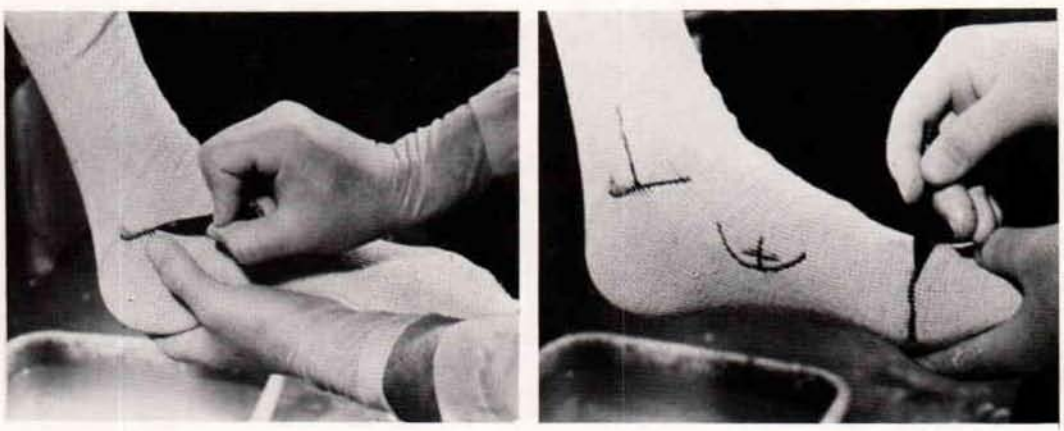
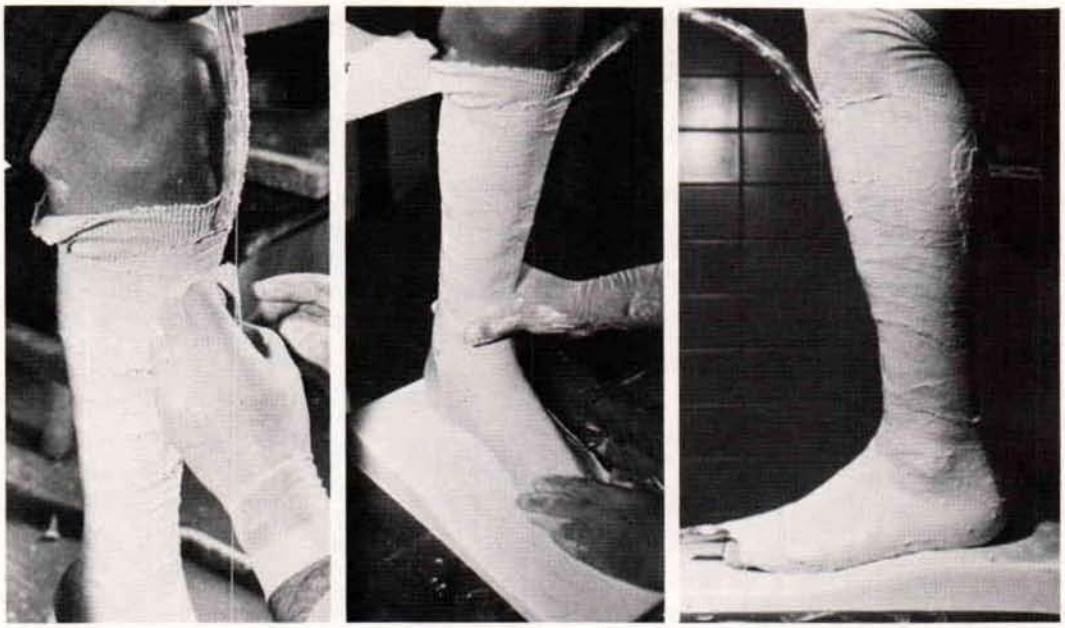


Fig. 5. Schematic drawing to show effect of heel height on function of an ankle-foot orthosis when heel height is changed after orthosis has been made. The vertical lines represent the weight-bearing line in each instance. Proper alignment should be achieved during casting and not by shoe modifications.

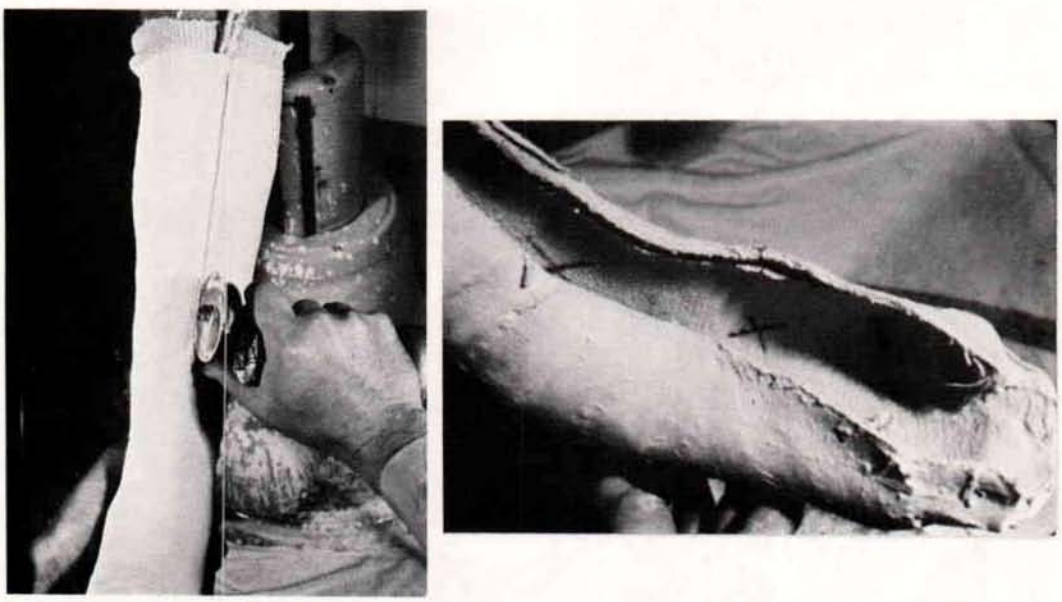
FABRICATION



Stockinet long enough to reach the mediotibial plateau is pulled over the foot and leg. Indelible pencil is used to mark the malleoli, the navicular, and the metatarsal heads.

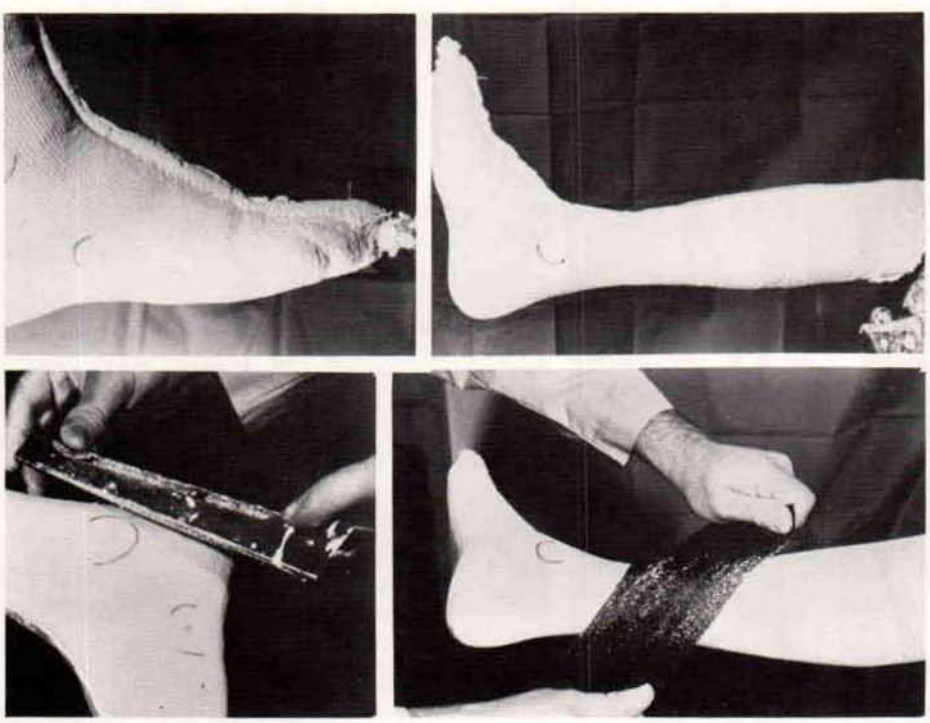


A vinyl or rubber tube is placed along the anterior aspect of leg to facilitate removal of the cast. Elastic plaster-of-Paris bandage is used to wrap the foot and leg. The appropriate size standard casting board is used, and the patient is sitting when the wrapping is carried out.

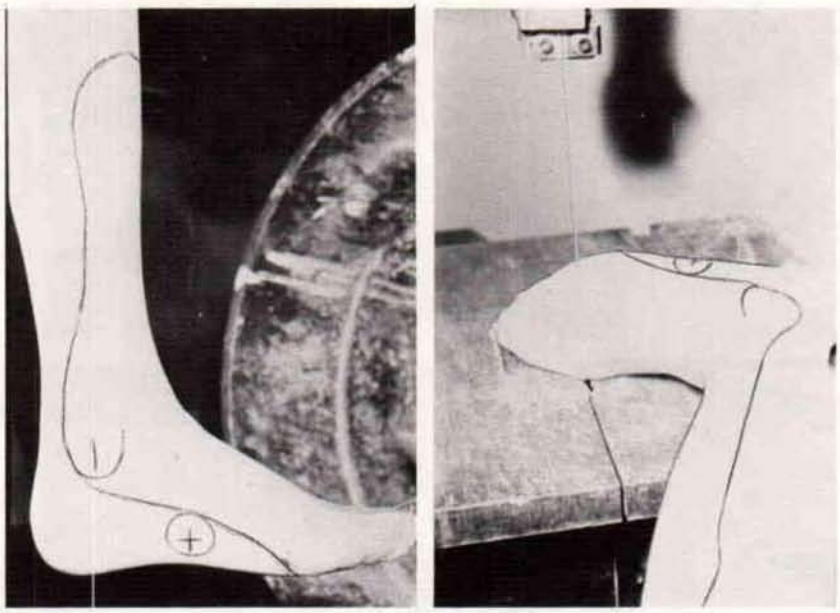


The cast is slit on the anterior aspect over the vinyl tube with a cast cutter, and removed from the patient. The marks showing locations of bony prominences and other points of reference are reinforced with indelible pencil and cast is closed and held together with staples or tape in preparation for pouring of the positive model.

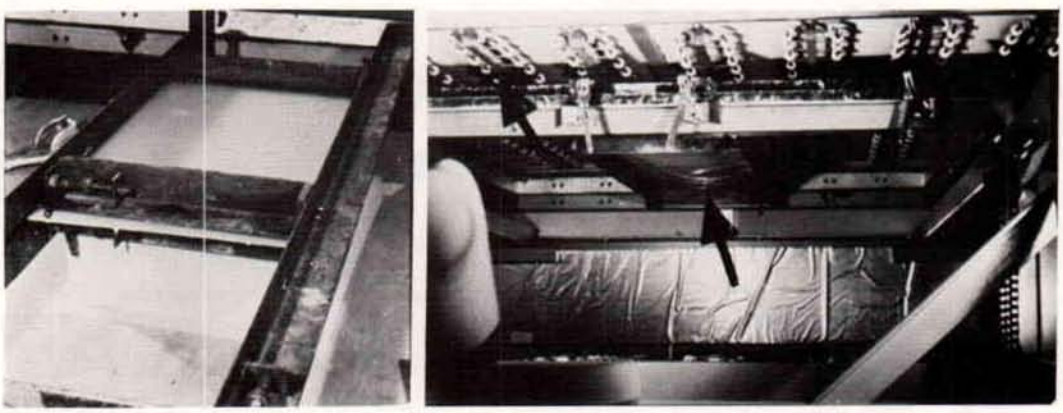
The cast is filled to produce a positive model. A mandrel is placed in the slurry so that it coincides as closely as possible with the centerline of the shank. When a cylindrical rod is used for a mandrel, it can be easily removed later if it is rotated a few times as the plaster hardens.



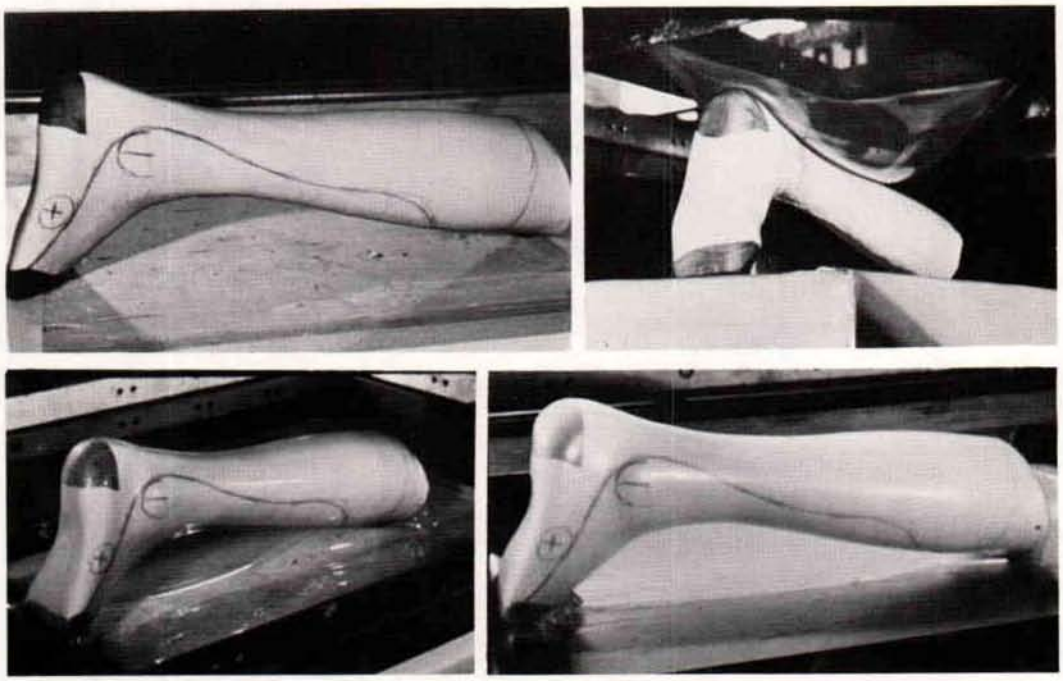
The ridge formed on the positive model as a result of the use of the vinyl tube is removed and the entire surface of the model is made smooth using conventional methods.



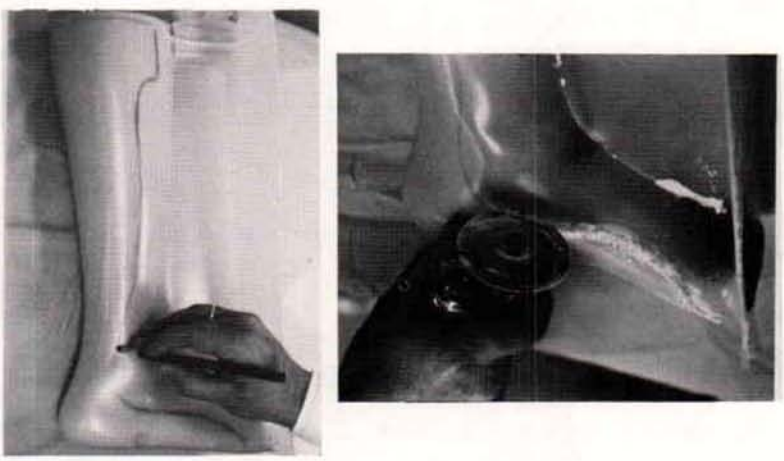
The first trimline is defined with indelible pencil. This line, of course, allows for more material to be left than is thought to be necessary in the finished orthosis. The toe section of the positive model is cut off in order to facilitate mounting the model in the vacuum-forming machine.



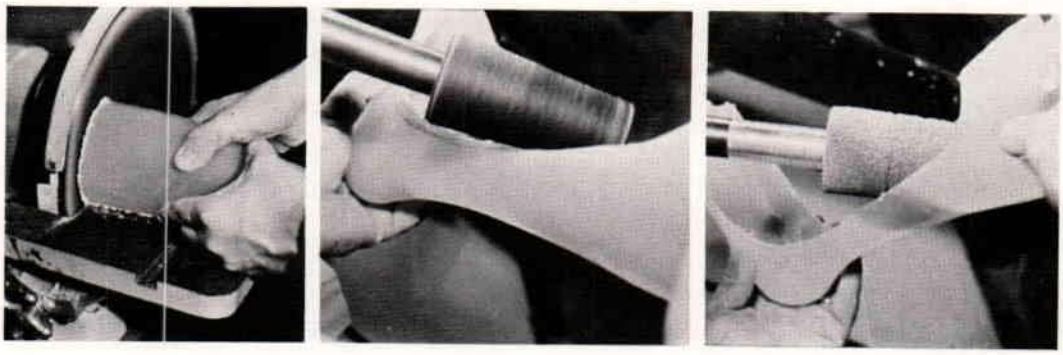
A sheet of polypropylene 12 in. x 24 in. is mounted in the frame of the vacuum-forming machine and subjected to a temperature of +400 deg. F until it sags to a level of about 2/3 the amount of the draw. As a general rule, AFOs for adults require polypropylene 3/16 in. thick; for children, 1/8 in. thick. The time required for heating ranges from 6 to 8 min., depending upon the thickness and temperature.



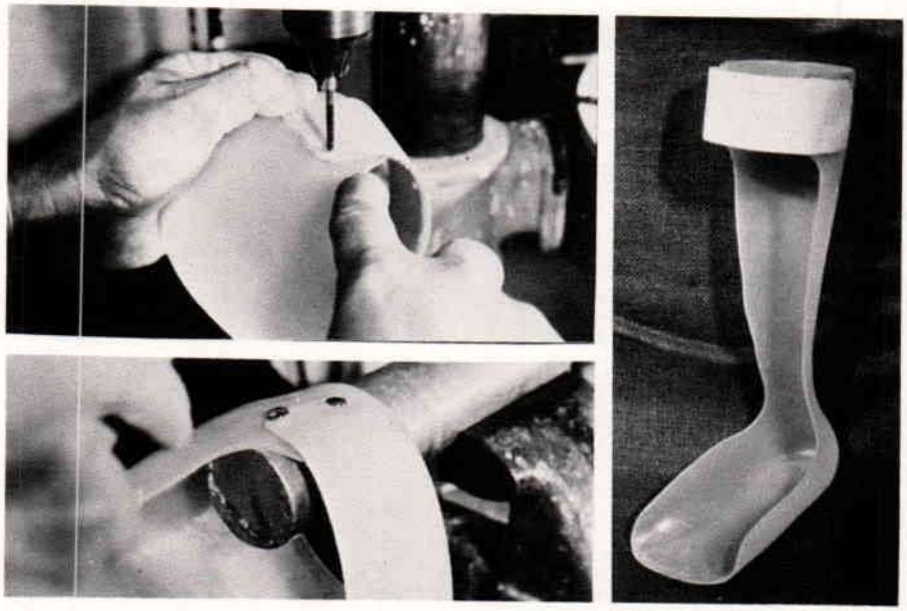
The softened polypropylene is lowered over the positive model and when the periphery of the polypropylene touches the table or platform holding the positive model so as to effect a seal, the vacuum is applied gradually until the model is covered entirely. Leaks caused by breaks can be sealed with molding clay.



The model and molding are removed from the vacuum-forming machine, and the initial trimline is reproduced on the outside of the polypropylene with a grease (china-marking) pencil since the marks on the positive model will not transfer automatically. A cast cutter is used to cut the molding along the trimline and thus permit removal from the model.



The edges of the orthosis are ground smooth using, as appropriate, the disc sander and the cone sander with coarse grit and TYCRO[3] cones.



Holes are drilled, Velcro straps are installed with rivets, and the orthosis is ready for initial fitting.

[3]Minnesota Mining and Manufacturing Co.

THE TRIMLINES

Location of the trimlines will depend upon the results of the assessment of the patient, thus on the patient's individual needs. The following rules apply in general:

- The proximal trimline (Figs. 2 and 6) is located approximately 3.8 cm. (1 1/2 in.) distal to the head of the fibula so that it will be clear of the peroneal nerve. The proximal trimline should encircle about 3/4 of the calf.
- The anterior trimline extends from the bottom of the junction of the Velcro strap and the orthosis to the midpoint of the shank. The anterior trimline should curve posteriorly as it proceeds distally to the midline of the shank (Figs. 2 and 6).
- The location of the ankle trimline affects the rigidity about the ankle joint more than any other single factor. Obviously, the further anterior that the trimline is located, the more resistance there is to rotation about the ankle.
- The foot trimline on the medial side extends through or slightly above the apex of the navicular. On the lateral side, the foot trimline extends slightly above the shaft of the fifth metatarsal.
- The metatarsal, or terminal, trimline (Figs. 2 and 6) lies along the apices of the metatarsal heads.
- All trimlines should be contoured and blended together smoothly. When a trimline is located slightly below a bony prominence, the area often becomes irritated, but when the trimline is located at the level of or slightly above the apex of a bony prominence, pressure areas do not occur as a rule.

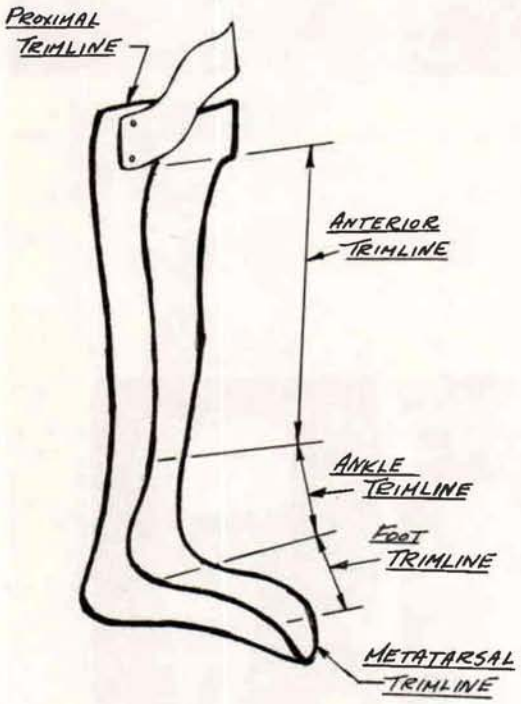


Fig. 6. Drawing to show location of various parts of the trimline of a typical molded AFO.

Until more definitive guidelines can be established for arriving at the rigidity desired, the location of the trimlines must be a result of the judgment of the orthotist. Some variations in trimlines to achieve desired effects can be seen in Figure 7. An attempt is being made to develop more definitive guidelines, especially for teaching purposes. Meanwhile, it is felt that competent orthotists will find that, with a little experience, quite satisfactory results can be achieved.

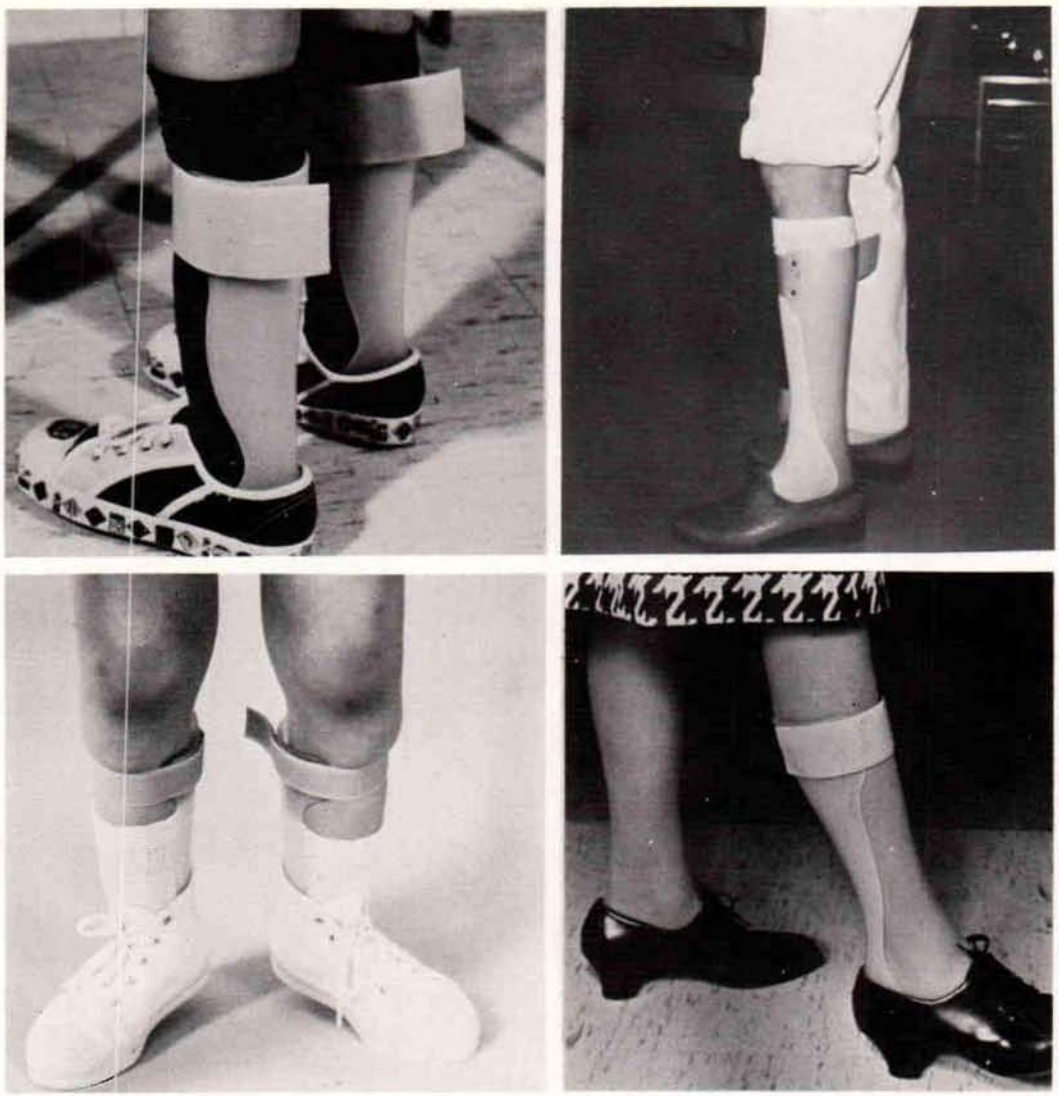


Fig. 7. Views that show variations in the trimline locations made to meet needs of the individual patient.

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1. Committee on Prosthetics Research and Development, *Report of seventh workshop panel on lower-extremity orthotics*, National Academy of Sciences, May 1970.

2. Finley, F. Ray, and Ray G. Wirta, *Rehabilitation biomedical engineering: orthotics design*, Krusen Center for Research and Engineering, Moss Rehabilitation Hospital, Philadelphia, Pennsylvania, May 1972.

Orthotics and Prosthetics, Vol. 29, No. 4, pp. 53–54, December 1975

TECHNICAL NOTE

ALLOWING NORMAL ADDUCTION OF FEMUR IN ABOVE-KNEE AMPUTATIONS

Virtually every article written in modern times on above-knee socket construction stresses the importance of support of the femur by the lateral wall of the socket. X-ray studies carried out at Fitzsimons Army Hospital since March 1974 show that very few above-knee prostheses built in the United States today achieve proper adduction of the femoral stump.

I have been amazed by the number of prostheses that are aligned so that the amputee is prevented from moving the femur into normal or equal adduction, because the prosthetic foot touches the sound foot while the femur is still in abduction. In *every* case, I found that, by simply realigning the knee and foot with respect to the socket, the amputee could bring the femur into a normal position.

After looking at X-ray pictures, it became evident to me that the head of the femur should be used as a starting point when aligning the AK prosthesis. I found the head of the femur to be located very near the center of the socket in the M/L dimension, and I now use that M/L center as a starting point in bench alignment.

A straight line from the center of the socket (viewed from the posterior) to the center of foot indicates the line of support. If the foot is located so that the support line intersects the distal end of the femur, a very close approximation of adduction equal to the contralateral side will occur. (I am sure every prosthetist knows that the distal end of the amputated femur is extremely close to the lateral wall of the AK socket.)

The only obstacle preventing adduction now is the prosthetic knee joint, which must be moved laterally to prevent it from touching the sound knee. I routinely place the center of the M/L axis of the knee on the support line during bench alignment. X-ray pictures confirm this position to be quite in accordance with the normal.

Aligning the distal femur over the center of the knee gives a definite advantage to wearers of hydraulic knees. A definite decrease in rotation is noticed when the distal femur is directly over the hydraulic cylinder.

A typical bench alignment of a long stump is shown in Figure 1. Photographs of a typical prosthesis aligned in the manner suggested here are shown in Figure 2. The socket brim is shown in Figure 3.

Adduction is difficult to maintain when the M/L dimension of the socket is excessive, because support of the femur is lost when weight is applied and the top of the socket moves laterally. When the M/L dimension is too large it is possible to have a socket with the lateral wall appearing to have 20 deg. adduction built in and yet the femur will be in abduction. This is not at all unusual; in fact, most United States AK limbs have this fault.

It would seem to me a serious study should be

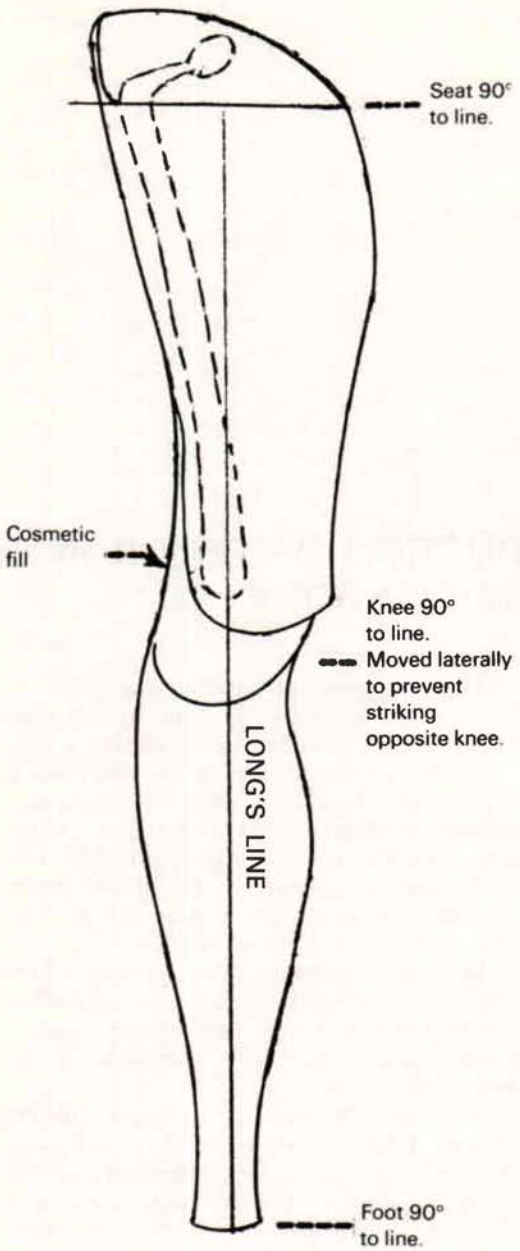

Fig. 1.

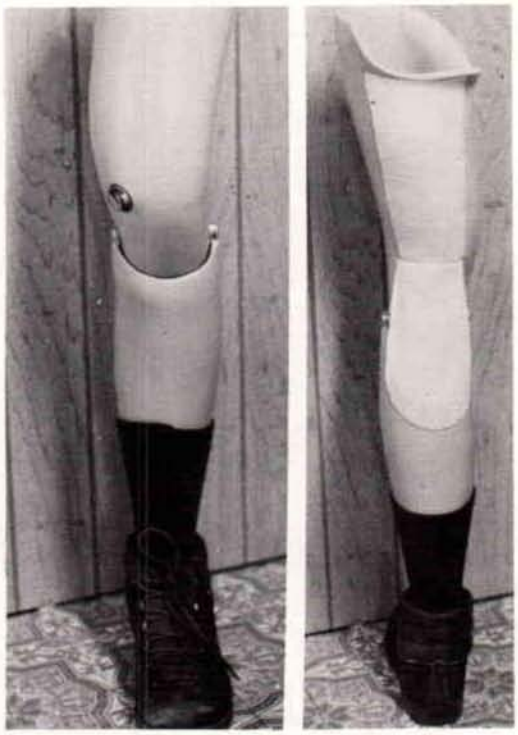

Fig. 2.

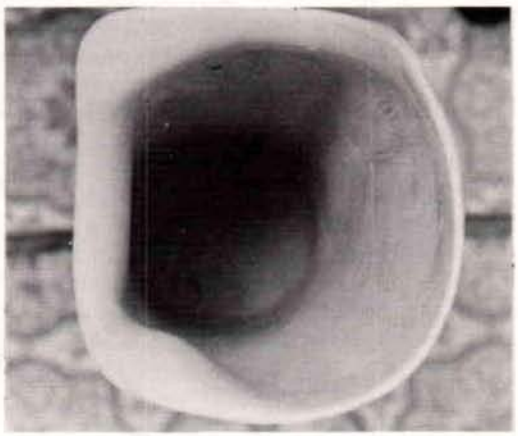

Fig. 3.

devoted to developing a method of measuring the stumps to give a more realistic shape to the top of the socket.

(Most IPOP casts achieve better adduction than regular sockets.)

This alignment permits the AK amputee to walk freely, securely, and without back pain, and side motion of trunk is greatly stabilized. Most amputees can balance on the prosthesis without shifting shin over heel. This is impossible with the stump in abduction.

Ivan Long, C.P.
8790 West 60th Avenue
Arvada, Colorado 80004

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.

METRIC SYSTEM
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 $\quad °C = \dfrac{°F - 32}{1.8}$

°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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1. THE ORIGINAL MANUSCRIPT AND TWO COPIES. If possible, the duplicate manuscripts should be complete with illustrations to facilitate review and approval.
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3. LEGENDS. List all illustration legends in order, and number to agree with illustrations.
4. ILLUSTRATIONS. Provide any or all of the following:
   a. Black and white glossy prints
   b. Original drawings or charts

   Do *not* submit:
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   b. Photocopies

PREPARATION OF MANUSCRIPT

1. Manuscripts must be TYPEWRITTEN, DOUBLE-SPACED and have WIDE MARGINS.
2. Indicate FOOTNOTES by means of standard symbols (*).
3. Indicate BIBLIOGRAPHICAL REFERENCES by means of Arabic numerals in parentheses (6).
4. Write out numbers less than ten.
5. Do not number subheadings.
6. Use the word "Figure" abbreviated to indicate references to illustrations in the text (. . . as shown in Fig. 14)

PREPARATION OF ILLUSTRATIONS

1. Number all illustrations.
2. On the back indicate the top of each photo or chart.
3. Write the author's name on the back of each illustration.
4. Do not mount prints except with rubber cement.
5. Use care with paper clips; indentations can create marks.
6. Do not write on prints; indicate number, letters, or captions on an overlay.
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Orthotics and Prosthetics, Vol. 29, No. 4, pp. 59–67, December 1975

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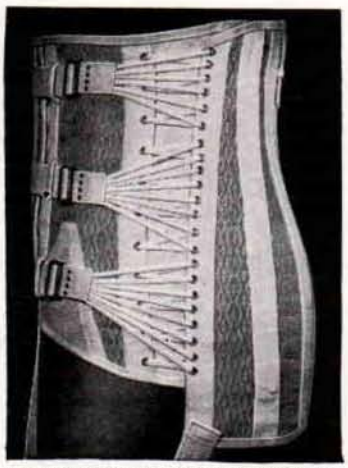

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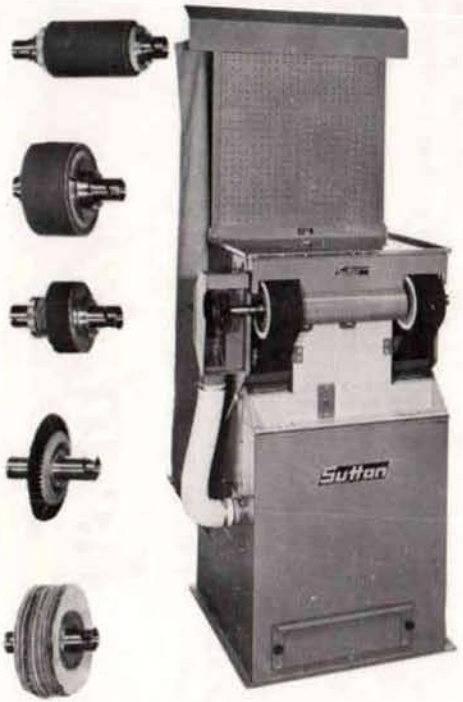

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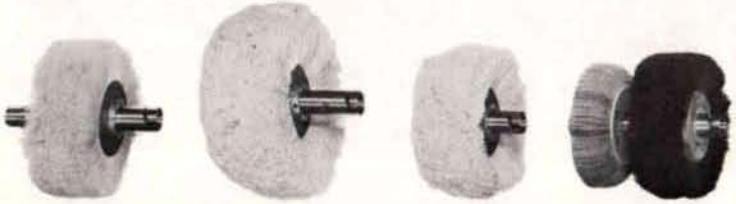

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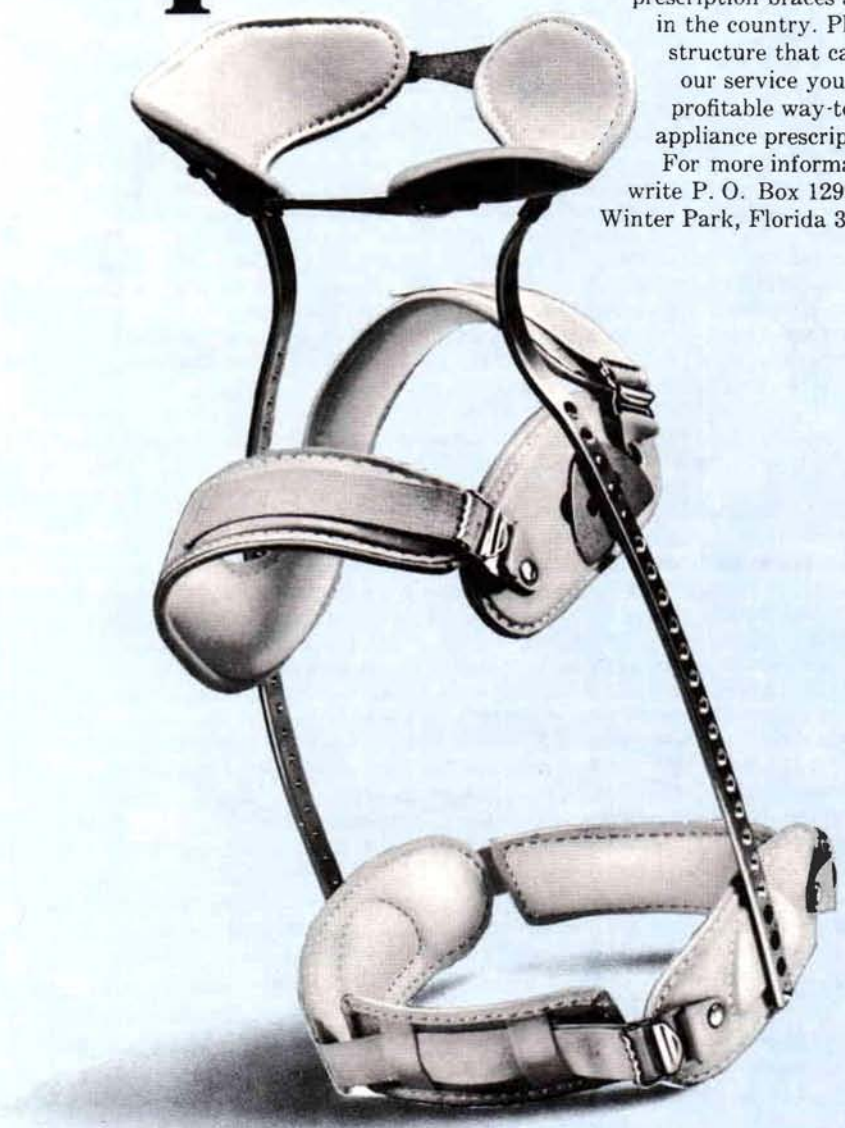

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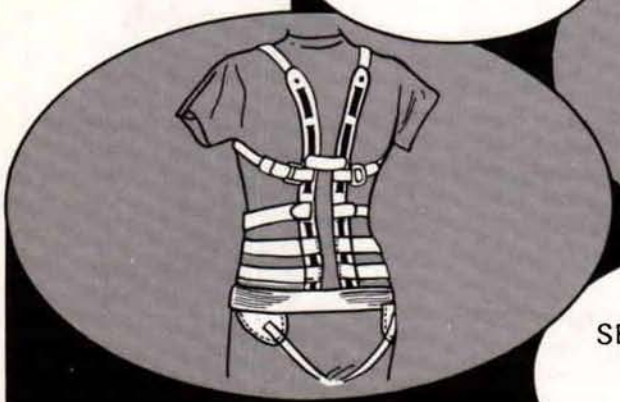

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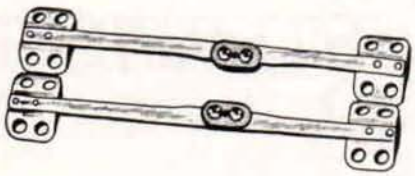

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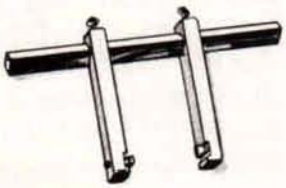
Small 16″ to 19″
Med. 19″ to 23″
Large 23″ to 28″

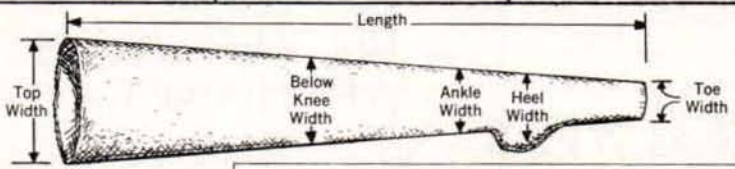
Lefts and Rights

ALIGNMENT JIG



Cast Boot
Small
Medium
Large

CLAMPS, Snaplock
Two necessary for each cast.



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Lycra/Spandex 4.75%
Closed Toe (not pulled up)
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35″	9″	5″	4″	6″	3½″
39″	9″	5″	4″	6″	3½″
42″	9″	5″	4″	6″	3½″
46″	9″	5″	4″	6″	3½″
46″	10″	6″	4½″	6½″	3½″

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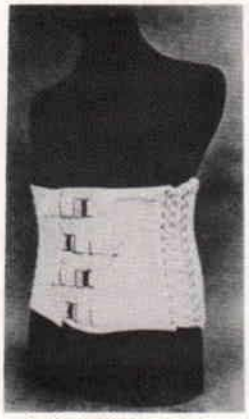

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