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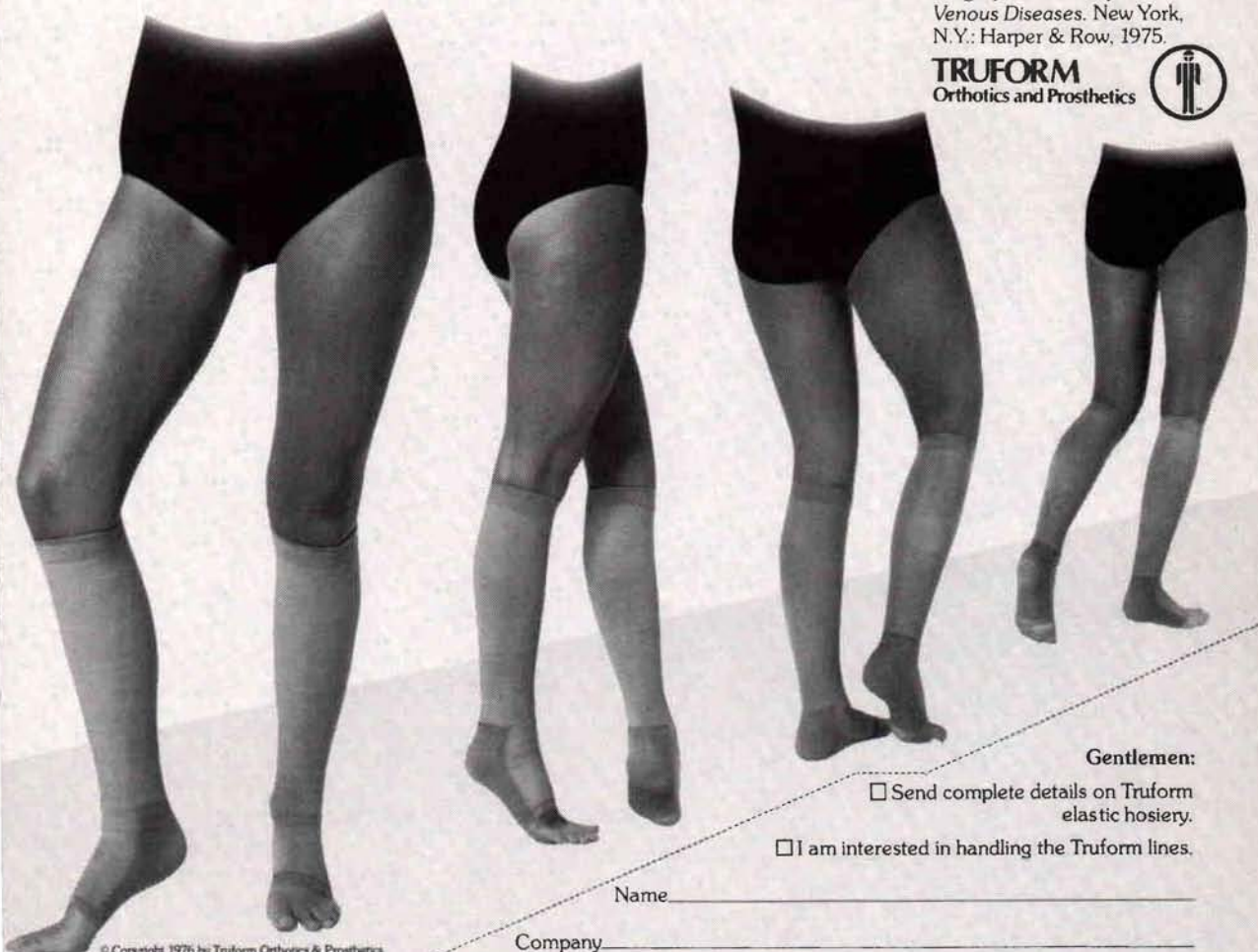

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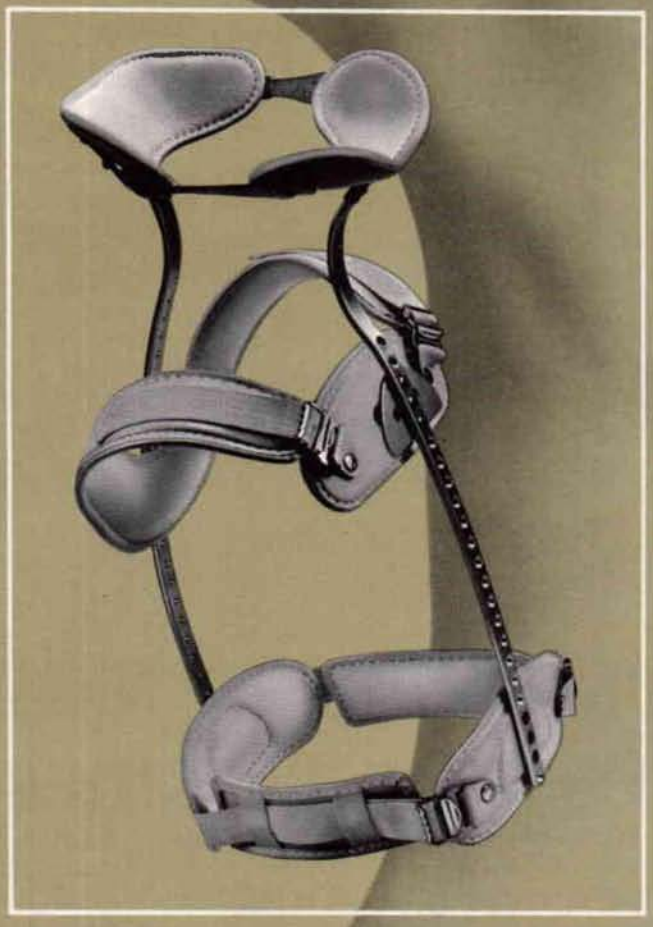

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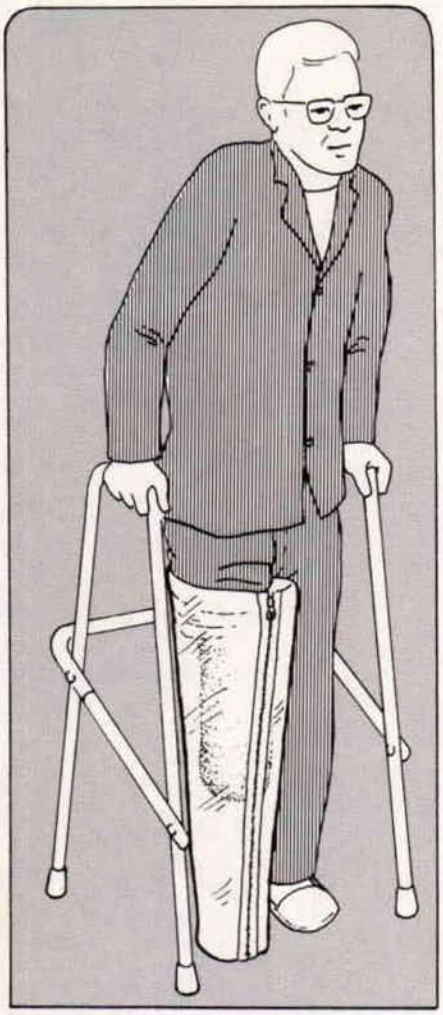

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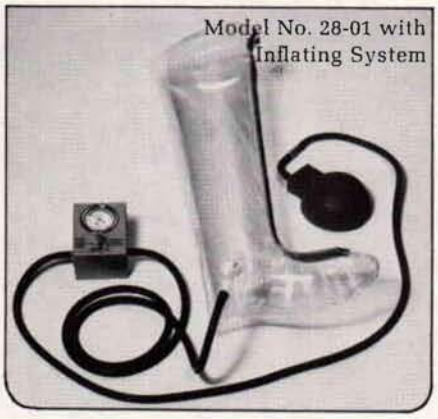

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Orthotics and Prosthetics, Vol. 30, No. 4, pp. 1–2, December 1976

TO CREATE IS NOT TO INNOVATE

All of us know people who are creative, intelligent, and industrious, but somehow their ideas never seem to get implemented. The fact is that creative thinking people are plentiful; nearly every business has a few of them. Theodore Levitt (*1*) stated in 1963 "The major problem is that so-called creative people often (though certainly not always) pass off on others the responsibility for getting down to brass tacks. They have plenty of ideas but little businesslike follow-through. They do not make the right kind of effort to help their ideas get a 'hearing and a try'." Creativity is the generation of ideas; their implementation is innovation. Orthotists and prosthetists are generally "creative" types, rather than "business" types, and usually they inherently have problems in implementing their ideas. Unfortunately, the result of this phenomenon is that other professionals who are more adept at implementation of ideas influence heavily the future of this profession.

A few years ago, a very talented research group developed an electrically powered prosthetic joint that received wide publicity, but to date the device has been used rarely outside of a research setting. The developers have blamed this on conformity in the orthotics-prosthetics field, where new ideas seem to be opposed routinely. The developers were definitely creative, but they failed to realize that the principal purpose of orthotics and prosthetics practices is action in the form of patient services. Few practitioners can find the time and money necessary to invest in a new technique that does not easily fit into the established system, or one that "rocks the boat". The orthotist-prosthetist's task is to handle a constant flow of patients and their problems, and to answer an unending series of questions on which decisions are made. To quote Mr. Levitt again, "everytime an idea is submitted to him, it creates more problems — and he already has enough".

So how do you implement your ideas? First, you should recognize that creativity does not automatically lead to actual innovation. Second, you should try to modify your idea so that it fits into an existing situation with as few changes in the "system" as possible. Third, the idea must be presented as reasonably and as responsibly as possible. Such factors as cost, risk, manpower, time, and comparison to existing systems should be detailed. Obviously, as ideas become more complex and more changes are needed in the organization, more detail is required. Fourth, the nature of the people you are addressing, and the nature of the profession at large, should be recognized and the amount of care that is put into the proposal adjusted accordingly.

You may rightly say that the truly creative person will be stifled if he must attend to all of these details before his idea is accepted. Often you will find a creative person, or a visionary, working side-by-side with someone whose knack is implementing his ideas; usually resulting in a synergistic relationship.

I feel that, within the last year or two, many outside groups (health professionals, government agencies, etc.) have begun to view our profession as static and unwilling to utilize new developments, such as powered prostheses, and to expand into related areas, such as powered wheelchair controls and other aids for the severely disabled. Our lack of knowledge of mechanical and electronics engineering is one reason that the orthotist-prosthetist balks at these new systems.

We hope that the university programs in prosthetics and orthotics will provide this knowledge in the future.

Also at fault is the lack of promotional and educational programs by the developers of some of these systems. In any case, a new specialist, the "rehabilitation engineer", will soon be on the scene to work with you and me in order to apply these new developments on patients — to implement new ideas. In the future, many orthotics and prosthetics practices may want to employ these individuals to help us expand our services.

In the meantime, however, we should demonstrate to the outside world that we are not resistant to changes, and that many of us are capable of expanding our services into other areas where assistive devices must be matched to the patient.

Now is the time for orthotists and prosthetists to assume a leadership role in the bionic age, but we can only assume this role if we accept new ideas and if our own ideas, based upon years of clinical experience, are accepted.

MICHAEL QUIGLEY

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Orthotics and Prosthetics, Vol. 30, No. 4, pp. 3–15, December 1976

A DYNAMIC ORTHOTIC SYSTEM FOR YOUNG MYELOMENINGOCELES

A PRELIMINARY REPORT[1,2]

John Glancy, C. O.[3]

Before discussing the approach that is in development at the Indiana University Medical Center for improving orthotics for myelomeningocele children, a few remarks about conventional orthoses in general use may serve as a point of reference.

CONTROL OF THE FOOT/ANKLE COMPLEX

The conventional double-upright, below-knee orthosis cannot control the paralyzed foot/ankle complex. Its principal weakness is the shoe. Because leather is a flexible material, especially when wet, and the intimacy of the fit of high-top shoes is decidedly suspect, a shoe cannot check unwanted mediolateral rotations about the subtalar joint of a flail foot. Acknowledgement of the control limitations of shoes led many years ago to the development of tee straps and mediolateral wedging.

Unfortunately, tee straps are also made of leather. The accumulation of scientific data by use of force plates has given us an understanding of the effect of floor reation forces upon foot balance. The validity of placing a flail foot upon an inclined surface, not parallel to the floor mediolaterally, to achieve a 'balanced' distribution of weight through the subtalar joint is questionable (*3*).

CONTROL OF THE KNEE JOINT

The double-upright, above-knee orthosis is generally used to control mediolateral and/or anterioposterior instability of the knee joint. Locks are added when buckling of the knees is the major problem. In any case, these orthoses are too heavy for the small paralyzed child. Also, the small paraplegic child cannot manipulate the locks and when he falls, getting to his feet again, on his own, is decidedly difficult for him.

CONTROL OF THE PELVIS AND TRUNK

Conventional orthotic means of controlling the pelvis and trunk include hip joints (both free and locking), metal pelvic bands and/or spinal orthoses. All of these components inhibit motion to a degree that many clinicians find unacceptable. They are bulky and heavy, and the smaller the child, the greater the effect. There being no other alternatives, these components have been used in varying combinations to serve two major purposes:

1. To prevent the development of hip contractures and excessive lordosis in growing myelomeningocele children.

2. To support the pelvis and trunk in the standing position in order that the very young child can develop a 'sense of balance' prior to attempting ambulation.

Within the past several years, prebalanced, platform orthoses such as the Parapodium (*6*) and the Swivel Walker (*7*) have been developed for young myelomeningocele children. Since platform orthoses have been available for such a relatively short time, I will continue to confine my remarks to what I have been referring to as

[1]Portions of this report were presented at a seminar on Current Trends in Myelomeningocele, held at the Plaza Inn, Kansas City, Kansas, April 5-7, 1976. The seminar was sponsored by The Children's Rehabilitation/University Affiliated Facility and the Section of Orthopaedics, University of Kansas Medical Center.

[2]This project was partially supported by a grant from the James Whitcomb Riley Memorial Association.

[3]Indiana University/Purdue University at Indianapolis, School of Medicine – Department of Orthopaedic Surgery – Division of Orthotics.

the 'conventional'. Does placing a myelomeningocele child in a static orthotic system, regardless of its design, contribute to his development of a 'sense of balance'? The question remains open.

How successful then has the combining of hip, pelvic, and trunk components been in the prevention of musculoskeletal deformities of the hips and spines of myelomeningocele children? I believe that it is fair to say that success with conventional orthoses in preventing the development of deformities has been, at best, a 'sometime thing'. What then, are the factors that explain such poor results? The following general conclusions were drawn and served to stimulate the developments now underway at Indiana University:

1. *The dilemma that the state-of-the-art of orthotics presents; namely that the cost of longterm protection of the musculoskeletal system of the growing myelomeningocele child is an unacceptable restriction to motion in terms of the immediacy of daily living. Conversely, removal of available means of longterm protection inevitably leads to an increase in deformities, followed by an inevitable decrease in daily activities.*

2. *The difficulty of getting the myelomeningocele child and/or his family to accept a conventional orthosis because it is an additional burden and/or restriction. We are all familiar with the proud parent's descriptions of how 'active' their young child is; "He can get anywhere he wishes by using his arms to slide across the floor, but when in his braces, he moves about very little."*

3. *The longer the growing child of conscientious parents is kept in conventional orthoses, the more obvious it becomes that the orthoses are* not *preventing further deformities and the rational for their continued use becomes less and less defensible.*

4. *The older a myelomeningocele child is when introduced to conventional orthoses, the less likelihood there is that he will wear them and derive advantages from them. Psychological reasons excepted, the crux of the dilemma is a lack of function. Experience in other areas seems to indicate that when a patient is able to function better when wearing an orthosis, he quite willingly accepts the unavoidable annoyances that are a part of wearing one.*

5. *Is cosmesis important in the final analysis? To those who have experienced the bracing of hundreds of patients from childhood through the teenage years and into adulthood, the answer is a very positive* yes, *especially to the individual whose condition will require the assistance of orthotic devices throughout his life. As the individual grows from early adolescence to maturity, appearance becomes his dominant concern, especially when functional improvement is marginal. To be physically 'different' than one's fellows in a private way, (under one's clothing), is a great deal less disconcerting than the* forced *public display (there being no other alternatives), of the particulars of one's physical differences. Those who are able to overcome the* lack *of choice are rare exceptions.*

Our answer to the failure of conventional below-knee orthoses to prevent deformities of the flail foot, once standing begins, is the "solid-ankle" orthosis (Fig. 1) (*2, 5*). The

Fig. 1. The "solid-ankle" orthosis.

rationale for the design of the "solid-ankle" orthosis is to achieve balance and gait as close as possible to the bilateral below-knee amputee by eliminating unwanted motion of the flail foot/ankle complex in all planes. During its development, consideration was given to: Control of knee flexion; lightness and durability; its being worn inside the shoe; and that the orthosis should be a single unit to ensure proper donning and function. The shell is molded of polypropylene and lined with Plastazote. The "solid-ankle" orthosis weighs from 6-8 ounces, depending upon the size of the wearer.

The "solid-ankle" orthosis prevents buckling of the knee (*4*, *6*) by inhibiting rotation of the tibia about the ankle axis. However, the myelomeningocele child presents further complications that were not readily apparent during the first few years that "solid-ankle" orthoses were used.

The complications arise because of three factors:

1. Stability of the knees depends upon the patient keeping his center of gravity forward of his knee axes.

2. The myelomeningocele child's lack of awareness of floor reaction forces below the level of deficit.

3. The total or partial lack of proprioception in all joints below the level of deficit.

The efficient control of the foot/ankle complex provided by the "solid-ankle" orthosis cannot prevent further deformities from developing in the hip and lumbar regions. (Fig. 2).

It is easier to visualize these children's problems by beginning with a child in the seated position wearing "solid-ankle" orthoses, (it being understood that, though not shown, the child must use his arms for assistance). Note the lordotic posture of the seated child in Figure 2A. Innervated muscles cross the hip joints anteriorly, but there are no muscles posteriorly to check the out-of-phase anterior rotation of the pelvis that occurs as the child descends to the seated postion. Once seated, there being no way for him to 'right' the lumbosacral angle of his pelvis, his lumbar spine must compensate by going into excessive lordosis. It is the only option he has for getting the CG of his trunk

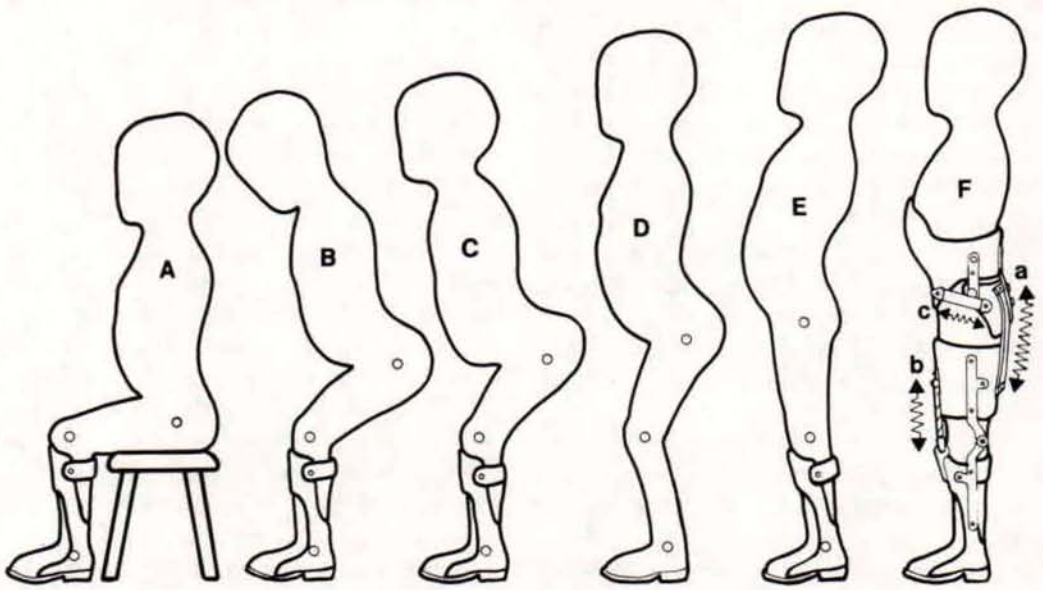

Fig. 2. Sketches to illustrate some of the problems of the myelomeningocele child.

over his buttocks and thus maintain sitting balance.

Figure 2 B shows the child beginning to rise to his feet and at this stage, his CG is well behind his knee axes and feet.

Figure 2C follows the ascent to a more extended position. The quadriceps are firing to extend the femurs about the knee axes. Since the tibiae are 'fixed' by the "solid-ankle" orthoses, the vasti muscles perform efficiently in their normal function of extending the femurs about the knee axes. However, because of the absence of opposition from the gluteus maximus and hamstrings that cross the hip joint, the rectus femoris becomes a powerful *unwanted* rotator of the pelvis in an anterior direction, bringing the trunk forward with it. It is apparent that, were this motion to continue, the youngster's CG would be forward of his feet and he would fall forward. The sketch attempts to catch the instant when the child is beginning to increase his lordotic posture to a much more severe degree — his only option for bringing his CG back over his feet. Unfortunately, his vasti muscles must now work constantly to prevent his femurs from going into further flexion about the knee axes. Unable to rest, they soon tire and the youngster will fall down backwards.

Figure 2D illustrates how it is possible for a child to maintain his balance without orthoses. Flexing his tibiae forward about his ankle axes permits him to bring his knee axes forward of his ankle joints. He is now able to maintain his balance by going into an excessive lordotic posture but with one all-important difference—he is able to keep his CG forward of his knee axes. Thus the burden upon the vasti muscles is dramatically reduced. The relationship of the development of hip and knee contractures and excessive lordosis to anterioposterior balance becomes apparent.

Figure 2E shows the posture that can be achieved with "solid-ankle" orthoses and posterior heel wedges when the orthoses are formed so that the feet are held in a neutral position with respect to the tibiae. The posterior wedge brings the knee and hip axes forward of the ankle axes. It is especially significant to note that the hip joints are *more forward* than either the knee or ankle joints. If these children's feet were cast in dorsiflexion, the result when standing, would be to force them into the posture shown in Figure 2D. When the feet are placed in equinus as in the technique used with poliomyelitis patients (*8*), the hip and knee axes are maintained directly over the ankle axes. The resulting vertical alignment of these joints makes anterioposterior balance of the trunk about the hip axes extremely precarious due to the ever present involuntary rotation of the trunk, i.e., 'swaying' in the anterioposterior plane. Were his joints aligned vertically, an increase in lordosis in the lumbar region would place his CG behind his feet and he would then have to flex his trunk forward about his hip axes in order to keep his CG over his feet. The forward placement of the hip joints, as illustrated, brings the mass of the trunk forward of the knee axes, thereby adding to their stability. The additional stability of the knee joints allows a relatively mild but essential increase of lordosis to bring the body's CG back over the base of support *without having to flex the hip joints to achieve balance.*

When "solid-ankle" orthoses are first applied, the result is very gratifying. Unfortunately, two or three years later, it is apparent that the battle to prevent hip contractures and increased lordosis is being lost. Although prevention of deformities to the foot/ankle complex and knee joints has been successful by eliminating the need for flexion in the ankles and knees to maintain balance, without his gluteus maximus muscles the only option the child has to maintain his balance is excessive lordosis. As time passes, the vertical load upon the lumbar spine is bound to increase the lordotic curve and without the opposing stabilizing force of the gluteus maximus muscles, the pelvis will be forced to rotate anteriorly. Growth, and the inevitability of bony deformation locks in the deformities.

Figure 2F shows portions of the present design of the dynamic orhtotic system that has been developed for the myelomeningocele

child. The components shown are for the L3 to L4 level. This system consists of the following bilateral components; "solid-ankle" shells, single lateral aluminum uprights with offset knee joints, polypropylene quadrilateral thigh cuffs and the thoracopelvic unit with its pivotable pelvic portion. Dynamic forces (generated by elastic materials), are applied to the system as the zig-zag lined arrows indicate. The letter 'a' identifies the pelvic extension assist; 'b' the knee extension assist; and 'c' the part designed to keep the lower paraspinal muscles on stretch. Note that no rigid members cross the hip joints and that the system is free of locking devices.

Figure 3 shows details of the design of the polypropylene thoracopelvic unit. The anterior view (Fig. 3A) shows the opening with a Velcro closure. Above and below the Velcro strap are .050-in. thick stainless steel strips that fit into nylon pieces that are slotted to receive them. This feature ensures that the portions on both sides of the opening cannot slide up or down or bulge in or out with respect to each other, thus preserving the integrity of the fit and also preventing any 'sawing' action upon insensitized portions of the patient's skin. The flattening of the abdomen is the same as used in the Milwaukee-Brace technique (*1*). The lateral views (Fig. 3B, C) also demonstrate the pivotal action of the pelvic portion and the sliding of

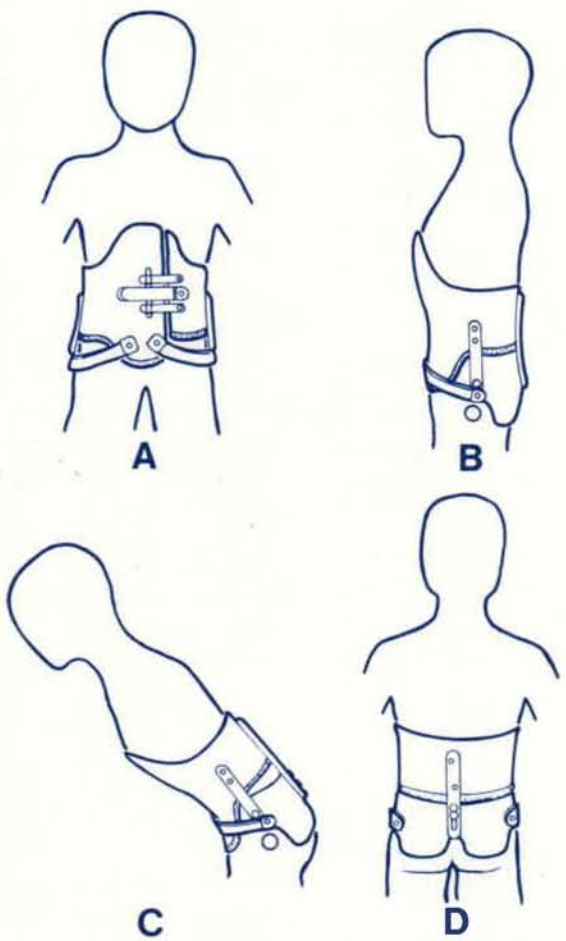

Fig. 3. Polypropylene thoracopelvic unit: A, anterior, B and C, lateral, and D, posterior view.

the slotted posterior polypropylene upright that allows the posterior of the thoracic portion to slide away from the pelvic portion to accomodate for approximately 1/3 of the elongation of the back as it flexes forward. Note the constant pressure that the pelvic portion is able to maintain upon the buttocks. The posterior view (Fig. 3D) shows the position of the slotted posterior upright, which serves as a stop to prevent excessive lumbar lordosis, when standing or sitting. Contrary to conventional spinal bracing, the *thorax* is used as the *foundation* from which the pelvis is controlled. The entire thoracopelvic unit weighs from 12 to 18 ounces, depending on the size of the child.

The schematic drawing in Figure 4 demonstrates the specific functional advantages that a myelomeningocele child realizes from the force generated by the elastic pelvic extension assist. To clarify the drawing, the elastic is not shown, but is represented by the zig-zag arrow from the proximal edge of the pelvic portion (a) to the distal edge of the quadrilateral thigh cuff (b) to illustrate its attachment points as well as its line of force. The system shown is designed for the youngster who has both medial and lateral hamstrings. Because stability of the knees is not a problem, neither the offset knee joints nor the knee extension assist units are necessary. However, the lack of innervation to the gluteus maximus muscles does leave us with the problem of preventing progressive lordosis. It is interesting to note that youngsters with the full complement of hamstring muscles do not develop *fixed* flexion contractures of the hips but do adopt a posture of hip flexion which appears to be an accomodation, for purposes of balance, to the excessive lordotic posture that they cannot avoid.

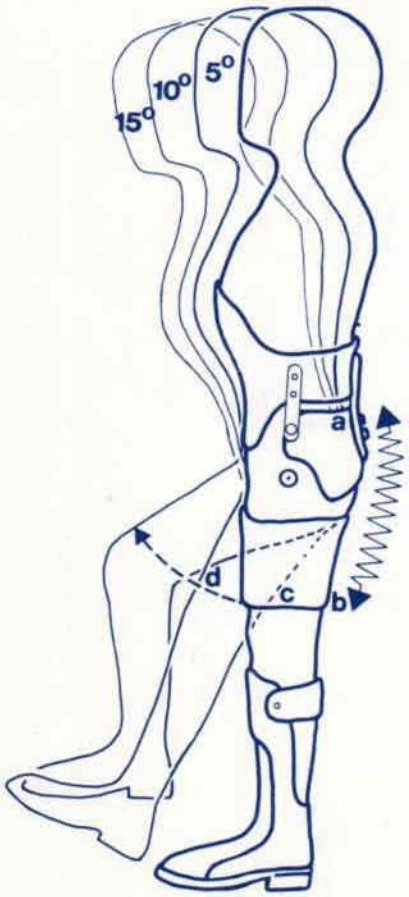

Fig. 4. Schematic lateral view of the orthotic system for the myelomeningocele child who has stable knee joints.

This system is based upon the premise that in these cases involuntary swaying of the pelvis is occurring constantly about the hip joints in the anterioposterior plane. It is assumed that tone in the abdominal muscles and anterior musculature crossing the hip joints is adequate to control involuntary rotation of the pelvis in the *posterior* direction. Because innervated anterior and posterior musculature crossing the knee joints is virtually complete, it is further assumed that both anterior and posterior tone is also present. The stability of the foot/ankle complex within the "solid-ankle" orthoses rules out involuntary anterioposterior swaying at either the ankle joints or at floor level. With these 'givens' present, the paralysis of the gluteus maximus muscles takes on a more subtle significance, i.e., the missing element to trunk balance appears to be the absence of the relatively light activity commonly referred to as 'tone', *not* the great power that these muscles are capable of generating. If then, a low magnitude of force is adequate to maintain equilibrium of the normal trunk, elastic materials seem to be a practical means of introducing dynamic

control of trunk balance. The key seemed to be to apply the dynamic force in a form that would mimic the action of the missing gluteus maximus muscles, i.e., in the form of an extension *moment* about the hip joints. All signs seem to indicate that whenever the gluteus maximus muscles are paralyzed involuntary sway about the hip axes in an anterior direction is the *trigger* that sets off excessive lordosis. The formula for determining how much force should be applied is based upon free-body analysis and has been published (*4*).

The trunk and leg motions depicted in Fig. 4 are to scale in order that one may relate various degrees of motion about the hip axes to the actual positions of segments of the body as they can be clinically observed. As an example, let us take the child who weighs 30 lbs. and whose trunk length (measured from the center of the hip joint to the top of the head) is 20 inches. A preload of four pounds of force (which would be generated by bilateral elastics, each one set for two pounds of force), would allow the child to voluntarily flex his trunk forward five degrees about the hip axes and would automatically extend it for him within the five degree range. Ten degrees of forward flexion would require a preload to each of the bilateral elastics of slightly more than four pounds to automatically return the trunk to a balanced position of extension. For 15 deg. of forward flexion a 6.25 lb. preload setting to each of the bilateral elastics would be needed to automatically return the trunk to equilibrium. The preload settings allow motions of the arms and head (which affect the position of the CG of the trunk), without upsetting trunk balance. The elastic pelvic extension assist, then, provides three functions simultaneously:

1. It maintains trunk balance.
2. It places soft tissues that cross the hip joints anteriorly on constant stretch to prevent flexion contractures.
3. It provides a "safety zone" within which the child may move his arms and head and/or flex his trunk without fear of jackknifing.

The child can overpower the extension moment at will, simply by flexing his trunk beyond the preset safety zone. As he flexes forward, he increases the distance between the attachment points at either end of the elastics and, in so doing, automatically increases the force of the elastics. The extra force then becomes a control mechanism for motion initiated by the child. For example, as the child lowers himself into a chair, the extension moment about the hip axes will automatically increase and by the time he is seated, the extension force will be 2 to 2.5 times greater than its preload setting. Conversely, the child has assistance rising from a chair from this extra force which automatically diminishes as he extends his trunk. Thus, the mechanical pelvic extension assist appears to be a reliable, albeit limited, replacement for some essential functions of the normal gluteus maximus muscles.

The young wearer is now placed in a learning situation. He *must* be an active participant in the control of his body, i.e., he must understand that voluntary movements of his trunk within the "safety-zone" are totally within his control, whereas *involuntary* movements beyond the safety range mean a loss of balance. It is felt that this learning process can be likened to that of a normal child, but with obviously narrower margins of safety to be mastered. The feeling is that the dynamic environment presented by the orthotic system makes it possible for the child to actually develop a 'sense' of balance and confidence. This last statement is based upon the fact that the intimate fitting of the thoracopelvic unit serves as a simple, but efficient, feedback system (in the form of alterations of pressure) to the wearer. He now can be aware of the occurence of motion when it *begins* instead of when it's too late for him to maintain control.

There are disadvantages. Elastic materials, as a source of external power, present strict limitations upon orthotic design. The most important to this discussion are:

1. The limits of available power in relation to the need.
2. The practical implications of specific characteristics of elastic materials, e.g., increased elongation means increased power,

whereas the limits of elongation mean limits to freedom of motion. To illustrate: The elastic material selected to assist pelvic extension must have a stretch range sufficient to permit activities such as sitting or climbing stairs and still deliver an amount of power that substantially aids, rather than inhibits, such activities.

3. Due to the need for constant force for purposes of balance, attention must be given to shear forces upon the skin of the thighs and lower trunk, especially upon areas that are insensitive. This consideration, when coupled with the characteristics of elastic materials, has the most profound influence upon the system's design.

Figure 4 also illustrates the functional advantages that the dynamic pelvic assist offers the myelomeningocele child in activities such as walking and climbing stairs. The contralateral limb is shown in 25 deg. of hip flexion, just prior to heel-strike, and again in 60 deg. of hip flexion, as if the foot had just been placed upon a step. As the limb in swing phase flexes forward, the distal attachment point, 'b', of the elastic on the posterior panel of the quadrilateral cuff draws away from the proximal attachment 'a'. This elongates the elastic as it conforms to the changing relationship between the thigh and the buttock and thereby increases its force. The increase in force from distal point 'b' to points 'c' and 'd' would be approximately 1.25 lbs. and 3.25 lbs., respectively, in addition to the preload which is constant. Using the previous example of a preload of 2 lbs. per elastic, the additional force of 1.25 lbs. amounts to a force equal to 35 percent of the trunk's actual weight acting to stabilize the pelvis and trunk on the stance phase side just before heel-strike. As the child who weighs 30 lbs. begins to lift himself up a step, the force to stabilize his pelvis and trunk is equal to 48 percent of the weight of his trunk.

Of equal importance, is the fact that the child receives in addition to these increased stabilizing forces, the benefits of reciprocation during such activities. This result is achieved because, although the pelvic panel allows free rotation in an anterior direction, the total force of the elastic upon the buttock on the swing-phase side (which is now actively resisting execssive anterior pelvic rotation, via the panel), transfers to the buttock on the stance phase side during mid-stance. Also, as the child raises the limb in stance-phase off the floor to lift himself up a step, the force transfers back to the buttock on the side of the weight-bearing limb now upon the step above him. Furthermore, he receives additional force with which to further stabilize his trunk, by flexing his trunk forward, which further increases the stretch upon both elastics, a motion that is 'in phase' with the normal sequence of events during the two activities described.

Previously, the problem of shear forces was mentioned as having a strong influence upon the system's design. Figure 5 shows the addition of free hip joints with posterior stops at 180 deg. Regretfully, the addition of hip joints is the best answer, to date, for two more complex orthotic challenges that some myelomeningocele children present.

First, youngsters with lesions above L3 to as high as T10: When anterior musculature that crosses the hip axes is also paralyzed, it is not possible for these children to stop their trunks from involuntarily rotating in a *posterior* direction as well as in the anterior direction. The free hip joints with posterior stops attached to the thoracopelvic unit prevent posterior rotation of the pelvis beyond neutral position about the hip axes, while providing the patient control of anterior pelvic rotation, without locking devices, by virtue of the dynamic pelvic assist. In order that the thoracopelvic unit may be worn separately, when deemed desirable and to make donning of the total system easier for parents and/or the child, a simple quick-release has been devised to disconnect the above-knee orthoses. The enlargment in the upper left of Figure 5 shows the precontoured polypropylene bar being removed from its slotted receptacle. Note how the polypropylene bar is springing back to its precontoured sharp radius which acts as a spring lock when passed through the slot and forced to press against the much more gradual radius of the thorax portion of the unit.

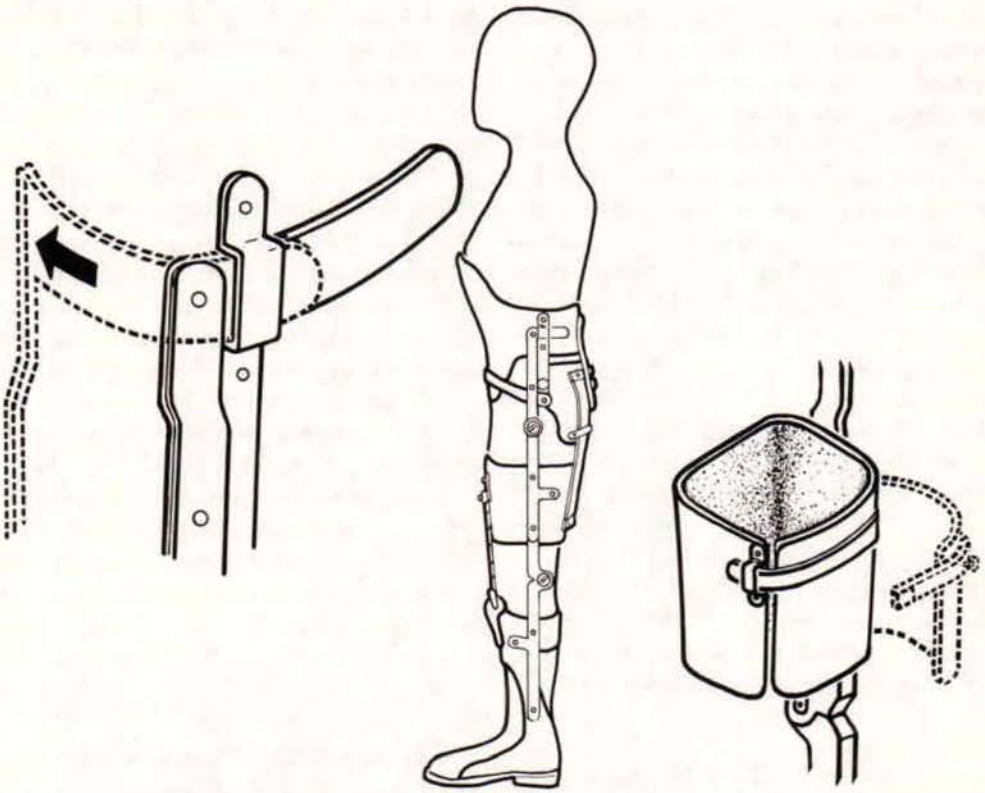

Fig. 5. Schematic lateral view of the components for lesions above L3 to T10: Upper left, detail of quick-release system. Lower right, detail of snap fastener for hinged anterior panel of the quadrilateral thigh cuff.

When the child's limbs are in the orthoses the limbs are rotated externally to insert and internally to remove the orthoses from the thoracopelvic unit.

Second, the larger and heavier children and/or the very young who may need additional help until they learn to use the as yet untried muscles that they do have, and the youngsters who have 5 to 10 degrees of hip flexion contractures *prior* to bracing: All of these cases present a common problem. They need greater amounts of force to keep their trunks upright without resorting to hip locks.

HIP CONTRACTURES

The factors that reflect the limitations upon design, when hip contractures are present, can be demonstrated by referring to Figure 4. In the previous example of the youngster who weighs 30 pounds with a trunk length of 20 inches, it was shown that a force in excess of 8 lbs. (slightly over 4 lbs. per elastic), is needed in order to provide the child with a 10 deg. zone of safety. If a youngster of the same weight and trunk length presents 10 deg. hip contractures, the same force would be required just to prevent the trunk from flexing further forward. However, this would be of little advantage to the child because his trunk balance would remain precarious due to the lack of a 'safety' zone. To provide him with a minimum safety zone of 5 degrees, it would be necessary to preset the force of each elastic to 6.2 lbs. for a total force of pelvic assist of over 12.4 lbs. If a 10 deg. safety zone were attempted, the total pelvic assist force required would be 16.4 pounds. The effect of one characteristic of elastic materials, i.e., an increase in elongation results in an

increase in force, cannot be ignored. Since, when seated, the initial setting or preload is known to increase by a minimum factor of 2, a 10 deg. safety zone setting would then be equal to a minimum of 32.8 lbs., well over the child's total weight. Due to this child's size, the length of elastic is limited to the distance from point "a" to point "b". To obtain such high preload settings from short lengths of elastic requires stretching the elastic so close to its limits of elongation that there would be insufficient stretch left to allow for activities such as sitting. There is no choice than but to reduce the preload setting and we must settle for a safety zone of 5 deg. or less. Such a slim margin of safety, especially when the trunk cannot achieve an upright posture, is too difficult for a small child to master. As we begin to understand the magnitude of the forces that these children's muscularly imbalanced bodies must contend with, the importance of dynamic bracing *before* contractures develop becomes apparent.

OVERWEIGHT CHILDREN

Overweight children without hip contractures, present less of a dilemma, yet their excessive weight does require design considerations that relate to shear forces. For example, a child with the same trunk length of 20 in., who weighs 40 lbs., requires a total preload setting for a 5- or 10-deg. safety zone of 6 and 11.1 lbs. respectively. The child with a preload setting for a 10 deg. safety zone, when seated, will have a total force of 22.2 lbs. acting between the thoracopelvic unit and the quadrilateral thigh cuffs. Although the weight of his trunk and the stretch length of the elastics may be sufficient to allow him to sit down easily, a downward force of 22.2 lbs. upon the thoracopelvic unit and an upward force of 11.1 lbs. upon each quadrilateral thigh cuff translates to shear forces upon the insensitive tissue of these areas that are too high to risk. A single lateral upright with a free hip joint, with or without a posterior stop, as the case requires, must be added bilaterally to eliminate the unwanted shearing effect. The free hip joints do not inhibit the function of the dynamic pelvic extension assist.

THE BEGINNERS

The very young, who are neither overweight nor have hip contractures, may require the addition of free hip joints with 180 deg. posterior stops in order to increase their safety zone to the maximum that the elastic materials will allow without risk of introducing unwanted shear forces to insensitive tissue. The hip joints can be removed later because the preload force maintaining the safety zone can be reduced to a smaller range once the child gains control. Also, as the child grows taller and heavier, longer elastics can be used and greater forces are available to assist him without restricting freedom of motion.

A FURTHER REFINEMENT

A quick release to separate the knee joints and "solid-ankle" portions of the KAFO's from the quadrilateral thigh cuffs is now in development, so as to make one system serve as both a day and bedtime orthosis. It is expected that the thoracopelvic unit, with its free hip joints extending to the quadrilateral cuffs, can be utilized as a night orthosis and the KAFO's added for daytime use. This design is intended for the youngster who has 5- to 10-deg. hip contractures *prior* to bracing. The idea is that hip contractures may be reduced during sleep when the dynamic pelvic extension assist would not have to contend with the force of gravity. Also, the corrective force acting about the hip axes would have an additional advantage because the anterior muscles and other tissues being stretched would be relaxed during sleep. Thus, contracted tissue could be placed on stretch around the clock. It is hoped that, with this approach, *time* and *growth*, two factors that have been so destructive in the past, may be brought under control for the myelomeningocele child.

CONSIDERATIONS FOR A VOLUNTARY HIP RELEASE MECHANISM

The preceding discussion may well prompt a question as to why the system does not provide a means for the wearer to voluntarily release the elastic tension of the pelvic assist. While it must be conceded that a greater preload could be applied and thereby increase the range of the safety zone if a release were available to disengage before sitting, the decision *not* to provide a release is a matter of 'tradeoffs'. A release would deny these children valuable automatic assistance and/or control in getting up and down from a chair, play position or fall. Furthermore, as has been discussed, without hip joints, the amount of increase to the preload is limited strictly by the necessity of avoiding unacceptable levels of shear forces. However, the importance of keeping the dynamics of the system fully automatic, without the need for conscious effort on the part of the young children for whom the system is designed, in my view, far outweighs all other considerations.

The enlargement in the lower right hand corner of Figure 5 shows details of the polypropylene snap fastener that secures the hinged anterior panel of the polypropylene quadrilateral cuff. Its purpose is three-fold:

1. To ensure that there is no slackness in the fit of the cuffs, especially in the anterioposterior plane, which would affect the accuracy of feedback being received from the system via the thoracopelvic unit.

2. To stabilize the anterior panels of the cuffs against the strong pull of the knee extension assist since the cuffs serve as the proximal attachment points for these units.

3. For ease of removal and donning and to ensure that the cuffs are worn in the same manner each and every time they are applied, because of their importance to the accuracy of feedback.

Figure 6 demonstrates how the knee extension assist unit is designed to 'program' its pivoting action so that the line of force generated by the elastic shockcord changes from an extension moment about the knee and mechanical axes, through both centers as the knee is flexed and drops below both centers to become an efficient knee *flexor* in the seated position. The advantage to the wearer is an assistive force equal to 15 percent of the weight of the trunk and thighs to aid in both rising from and descending into a chair. The flexion moments stabilize the lower legs, when rising from a chair, until the point when a sufficient amount of the body weight is above them to take over and also it prevents the occurrence of unwanted extension of the lower legs when seated. Figure 7 shows a detailed view of the dynamic knee extension assist unit.

Experience to date with the dynamic orthotic system indicates that it is technically feasible to improve the effectiveness of the orthotic contribution to the overall management of myelomeningocele children. From a biomechanical point of view, the key appears to be *prevention* by introducing dynamic external forces sufficient to maintain balance and control, in a manner that enables innervated muscles to develop to their full potential in both strength and function without causing deformities. It can be demonstrated that a surprisingly low magnitude of force is sufficient to maintain balance about a joint that is muscularly imbalanced and which has a normal range of motion. Also, as has been shown, adding a mild increase to the balancing force across a mobile joint provides a range or 'zone of safety' that allows movement within the safety range without jeopardizing the patient's overall balance. However, even a mild degree of contracture, e.g., about the hip joints, brings kinetic forces into play that add to the dimensions of the problem of maintaining balance with sufficient freedom to perform daily activities. Early application of a dynamic system would seem to suggest the direction toward prevention of musculoskeletal deformities that develop after birth.

SUMMARY

A lightweight, dynamic orthotic system has been developed for the myelomeningocele

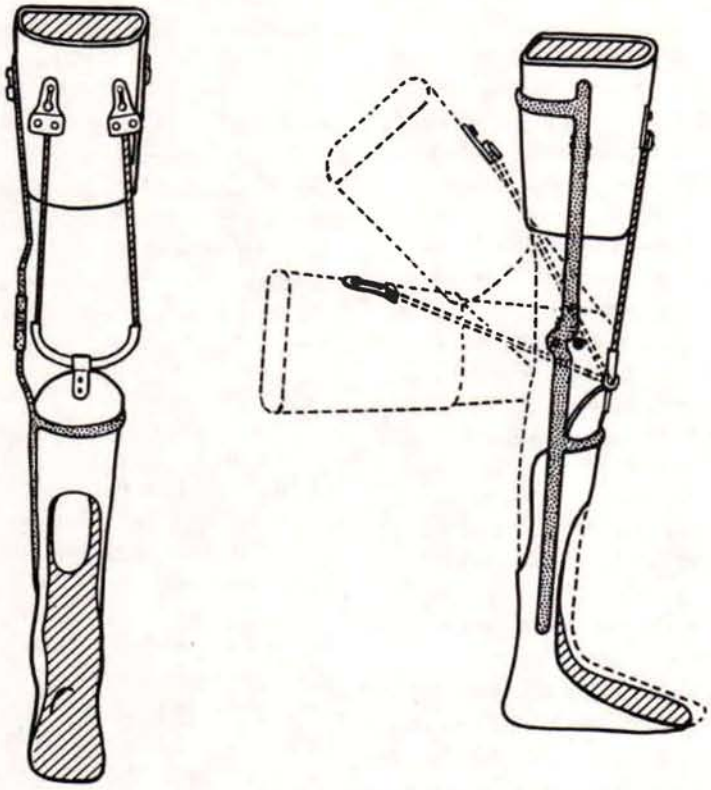

Fig. 6. Schematic of the dynamic knee extension assist: Left, anterior view showing the unit attached to the orthosis. Right, lateral view illustrating the pivotable action of the knee extension assist unit that shows how the force changes from an extension moment in the upright position to a flexion moment in the seated position. The solid dot represents the axis of the anatomic knee joint.

child. The system is modular in design in order that a variety of components can be 'plugged' into it to meet the multiple needs that each level of lesion presents. The system is fully automatic in that it provides the wearer the following functions, without conscious effort on his part: overall body balance parameters with freedom of movement for daily activities; dynamic forces that act reciprocally during gait; assistance and/or control to and from a chair, play position and in climbing stairs; assists pelvic and knee extension without locking devices. The system has additional uses, e.g., the dynamic pelvic extension assist components can be used post-operatively to maintain surgical releases of hip flexion contractures and/or as bedtime orthoses to reduce mild hip contractures during sleep. Also, the knee extension assist components can be used in either manner for knee flexion contractures. The overall design as well as the rationale for its use is described in detail.

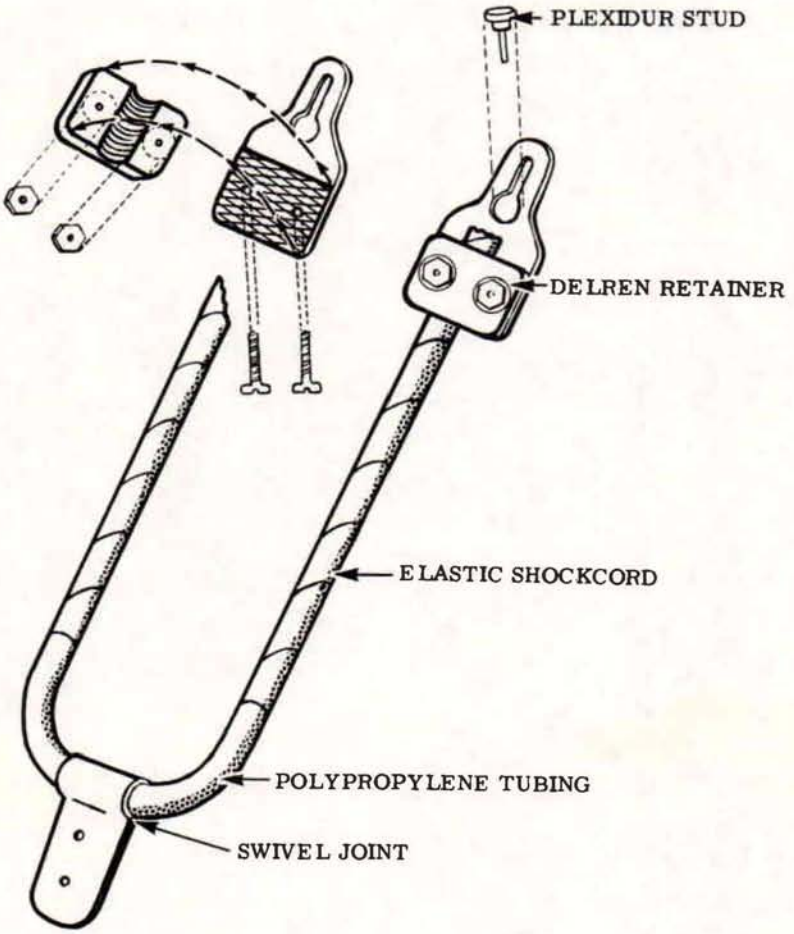

Fig. 7. Details of the dynamic knee extension assist unit.

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THE RUBBER SLEEVE SUSPENSION FOR BELOW-KNEE PROSTHESES

Joseph P. Giacinto, C.P.[1]

In various areas of the world there seem to be preferences for just one of the many types of prosthetic devices that have been developed to solve a given problem. This we can attribute to "human nature." Perhaps we ourselves may be from the same mold, and therefore would like to share one of our experiences with you.

In spite of the various auxiliary suspension aids that have been developed for below-knee prostheses (*6*) friction between the amputee and the prostheses still remains a problem. Aids such as stump socks, shoulder straps, fork straps, thigh lacers, condylar straps (*6*), condylar wedges (*2, 5*) and suction (*4*) with valves (*3*) have been used and all with very good results. We have one other: the rubber-sleeve suspension.

THE RUBBER SLEEVE SUSPENSION

The rubber sleeve for suspension of the below-knee prosthesis was conceived and developed at the Orthotic and Prosthetic Facility, University of Michigan Medical Center, Ann Arbor, Michigan.[2] On a below-knee prosthesis, a rubber sleeve is applied so that it extends from the prosthesis to the patient, and in a manner that a negative pressure is developed (*1*) to create a suction-type below-knee prosthesis. Well over 450 patients have been fitted with this type of suspension here during the past six years.

The rubber sleeve can be adapted to any below-knee prosthesis that does not have mechanical joints or straps attached.

The first reactions and comments from patients that have worn other suspension aids before using the rubber sleeve are that piston action is absent or minimal, the prosthesis feels lighter in weight, and it seems to be more of a intimate part of him.

The rubber sleeve should be applied to the prosthesis so that the bulge of the rubber is in the area of the patella. It should be rolled or pulled up (Fig. 1) onto the flesh to cover approximately 5 to 10 cm. of the upper thigh, well above the stump sock (Fig. 2). The patient is taught to expel all of the air that is captured within the sleeve so that it fits snugly with no creases or wrinkles (Fig. 1). This constitutes a suction-type below-knee prosthesis (*7*).

The rubber sleeve is available in three sizes

Fig. 1. Application of the rubber-sleeve suspension for a BK prosthesis.

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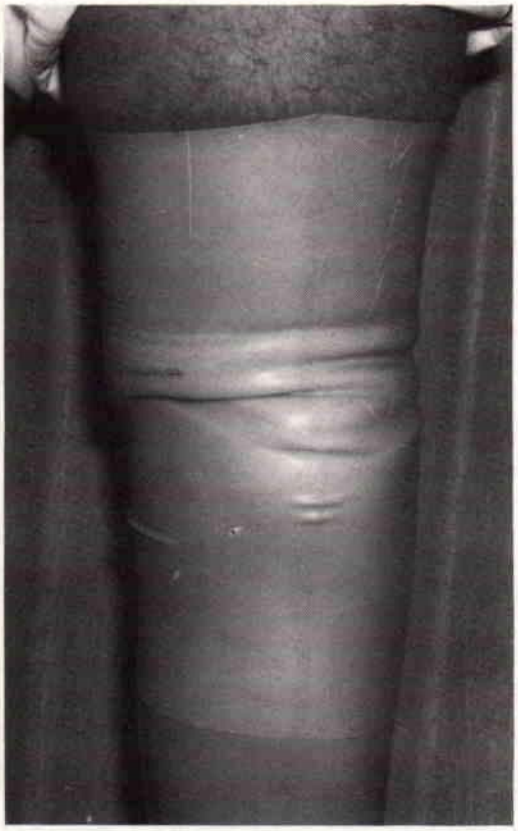

Fig. 2. The rubber-sleeve suspension in place.

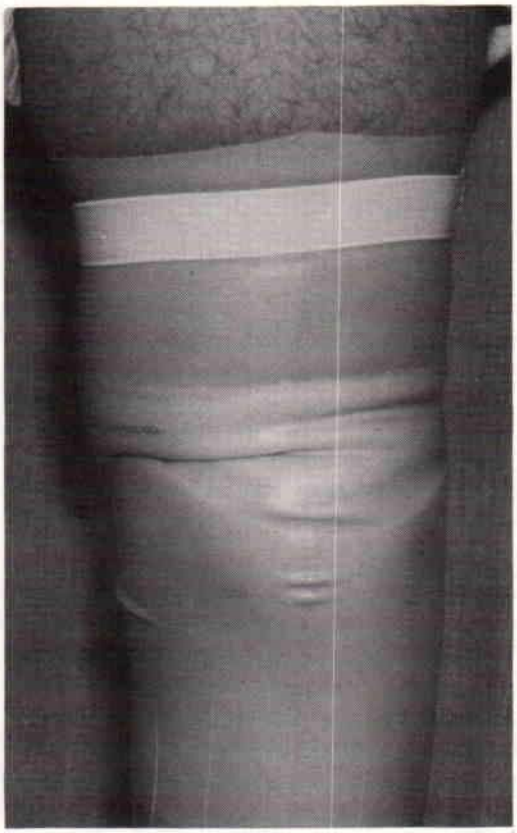

Fig. 3. A lightweight elastic garter with Velcro tabs may be used if the sleeve is oversize.

The patient should have several rubber sleeves on hand. Three or four should cost no more than a "U" strap and waist belt. When the rubber sleeve is torn or punctured, the only suspension remaining is due to the elasticity of the rubber and there will be some slippage and piston action.

Good hygiene is necessary. The rubber sleeve that comes in contact with the flesh, should be cleaned with rubbing alcohol and powdered with talcum. When an insert is used and becomes damp with perspiration, it should be dried by taking it out of the prosthesis and set out to dry in the air.

Because the patient is fitted with stump socks as in conventional methods, bony prominences and hypertrophy are of little problem. The patient's prosthesis can be padded in the usual method and with the rubber sleeve suspension still be a suction prosthesis.

We have yet to see an edematous condition caused by the rubber sleeve and suction. Heat and perspiration give no more problems than do suction socket prothesis for the above-knee amputees when kept clean.

We hope this article will help many prosthetists in their prosthetic suspension and fitting problems, and thus many patients.

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MANAGEMENT OF SHORT ABOVE-KNEE AMPUTEES[1]

W. G. Biedermann[2]

The supply of prostheses for short above-knee amputees is still not a routine matter even today. A number of examples of the difficulties which may be encountered by the practitioner, and the possibilities of overcoming them will be demonstrated.

I wish to stress that I am not in favor of mechanical hip joints because of their weight and generally unsatisfactory position when the wearer is sitting, and use them only in a limited number of cases when there is stiffness in the hip joint. The weight is of particular importance in the case of short stumps. Whenever feasible, an attempt should be made to find other solutions for this group of amputees. It is necessary, however, that all the techniques commonly used above the knee be modified, since short stumps demand their own rules.

Oval or quadrilateral shapes are of no help, the more so since it is still not known which shape is ideal. The short stump with muscular conditions that are unfavorable for the fitting of prostheses requires a special, individual shape. Because of the small area and poor leverage of the stump, the structural conditions are necessarily different also. The abduction and flexion contractures usually present reduce security and must be taken into account in the structure, unlike in standard leg prostheses. Greater security can be achieved by increased plantar flexion, and by use of special joints that are as physiologic as possible. The amputee with the short above-knee stump requires far greater security, since the features of his stump never offer optimal conditions. It will be demonstrated on a number of cases how individually different short stumps can be managed, although I am aware that these developments are not yet definitive. Because of the shape of the total-contact socket in which the medial and lateral brims work comprehensively, and in which, in some circumstances, the gluteal musculature is also partially included, a synthetic material is especially advantageous. The use of contact rings under loading is necessary for making the plaster impression.

Case 1

The first patient is a 65-year-old man, 193 cm. tall, with a right short stump that was reamputated in 1971 by Dr. Dederich in Bonn. The patient had worn a prosthesis constantly, yet the condition of the stump was extremely poor. Because of the myoplastic operation, the function of his stump was improved considerably and therewith initial conditions were more favorable for further prosthetic management (Fig. 1). Upon flexion, the shape of the stump changes appreciably (Fig. 2), which makes the management of short stumps difficult. There is also a slight flexion contracture. For this patient a socket was made of a thermoplastic material.

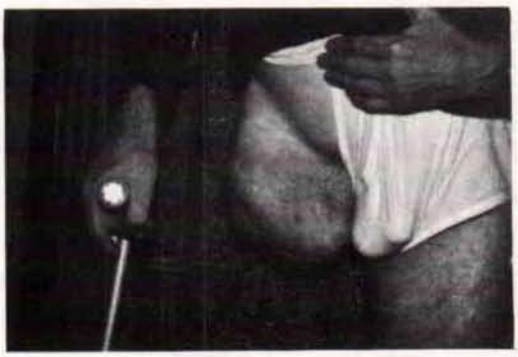

Fig. 1. A 65-year-old patient with a surgically corrected short above-knee stump.

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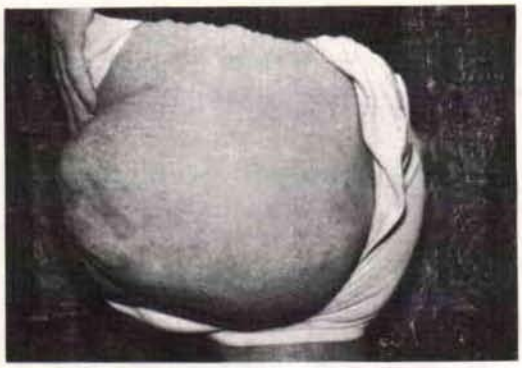

Fig. 2. Changes in the shape of the stump upon flexion.

The lateral edge was left very high (Fig. 3), in the manner of a Silesian bandage, so as to ensure good lateral control and at the same time correct fixation during flexion. In addition, the Silesian bandage provides a high point that is ideal for attachment of the pelvic strap so that it does not slip down even in extreme positions. In order to reduce uncertainty when walking, a hydraulically operated MH-type Habermann foot was used. The long adjustable part was used as the knee. This knee-foot combination proved to be a very satisfactory solution. The socket is actually fairly conventional in its basic shape. It differs from the conventional only by the drawn-up outside wall. Figures 4 and 5 show the finished prosthesis. To provide sufficient alignment stability the socket was placed far anteriorly. The flexion contracture was

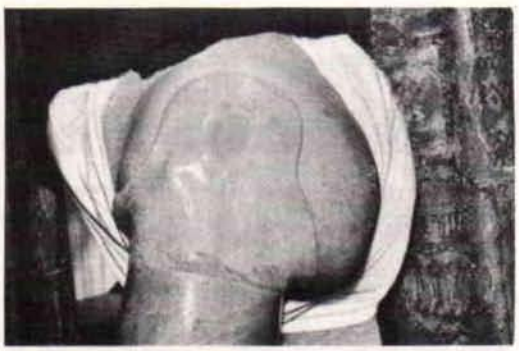

Fig. 3. The socket for the patient shown in Figure 1. The lateral wall is extended proximally to improve stability and control.

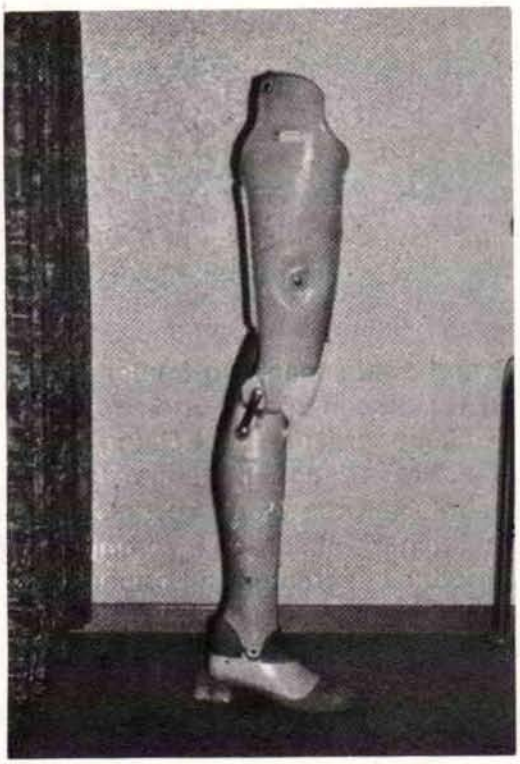

Fig. 4. Finished above-knee prosthesis for the patient in Figure 1. A Lang knee and a type MH Habermann foot are used.

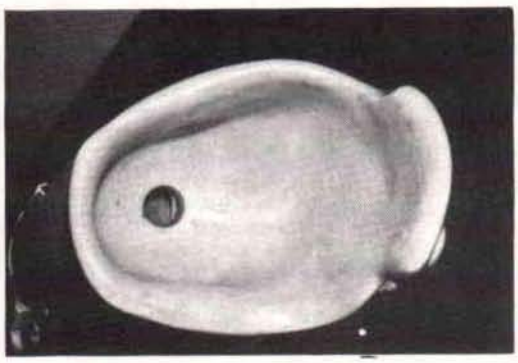

Fig. 5. Proximal view into the socket.

taken into account when positioning the socket with respect to the centerline of the thigh, so that the artificial leg can hardly be distinguished from the outside from a normal above-knee prosthesis.

Case 2

The second patient was a 55-year-old man with an amputation on the left and who had worn a prosthesis continuously for some time, but had been unable to wear one at all for the past three years owing to severe pains in the stump. The stump is very short and consists essentially of fatty tissue at the distal end (Fig. 6). The amputee could not bring himself to have

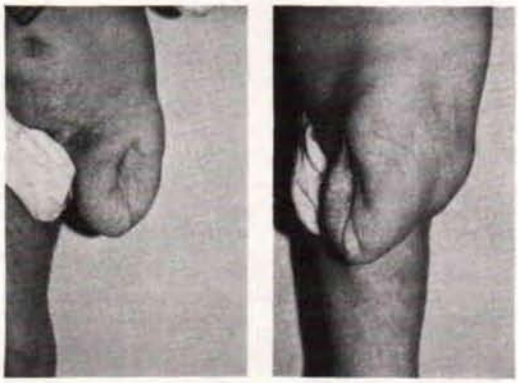

Fig. 6. Front and lateral views of a 55-year-old amputee with short stump.

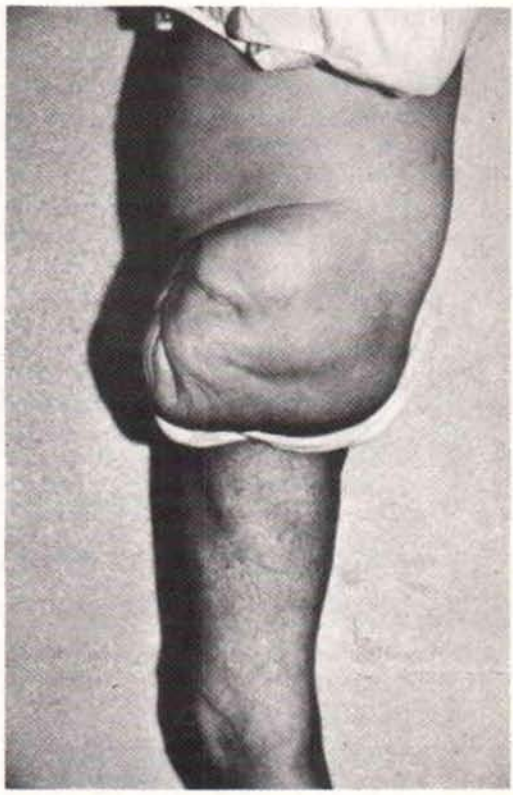

Fig. 7. Changes in the stump as a result of hip flexion.

a revision performed but as a result of wearing crutches had involved his hands so severely that some solution had to be found to make it possible for him to resume wearing a prosthesis. In this case, also, there was a complete change in the shape of the stump during flexion (Fig. 7). The injured man had been furnished earlier with a wooden socket, a Röck knee, a single-axis foot and a Silesian bandage; later, with a plastic total-contact socket, Lang knee, and a Greissinger foot. He was then furnished with a plastic total-contact socket (Fig. 8) which incorporated a highly elaborate Silesian bandage. The artifical leg itself was constructed with modular components, using a Greissinger foot (Fig. 9). This type of prosthesis was a good starting point not only with respect to cosmesis, but also with respect to weight. Despite the wobbly fatty mass at the end of the stump, the total-contact socket proved to be satisfactory, since this tissue is also used for adhesion and

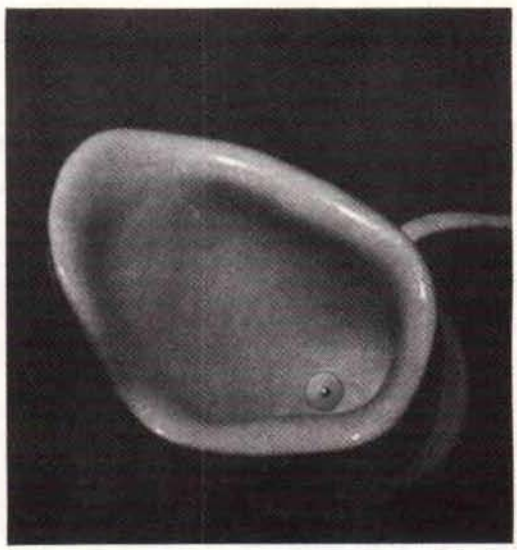

Fig. 8. Total-contact socket with a high lateral wall for the patient shown in Figure 6.

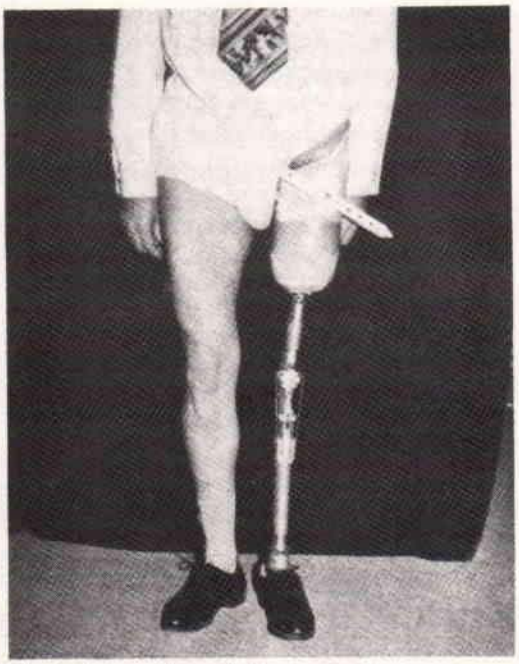

Fig. 9. Patient with prosthesis using modular components.

weight-bearing. A further advantage of the total-contact socket is that slipping out of the socket during flexion is prevented (Fig. 10). Since the stump is very bony proximally, the trochanter was very generously embedded, particularly since it only revealed a narrow weight-bearing surface at the upper edge. Figure 11 shows the finished prosthesis with cosmetic covering.

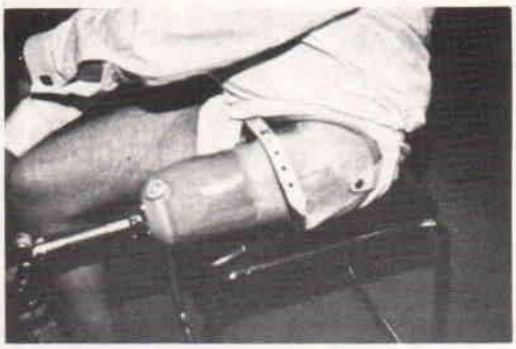

Fig. 10. The total-contact socket prevents the stump from slipping out during flexion.

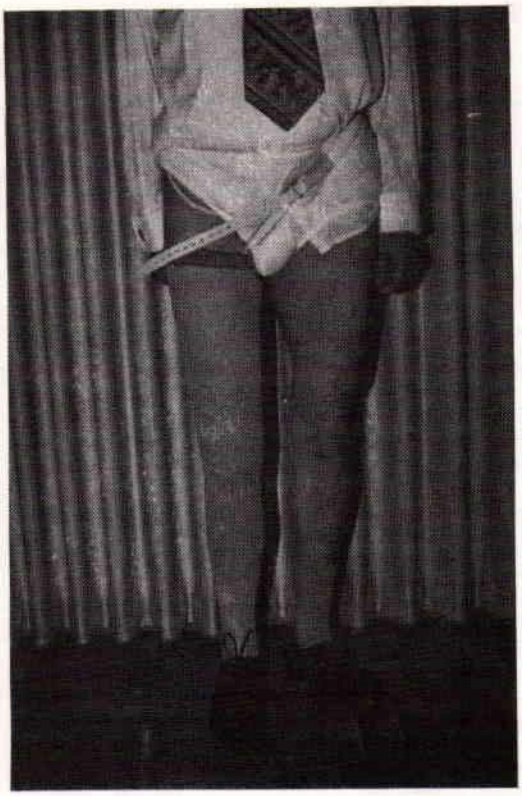

Fig. 11. Finished prosthesis with cosmetic cover.

Case 3

Case 3 is 60 years of age, and although he lost a leg in World War II he was fitted here for the first time. The stump has several scars resulting from a gunshot wound (Fig. 12). He also has large scars on the lower abdomen and spine. This man had never been furnished with a prosthesis and had moved around for 30 years on crutches. At the time of fitting he was almost incapable of movement. His only chance of avoiding a wheel chair was to use a prosthesis. In his case everything that could cause trouble and anxiety had coincided: not only are there scars, but also splinters are still in the stump. In addition, severe flexion and abduction contractures are present. Therefore, at first it was not known whether this man could be fitted with a prosthesis at all.

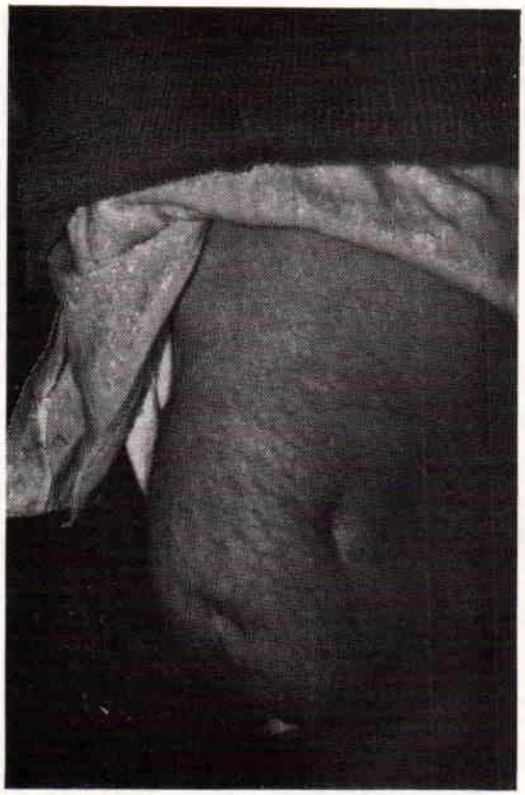

Fig. 12. A 60-year-old amputee with a short stump covered with scars.

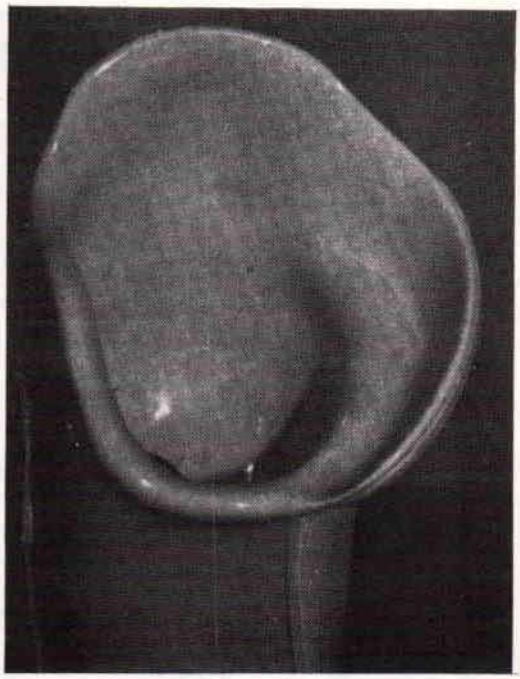

Fig. 13. Complete total-contact socket for the patient shown in Figure 12. Provision is made for partial inclusion of the gluteus.

Finally he was fitted with a plastic total-contact socket which rose very highly laterally and anteriorly, and partially included the gluteus (Fig. 13). A further problem was that in order to bend, the spine had to be exposed, since otherwise the stump would have been levered out over this pressure point in a bending position. The bed of the ischial tuberosity had to be cushioned with a leather covering (Fig. 14) so as to counteract the sensitivity of the tuber. A Lang knee and a Greissinger foot were used.

First we let the patient move about with the artificial leg in the unfinished state (Fig. 15). It can be seen from the various cuts that the position was changed several times, since the original flexion contraction decreased with increasing use and as the patient acquired an

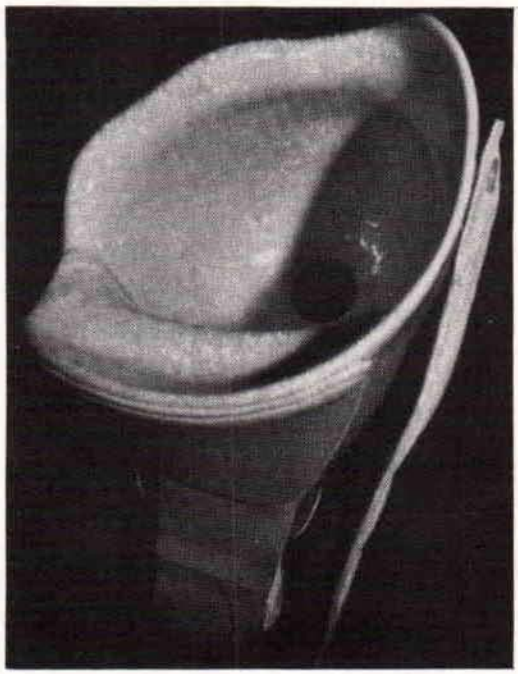

Fig. 14. Provisions made for cushioning the ischial shelf.

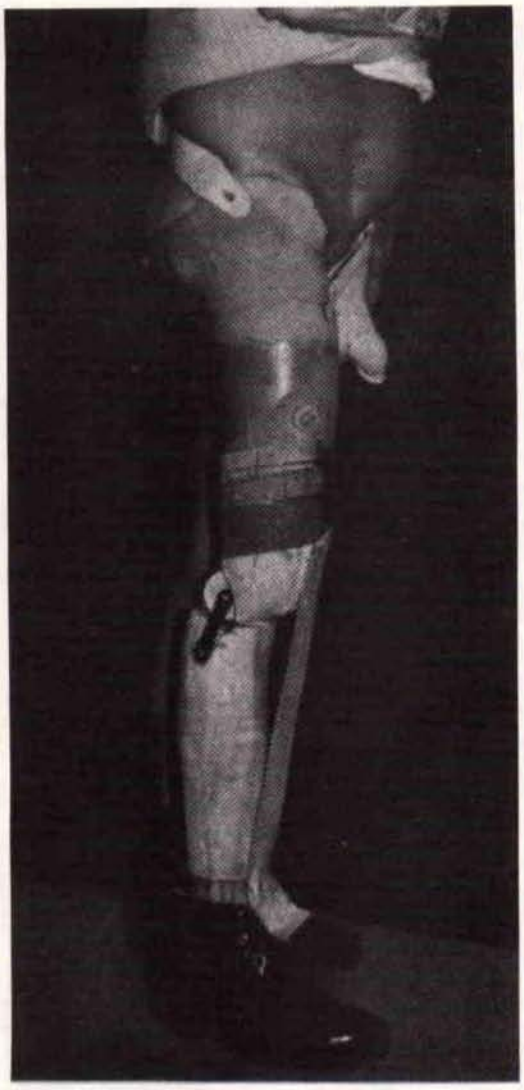

Fig. 15. Unfinished artificial leg for the amputee shown in Figure 12. A Lang knee and Greissinger foot are used.

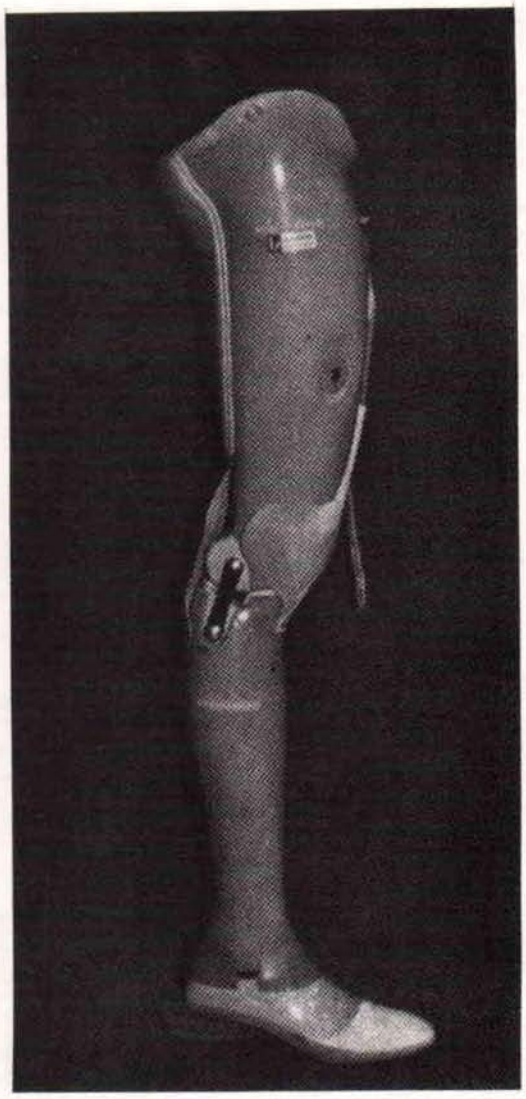

Fig. 16. The finished prosthesis.

increased sense of security on the ground, so that it was even possible to omit the exaggerated safety position of the knee. For suspension a simple pelvic belt was used. I am now of the opinion that a suspensory belt with rotating straps should never be used for short stumps.

This case involved much work, but also gave much pleasure, since a successful fitting encourages new experiments. The finished prosthesis is shown in Figure 16.

Case 4

Case 4 is a triple amputee with a short above-knee stump on the left (Fig. 17), as well as an above-the-elbow amputation on the left arm and below-the-elbow on the right. The patient has Sauerbruch prostheses on both arms. Of necessity the amputee is a constant wearer of prostheses, since without them he is completely helpless. In addition, the patient lives in a

mountainous area and for walking he depends on swinging motion and balance from his arm prostheses. In other words, this amputee, in order to be able to walk normally and safely, needs not only his leg prosthesis, but also his artificial arms. The patient is rehabilitated to the point that he lives completely independently; he

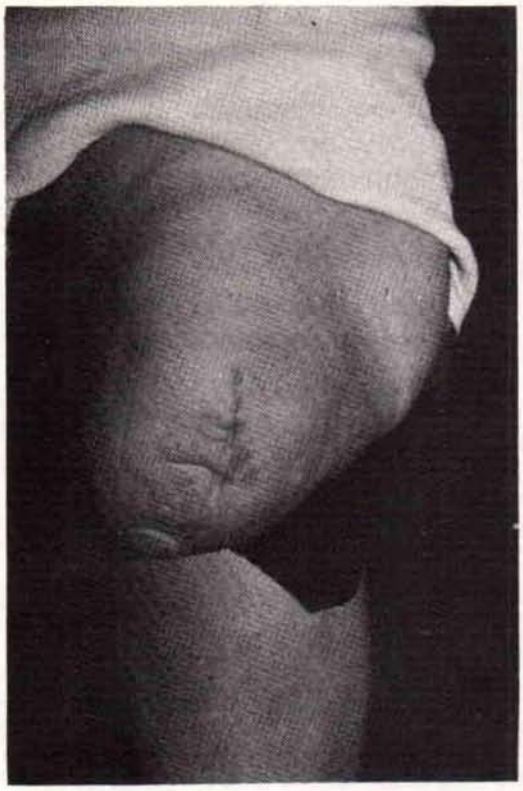

Fig. 17. Patient with a left short above-knee stump. The patient also had his left arm amputated above the elbow and the right one below the elbow.

is employed and drives by car alone to his place of work. The condition of his stump is very good despite its short length and there is no problem with the muscular covering. The stump changes in shape only slightly between the extended and flexed positions (Fig. 18).

The patient was fitted with a plastic total-contact socket with a pelvic belt, Lang knee, and Greissinger foot (Fig. 19). In order to reduce the weight of the prosthesis the lower

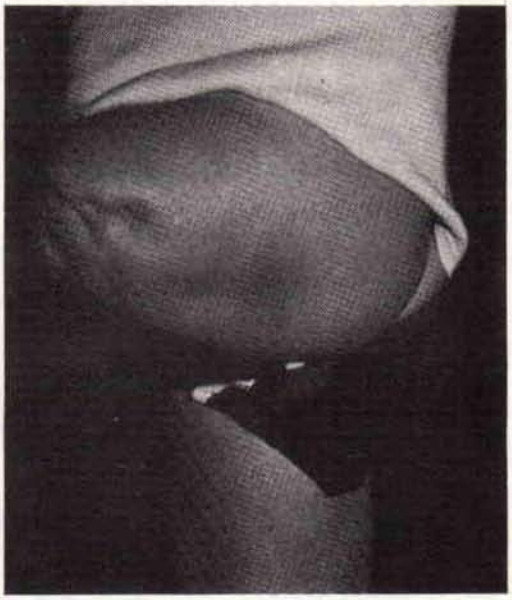

Fig. 18. The stump of the amputee shown in Figure 17 reveals only a slight change in shape upon flexion.

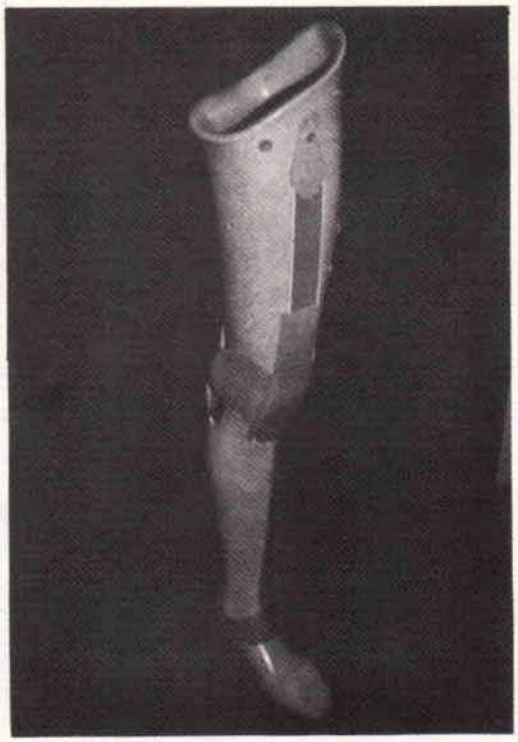

Fig. 19. Leg prosthesis prepared for the triple amputee. A plastic total contact socket, a Lang knee and a Greissinger foot are used.

shank was also molded of plastic. The foot was locked in a pronounced dorsal position and the upper socket was placed far posteriorly for security. In this case the shape of the above-knee socket was conventional and differed little from a standard prosthesis (Fig. 20).

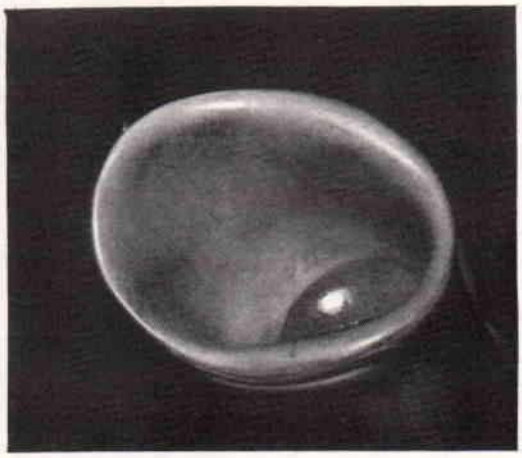

Fig. 20. Top view of the conventional above-knee socket for the patient shown in Figure 17.

Case 5

The last case to be presented is that of a 40-year-old amputee who lost both his legs in an accident (Fig. 21). He has a good, normal stump on the right, but on the left the stump is extremely short, scarred, and has to be regarded as very difficult to fit. The amputee moved around on his hands, and had worn no prosthesis for 10 years. Yet he was very mobile and active.

Technically, the fitting of the right stump presented no difficulties. For the right an artificial leg was constructed with a plastic upper part, a Röck knee, and a Greissinger foot (Fig. 22). For the short stump a synthetic above-knee total-contact socket with a Jüpa knee and a Greissinger foot was used (Fig. 22). A Jüpa knee was selected because the knee can be locked under loads to give a greater feeling of security.

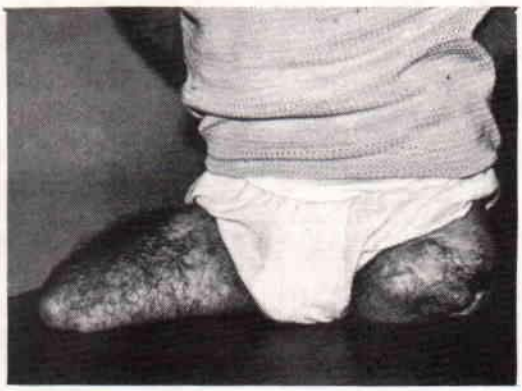

Fig. 21. Patient with bilateral thigh amputations with an extremely short, severely scarred above-knee stump on the left.

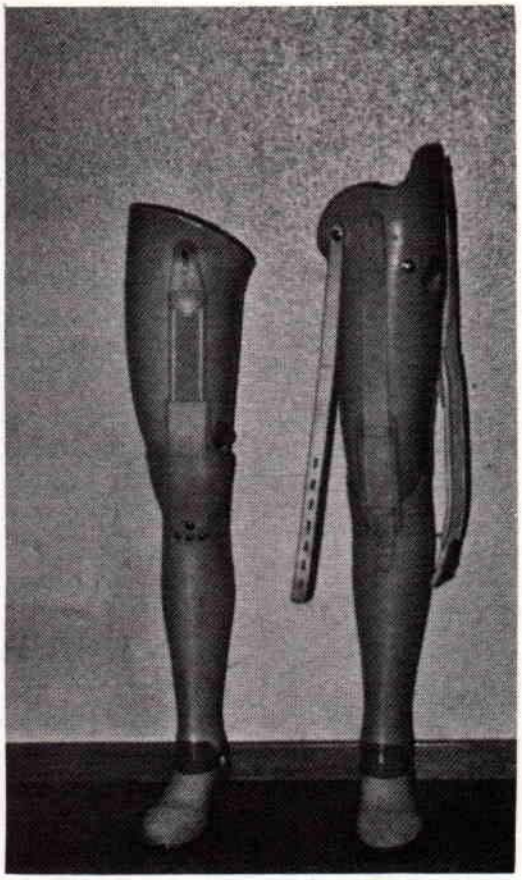

Fig. 22. Prostheses prepared for the bilateral above-knee amputee. On the right a prosthesis with a Jüpa knee and Greissinger foot is used.

In order to improve the control, the entire posterior portion of the stump from the trochanter to the gluteus was embedded very high and the anterior side was extended proximally. The spine required particular attention (Fig. 23). Because it is doubtful that the prosthesis could be suspended by the stump alone while bending or moving, a simple suspension strap with a pelvic belt was used (Fig. 24).

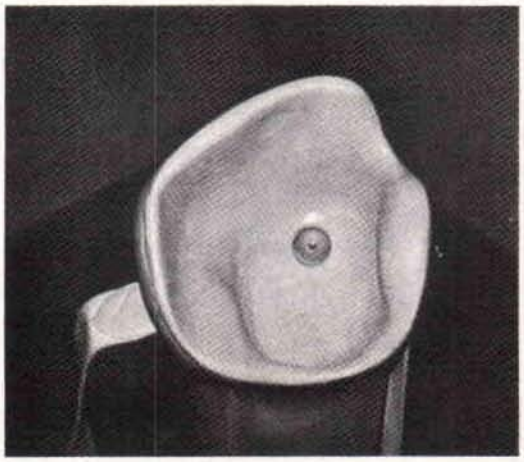

Fig. 23. Top view of the socket for the short stump. To improve control, the area from the trochanter major to the gluteus and the anterior side are embedded high proximally.

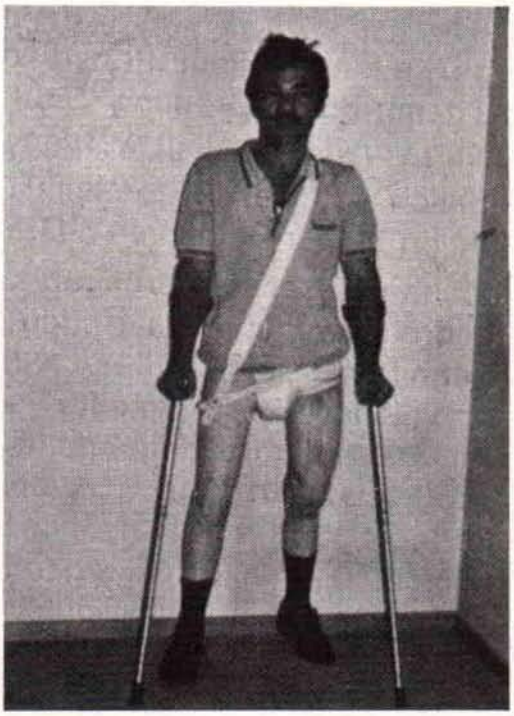

Fig. 24. Bilateral leg amputee shown in Figure 21 with both prostheses. A suspension strap connected to a pelvic band aids suspension for the prosthesis for the short stump.

During the fitting of this patient the question naturally arose whether, if he was to be furnished with two prostheses, he should be fitted in accordance with his former height or not. In testing the rough prototypes we first used prostheses that were 15 cm. shorter. However, at the patient's request, after several fittings the prostheses were adjusted to his former height, since it was his desire not to have to view everything from below as he had done for 10 years, but to be of normal height again. This is important in rehabilitation. From the beginning of wearing his prostheses the patient was able to move forward using crutches without outside help.

CONCLUSION

In closing, the question arises whether the fitting of short above-knee stumps still represents a problem today. In general the question must be answered in the affirmative, since each case is different. Each case presents a considerable degree of difficulty and in each it is necessary to analyze the individual conditions of the stump to make the best of what is left. We do not always achieve the desired success, but even partial successes are on the positive side. In our profession, in particular, and in this age of mass production, the supply of prostheses to difficult or even "hopeless" cases is a rewarding task. We are indebted to the past greatness of our profession for solutions to such difficult problems.

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USE OF THERMOPLASTIC COMPONENTS IN TEMPORARY PROSTHESES
A PROGRESS REPORT

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David Knighton[1]

In December of 1974, we reported on our early work with polyvinylchloride (PVC) pylon systems for the below-knee amputee (*1*). Since then, development of this system has continued and been expanded along other lines. The purpose of this article is to report our work and experience with the PVC-pylon system to date.

BASIC BK PYLON

At the end of the previous report, preliminary work on a new pylon system was described. The new system (Fig. 1) has been standardized and is being used routinely in our clinical practice. A female receptacle is fabricated from a slip-type, double-ended coupling that is commercially available. One end of the coupling is shortened and three pieces of galvanized pipe-strapping are attached with pop-rivets for lamination into either a standard plaster-of-Paris rigid dressing or a plastic PTB socket. A PVC pylon and an ankle plug (turned from PVC on a lathe) are attached in the manners previously described (*1*).

With experience, it is possible to attain satisfactory alignment at the time of initial fabrication. Subsequent changes can be made easily, though, with a heat gun when necessary.

These pylons have been used by patients for periods up to and exceeding six months with only two incidences of failure (one of them inexplicably in the case of a very petite young lady). The only problem has been the occasional development of an annoying squeak between the receptacle and the pylon. Work continues on this problem.

COSMETIC COVER

Clinical experience has demonstrated the desirability of extended periods of treatment with temporary prostheses. In this instance, cosmetic appearance of the prosthesis can become an important factor in patient acceptance. The need, therefore, is for an inexpensive, cosmetic covering that can be fabricated quickly for use with the basic pylon. Considerable success has

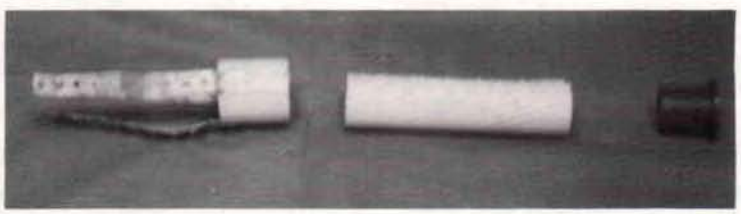

Fig. 1. Components of the new pylon system. From left to right: the female receptacle with galvanized iron straps, the PVC tube, and the PVC ankle plug.

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been attained by foaming a covering in place using a standard polyurethane foam (Otto Bock 617H12). This foam can be given color with standard laminating pigments, and it can be shaped and smoothed in the routine fashion. A separator plate is used between the SACH foot and ankle plug (Fig. 2). A support hose or length of colored stockinette can be applied over the shaped foam covering (Fig. 3). The entire procedure can be carried out in 30–45 minutes making it feasible to be done while the patient waits (Fig. 4). However, for various reasons (basically lack of patient demand) this has not been carried out routinely.

Flexible foam (Otto Bock 617H32) has been used experimentally in one instance to fabricate a cover, but due to the small cross-sectional area between the outer wall and the socket and pylon lying beneath, very little resiliency was achieved and the experiment was abandoned.

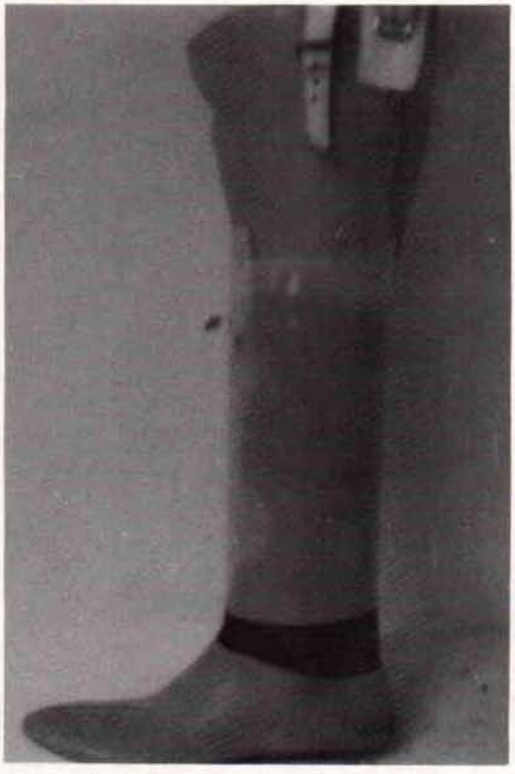

Fig. 2. Lateral view of prosthesis during fabrication, just after foaming cosmetic filler in place.

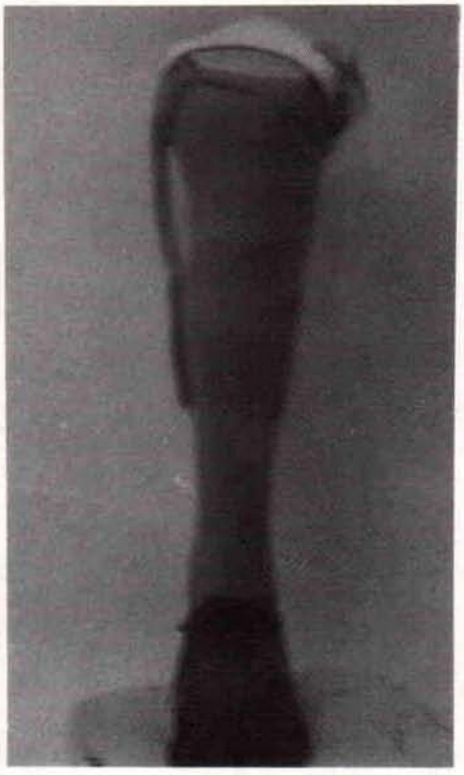

Fig. 3. Finished prosthesis with support hose in place.

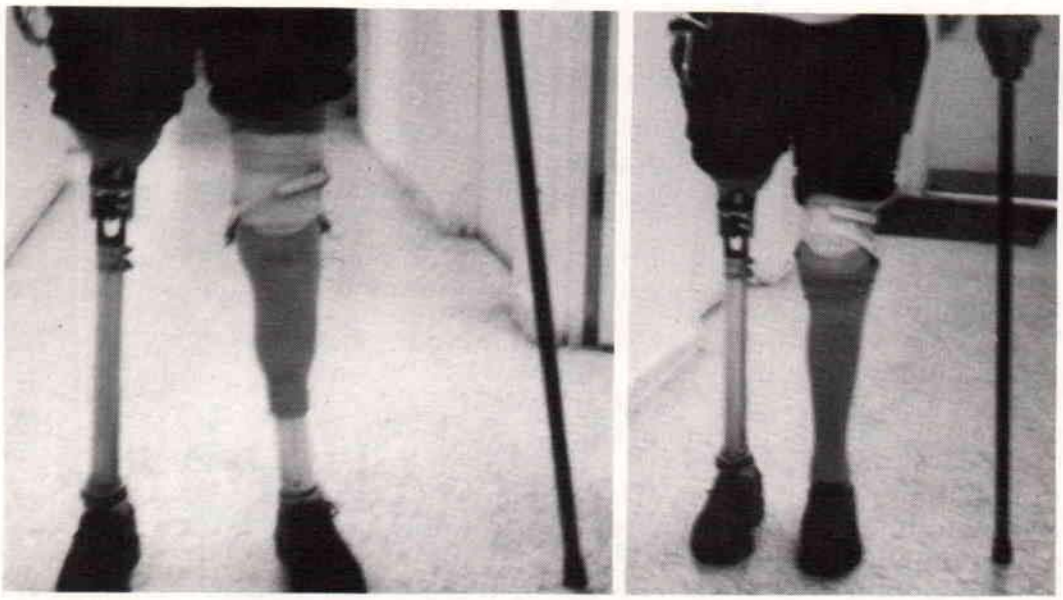

Fig. 4. Two views of the same BK prosthesis. Left, without foamed cosmetic filler; right, with filler and support hose for cosmetic cover.

PYLON SYSTEM FOR THE BELOW-ELBOW CASE

In the past, we have routinely used Polysar® tubing to connect a wrist unit and a polyester socket in the fabrication of temporary below-elbow prostheses. Recently, in the fashion described by Sumida and the group at the Child Amputee Prosthetics Project at UCLA (*2, 3*), we have been experimenting with the substitution of Polysar with a PVC system, using commercially available 3/4-in. I.D. tubing and fittings (Fig. 5). A female receptacle is fabricated in the same manner as the one used with the BK prosthesis and laminated into the socket. Control is achieved at the wrist by splitting the tubing, applying a hose clamp, and using a specially fabricated adaptor applied to the stud of the terminal device (Fig. 6).

Fig. 5. Temporary below-elbow prosthesis using a PVC tube for structure.

Experience with this system has been confined to one case and thus far is inconclusive, but it is hoped it will prove to be a practical alternative to the rather expensive and often unsightly Polysar. Certainly it should be more practical for use with rigid dressings than the techniques used earlier of attaching metal straps to a wrist unit and wrapping the assembly into a rigid dressing with plaster-of-Paris bandage.

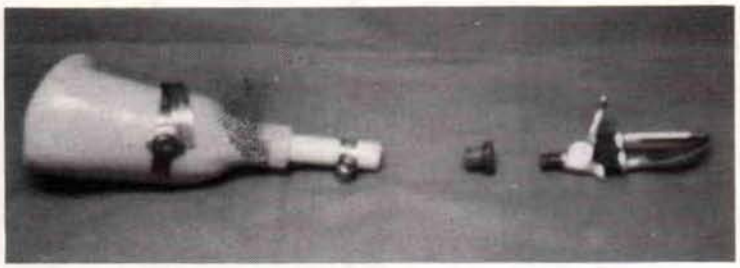

Fig. 6. Components of the temporary below-elbow prosthesis.

CONCLUSIONS

With the emphasis in contemporary prosthetics on the use on temporary or preparatory prostheses in the early periods of stump maturation and with what seems to be a trend (at least in our clinic setting) to even longer periods of use of such systems, the need is for readily fabricated, inexpensive, and durable prostheses. This article describes one such approach and possible elaborations on it. Conceivably, the goals of inexpensive and rapid fabrication would be facilitated if it proved to be practical to use a vacuum formed socket (not necessarily of polycarbonate) with the system and this matter is currently under consideration (Fig. 7).

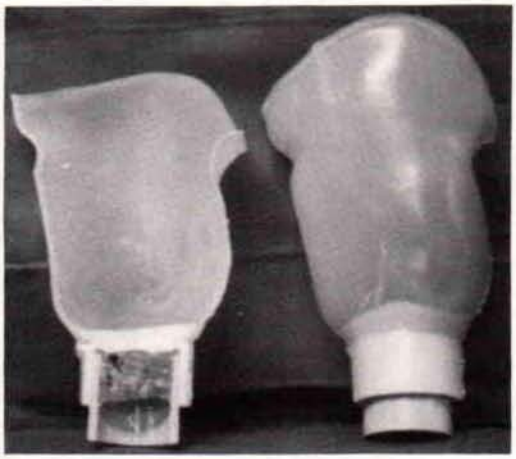

Fig. 7. Cutaway sections of vacuum-formed below-knee socket used in experiments for development of improved, inexpensive attachment methods.

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APPLICATIONS OF TRANSPARENT SOCKETS

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Among recent trends in prosthetics, perhaps the most significant is the transfer of the prosthetist's work from the "bench" to the clinical setting where he has a far greater interaction with patients. In many clinics this move has been facilitated by the application of modular prostheses (*6*), by introduction of improved plaster techniques (*1*, 2), and by use of transparent check sockets made with vacuum techniques (*4*, *5*). Through improvement of fitting techniques, use of check sockets provide substantial benefits to both the patient and prosthetist (*3*).

The transparent sockets are molded from strong, clear plastic in a heated, soft state. Almost any type of socket can be formed (Fig. 1). The use of transparent plastic provides a relatively inexpensive and expeditious means of direct visual assessment of the total-contact fit of the socket with respect to the stump. Because direct observation of the stump-socket interface

Fig. 1. Polycarbonate transparent sockets. Illustrated from left to right are PTB, PTS, Below-Elbow, Short Above-Elbow, Above-Knee and Symes.

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under static and dynamic conditions is possible, some of the obstacles to optimum fit and training such as stump pain and skin breakdown are substantially reduced.

Caution must be exercised when the patient is allowed to walk in vacuum-formed sockets because repeated application of loads can lead to work hardening and stress failure of the plastic. In short-term applications, this is no problem, particularly when the load-transmitting areas of the socket are reinforced.

MATERIALS AND METHODS

Using pre-heated sheets of Lexan, a polycarbonate developed by General Electric Company, transparent sockets are molded by atmospheric pressure over a positive plaster model of the patient's stump as described by Mooney and Snelson (*4*). Lexan is a strong thermoplastic material with a tensile strength of 6,000 N/cm^2 (8500 psi). Its very high impact strength, light weight, and transparency make polycarbonate highly desirable for fabrication of prosthetic check sockets.

Unfortunately, polycarbonate is hydroscopic. Any water absorbed from air results in bubble formation when the material is molded at elevated vacuum-forming temperatures, thus leading to significant loss of transparency and decreased tensile strength after molding is completed. This tendency toward bubble formation is augmented further by expanding volatiles trapped in the sheet of polycarbonate. Drying and dehydration by heating the sheet material at 135 deg. C. (275 deg. F.) for 36 hours prior to molding will substantially eliminate this undesirable characteristic. Although lower temperatures are also effective, much longer times are required for drying.

Depending on the size, strength, and durability requirements of the socket, a blank sheet of polycarbonate of either 10 mm. (3/8-in.) or 13 mm. (1/2-in.) thickness is selected. It is then cut to a size of 23 x 23 cm's. or 30 x 30 cm's. (9 x 9-in. or 12 x 12-in.) depending upon the size of the socket to be formed. Commonly, molding is initiated at 235 deg. C. (450 deg. F.) at 500-700 mm. (19–27-in.) of mercury vacuum.

Vacuum molding requires no elaborate equipment. Indeed, all needed machinery is readily available in most prosthetics facilities. A low-temperature oven for drying the plastic, a high-temperature oven for softening the plastic, a work stand, a vacuum pump, a ballast tank, valves, and tubing are the only items required. Figure 2 shows schematically the arrangement of the work stand and pumping system with a cross-sectional view of a plaster positive stump model in place.

Soft polycarbonate is pulled by hand over the positive model and contact-sealed against the circumference of the support surface before application of the vacuum through the central port. Firm contact and a tight seal will assure accurate reproduction of the positive model. The support surface on which the positive model rests is disconnected easily and, thus, is interchangeable. Three sizes of 18, 20 and 24 cm. (7, 8 & 9.5-in.) diameter plywood surfaces with soft rounded edges are used to support the various plaster models of the more commonly seen socket sizes. The diameter of the table should be no more than 5 cm. (2-in.) greater than the largest proximal diameter of the positive model. Proper selection of the table with respect to the diameter of the model will minimize "webbing" of the softened plastic and will result in a socket with a more uniform wall thickness.

The work table, pump, and vacuum tank may be arranged on a compact frame and should be located near the ovens in an easily accessible position. A simple arrangement is illustrated in Fig. 3.

At the start of the fabrication process, the positive plaster model is covered with a thin nylon stockinet to aid in air evacuation and positive mold breakout after socket formation. A blank sheet of polycarbonate that has been dried adequately is placed in a square metal

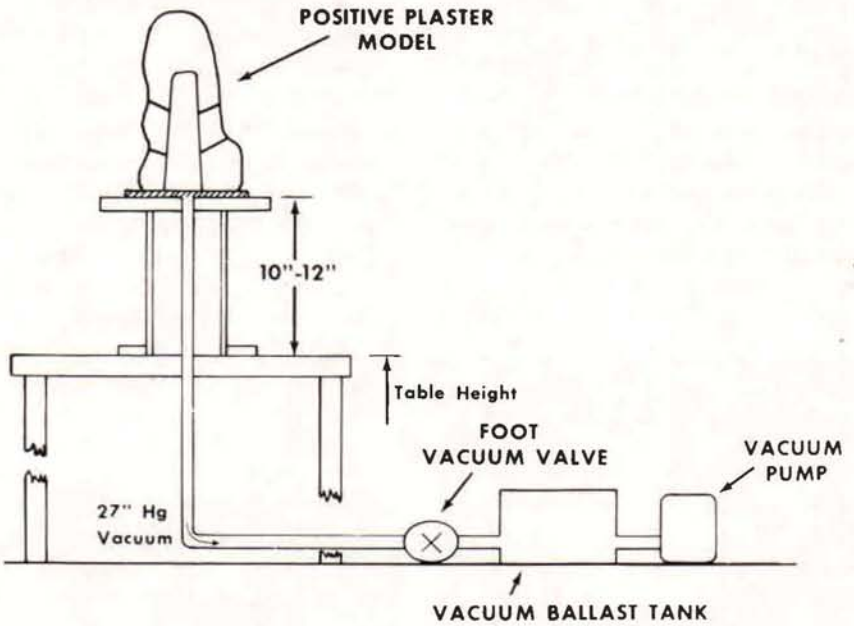

Fig. 2. Schematic drawing of the vacuum forming apparatus. Polycarbonate on the positive stump model is not shown.

Fig. 3. Work stand with stump model on 20 cm. (8-in.) diameter support surface. Two controls are provided, the vacuum foot valve close to the floor and the electrical on-off switch for the pump near the top. Parts of the pump and ballast tank are visible at the center.

frame which is placed horizontally in a high-temperature oven. When the material begins to melt and sags to a point between 1/2 and 2/3 of the length of the plaster model, it is removed from the oven, turned over, and rapidly pulled down over the full length of the model. When contact and seal are established between the plastic and the edge of the support table, the pump is turned on with a foot-actuated valve and air is evacuated between the plaster model and soft polycarbonate. The higher atmospheric pressure exterior to the plastic surface forces the polycarbonate against the model, reproducing all of its surface detail.

During subsequent cooling, the transparent plastic hardens. The plaster model is broken out, and thus lost in the process. When further modification is required, a second plaster model may be formed from the first transparent socket. Following setting of the plaster, the first polycarbonate socket is cut away so that it can be removed without damaging the new model which is modified appropriately for formation of a subsequent transparent socket.

CLINICAL APPLICATIONS

BELOW-KNEE PROSTHETICS

In the past, transparent sockets have been used principally as check sockets to eliminate guess work in the fitting of prostheses. More recently, they have been used to help control edema. In an attempt to see if stump wrapping can be avoided several patients have been fitted with temporary prostheses, as shown in Fig. 4, with a removable transparent socket. The socket can be applied to the stump at night and held in proper apposition to the limb by a waist suspension system. The outer portion of the temporary prosthesis is a laminated polyester shell that provides reinforcement of the transparent socket and distributes on socket the load that is transmitted through the polyvinylchloride pylon. To facilitate insertion and removal of the socket, a slit is cut in the polyester shell with a slip-out rubber wedge laminated under it. Figure 5 shows the socket being inserted into the shell while the rubber wedge is held in place.

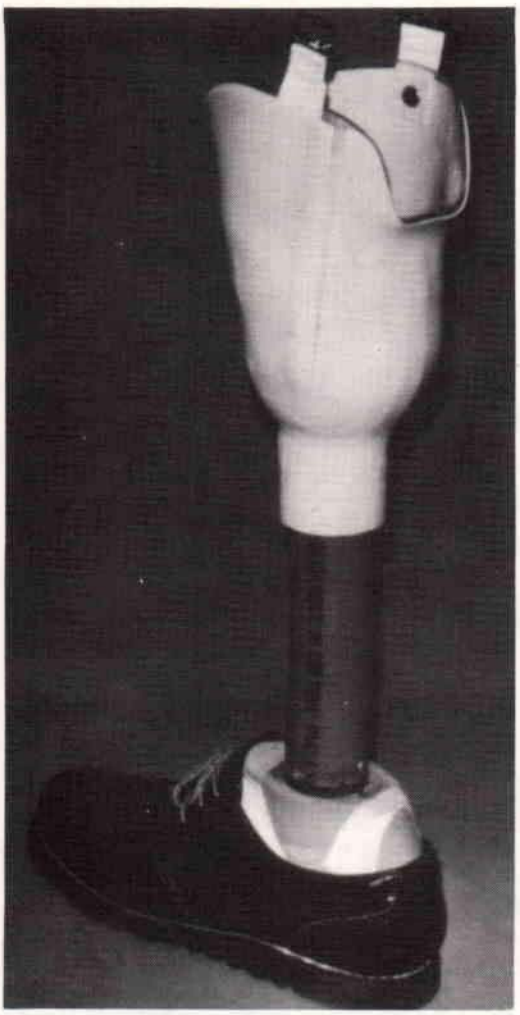

Fig. 4. Temporary BK prostheses with removable socket for edema control at night. The polycarbonate socket is enclosed in a polyester laminated outer shell for reinforcement. Note the split outer shell to facilitate removal. Not shown are the Velcro or tape used to close the split wall after reinsertion of the removable socket.

Fig. 5. Insertion of the socket into the outer shell while the rubber wedge is held in place just under the lateral slit. Suspension of the socket during the night is achieved by buckling the medial and lateral billets to a pelvic belt.

BELOW-ELBOW PROSTHETICS

Transparent sockets have been used in the fitting of externally powered, myoelectrically controlled below-elbow prostheses. Their use has facilitated proper fit and suspension over the humeral condyles and the olecranon. Concomitantly, the visibility of the socket has helped in properly locating the control electrodes over points of maximum electrical activity of wrist flexor and extensor muscle groups (Fig. 6). Consequently, even in the presence of extensive scarring and other skin alterations, critical electrode positioning can be accomplished successfully.

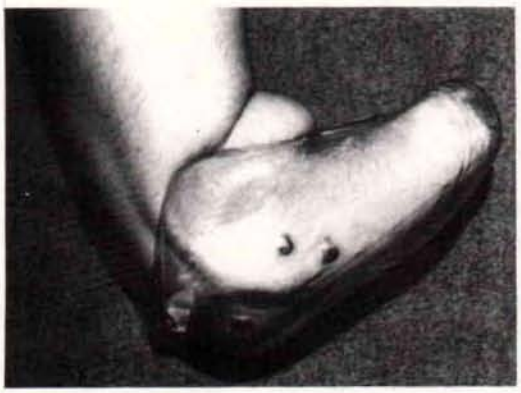

Fig. 6. Transparent below-elbow socket for myoelectric prosthesis. The points of maximum electrical activity marked on the skin are transferred to the outside of the socket after the muscle groups have conformed to and been distributed in the socket.

CONCLUSION

This clinical work, supported by the University of Virginia Orthopaedic Research Fund, has demonstrated the versatility and usefulness of vacuum-formed transparent sockets. The transparent sockets may provide visual assessment of static and dynamic socket fit, inspection of proper weight-bearing distribution across the stump surface, and more accurate determination of relief for bony prominences. Furthermore, transparent sockets are helpful in the accurate location of joint rotation centers and electrodes in the socket wall.

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ABOVE-KNEE POLYPROPYLENE PELVIC JOINT AND BAND

Erich Fischer, C.P.[1]

Suspension systems for the above-knee amputation prostheses that do not use suction are usually available in either of two extremes, the flexible webbing type of Silesian belt or the rigid steel pelvic joint and band with an attached belt. Both designs have their indications and are widely used. At the VAPC a thin flexible spring steel pelvic joint and band combination has been used for many years, but the frequency of breakage, noise, and binding at the joint has been high.

It is the purpose of this paper to present a sturdy suspension system for above-knee prostheses. It is fabricated of polypropylene and provides a degree of flexibility somewhere between the presently employed extremely rigid and hyper-flexible systems.

For more than a year now the flexible polypropylene pelvic joint and band (Fig. 1) have been used at our Center. This design provides the prosthetist and the patient with a suspension component with a flexibility between the supple Silesian belt and the rigid steel pelvic joint and band.

The polypropylene unit has been used in various combinations for patients up to this time, and has specific advantages over the spring steel. The joint does not "bind", and neither breakage nor noise is a problem. Maintenance is insignificant. Weight has been reduced greatly, the polypropylene system weighing only about one-fourth that of the rigid metal pelvic joint and band.

The flexibility of the band allows it to spread and thereby provides greater comfort during sitting. When the stump is very short and the patient finds the stability inadequate, a combination of steel pelvic joint and polypropylene band

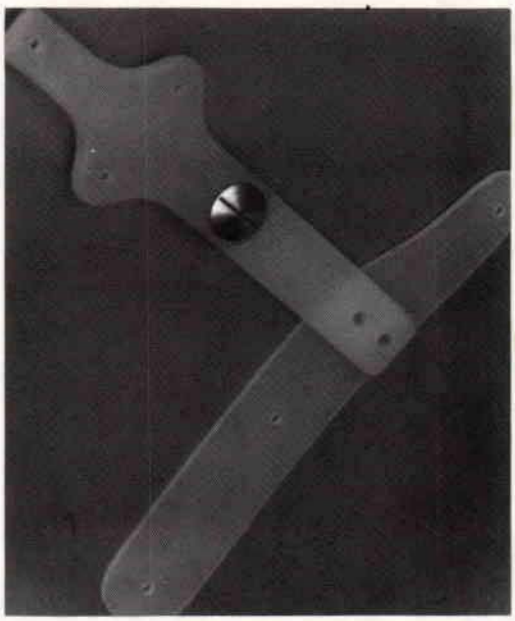

Fig. 1. The polypropylene pelvic joint and band, assembled.

may be employed, allowing for greater comfort on sitting, and yet providing stability when standing and walking.

At present we are fabricating the pelvic joint (Fig. 2) from ¼-in. thick polypropylene sheet and using a stainless steel shoulder bolt and nut for the joint. Nyliner bearings ensure long wear and smooth operation. They can be replaced easily when necessary. The band is fabricated from ⅛-in. thick polypropylene sheet. The joint and band can be shaped readily with a heat gun and attached to the socket with rivets (Fig. 3).

Other similar thermoplastic materials can be used in fabricating the pelvic joint and band. The author has also used high density polyethylene for the band which can be shaped without the use of heat.

U.S. Manufacturing Company, 623 South Central Avenue, Glendale, California 91209 has

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

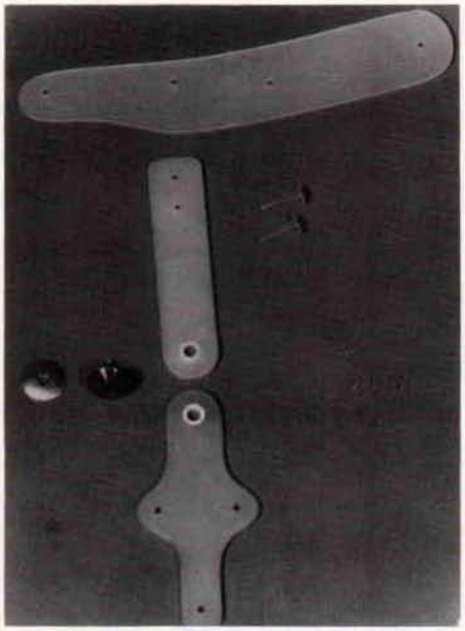

Fig. 2. Components of the polypropylene pelvic joint and band.

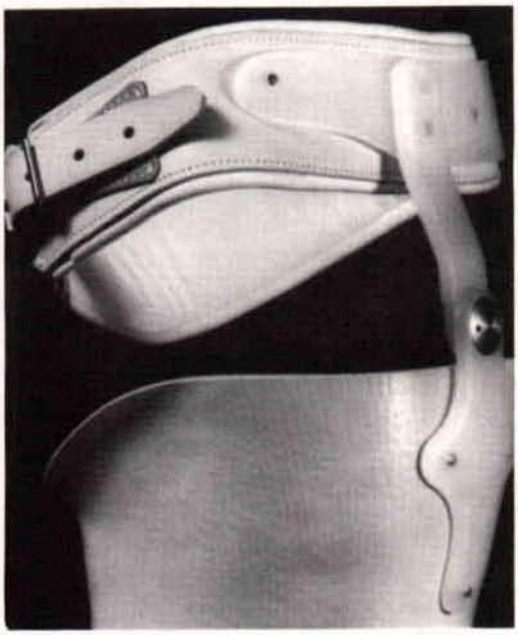

Fig. 3. The polypropylene pelvic joint and band in place on an above-knee amputee.

indicated to the VAPC that they wish to manufacture the joint and band.

It is the author's opinion, and that of his colleagues at the VAPC, that the polypropylene pelvic joint and band are a useful addition to the armamentarium of the prosthetist.

ACKNOWLEDGMENT

The author wishes to extend his appreciation to Gustav Rubin, M.D., FACS, for his valuable assistance in carrying out this project.

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THE EFFECTS OF HEEL HEIGHT AND ANKLE-FOOT-ORTHOSIS CONFIGURATION ON WEIGHT LINE LOCATION: A DEMONSTRATION OF PRINCIPLES[1]

Thomas M. Cook, M.S.[2]
Barbara Cozzens, B.S.[2]

The purpose of most lower-limb orthoses and prostheses is to impose or control sets of forces on the body to provide stability, mobility, and deformity prevention (*3*). A review of current literature in the field discloses numerous comments about the lower-limb "weight line" along with proposed guidelines regarding its proper location for optimal alignment. Although there are many "rules of thumb" for the ideal weight line location, the origins of many of these rules are unclear (*1*) and their applicability to various pathological conditions has yet to be subjected to scientific scrutiny.

From a theoretical point of view, to maintain a stable upright standing posture, the torques, or moments, about the supporting joints must be in equilibrium, i.e., the forces in the muscles and ligaments multiplied by the perpendicular distances to each joint center must be equal and opposite to the supported weight multiplied by the perpendicular distance between the weight line and the joint center. The counteracting muscle force requirements can be altered by increasing or decreasing the magnitude of the weight borne on that limb (at the cost of affecting the other limb or an assistive device) or by changing the distance from the weight line to the joint axis, or by a combination of both. The second approach, namely, controlling the weight line location, is the goal in most applications of ankle-foot orthoses (AFO's).

As recently stated by Stills (*4*), the two factors that control the weight line location in AFO applications are: 1) the position of the foot part of the orthosis with respect to the shank part of the orthosis; and 2) heel height. The purpose of this paper is to use actual weight line measurements to examine the interrelationship of these two factors.

MATERIALS AND TECHNIQUE

The measurement technique used an optical beam splitter[3] to superimpose the weight line on the lens of a camera used to photograph the subject (Fig. 1). The subject, in this case, a normal 25-year-old female, stood on a strain gage type of force plate that provided signals necessary to display on a large television monitor a vertical line that represented the weight line. The horizontal location of the line represented the vertical projection of the resultant force, that is, the weight line, on the platform with an accuracy of less than ± 0.5 centimeters. Appropriate signal conditioners and an analogue circuit calculated the force line location. (This technique can be used to assess medial-lateral alignment in either leg as well as the total body "weight line.")

The subject wore three different molded polypropylene ankle-foot orthoses (Fig. 2) (*3*) with each of three shoes with a different heel-height. The foot-shank angles of the orthoses were 75 deg., 90 deg., and 105 deg. and the sole-to-heel height differences were 0.9, 2.5, and 5.7 centimeters.

[1]This work was supported in part by Grant Number 23P-55518 from the Rehabilitation Services Administration, Department of Health, Education and Welfare.

[2]Rehabilitation Engineering Center, Moss Rehabilitation Hospital, Temple University, 12th Street & Tabor Road, Philadelphia, Penn. 19141

[3]Edmund Scientific Co., Barrington, N.J.

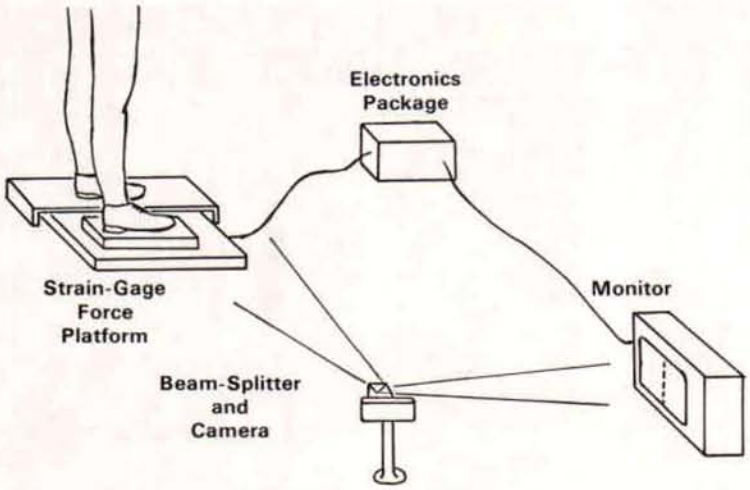

Fig. 1. Schematic of the equipment used to superimpose weight-line on photograph of a subject at rest.

Fig. 2. Molded polypropylene ankle-foot orthosis used in the demonstration. Left, AFO with ankle portion in dorsiflexion; Center, AFO with ankle portion in neutral position; Right, AFO with ankle portion in plantar flexion.

FINDINGS

Figure 3 shows the position of the weight line when the subject stands without shoes in a relaxed condition on the force platform. This line represents only the location of the weight borne on the right leg. The left leg is supported by an over-platform which is not in contact with the force plate.

Figure 4 shows the force line locations with the various heel-orthosis combinations. Column

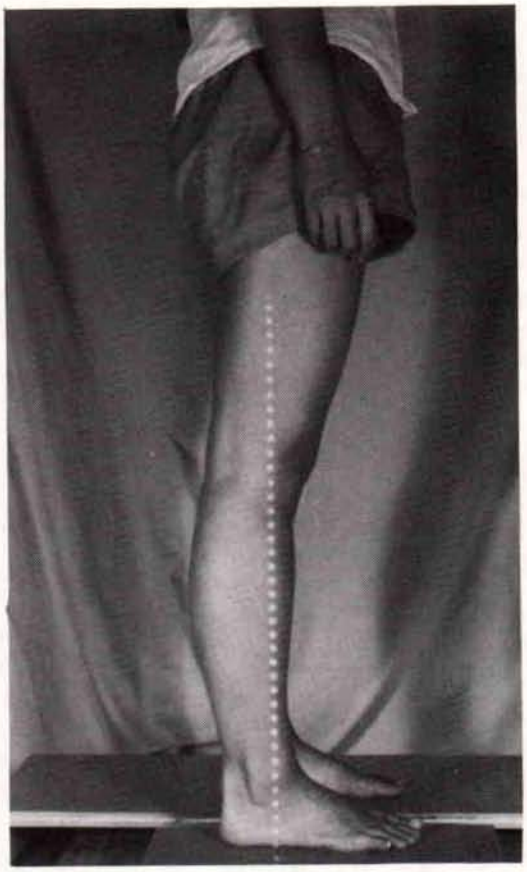

Fig. 3. Weight-line superimposed on photograph of subject standing without shoes.

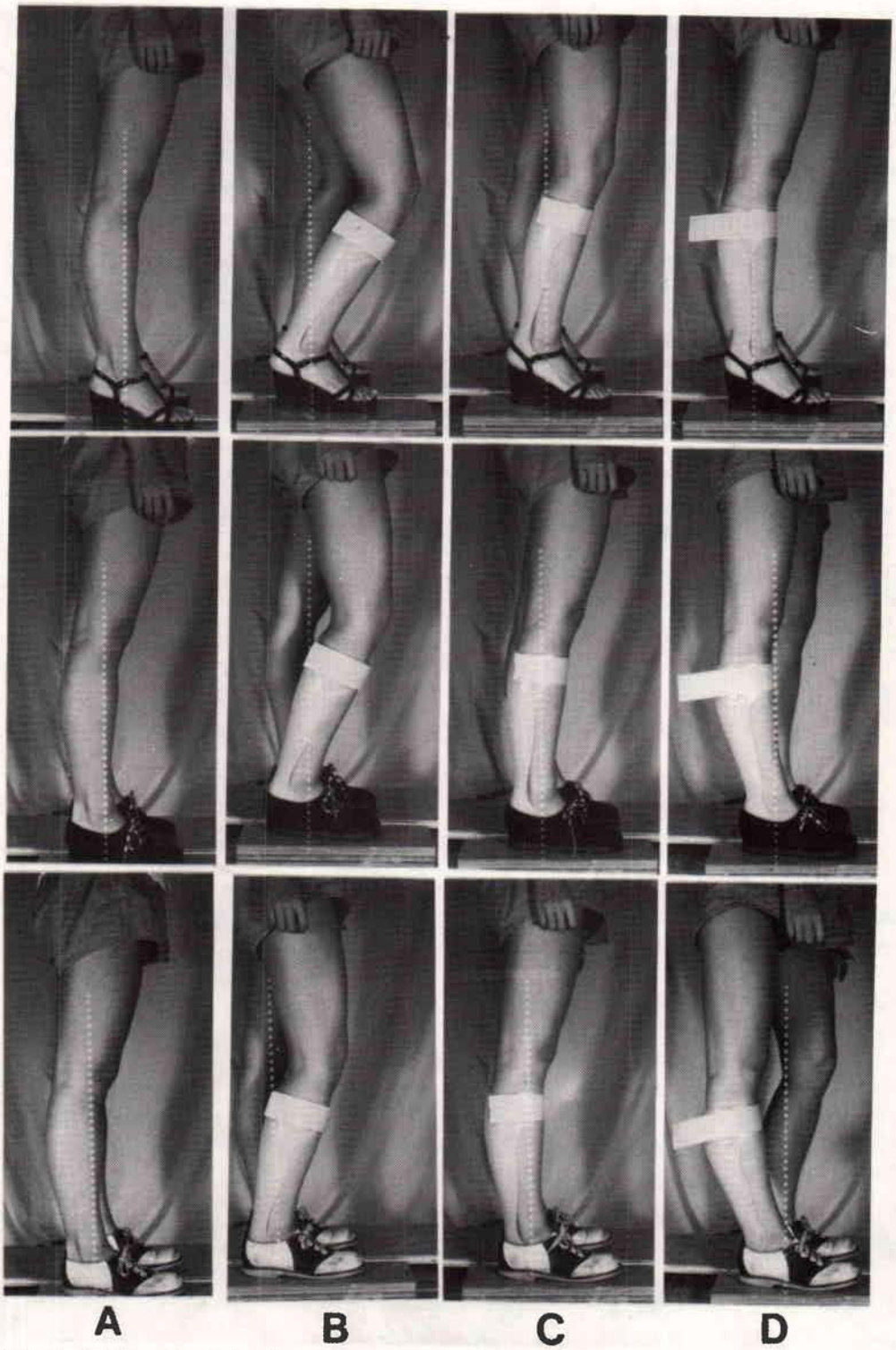

Fig. 4. Matrix of photographs showing effects of heel height and orthosis configuration on the patient in a standing position.

A shows the weight line with various heel heights prior to any orthotic application, and it can be seen that the weight line relationships at the ankle, knee, and hip appear unaffected by heel height. This is in agreement with previous findings by Hellebrandt (*4*).

In column B the subject is wearing the AFO with 15 deg. of dorsiflexion. In all instances the weight line is anterior to the ankle and posterior to the knee producing a knee flexion moment. The highest heel height causes the subject to stand with the left as well as right knee flexed.

In column C, the AFO having a 90 deg. shank-foot ankle relationship and the lowest heel height results in an alignment apparently identical to the non-braced condition. Higher heels again shift the force line posterior to the knee.

In column D, the AFO with 15 deg. of plantar flexion has been applied and the highest heel height seems to result in optimal alignment while the lower heels shift the force line anterior to the knee causing an extension moment. It should be noted that the location of the weight line under the foot seems to be affected very little except in the extreme cases of heel-orthosis combination.

DISCUSSION

Although the findings on only one subject are presented here, several other normal subjects have been examined also with substantially identical results. The principal implication from these pictures is that heel height and AFO configuration must be closely matched if optimal biomechanical alignment is to be effected. Both orthotist and patient must be highly cognizant of the fact that a particular orthosis requires an exacting match-up of heel height. The increasing use of thermoformed orthoses which fit inside standard street shoes will increase the likelihood that patients may use shoes with heel heights different from that for which the orthosis was specifically designed. This tendency is probably greatest among younger patients who are more style conscious.

The technique presented here seems to offer considerable potential for examining a wide range of orthotic and prosthetic devices, at least from the viewpoint of static alignment, and a number of such investigations are currently being undertaken. How static alignment relates to the dynamic activity of locomotion is a more crucial and complex question for the biomedical scientist (*3*).

The practitioner will ask how such a procedure, with its elaborate electronic and photographic features, could be implemented in the clinical setting. Work is currently being done at the Krusen Research Center, Moss Rehabilitation Hospital to develop a simpler device for clinical use. It is our aim to provide a practical device that will enable the clinician to apply more precisely general principles, such as those demonstrated above, to individual patients in order to provide maximal stability, mobility, and deformity prevention.

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A DYNAMIC PLASTIC ELBOW-EXTENSION ORTHOSIS FOR REDUCTION OF FLEXION CONTRACTURES

Herbert Goller, Ph.D., C.P.O.[1]
March Enders, M.D.[1]

The function of the elbow is to be subservient to the hand in the sense that the elbow enables the hand and fingers to be placed at desired distances from the body. Two types of motion are required to position the hand. The first, and principal one, is the shortening and lengthening of the distance the hand is positioned from the body. The hand may be brought out at full arm's length, as when the elbow is in full extension, or it may be brought up against the shoulder, as it is when the elbow is fully flexed. The second motion of the elbow adjusts the position of the hand in relation to the transverse plane by either pronation or supination.

The ulnar-humeral joint, a pure hinge-type joint, allows extension and flexion motions of the elbow. Extension of the elbow is supplied by the three heads of the triceps, and some slight effort is attributed to the anconeus. The brachialis and the biceps are the two principle flexors, with the brachioradialis considered an auxiliary flexor.

Spinal cord trauma, as well as a variety of neuromuscular diseases, may cause paralysis of the elbow extensors. Although gravity does assist in elbow extension, the patient that has normal or near-normal elbow flexor strength frequently suffers elbow flexion contractures owing to a lack of active elbow extension. This happens usually in quadriplegics with a lesion at the C5-6 neurological level, in which the elbow flexion is in the range of good to normal strength and the elbow extensors are absent. A well-planned program of daily exercise is required to prevent the formation of progressive contractural deformities that frequently occur when the patient no longer has physical therapy after discharge from the hospital.

Various techniques and methods are used for the post-contracture treatment. Several surgical techniques which transpose muscles to improve the extensor power of the elbow may be performed. Serial plaster splinting of the elbow with forced extension while casting is a simple orthotic approach. Various orthotic appliances made of metal, leather and fabric are described in the literature. These appliances have extension moment supplied by metal springs, elastic fabric, or a machine screw. A slow steady pressure is desired so that the fibrous tissue atrophies, stretches, and, in effect, grows longer. Those designs that cause an intermittent pressure such as from a sudden passive manipulation may cause the fibrous tissue to hypertrophy if the extension is applied inexpertly.

Current research at certain Rehabilitation Engineering Centers gives consideration to the elbow flexion contracture. The Annual Progress Reports of Rancho Los Amigos briefly describe research on powered orthoses that are used to extend the contracted elbow periodically. Electrical stimulation of the patient's extensor muscles to overcome the contracture is also receiving their attention.

DESIGN OF THE ORTHOSIS

A review of the existing orthotic designs for elbow extension indicated that a new improved design was needed. It was felt that the use of modern plastics, with their inherent advantages such as lightweight, ease of fabrication, and appearance, should be tried.

[1]George Washington University, Washington, D.C.

A biomechanical analysis of the problem indicates that the simplest force system that will produce the required extension moment about the elbow joint is that shown in Figure 1. This resultant force system is that of two couples, one proximal and the other distal to the elbow joint.

By positioning a single thermoplastic band into the shape shown in Figure 2, the force system described in the previous paragraph may be implemented. The band is formed into an elbow extension position that exceeds the contracted angular position of the anatomical elbow. When positioned on the elbow with contracted elbow flexors, the result is an extension moment force system.

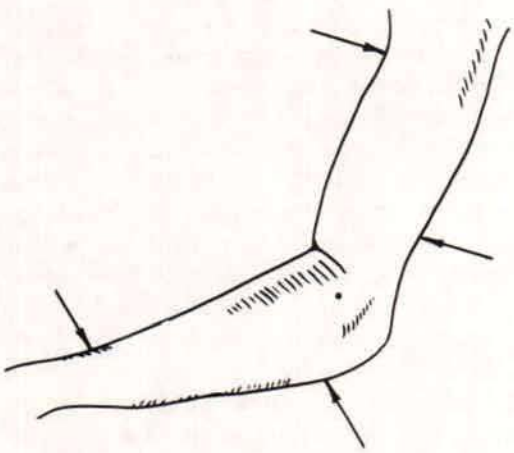

Fig. 1. The simplest force system that will provide an extension moment about the elbow. Note that two couples are provided: one distal to the joint; the other, proximal.

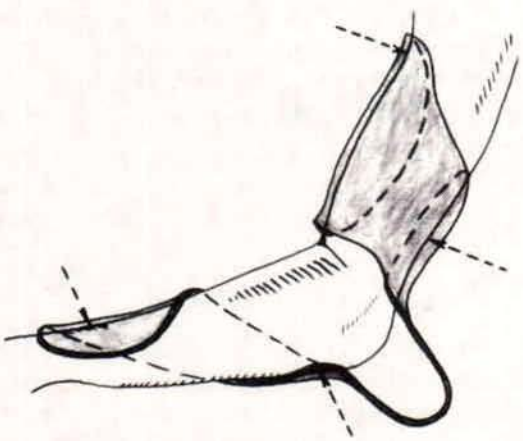

Fig. 2. A single plastic band shaped to provide the force system shown in Figure 1 and thus produce an extension moment about the elbow.

The plastic used in the construction of the orthosis is Plexidur,[2] a thermoplastic that offers the desired mechanical and physical properties. Its amber color offers cosmetic appeal.

The bow shape of the orthosis over the olecranon process of the elbow serves three purposes. First, it insures that no contact is made over the pressure sensitive region of the olecranon process and the lateral and medial epicondyles. Second, it serves to reduce the stress concentration that would be present if a sharp change in direction of the plastic were formed over the relatively prominent olecranon process. Third, the bow shape is used to adjust the pre-extended position built into the orthosis. Localized heating of the bow with a heat gun allows angular adjustment to be made easily. This permits the orthotist to adjust the extension moment as the flexion contracture is gradually reduced over a period of time.

Assistance is needed in donning the orthosis, which must be forced into a flexed position but put on the patient's arm with a twisting motion. Donning and removal of the orthosis is a relatively simple procedure after one or two trials. The unwinding action of the spiral and also the fit of the shell over the anterior aspects of the arm and forearm keep the orthosis stable and properly positioned.

FABRICATION

The initial step in fabrication is to obtain a plaster cast of the patient's arm with the elbow extended to the limit of the contracture. The arm is positioned with the forearm supinated, and prominences and landmarks are marked with an indelible pencil. The cast is removed carefully and filled with plaster of Paris to provide a positive model.

[2]Rohm & Haas Company, Darmstadt, Germany

The positive model is smoothed and undercuts and buildups added for flares at the location of trim lines. As shown in Figure 3, a cylindrically shaped piece of plaster of about five centimeters diameter is added over the olecranon process for formation of the bow shape. A paper pattern is cut and fitted on the cast, and then used to lay out the thermoplastic from 4 mm.-thick sheet stock, as shown in Figure 4. The edges of the thermoplastic are sanded and polished.

The thermoplastic is heated in an oven to a temperature of 140 deg. Celsius for approximately ten minutes. The plastic is removed and wrapped over the positive model and held in place with an elastic bandage. After the plastic has cooled to room temperature, it may be removed from the mold, as shown in Figure 5.

The orthosis is extended prior to fitting by locally heating the 5 cm. diameter bow with a heat gun, and carefully extending the orthosis to the desired angular displacement. The amount

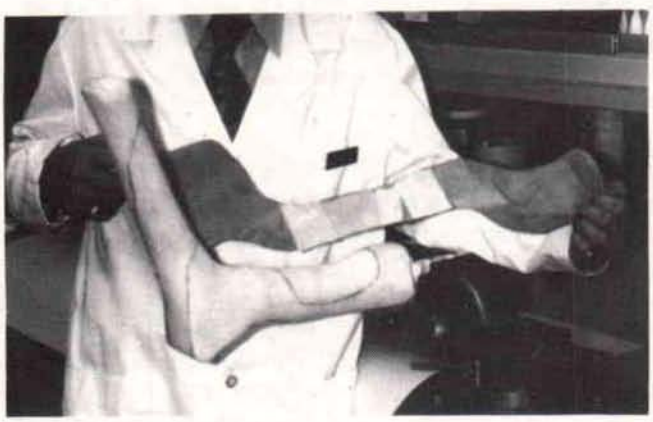

Fig. 3. The cast has been modified by building up a five-centimeter diameter piece over the olecranon to provide for formation of the connecting bow. A paper pattern for the orthosis is developed over the model.

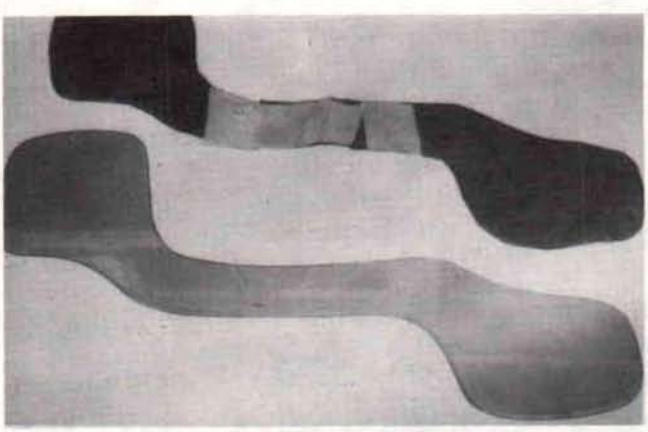

Fig. 4. The paper pattern is used as a guide for cutting the Plexidur flat stock.

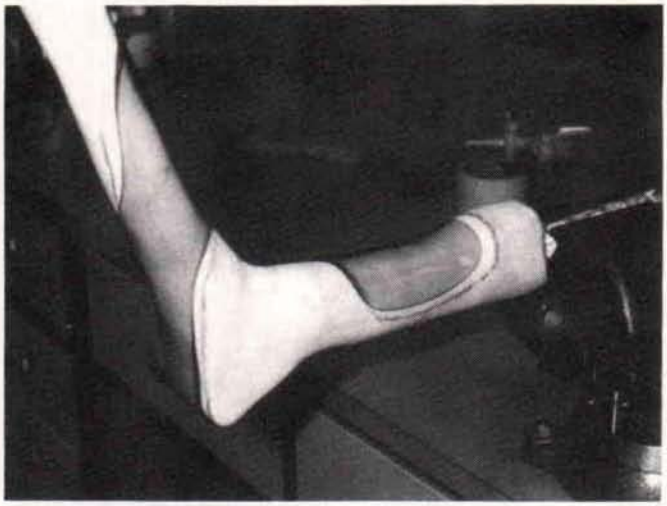

Fig. 5. The orthosis just before removal from the model.

of extension will depend on the patient and the severity of the contracture but an initial extension of approximately twenty degrees beyond the contracted position has been found to be satisfactory. The orthosis is shown on a patient's arm in Figure 6.

DISCUSSION AND RESULTS

A total of five plastic elbow extension orthoses have been fitted to patients. The overall acceptance of this device by the patients has been good.

The functional performance of the orthosis to stretch out the elbow flexion contracture is shown in Figure 7, which graphically illustrates the gonimetric record of the elbow joint extension over the first five months of treatment. The slopes of the curves indicate that a reduction of approximately two degrees of contracture per week can be obtained with the orthosis during the early months of treatment.

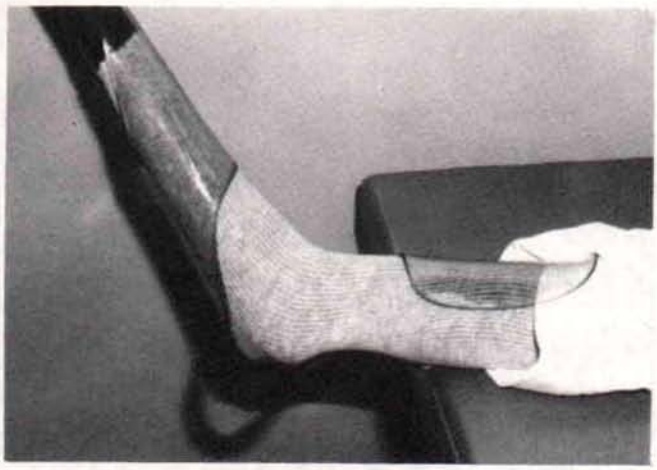

Fig. 6. The orthosis on the patient.

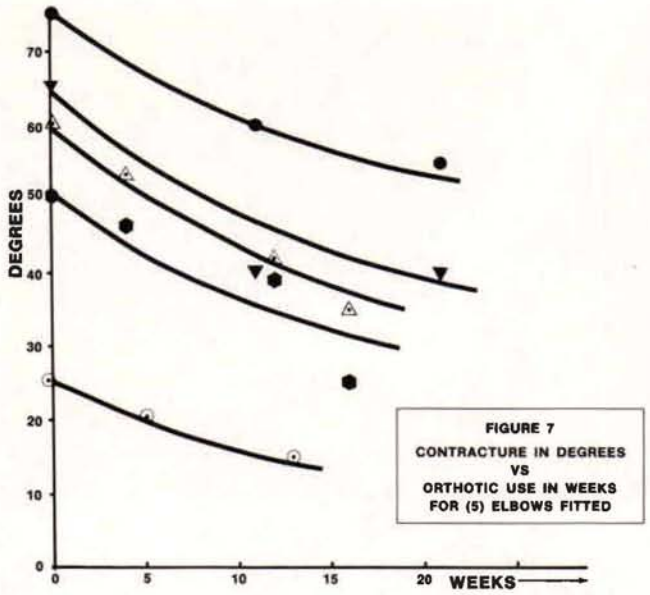

FIGURE 7
CONTRACTURE IN DEGREES
VS
ORTHOTIC USE IN WEEKS
FOR (5) ELBOWS FITTED

Some of the advantages of the new plastic orthosis described in this paper over the conventional designs are improved cosmesis, lighter weight and less bulkiness. The plastic orthosis also offers the advantage of being relatively easily removed for bathing. The appliance is also easily maintained in a clean and appealing condition.

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NEW PUBLICATIONS

EVALUATION OF THE ORTHO-WALK TYPE B PNEUMATIC ORTHOSIS ON THIRTY-SEVEN PARAPLEGIC PATIENTS. Committee on Prosthetics Research and Development, National Academy of Sciences, 2101 Constitution Avenue, Washington, D.C. 20418, USA. 71 pp.

This is a report of a clinical study coordinated by the Committee on Prosthetics Research and Development of the National Academy of Sciences on the pneumatic bilateral lower-limb orthoses known as the Ortho-Walk, Type B. Trials were made with thirty-seven paraplegic subjects in both Veteran Administration and civilian spinal-cord centers.

The summary and recommendations are:

"Based upon data collected from seven hospitals and 37 paraplegic patients with thoracic lesions, the following conclusions and recommendations were made concerning the Ortho-Walk Type B pneumatic orthosis.

ADVANTAGES

- Less weight than conventional orthoses.
- Temporary positive psychological reaction.
- Cost is reasonable when used as stock item on many patients.
- Adjustability and adaptability allows rapid application in the early treatment phase.
- Possible energy expenditure savings during ambulation.
- Adequate trunk support.

DISADVANTAGES

- Difficult and time-consuming to don.
- Inflation-deflation system is impractical.
- Provides less support to lower limbs than conventional orthoses.
- Cosmesis is poor, especially for women. Conventional orthoses are more cosmetic.
- Mobility is decreased. Stair-climbing and transfers are more difficult than with conventional orthoses.
- Maintenance and repairs must be made by the manufacturer, necessitating long waits due to mailing time.
- Cost is higher than conventional orthoses when used for one patient only.

Recommended Improvements

1. Improvement of the inflation-deflation system by miniaturizing the pressure source. An alternative solution may be the addition of knee and hip joints.
2. More ankle stability be provided, specifically to prevent dorsiflexion.
3. Ventilation of the suit.
4. The color of the suit (blue) is not cosmetic and could be improved.

Recommendations for future studies

1. A study of the pressures at the interface between the pneumatic suit should be made. Redness usually occurs over bony areas when the suit is worn and clinicians are fearful that long-term standing may lead to skin breakdown.
2. More basic bioengineering is needed if pneumatic orthoses are to be made practical and be universally accepted.
3. Evaluation of other systems of mobilizing paraplegic patients should be made, such as that undertaken by Lehman *et al.* (*6*) on the Craig-Scott design knee-ankle-foot orthosis, which is used successfully in the Denver area. A quantitative study would be useful to compare penumatic orthoses with standard knee-ankle orthoses used in paraplegic ambulation.
4. A careful study should be done to verify and document reported psychological and physiological benefits accruing from the use of various orthoses.

CONCLUDING STATEMENT

The concluding statement of the participants in the evaluation was that the Ortho-Walk Type B pneumatic orthosis has the potential for being a tool in the early rehabilitation of spinal-cord-injured patients. Its relative adaptability and ease of use in the early rehabilitative phase has the advantages of a temporary psychological lift, early screening of potential candidates for orthoses, and possibly aiding and improving the physical status of these patients. When considering candidates for the pneumatic orthosis, individualization of patients is a must, just as with conventional orthoses.

This study has shown that, as with all new ideas, the Ortho-Walk is not a panacea. Many improvements must be made before it can equal the merits of conventional metal orthoses. The idea of pneumatic splinting for paralyzed people is a feasible one, but it still needs much work. The fact that a private manufacturer can research, develop, and market such an innovative device without outside support is an accomplishment in itself. Ideas such as the Ortho-Walk can stimulate research teams in different disciplines to work jointly for the ultimate goal of better care for the spinal-cord-injured patient."

CLINICAL EVALUATION OF RANCHO LOS AMIGOS HOSPITAL/MEDTRONIC, INC. Implanted Neuromuscular Assist Device, Committee on Prosthetics Research and Development/Committee on Prosthetic-Orthotic Education, National Academy of Sciences, 2101 Constitution Avenue, Washington, D.C. 20418, 105 pp., Gratis.

This is a report on the application of an implanted electrical device for the stimulation of the peroneal nerve to provide dorsiflexion and eversion in "drop foot" conditions.

Thirty-seven cases were provided with the device in seven clinics. After appropriate follow-up studies the participants in the evaluation program made the following conclusions and recommendations:

"1. The implanted receiving component of this functional electrical stimulation system is sufficiently reliable to be appropriate for general use. Reliability problems exist in the transmitter, but this component as well has been considerably improved.

2. The clinical use of functional electrical stimulation to give neuromuscular assistance by means of an implanted device is practical and potentially offers better care to selected patients with a lesion of the motor cortex than is currently available. Its development should be continued.

3. The Rancho Los Amigos Hospital/Medtronic, Inc. neuromuscular assist device should be made available to a restricted but progressively increasing number of clinical teams who should have specific training in patient selection and management. It was suggested that this should be by a system of Regional Study Groups established by the industry."

THE KNEE JOINT — RECENT ADVANCES IN BASIS RESEARCH AND CLINICAL ASPECTS. Editors: O.S. Ingwersen, B. Van Linge, Th. J. G. Van Rens, G. E. Rösingh, B. E. E. M. J. Veraat, and D. Le Vay. ISBN 90-219-0254-0, International Congress Series No. 324: Excerpta Medica Amsterdam, Excerpta Medica American Elsevier, New York 1974, 329 pages.

This beautifully printed book represents the proceedings of an International Congress on the Human Knee Joint held in Rotterdam, the Netherlands, September 13–15, 1973. It consists of 59 separate papers, an author index, and a subject index. The papers are grouped under six headings:

I. Biomechanics of the Knee Joint
II. Degenerative Joint Disease
III. Trauma of the Knee Joint

IV. Congenital and Postural Deformities
V. Transplantation of Cartilage and Bone
VI. Arthroplasty of the Knee Joint

"The Knee Joint" should serve as a reference book for many years for serious students of the human knee. Most clinicians will also find a good portion of the proceedings valuable and well worth their study.

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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	To	Multiply by
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.

Fracture bracing

O.H.C.

*The knee disarticulation prosthesis was designed and developed at the Orthopaedic Hospital in Copenhagen, Denmark.

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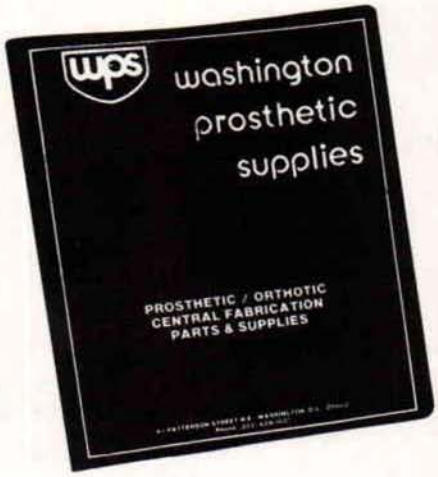

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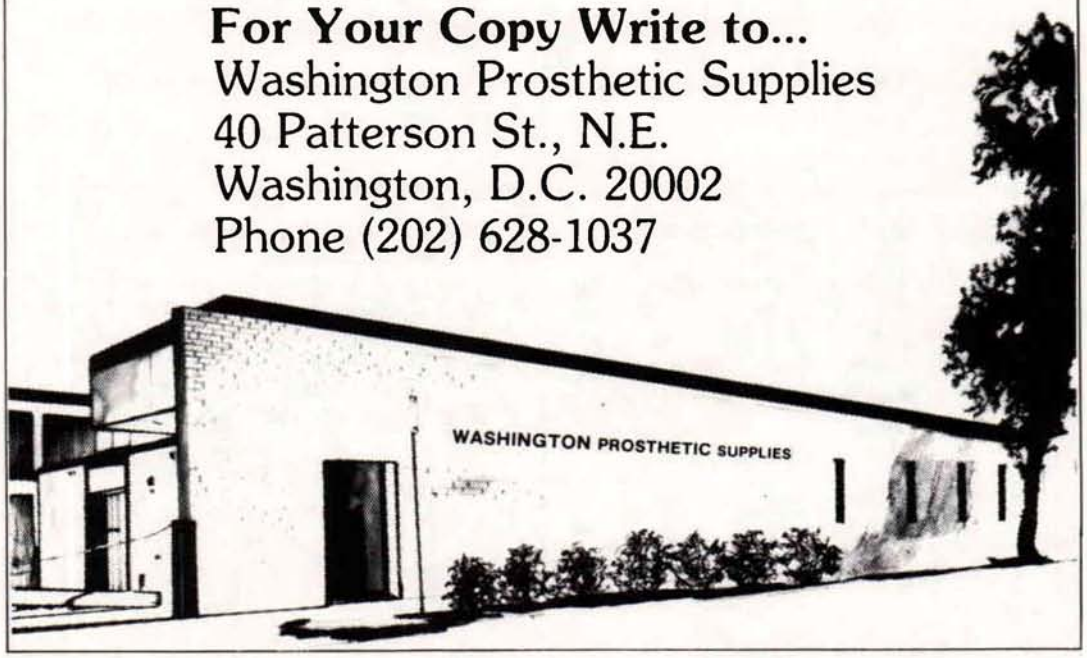

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