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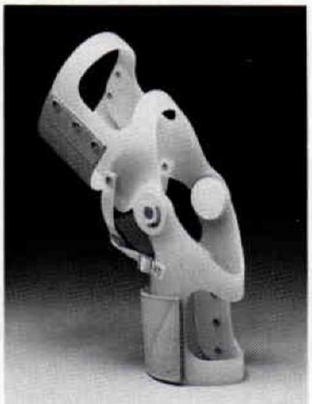

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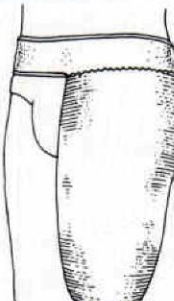

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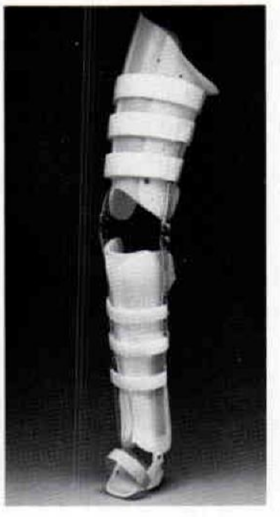

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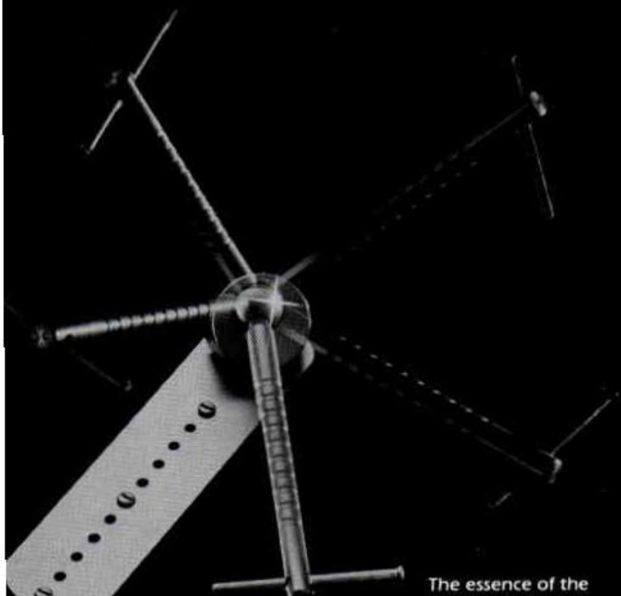

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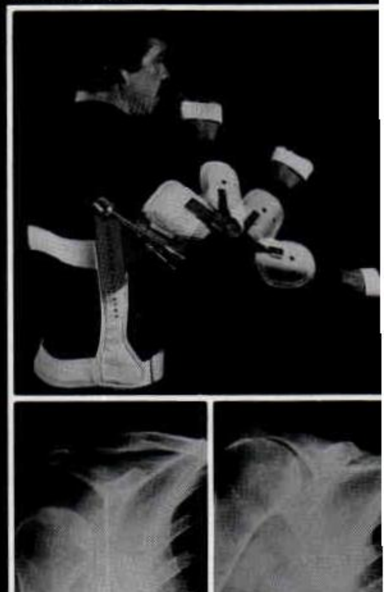

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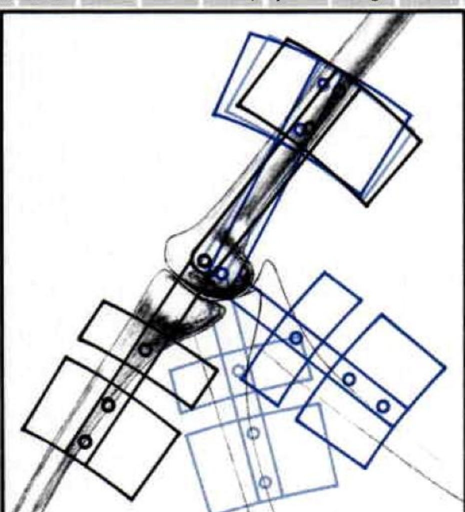

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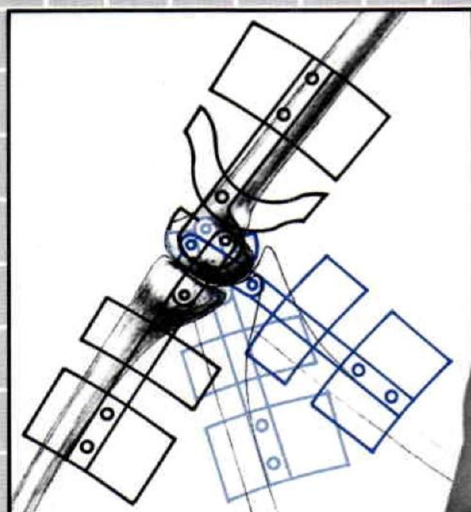

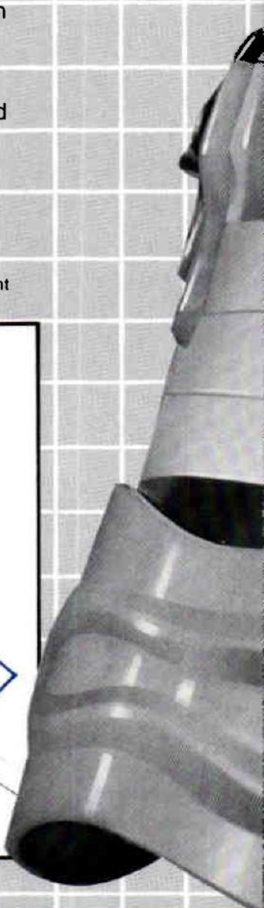

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1984

October 15–21, AOPA General Assembly and International Congress, Fontainebleau Hilton Hotel, Miami Beach, Florida. Contact: AOPA National Headquarters, 703-836-7116.

October 21–23, "Certification for Rehabilitation Nurses: Applying Advanced Concepts to Practice," a seminar offered by the Rehabilitation Nursing Institute, Cincinnati, Ohio. Contact: RNI Seminar Department, 2506 Gross Point Road, Evanston, Illinois 60201; tel. 312-475-7300.

October 23–25, International Conference on Rural Rehabilitation Technologies, University of North Dakota. Contact: Don V. Mathsen, Box 8103, University Station, Grand Forks, North Dakota, 701-777-3120, or Charles M. Page, Box 8202, University Station, Grand Forks, North Dakota 58202, 701-780-2494.

October 23–27, IFAS '84, the 18th International Trade Fair for Hospital and Medical Supplies, Zurich, Switzerland. Contact: Joachim Schafer, Executive Director, TEAM, P.O. Box 3092, 265 Varsity Avenue, Princeton, New Jersey 08540. Telephone: 609-452-2895.

October 24–28, 38th Annual Meeting, Academy for Cerebral Palsy and Developmental Medicine, Washington, D.C. Sheraton Hotel. Contact: American Academy for Cerebral Palsy and Dev. Med., Ste. 1, 2315 Westwood Avenue, P.O. Box 11083, Richmond, Virginia 23230, 804-355-0147.

November 3, Midwest Chapter of the Academy Fall Seminar, Northwestern University, Chicago, Illinois.

December 2, Northwest Chapter of the Academy Meeting and Seminar, Red Lion Inn, Portland, Oregon. Contact: William E. Teter, CO, 208-342-4659.

1985

January 24–29, American Academy of Orthopedic Surgeons Annual Meeting, Las Vegas, Nevada.

January 30–February 3, Academy Annual Meeting and Scientific Seminar, Cathedral Hill Hotel, San Francisco, California. Contact: Academy National Headquarters, 703-836-7118.

February 9, Midwest Chapter of the Academy Prosthetics Workshop, Northwestern University, Chicago, Illinois.

April 11–13, Association of Children's Prosthetic and Orthotic Clinics (ACPOC) Annual Meeting and Scientific Sessions, New Orleans. Contact: Dr. Robert Tooms, tel. 901-525-2531.

April 12–13, New York State Chapter of the Academy seminar, The Hotels at Syracuse Square, Syracuse, New York.

April 18–20, AOPA Region IV Annual Meeting, Wilmington Hilton Hotel, Wilmington, North Carolina.

April 20, Midwest Chapter of the Academy Spring Seminar/Social Event.

May 2–4, AOPA Region V Annual Meeting, Holiday Inn, Cleveland, Ohio.

May 8–11, AOPA Regions VII, VIII, X, and XI Combined Annual Meeting, Tucson, Arizona.

May 16–19, AOPA Regions I, II, and III Combined Annual Meeting, Hyatt Regency Inner Harbor, Baltimore, Maryland.

June 4–8, Orthopädie & Rehn-Technik 85 International Trade Fair and Congress, Messe Essen, W. Germany.

June 7–9, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting, Reno, Nevada.

June 20–23, AOPA Region VI, Midwest Chapter of the Academy Combined Annual Meeting, Arlington Park Hilton, Arlington Park, Illinois.

June 24–28, RESNA 8th Annual Conference on Rehabilitation Technology, "Technology—A Bridge to Independence," Peabody Hotel, Memphis, Tennessee. Contact: RESNA, Suite 402, 4405 East-West Highway, Bethesda, MD 20814, 301-657-4142.

September 13–15, Fifth Annual *Advanced* Course in Lower Extremity Amputation and Prosthetics, Nassau County Medical Center, East Meadow, New York. Contact: Lawrence W. Friedmann, M.D., Chairman, Dept. of Physical Medicine and Rehabilitation, Nassau County Medical Center, 2201 Hempstead Turnpike, East Meadow, NY 11554; (516) 542-0123.

October 15–20, AOPA Annual National Assembly, Town and Country Hotel, San Diego, California. Contact: AOPA National Headquarters, 703-836-7116.

1986

January 27–February 2, Academy Annual Meeting and Scientific Seminar, MGM Grand, Las Vegas, Nevada. Contact: Academy National Headquarters: 703-836-7118.

February 20–25, American Academy of Orthopedic Surgeons Annual Meeting, New Orleans, Louisiana.

April 8–11, Pacific Rim Conference, Intercontinental Hotel, Maui, Hawaii.

April 17–20, AOPA Region IV Annual Meeting, Orlando, Florida.

June 6–8, AOPA Region IX, COPA, and the California Chapters of the Academy Combined Annual Meeting.

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January 22–27, American Academy of Orthopaedic Surgeons Annual Meeting, San Francisco, California.

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Orthotic Control of Ground Reaction Forces During Running (A Preliminary Report)

by John Glancy, CO

This discussion of the control of ground reaction forces is presented on the assumption that the best approach to improving the runner's environment is to prevent injuries by the prophylactic use of orthotic devices. It is now readily acknowledged that ground reaction forces can adversely affect the foot/ankle complex when one or more of a variety of abnormal conditions are present. Also, there is equal acceptance that ground reaction forces when malfunction of a part of the foot/ankle complex is present, can have debilitating effects higher up in the kinetic chain. This discussion will be limited to the effects of ground reaction forces to the foot and ankle during the heel-to-toe running cycle.

The prevention of running injuries by orthotic means may seem academic with respect to runners attending sports medicine clinics who come for immediate treatment of injuries they have already sustained. Under these circumstances, the use of orthotic devices may still be thought of as prophylactic if the cause of a particular injury is determined to be biomechanical in nature and if the device successfully arrests or reverses the causative condition and/or prevents reoccurrence of the injury.

There are estimated to be over 25 million adult Americans now running or jogging regularly.[2 (Part 1)] If the patients referred to the University of Indiana Division of Orthotics are representative of runners in general, many of their injuries are related to foot and/or ankle conditions that were present prior to their running activities. These conditions are often reported to have been asymptomatic before the patient took up running. In fact, the majority report that they do not experience symptoms except when they run or soon after a run. The increase in impact with the ground when running, by a factor of two to three times body weight at heel strike, is known to be related to the development of symptoms. Burdett's study predicts peak Achilles tendon forces ranging from 5.3 to 10 times body weight.[3] Also related are pre-existing biomechanical "imperfection(s)" of which the runner has been unaware and/or a long-forgotten prior injury has led to malalignment of the foot or ankle and which predisposes the runner to injury.[2 (Part II, p. 26)]

Sports medicine patients are unique in terms of individual motivation. They dislike hearing that they must give up or slow down their weekly running schedule, even for a few days. Often, many attempt to "run through" an injury, causing the condition to become chronic or resulting in a more severe level of injury. This mind-over-matter attitude, when pathomechanical factors are involved, coupled with attrition, can lead to a no-win situation, both immediate and long-term.

Since the general fitness of most runners is well above average, their very fitness often masks the cause of their complaints which, as a consequence, can be very subtle and difficult to define. Dealing with these subtleties sharpens one's powers of clinical observation. The insights and developments reported in this paper are to a substantial extent the product of the patience, persistence, conscientiousness, and accurate feedback of the runners themselves. I am indebted to them because the benefits from these new insights and developments are already being shared by orthotically handicapped patients. For example, the promising results to date with the flexible polypropylene insert and cushion heel wedge (Figures 6 and 7) for juvenile rheumatoid arthritis patients are especially gratifying.

THE RUNNING CYCLE VERSUS THE WALKING CYCLE

The purpose of comparing the walking and running cycles is twofold: first, to identify differences in kind from differences in degree between the two cycles; second, once identified, to try to understand the biomechanical relevance of either type of difference as regards injuries to the lower limb while running, particularly to the foot/ankle complex.

The stance phase is reduced from approximately two-thirds of the walking cycle to one-third in the running cycle. As one might anticipate, whether during walking or running, the velocity, cadence and stride length increase as the speed of gait increases. Also, as the gait speed increases, the period of stance phase decreases, and the period of swing phase increases in the walking cycle. As a result, the greater the increase in gait speed, the less time the foot spends on the ground.[20, 21] An increase in the speed of gait, then, in spite of its effect upon velocity, cadence and stride, is not the feature that differentiates the running cycle from the walking cycle.

It is the absence of a double-support phase that distinguishes running from walking. There is a period during the running cycle when both feet are off the ground. This period has been named the non-support or "float" phase.[20] What motion, exclusive to running, eliminates the double-support phase? None, per se. An increase in the magnitude of the thrust at push-off by the contralateral limb, generated principally by the gastrocnemius and soleus muscles, "lifts" the body and makes a float phase possible. The forward velocity attained is also an important contributor to the float phase.

From an orthotic standpoint, the replacement of the double-support phase of walking with a float phase when running is the most significant biomechanical feature of the running cycle. It is this biomechanical feature, above all, that makes the control of ground reaction forces important for the following reasons:

- The orientation of the foot/ankle complex to the ground at lift-off determines the balance and direction of the body as it ascends, particularly with respect to the line of progression.
- Without the double-support phase, a person's ability to shift his body's weight efficiently and economically during stance phase is greatly reduced. Adaptation to surface conditions, topography, and fatigue, for example, must be made rapidly upon one foot at a time. This restriction affects both lift-off and heel-strike.
- The orientation of the foot/ankle complex to the ground, when receiving the body as it descends from float phase to stance phase, has a direct bearing on running injuries.
- During its descent, the body is in "free fall." The resulting impact to the lower limb is reported to be 2.5 to 3.0 times the body's weight.[4, 20, 21] Obviously, there is no way of altering the

body's rate of descent for a given running speed. However, the floor reaction forces are always equal and opposite to the resulting vertical, AP shear, ML shear, and torque forces generated at any particular speed. Control of one or more of these floor reaction forces is a viable means of controlling the alignment and/or phasic motions of the foot/ankle complex throughout the stance phase, thereby preventing injury.

Several studies report that the vertical force passing through the foot reaches its highest peak just before lift-off.[4, 20, 21] As a consequence, from an orthotic point of view, the mid-foot is especially vulnerable between heel-rise and lift-off. It is during this period of the running cycle that protection from injury is most difficult to provide without interfering with the dynamics of running.

With biomechanics foremost in mind, then, the only difference in kind between walking and running is the transition from a double-support phase to a float phase. All the motions that are inclusive to normal stance and swing phases of the walking cycle are the same that occur during the stance and swing phases of the heel-to-toe running cycle.

However, there are differences in degree of these motions as they occur within the running cycle. The principal difference is that the range of these motions increase during running, reflecting additional differences in degree in gait, velocity, cadence, and stride length. There is also a difference in degree with respect to the impact and stress to which the lower limbs are subjected between walking and running.

Because the same number and kinds of motion occur in the same sequence in both cycles, the absence of a double-support phase forces the runner to perform the phasic motions of the stance phase without assistive substitution from the contralateral lower limb. Any attempts to substitute one motion for another are not feasible, since each motion—in proper sequence—is essential to a running gait free of injury. As a consequence, when one or more of three conditions are present within the foot/ankle complex, the probability of injury becomes a distinct biomechanical possibility. The three conditions are:

- *Hypermobility,* i.e., excessive range of one or more phasic motions,
- *Hypomobility,* i.e., less than normal range of one or more phasic motions, and
- *The loss* of one or more phasic motions.

Whether resultant injury from these conditions is immediate or long term will be discussed later, along with suggested orthotic management for each.

Numerous questions arise as a result of comparing the walking and running cycles. For example, 70 percent of the runners seen in our Sports Medicine Clinic cannot evert their heels, either passively or actively. There is nothing in the literature to indicate that the absence of this motion is as prevalent in "normal" feet among the walking population. Is it not reasonable, then, to presume that the loss of heel eversion is acquired and that this loss is a product of long term distance running?

Is "cavus" necessarily the best term to use to describe the high longitudinal arch of a runner? Dorland's *Medical Dictionary* defines "pes cavus" as "exaggerated height of the longitudinal arch of the foot, present from birth or appearing later because of contractures or disturbed balance of the muscles."

In an otherwise normal foot, is the presence of a congenital high longitudinal arch (Dorland's "exaggerated height") considered an abnormality? When the subjects are runners, the literature does not associate abnormality with this condition. In fact, substantial evidence has been reported, regarding the demands that running places upon the foot/ankle complex, that has led to a consensus that a high longitudinal arch is favorable to runners. This consensus is based on evidence related to studies of the subtalar joint.[11, 14, 15, 16, 18, 22]

In essence, the more vertical the angle of inclination of the A-P axis of the subtalar joint (*i.e.,* the higher the longitudinal arch), the less pronation or supination of the foot. The correlation between excessive

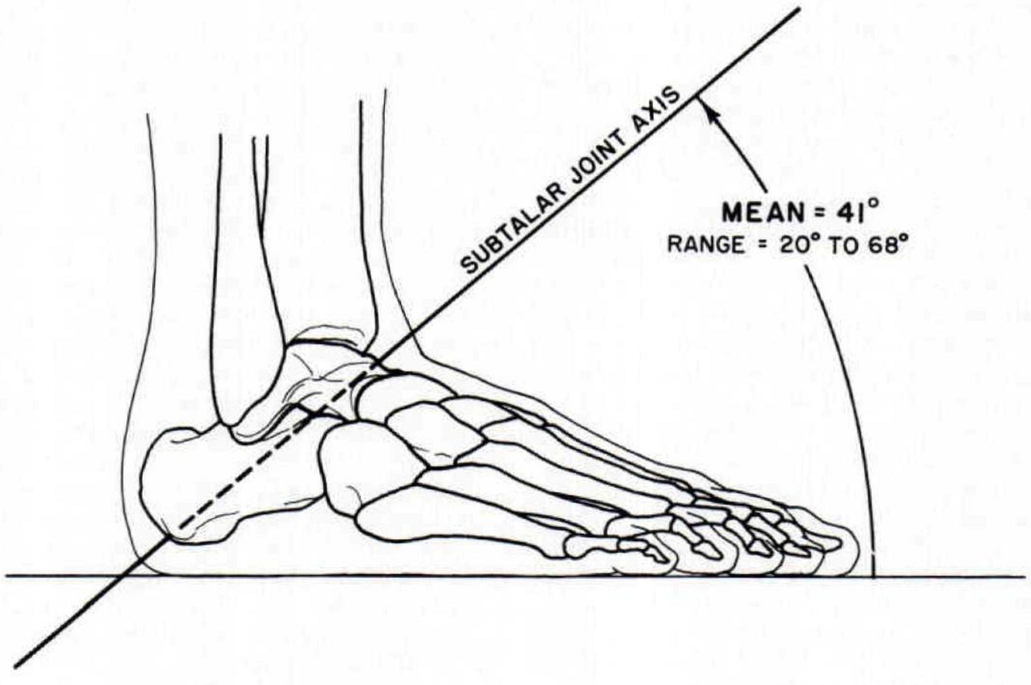

Figure 1. Angle between the axis of the subtalar joint and the horizontal (From Inman [14] in: *Bull. Prosth. Res.*, BPR-10-11, 1969, p. 137). Reprinted with permission of the Veterans Administration.

pronation and injury to runners is well established, hence the broad consensus (Witness the attention to "heel control" by running shoe manufacturers). The answer thus appears straightforward—a high arch is an asset to runners.

However, I have experienced difficulty accepting this answer, particularly with respect to long distance running as a long term activity. How high is high? What precisely is an "exaggerated height?"

First, how high is high is another way of asking "What is the norm?" Isman and Inman[16] reported a mean of 41 degrees for the ankle of the axis of the talocalcaneal joint with respect to a horizontal plane (Figure 1). They found the range of inclination of this axis to be 20 to 68 degrees.[16] It would appear safe to assume, then, that one may be born with quite a "high" longitudinal arch and still be within the bounds of normality. It would also appear that Dorland's term, "exaggerated height," to describe a cavus foot is misleading. Two of Webster's definitions of "exaggerate" are "overstate," and "to enlarge or increase, especially beyond the normal." When the term "cavus" is used with reference to runners' feet, it overstates the case, in light of Webster's second definition.

This reference to abnormality brings us to the non-congenital, or "acquired," cavus foot due to "contractures or disturbed balance of the muscles" (Dorland's words). The musculature of the plantar aspect of the foot must be stronger than that of the dorsal aspect to cause the formation of a cavus foot. Such was frequently the case with growing children stricken with poliomyelitis.

Runners do not "acquire" cavus feet. The runner with high longitudinal arches was born with them. The A-P angle of his subtalar joints is more perpendicular than horizontal with respect to the ground. This is a condition that is generally considered favorable because it provides a built-in restriction to pronation of the mid-foot. A high arch is better suited to resist the superincumbent weight of the body. However, there is also one important disadvantage of a high arch—its relative rigidity makes it a less efficient shock absorber for the greater magnitude of impact forces that are generated by running. There is, then, a trade-off—hypomobility of the longitudinal arch versus a reduction of shock absorbency. Is this trade-off advantageous to the long distance, heel-to-toe runner over the long term?

As runners with high longitudinal arches began to appear with regularity at the Sports Medicine clinic, the clinical staff's unanimity of opinion with regard to the advantages of cavus feet began to fragmentize. From an orthotic overview, this puzzling question arose: Is not the possessor of so-called cavus longitudinal arches in actuality the possessor of what may best be described as feet with functional forefoot drop? The question is directed to the heel-to-toe runners only, i.e., Levels I, II, and III—or 98 percent of all runners.[2, Part III] A quite different set of biomechanical circumstances applies to the long distance (Level IV, "Elite") forefoot runner.

A GENERAL RULE

Assume the proposition that we are now dealing with two distinctly different types of feet, neither of which can be said to be abnormal *per se*. Structurally, the difference is one of degree between the normal non-cavus and the normal cavus foot. However, there are distinct functional differences resulting from these structural variances which have been overlooked.

One functional difference concerns the synchronous relationship of pronation of the longitudinal arch of the foot to transverse internal rotation of the tibia and, conversely, the raising (supination) of the arch to transverse external rotation of the tibia. From an orthotic viewpoint, an important feature of these motions is the fact that, as a general rule, blockage of the one automatically blocks its synchronous mate. That is, inhibition of phasic midfoot pronation also inhibits phasic internal rotation of the tibia, and vice versa. The same applies to phasic, midfoot supination and external rotation of the tibia.[21] There are exceptions and these exceptions are directly related to structural differences.

The initial pronation of the longitudinal arch (as a result of eversion of the heel at heel-strike) is a purely passive mechanism which is initiated by contact with the ground. The limitations of the range of either heel eversion or pronation are not dependent upon muscle control. The range of both motions is controlled, in order of importance, by the congenital placement of the axes of the subtalar and transverse tarsal joints, the geometry of their articulating surfaces, and their connecting ligaments.[21] Both motions are integral parts of the heel-toe running cycle; their occurrence at the beginning of the stance phase sets the alignment of the foot/ankle complex and, in so doing, affects all that follows.

There are great individual variations in the angle of the sagittal axis of the subtalar joint. These variations (Figure 1) alter the relation between the amount of pronation and supination of the foot and the amount of internal and external rotation of the tibia in the transverse plane.[14, 16] These tibial rotations are affected by the variations in the tangent of the angle of inclination in the sagittal plane. When the leg is vertical and the foot is at a right angle to it (flat on the ground), and the axis of the subtalar joint is 45 degrees, the internal-external rotations of the tibia would be equal in magnitude to their respective pronation-supination motions of the longitudinal arch.[14] It is because of this one-to-one relationship of these motions, at this angle of inclination of the axis, that 45 degrees was chosen as the benchmark for our general rule. The A-P angle of the subtalar joint in a living subject cannot currently be determined. Nevertheless, the benchmark gives a factual point of reference in order to discuss the relationship of structure to function.

When the angle of inclination of the A-P axis of the subtalar joint (as viewed in the frontal plane) is closer to the horizontal

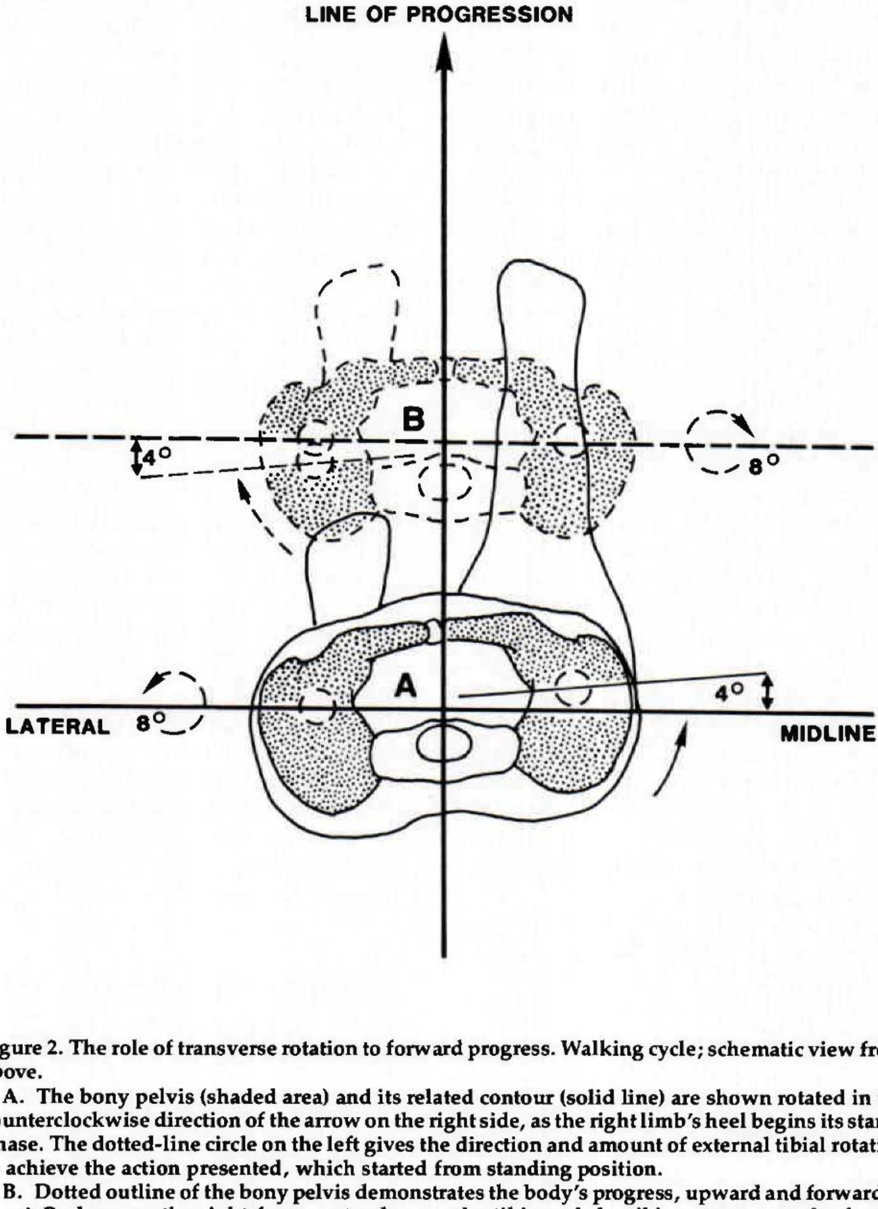

Figure 2. The role of transverse rotation to forward progress. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its related contour (solid line) are shown rotated in the counterclockwise direction of the arrow on the right side, as the right limb's heel begins its stance phase. The dotted-line circle on the left gives the direction and amount of external tibial rotation to achieve the action presented, which started from standing position.

B. Dotted outline of the bony pelvis demonstrates the body's progress, upward and forward in the A-P plane, as the right femur extends over the tibia and the tibia rotates over the foot to mid-stance. The dotted-line arrow on the left shows the clockwise forward rotation of the pelvis to the lateral midline and return to the standing position, as indicated by the placement of the left forefoot's dotted outline. The dotted-line circle on the right shows the eight degrees of external clockwise rotation, half its range, which reverses the tibia's internally rotated position taking the pelvis *et. al.* with it. The preceding eight degrees of internal tibial rotation, having occurred between foot-flat and mid-stance, are not shown.

(less than 45, but not lower than 20 degrees), the greater is the amount of pronation and supination for a given amount of internal or external rotation in the transverse plane[16] (Pes planus being representative of the abnormal non-cavus foot). Also, it can be demonstrated that when the oblique angle of the subtalar joint is closer to the vertical (more than 45 degrees) when viewed from the side, the magnitude of axial rotation to pronation and supination of the midfoot is greater.

It is in the normal cavus foot that we will find exceptions to the general rule. ("Normal" is here defined as an A-P obliquity of the axis of the foot's subtalar joint from 46 to 68 degrees to the horizontal.) The correlation between axial rotations to pronation/supination breaks down when the axis' A-P angle is greater than 45 degrees. For example, a given amount of tibial internal or external rotation effectuates less respective pronation or supination of the midfoot, the closer the axis of the subtalar joint is to the vertical. However, in spite of the fact that the range of pronation/supination is anatomically restricted in such cases, ground reaction forces being transmitted through the cavus foot (especially the vertical component) translate into a disproportionally larger transverse rotary force to the tibia.[14]

The more vertical the A-P angle of the subtalar joint axis, the less the amount of pronation possible. This fact provides insight into the biomechanics unique to the forefoot runner. The forefoot runner's gait does not seem to require depression of the longitudinal arches; they would be subject to injury were pronation to occur with each stride. Suppose we were to assume that the absence of pronation also eliminates the need for internal rotation of the tibia (and femur) in the transverse plane between foot-flat and mid-stance. Such an assumption seems to be valid, since heel-strike—the "trigger" of pronation—is bypassed by forefoot runners. Obviously, external rotation of the tibia (and femur), as well as pelvic rotation in the transverse plane, must be modified under such circumstances to enable the forefoot runner to effect an efficient stride. However, it would appear that a like accommodation cannot be achieved during a heel-to-toe running gait (Figure 5).

Another functional difference presented by the normal, so-called cavus foot is related to the hypomobility of the longitudinal arch of the foot with the condition here referred to as functional forefoot drop. This hypomobility of the midfoot necessitates an adjustment in order to effect heel contact in the standing and/or midstance position. The patient is unconscious of making this adjustment, which he performs by rotating his tibia anteriorly or posteriorly in the A-P plane (Figure 3). The consequences of this adjustment to a runner are discussed in detail under Condition IV.

One might well ask whether a heel-to-toe running gait should be advised for the beginner whose feet have functional forefoot drop. Also, it would appear that current practices of shoe selection and fitting of growing, active youngsters should be reexamined. The impact of this finding upon the design of commercial running shoes would appear to depend on the percentage of feet in the running population with functional forefoot drop. If the percentage is found to be substantial, it would be feasible for manufacturers to provide running shoes similar to the modified shoe shown in Figure 9.

EXTRAPOLATIONS FROM THE BIOMECHANICS OF THE WALKING CYCLE TO THE RUNNING CYCLE

Why should the orthotist find the synchronous character of certain motions of the tibia, hindfoot, and midfoot of special interest, when his patient is a runner? The three major reasons for his interest have been mentioned previously: first, the conviction that the foot/ankle complex is most vulnerable and most difficult to control between heel-rise and lift-off; second, the absence of a double-support phase in the running cycle leads to the presumption that there are no viable substitutions for hypermobility, hypomobility, or the loss of a particular motion, and third, the fact that the axis of the subtalar joint (the primary control mechanism of the synchronous motion) cannot be identified in living subjects. To determine the importance of each of the above to the running cycle, the orthotist must temporarily divert his attention further up the kinetic chain.

During running, the impact at heel-strike is 2.5 to 3.0 times body weight as compared to 1.2 times body weight during walking. This increase in vertical force can be assumed to generate a proportionately greater eversion moment during running. Because eversion of the heel is the key to midfoot pronation, it is reasonable to assume that its range also increases proportionately to the increased moment. With an increase in the range of pronation, there is a corresponding increase in the range of internal rotation of the tibia. This action would fall within the general rule, as the rule applies to the non-cavus foot.

When analyzed from an orthotic point of view, it becomes apparent that motions in the transverse plane are especially important to runners. The synchronous motions of depression of the longitudinal arch and internal rotation of the tibia, as well as the reverse motions of the raising of the arch with its synchronous external rotation of the tibia, are directly related to the length of stride of a heel-toe gait. What happens when these phasic motions are inhibited?

In order to answer this question, a closer look at the normal walking cycle from a fresh perspective was needed. We begin with a single step.

Imagine looking down from above on the lower half of a body as it begins to walk from a standing position (Figure 2). The left limb is in the mid-stance position as the right heel is about to contact the ground and begin its stance phase of the cycle. In the mind's eye, remove all portions of the skeleton above S1. We now look directly down on the pelvis and lower extremities. Draw an imaginary line in the sagittal plane which parallels the lower limbs. The line should be equidistant between both limbs and pass through the center of the first sacral vertebrae. This line represents the line of progression. Draw a second imaginary line perpendicular to the line of progression, so that it crosses the line of progression and passes through the center of the head of the left femur. This second reference line serves as the body's imaginary lateral midline (as viewed in the frontal plane). If the lower half of the skeleton were in a standing position (that is, if the left limb were to come to a stop and assume the standing position, instead of continuing forward to complete its stride in the cycle), the lateral midline would then pass through the center of the head of both femurs (Figure 2).

The total amount of external transverse rotation of the tibia is known to occur between mid-stance and push-off, whereas half the matching internal rotation occurs during the swing phase and is completed between foot-flat and mid-stance. All of the muscles within the limb in stance phase which are affected by these transverse motions are placed on stretch (eccentric contractions). None of them contributes any of the force necessary to effectuate transverse rotation during the stance phase. The role of the musculature, then, is to control external forces acting upon the limb in stance phase.

What is the significance of this arrangement to running? In seeking possible answers, it is first necessary to define the neutral position of the tibia and femur with respect to the transverse plane, as they relate to each other and to the foot. (Reference to the anatomic position would be confusing, since the tibiae are internally rotated when standing.) For this discussion, then, their neutral position is here defined as that position when all muscles within the limb are at their normal rest lengths, i.e., no muscles are elongated as a result of torque forces, nor are any contracted to generate torque forces. This definition immediately raises another question: At what instance(s), if any, during the walking and/or running cycle, are the tibia and femur in the neutral position? I submit that there can only be two instants when such is the case: just prior to heel strike, and just as the forward progression of the pelvis *et. al.*, moving in the transverse plane, reaches the lateral midline of the body at mid-

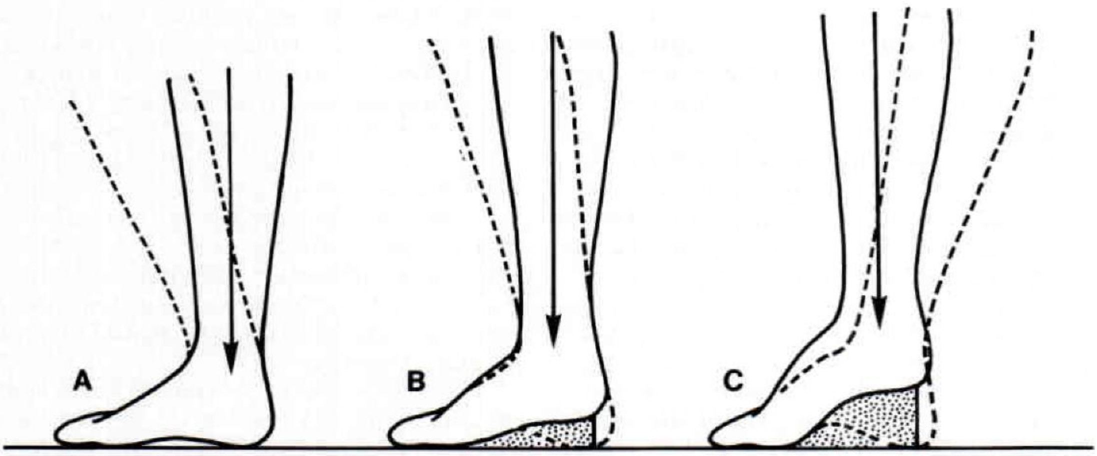

Figure 3. Foot/ankle complex's adjustments to forefoot drop to achieve standing balance in the A-P plane:

A. Tibia is vertical to a non-cavus foot and floor (solid line) for A-P balance. Downward arrow represents CG. Dotted line shows phasic 15 degrees of anterior rotation of the tibia prior to heel-rise at mid-stance.

B. Relationship of moderate forefoot drop to floor when tibia is in vertical position (solid line). Heel cannot reach the floor. Dotted line shows the two adjustments that allow the heel to make contact with the floor: 1. The midfoot and hindfoot are lowered together by rotation at the MP joints (the midfoot is hypomobile); 2. The tibia is rotated anteriorly to the degree necessary to restore standing balance in the A-P plane.

C. Solid-line drawing depicts severe fore-foot drop. Range of anterior tibial rotation is insufficient to achieve heel contact with the floor. The tibia must rotate posteriorly in order to gain the range required for restoring A-P balance. Unfortunately, hyperextension of the knee joints must accompany this adjustment.

A Glaubitz modification under the midfoot and heel (shaded areas, B and C) restores the tibia's normal perpendicular relationship as well as the full functional anterioposterior range to the tibia.

stance (Figure 4-A). It is only during these two instants that all muscles are at their normal rest lengths, due to the position of the tibia and femur whose actions affect them (with respect to motion in the transverse plane). Throughout the rest of the cycle, these same muscles are elongated beyond their normal rest lengths as each, to one degree or another, is "wound" around the tibia and/or femur. Once the foot has left the ground, these muscles are now free to "snap back" to their normal rest lengths. However, it is important to note that they do not go beyond their normal rest lengths ("snap back" being a passive activity), for to do so they would have to contract to generate the necessary force for such action. There is no evidence that concentric contractions are initiated by any of the extrinsic muscles of the foot to effectuate internal tibial rotation during the swing phase.

At heel-strike, then, both the tibia and femur are in the neutral position with respect to the transverse plane. Yet, only one-half of the full range of internal rotation has been completed (*i.e.*, the "snap-back" half of the range needed to match the full range of external rotation which had immediately preceded it). The other half of internal rotation occurs between foot-flat and mid-stance. Why? The answer lies further up the kinetic chain. It is the position of the pelvis at this time in the cycle, with respect to the lower limb which has just begun its stance phase, that points to the answer. In this case, maintaining a straight line of progression and economy of motion are the primary functions of nature's arrangement. The pelvis, with respect to the limb beginning its stance phase, is obliquely behind with references to the line of progression as viewed from above (Figure 2).

It is known that during fast walking, the increase in magnitude of the transverse rotary motions of tibia, femur and pelvis can exceed 50 percent of the range of average walking speed.[17] As our interest is in running, we will use the following ranges throughout the remainder of this discussion: tibia, 16 degrees; femur, 16 degrees; and pelvis, eight degrees. All of these convenient even numbers are within normal ranges for fast walking, and are, therefore, conservative estimates for running. It is not the magnitude of these motions alone that should attract our attention, because their direction and the phasic period(s) within the walking cycle in which they occur are equally important.

As walking speed increases, the magnitude of these motions increases proportionally. Therefore, it would seem to follow that during running, the proportional increase would be even greater, with cadence having the same relevance as it does during walking. A major component of cadence (when speed is the consideration) is an increase in stride length. Walking speed can be increased by stepping up the cadence without an increase in stride length. With the latter gait, however, Point A to Point B may be reached in a shorter time, but the cost in energy expended to distance travelled is disproportionately high. In order to cover more ground in less time (since increasing hip flexion alone results in an inefficient over-stride), a primary mechanism for increasing stride length is to increase transverse rotation of the tibia, femur, and pelvis.

Returning to the walking cycle, we pick it up as the left limb begins to rotate forward in the sagittal plane. The weight being borne by the right foot increases a-pace with the left limb's forward progression to mid-stance (Figure 4). From this point on, the body is being fully supported by the right foot, throughout the swing phase of the left limb. However, with all the body's weight positioned behind and medial to the right foot, forward rotation of the pelvis in the transverse plane is seriously compromised, unless accompanied by the phasic internal rotation of the tibia in the transverse plane, between foot-flat and mid-stance (Figure 5). In turn, the internal rotation of the tibia cannot occur without placing the extrinsic muscles of the foot on stretch as the tibia turns.

Keeping the values assigned previously, the normal course of events, with respect to the line of progression, would be eight degrees of internal rotation of the tibia to accommodate for the oblique, posterior position of the pelvis, which is four degrees counterclockwise to the right limb at foot-flat. Normally, the first eight degrees of external tibial rotation that occurs immediately following heel-rise at mid-stance, reverses the eight degrees of internal rotation the tibia was in, and thereby, along with the femur, reverses the four degrees the pelvis was in, thus bringing the pelvis to the lateral midline. The external rotation of the tibia and femur continues on to the completion of their respective ranges of 16 degrees each. The last eight degrees of external tibial/femoral rotation effectuates a four degree rotation of the pelvis forward of the lateral midline. The completion of the 16 degrees of external tibial/femoral rotation is reached at push-off. The muscles return to their normal rest length during the swing phase, causing the tibia and femur to return to their "neutral" positions (Figure 4).

Applying the same assigned values, what effect would either a partial or total absence of the lowering of the longitudinal arch and the synchronous internal tibial rotation have upon the running cycle? What are the conditions that can inhibit these two phasic motions during the heel-to-toe gait of the runner with non-cavus feet? Any condition that prevents depression of the longitudinal arch at mid-stance can be a causative factor.

For instance, when an insert is well molded to encompass the heel and longitudinal arch in their neutral positions and is sufficiently rigid to receive the full weight of the body at mid-stance without distorting, depression of the arch between foot-flat and mid-stance is blocked, as is the synchronous tibial internal rotation. The immediate result is a shortening of the stride length of the contralateral limb which is in swing phase. This is caused by denying the latter half of the full magnitude of normal transverse internal rotation of the tibia at the talocalcaneal (subtalar) joint. The heel everts at heel-strike, which normally "unlocks" the midfoot with respect to the hindfoot, but the rigid insert, as it firmly encases the hindfoot and midfoot together, maintains a neutral relationship between the two by mechanically "locking" both together. Albeit, the mid-foot still pronates with the eversion of the heel, but the rigidity under the arch prevents its phasic depression. Thus, two important motions, depression of the longitudinal arch and internal rotation of the tibia, are inadvertently eliminated from the normal sequence of events within the walking and/or the heel-to-toe running cycle(s) of non-cavus feet.

How does the elimination of these two motions effectuate a shortening of the stride of the limb in swing phase? Visualize a limb in stance phase that has been denied the two motions discussed as mid-stance is reached (Figure 5). At this time, the triceps surae of the contralateral limb pushes the contralateral limb off into its swing phase. This thrust provides the power to rotate the pelvis forward in the transverse plane about the vertical axis of the limb in stance phase. However, the tibia is in a neutral position, *i.e.*, the phasic internal tibial rotation that would normally occur between foot-flat and mid-stance has been effectively blocked by the rigid insert. The pelvis, taking with it the limb in swing phase, begins moving forward in the transverse plane about the vertical axis of the limb in stance phase. At this period of the cycle, the first half of the tibia's full range of external rotation is now superfluous, *i.e.*, it is not needed to rotate the tibia to the neutral position as mid-stance is reached because the tibia is already in the neutral position.

The force that the left triceps surae is able to generate, in order to pivot the pelvis *et. al.* forward under the circumstances just described, is more than equal to the task. The point of application of this force, at the left outer rim of the pelvis, gives it an enormous mechanical advantage. The weight of the pelvis *et. al.*, plus the required additional momentum, would apply a greater amount of torque than usual upon the muscles of the right limb that would increase the amount of their elongation to allow the tibia and femur to rotate beyond their normal ranges.

However, in this case, the latter eight

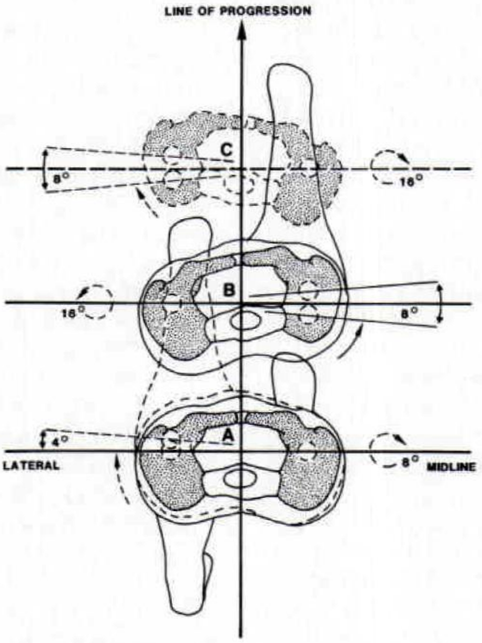

Figure 4. The contribution of rotations in the transverse plane to symmetry of stride. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its related contour (solid line) are shown in the mid-stance position, as indicated by the solid-line right foot and left limb, with its flexed knee, at the midpoint of its swing phase. The dotted outline of the pelvis and left limb have continued forward to heel-strike, completing the swing phase as shown by the dotted arrow on the left side. The dotted-line circle on the right gives the direction and amount of rotation of the right tibia to achieve the action presented. The preceding eight degrees of internal tibial rotation, having occurred between foot-flat and mid-stance, are not shown.

B. The bony pelvis and its related contour depicts the body's progress, upward and forward in the A-P plane, as the left femur extends over the tibia and the tibia rotates over the foot to mid-stance. The arrow on the right side shows the now counterclockwise forward rotation of the pelvis, from its previous position four degrees posterior to the lateral midline at foot-flat, on through to four degrees forward of the lateral midline, as the right limb completes its swing phase. The dotted-line circle on the left gives the direction and amount of rotation of the left tibia (the total range) to achieve the action presented. Again, the preceding eight degrees of internal tibial rotation, so essential to the symmetry of stride, could not be shown.

C. The dotted outline of the bony pelvis demonstrates that the upcoming full stride should be a mirror image of the preceding full stride (shown in B)—the same amount of upward and forward progression of the body through space in the A-P plane, the same amount of pelvic and tibial rotation, but both now in a clockwise direction. Though not shown, the preceding internal tibial rotation was in a counterclockwise direction, the opposite of B's stride.

Note: Since our primary interest is the foot/ankle complex, all reference to transverse rotations of the femur were dropped. However, it is understood that internal/external rotations of the femur are phasic with and closely mimic the range and direction of the tibia throughout the walking and heel-to-toe running cycles.

degrees of external rotation would force the affected muscles to stretch twice their usual required elongation. The stretching of these muscles is a passive action. It is known that muscle fiber cannot be passively stretched beyond 60 percent of its normal rest length without rupturing.[27] The mechanical advantage of the triceps surae's action upon the pelvis is such that the muscles within the right limb can be stretched quite easily beyond the danger point. All else being equal, however, tearing at a muscle tendon junction, especially in persons over 30, is more likely to occur than rupture of muscle fibers. In younger individuals, evulsions occur more easily than ruptures of tendons.[27] Figure 5 also demonstrates that there is another very practical reason for limiting the range of transverse rotations for the purpose of lengthening one's stride under the circumstances just described.

It is now evident that rotation of the tibia, femur, and pelvis in the transverse plane can be directly related to the length of stride. The amount of increase to stride length that transverse rotation can safely contribute is directly related to the fact that these rotations cannot occur without placing a large number of muscles, within the limb in stance phase, on passive stretch (eccentric contractions). The mechanical advantage the action of the triceps surae of the contralateral limb has upon the pelvis, suggests that tendon tissue within the limb in stance phase could be subjected to excessive stress.

A substantial increase in push-off thrust of the triceps surae and anterior tibialis of the contralateral limb would now be necessary. This increase in thrust at push-off would be necessary to replace the assistive force that would have been supplied in the form of phasic eccentric contractions by the musculature of the limb in stance phase. This would seem to be a contradiction of a previous statement that there is no evidence that the musculature of a limb contributes to rotation in the transverse plane during stance phase. Such is not the case. The primary biomechanical function of all eccentric contractions is control. Nevertheless, when a muscle is placed on stretch beyond its normal rest length, its elastic properties enable it to store energy in a manner not unlike an elastic band when stretched. Like the elastic band, a muscle utilizes the energy thus stored to return itself to its normal rest length. Having gained the desired control, nature also uses the energy resulting from the eccentric contraction to return the tibia and femur to their neutral position during swing phase. It is the absence of this secondary source of energy, due to the loss of internal tibial rotation between foot-flat and midstance (bearing in mind that the foot is receiving a rapid increase in load during this period), that necessitates an increased thrust by the contralateral limb. There is an immediate cost, an increase in energy expenditure.

The absence of internal tibial rotation also eliminates placing the musculature of the limb in stance phase into the necessary amount of eccentric contraction to control the tibial/femoral counterclockwise rotations (as viewed from above). As seen from above, the pelvis, as in normal circumstances, would still be positioned obliquely four degrees posterior to the lateral midline of the right limb at mid-stance. Once the pelvis is posterior to a limb in stance phase, the magnitude of external tibial rotation must be limited to one-half of the representative 16 degrees to avoid injury.

From the foregoing analysis, we arrive at these conclusions: Actual forward progression of the pelvis, through space via the transverse plane, is a primary function of external tibial/femoral rotation. The functional range of external rotation for a step forward (using the previously assigned values) is eight degrees (Figure 2). The functional range for a full stride forward (using the same assigned values) is 16 degrees (Figure 4). The occurrence of internal tibial/femoral rotation between foot-flat and mid-stance is a prerequisite to extending a step into a stride.

To inhibit internal rotation of the tibia—that normally occurs between foot-flat and mid-stance—is decidedly not in the best interest of a runner. To do so causes a dilemma because his desire for speed

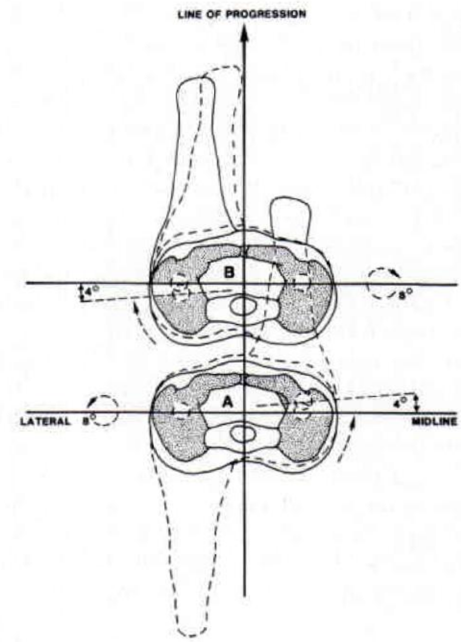

Figure 5. Effect upon stride length when phasic depression of the longitudinal arch is inhibited. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its outer contour (solid line) are shown in the standing position. Dotted-line contours show right limb at heel-strike and left limb at heel-rise, *i.e.*, the beginning of the double-support phase. The pelvis has rotated forward (counterclockwise) in the transverse plane about the left limb, as shown by the dotted arrow on the right side. The range and direction (counterclockwise) of the tibia's rotation are represented by the dotted-line circle on the left side.

B. The left limb has pushed off. The body has progressed through space, upward and forward in the A-P plane, as the right femur extends over the tibia and the tibia rotates over the foot to mid-stance. The solid-line contour of the pelvic region, the heel-strike position of the left limb, and the forefoot portion of the right foot (indicating mid-stance) demonstrate an arrested forward advance of the pelvis at the lateral mid-line following the phasic external rotation of the tibia and femur in the transverse plane. Such an arrest of forward advance of the pelvis can result from wearing a molded rigid insert which inhibits phasic depression of the longitudinal arch and its synchronous medial tibial rotation in the transverse plane, between foot-flat and mid-stance. As a result of the blockage of phasic medial rotation (counterclockwise) of the right tibia (not shown), its external range is shortened by half to eight degrees (dotted-line circle on right side) because the tibia has been out-of-phase in a neutral position from foot-flat to mid-stance. The dotted-line contour of the pelvic region and left limb demonstrate that any attempt at the full range of external rotation (clockwise) of the right tibia cannot effectuate forward progress of the pelvis beyond the lateral midline. Note how straightforward progression would be jeopardized.

urges him to a full-length stride, yet inhibition of internal tibial rotation dictates a shortened stride in order to avoid injury.

We have found why the latter half of internal tibial rotation in the transverse plane occurs between foot-flat and mid-stance. It is the key to maintaining control of the forward advancement of the pelvis during a heel-to-toe gait. The tibia's completion of the latter eight degrees of internal tibial rotation, between foot-flat and mid-stance, ensures a straight line of progression and symmetry of stride. Eight degrees of external tibial rotation reverses the internal tibial rotation, thus bringing the trunk to the lateral midline. Eight degrees of external tibial rotation brings the trunk and contralateral limb to heel-strike position. Symmetry is achieved during each swing phase by a form of "catch-up" previously referred to as "snap-back."

CONDITIONS I–VIII

The following is a list of eight conditions of the foot/ankle complex which appear with regularity in our Sports Medicine Clinic. A discussion of the variety of complaints made by patients who presented one or more of these eight conditions would be beyond the scope of this report. Orthotic management of their symptoms was clinically determined by the process of elimination. That is, all known possible causes (other than biomechanical ones) for each of the patients' complaints were judged to be unrelated. Each condition listed will be discussed individually from an orthotic point of view, followed by the orthotic solution and its rationale.

I. *Slight to mild excessive pronation in the standing position*

The patient has sufficient flexibility to raise his longitudinal arch voluntarily, while standing, without raising his heel or forefoot from the floor. When viewed from the back, his Achilles tendons indicate that the usual related heel eversion occurs. The patient is free of symptoms in all activities except running.

II. *Severe pronation in the standing position*

Hypermobility: Patient can raise the longitudinal arch voluntarily with ease. Transverse external/internal rotation of the tibia and femur are very apparent with the raising and lowering of the arch. The knee also shifts in and out of a valgum position with voluntary lowering and raising of the arch. Severe eversion of the heels is evident. The patient reports that his feet are often bothersome during the business day.

III. *Moderately rigid flat feet, usually in conjunction with bowed tibia*

This condition is usually reported to have been asymptomatic until the patient began running.

IV. *Functional forefoot drop ("cavus" foot), unilateral or bilateral*

Previously unknown to patient. The condition is hidden by involuntary, excessive dorsiflexion or plantar flexion at the ankle joint in order to effect heel contact in the standing and mid-stance positions. Patient was without symptoms prior to taking up running.

V. *Limited dorsiflexion*

Hypomobility: inability to rotate tibia forward beyond the neutral or mid-stance position. Not responsive to Achilles tendon stretching exercises.

VI. *Loss of eversion of heel*

Hypomobility: os calsis cannot evert beyond neutral position.

This condition is most common among the runners seen in our clinic.

VII. *Heel in fixed inversion*

Hypomobility: range of fixed inversion of os calsis seen as much as 20 degrees from neutral position. Amount of inversion may or may not be the same bilaterally. The patient is symptom free during activities other than running.

VIII. *Hypermobile transverse tarsal joint*

Abnormal pronation of midfoot occurring in this joint. This condition has been seen in isolation. When not a part of a gen-

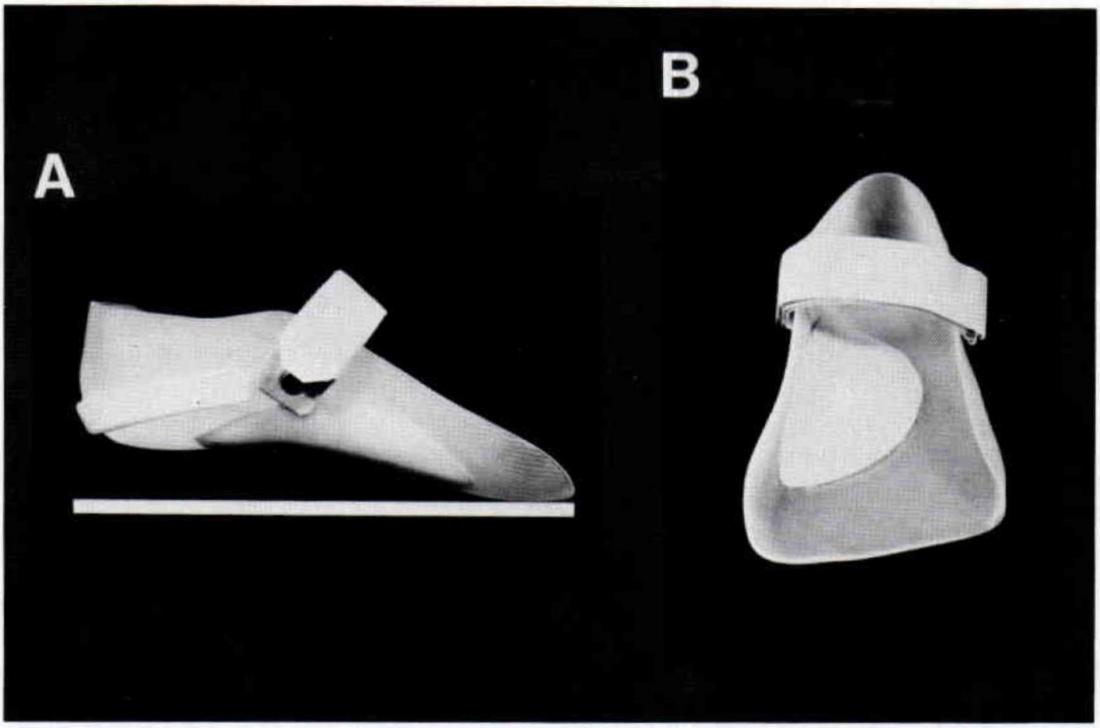

Figure 6. The Flexible Polypropylene Insert: A. Viewed from the medial side, showing the molded reinforced heel portion, the Velcro® instep control strap and the PPT foam longitudinal pad seen through the thin polypropylene. Note the amount of forefoot drop. B. Three-quarter anterior view.

eral condition of hypermobility, it seems to be the result of long-term running, following the loss of phasic eversion of the heel and/or limited anterior tibial rotation due to functional forefoot drop.

CONDITION I

Comments from an orthotic point of view

If a patient has normal range of eversion/inversion of the os calsis and is asymptomatic throughout his daily activities, other than running, his range of pronation when standing is directly related to the angle of his subtalar joint axis and therefore natural to him. However, introduce to such a pair of feet a regimen of running, which automatically increases the impact to the lateroposterior border of the heel by a factor of 2.5 to 3.0 times body weight, and injury would seem to be only a matter of time.

This occurs for the following biomechanical reasons: The increased impact to the lateroposterior border of the os calsis increases the moment of heel eversion by the same factor of 2.5 to 3.0. It is reasonable to assume that such a force causes a greater degree of eversion of the heel to occur, particularly as supportive musculature tires during a long run. The increase in the range of heel eversion automatically increases the amount of pronation of the midfoot.

The angle of the axis of the subtalar joint normally determines the "neutral" position of the longitudinal arch. However, given repeated applications of force to the os calsis (of the magnitudes known to occur when running) for several thousand cycles in rapid succession and as muscles fatigue, such forces are more than likely to weaken the ligaments supporting the multiple jointings within the midfoot. Since hindfoot eversion and midfoot pronation are passive motions, i.e., they are not initiated by muscular activity but by external floor reaction forces, control of the range of either motion is determined by the A-P angle of the subtalar joint axis and by the geometry of the articulations involved. An increase of the interspaces between these articulations, due to weakened ligaments, can lead to serious breakdown of the biomechanical checks that keep these motions within normal ranges. That attrition can be an important contributing factor to the cause of injuries, when biomechanical malalignment and/or weakened ligaments are present, is a conclusion that is hard to dismiss.

Orthotic solution to Condition I

1. A longitudinal arch pad is cemented to the upper surface of the removable innersole of a new running shoe(s). It is important to this conservative treatment that

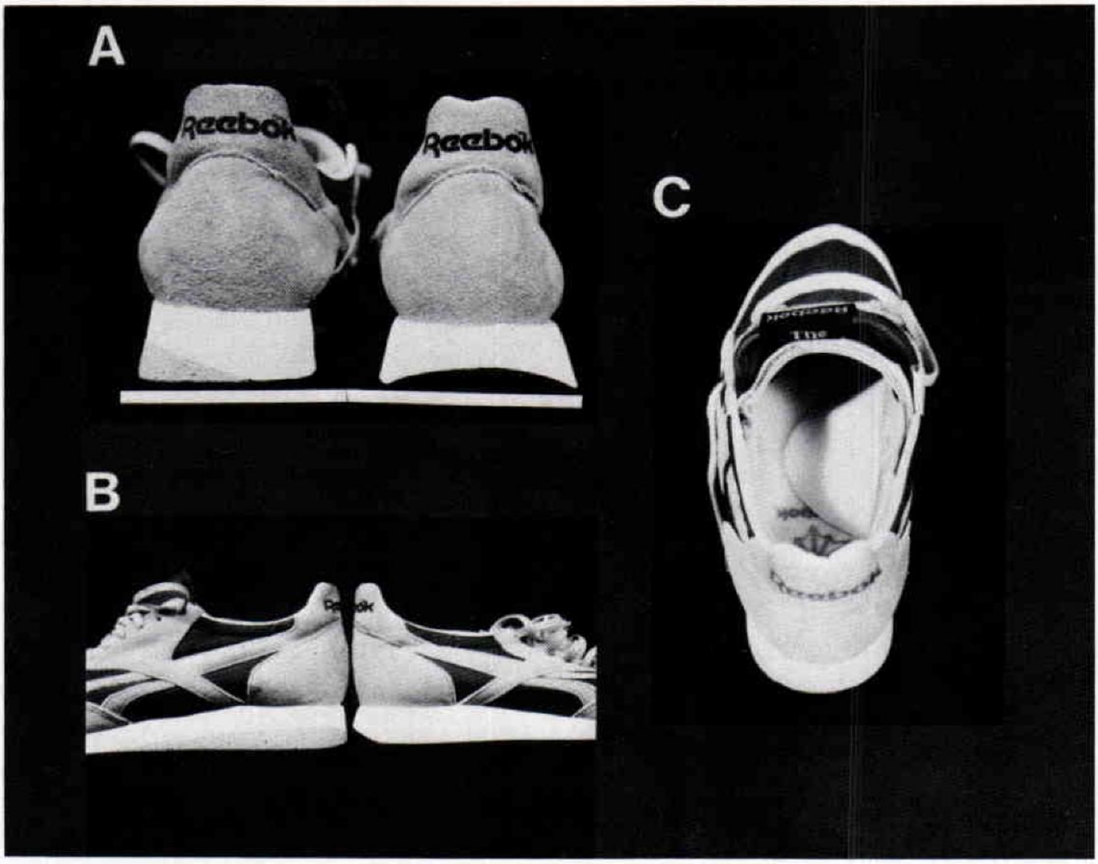

Figure 7. The Lateral Cushion Heel Wedge. A. Posterior view. The cushion wedge is shown on the posteriolateral portion of the left shoe. Note the downward inclincation to the lateral side. B. Lateral view showing the tapering of the cushion wedge to a zero point, distal to the fifth metatarsal head. C. A flexible polypropylene insert shown in a running shoe. The thinness of the polypropylene is evidenced by the design on the shoe's innersole showing through the bottom of the insert.

the running shoes, if previously worn, have not been distorted by the patient's pronated stride. If the running shoe does not have a removable innersole, the arch pad is cemented onto the inner surface at the proper location. The arch pad is made of 0.5-inch thick PPT® foam.

2. A lateral cushion heel wedge is applied to the lateral border of the running shoe(s) (Figure 7). It is also important not to attempt to apply the cushion wedge to a distorted shoe(s). (See Orthotic solution to Condition II for a description of the lateral cushion heel wedge.)

Rationale for use of lateral cushion heel wedge for Condition I

The PPT® arch pad offers dynamic control of the compressive effect of the body's weight upon the foot during mid-stance, but it cannot control excessive pronation at any point between foot-flat and mid-stance, because pronation has already occurred by then, i.e., at heel-strike, along with the eversion of the heel. The lateral cushion heel wedge controls heel eversion at heel-strike by limiting its range to normal dynamic requirements of the runner's cycle. (See Condition II for a description of how the lateral cushion heel wedge functions.)

CONDITION II

Comments from an orthotic point of view

We have just concluded that the purpose of the latter half of internal tibial rotation occurring between foot-flat and mid-stance is related to bringing the pelvis (when obliquely posterior to the limb in stance phase) forward to the lateral midline. In the lateral midline position, the pelvis is perpendicular to the line of progression, and the tibia and femur are in their neutral position in relation to the line of progression as well as to the right foot in stance phase (Figure 4).

From its lateral midline position, the pelvis continues its transverse rotation about the right limb, forward of the lateral midline. The left foot now completes its stride as its heel contacts the ground. The force generated by the weight of the pelvis and its forward momentum has externally rotated tibia and femur in the right limb simultaneously. As the double support period is reached (as viewed from above), the pelvis' position, with respect to the right limb, is forward, *i.e.*, the reverse of its relationship in the preceding double support period. Thus, the symmetry of internal/external rotation of the tibia and femur, which is essential to a straight line of progression, is maintained throughout the cycle. If one is in doubt as to the contribution of these transverse rotations to economy and efficiency of gait, observe a toddler walking or running. His side-to-side wobbling gait is largely due to the lack of (as yet unlearned) rotary motions in the transverse plane.

Excessive depression of the longitudinal arch is accompanied automatically by excessive internal rotation of the tibia. This condition places the foot's related extrinsic muscles continuously on stretch throughout the stance phase. This constant stretching during weightbearing causes permanent elongation of the muscle tissue. Unlike a normal, in-phase, eccentric contraction, the rest length of the muscle tissue is overextended, thus reducing the control associated with eccentric contractions. The motive power of concentric contractions is also seriously compromised. The most difficult dysfunction to treat, that results from such hypermobility, is the hindfoot's loss of control of the midfoot between heel-rise and lift-off.

Orthotic solution to Condition II

A flexible polypropylene shoe insert is vacuum-formed of 1/16 inch polypropylene over a plaster of Paris model of the patient's foot. The form is trimmed to cup the entire heel. The medial and lateral trim lines are extended to encompass approximately one-third of the dorsal surface of the foot. The distal trimline is at the distal edge of the metatarsal heads, in order to permit freedom for full flexion of the MP joints and to avoid any impingement of the metatarsal heads by the edge of the insert. Before forming the polypropylene, a stock, preformed, firm rubber longitudinal pad is cemented to the model under the arch position. Once the polypropylene is formed, a longitudinal pad of the same shape (3/8 to 1/2 inch thick at its center) is made of PPT™ foam. The trimmed insert is placed over the model and the outline of the firm rubber pad is traced on its outer surface. The PPT™ foam longitudinal pad is cemented to the inside of the insert within the outline. A Velcro® strap is attached to cross over the proximal portion of the instep when additional control is needed (Figure 6).

Out-of-phase flexibility of the 1/16 inch polypropylene insert is controlled by a snugly laced running shoe. Since the uppers of running shoes are made of woven materials, when drawn in by the laces the upper hugs the insert firmly from the sole level up about the sides and over the top of the instep. The flexibility of the materials used in running shoes is such that the uppers quickly assume the shape of malaligned feet, whereas the polypropylene insert, although flexible, is not stretchable and will not assume unwanted shapes. There is a "marriage" of the design characteristics of both the insert and the running shoe, each complementing the other to ensure the integrity of hindfoot to midfoot throughout the running cycle. Thus, the normal ranges of phasic motions are not inhibited, but abnormal magnitudes and/or out-of-phase motions are not permitted.

To achieve the control just described, it is essential that the dynamics of both heel and midfoot be free to perform their motions in proper sequence and within their normal ranges throughout the running cycle. This is accomplished by encompassing the heel and midfoot in a single, intimately fitting polypropylene form of the foot in the neutral weightbearing position. The mechanical encasement of heel and midfoot of a severely pronating foot mimics the normal funciton of the heel at heel-strike, *i.e.*, the midfoot and forefoot follow the direction of the heel at heel-strike, but the normal amount of depression of the longitudinal arch will not occur prior to the rapid buildup of weight between foot-flat and mid-stance. The intimately formed insert holds the hypermobile midfoot to its normal relationship to the heel without inhibiting the dynamics of a normal amount of depression of the arch, which is allowed as the PPT™ pad under the arch compresses. With controlled depression of the midfoot, synchronous internal tibial rotation also occurs in the proper phase of the cycle. During this same period, the insert offers sufficient resistance to prevent destructive magnitudes of either of these two motions.

As the heel begins to rise and the center of gravity moves over the forefoot, the normal action of "locking" midfoot to hindfoot by the inversion of the heel is assimilated by the intimate encasement of both these regions of the foot. The design again mimics nature by ensuring that, as the heel inverts, the midfoot must go with it—*i.e.*, the longitudinal arch rises and the tibia synchronously rotates lateralward, in the transverse plane, in their normal sequence in the cycle. Thus, excessive pronation is prevented from occurring, without inhibiting normal dynamics, at a time in the cycle that is particularly destructive. The control just described is particularly important in cases where the transverse tarsal joint is hypermobile because, as the heel inverts and the tibia rotates externally, the midfoot will continue to pronate out-of-phase. In such situations, the forces acting upon the extrinsic muscles that insert onto the midfoot are of a high magnitude, and these muscles are in danger of being stretched beyond their passive limits. Also, the flexibility of the insert's distal portion allows normal forefoot abduction to occur, in phase, without interference.

Although the design of the flexible polypropylene insert returns control of the midfoot to the heel without interfering with normal dynamics of the foot, there is one other essential control that it cannot provide. The insert cannot control the magnitude of the impact to the posteriolateral border of the heel, which is generated by the ground reaction force at heel-strike. In short, neither this nor any other insert can control eversion of the heel. This is because the motion is passive, that is, the motion is caused by an external force prior to the

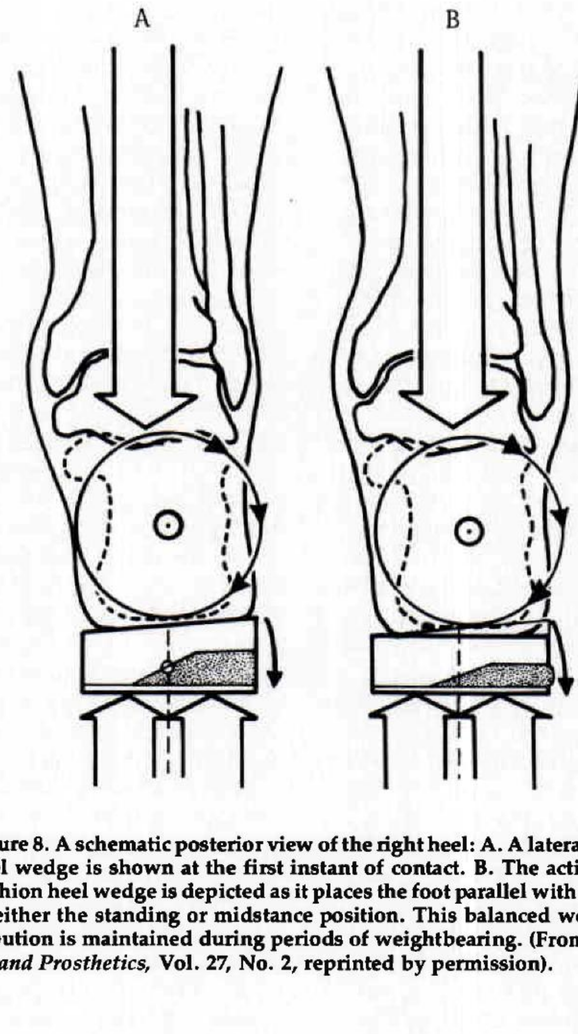

Figure 8. A schematic posterior view of the right heel: A. A lateral cushion heel wedge is shown at the first instant of contact. B. The action of the cushion heel wedge is depicted as it places the foot parallel with the floor, in either the standing or midstance position. This balanced weight distribution is maintained during periods of weightbearing. (From *Orthotics and Prosthetics,* Vol. 27, No. 2, reprinted by permission).

weightbearing phase. The floor reaction force will evert the heel to whatever degree allowed by the axis of the subtalar joint, regardless of the surface contouring within the shoe on which the heel rests.

Orthotic solution to Condition II: The lateral cushion heel wedge (U.S. Patent No. 3738373)

A lateral cushion heel wedge[9] (Figure 7) is incorporated into the shoe heel, in conjunction with the flexible insert, to control unwanted effects from impact at heel-strike. The cushion heel wedge is an integral part of the dynamic control system for the hypermobile foot. The control provided by the wedge is twofold: it reduces the magnitude of impact, and it limits heel eversion at heel-strike to normal dynamic requirements (Figure 8).

Design Rationale

Whereas, conventionally, a solid medial heel wedge is used to block valgus motion of the heel statically by placing the heel in a position of inversion upon weightbearing, the action of the lateral cushion heel wedge is diametrically opposite in function. When two shoes, one with a solid medial wedge and the other with a lateral cushion heel wedge of equal height, are placed side by side on a flat surface and viewed from behind, the medial side is higher on the shoe with the medial wedge, but the lateral side is higher on the shoe with the lateral cushion heel wedge. At heel-strike, the foot of the runner wearing the solid medial heel wedge is positioned upon an unyielding, inclined plane which prevents the heel from achieving a balanced or neutral position in the M-L plane. The foot of the same runner wearing a lateral cushion heel wedge would experience the following:

The instant the posteriolateral border of the shoe heel contacts the ground, the lateral cushion heel wedge begins to compress as the superincumbent weight from above rapidly increases. As the cushion wedge compresses, it absorbs and thereby delays for an instant the progress of the initial weight passing through it down to the ground.

Within that first instant, the lateral half of the heel due to its greater thickness, contacts the ground slightly before the solid medial half has made contact. By the time the cushion wedge has compressed to a level parallel with the ground, the solid medial half of the heel has also made contact.

The lateral cushion heel wedge cannot change the rate of descent of the body's weight to either side of the heel. It can, however, and does, effect a change in the amount of weight that reaches the ground at a given instant in time, *e.g.*, at the instant of heel-strike. The change is due to the "storing" of an unknown amount of weight by the lateral cushion heel wedge. The result is a slight difference in the amount of force making contact with the ground at any given instant in time, between heel-strike and heel-rise, through the medial and lateral halves of the shoe heel. In turn, the floor reaction force is proportionately imbalanced either side of the shoe heel. With the medial half of the heel receiving a greater force than the lateral half, a force-couple is produced which acts dynamically to maintain the os calsis in a position parallel to the ground (Figure 8).

As the tibia rotates anteriorly and internally following foot-flat, bringing the body's center of gravity forward and medialward over the midfoot, the load upon the os calsis increases rapidly. However, this rapid buildup of weight upon the shoe heel cannot affect the imbalance of ground reaction forces caused by the lateral cushion heel wedge, once the force-couple, acting upon the subtalar joint, is activated at the instant of heel-strike. With the preponderance of ground reaction force now passing medial to the axis of the subtalar joint, a moment of inversion is now acting upon the subtalar joint, instead of the phasic eversion moment that would otherwise be generated at heel-strike. The heel does not go into inversion, however, because the superincumbent weight and the floor reaction force on the medial side of the heel have equalized.

In order to produce a force-couple to control dynamically the eversion moment about the A-P axis of the subtalar joint at heel-strike, the action of the cushion heel wedge must be very quick. For example, Cavanaugh and Lafortune[4] report that a runner, running at a speed of six minutes a mile, travels from heel-strike to midstance in 42 milliseconds (ms). The mean ground contact time for 12 heel-toe runners tested was 188 ms per each foot at the same speed.[4]

A most pertinent finding from the same study was that the Z force was not recorded before a magnitude of 50 Newtons (110.23 lbs) was reached. This magnitude of vertical force was reached approximately two ms after heel-strike. These figures indicate that the lateral cushion heel wedge must compress much faster than two ms in order to shift successfully the preponderance of oncoming superincumbent weight to the medial side of the heel, before rapid buildup can overpower the lateral cushion wedge and result in a failure to control eversion of the heel. The speed with which the lateral cushion heel wedge must react at heel-strike is indicative of the need for the wedge to be incorporated into the shoe heel. For instance, were the cushion wedge placed inside the heel of the shoe, or under the heel portion of an insert, the delay in reaction time, in either case, would be sufficient to render it useless to runners.

CONDITION III

Comments from an orthotic point of view

The oblique angle at which the body's weight is received by the feet is due to the bowed tibiae. Were nature not to make this accommodation, the feet would be forced to receive the superincumbent weight in an untenable varus position. The oblique angle of the tibiae in a medial direction dictates that the feet accommodate by pronating. The lateral to medial direction, from which such feet receive the body's weight, due to the tibiae's lateral bowing, generates a moment that everts the os calsis about the axis of the subtalar joint, a moment of greater mechanical advantage than under conventional circumstances. The pronation accompanying bowed tibiae is both necessary and "natural." Consequently, no attempt should be made to alter the alignment of the foot/ankle complex which the patient presents when standing.

With few exceptions, these patients were asymptomatic prior to taking up running, and they continue to be without discomfort except when running. What phase(s) of the running cycle are most abusive to the runner with bowed tibiae and pronated feet? Decidedly, at heel-strike and, if the transverse tarsal joint is weakening, from heel-rise to push-off. At heel-strike, the moment to evert the heel is 2.5 to 3.0 times the runner's body weight versus 1.2 times when walking. Since these individual's heels are already aligned well beyond the normal eversion range at impact, with respect to their tibiae, it is reasonable to assume that heel-strike is the most damaging phase of the cycle to the foot/ankle complex. Once the heel leaves the ground and begins to rotate into inversion to lock the midfoot in preparation for push-off, the longitudinal arch, already depressed beyond the normal range of conventional alignment, may remain depressed if the os calsis' inversion action cannot control the midfoot due to a loose transverse tarsal joint.

Orthotic solution to Condition III

- Bilateral, lateral cushion heel wedges to running shoes, which will check further eversion of the os calsis at heel-

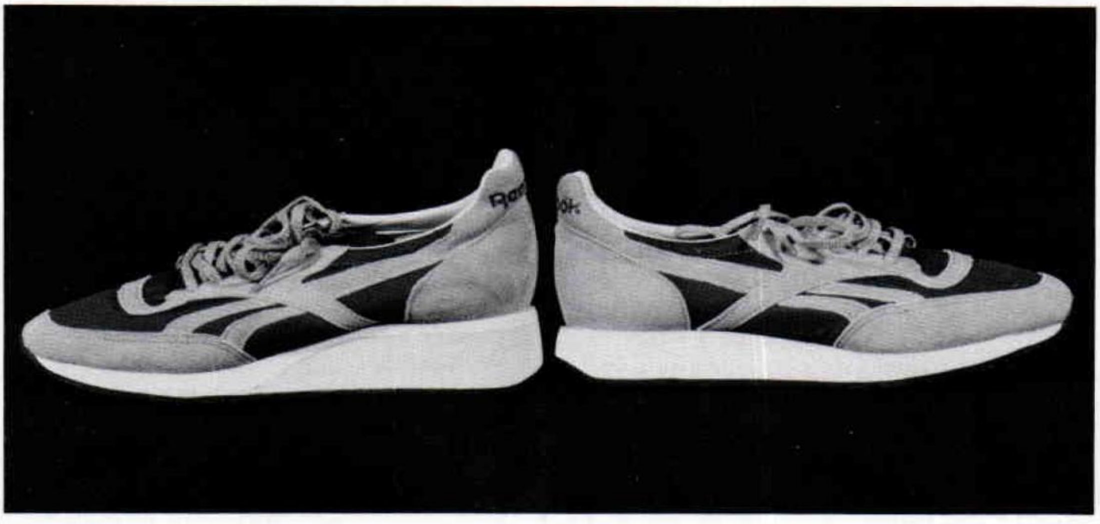

Figure 9. Lateral view of running shoe with Glaubitz modification for functional drop foot on left. Standard shoe on right.

strike in the manner previously described.

- Commercial Spenco® compressible arch supports to resist further unwanted depression of the longitudinal arch, if required, for the reason previously stated. A firmer arch support would be too restrictive, uncomfortable and, not incidentally, would inhibit the dynamics natural to such feet.
- If the above fails to give relief, the problem is a lack of anatomic control at lift-off. A flexible polypropylene insert with reinforced heel portion and Velcro® instep strap is required to restore contol of the midfoot (Figure 6).

CONDITION IV

Comments from an orthotic point of view

This condition is not easy to detect because it is natural to the patient, and he/she has throughout his/her life unconsciously masked the condition in order to effect heel contact in the standing and/or midstance positions. This the patient does in either of two ways: 1) anterior tibial rotation, the more common adjustment, is accompanied by posteriorward rotation of the pelvis to bring the center of gravity back over the feet to maintain postural balance in the A-P plane (When viewed from the side, the patient's posture is mildly reminiscent of the balancing stance of the paraplegic); 2) posterior tibial rotation is accompanied by hyperextension of the knees to bring the center of gravity forward over the feet to maintain balance in the A-P plane (Figure 3).

To the heel-to-toe runner, the unconscious postural accommodation for functional forefoot drop, *i.e.*, anteriorward rotation of the tibia at midstance, is significant for both their immediate and longterm effects. An immediate result of this accommodation is a shortening of the stride length of the contralateral limb, because at midstance the tibia of the limb in stance phase, which is normally identical to standing alignment, may have no tibial anterior rotation range left. The amount of anterior tibial rotation left at midstance is proportional to the amount of forefoot drop. For example, for each 1/16 inch the plantar surface of the forefoot is lower than the heel's plantar surface, the tibia will rotate forward approximately two degrees to achieve a plantigrade position. Thus, a ¼ inch difference between the forefoot and the heel requires approximately five degrees of anterior tibial rotation, a ½ inch difference requires approximately ten degrees, etc. The greater the amount of forefoot drop, the more of the normal maximum range of 20 degrees of anterior tibial rotation (dorsiflexion) about the talocrural joint[13, p .28] is used out-of-phase to achieve balance in the A-P plane during midstance.

During normal walking or heel-to-toe running, between midstance and toe-off, the tibia rotates anteriorly 15 to 20 degrees before the heel leaves the ground. The tibia of a runner with a ½ inch forefoot drop, for example, is already in a position of ten degrees of anterior rotation at midstance. If he is running fast, it is reasonable to assume he will utilize the full 20 degree range of tibial anterior rotation as he extends the stride length of the limb in swing phase. Consequently, since 50 percent of the total range has previously been used in an out-of-phase manner, the stride of the limb in swing phase will be shortened proportionally. The runner has but one option to make up for the deficiency in stride length—he must extend the weight of his body vertically by "vaulting" over the MP joints of his forefoot on the stance phase side. Preceding the vaulting, the heel must rise prematurely, which action prolongs the time (by 50 percent) that the midfoot must bear the brunt of the body's weight during the running cycle. The increased stress placed upon the transverse tarsal joint under such circumstances is not hard to imagine, especially as the extrinsic muscles of the foot tire during a run. Also, it is not unreasonable to assume that energy expenditure increases substantially due to having to raise the body higher with each cycle of the run—a use of energy that is counterproductive to efficient forward progression. Functional forefoot drop ("cavus" foot) also subjects the Achilles tendon to severe stretching which can cause tendonitis and, if ignored, microtearing of the tendon.

When the condition is unilateral (which in our experience has been infrequent), there is an additional factor to be considered, *i.e.*, once the heel of a limb with forefoot drop is off the ground, the involved limb is functionally longer than its opposite member. Since the full range of in-phase anterior tibial rotation is blocked, the normal 15 to 20 degree range of knee flexion that simultaneously occurs between heel-rise and toe-off is limited to a proportional degree; hence the functionally 'longer' limb and further need to vault over the MP joints.

This functional leg length discrepancy and the vaulting that accompanies it causes an asymmetrical running gait which is manifested in three important ways: The involved limb causes the uninvolved limb to drop lower to achieve heel-strike, thereby increasing the impact; the trunk will flex laterally to the low side and place abnormal stress upon the hip abductors on the involved side, particularly the gluteous medius muscle; the stride length of the uninvolved lower limb will be shorter. When this condition involves both lower limbs, stride length is shortened bilaterally, and a highly wasteful increase in energy expenditure due to bilateral vaulting is a consequence.

Orthotic solution to Condition IV

An adaptation to the Glaubitz shoe modification[8, 24] is used to restore the tibia to its normal position perpendicular to the foot and ground when standing and/or at midstance during the cycle. The construction of running shoes is such that they do not lend themselves to the conventional Glaubitz forefoot drop modification (Figure 9).

For mild forefoot drop, a lift contoured to the heel and midfoot is placed inside the shoe. When the patient is standing or the limb is in the midstance position with the Glaubitz modification, the aphasic anterior tibial rotation adjustment is no longer necessary in order for his heel to make contact with the ground. Thus, the normal range of anterior tibial rotation can now occur in phase, *i.e.*, between midstance and toe-off, because the tibiae's range is no longer dissipated by an out-of-phase motion to lower the heel to make contact with the ground.

When the forefoot drop is unilateral, a matching lift is applied in or to the shoe on the non-involved side. This is necessary to level the pelvis. It is also necessary to prevent the development of genu recurvatum in the involved limb. These heel lifts are made of firm, lightweight crepe. A softer material is not used because, when resting upon the average running shoe's .75 to 1-inch polyvinylacetate (PVA) foam midsole, the additional softness causes M-L instability.

CONDITION V

Comments from an orthotic point of view

When the foot is non-weightbearing, it cannot dorsiflex, either passively or actively, beyond the neutral position. The runner is unable to rotate the tibia(e) forward beyond the neutral position at midstance, causing premature heel-rise and the necessity to vault over the M-P joints. The tight Achilles tendon does not respond to stretching exercises. This condition results in a shorter stride length and an increase in the percentage of time within a cycle that the midfoot must support the superincumbent weight. Great stress is placed upon the transverse tarsal joint(s), particularly as the extrinsic and intrinsic musculature of the foot fatigue. Aphasic pronation is likely to occur following heel-rise in the attempt to compensate for the absent phasic anterior tibial rotation.

Orthotic solution to Condition V: The rocker bar

The rocker bar or "rollover"[23] is used for the same purpose as it is used generally, *i.e.*, to relieve stress upon the transverse tarsal joint by mechanically raising the heel at the proper time within a walking cycle. This is

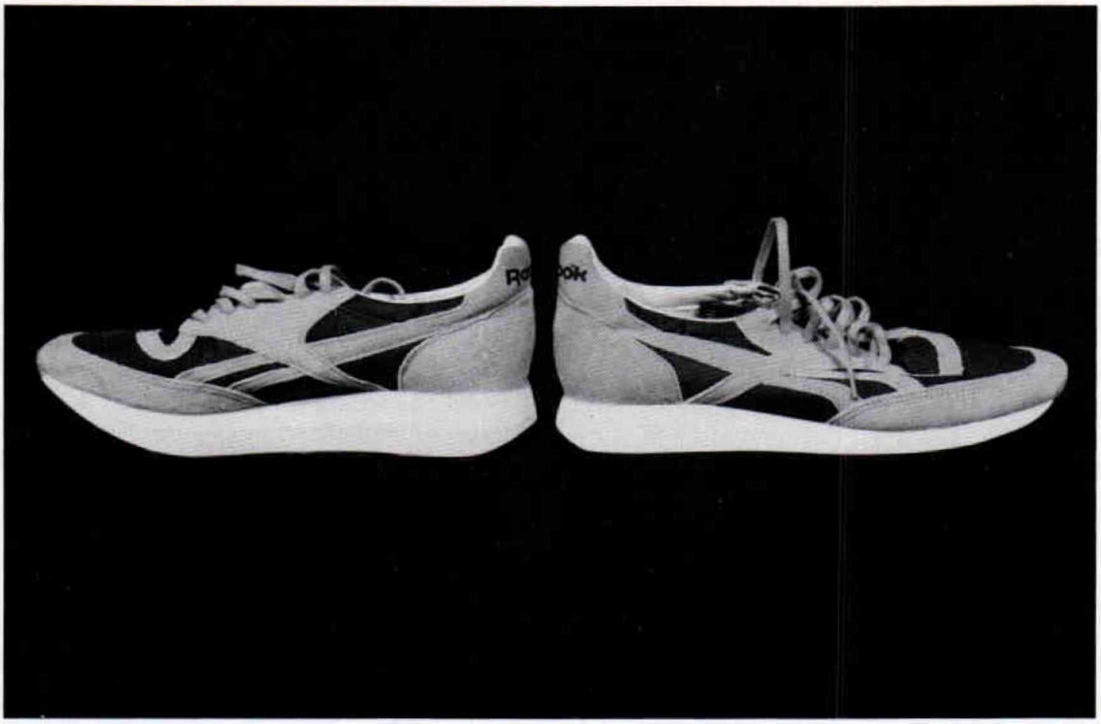

Figure 10. Lateral view of running shoe with rocker bar or "rollover" on left. Standard shoe on right.

done by 'rolling' the body's weight smoothly onto the forefoot (Figure 10). This action of the rocker bar is of primary importance to a runner, as it eliminates the necessity to vault over the M-P joints and restores the normal length of his stride. The timing of heel-rise is regulated by proper positioning of the rocker bar's apex with respect to mid-stance and the M-P joints.

Applying a rocker bar to a running shoe is not as simple a procedure as applying one to a conventional shoe. The process is complicated by two features of running shoes: their gridded contact soles and the need to preserve these soles' high friction surface, and the "wedgie"-like flatness of their contact surface.

The preservation of these two features, friction and flatness, necessitates removal of the gridded sole and the addition of a full-length, ½ to ⅝ inch thick, lightweight, firm crepe to the midsole. The addition is then sanded to form an apex directly behind the metatarsal heads to zero inches at the toe end and angled at the heel end to match the original heel contour. The gridded friction sole is then recemented over the added crepe rocker midsole. The rocker should raise the heel a minimum of 15 degrees (Figure 10).

Care must be taken not to introduce M-L instability by increasing the sole height to a point that magnifies the M-L moments acting upon the foot during stance phase. To prevent unwanted M-L instability, grind away an appropriate amount of the PVA foam midsole before cementing the firm crepe material to it. Unless there is a leg length discrepancy, a matching lift must be added to the opposite shoe in order to maintain a level pelvis.

CONDITION VI

Comments from an orthotic point of view

The os calsis cannot evert beyond the neutral position. This condition is very common among runners seen in the Indiana University Sports Medicine Clinic. All but "elite" runners have a heel-toe running gait. A three-year videotape study of the Boston Marathon showed that 98 percent of the participants "generally strike the ground on the outer heel."[2, Part III]

Since depression of the midfoot (longitudinal arch) is known to have a shock absorption function, and that eversion of the heel automatically places the foot in a pronation mode, prior to the occurrence of the depression, it is reasonable to assume that everting of the os calsis at heel-strike is an integral part of the shock absorption mechanism of the foot. It would appear to follow, then, that inability to evert the os calsis may seriously inhibit the shock absorption function of the foot. Furthermore, it is likely that detrimental consequences may affect the entire kinetic chain, particularly with regard to attritional factors over the long term. When one considers the force with which a runner's heel strikes the ground, it is difficult not to reach the conclusions expressed above.

Orthotic solution for Condition VI: The lateral cushion heel wedge

It has already been described how the lateral cushion heel wedge prevents excessive pronation, when used in conjunction with an insert, by dynamically controlling heel eversion beyond normal range. The rationale for the use of the lateral cushion heel wedge, when eversion of the heel is not anatomically possible, is based upon its ability to reduce the magnitude of impact at heel-strike.

Prior to heel-strike, the descent of the body may be said to be "falling free." The first point of contact is the lateroposterior border of the heel. The ground reaction force, at this point lateral to the axis of the subtalar joint, generates a moment of eversion to the heel. The magnitude of this moment is linked to the velocity of the body's fall. For example, the force of impact to the lateroposterior border of a shoe heel of the currently best shock absorption rated running shoe[6] absorbs all but 1.62 Gs of the impact force, which is normally at least 2½ times the runner's weight. The kinetic chain is subjected to a shock of 1.62 times the body weight when this shoe is worn, instead of 2½ times the body weight.

A runner wearing shoes with the above rating, for example, who weighs 150 pounds, strikes the ground with an impact of 375 pounds. Were he wearing a running shoe incapable of absorbing any portion of a 2.5 G force, the ground reaction force would be equal to the full impact of 375 pounds.

Let us assume that the lateroposterior point of contact of his shoe heel is a half-inch from the axis of his subtalar joint. Multiplying the half-inch radius to the force line by 375 pounds produces a moment of 187.5 inch pounds, or 15.6 foot pounds. The shoes with the above rating are said to absorb all but 1.62 Gs (or 35 percent less) of the 375 pounds, which would reduce the ground reaction force to 244 pounds.[6] When the 244 pounds is multipled by the half-inch radius, the product is now 122 inch pounds, or 10.1 foot pounds of moment to evert the heel. Whether it is 15.6 or 10.1 foot pounds of moment acting to evert a heel that anatomically cannot evert, either one is very jarring to the kinetic chain. A distance runner may run 130 km (80 miles) per week in training, which would subject each lower limb to approximately 40,000 such impacts over each seven-day period.[4]

The lateral border of a running shoe with a lateral cushion heel wedge is 3/16 inch thicker at its heel portion than at the shoe heel's medial border. This lateral wedge tapers to zero inches to a point just behind the head of the fifth metatarsal (Figure 7). From the instant of contact at heel strike, the posteriolateral border of the cushion wedge compresses rapidly, well before the first peak load is reached. The extra thickness of cushion provides an instant delay in transfer of the initial superincumbent load through the lateral portion of the shoe heel to the ground. As previously described, within the minute time frame that immediately follows heel-strike, an imbalance of floor reaction forces is effectuated by the cushion's action, *i.e.*, a higher magnitude under the medial side, there being no delay of the descending superincumbent load on its side.

However, it is the lateral border of the heel that first makes contact with the ground and therefore, the first imbalance of floor reaction forces is lateral to the subtalar axis and generates a moment of eversion. Also, the time frame in which this initial imbalance operates is extremely short, since the superincumbent weight is still falling at a rapid rate.

Whether or not the os calsis is anatomically able to evert, a moment to evert it is generated at each heel-strike. However, the lateral cushion heel wedge has served its purpose, for as it continues to compress, the medial portion of the heel is receiving a rapid and equal increase in vertical load. Since the free-fall weight has been uninterrupted on the medial side from the beginning, by the time the cushion has compressed to the point that the shoe heel is parallel to the ground, the floor reaction force is of greater magnitude on the medial side of the subtalar joint axis. The result is a moment of inversion to the heel which is in effect throughout the short time the heel remains in contact with the ground.

Thus, the lateral cushion heel wedge efficiently reduces the initial shock at heel-strike and simultaneously reduces the eversion moment to the heel, while maintaining the heel parallel to the ground during weightbearing.

The sequence of events just described occurs so rapidly during running that the role of the lateral cushion heel wedge for Condition VI is exclusively that of a very efficient shock absorber. This shock absorber effect is due to the anatomical inability of the os calsis to respond to a relatively light eversion moment which the wedge effectuates in the first instant of heel contact. The shock absorption is twofold: The subtalar joint is relieved of a high magnitude jolt in the form of an eversion moment, and the rest of the kinetic chain is relieved of a severe vertical jolt normally generated by the ground reaction force at heel-strike.

CONDITION VII

Comments from an orthotic point of view

The amount of fixed inversion of the os calsis seen in clinic has been as much as 20 degrees. The amount of inversion may or may not be the same bilaterally.

Biomechanically, this condition presents the same problems as Condition VI, limited heel eversion. However, it is quite possible that the eversion moment normally generated at the instant of heel-strike may cause additional premature inversion of the heel due to its inverted position. If this is the case, the lateral cushion heel wedge can inhibit this out-of-phase heel motion by reducing the impact at heel-strike.

Orthotic solution for Condition VII: The lateral cushion heel wedge

The rationale is the same as that for Condition VI.

CONDITION VIII

Comments from an orthotic point of view

When standing, the patient presents an abnormally pronated midfoot, but when viewed from behind the os calsis is not everted (the Achilles tendon is perpendicular to the ground). The integrity of hindfoot to midfoot has been lost. The most serious consequence of this condition is loss of the heel's ability to "lock" the transverse tarsal joint as it rotates into inversion following mid-stance. The result is that as the hindfoot inverts and the limb simultaneously rotates externally in the transverse plane, the midfoot will rotate medially in the frontal plane as the center of gravity advances over the midfoot.

With the flexing of the metacarpophangeal joints to the toe-off position, the center of gravity moves forward over the forefoot, the region where maximum vertical forces occur at push-off.[4] It is at this end of the stance phase that the runner's pronated midfoot is most vulnerable. The "Spanish windlass" action of the plantar aponeurosis, which passively raises the longitudinal arch as flexion occurs at the MP joints, is also inoperative when hindfoot/midfoot integrity is lost.

Orthotic solution to Condition VIII: The flexible polypropylene insert

The rationale and use of the flexible polypropylene insert are essentially the same as described for Condition II, severe pronation of the midfoot. However, due to the hypermobility of the transverse tarsal joint, restoring control of the midfoot to the hindfoot requires two additional components (previously mentioned) to the flexible insert design: a corrugation-like reinforcement is incorporated into the heel portion when the 1/16 inch thick polypropylene insert is vacuum-formed; a Velcro® strap which passes over the proximal area of the instep is attached to the insert (Figure 6). The purpose of adding one or both components is to prevent mediolateral spreading of the heel portion of the insert, which reduces the intimacy of fit, resulting in loss of control. The amount of laxity present in the transverse tarsal joint is usually a reliable indicator of whether the Velcro® instep strap will be necessary. Also, the Velcro® strap is mandatory for heavy individuals. Molding the insert out of thicker polypropylene for a heavier person is neither feasible nor tolerable, because its rigidity interferes with the dynamics of the running cycle.

During the past two years, 30 runners seen in our Sports Medicine Clinic have been using the flexible polypropylene insert with gratifying success. There are 65 runners now using the lateral cushion heel wedge; many of them have worn out several pairs of running shoes with cushion heel wedges during this period.

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A New Orthosis for Fixation of the Cervical Spine—Fronto-Occipito-Zygomatic Orthosis

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Mitsuru Oh-Hama, M.D.
Hikosuke Shingu, M.D.

INTRODUCTION

Most of the cervical orthoses for long-term fixation due to cervical spinal injury and other disorders, the Halo orthosis excluded, not only restrict the mouth movement to speak and to eat, but also give insufficient fixation against rotation because they have the fixing points on the mandibular occiput and body. A cervical orthosis was devised which does not fix on the mandible, to allow the mandible to move freely and which stabilizes the head by supports fixing on the frontal bone, occipital bone, and both the zygomatic bones and connecting to the trunk. The orthosis has been used for many patients with cervical diseases and has received favorable comment in our clinic since it is easily handled and provides excellent fixation.

FEATURES AND MERITS OF THE ORTHOSIS

If a patient in our clinic with a cervical spine disorder requires a non-invasive brace for firm fixation, the Fronto-Occipito-Zygomatic Orthosis—abbreviated as FOZY Orthosis hereinafter (Figures 1 and 2)—is considered to be the best orthosis of various cervical supports due to the following advantages.

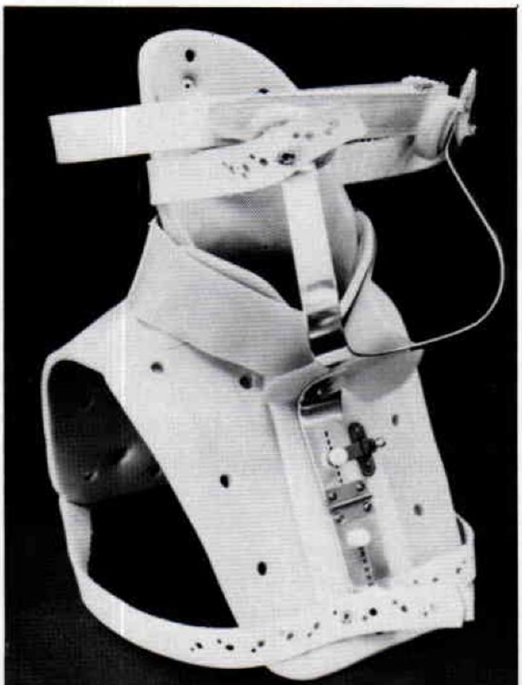

Figure 1. The FOZY Orthosis.

The FOZY Orthosis can limit the movement of the cervical spine in all directions, including rotation of the upper cervical spine, and therefore can be used for pa-

Figure 2. The anterior and the posterior parts.

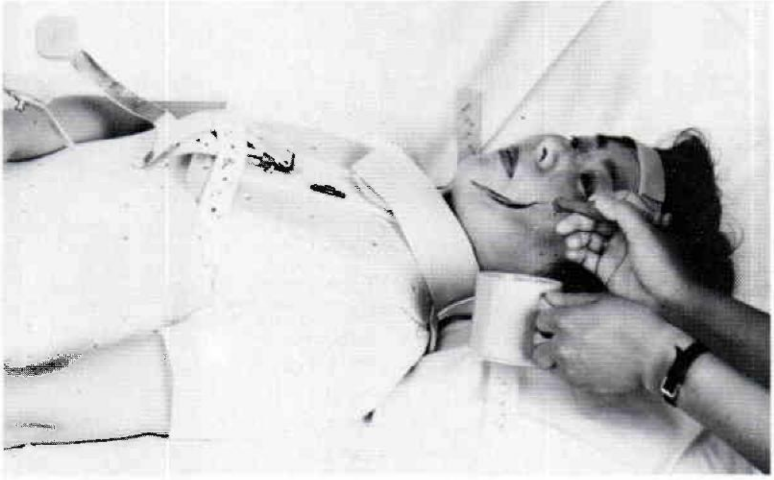

Figure 3. Taking a meal in a supine position.

tients with diseases affecting the upper cervical spine. As this orthosis has the fixing points on the frontal bone, occipital bone, and zygomatic bones, the mandibular bone has no limitation. The anterior part of the Superplast® support is cut short so the mouth can open freely to speak and eat.

The orthosis can be put on and off in the supine position so easily that it is adequately applied to patients with cervical spinal injuries. The support for the zygomatic bones can be adjusted to an adequate angle and be removed, depending on the patient's conditions, by use of a hinge and a screw (Figures 4 and 5).

For patients after the acute stage, it can be used as an orthosis similar to the Philadelphia type. It is light, easy to don and doff, and can be easily fabricated.

When putting it on a normal adult, the residual movable range is about ten degrees in every direction (In nodding action, the upper cervical spine has a limited motion within ten degrees only). The FOZY Orthosis is useful for long-term fixation after treatment with Halo-Pelvic or Halo-Jacket traction, and it is also widely applied for fixation of the middle and lower cevical spine. It is also useful after anterior fusion of the cervical spine.

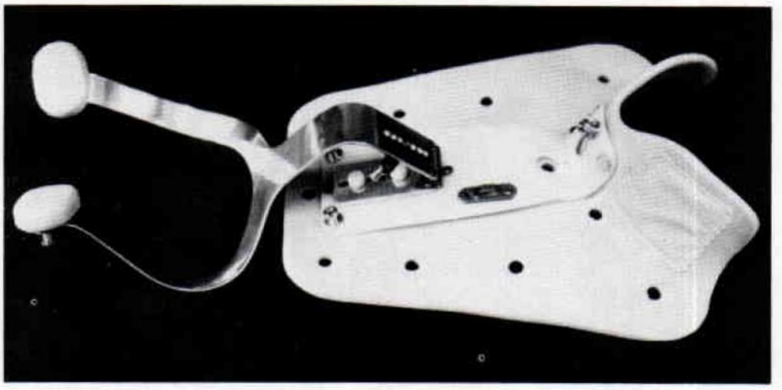

Figure 4. Adjusting the angle of the zygomatic support.

Figure 5. Removal of the zygomatic support.

DESIGN OF THE FOZY ORTHOSIS

The FOZY Orthosis consists of a firm cervical support provided with a Y-shaped support for both zygomatic bones and circumferential belts for the anterior head fixation. The orthosis is composed of two parts: an anterior part that fixes on the cheeks and the anterior thorax (Figure 8); and a posterior part that fixes the posterior head, the neck and dorsal surface of the body (Figure 9). The surface against the skin of both parts is covered with Superplast® of 6-10mm thickness and the supporting stem along the median line of the body is made of thermoplastic Subalsorene® of 3-4mm thickness. A Y-shaped cheek supporting plate made of light alloy (duraluminum plate of 2mm thickness) is attached to the front stem, and cheek pads are applied to its top. This supporting plate is adjustable to change the angle and height of the pads. It also has a hinge and screw to allow easy removal. The posterior part has four pairs of Velcro® straps; head belts support the anterior cranium; zygomatic belts are connected to the cheek supports; shoulder belts fasten the clavicular region and chest belts connect the sternal piece of the anterior portion. Its total weight is only 450 grams.

It is preferable to fabricate the orthosis over a positive plaster model of the patient. If modeling is impossible in an acute stage, it can be made based on the circumference measurements of head, neck, and chest, and on the distance between both cheeks.

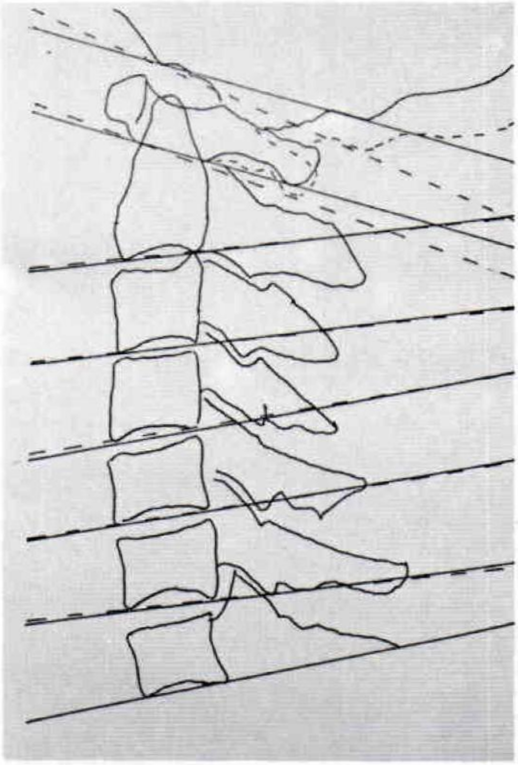

Figure 6. A sketch of an X-ray picture of vertebrae after application of the FOZY Orthosis to a normal adult. The solid line represents flexion and the dotted line represents extension. It shows that large movement is well controlled but upper cervical vertebrae remain able to move 5-10 degrees.

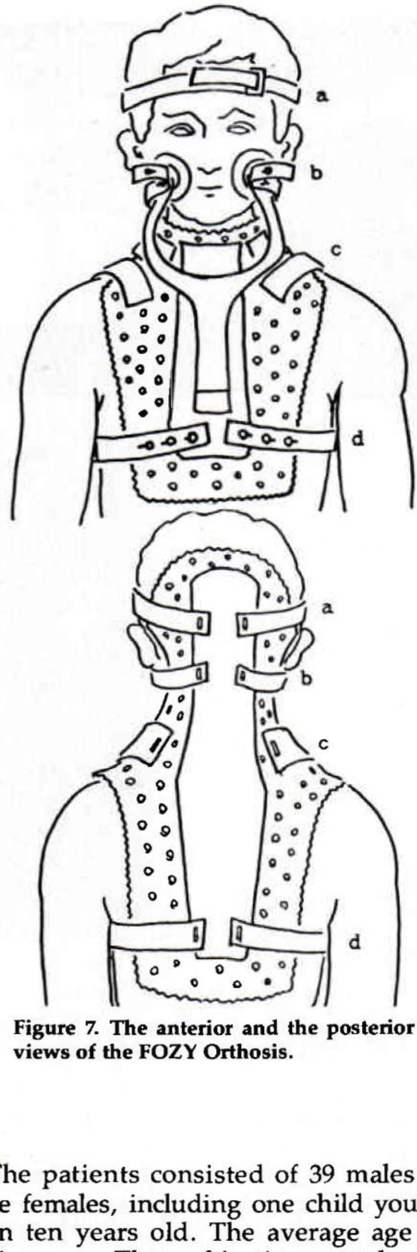

Figure 7. The anterior and the posterior views of the FOZY Orthosis.

CLINICAL EXPERIENCE AND INDICATIONS

The FOZY orthoses were fitted to 48 patients with cervical spine involvement. They were used in the conservative treatment of Hangman's fracture, dental process fracture of the axis, vertebral arch fracture and injuries of the intervertebral disc with spinal paresis in seven cases. They were also used for fixation after the anterior fusion for cervical subluxation and fracture, spinal cord injuries and cervical spondylosis in 27 cases, as well as for one-side opening spinal canal enlargements and for patients with cervical spinal canal stenosis in 14 cases.

The patients consisted of 39 males and nine females, including one child younger than ten years old. The average age was 52.1 years. The subjective vertebrae of FOZY Orthosis application were from the atlas to the second thoracic vertebrae and the application period was from one to three months. During this period, the patients were subjected to mat, standing, and ambulation training. Thus, rehabilitation for patients with cervical spine injuries could be actively progressed. Of the 48 participants, no cases had increased injury

Figure 8. The anterior portion.

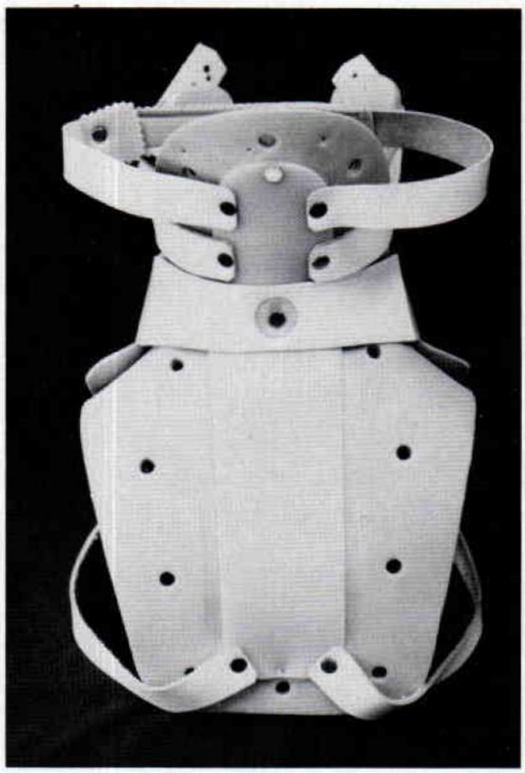

Figure 9. The posterior portion.

to the cervical spine or dislocation of grafted bone in the period of orthosis attachment. After discontinuing the orthosis, the range of cervical motion was almost normal, except for the patients with vertebral enlargement.

SUMMARY

Though the upper cervical vertebrae can move in five to to ten degrees merely to nod, the FOZY Orthosis can fix the cervical spine in every direction, including rotation, and also firmly limit the movement of the upper cervical spine. It can thus be applied to many diseases of the upper cervical spine, to cervical spinal injuries in an acute stage, and for external fixation after anterior spinal fusion. Since it is easily handled and light, early rehabilitation can be performed actively.

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Psychological Adaptation to Amputation: An Overview

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Jana Poole, O.T.R.

INTRODUCTION

There are approximately 400,000 amputees in the United States. It has been estimated that approximately 50,000 new amputations are performed each year.[13, 15] Most amputees are males in their early sixties, approximately 90 percent of the amputations involve the lower limb, and approximately two-thirds of lower limb amputations are necessitated by diabetes mellitus and/or peripheral vascular disease. Although there are circumstances when the amputation of a chronically painful and/or infected limb may be a welcome therapeutic alternative to a patient, the resulting disability is never welcome. In addition, although amputation of a limb is sometimes a lifesaving measure, it exacts a physical and psychological price from the patient. Since an increasing number of amputations are performed each year due to the expanding geriatric population,[13, 15] the process of psychological adaptation of patients to amputation seems to be a subject worthy of further investigation.

Our paper will review and synthesize the literature devoted to the psychological and emotional reaction of patients to amputation and integrate this information with our own experience in 248 patients with 368 amputations over the last five years. This report is primarily directed towards non-psychiatric allied health professionals, who, we feel, have the most impact in preventing disability and in reducing the need for professional psychiatric intervention.

ADAPTATION TO AND ACCEPTANCE OF AMPUTATION

The psychological reactions by patients after surgical or traumatic amputation of a limb are both varied and complex. Based upon our own experience and a review of the existing literature, we feel that an individual's adaptation to his/her loss of a limb can be artificially divided into a preoperative stage and three postoperative stages.[4, 5, 8, 17-20, 22] In all four stages of adaptation, certain emotions and defense mechanisms can be universally identified. These psychological phenomena are normal and help the amputee in adapting to a new body image, however, the overutilization of these mechanisms can result in more disability than can be accounted for strictly by the physical loss of the limb.

The first, or preoperative stage, of adaptation for the amputee begins with the patient's realization that the loss of limb is a possibility. This realization may or may not coincide with the first formal doctor-to-patient presentation of amputation as a therapeutic treatment. This stage is necessarily absent in some patients, for example, with emergency amputation due to traumatic injury. Grief is the universally identified reaction in those patients being told that they must lose a limb.[10, 20, 21] In addition to the grief reaction, the preoperative stage usually includes concerns about: 1) pain; 2) financial difficulties; 3) general health; and 4) future functional capabilities at home or on the job.[12] The overall psychology of the amputee is modified by how he/she perceives the pending amputation, which in turn is modified by variables such as the patient's culture, background, family and community.[3, 10, 21] Questions regarding the exact nature of function and use of a prosthesis, future sexual function, and even disposal of the amputated limb, are all questions which are also prominent in the minds of potential amputees.[3, 21, 24]

The second, or immediate postoperative, stage is a relatively short period of time which begins with the patient's first postoperative realization that the limb is no longer present and ends during the early phases of postamputation rehabilitation. Randall, *et. al.* in their study of 100 amputees, noted that the immediate reaction to amputation was modified by the circumstances surrounding the assault on the limb.[22] Those patients who sustained injuries in battle or in the line of duty, where loss of life was likely, were found to have a more optimistic future outlook immediately after amputation than those individuals who sustained their injuries through carelessness or unfortunate accidents where loss of life was not a significant risk.

However, the early acceptance of amputation as it relates to the mechanism of amputation injury/loss, does not seem to be a significant factor in the ultimate rehabilitation and acceptance of the disability.[4] In fact, it has been suggested that those showing the best early acceptance may have delayed depressive reactions upon return to society.[21] Our experience, in over 300 amputations (Tables 1 and 2), correlates with studies by Friedmann[4] and Randall, *et. al.*[22] We feel that the immediate response to amputation correlates well with the cause of limb loss, and that, in most cases, the early acceptance of an amputation does not seem to be a significant factor in the ultimate rehabilitation and acceptance of the disability.

As patients move from the second stage, (immediate postoperative), to the third stage, (inhospital rehabilitation), denial gradually replaces grief as the prominent feature in a patient's adaptation to amputation.[19] Euphoric mood, regression and withdrawal are mechanisms used by patients to deny both anxiety and the challenge of adjustment to reality.[17] Patients often deny their injury with statements and demonstrations of physical prowess, such as wheelchair racing in the hallways, boisterous behavior on the ward, and jokes about their respective physical injuries.[2, 3, 17]

Also seen during this transition period is a process described by Parkes[19] as "pining," wherein an amputee grieves for the lost limb, a process primarily represented through the amputee "pining" for those aspects of life lost with the loss of limb.

The inhospital, postsurgical, rehabilitation adjustment period can be made more complex by surgical limb revision, manipulation, prosthesis fitting and training, and adjustments to friends and relatives from whom the patient was separated during the first and second stages. It is at this point in the adaptation process that the patient begins to feel deeply depressed, insecure, uncertain, apathetic, and preoccupied with limb loss.[4, 20, 22]

Many of these feelings arise from the patient's interaction with those people close to him/her. Insecurity and anxiety stem in part from the amputees' concern over the anticipated reaction of loved ones and the actual sympathy that they eventually receive from them.[22] It has been suggested that sympathy serves as a reminder to the patients of their amputation and that empathy is both more appropriate and supportive.[4, 22] Most authors agree, however, that although depression, anxiety, and feelings of self pity are prominent during the third stage, the need for formal psychiatric intervention is indicated in relatively few patients.[17, 20] That is not to say that there are not significant problems worthy of professional psychiatric attention in patients during this early adaptive process, rather it is to emphasize the need for supportive intervention on the part of those non-psychiatric personnel (prosthetists, therapists, nurses, etc.) involved with amputee postsurgical care and rehabilitation.

The fourth or final stage of adaptation begins with the patient's return home, usually several weeks after amputation. While leaving the hospital represents some evidence of recovery to the amputee, it also forces upon him/her the more harsh realities of disability. By the time of hospital discharge, most patients have undergone some prosthetic fitting and many have actually begun or are well adapted to ambulation as an amputee. However, upon returning home, the amputee is abruptly faced with a marked decrease in supportive help (previously provided by hospital personnel) and a marked increase in demands that manifest his/her disability, both physically and emotionally.[10, 20]

It would appear that the amputee's return to home is a crucial turning point in the adaptation/rehabilitation process. Available evidence suggests that the amputee will either successfully adapt during this final phase and learn to live with his/her disability, or will fall back into a pattern of psychological behavior which represents a continuation of the third stage (denial) of adaptation.[20, 22]

Certain demographic factors may also play a role in ultimate social adjustment. While younger patients may have more difficulty in initially adapting to a "new body image," older patients tend to have more difficulty with longterm social adjustment.[20, 22] In addition, single individuals have more difficulty than married individuals and lower extremity amputees have more difficulty than upper extremity amputees.[20, 22] Not surprisingly, those individuals with multiple amputations have more difficulty than those individuals with only unilateral amputations. Finally, it is of interest to note that no real differences have been identified between male and female amputees in terms of social adjustment after amputation.[19]

The experience of the authors is in agreement with the studies by Parkes,[18-21] Randall, *et. al.*[22] and Reinstein, *et. al.*[23] with respect to age, sex, marital status, number of amputations, and upper versus lower limb amputations as each of these factors relate to the process of rehabilitation. A large part of the psychological reaction to the fourth stage is secondary to environmental influences largely out of the control of the patient. With proper support and aggressive rehabilitation, the final stage of social adjustment for the amputee is successful in time, whereas without support, social adjustment is seriously impaired.[13-15]

GUIDELINES FOR THE PSYCHOLOGICAL MANAGEMENT OF AMPUTEES

It is unfortunate that many times, amputation surgery, as a therapeutic alternative in the management of patients with limb threatening problems, has been looked upon with a jaundiced eye in relationship to seemingly preferable limb salvage procedures. Of particular interest, therefore, is a recent study by Sugarbaker, *et. al.* who did a quality of life assessment in 27 patients, roughly half of whom had limb sparing surgery plus radiation and chemotherapy for limb cancer, while the other half had amputation surgery and chemotherapy. It was the initial hypothesis of this group that limb sparing surgery resulted in a higher quality of life as compared to those patients treated with amputation; however, this hypothesis was not demonstrated and, in several instances, the amputee group had higher quality of life scores than their counterparts in the limb salvage group.[25] Overall, the quality of life assessment for both groups was not found to be significantly different.[25]

Tucson VA Medical Center and University of Arizona 7/1/77–6/30/82	
Number of Patients	248
Number of Amputations	368
Primary	300
Preparatory	54
Revisions	14

Table 1.

In our opinion, and that of Bowker,[1] amputation surgery should be viewed by all involved personnel as a reconstructive, not a mutilating procedure. A team approach is optimal in amputee rehabilitation and should include the surgeon, ward team, surgical nurses, prosthetist, physical therapist, occupational therapist, social worker, vocational counselor, and, if indicated, a psychiatrist or psychologist.[6, 14, 21]

The psychological preparation of the potential amputee should begin as early as possible, and preferably, should begin preoperatively (Stage I).[1, 4, 7, 10] When the amputation becomes a possibility, not a probability, the patient should be informed and the entire amputation rehabilitation process should be discussed. Open communication is essential and should specifically address the hows, whys, and wherefores of the operation itself, disposal of the amputated limb or part, expected phantom phenomena and phantom pain, sexual and social readjustment, prosthetic fitting and training, and the process of amputee rehabilitation.[12, 21]

Surgeons or rehabilitation teams who keep the possibility of amputation from their patients until it becomes inevitable, are left with far less desirable conditions for successful rehabilitation than surgeons or rehabilitation groups who integrate the patients and families into an early program of planned prosthetic rehabilitation.[1, 21] The authors would suggest that keeping the possibility of amputation from the patient until the last possible moment serves only to cultivate the attitude in the patient that the amputation is a treatment failure

Tucson VA Medical Center and University of Arizona
Primary Amputations
7/1/77–6/30/82

Amputation Level	Number of Amputations
LOWER LIMB	
Toe	49
Transmetatarsal	10
Symes	22
Below Knee	124
Through Knee	15
Above Knee	61
Hip	1
UPPER LIMB	
Partial Hand	1
Below Elbow	9
Above Elbow	7
Shoulder	1
TOTAL	300

Table 2.

and not a lifesaving or reconstructive surgical procedure. Integration of the family into the support group for the amputee is also recommended as early as possible (Stage 1). The authors believe that such family involvement decreases some of the social problems that an amputee would otherwise face on his return home (Stage 4).

In many institutions, amputation surgery is delegated to junior house officers, often without supervision. Such an approach, in the opinion of the authors, compromises optimum rehabilitation results. Amputation surgery should command the attention of the senior surgical staff. In our program, the senior surgical attending is directly involved in the performance of all amputations and supervises the entire process of amputation rehabilitation. A poorly performed amputation almost guarantees poor rehabilitation. While a well performed amputation does not guarantee a successful rehabilitation outcome, it certainly makes successful rehabilitation more possible.

Early prosthetic fitting and amputee rehabiliation (Stages 2 and 3) are vital to a patient's successful physical, psychological, and emotional recovery, both from a short-term and longterm standpoint.[9, 11, 13-18]

Early prosthetic fitting and rapid rehabilitation enable the patient to incorporate all of his physical and emotional efforts into recovery from the earliest possible moment, rather than allowing the patient to focus only on disabilities and pain.[4] An important corollary to this principle is the early introduction, to the potential amputee, of the patients who have undergone similar amputations and successfully adapted to their prosthesis and their social environment.[16] The experience of the authors supports the view expressed in the literature, that the introduction of a successful amputee patient to a potential amputee has been very helpful in the rehabilitation of the latter individual.[4, 10, 21]

CONCLUSION

Of utmost importance in the rehabilitation of any amputee is the realization that the rehabilitation process is a lifelong effort. It is only with concerted effort that the rehabilitation team will be able to provide the necessary reassurance for each amputee in order to get them through the gate of the rehabilitation process.[1] It is our contention that with a better understanding of an amputees' psychological and physical needs, they need not become more disabled than necessary by the loss of their limb alone. In addition, it is the authors' contention that professional psychiatric intervention is required for relatively few amputees, if allied health personnel play a continually active role in the rehabilitation process.

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NOTES

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A Multi-Adjustable Cervical Orthosis for Use in the Treatment of Torticollis (A Case Study)

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INTRODUCTION

In the treatment of torticollis, the use and design of cervical orthoses is varied depending on the severity of the deformity, surgical intervention, if any, and continued follow-up care relative to the initial success towards its resolution. Orthoses may be simple collars or poster-type braces. In certain instances, intricate custom devices may be most appropriate, such as described by Townsend.[1] For the most part, these orthoses act to control the deformity but are often limited in their adjustability, comfort, and ease of use, especially in those instances when the physician has sole contact with the device. It was the need to incorporate these design requirements into a functional unit that prompted the suggestion of a new orthosis. The orthosis described in this paper provides an alternate means to orthotic treatment of torticollis and related conditions.

CASE HISTORY

The patient for whom the cervical orthosis was designed was, at the time of the treatment, a 19 year-old female with a diagnosis of congenital torticollis, significant enough to produce feelings of self-consciousness when in public. Upon examination and evaluation, it was decided that a surgical release of the tight sternocleidomastoid muscle on the left side of her neck would be the treatment of choice. Post-surgical care was to include an orthosis to maintain the new position and to prevent recurrence of contracture. The orthosis was to be worn for 24 hours a day for the first three weeks, with weaning of the orthosis over the next three months.

In past instances, a SOMI[2] type orthosis has been prescribed for use, but not always without problems. Patient tolerance and ultimate rejection is a problem, as well as is skin breakdown. In addition, the fitting of this type of brace varies from practitioner to practitioner. If the patient is seen in the doctor's office, adjusting the orthosis would prove difficult without the orthotist present (it would be necessary for the patient to return to the orthotic practitioner's office). Therefore, a custom total contact support that would provide better patient acceptance and eliminate skin irritation, along with a design that would allow the physicians to make position adjustments in the office, was indicated.

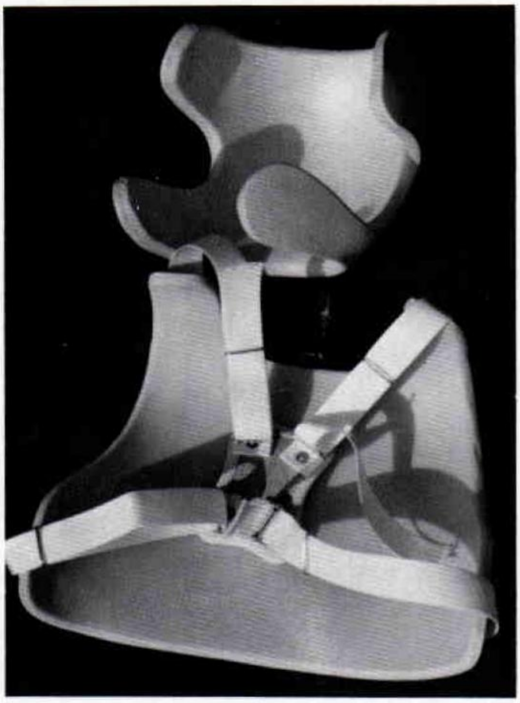

Figure 1. A multi-adjustable cervical orthosis (M.A.C.O.), anterior view.

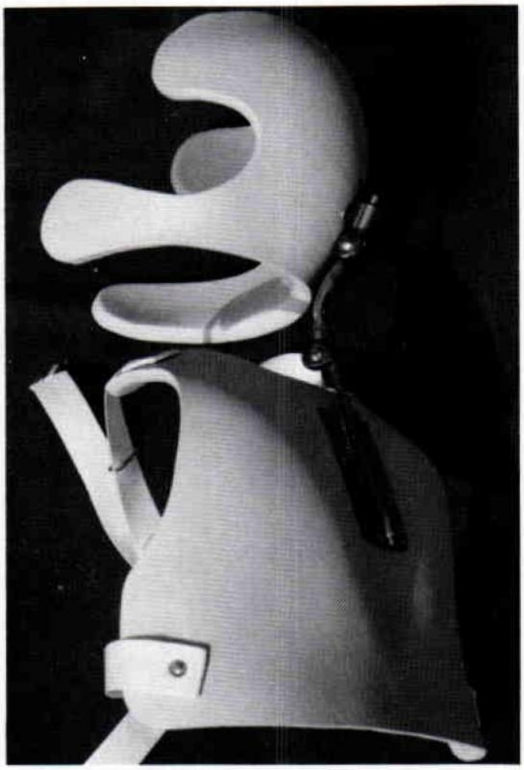

Figure 2. A lateral view of the orthosis.

FABRICATION AND FITTING

The patient was seen in the office one week before surgery was scheduled in order to take a plaster negative impression from which the orthosis would be fabricated. The patient was seated on an examination table and stockinette was placed on her in a two-step procedure. Two separate pieces of stockinette were pulled over her head and down onto her torso, extending down to her waist; slits were cut for the arms to extend through. Two smaller pieces, each sewn at one end, were pulled over her head, extending down to her sternal area, with slits placed up the sides so the smaller pieces would provide a smooth transition to the larger torso pieces. Holes were cut for the eyes and nose. Using an indelible pencil, both ears were outlined on the stockinette. The patient was placed in as close to a neutral position as possible, though with the design of this orthosis, correction during the casting stage is not necessary.

Strips of four-inch wide plaster bandage were cut, and, starting with the torso, were applied in anterior and posterior sections, extending to the crests. Additional strips were applied to the neck and head, covering everything but the face. The edges of the sections were coated with Vaseline® for ease when separating from each other. The strips and stockinette were removed and put together, ready to be poured to form the positive model. The positive model was carefully smoothed, and build-ups were placed in the areas of the scapulae.

For materials, one-eighth inch kydex plastic was used because of its strength and ease with which it can be adjusted with the application of heat. For comfort, ¼ inch plastazote was added for an interface. There is only a posterior plastic section to the brace on the torso with trimlines as shown (Figures 1 and 2). One-inch dacron straps are riveted to the corners of the posterior torso piece. Anteriorly, four one-inch slide buckles are attached to a leather sternal piece, trimmed to prevent impingement on the breasts. This leather piece will act as the attachment point of the dacron straps. This design will keep the orthosis lightweight and less bulky (Figure 3). The headpiece is trimmed to provide as much contact with the head as possible without pressing on the ears. The extensions anteriorly provide contact above the mandible, acting to maintain and control the motion of the head within the plastic (Figure 4).

The key aspect of the orthosis is the use and positioning of the ball joints incorporated into the system. These are two single joints that are riveted and welded together to form one unit (Figure 5). The motion of the two joints together act to more closely approximate normal cervical motion. The inclusion of Allen set screws enable the unit to be locked in any position, in any plane. In order to further enhance the rotational capacity and extendibility of the joints, additional modifications were made. Stainless steel plates are riveted onto the torso and head piece. Onto these plates are welded steel sleeves, into which the ends of the joints slide. The sleeves are drilled and tapped, with Allen set screws added. The finished unit now has motion in all planes, with adjustability at four separate points and the capacity to lengthen or shorten the height of the head piece (Figure 6). The finished orthosis was fit on the patient one week following surgery, with instructions to wear the orthosis 24 hours a day, removing it only for bathing. This was done for three weeks, after which the patient would gradually be weaned from the orthosis.

FOLLOW-UP

The patient was seen for a one-week follow-up. She had been following instructions as to its use and accepted it as part of her treatment. She related that she had only minor problems that were alleviated by minor postural changes. This included sleep time. There were no signs of skin irritation from the plastic or padding and

Figure 3. View showing the orthosis on the patient (anterior view).

Figure 4. A lateral view of the patient wearing the orthosis.

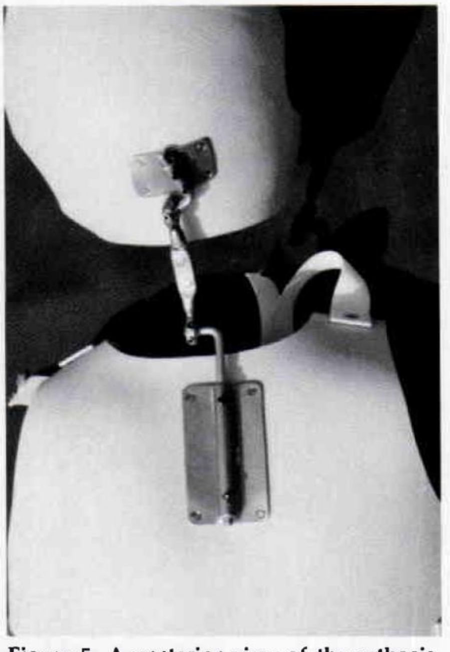

Figure 5. A posterior view of the orthosis. Note the ball joints and length adjustment rods.

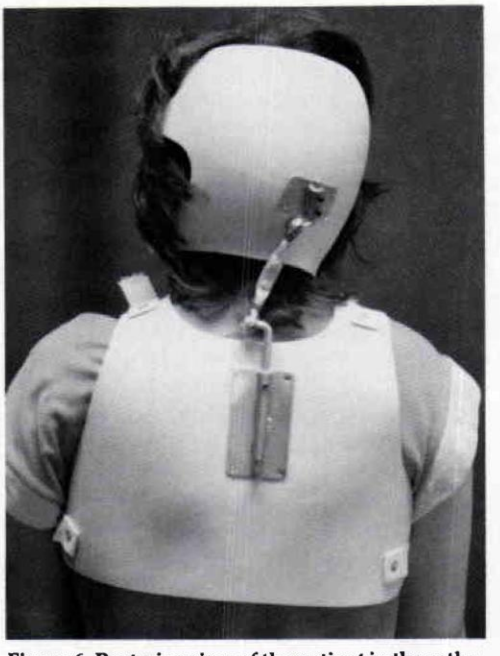

Figure 6. Posterior view of the patient in the orthosis, showing adjustments.

the patient stated the orthosis was comfortable. She had no problem putting the orthosis on and had maintained good position inside the orthosis. A subsequent follow-up two weeks after this visit found the patient still maintaining good position and tolerating the orthosis well. The two month follow-up was equally successful.

CONCLUSION

The problems of cervical control, patient tolerance, and ease of adjustability in a cervical orthosis have been addressed in this study. The cervical orthosis described above was found to offer excellent control of the head in all planes, especially for rotational positioning. Adjustments were done simply and quickly, with good holding power from the Allen set screws. Patient acceptability in this case was good, and comfort was maximized with total contact support and use of padding material.

One major advantage of this orthosis, due to the adjustability of the joint, allowed presurgical casting and fitting while deformity still existed, eliminating a painful casting procedure and delays in orthotic delivery. Correction of the deformity need not be done in the casting stage, thus making presurgical fitting possible, when necessary. The application of this orthosis may prove beneficial in other instances when such cervical control is desired under similar circumstances. We are further evaluating its use in other cervical anomalies.

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[2]SOMI orthosis is manufactured by United States Manufacturing Company in Pasadena, California.

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A Rationale for Treatment of Complete Brachial Plexus Palsy

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INTRODUCTION

Injuries to the brachial plexus present major problems in diagnosis and treatment. Often, associated life threatening injuries initially overshadow a brachial plexus injury. The evaluation and treatment of these patients involve many health care team members. Close cooperation and communication among all parties, including the patient, is essential. Once a definitive diagnosis is made, the majority of the elective treatment involves the prosthetist-orthotist, therapist, and vocational counselor or social worker, in cooperation with the managing orthopaedist.

This paper reviews our existing knowledge of complete brachial plexus palsy and methods of treatment. The elective care described pertains to those complete injuries which present no opportunity for return of normal nerve function. Since few centers treat large numbers of these injuries, this discussion will be based in part upon the data and experiences of other clinicians.

MECHANISMS OF INJURY

The two major causes of brachial plexus palsy are childbirth complications and motor vehicle accidents. During childbirth, downward traction on the shoulder increases the angle between the head and shoulder, resulting in injury to, or avulsion of the upper (C-5 and C-6) roots. Upward traction on the shoulder increases the angle between the arm and lateral thoracic wall, injuring the lower (C-8 and T-1) roots. Complete injuries involving the entire plexus (C-5 to T-1) may occur.

Motor vehicle accidents, especially motorcycle accidents, cause most brachial plexus injuries. Fletcher[1] reported 180 brachial plexus injuries and noted that 81 percent of the patients were under the age of twenty-four years and that 77 percent of the injuries resulted from motorcycle accidents. The injury may result from traction across the arm or across the head. Barnes[3] stated that root tension varies with position of the arm, elevation tenses the lower roots, while adduction tenses the upper roots. When the shoulder is forcibly depressed with the arm by the side, as may occur in a motorcycle accident, the greatest tensile stress falls on the upper roots. During arm abduction and extension, the axillary portion of the plexus, particularly the posterior cord, may be stretched across the head of the humerus as it dislocates anteriorly and inferiorly. When the abducted limb is forced behind the trunk and the head is thrust to the opposite side, tensile stress is exerted on all roots, and the most severe brachial plexus lesion, a complete palsy, may result.

Traction across the head may also be an important mechanism of brachial plexus injuries. In Fletcher's series,[1] most motorcycle accident victims wore helmets and were involved in head-on collisions. The head was forced laterally, away from the shoulder, injuring the plexus.

CLASSIFICATIONS OF INJURY

Brachial plexus injuries may be classified by the roots involved, by division of plexus injured, and by Sunderland's severity of injury to specific nerves.[5] Sunderland's five degrees of injury best correlate with prognosis for recovery of the injured nerve. A first degree injury produces temporary loss of nerve conductivity at the site of injury with loss of motor function and muscle tone, and a reduction in proprioception. First degree injuries recover completely and spontaneously. Second degree injuries involve the fascicles, resulting in complete loss of motor, sensory and sympathetic functions. Axon regeneration proceeds distally from the site of injury with proximally innervated muscles returning first. Third degree injury results in interruption of the internal structure of the fascicles. Regenerating axons are not aligned with appropriate tubules and clinical recovery is never complete. Fourth degree injury results in disruption of all fascicles. Complete loss of motor, sensory and sympathetic function occur and no motor or sensory function return spontaneously. Fifth degree injury is severance of the nerve trunk with loss of motor, sensory and sympathetic function. Although neuromas form, no neurologic recovery is possible.

EVALUATION AND DIAGNOSIS

Early, accurate assessment of the plexus injury is necessary but difficult. It requires various neurological examinations and tests, most important of which is a thorough physical examination. Additional tests, including myelograms, electromyographs and nerve conduction velocities, are helpful, but require experience and interpretative skills. A relatively new technique developed by Dr. Steven Jones in England, the spinogram, involves stimulation of the peripheral nerves at the wrist while recording over the plexus at the root of the neck. This non-invasive test departs from the usual procedure of stimulating proximally and recording distally in order to demonstrate preganglionic or root avulsion injuries.

Pain following brachial plexus injury is a common problem. Pain is often not experienced until two or three weeks after injury, and increases in intensity until it reaches its peak about six weeks post-injury. It may persist at this level for years. The pain may be described as burning, crushing, stabbing or like severe electric shocks. According to Wynn-Parry,[2] the presence of severe burning pain indicates a preganglionic lesion with root avulsion from the spinal cord. Dermatomal pain distribution correlates with the avulsed root.

NON-OPERATIVE TREATMENT

Treatment begins with physical therapy to prevent joint stiffness, prevent soft tissue contractures, and assist in relief of pain. Pain relief is a monumental challenge, and neuromodulation may be beneficial in some patients. Splints are used to prevent joint contractures, or to optimize limb function. These programs require careful reevaluation of the patient at regular intervals to determine changes in muscles and joints. The therapist and orthotist share in encouraging the patient to comply with the prescribed regimen. The exercise program can often be performed by the patient or family at home and only requires checks by the managing team to assess progress.

Careful attention to detail and accurate communication among the team members will clarify goals and alert the physician to a change or lack of progress. Decisions are made by the patient based on a sound understanding of all options, thus the patient becomes the controlling factor in care management.

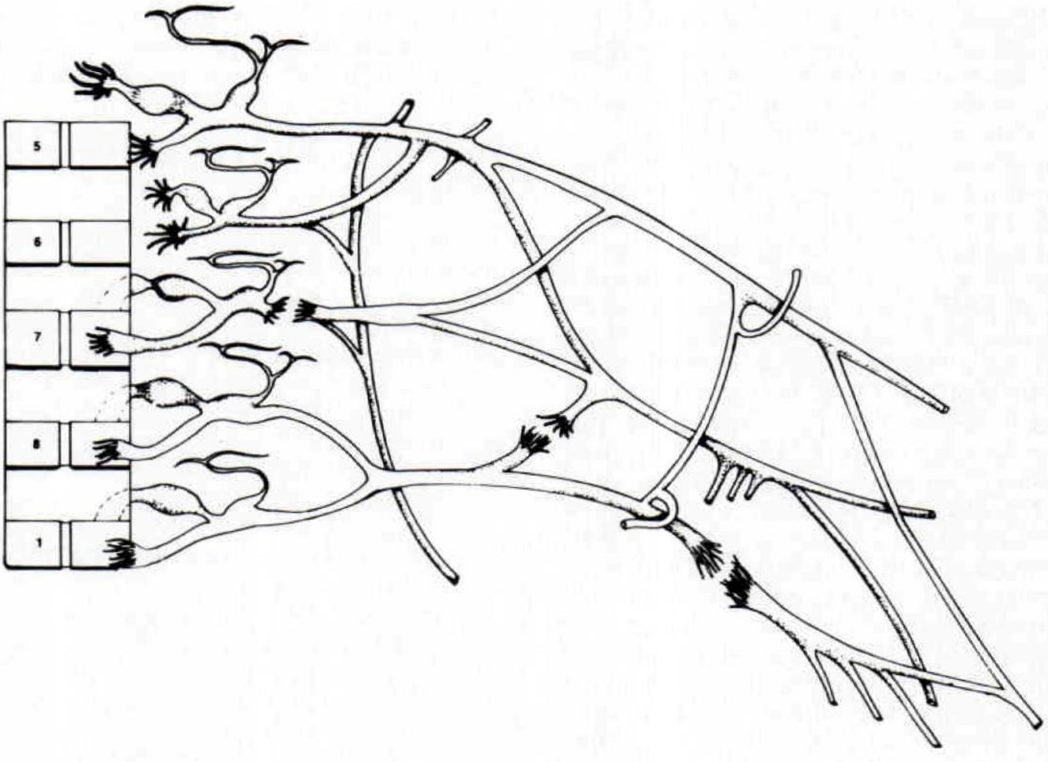

Figure 1. A representation of a complete brachial plexus injury.

OPERATIVE TREATMENT

A complete brachial plexus lesion implies that all parts of the brachial plexus are involved: all five roots, all three trunks, all three cords, or a combination of root, trunk, and cord lesions (Figure 1). If no neurologic combination of recovery has occurred by six weeks after injury, and if physical findings (Horner's sign), paraspinal EMG's, or cervical myelography indicate a preganglionic component to the injury, the prognosis for recovery is poor. The treatment described will concern only the severe, complete injuries.

Given a permanent complete plexus palsy, the pivotal issue is whether the patient would become a successful prosthetic user. Patient sex, age, limb dominance, associated injuries, motivation, experience with mechanical equipment and the support and experience of the medical care team may influence this decision.

A positive relationship may exist between prosthetic use and the amputation of the patient's dominant hand, especially for those who are unable to transfer dominance to the non-dominant hand. Ransford and Hughes[11] state that if the patient is a manual worker, he or she will more likely use a prosthesis if he or she has difficulty converting hand dominance. They reviewed twenty cases at ten years. Thirteen patients were supplied prostheses, seven were dominant and six non-dominant. Only two of the seven were true prosthetic users. Since only two of 20 cases resulted in true prosthesis users, they recommended amputation and prosthetic fitting only if the dominant limb was affected.

The treatment plan is simplified if the patient is clearly not destined to use a prosthesis. No surgery may be indicated. The patient may elect to retain the limb for cosmetic reasons. If the patient is athletically inclined or if the flail nondominant limb is a nuisance, above elbow amputation is an accepted option.[11] It may also be indicated for the dominant limb in the patient who will not be a prosthetic user, who has carefully considered the alternatives and who requests the procedure for convenience (Case 1).

Careful consideration of surgical alternatives (including above elbow amputation and shoulder arthrodesis) is important for the potential prosthesis user (Case 2). Rorabeck[10] stated that amputation and fitting done within one year after injury are more likely to result in successful prosthetic fitting than are alternative approaches. He evaluated nineteen patients, fourteen with above elbow amputation alone, and compared them to five patients with above elbow amputation and shoulder arthrodesis. Only one of the five returned to gainful employment while six of fourteen returned to work. Yeoman and Seddon[10] believe that combined amputation and arthrodesis are the treatment of choice within two years of the injury. They reported on seventeen cases of above elbow amputation and shoulder arthrodesis. They compared their results to either no surgical treatment or to total limb reconstruction, but not to above elbow amputation with early prosthetic fitting. For those using a prosthesis, the average interval between injury and amputation was sixteen months. For those not using a prosthesis, it was three and a half years.

Wynn-Parry[2] reported on fourteen patients who underwent above elbow amputation and arthrodesis within six months of injury. Of these, ten returned to work within one year. Further follow-up revealed that these patients were working without their prosthesis, leading Wynn-Parry to a more conservative attitude towards early amputation.

Ransford and Hughes[11] felt that shoulder arthrodesis was necessary for the true prosthesis user. Because the true prosthetic user is rarely seen in clinical practice they recommended the procedure cautiously. They noted that arthrodesis of the shoulder produces potential for skin irritation over bony prominences, but that the procedure resolved the problem of humeral head subluxation. Prosthesis fitting must be delayed until after fusion has occurred.

PROSTHETIC FITTING TIME

The elapsed time between elective amputation and initial prosthetic fitting is important. Burkhalter[13] believes that early or immediate fitting does not adversely affect wound healing and helps maintain the two handed pattern for activities of daily living. However, only three of the eighty-seven patients in his series had brachial plexus injuries and none of these were using a prosthesis at follow-up. These data suggest that the patient with a brachial plexus injury may differ from other amputees treated similarly.

Rorabeck[10] states that amputation and fitting should be done within one year of injury, suggesting that the two parts together play an integral role in successful prosthesis wearing and return to work.

Leal and Malone[14] report that myoelectric fitting decreases rehabilitation time when compared with conventional immediate fitting. This suggests the important factor is prosthetic control. Patients who used to support this conclusion were all working prior to injury and returned to work after fitting. However, no job description or dominant hand data were reported. Additional follow up is needed to clarify long term results.

PATIENT SATISFACTION

Perhaps the most interesting and perplexing data reported deals with patient satisfaction. Fletcher[1] reported on seventy-three patients contacted by questionnaire after one year post-amputation. Ninety-one percent reported wearing the prosthesis regularly at work, and all were glad they chose to have the arm amputated. We are not told what procedures each patient underwent.

Brewerton and Daniels[12] reported that at one year post-injury only 16 percent of the patients recalled talking with their managing physician about long-term options and outcomes. They emphasized the existence of this void in the care of the brachial plexus injured patient.

CASE REPORTS

Case 1:

A 27-year-old male employed as an unskilled laborer suffered a complete avulsion of his nondominant limb brachial plexus. No spontaneous recovery occurred in six months. When offered amputation with early prosthetic fitting, he replied "I wouldn't use it if I had one." The patient later requested amputation since the arm was "always in the way." An uncomplicated above elbow amputation was completed eighteen months after the patient's accident. Since no prosthetic fitting was planned, shoulder arthrodesis was not performed. He described mild pain prior to and unchanged since the operation.

Since the accident he has remained unemployed, is now divorced and is currently residing with his parents. He has adapted to one handed activities of daily living. He believes that his care was satisfactory.

Case 2:

A 34-year-old male semiskilled service station attendant sustained a complete brachial plexus avulsion. He suffered a C5 through T1 preganglionic injury to his dominant limb. He was advised of his poor prognosis. No recovery had occurred within fourteen months. He felt the arm was a nuisance and requested amputation, but sincerely wanted a prosthesis to aid him in his hobbies and with his wheel repair business. An above elbow amputation was completed fourteen months following his injury. A shoulder fusion was not performed, allowing early prosthetic fitting. He was fitted with a conventional above elbow body-powered system. He is able to control the elbow position and the terminal device. He uses both a stainless steel terminal device and an Otto Bock cosmetic hand.

Since his injury, the patient has changed extremity dominance and can eat, write and work in his shop. He stated that he was never athletic but enjoyed fishing and fishing reel repair. The patient is satisfied with his treatment, and continues to gain skills with his above elbow prosthesis.

CONCLUSIONS

Brachial plexus trauma results in a spectrum of palsies, including the severe, complete plexus palsy. When this injury includes a preganglionic component, the prognosis for recovery is poor. Early, accurate diagnosis is critical to planning treatment and counseling the patient. Nonoperative treatment of the complete brachial plexus palsy, under the supervision of the physical therapist, includes neuromodulation for pain control and prevention of joint contractures. Operative treatment includes above elbow amputation in the nonprosthetic user. For the potential prosthesis user, above elbow amputation and/or shoulder arthrodesis may facilitate prosthetic fitting and use.

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Cerebral Palsy: A Complex Brain Disorder

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Cerebral palsy is a relatively common brain disorder found worldwide which occurs shortly before, during, or shortly after birth. The condition presents itself early in childhood, usually before three years of age, and involves voluntary muscle dysfunction. The degree of physical complication affecting each cerebral palsy child determines the courses of treatment initiated.

Separate physical complications are attributed to each different type of cerebral palsy. The three different groups of cerebral palsy patients, the antenatal, the natal, and the postnatal, and the recommended treatments will be discussed in this paper. The sub-classes involved within each of these groups consist of the spastic, athetoid, and the ataxic cerebral palsy child.

Due to the complexity of cerebral palsy, its treatment, and the varying degrees of severity, the psychological complications for the cerebral palsy patient and his family must also be noted.

STATISTICS AND CAUSES

There appears to be some disparity in the statistics reviewed on the occurence of cerebral palsy. The United Cerebral Palsy Association estimates that there are 600,000 people afflicted with this condition in the United States. Approximately one baby in every 170 newborns has cerebral palsy.[1] The National Paraplegic Foundation estimates that there are 750,000 individuals with cerebral palsy in the United States.[2] It has also been estimated that every year for each 100,000 in population, six children will be born with cerebral palsy.[3]

This figure disparity is obviously significant, however, even the lowest number estimated is sufficiently high to suggest that there is a large minority of cerebral palsy patients both in the United States and around the world.

Within this large group there exists three major classes of cerebral palsy patients. The antenatal (before birth) group consists primarily of the congenital type of cerebral palsy. Some of the causes of this disorder range from infectious disease, syphillis, inherited metabolic defects, and toxemia during pregnancy. The natal (during birth) group is the most frequent type of cerebral palsy, involving trauma at birth, hemorrhage, anoxia, heavy sedation of the mother, or hypoxia secondary to winding of the umbilical cord around the baby's neck. The postnatal group is from infectious and traumatic lesions such as encephalitis, meningitis, vascular accidents, and Rh incompatability occurring shortly after the birth.

Within these three separate categories of cerebral palsy are sub-classes of muscular dysfunction and conditions described as spastic cerebral palsy, the athetoid cerebral palsy, and the ataxic cerebral palsy.

SPASTIC CEREBRAL PALSY

In the spastic group of cerebral palsy children the most common symptoms are an awkward, irregular type of gait. Most of these victims are hemiplegic wherein the cerebral palsy affects one side of the body involving both limbs. Other segments of the body may be afflicted, but are less common: monoplegia (one limb), diplegia (both lower limbs), and quadraplegia (all four limbs).

Of all the different types and complexities of cerebral palsy, approximately 60 percent of the cases fall into the spastic category. This lack of muscular control and balance makes the cerebral palsy individual stand out in a crowd. Therefore, sometimes orthotic assistance is sought to correct unsightly posture and gait.[4]

ATHETOID CEREBRAL PALSY

It is estimated that 20 percent of cerebral palsy patients are in this category. The characteristics of athetoid cerebral palsy are uncontrollable movement of the face and all four limbs. There is also difficulty with speech and swallowing. The individual usually attempts voluntary movement and cannot, which brings about emotional tension. This tension, however, is usually absent during sleep.

ATAXIC CEREBRAL PALSY

In the ataxic cerebral palsy patient, there is no spasticity, nor is there athetosis. The gait is unsteady and the appearance of falling is very noticeable. Basically, there is a disturbance of muscular coordination since the brain lesion is primarily cerebellar. The child's intelligence is usually not affected in these cases.

The complexities of cerebral palsy and its occurences are variable and the integration of the different types, sub-classes, and it causes are immeasurable. There are subdivisions within these major divisions of cerebral palsy which are not elaborated upon in this paper. These groups are "characterized by outstanding clinical manifestations," all dependent upon the location of the brain lesions.[5]

These physical complications all become part of the medical team treatment picture as the child progresses.

TREATMENT AND PSYCHOLOGICAL CONSIDERATIONS

Initially the cerebral palsy baby may appear normal and show no outward signs of disability. But usually by the age of three months some symptoms will begin to appear. The first step in any treatment situation should be the setting of realistic goals based upon the individual potential of the cerebral palsy child.

Since the lesion involvement can be mild to severe, the recommendations of the family physician usually consist of a multiple team treatment approach. The management of the cerebral palsy child is a family affair, but medical team members such as the rehabilitation physician, surgeon, therapists, psychologists, and orthotists play a major role as well. These professional people must have compassion, understanding, and a caring attitude in conjunction with realistic goals.

The mentality of approximately 75 percent of these individuals with cerebral palsy is well below average. Most of the mentally deficient cerebral palsy children are in the spastic quadraplegic group. Thus, knowledge of these immediate limitations helps the professional to assist the family in dealing with the non-institutionalized as well as the institutionalized cerebral palsy child during the early years.

A small percentage of the cerebral palsy children are able to function at home and school with somewhat minimal dysfunction. However, a larger percentage of these children require substantial assistance physically, emotionally, and psychologically. Again, the degree of muscular dysfunction depends upon the amount and severity of brain lesions. "No two patients have identical symptoms."[1] This complexity suggests careful psychological considerations.

As mentioned earlier, the parents of the cerebral palsy child are seldom aware of the problem with their child because the disorder can sometimes go undetected for several months.

Therefore, the parents, as well as the cerebral palsy child, need special psychological attention. They experience disappointment, guilt, and anxiety when the first signs of the brain disorder appear.

Some parents have great difficulty accepting the reality that their child will never be "normal." This is when the team treatment approach becomes essential.

The psychological needs of the cerebral palsy child are assessed depending upon the age of the child and the potential of mental development. Of course, the less severe the palsy is, the greater the chance for social competence and interaction.

Because the cerebral palsy individual's disability is visible, we frequently assume that they also suffer from some other unseen defects such as mental retardation, deafness, etc.[6]

However, this is an arguable fallacy in most instances of disabilities. Certain types of cerebral palsy children suffer a mental deficiency, yet often times they are comprehensive, teachable, and mentally functioning individuals with a physical disability.

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"Zero-Position" Functional Shoulder Orthosis for Postoperative Management of Rotator Cuff Injuries

Jiro Ozaki, M.D.
Ichiro Kawamura

INTRODUCTION

Many shoulder orthoses such as the airplane splint, the abduction splint, and the Velpeau bandage have been widely used for injuries. These orthoses, however, sometimes cause shoulder contracture and muscle imbalance. The shoulder joint possesses the widest range of motion and the most varied movements of any joint in the human body. The "scapular plane" has been accepted as the reference plane for the mechanism of the shoulder joint[2] (Figure 1). Codman (1934)[1] pointed out a very natural position for the human arm when the body is recumbent, and called this position a subordinate pivotal position. In this position, the axis of the humerus is in line with the axis of the spine of the scapula, and the head and the neck of the humerus is in the same plane (Figure 2). Saha (1961)[5] has designated this point as the "zero-position," because muscular rotatory forces acting upon the humerus at this position are almost zero. Ozaki (1980)[4] performed cineradiographic and radiographic studies, and concluded that the scapular plane should be inclined forward at an angle of 30 to 45 degrees to the frontal plane, and that in the zero-position the humerus must be elevated to 150 degrees in the "scapular plane," with individual variations. On the basis of these biomechanical concepts, we designed our zero-position functional shoulder orthosis for the postoperative management of rotator cuff injuries.

FABRICATION PROCEDURE

The orthosis consists of a pelvic support, a thoracic support, an upright bar, and an arm support (Figure 3). The pelvic girdle is made of 4mm subortholene plastic sheet. The iliac crest is the most important point of this orthosis—if the upright support fails to hold the shoulder in position after it is fitted to the patient, it will result in serious problems for the shoulder joint. The length of the upright bar can be changed because the distance from axilla to iliac crest varies according to the angle of elevation and depression of the arm. The up-

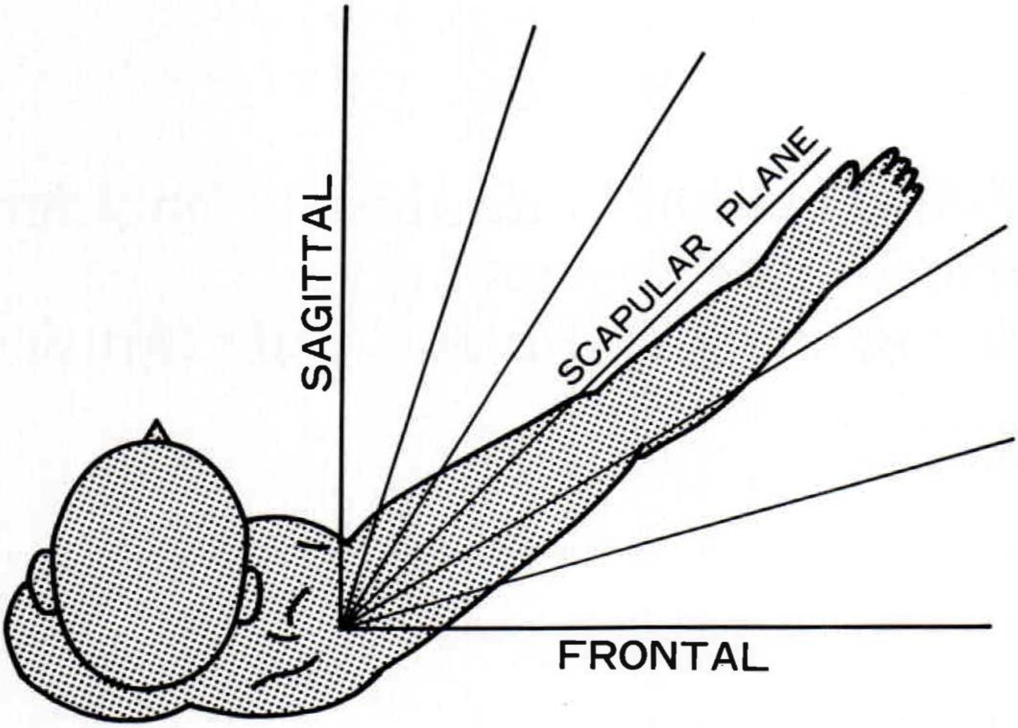

Figure 1. The scapular plane.

right bar can be rotated on a horizontal plane and fixed in any position. The elevation angle of the shoulder joint can be changed by a mechanical joint and it can be fixed in any position. The distance of axilla to elbow joint changes when the shoulder joint is lowered so it also has to be adjustable. The angle of internal and external rotation of the humeral axis is very important for the zero-position. For a relaxed position, the shoulder has to take a neutral position; therefore, we have provided a rotation mechanism. The angle of the elbow joint is also important for a relaxed position. It should be flexed slightly. At this joint, the angle of flexion can be changed and fixed in any position and the lower arm support can be removed when necessary for exercising the forearm.

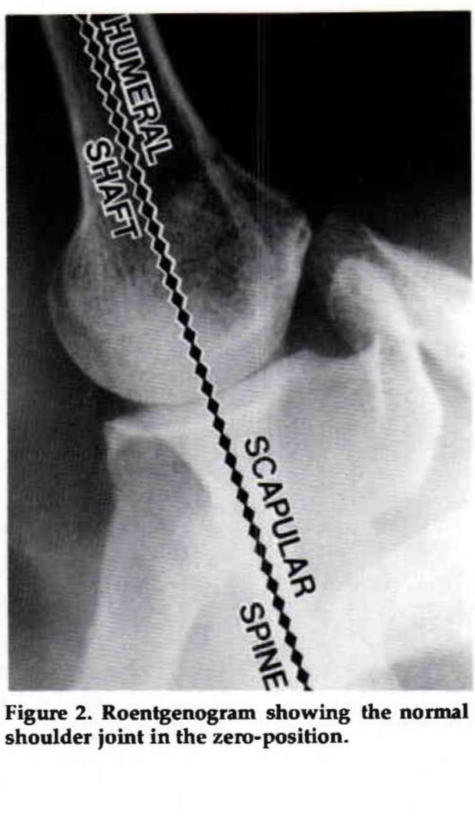

Figure 2. Roentgenogram showing the normal shoulder joint in the zero-position.

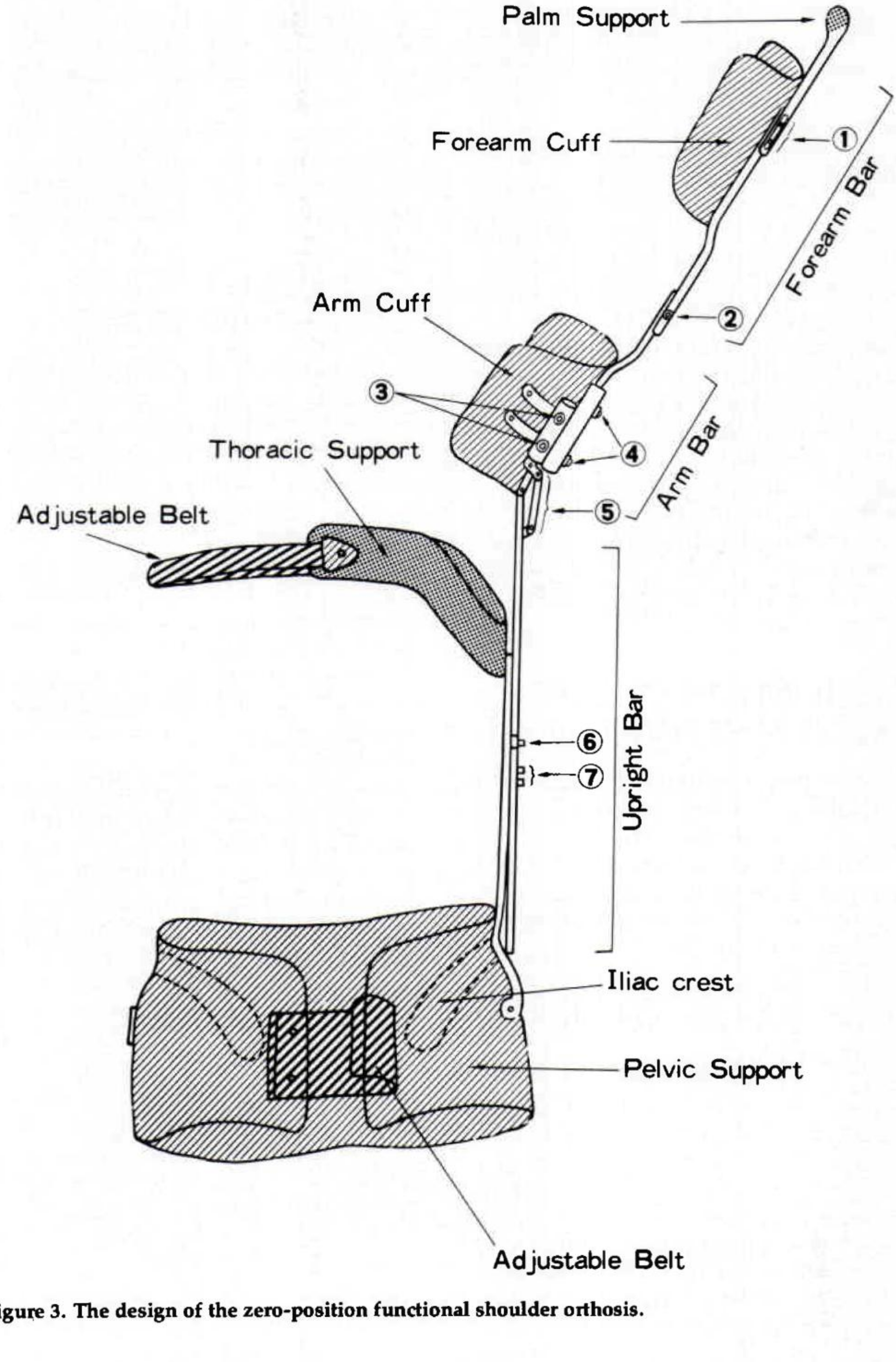

Figure 3. The design of the zero-position functional shoulder orthosis.

THE CASTING PROCEDURE

The plaster negative should be made before the operation because the orthosis must be fitted to the patient after only a few days and it is difficult to make the cast after the operation.

The casting should be made from the metacarpal phalangeal joint of the hand to the pelvic girdle in the zero-position. As we pointed out, the most important point of this orthosis is the iliac crest, so be careful to create the proper shape for this area in making the plaster negative.

The arm should be elevated to 150 degrees and flexed forward 30 to 45 degrees from the frontal plane. The palm of the hand should face inwards to give neutral rotation of the humerus, and the elbow joint should be slightly flexed to give a more relaxed position of the shoulder joint.

If it is impossible to make the plaster negative for some reason, fit the patient using the Boston Brace™ body jacket or something similar.

MODIFICATION OF THE PLASTER POSITIVE MODEL

For convenience of modification, provide marking at the base of the axilla and cut open the trunk part and the upper limb. The basis for ensuring zero-position is a well fitted pelvic girdle, so it must be deeply cut away in the waistline as with a Milwaukee Brace girdle.

PLASTIC LAMINATION AND FITTING

The pelvic girdle and each cuff for the upper limb are made from 4mm subortholene plastic sheet. The distal trimming line of the pelvic girdle should extend low enough posteriorly to just clear the chair when the patient sits. Proximally, it should not touch the bottom edge of the 12th ribs. The anterior distal edge of the pelvic girdle must cover the anterior superior iliac spines laterally, but medially only to the top of the symphysis pubis.

The upper arm and forearm cuffs should be trimmed so that the shoulder joint can be adducted completely and the elbow joint flexion angle is not disturbed. Fix the 5/8″ upright to the pelvic girdle and the thoracic metal band. Weld a 1/2″ diameter metal tube to the bar. Assemble each cuff for the upper limb and the shoulder joint, and then connect them to the tube. Cut off any unnecessary material as far as possible.

CLINICAL CONSIDERATIONS

Concerning shoulder immobilization after the operation for rotator cuff injuries, several methods such as Plaster of Paris spicas and the Velpeau bandages have been used, but they sometimes cause contracture of the shoulder joint. The use of the zero-position as the position for postoperative immobilization of rotator cuff injuries is not a common practice, but it is particularly helpful in the postoperative program for regaining functional movement.[3] As the zero-position tends to be a point of convergence for elevation in the scapular plane, the patient who works for greater range of motion from the zero-position is able to initiate movement readily in any plane of elevation. Furthermore, since the deltoid, the supraspinatus, and the infraspinatus are relaxed, the zero-position is the most favorable position to encourage physiological repair of lesions in and about the greater tuberosity. Also, the scapular plane is the reference plane for the movements of flexion and extension, abduction and adduction at the humeroscapular joint, and in this plane the rotator cuff is not subjected to a twisting strain. These are the reasons why we have used our zero-position functional shoulder orthosis for postoperative management of rotator cuff injuries.

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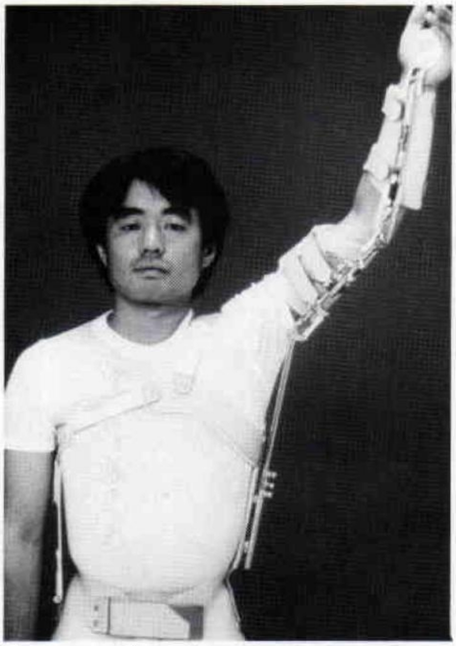

Figure 4-A. The orthosis is applied to maintain the zero-position. An anterior view.

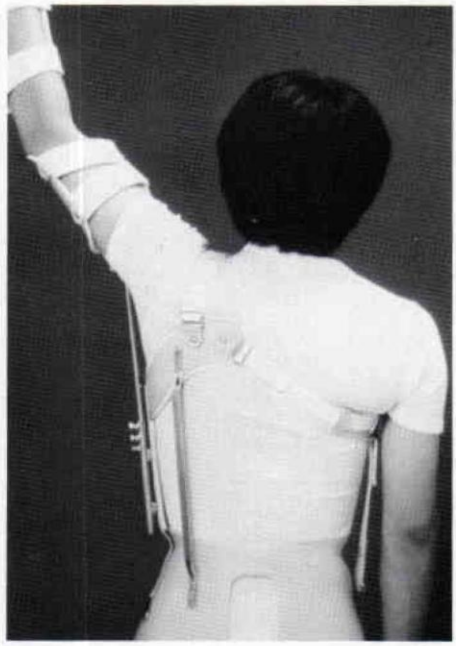

Figure 4-B. A posterior view.

POSTOPERATIVE MANAGEMENT OF ROTATOR CUFF INJURIES

Immediately after the successful repair of the rotator cuff injury, the zero-position of the shoulder should be maintained by skin traction while the patient rests in bed. After three days, the functional shoulder orthosis, which has been made to order preoperatively, is applied to maintain the zero-position on the scapular plane (Figures 4-A, 4-B). At the beginning of the third postoperative week, the upper limb in the orthosis is extended at 100 degrees abduction on the scapular plane and the patient is allowed to start gradual actively-assisted abduction exercise of the arm.

From the fourth to the sixth week, when the patient is able to perform active elevation in the range of 60 degrees to 150 degrees, the abduction angle of the orthosis can be decreased gradually to 30 degrees. Mass movement exercise involving circular motion is indicated. Two to three months after surgery, the orthosis is removed. At this stage the patient is able to use a full range of elevation, and at three to six months, the patient has made maximum recovery.

For the postoperative management of rotator cuff injuries, we have fitted this orthosis on more than 75 patients. As a result of being able to gradually decrease the elevation angle from the zero-position, it was recognized that pain which had occurred in the patients of the plaster cast immobilization group in the zero-position was reduced. Therefore, early healing and excellent results were achieved in nearly all cases.

CONCLUSION

On the basis of the biomechanical concepts of the zero-position and the scapular plane, the authors have designed this zero-position functional shoulder orthosis. It has been successfully fitted to more than 75 patients for the postoperative management of rotator cuff tears. Clinical trials of this orthosis can be extended to the postoperative management of other shoulder conditions, but in case of recurrent shoulder dislocation, posttraumatic shoulder dislocation, and inferior and multidirectional instability of the shoulder, it should not be used to immobilize at the zero-position, because glenohumeral dislocation may frequently be encountered due to glenohumeral instability such as Bankart, or Hill-Sacks lesions, and glenoid dysplasia.

NOTES

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Technical Note:

A New Concept in the Fabrication of Double Upright Orthoses

Karl W. Bremer, Jr., C.O.

INTRODUCTION

The concept of having the uprights for a double upright knee ankle foot orthosis (K.A.F.O.) formed using templates and a pneumatic pressure system has intrigued the author for a number of years. Many ideas have been pursued in an attempt to develop a system that was economical, efficient, easy to use, in need of little maintenance, took up a relatively small amount of space, and could be set up on a standard work bench.

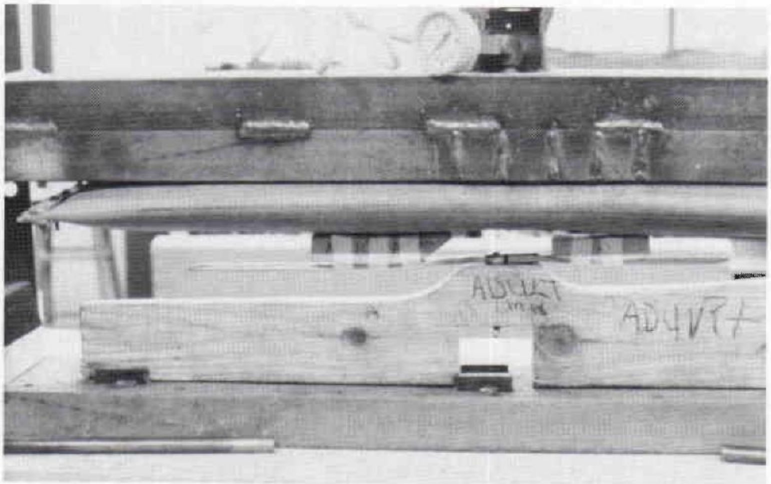
Using the pneumatic forming system in conjunction with carbon fiber composite material, which is used for the bands of ankle foot orthoses (A.F.O.) and K.A.F.O.'s, has made it possible to achieve substantial savings in time. A single K.A.F.O. can be produced in two to 2½ hours time, which is approximately one-third the time otherwise necessary. The orthotist or orthotic technician, requiring less skill, is capable of producing three or four double upright K.A.F.O.'s a day.

THE PNEUMATIC PRESS

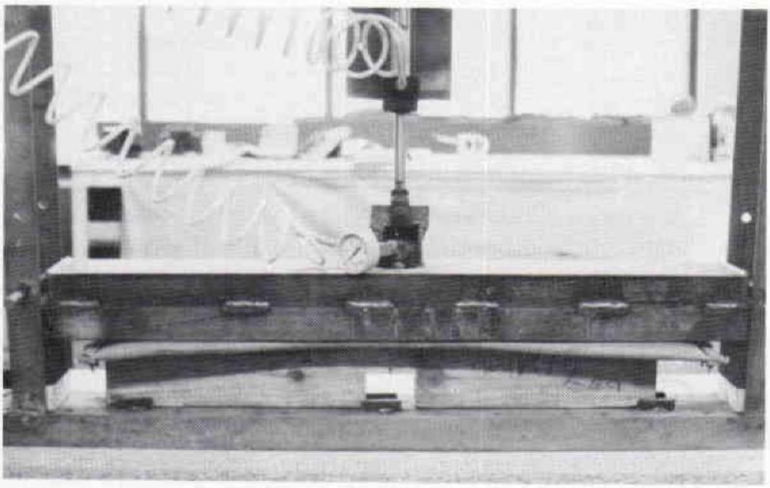
The Press* consists of an air bag mounted in a steel support frame. The bag (a section of fire hose) is six inches in diameter and 36 inches long, sealed at both ends, and connected to the shop air system by a nozzle in the middle. At 100 p.s.i. (pounds per square inch), 4,320 pounds of pressure is exerted in the expanded state. This is sufficient for bending stainless steel uprights up to 3/16" by 3/4" in size. Aluminum uprights are annealed for ease of forming and to lessen the chance of breakage. For annealing, the uprights are heated with a torch and the temperature checked, using white soap, which turns a dark brown, or wood, which skates across the surface like soap, and then quenched in cold water for faster cooling. Air cooling also works, and of course 2024T3 grade aluminum alloy regains most of its hardened properties in 24 hours.

A series of ten templates ranging from infant to adult sizes are used to form the uprights (Figure 1). In 80 percent of the cases, the one standard adult template is used. Use of a template not only speeds the process, but also diminishes marring of the surface and rotation about the long axis of the upright. For any given patient, 90 percent of the required bends can be formed using a standard template. The remaining ten percent of the contouring is done by hand (Figures 2–7).

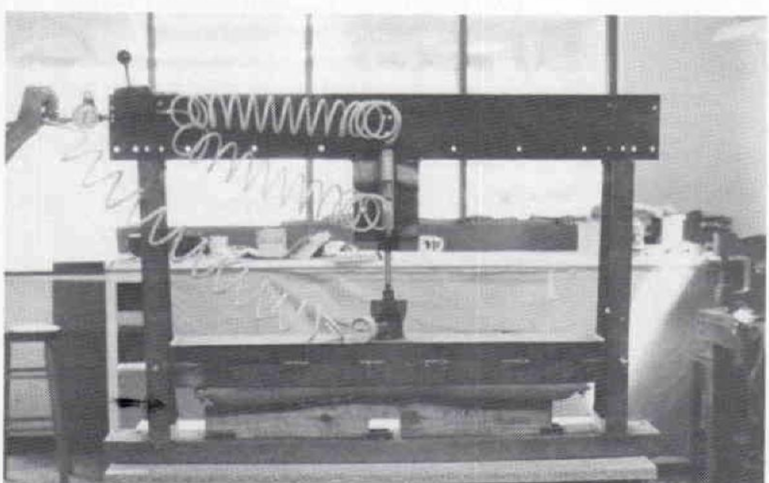
The benefits of using the pneumatic press become evident when forming bilateral K.A.F.O.'s with double uprights, providing the lower extremity tracings have a standard non-atrophied shape. In conjunction with the use of the carbon fiber material, a substantial time savings is achieved. This reduces the time required for completing the orthosis to a total of approximately four and one half to five hours. The four uprights are pressed and shaped within a 20 to 25 minute period.



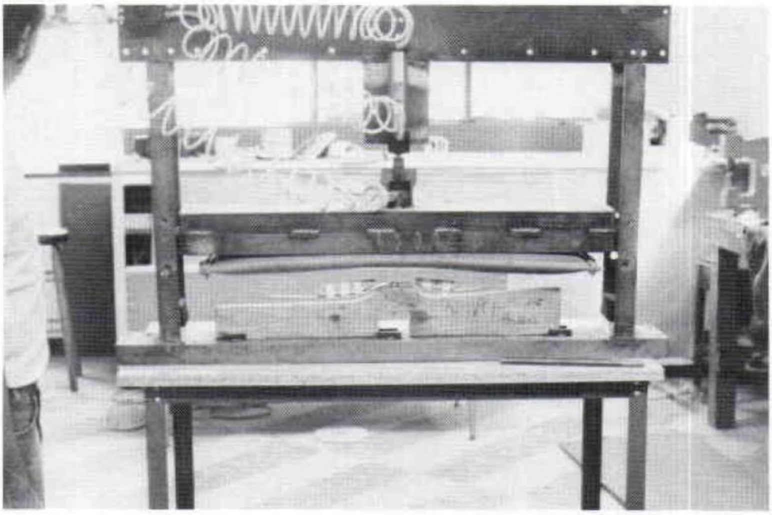
Step 1. In this case, the standard adult template is used with the aid of a pair of compression blocks. These are used when more pronounced bends are necessary. Upright has been annealed and bolted to the template.



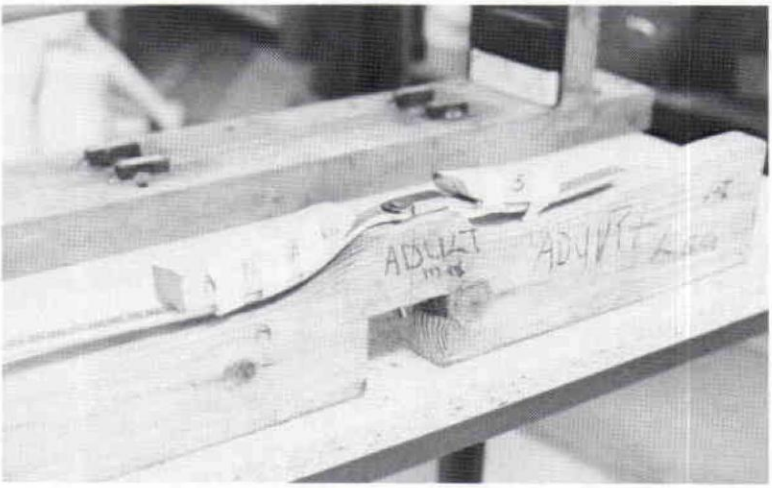
Step 2. The pneumatic bag is lowered into position and the steel safety pins are inserted.



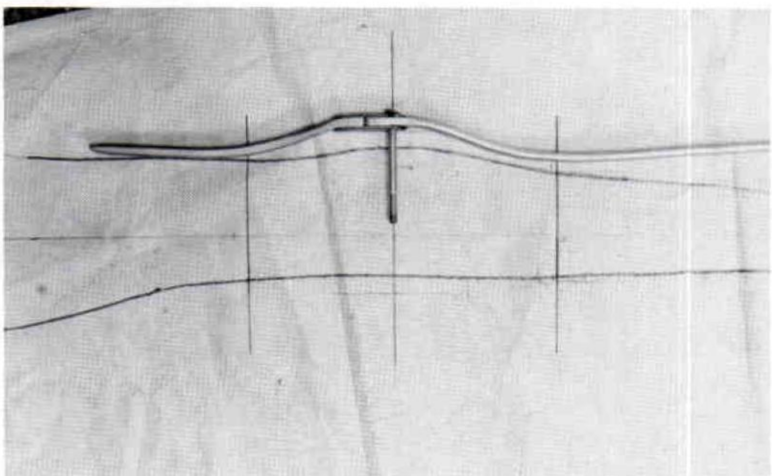
Step 3. Air is forced into the air bag at 80 P.S.I. (Notice expansion of the bag).



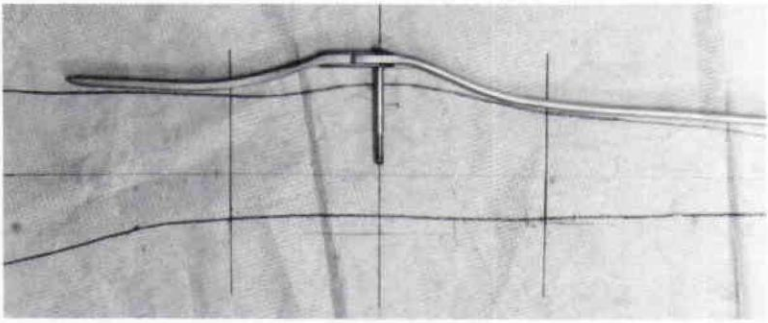
Step 4. Air bag is raised.



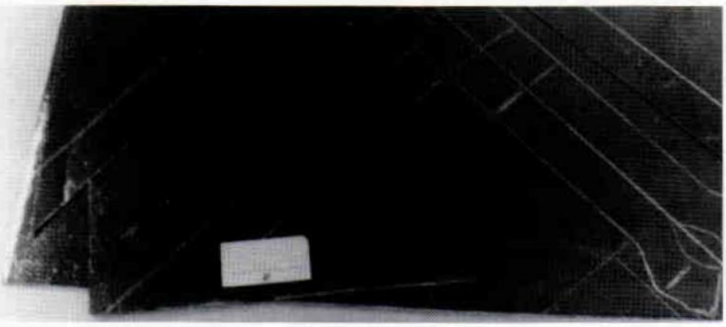
Step 5. The template is removed from the press.



Step 6. Upright is removed from the template and checked against the tracing. Distal bend at calf band will be done by hand.



Step 7. The upright has been corrected and checked against the tracing. Time from Step 1 to Step 7 is 10 minutes.



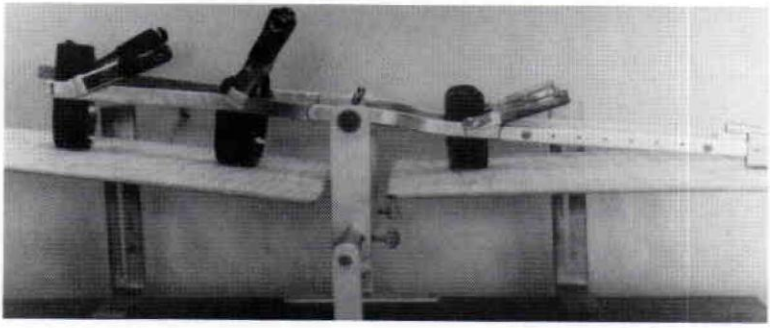
Step 8. Bands are predrawn on carbon fiber to minimize waste. An average of 20% of the carbon fiber material is wasted.

SUBSTITUTION OF CARBON FIBER COMPOSITE BANDS FOR ALUMINUM BANDS

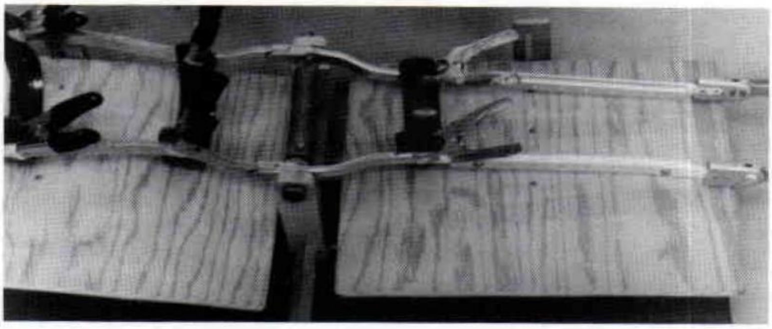
The use of carbon fiber composite is the result of several years of experience looking for a substitute for aluminum bands. Initially, 1/8" nyloplex was used. Nyloplex is easy to shape, and has beneficial rigidity and memory properties. However, we found that it tended to crack when riveted with metal rivets and when subjected to prolonged stress. As a substitute we tried 1/8" and 3/16" polyvinyl chloride. It is not brittle like nyloplex and otherwise is easy to shape and has good memory. However, its lack of rigidity rendered it unacceptable.

Experience with carbon fiber composite inserts, to reinforce the ankle sections of L.S.U. Reciprocating Gait Orthoses, led us to use carbon fiber composite material[†] in sheet form (Figure 8). To date, this has proven to be the material of choice. Bands ranging in width from 1¼"-1½", depending upon patient size, are laid out on a sheet of the composite material, so as to minimize waste. The bands are cut out using a bandsaw with a skip tooth blade. The two-ply carbon sheet is adequate for small to medium size orthoses on children up to age 12. On larger patients the three-ply sheet is recommended. To minimize the possibility of wrinkles developing when the bands are formed, the bands should be cut out on a bias so the fibers are at a 45° angle relative to the direction of bend. This will also cause the edges to radiate outward from the patient and appreciably increase the resistance to torque. So far, only one instance of breakage has occurred amongst 150 bands. In this instance, the two, two-ply bands on a K.A.F.O. failed. These were changed to a pair of three-ply bands without further incident.

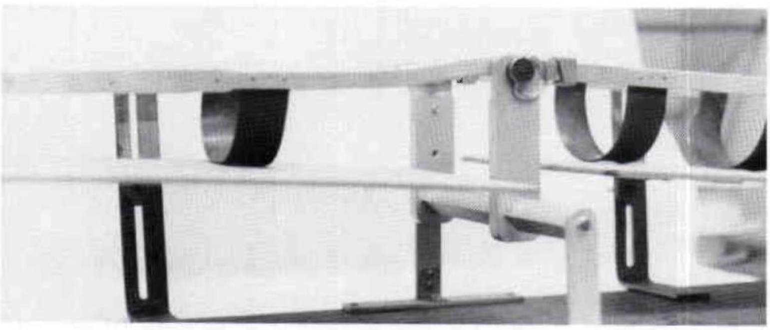
To form the bands, a special device is used. The properly contoured uprights are attached to the extensions and stirrup. They are then mounted in the knee joint alignment fixture which has been set to the proper medial-lateral diameter as dictated by the patient's measurements (Figures 9–11). The depth and tilt of the thigh and calf bands are established by setting the



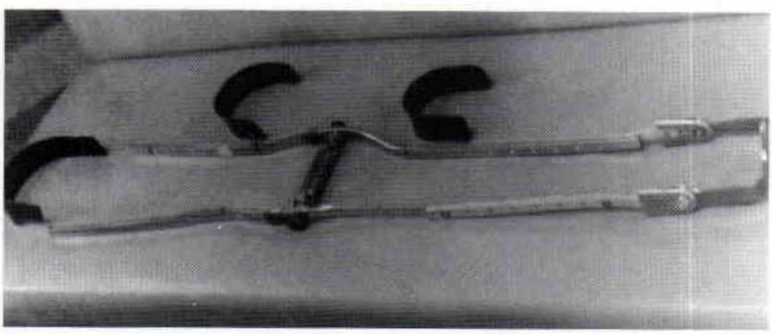
Step 9. Uprights are transferred to the K.A.F.O. jig. Bands are cut out, heated with a heat gun, and clamped into position. The depth of the bands is predetermined with the two horizontal tables.



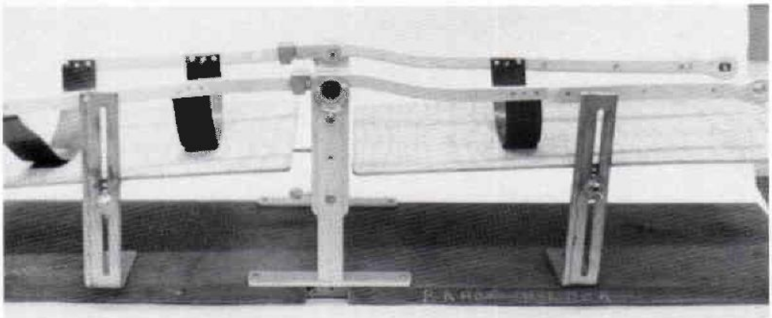
Step 10. Top view. Bands will be removed, trimmed, and attached to uprights. Orthosis is then ready for leather. Time: 2 hours.



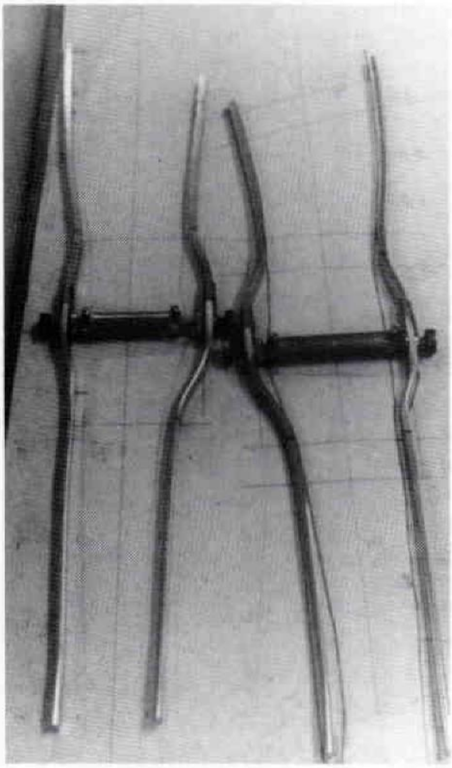
Step 11. Close-up view.



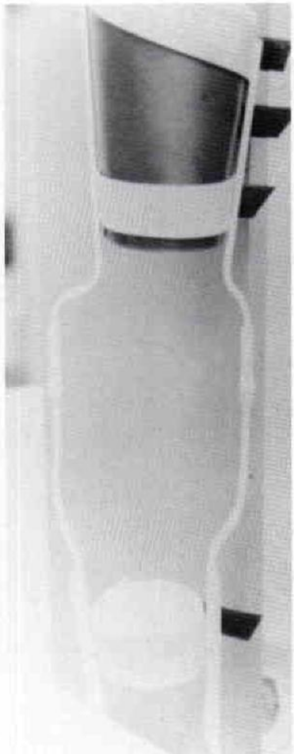
Step 12. Bands finished and ready for attachment.



Step 13. Bands finished and riveted to the uprights. If not satisfied with the shape of the bands, adjustments can be made by re-heating.



Step 14. Uprights for bilateral K.A.F.O.'s that have been pressed out. Time: 30 minutes at this step. This includes annealing, pressing, and assembling.



Step 15. Carbon fiber bands on finished K.A.F.O. patient has severe damage to the knee joint.

Step 16. Completed child's double upright K.A.F.O. with carbon fiber bands.

two horizontal tables. The bands are heated, secured in place with spring loaded clamps, and allowed to cool for 2½ minutes (Figures 12, 13).

COST AND TIME COMPARISONS

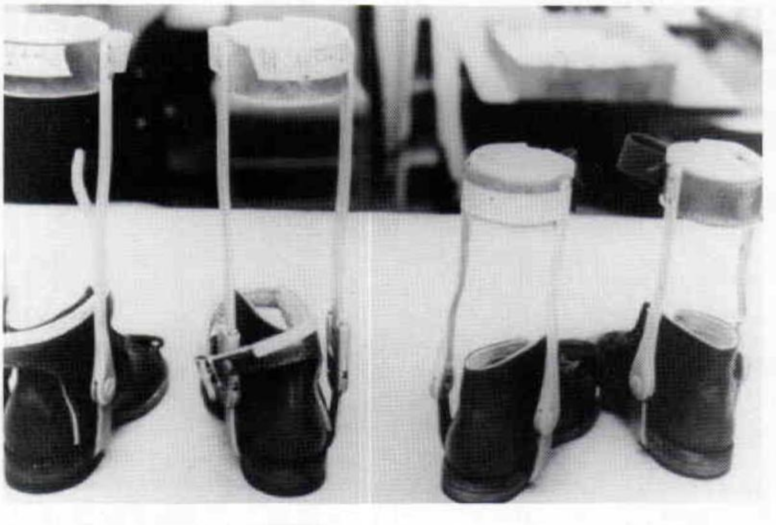
A carbon fiber band costs about $7.50. Five minutes or less is needed to put the three bands in place, ten minutes is needed to finish and attach the bands, for a total of 15 minutes.

An aluminum band costs about $2.50 per band. It takes about 15 to 20 minutes to put each band in place or about one hour to position and attach all three.

It takes approximately two to two and one half hours to fabricate a K.A.F.O. using the pneumatic press and carbon fiber composite bands, versus six hours to fabricate a K.A.F.O. using the conventional technique and materials.

Step. 17. (above) Carbon fiber bands on genu valgum patient. Single upright K.A.F.O.'s with aluminum pelvic band.

Step 18. (right) Use of carbon fiber composite bands in double upright A.F.O.'s.



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*Available from Bremer Brace Company, Inc., 1724 South Orange Avenue, Orlando, Florida 32806.

†Available from Durr-Fillauer Medical, Inc., Orthopedic Division, 2710 Amnicola Highway, Chattanooga, Tennessee 37406.

REVIEWS

by Charles H. Pritham, C.P.O.

A Guide to Controls; Selection, Mounting, Applications, Rehabilitation Engineering Center, Children's Hospital at Stanford, 520 Willow Road, Palo Alto, California 94304. 144 pages, 1982.

The challenge of providing the severely disabled with mobility, communication, and control of their environment is one that has gained increased attention in the last ten years. Great potential in meeting this challenge is afforded by the growing availability of such devices as powered wheelchairs, personal computers and environmental control systems. One of the most daunting aspects of the matter, however, has been the business of providing an individual with control of the various devices intended for his assistance. As its title suggests, this publication is intended as a guide to the subject.

The book is assembled in a loose leaf format so that it can be readily updated and expanded. It is essentially a survey of commercially available controls with suggestions as to their application and mounting. Suggestions are given for performing a systematic evaluation of an individual's needs and capabilities. The book does not purport to be all inclusive and purposely excludes most of the very sophisticated and specialized devices at the far end of the spectrum. It does include, however, extensive lists of references and reference and information centers to be contacted for further information. While it may seem simplistic to some, it should prove to be of considerable assistance to those who are confronted with the problem of providing even the simplest function without backup or resources.

Splinting in Hand Therapy, Erik Moberg, M.D., Ph.D., Carl-Goran Hagert, M.D., Ph.D., Ulla Nordenskiold, Monica Traneus, and Brigit Svens. Thieme-Stratton, Inc., 381 Park Avenue South, New York, New York 10016. 88 pages and index, softcover, Feb. 1984, $19.50.

This is the English edition of a book originally published in Swedish and subsequently in German. The authors describe it as being intended specifically for the personnel—surgeon and therapist—working in a small unit remote from large specialized centers. As such, and given its length, it might best be characterized as presenting a problem-solving technique rather than a series of specific prescriptions for specific problems.

The book reviews a number of basic principles, criteria, and objectives while presuming a certain depth of knowledge and experience. One of the points the book stresses is the need for mutual support and cooperation of surgeon and therapist. Strikingly absent is any mention of a role to be played by an orthotist. This seems to be more typical than not of hand therapy in general, however. Not too surprisingly, the emphasis is given to use of such readily worked materials as plaster and low temperature plastics. The authors' unabashed preference for plaster is explained, and the problems of it and of plastic are explored.

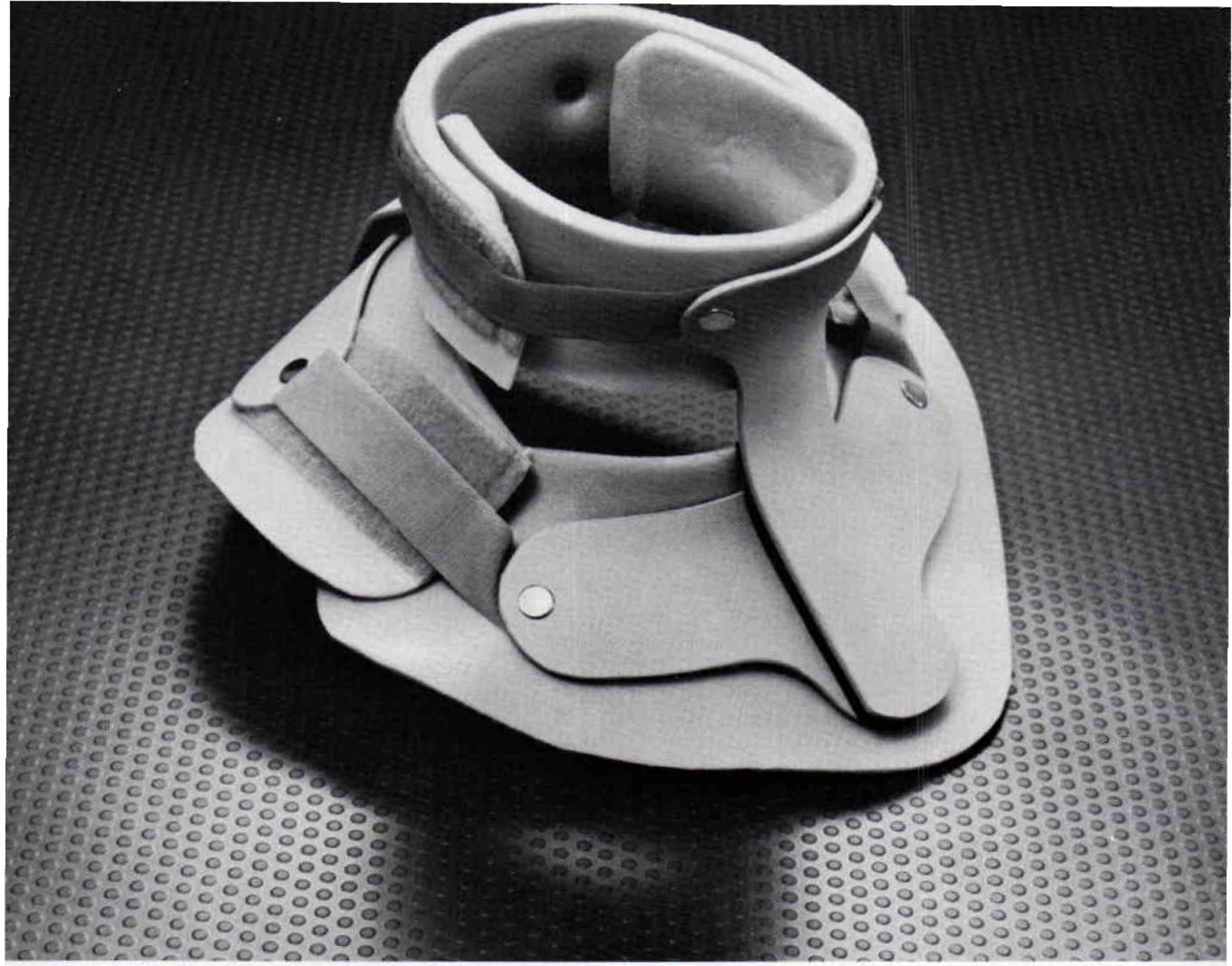
This volume should be read and analyzed in an objective fashion by any orthotist attempting to identify a role for himself in the area of upper limb orthotics. In particular, the reasons why the authors apparently prefer to work without the assistance of an orthotist should be identified.

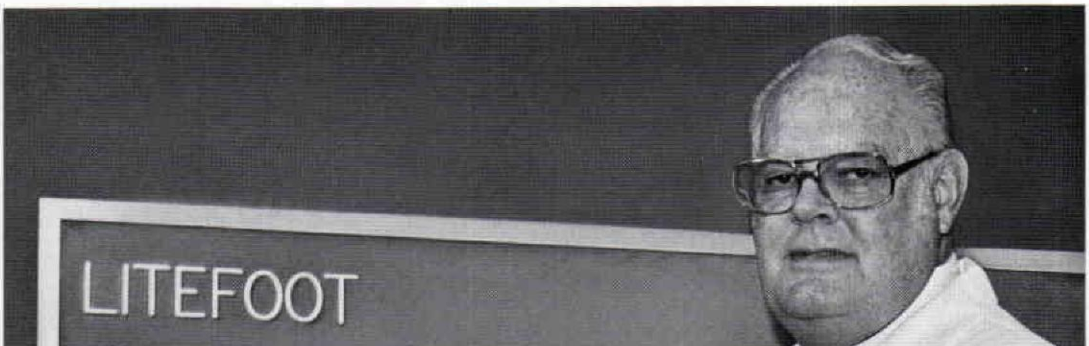
The Cervical Spine, The Cervical Spine Research Society, J.B. Lippincott Company, The Health Professions Publisher of Harper and Row, Inc., East Washington Square, Philadelphia, Pennsylvania 19105. 544 pages and index. February 1983, $57.50.

The book is a compilation of articles by 53 contributors from around the world. In a comprehensive fashion it reviews the subject, covering such matters as anatomy, roentgenographic evaluation, physiology, biomechanics, and neurological evaluation. It also considers such causes for concern as fractures and dislocation, infections, injuries of the cervical cord, tumors, and degenerative disorders. Surgical approaches and techniques are also discussed.

Of particular interest to orthotists is the chapter on use of the halo-vest system and the chapter that compares the effectiveness of various cervical orthoses. This latter chapter is particularly interesting because it summarizes much that has been published in the literature and provides an objective comparison based on various means of measuring motion for various segments of the cervical spine. The use of cervical orthoses is also mentioned where appropriate in relation to specific conditions.

Any orthotist interested in learning more about the cervical spine should find this text useful.

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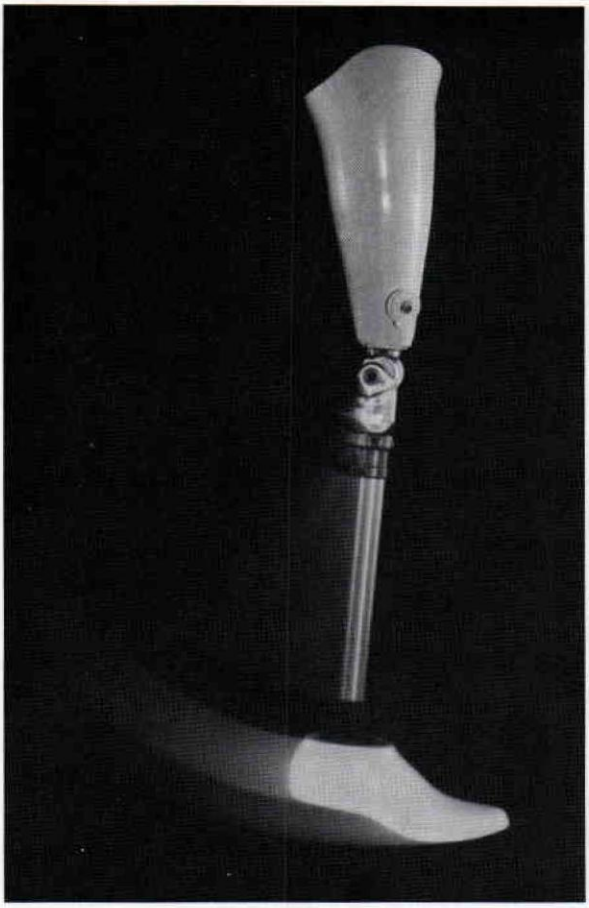

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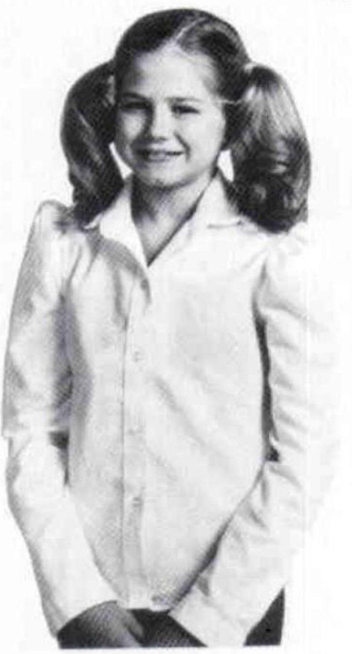

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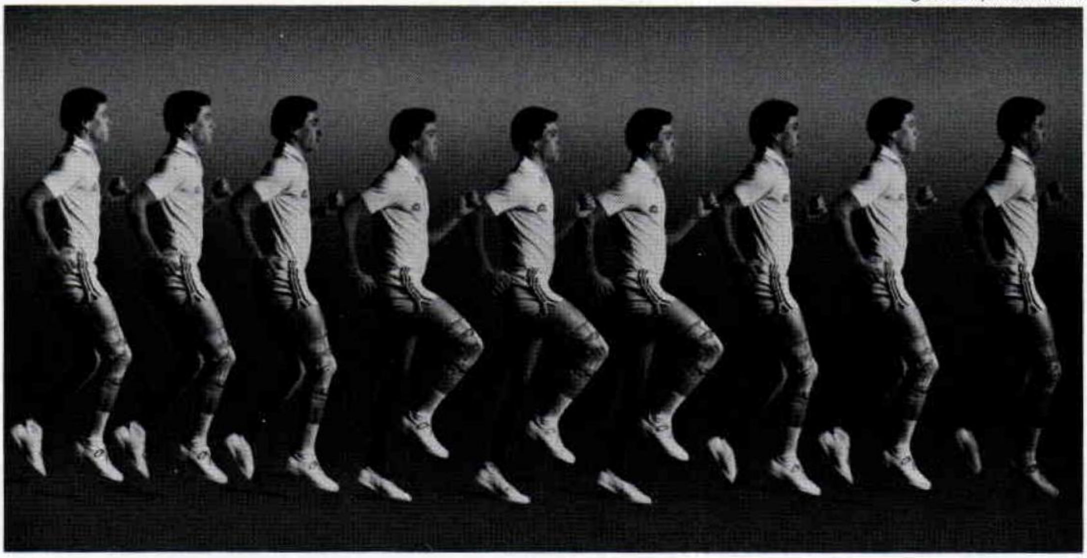

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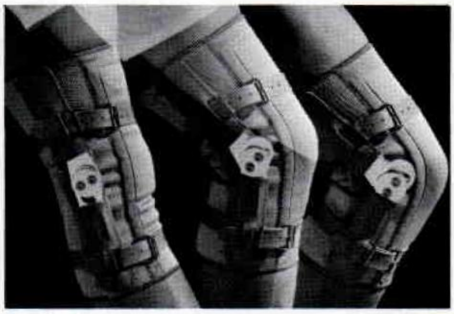

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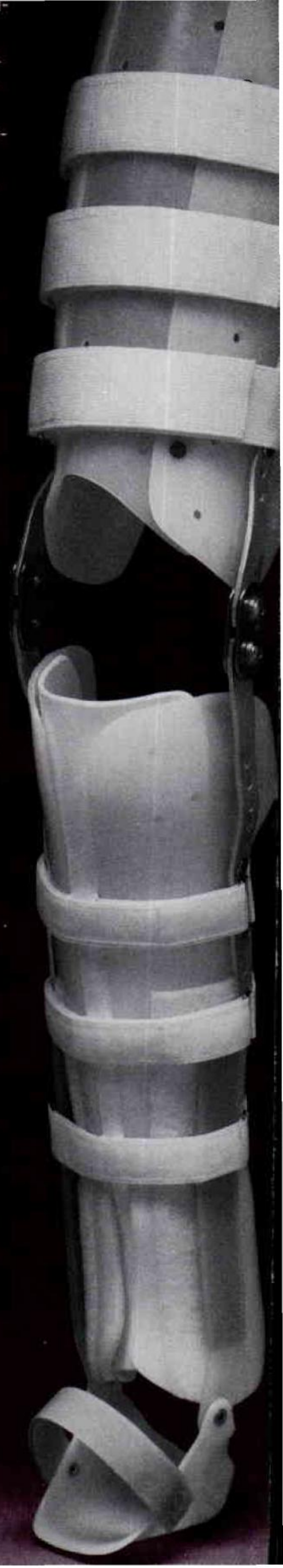

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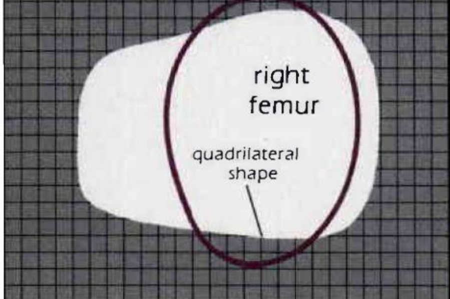

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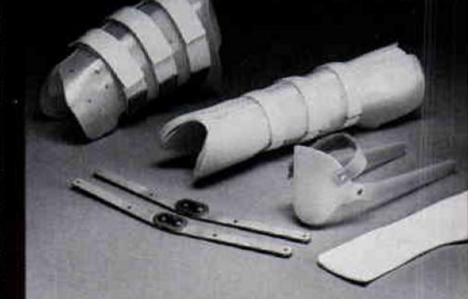

*U.S. Patent 4,320,748

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