Chapter 11

ORTHOTICS FOR THE WOUNDED COMBATANT

JUSTUS F. LEHMANN, M.D.*

**Professor, Department of Rehabilitation Medicine, University of Washington, Seattle, Washington 98195*

INTRODUCTION

The science of orthotics deals with the design and use of orthotic devices, that is, orthoses. Orthoses provide joint immobilization, augment weakened muscles, assist in normal joint biomechanics, and can improve gait and the ability to conduct activities of daily living (ADL). This chapter reviews upper and lower extremity orthoses that may be prescribed to treat war-injured military personnel.

Orthoses for upper extremities include those that assist the veteran who suffers from paralysis or paresis resulting from peripheral nerve injuries; from upper motor neuron lesions, such as traumatic brain injury (TBI); and spinal cord injury (SCI). Specific discussions will address orthoses commonly used by hemiparetic patients whose conditions result from TBI or stroke and orthoses commonly used by quadriplegic patients from high level SCI. The designs of upper extremity orthoses offer great flexibility, especially for the hand. Standard designs of hand orthoses can be modified and customized to provide for the need of the individual patient.

The use of lower extremity orthoses as treatment for neuromuscular conditions will be similarly discussed, beginning with orthoses for paralysis or paresis resulting from peripheral nerve or upper motor neuron lesions (such as in TBI or SCI) or any combination of these. The biomechanical function of the orthosis will be clearly related to specific gait problems. Factors such as appropriate adjustment of the orthosis, reduction of energy consumption, and functional improvements will be addressed. This discussion will include ankle-foot orthoses (AFOs), knee-ankle-foot orthoses (KAFOs), and hip stabilization orthoses. In addition, lower extremity orthoses designed for the alignment of injured joints or bones will be discussed in terms of maintaining alignment and limiting weight bearing.

To manufacture properly fitting and functioning orthoses, technical skill is required and a laboratory must be available. Therefore, only a limited number of the devices described can be used close to the war theater. Most of them should be used in treatment programs that primarily deal with recovery from the injuries.

UPPER EXTREMITY ORTHOSES

Conditions for which orthoses are commonly prescribed include paralysis or paresis caused by (1) peripheral nerve injuries; (2) upper motor neuron lesions, such as TBI; and (3) high level SCIs. In some cases, an orthosis may be used for any of the described applications. Therefore, orthoses for all three conditions[1-5] will be discussed here. For an upper extremity orthosis to be useful, two major functions must be provided: (1) grasp (and release) and (2) reach.

Hand Orthoses

Hand orthoses are primarily designed to provide grasp and release. The following six grasp functions of the hand are used frequently in daily activities: (1) fingertip prehension or palmar prehension; (2) lateral pinch; (3) 3-jawed chuck; (4) large grasp, spherical object; (5) large grasp, cylindrical object; and (6) hook grasp, as for carrying a suitcase (Figure 11-1).

Basic Principles of Attaching Orthoses to the Hand

Due to the greater mobility of a hand and upper extremity in comparison to a lower extremity, a major problem is the firm attachment of the orthosis so it can be used for functional purposes, as shown in Figure 11-2. The basic orthosis is positioned diagonally across the palm of the hand to accommodate a firm grip. It embraces the area of the fifth metacarpal and passes over the dorsum of the hand just beyond the third metacarpal. The third metacarpal is the most firmly anchored component of the palm of the hand, whereas the fourth and fifth are highly mobile in the palmar dorsal direction. To keep the orthosis from sliding off toward the ulnar side, the orthosis is anchored to the side of the second metacarpal; this is called a radial extension (see Figure 11-2). This type of orthosis, without an extension across the wrist joint, is called the basic opponens orthosis. It is held by a strap extending from the dorsal portion of the brace across the wrist proximal to the heel of the hand (Figure 11-3). If the orthosis extends across the wrist into the forearm, it is called a long opponens orthosis. Modification of these orthoses will substitute for many of the grasp functions of the hand.

Orthoses can substitute for grasp by (*a*) stabilization of unstable joints where motion is not needed, (*b*) transfer of power from available muscles

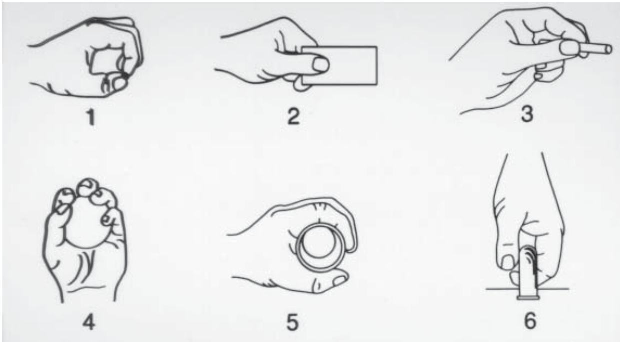


Fig. 11-1. Hand prehension patterns: (1) fingertip or palmar prehension; (2) lateral prehension; (3) 3-jawed chuck; (4) large object, spherical grasp; (5) large object, cylindrical grasp; (6) hook or snap. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965: slide 32: Various commonly used types of grasp.

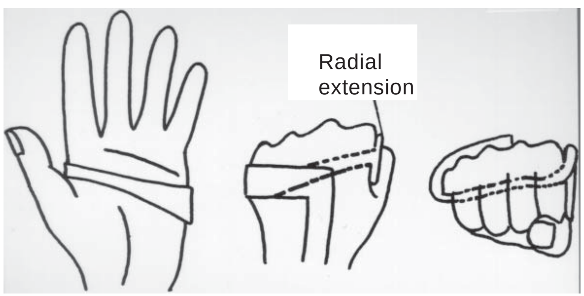


Fig. 11-2. Basic hand orthosis with radial extension.

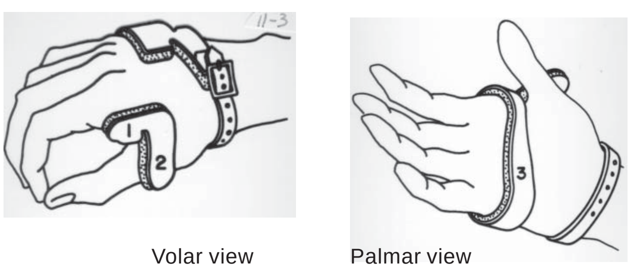


Fig. 11-3. Short opponens hand splint; volar view (left) and palmar view (right): (1) radial extension; (2) opponens extension; (3) palmar arch. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

Fig. 11-4. Basic opponens splint with radial extension, opponens extension, and "C" bar (arrow). Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas;1965.

to move joints that are otherwise immobilized by the paralysis of the primary movers, and (*c*) use of external power.

If no possibility exists to transfer power from a strong muscle to a weaker muscle group, external power may be needed. Two sources of power commonly used are (1) compressed carbon dioxide (CO_2), or (2) an electric motor. The application of power is controlled by microvalves or switches that require minimal force to operate and can be positioned at any place where adequate muscle function exists. A small myoelectric signal may also be used through a microprocessor to proportionally control the external electric power.

Orthoses for Specific Functions

Basic opponens orthosis. This device substitutes for opponens and abductor pollicis brevis functions as they occur in median nerve lesions. To substitute for the absent or paralyzed opposition of the thumb to the fingers, the basic opponens orthosis, made of aluminum, is equipped with an opponens extension (see Figure 11-3). The opponens extension slides the thumb, by means of pressure on the first metacarpal, into opposition to the index and third finger pads. However, it does not move the thumb out of the palm. Therefore, grasp is only possible in the presence of function of the abductor pollicis brevis, which abducts the thumb perpendicular to the plane of the palm. If this muscle is weak or absent, another addition to the opponens splint (Figure 11-4) is the so-called "C" bar, which keeps the thumb out of the palm of the hand. With these modifications, usually a palmar or fingertip prehension can be produced, as well as a 3-jawed chuck. These modifications would oppose the thumb to the tips of the index and third fingers. Thumb flexors or finger flexors must be available for grasp and the finger extensors are needed for release.

The same functional orthosis can be produced by using plastics. By positioning the thumb against the fingers in the same fashion, an abducted and opposed thumb post is created (Figure 11-5). Many other modifications of this orthosis using heat-formed (formed when hot) plastics are frequently used in occupational therapy departments.[6] With further attachments to produce restraining forces, the basic opponens orthosis may be tilted off the hand, without added stabilization against the forearm. Because of this, a long opponens orthosis should be used (Figure 11-6).

Finger and metacarpophalangeal (MCP) extension attachments for conditions caused by radial

nerve lesions. One attachment may be used to substitute for the interphalangeal (IP) joint extension. This orthosis (Figure 11-7) uses rubber band pulls with plastic sleeves to assist extension of the finger joints. Proximal or distal IP joint support is dependent on sleeve position. The bar just distal to the MCP joints serves to stop the MCP joints from going into hyperextension, due to the constant pull of the rubber bands. To prevent hyperextension of the IP joints, it is essential that the rubber band tension is finely tuned to provide just enough force to extend the fingers. Clearly, the force of the rubber band pull must be deducted from the finger flexor force, since the fingers have to work against the rubber band. Similarly, the extension of the MCP joints can be supported. For example, this orthosis should be used in cases of radial paralysis with good finger flexor strength, that is, intact ulnar or median nerve innervated flexors superficialis and profundus.

Long opponens splint with intrinsic bar (Figure 11-8). Normally, IP joints of the fingers are extended by the lumbricals and interossei, which also flex or stabilize the MCP joints; the extensor digitorum longus extends and, when unopposed, hyperextends the MCP joints. In the absence of these intrinsic functions, the extensor digitorum can be used to extend the IP joints for grasp release, provided that the intrinsic bar of the orthosis keeps the MCP joints in slight flexion. The unopposed action of the extensor digitorum would hyperextend the MCP joints associated with slight flexion of the IP joints.

The range of orthosis flexibility is illustrated by the following: a patient with a weakness of the extensor digitorum, with strong flexors digitorum

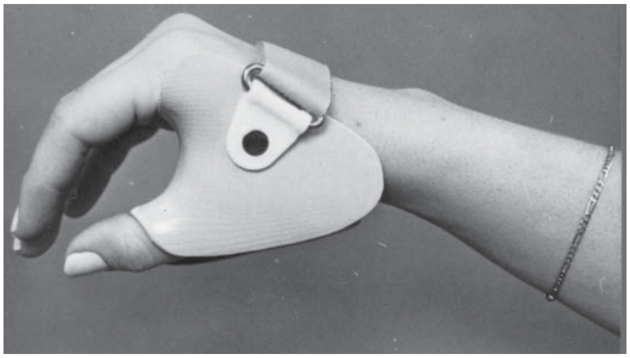


Fig. 11-5. Plastic basic opponens splint (Engen), keeping the thumb in opposition to the second and third fingers and abducted out of the plane of the palm.

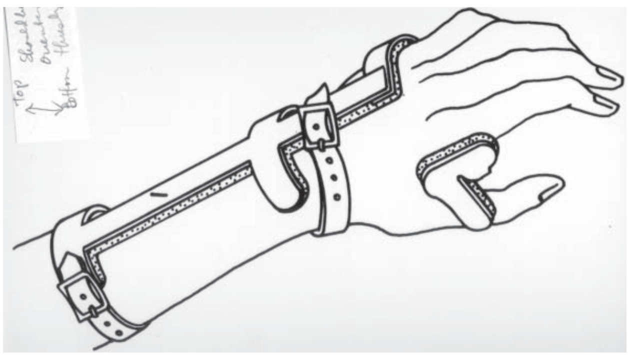


Fig. 11-6. Long opponens hand splint with forearm piece (1) and radial extension and opponens extension. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

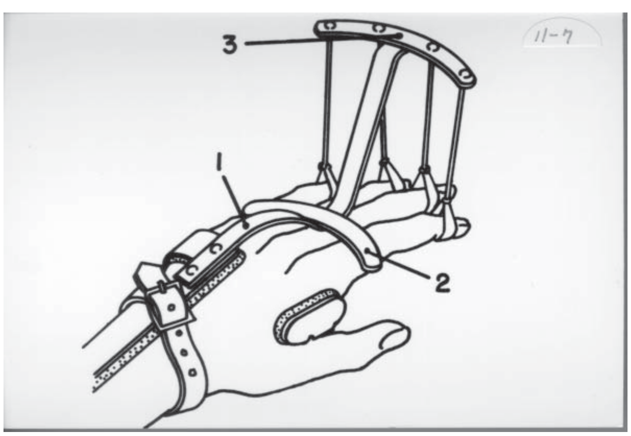


Fig. 11-7. IP extension assist with MCP stop: (1) allows detachment from the long opponens splint; (2) MCP extension stop; (3) outrigger with rubber band IP joint extension assist. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

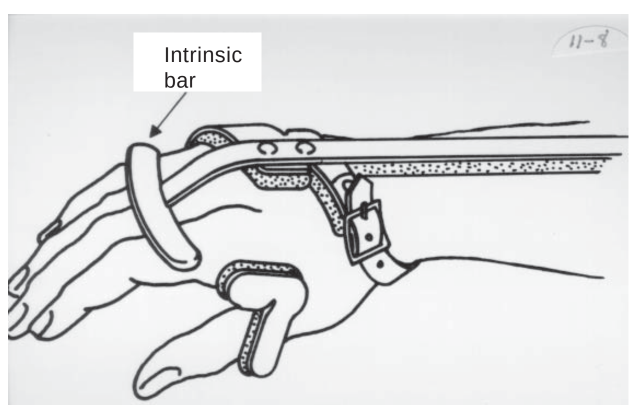


Fig. 11-8. Long opponens splint with intrinsic bar. Reprinted with permission from Anderson MH: *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

profundus and superficialis and available intrinsic musculature (such as palmar and dorsal interossei and lumbricals) has difficulty releasing the grasp. The patient may try to use the intrinsic muscles to extend the IP joints, at the same time, however, enacting unopposed maximal flexion of the MCP joints. If a wrist support orthosis is combined with a palmar piece that extends just beyond the MCP joints and limits flexion of these joints, the opening of grasp can be accomplished by the intrinsic muscles. These extend the IP joints, and the limitations of the MCP joint still allows a degree of functional grasp.

Opponens orthosis with holders for commonly used utensils. Multiple lesions may lead to conditions where this type of orthosis is needed. They may include upper motor neuron lesions, such as with SCI, or the combination of several peripheral nerve lesions. In these situations, adequate force of grasp and prehension are not available for effectively using these types of utensils (Figure 11-9). Attachments to an orthosis can hold such items as pencils, spoons, toothbrushes, and so forth.

Flexor Hinge Orthoses

Flexor hinge orthoses are primarily used in quadriplegia resulting from SCI.

Wrist extensor driven flexor hinge orthosis. Wrist extensor-driven flexor hinge orthoses[7] are shown in Figures 11-10 through 11-13. This orthosis stabilizes the thumb rigidly in opposition to the index and third finger. It stabilizes and immobilizes the IP joints of the second and third fingers in a semiflexed position for grasp and allows movement only in the MCP joints. As the fingers are flexed at these joints, the 3-jawed chuck pinch is produced, and, as they are extended, the grasp is released. The motor for this movement comes from the radial wrist extensors. The proximal part of the orthosis

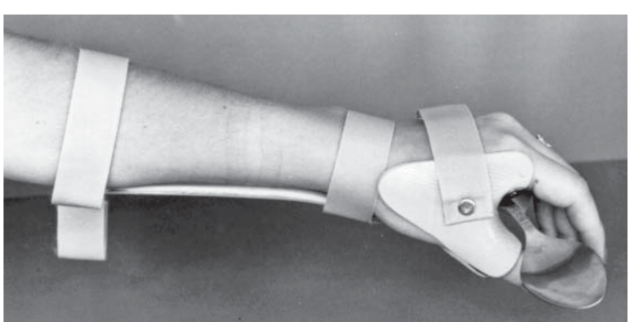


Fig. 11-9. Spoon fitted with a spring clip holder.

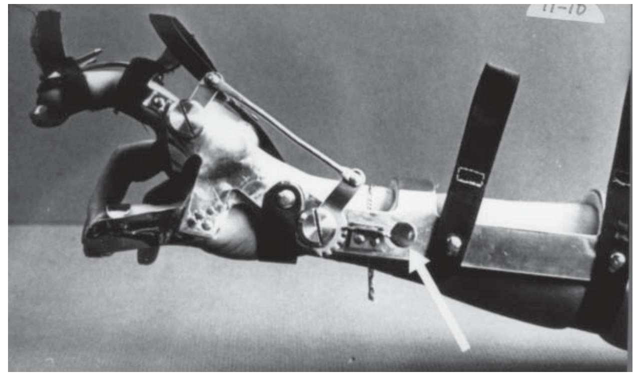


Fig. 11-10. Metal flexor hinge splint with adjustable ratchet controlling wrist angle at closing and opening: with fingers open; arrow indicates button to adjust ratchet.

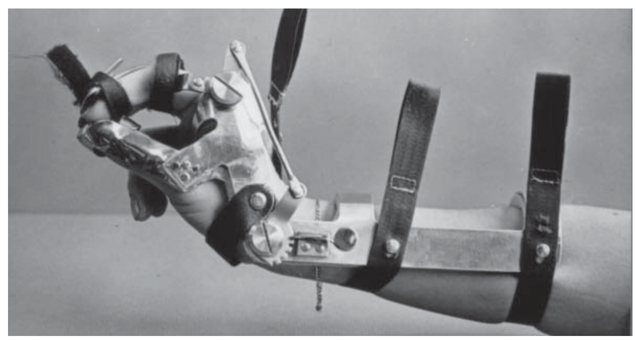


Fig. 11-11. Metal flexor hinge splint with adjustable ratchet controlling wrist angle at closing and opening: with fingers closed.

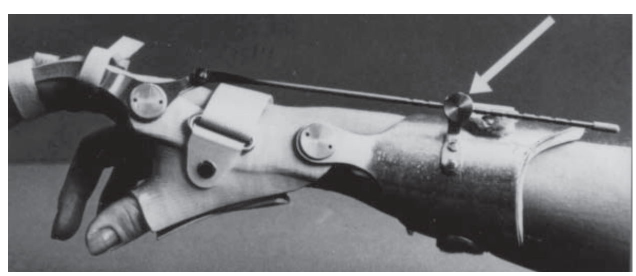


Fig. 11-12. Wrist-driven plastic flexor hinge splint with adjustable linkage across wrist to control extension and flexion angles at closing and opening (Engen); with fingers open, arrow indicates button to adjust the length of the linkage.

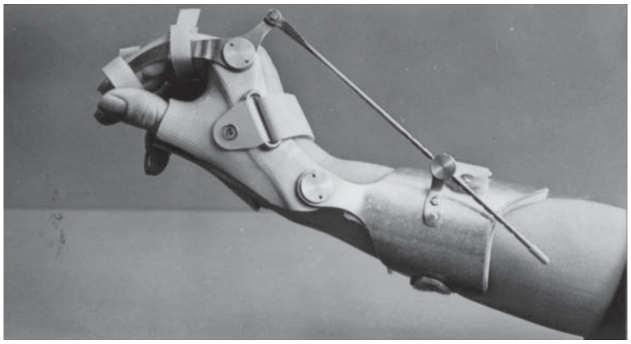


Fig. 11-13. Wrist-driven plastic flexor hinge splint with adjustable linkage across wrist to control angle at closing and opening (Engen): with fingers closed.

is connected with a hinge across the wrist, with the part of the orthosis extending over the forearm, as in the long opponens splint. The transfer of power from the wrist extensors occurs by metal rod linkage across the wrist, which produces grasp on extension of the wrist and release on flexion of the wrist (see Figures 11-10 through 11-13) This splint is especially useful for SCI patients who have some function at C5-C6, that is, the innervation of the radial wrist extensors, while the ulnar wrist extensors and the wrist flexors innervated by lower spinal segments are paralyzed. Therefore, grasp-release through wrist flexion is often provided by gravity or, on occasion, by rubber band pull that would oppose, in turn, the grasp movement produced by wrist extension. Because the radial wrist extensors receive the innervation from a higher level of spinal cord innervation, they are often stronger than the ulnar wrist extensors. Thus, a strong tendency exists to drive the wrist into radial deviation as it is extended. To prevent binding of the joint at the wrist, a flexible portion is incorporated into the orthosis at the wrist to allow this radial deviation to occur. Use of all flexor hinge-type orthoses necessitates some protective sensation between the thumb and index finger.

Shoulder harness driven flexor hinge orthosis. This orthosis is used in the presence of complete paralysis of the hand. Necessary physiologic conditions are some protective sensation in the fingertips and the voluntary ability to reach so that the hand can be positioned for grasp. Shoulder motion pulls the cable out of the housing and transfers the power from the shoulder musculature to the orthosis to release the grasp. Closure of the grasp is produced by rubber band or flexor spring pull (Figure 11-14). Since the rubber band would pull the fingers together all the time and, therefore, produce an ischemic necrosis of the tips, a mechanical release is provided which can stop the finger flexion with the fingertips apart. A quick pull release by the harness would overcome the stop and close the fingers.

External power driven flexor hinge orthoses.[8] The two sources of external power commonly used in flexor hinge orthoses are electrical motor drive or compressed carbon dioxide with artificial muscle (McKibben) as the motor. The electrical drive can be controlled by a switch, which the patient may activate by any muscle available. It can also be activated by an available, although weak, electromyographic signal of a voluntarily controlled muscle. By way of a microprocessor, this signal can proportionally control the motor drive.

The compressed CO_2 is provided in a metal cylinder. The gas is used to inflate an artificial muscle (McKibben), which consists of an inner rubber tube and an outer helically woven stocking. As the rubber tube is inflated and expands, the sleeve shortens in a similar fashion as a muscle. Inflation of the muscle produces a 3-jawed chuck grasp. The grasp release after deflation can be produced by gravity, spring, or rubber band pull. Inflation and deflation of the muscle are controlled by microvalves, which can be manipulated by any minimal voluntary power available any place. Like the other models of flexor hinge splints, the basic mechanical design remains the same, and this power driven orthosis should be used only in the presence of some protective sensa-

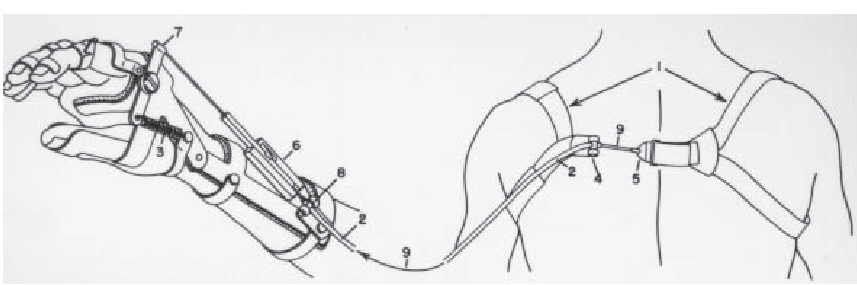


Fig. 11-14. Shoulder-driven flexor hinge splint. 1 = "butterfly" or shoulder-to-shoulder double loop harness; 2 = Bowden cable; 3 = flexor spring; 4 = leather retainer; 5 = stainless steel hanger; 6 = pressure relief control; 7 = finger piece operating lever; 8 = crossbar and anchor for cable housing; 9 = cable. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

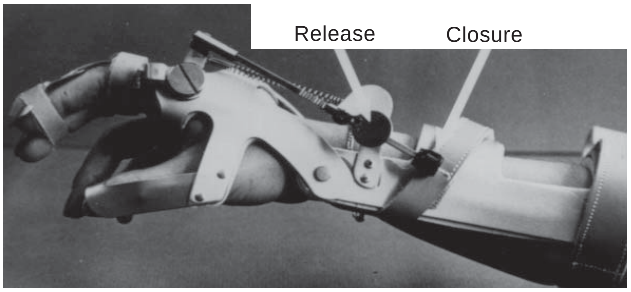


Fig. 11-15. Ratchet splint with push button for closing and pressure button for release; with fingers open.

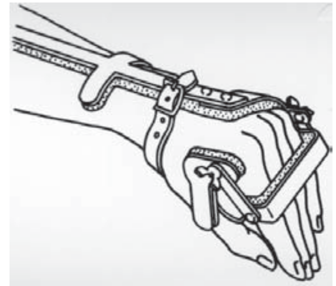


Fig 11-17. Adjustable MCP flextion control, "knuckle bender." Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

tion of the fingers and with a functional ability to reach. The power sources, electric batteries and gas cylinders, are usually attached to the wheelchair.

Passive prehension orthosis (flexor hinge splint, with ratchet control). This splint (Figures 11-15 and 11-16) allows a passive closing of the fingers by pushing a lever with a ratchet. Any force of prehension can be maintained, due to the ratchet stop. A button is pushed to open the ratchet lock, and grasp is released as a result of a spring pull. The force to push the ratchet lever to close the fingers has to be provided by the other extremity or by pushing against the lapboard or the wheelchair. This orthosis is commonly used for high level SCI patients who have no wrist extension and therefore cannot use the wrist-driven flexor hinge splint. Controlled prehension and release has also been achieved using functional electrical stimulation (FES),[9] although its use is uncommon.

Orthoses for Reduction of Contractures

Contractures can develop in cases of long-standing paralysis with partial immobilization of the joints of the hand and wrist, or as a result of prolonged casting. Contractures can limit either flexion or extension, depending on the position at which the joint has been immobilized.[10] In addition to physical therapy, a so-called "knuckle-bender" orthosis can be used to apply a long-term stretch to reduce the contracture (Figure 11-17).[11] There are many models constructed of metal, plastic, or springs, which include the knuckle-benders, or Bunnell splints, and allow endless varieties of positions for the fingers and wrist. These orthoses were frequently used for the Persian Gulf War casualties, and incorporated the wrist-hand orthotic base and outriggers with rubber bands for prolonged static stretch.

Interphalangeal Joint Stabilizers

If there is any instability of an abnormal IP joint, an IP stabilizer can be used to make the hand more functional by preventing movement at an IP joint (Figure 11-18).

Wrist Orthoses

Volar Wrist Flexion Control Orthosis

The volar wrist flexion control orthosis (cockup splint) is usually made out of plastic (Figure 11-

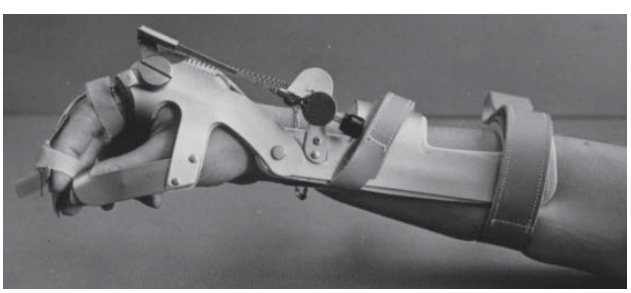


Fig. 11-16. Ratchet splint with push button for closing and pressure button for release; with fingers closed.

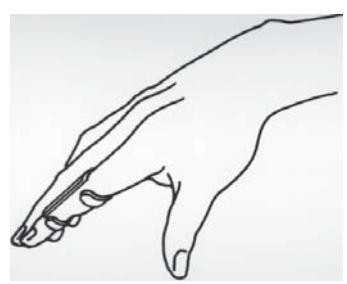


Fig. 11-18. IP joint stabilizer. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

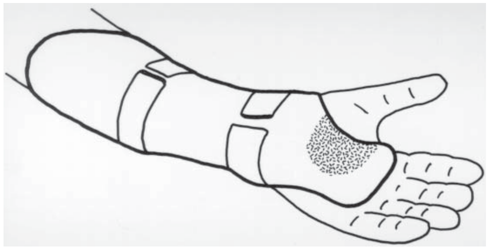


Fig. 11-19. Volar wrist flexion control orthosis, acrylic polyvinyl chloride.Photograph: Courtesy of DeRoyal LMB, San Luis Obispo, Calif.

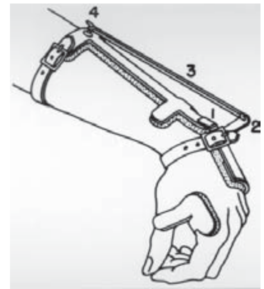


Fig. 11-20. Action wrist with dorsiflexion assist. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

19).[12,13] It consists of a rigid volar formed section, which continues into the palm. Straps over the dorsum of the forearm hold the orthosis in place. It supports the wrist in approximately 30° of dorsiflexion. In the presence of weak wrist dorsiflexion, the unopposed finger flexors pull the wrist into maximal flexion, and thereby allow maximal shortening of the flexor muscles. In this shortened position on the length-tension diagram, the strength of grasp is significantly reduced. This orthosis allows full grasp strength by limiting finger flexor excursion through immobilization of the wrist, and is used in cases of weakness of the wrist extensors.

Wrist Extension Assist (Action Wrist Extension with Dorsiflexion Assist)

A wrist extension assist orthosis consists of a forearm portion applied to the dorsum of the forearm with a hinge at the wrist. The portion of the orthosis over the hand distal to the hinge is similar in construction to the basic opponens orthosis, but without attachments. The spring or rubber band passing across the wrist dorsally assists the wrist extensor muscles. The force provided by the rubber band or spring extension combined with the remaining wrist extensor musculature strength must stabilize the wrist against the considerable forces of the finger flexors pulling the wrist into flexion, a condition which is rare. A prototype example is the long opponens splint with a hinge at the wrist and rubber band or spring extension assist (Figure 11-20). This orthosis can also be used for radial nerve injuries when it is combined with MCP extension assists.

Devices to Provide Reach

In many conditions, such as those caused by high level cervical spinal cord lesions, the patient may not have enough voluntary muscle control to place the hand in a position of grasp or to transfer objects, once grasped, from one location to another. At the same time, many of these patients are not ambulatory and are confined to a wheelchair. Therefore, the most common devices used for support of the arm, allowing some limited reach, are ball bearing mobile arm supports[14] and overhead slings, all of which are attached to a wheelchair.

Ball Bearing Mobile Arm Support

The support bracket for the ball bearing mobile arm support[15] (Figures 11-21 through 11-24) is attached to the steel tubing on the back of the wheelchair. The incline of the ball bearing in the sagittal plane can be adjusted; a swivel arm is inserted into this bearing. Another bearing is fixed at the distal end of this arm, into which the distal swivel arm is attached. Both ball bearings may have stops to prevent undesirable and uncontrollable motions. A trough that supports the forearm is attached to the distal swivel arm, and a dial attached to this trough keeps the forearm from slipping out of the trough support. The trough, in turn, is allowed to swivel up and down around a bearing located close to the center of mass of the forearm (Figure 11-25). By this design, which replaces active elbow flexion and extension, only minimal force is required to alter the balance of the trough and cause the hand to move down or up.

In addition, a mechanical guide can be installed so that the trough moves the hand down and, thus, into pronation for easy grasp. As the trough swivels the hand up, the hand is guided into supination for the feeding motion (Figure 11-26). As the trough moves the hand down, the guide moves the hand in pronation for easy grasp.

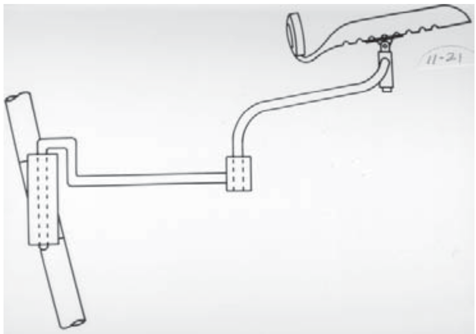


Fig. 11-21. Ball bearing mobile arm support, side view. Reprinted with permission from Wilson DJ, McKenzie MW, Barber LM. *Spinal Cord Injury: A Treatment Guide for Occupational Therapists*. Thorofare, NJ: Charles B. Slack; 1974.

Fig. 11-24. Patient using ball bearing mobile arm supports and hand orthoses to stack coins.

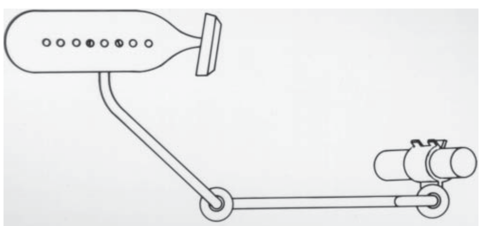


Fig. 11-22. Ball bearing mobile arm support, top view. Reprinted with permissin from Wilson DJ, McKenzie MW, Barber LM. *Spinal Cord Injury: A Treatment Guide for Occupational Therapists*. Thorofare, NJ: Charles B. Slack; 1974.

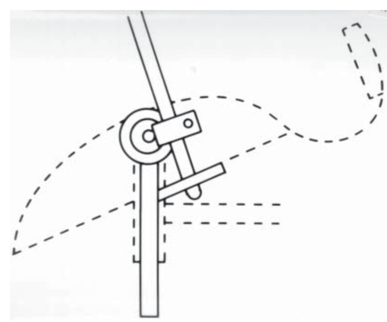


Fig. 11-25. Ball bearing mobile arm support, outside rocker arm assembly.

Fig. 11-23. Patient using ball bearing mobile arm supports and hand orthoses to write.

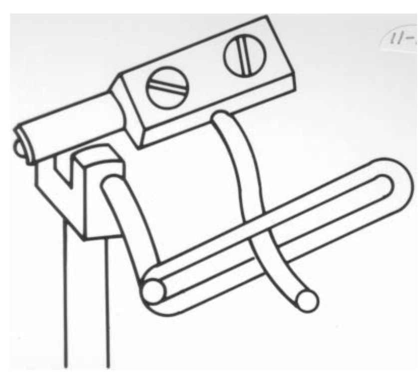


Fig. 11-26. Ball bearing mobile arm support, supinator assist.

If the ball bearing at the back of the wheelchair is adjusted so that the distal portion of the attached swivel arm points down and in a plane away from the body, gravity will extend the elbow and arm, moving the hand away from the body. If the rear ball bearing is adjusted so that the distal portion of the swivel arm is up, gravity will flex the elbow, and bring the hand close to the body. Therefore, the ball bearing support moves on the principle of an inclined plane. If the patient has musculature to pull the arm to the body, the plane inclines away from the trunk and helps to extend the reach. If the patient has no musculature to bring the hand close to the body or the mouth but has extensor musculature to push the hand and arm away from the body, the plane is adjusted to incline toward the body. In either case, if the adjustment is fine-tuned, only minimal forces are used to produce the desired movement. This type of substitution for a poor voluntary reach function is typically used in combination with some of the hand orthoses previously described.

The adjustment of this ball bearing mobile arm support, as well as the support trough for the arm, is critical for maximal utilization of the minimal voluntary force the patient will have available. Those who can benefit from this orthosis include patients with weak C-5 innervated muscles, brachial plexopathy, and SCI. All of these conditions occurred among Persian Gulf casualties.[16]

Forearm Orthosis with Friction Joints (Friction Feeder)

The same basic design, as in the ball bearing mobile arm support, can be used for a different purpose. The important design modification is that the ball bearing joints are replaced by joints with adjustable friction. The forearm orthosis with friction joints is able to control involuntary movements that often occur in various types of brain injuries. For instance, in conditions caused by cerebellar or cerebellar-pathway lesions, the orthosis is able to dampen ataxia and dysmetria, which enables the patient to get better controlled motion (Figure 11-27). In the same way, involuntary tremors, which often occur as a result of lesions of the basal ganglia, can be controlled.

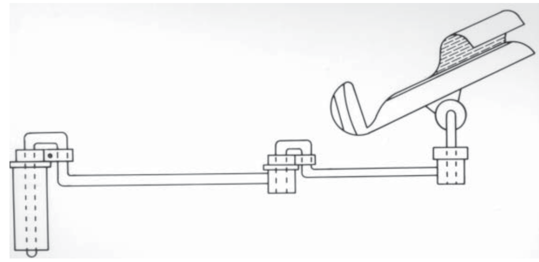


Fig. 11-27. Friction feeder, side view.

Suspension Sling Arm Support

The suspension sling arm support uses essentially the same trough design as is used in the ball bearing mobile arm support, and allows swiveling of the hand down or up. The arm can also be supported (Figure 11-28) by a sling at the elbow and one at the wrist. The slings are suspended by a horizontal bar, which allows balance in favor of hand elevation or downward movement. This swivel bar is attached to the overhead support. In case of availability of muscles that depress the shoulder, it is sometimes useful to insert a spring between the overhead support and the swivel bar. The overhead support is attached by a metal bracket to the wheelchair. The patient with the least amount of voluntary force uses the trough suspension, while one with the most voluntary movement may use the sling suspension with the spring inserted to utilize the availability of shoulder depressors. With the straight, nonextendable suspension from above, these slings work on the principle of a pendulum. The upper extremity is supported overhead against gravitational forces (see Figure 11-28). Reaching movements can be accomplished by pushing the arm and hand forward or pulling them back. The arm follows the movement of a pendulum. There-

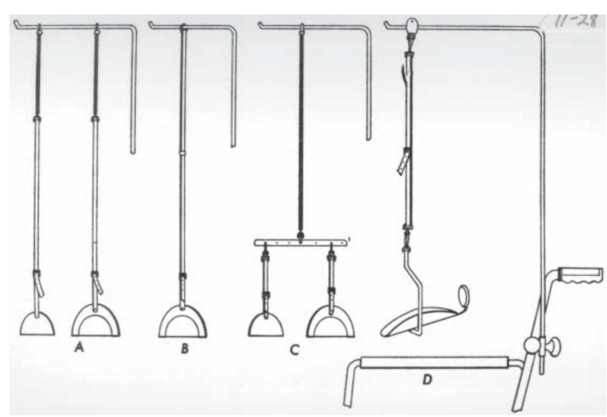


Fig. 11-28. Sling suspensions. **A**: double sling; **B**: single sling; **C**: sling with rocker arm; **D**; sling with rocker trough. Reprinted with permission from Long C, Schutt AH. Upper limb orthotics. In: Redford JB, ed. *Orthotics Etcetera.* 3rd ed. Baltimore, Md: Williams & Wilkins; 1986.

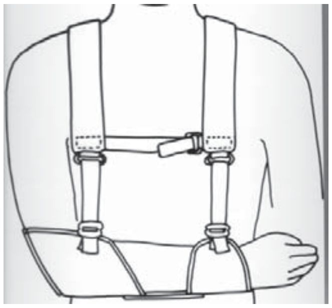


Fig. 11-29. Harris hemisling. Reprinted with permission from Brooke MM, de Lateur BJ, Diana-Rigby GC, Questad KA. Shoulder subluxation in hemiplegia: Effects of three different supports. *Arch Phys Med Rehabil.* 1991;72:582-586.

fore, some minimal lift and minimal force is required to push the arm forward or backward in relation to the rest position. To get adequate reach, these devices require significantly more voluntary muscle control than the ball bearing mobile arm support. In cases of brachial plexopathy, as well as SCI, it has been found that the suspension sling arm support increases shoulder range-of-motion, exercises the shoulder, and also provides a degree of early ADL independence.

Static Orthosis in Support of Shoulder and Elbow

Appliances commonly used on surgical services such as shoulder abduction orthoses, airplane splints, plaster casts, and other immobilization devices will not be discussed in this section.

Different types of shoulder sling supports are used in conditions which result from lack of voluntary control or reduction of muscle strength or tone of the deltoid or other glenohumeral joint musculature.[17-20] For instance, in central nervous system lesions that lead to hemiplegia, subluxation of the humeral head out of the glenoid of the scapula may occur from lack of shoulder support against gravity; thus, subluxation may be associated with pain and other complications. In a recent evaluation[21] of sling supports of various designs, the Harris hemisling (Figure 11-29) was found by radiograph measurements to give the best vertical correction, whereas the Bobath sling was not as effective and distracted the humeral joint horizontally. Sling designs similar to the Harris hemisling, such as the multiple and single strap designs (Figure 11-30), may be used in support of the forearm in the absence of elbow flexors. Other slings, such as the vertical arm sling, may support the arm, but allow elbow extension. Similar systems have been reviewed by Wynn Parry.[22,23]

Williams and colleagues[24] compared hemiparetic patients with shoulder subluxation treated with a Bobath shoulder roll to patients with the same condition treated with the Anderson shoulder ring. There was no difference between the two methods of management. However, greater subluxation occurred without any treatment.[24] Orthotic shoulder stabilizers for trapezius and serratus anterior weakness, stabilizing the shoulder, have been recommended by Villanueva[25] and by Truong and Rippel.[26]

Functional Orthoses for Shoulder and Elbow

Functional bracing of the shoulder and elbow is limited, since isolated functional weaknesses are rare, and only a few conditions lend themselves to bracing these two joints in an ambulatory patient. Also, various surgical approaches can be used, such as fusion of the glenohumeral joint or muscle and tendon transfers. Patients with high level SCIs and more widespread functional loss are often not ambulatory; therefore, the sling suspension or ball bearing mobile arm support must be used. Brachial plexus lesions often result in both motor and sensory losses and, therefore, bracing might not be the correct solution in the absence of protective sensation.

Basic Support of Shoulder and Elbow Orthoses

Development of an adequate support for the attachment of shoulder and elbow orthoses poses basically the same problem as in the hand: the need of firm attachment to allow proper function.[27,28] The shoulder and shoulder girdle are highly mobile and

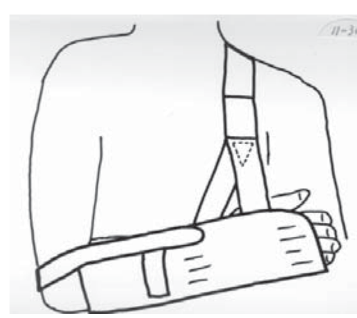


Fig. 11-30. Single strap design for shoulder support. Photograph: Courtesy of DeRoyal LMB, San Luis Obispo, Calif.

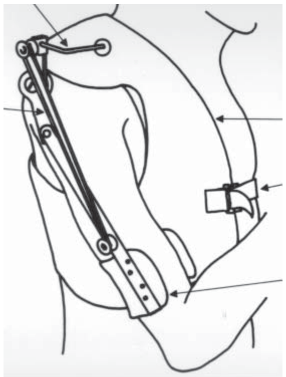


Fig. 11-31. Plastic shoulder cap design with rubber band shoulder flexion assist. Reprinted with permission from Anderson MH. *Upper Extremity Orthotics*. Springfield, Ill: Charles C Thomas; 1965.

Fig. 11-33. Axillary shoulder suspension hoop.

Fig. 11-34. By-pass shoulder suspension hoop.

do not readily allow direct application of orthoses[29]; therefore, a basic support structure has to be designed.

The basic support structure for the shoulder and elbow orthoses can be made of plastic in the form of a shoulder cap design that encloses half of the thorax below the axilla and covers the shoulder, and thereby allows freedom of shoulder motion. The cap is held in place by a chest strap or straps, and a U-shaped bar is attached to the cap of the plastic orthosis (Figure 11-31). Other modifying orthoses are ultimately attached to this U-bar. An alternative for this basic support structure is a pelvic band that encircles the trunk between the iliac crest and the greater trochanter. This band is rigid on the side of the orthosis and contains a soft closure on the opposite side. To this pelvic band, three rigid structures may be attached and held against the chest by a chest strap. These are the straight, the axillary, and the shoulder bypass suspension hoops (Figures 11-32, 11-33, and 11-34, respectively).

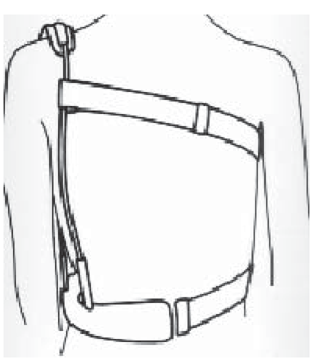


Fig. 11-32. Straight shoulder suspension hoop.

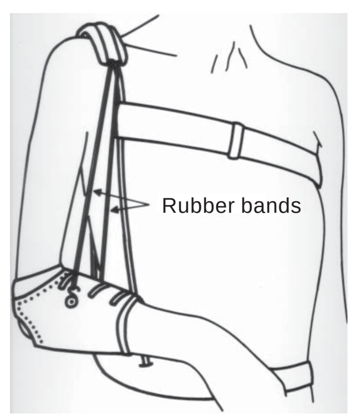


Fig. 11-35. Straight shoulder suspension hoop with forearm cuff elbow flexion assist.

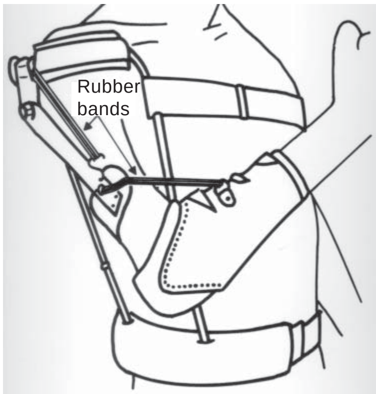


Fig. 11-36. Shoulder flexion assist with forearm cuff and elbow flexion assist.

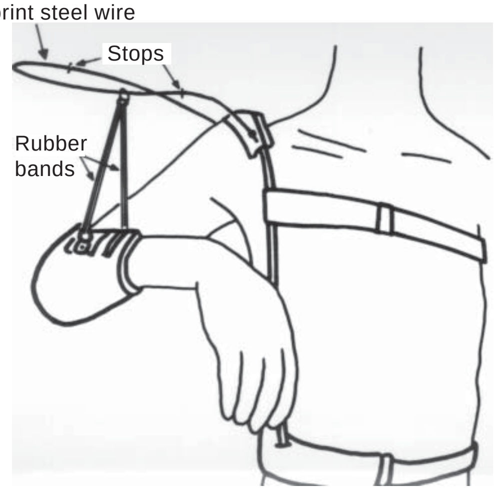


Fig. 11-37. Straight shoulder suspension hoop with abduction outrigger.

Orthoses are attached to these suspension hoops, as in the plastic design with the U-bar. As an example, Figure 11-35 shows a straight shoulder suspension hoop with an elbow flexion assist, that consists of a forearm cuff and rubber bands to support it. Figure 11-31 shows an orthosis with shoulder flexion assist that includes the rubber band. It is attached to the U-bar of the plastic base. Figure 11-36 shows an orthosis that is a combination of a forearm cuff elbow flexion assist and a shoulder flexion assist. As a basic support structure, the shoulder bypass suspension hoop was used. A final example of the approach to bracing the shoulder is shown in Figure 11-37. A shoulder bypass hoop is used for the attachment of an outrigger to assist shoulder abduction through a rubber band supported forearm cuff. Any movement against the rubber band support requires enough voluntary force to overcome the rubber band tension.

These examples show how components of available orthoses should be put together to fit the individual patient. Often, imagination is necessary to modify and even invent new designs to meet a particular patient need.

LOWER EXTREMITY ORTHOSES FOR NEUROMUSCULAR CONDITIONS

Ankle-Foot Orthoses

AFOs are the most commonly used braces for lower extremities. They are used in patients with TBI, SCI, and peripheral nerve injury. The AFOs substitute for loss of physiologic function, and additional walking aids are rarely needed unless there are other associated problems with a lesion that go beyond the paralysis or paresis of the leg and involve more widespread lower extremity groups, generalized spasticity, balance problems, or significant sensory feedback loss.

Components

AFOs can be made of metal or plastic. The metal orthoses discussed in this section are constructed of uprights of steel or aluminum that are connected on top and posteriorly by a rigid metal padded calf band with a soft front closure (Figure 11-38). They are attached below by an ankle joint to a stirrup incorporated into the sole of the shoe. The stirrup is often combined with a rigid metal sole plate which is riveted to the stirrup and extends to the metatarsal head area (Figure 11-39). Commonly, stops are attached to the ankle joint (Figure 11-40). There may be a posterior (plantar flexion) stop to substitute for the foot dorsiflexors and support the toe during the swing phase. These stops can be made rigid by the insertion of a metal rod into the posterior channel of the ankle joint, which engages the flange of the stirrup at an angle adjustable with a set screw. A more substantial assist for weak dorsiflexors would be a spring wire dorsiflexion assist orthosis, where a spring can be used instead of the rigid metal rod (see Figure 11-40).

The anterior stop (dorsiflexion stop) uses an adjustable pinstop, where the pin engages an anterior

Fig. 11-38. Metal orthosis constructed out of bar stock metal uprights of steel or aluminum with Klenzak-type joint and posterior plantarflexion stop.

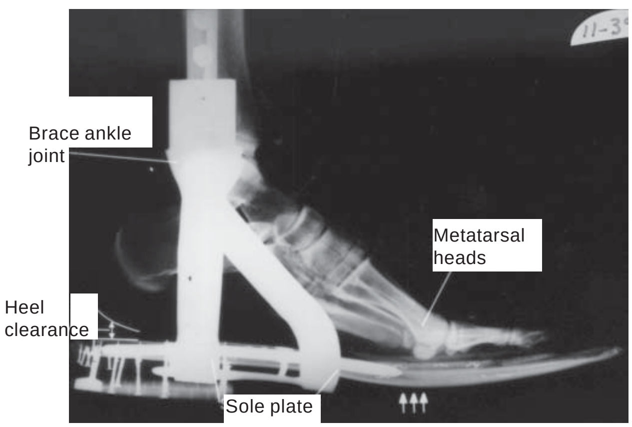


Fig. 11-39. Radiograph of a foot in an ankle-foot orthosis.

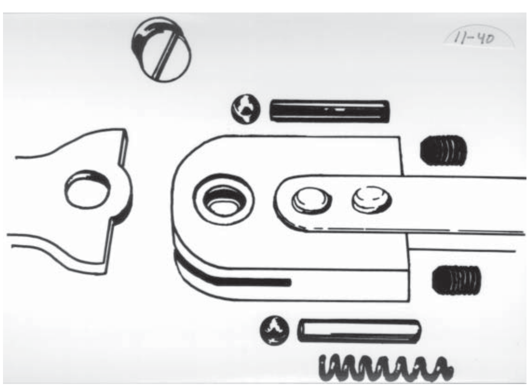


Fig. 11-40. Double-stopped ankle joint. The top stop limits dorsiflexion and the bottom stop limits plantar flexion. Reprinted with permission from Lehmann JF. Lower limb orthotics. In: Redford JB, ed. *Orthotics Etcetera.* 3rd ed. Baltimore, Md: Williams & Wilkins; 1986.

flange in the stirrup at a set angle. This stop should be used only in combination with the sole plate to the metatarsal head area; a spring assist will be too weak. This stop substitutes for the foot plantar flexors acting during pushoff. Figure 11-40 shows a Becker type ankle joint. The same stops are also commonly used in a Klenzak type joint (see Figure 11-38). Plantar flexion and dorsiflexion stops can be used together in the so-called double-stopped joint (see Figure 11-40).[30]

Indications

Most conditions requiring AFOs are due to weakness caused by upper and lower motor neuron lesions. The brace will accommodate spasticity and muscle imbalance. Many of these patients are able to walk without the orthosis; however, they are in danger of falling either because they trip as a result of the toe drag, or because mediolateral instability causes them to turn an ankle and fall.[31] Reasons for applying orthoses are to (*a*) provide mediolateral stability, and substitute for the lack of pushoff; (*b*) minimize knee instability during the early stance and increase stability during the late stance; (*c*) provide toe pickup during the swing phase; (*d*) approximate a normal gait pattern and, thus, reduce energy expenditure; and (*e*) prevent the development of deformities by improper weight bearing on weakened muscles.

Biomechanical Function

The biomechanical function of AFOs will largely depend on their design, irrespective of whether they are plastic or metal. Therefore, the basic biomechanical functions will be discussed with the standard double upright metal orthosis with stops at the ankle as a prototypical model.

Mediolateral stability. The orthosis in Figure 11-38 usually provides adequate mediolateral stability unless there is a strong tendency at the ankle to invert or evert. To prevent inversion (varus) of the foot, a T-strap may be attached to the sole of the shoe at the level of the instep, passing inside the lateral upright of the AFO and over the lateral malleolus, and cinched outside around the medial upright (Figure 11-41). As a result, the protruded lateral portion of the ankle is forced in line with the shoe and foot below and the calf band above, correcting the inversion. This type of correction is often needed in hemiparesis produced by stroke or TBI.

In eversion deformity (valgus) of the foot, the T-strap is applied to the shoe directly below the me-

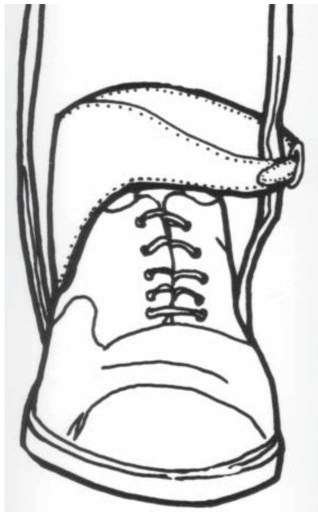


Fig. 11-41. Metal orthosis constructed out of bar stock metal uprights of steel or aluminum with Klenzak-type joint and medial T strap with posterior plantar flexion stop. Reprinted with permission from Lehmann JF. Lower limb orthotics. In: Redford JB, ed. *Orthotics Etcetera.* 3rd ed. Baltimore, Md: Williams & Wilkins; 1986.

dial malleolus and runs over the medial malleolus. It is cinched outside the AFO lateral upright to align the ankle with the shoe and foot below.

Posterior stops. The posterior stop substitutes for the weak ankle dorsiflexors during the swing phase to provide toe clearance. If there is an imbalance between the plantar flexors and the dorsiflexors, as with many conditions resulting from upper motor neuron lesions, a rigid pinstop may be needed to provide the force to pick up the toe against a strong spastic calf. If there is a flaccid paralysis, as happens with peripheral nerve injuries, a spring stop or spring wire brace may be adequate. Any of these posterior plantar flexion stops or springs supporting the toe during swing will produce more resistance to plantar flexion than the normal lengthening contraction of the dorsiflexors when the foot moves from heelstrike to the footflat position. Therefore, they all produce an increased bending moment at the knee during the heelstrike and early stance that the patient has to overcome by voluntary knee extensor musculature. The bending moment is caused when the action line of the ground reaction force (Figure 11-42) falls behind the knee and thus creates the bending moment. The ground reactive force line starts from the center of pressure, that is, the heel contact against the ground, while the patient rocks over the posterior extremity of the heel during the heelstrike phase. The amount and duration of bending moment at the knee depends on the amount of force in resistance to plantar flexion. Therefore, the greatest bending moment is created by the pinstop, rather than by spring assists.[32-34]

The duration and magnitude of the bending moment will also depend on the angle at which the pinstop engages the flange of the stirrup and stops plantar flexion. The more the foot is stopped in dorsiflexion, the greater the bending moment and its duration. Therefore, whenever a spring assist is adequate, a pinstop should not be used because the bending moment at the knee would be increased and would require more extensor muscle force to keep the knee from buckling. However, in special cases, where the gastrocnemius-soleus drives the foot into forceful plantar flexion, the pinstop must be used even though the voluntary knee extensor force may be marginal. In these cases, the adjustment of the stop will be changed to allow the foot to be stopped in more plantar flexion. This will reduce the bending moment at the knee and its duration during heelstrike. However, if the posterior stop is adjusted to allow enough plantar flexion so that the bending moment at the knee is reduced to keep the knee stable, it is possible that this angle will not allow adequate toe clearance during the swing phase; therefore, another method must be used. The posterior pinstop will be adjusted to stop the ankle in more dorsiflexion, allowing adequate toe clearance. In this case, the orthosis can be modi-

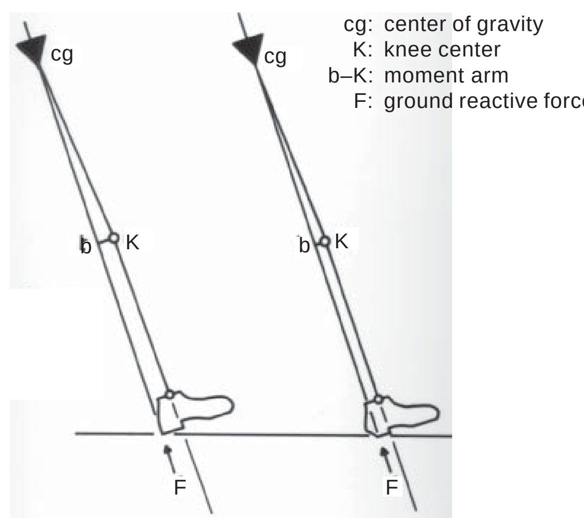


Fig. 11-42. Reduction of knee bending moment during heelstrike phase by heel cutoff. Reprinted with permission from Lehmann JF. The biomechanics of ankle foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

fied either by cutting off the posterior aspect of the heel at 45° (see Figure 11-42) or by inserting a cushion wedge into the heel. In both cases the location of the ground reactive force is moved forward. Thus, the extension of the force line comes closer to the knee, the moment arm gets smaller, and the bending moment is reduced so that the patient, in spite of marginal extensor musculature, can complete the heel strike phase without buckling and can adequately clear the toe during the swing phase of gait.

Anterior stops. The anterior (dorsiflexion) stop should use only a pin; no spring assist will be strong enough to assist with pushoff.[30] The dorsiflexion stop is used in combination with the sole plate riveted to the stirrup and extending rigidly to the metatarsal head area (Figure 11-43). As the center of gravity of the body moves forward, the ground reactive force line falls in front of the ankle. Dorsiflexion at the ankle is stopped; therefore, the foot pivots over the metatarsal head area, that is, the end of the sole plate, and raises the heel (Figure 11-44). This simulates pushoff by advancing the center of gravity of the body and pelvis upward and forward. As a result, the ground reactive force line is in front of the knee joint with the moment arm also in front (Figure 11-45). Therefore, a stabilizing extension moment is created in the latter part of the stance. This moment is of greater magnitude and longer duration if the dorsiflexion stop is adjusted in more plantar flexion. At the same time, the toe clearance is less.[31,32] Therefore, there is a trade-off between knee stability and toe clearance.

The extent to which these orthoses can restore the normal gait pattern and pushoff is shown in the example using the dorsiflexion stop with sole plate to restore pushoff in case of tibial nerve paralysis.[35] The normal timing of gait events is shown in Figure 11-46. The changes after a temporary tibial nerve block paralyzing the plantar flexors show that the heel comes off too late, lengthens the midstance, and shortens the pushoff phase. Restoration of normal timing of heeloff can be produced by an AFO with the anterior dorsiflexion stop adjusted to 5° of plantar flexion. The restoration is less complete with the orthosis adjusted to 5° of dorsiflexion. Figure 11-47 shows the normal progression of the center of pressure in relation to the ankle (black triangle curve). The center of pressure moves from behind the ankle, through the ankle, and then rapidly forward, in front of the ankle. This rapid forward movement is due to the gastrocnemius-soleus action resisting further dorsiflexion as the center of gravity of the body moves forward. Figure 11-47 shows that, in

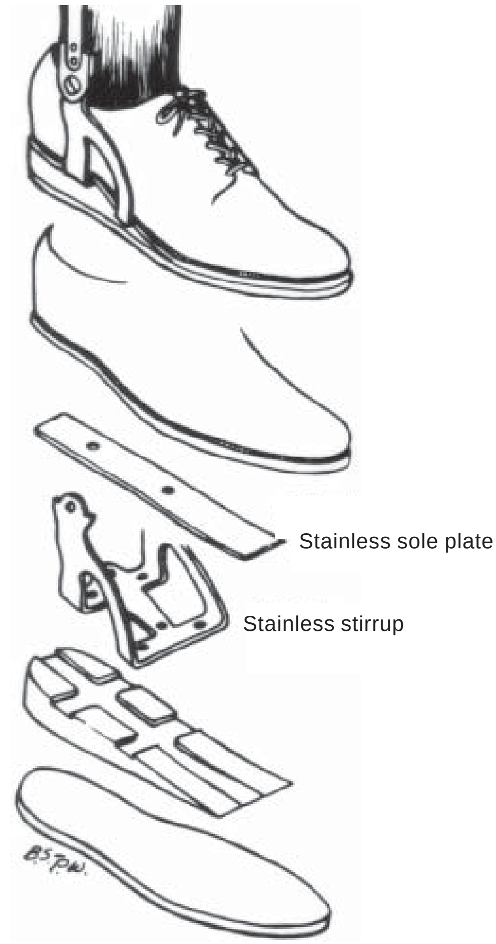


Fig. 11-43. Exploded view of stirrup, sole plate and shoe construction. Reprinted with permission from Lehmann JF, de Lateur BJ, Warren CG, Simons BC, Guy AW. Biomechanical evaluation of braces for paraplegics. *Arch Phys Med Rehabil.* 1969;50:179-188.

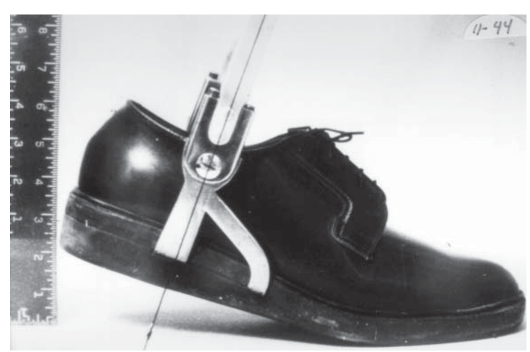


Fig. 11-44. Heel rise of brace with posterior and anterior stop and rigid sole plate. Reprinted with permission from Lehmann JF, de Lateur BJ, Warren CG, Simons BC, Guy AW. Biomechanical evaluation of braces for paraplegics. *Arch Phys Med Rehabil.* 1969;50:179-188.

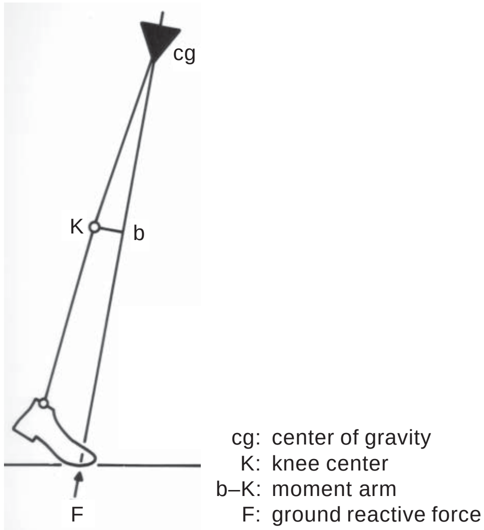


Fig. 11-45. Knee extension moment during pushoff. Reprinted with permission from Lehmann JF. The biomechanics of ankle foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

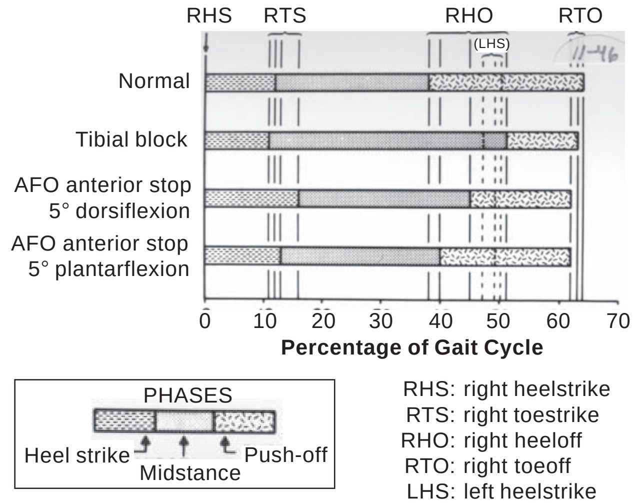


Fig. 11-46. Mean timing of gait events for six subjects. Reprinted with permission from Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Ankle-foot orthoses: Effect on gait abnormalities in tibial nerve paralysis. *Arch Phys Med Rehabil.* 1985;66:212-218.

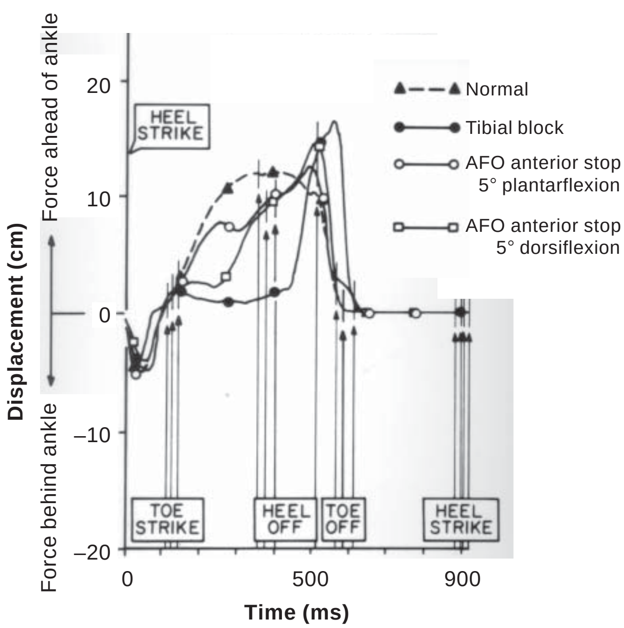


Fig. 11-47. Vertical force moment arm with respect to the ankle versus time. Curves represent the mean of six trials of one subject. Reprinted with permission from Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Ankle-foot orthoses: Effect on gait abnormalities in tibial nerve paralysis. *Arch Phys Med Rehabil.* 1985;66:212-218.

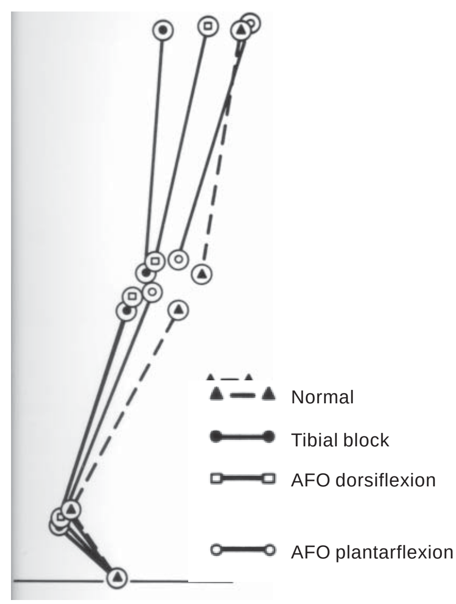


Fig. 11-48. Mean position of right lower limb at time of left heelstrike. Stick figures represent the means of six trials of one subject. Circles represent markers placed on the greater trochanter, lateral epicondyle, fibular head, lateral, malleolus, and fifth metatarsal head. Reprinted with permission from Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Ankle-foot orthoses: Effect on gait abnormalities in tibial nerve paralysis. *Arch Phys Med Rehabil.* 1985;66:212-218.

the absence of gastrocnemius-soleus action, the patient holds back the center of gravity of the body. This is done to prevent the center of pressure and the force line from moving ahead of the ankle and producing unstable dorsiflexion (black dot curve). Forward movement of the center of pressure occurs only after the weight bearing has started on the opposite extremity. The open circles in Figure 11-47 show significant improvement when the adjustment of the dorsiflexion stop is at 5° of plantar flexion, with less improvement when the adjustment of the dorsiflexion stop is at 5° of dorsiflexion (open squares). Because pushoff is impossible with the paralysis of the tibial nerve and without braces, the forward movement of the center of gravity and the hip does not occur (Figure 11-48).

Full restoration of the forward movement occurs with the adjustment of the dorsiflexion stop at 5° of plantar flexion; partial restoration with adjustment at 5° of dorsiflexion. As a result of the lack of advancement of the pelvis and the lack of pushoff with tibial block,[35] the step length on the opposite side is reduced as compared with normal gait. The mean step length (in meters) during four conditions for the left (unblocked) side is shown below:

1. Normal (0.71)
2. Tibial block (0.55)
3. AFO 5° DF (0.66)
4. AFO 5° PF (0.63)

Conditions had a significant effect on step length ($p \leq 0.001$). Step length for the normal condition and for both AFO conditions was significantly greater than step length for the block condition. The normal condition step length is greatly improved by the AFO's dorsiflexion stop. The influence on knee stability by the nerve block is shown in Figure 11-49. Because the center of gravity of the body and, therefore, the ground reactive force line is held back at the ankle, the knee center moves well forward on the force line. This creates a large moment arm and a large bending (flexion) moment at the knee during stance. The bending moment is so great it requires considerable knee extensor force to prevent the knee from buckling. This is optimally corrected by the adjustment of the dorsiflexion stop to 5° of plantar flexion and also somewhat improved by the adjustment to 5° of dorsiflexion (see Figure 11-49). While this example shows that better correction of the paralyzed posterior calf can be achieved by using a dorsiflexion pinstop in combination with sole plate to the metatarsal head area, adjusted to 5° of plantar flexion, there is still a tradeoff with toe clearance during the swing phase. In this case, a compromise must be achieved.

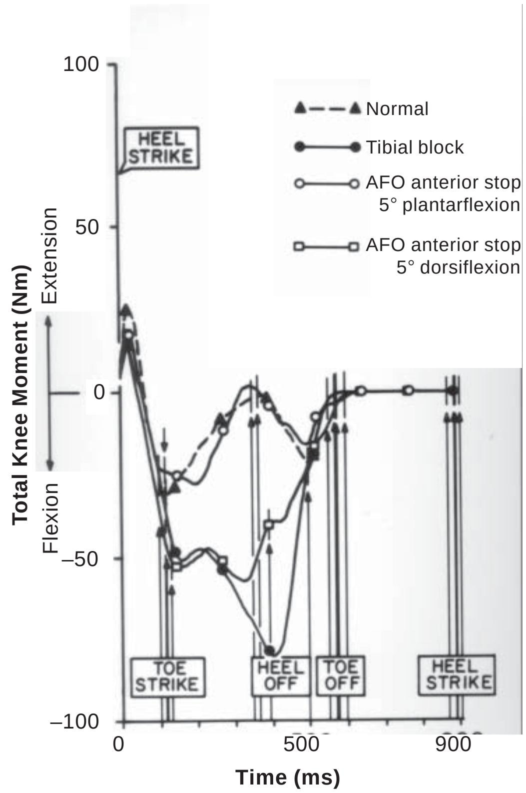


Fig. 11-49. Total knee moment versus time. Curves represent the mean of six trials of one subject. Reprinted with permission from Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Ankle-foot orthoses: Effect on gait abnormalities in tibial nerve paralysis. *Arch Phys Med Rehabil.* 1985;66:212-218.

Design and Suitability

To understand the biomechanical function of any AFO, the design must be studied to determine its suitability for a given patient. In addition, a manual test of the maximum plantar flexion and dorsiflexion force the orthosis can resist should be conducted to decide whether or not the orthosis is adequate to control the forces created by the individual patient. It is always desirable to use the correction with minimal force, since over-bracing usually makes it more difficult for the patient to walk and maintain knee stability. With this caveat in mind, the biomechanical function of two brace designs will be discussed.

A plastic ankle-foot orthosis (PAFO)[36,37] has been developed using lamination techniques (Figure 11-50). The rigid ankle of this orthosis is equivalent to a double pinstop ankle joint; the rigid extension of the plastic to the metatarsal head area is equivalent to the metal sole plate. The orthosis is highly ac-

Fig. 11-50. Plastic laminated (Seattle) orthosis. Reprinted with permission from Lehmann JF. Biomechanics of ankle-foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

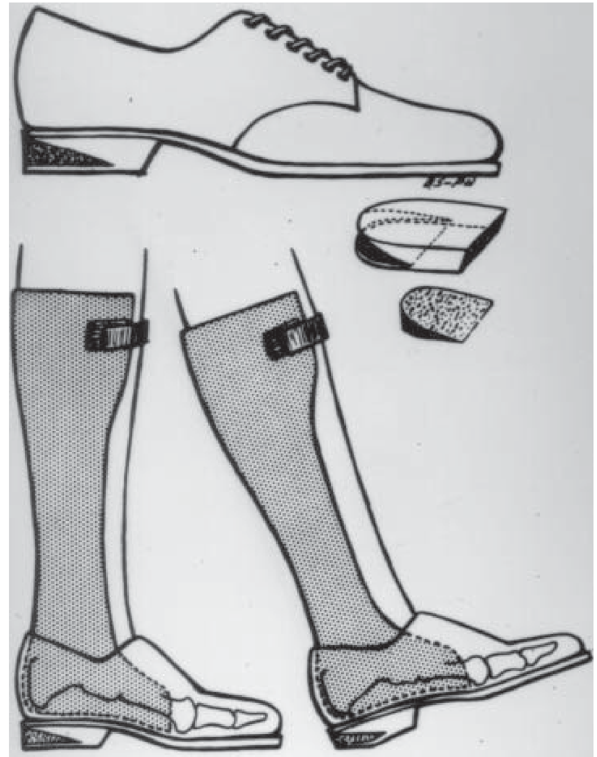


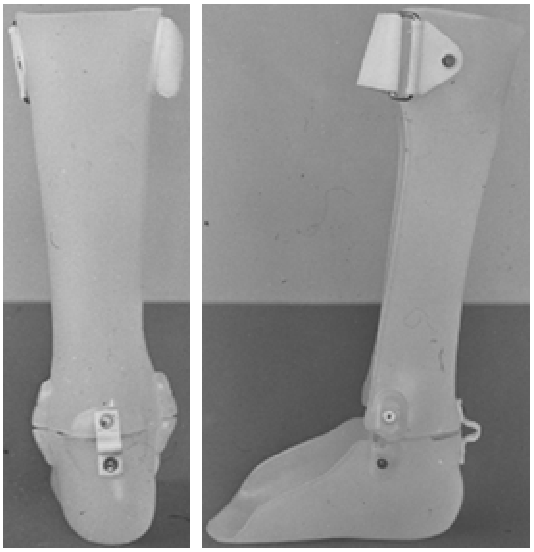
Fig. 11-51. Plastic laminated orthosis with heel cushion wedge, showing extension to metatarsal head area. Reprinted with permission from Lehmann JF. Biomechanics of ankle-foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

ceptable cosmetically (see Figure 11-50) and can be hidden under the stocking. However, it is critical that the ankle is fixed at the correct angle because of the tradeoff between knee stability and toe clearance. Because the trim lines (Figure 11-51) are in front of the ankle, the ankle is rigidly immobilized to provide maximum mediolateral stability. Shoes can be changed with this orthosis as long as the heel and sole height remain the same. The bending moment at the knee during early stance can be minimized by putting a cushion wedge into the heel or by a 45° cutoff of the heel (see Figure 11-51)

The manufacturing of this PAFO requires that a plaster cast be made with the correct angle at the ankle, with consideration given to the heel and sole height of the shoe. After this cast is removed, it is filled with plaster of Paris to produce a positive mold over which the plastic orthosis is laminated and then trimmed. If there is any mistake in casting, inversion, eversion, plantar flexion or dorsiflexion of the foot cannot be corrected, and the entire process has to be repeated. The use of wedges for correction of the ankle alignment typically results in an additional source of error.

This orthotic design has been advocated by Rancho Los Amigos Orthotic Department, but using heat-formed plastics with vacuum-molding techniques. Because the heat-formed plastics are thinned out most over the ankle area, the orthosis may become too soft to resist dorsiflexion and "open up" or "give way" at the ankle during pushoff. This does not provide adequate substitution for the forward and upward movement of the pelvis. To correct this problem, a thicker sheet of plastic should be used, or a boomerang-shaped carbon composite with beveled edges added over the positive mold.[38] This is so the hot plastic will flow over the beveled edges and incorporate these carbon composite pieces flush with the orthosis, thus providing a rigid equivalent to the dorsiflexion stop.

Whereas the PAFO described above is equivalent to a metal brace with an anterior and posterior rigid pinstop and a sole plate to the metatarsal head area, and does not allow any ankle motion, an articulated PAFO has been designed with functional adjustability of these stops. This orthosis consists of a separate portion for the foot and for the shank, connected by a hinge joint (Figures 11-52 and 11-53).[39] The angle at which dorsiflexion is stopped can be adjusted by the length of the posterior strap between the upper and lower parts of the orthosis.



Figs. 11-52 and 11-53. (**11-52,** left) Plastic articulated ankle-foot orthosis (AFO), rear view. (**11-53,** right) Plastic articulated AFO, medial view. **52** and **53:** Reprinted with permission from Lehmann JF, de Lateur BJ, Price R. Ankle-foot orthoses paresis and paralysis. *Phys Med Rehabil Clin North Am.* 1992;3:139-159.

The plantar flexion stop equivalent is provided by abutting the thickened edges of the upper portion and the lower portion posteriorly, so that they engage on plantar flexion. Trimming these edges produces more plantar flexion. Mediolateral stability is almost the same as in the other models of PAFOs.

Evaluation of AFOs by Biomechanical Principles

To evaluate an AFO, the sole and heel of the foot should be flat on the floor in quiet standing and during midstance. That it is flat can be best tested by using a thin piece of cardboard pushed between the sole of the shoe and the floor from the front, the back, and the lateral and medial sides. If the brace is tilted, the cardboard will move in too far from one or the other side. This fault can be corrected by bending the uprights or adjusting the angle at the ankle by the pinstops. In the PAFO, recasting is necessary. During the early stance (heelstrike phase), the knee should be observed for any tendency to buckle. If there is instability at the knee, the plantar flexion pinstop may be replaced by a spring, as in flaccid paralysis. If the pinstop is needed, it may be adjusted in too much dorsiflexion to create the bending moment at the knee.

During the latter part of the stance (pushoff), the knee should be observed for any tendency toward hyperextension. If that is the case, the pinstop should be readjusted at less plantar flexion or more dorsiflexion. During the swing phase, toe clearance should be checked for adequacy, and the stops readjusted if necessary.

In any orthosis that allows some plantar flexion and dorsiflexion, pistoning of the orthosis on the leg will occur unless the axis of the anatomical ankle (talocrural joint) is in the same location as the axis of motion in the orthosis. Therefore, it is desirable to line up the axis of the ankle joint of the double upright metal orthosis to coincide with the axis of the anatomical ankle, identified best by connecting the two tips of the malleoli (Figure 11-54).[40] In PAFOs with motion at the ankle, pistoning may be unavoidable except in the VAPC shoe clasp orthosis (Figures 11-55 and 11-56) where the pistoning of the posterior plastic rod is absorbed in a sleeve of the calf band.

In conclusion, the plantar flexion stop should use the minimum force required to prevent plantar flex-

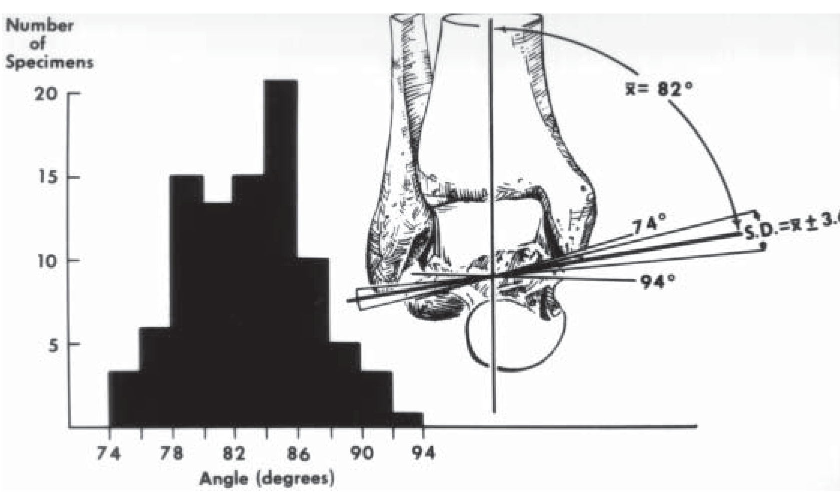


Fig. 11-54. Variations in angle between midline of tibia and empirical axis of ankle. The histogram reveals a considerable spread of individual values. Reprinted with permission from from Inman VT, Ralston HJ, Todd F. *Human Walking*. Baltimore, Md: Williams & Wilkins; 1981.

Fig. 11-55. The VAPC shoe clasp orthosis. Reprinted with permission from Lehmann JF, de Lateur BJ. Lower extremity orthotics. In: Kottke FJ, Lehmann JF, eds. *Krusen's Handbook of Physical Medicine and Rehabilitation.* 4th ed. Philadelphia, Pa: WB Saunders; 1990.

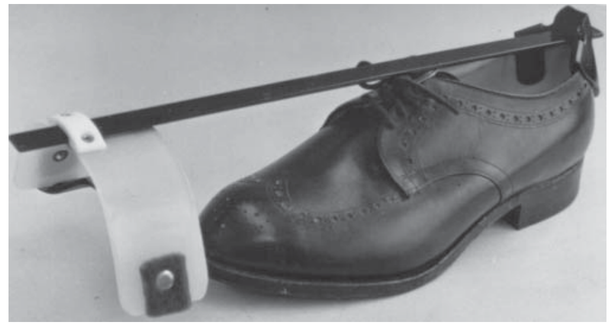


Fig. 11-56. The VAPC shoe clasp orthosis showing no resistance to dorsiflexion. Reprinted with permission from Lehmann JF, de Lateur BJ. Lower extremity orthotics. In: Kottke FJ, Lehmann JF, eds. *Krusen's Handbook of Physical Medicine and Rehabilitation.* 4th ed. Philadelphia, Pa: WB Saunders; 1990.

ion, which results in toe drag. To minimize the bending moment at the knee that must be overcome with a voluntary effort, if adequate, a spring should be used. If a plantar flexion stop (posterior stop) is needed to provide the adequate forces against the plantar flexion during swing, a tradeoff exists between the amount of toe pickup and knee instability during the early stance at heelstrike. The less the stop is fixed in dorsiflexion and the more it is fixed in plantar flexion, the greater the knee stability at heelstrike. In the clinical setting, posterior stops should be used to provide minimum yet safe toe clearance during the swing phase and at the same time avoid unnecessary knee instability during heelstrike. Also, the anterior stops (dorsiflexion stops) should help support the knee during the latter part of the stance phase, especially in patients with marginal knee stability using the AFO.

Metabolic Requirements

Energy expenditure can be measured in terms of rate of energy consumption (cal/min/kg body weight) or in terms of efficiency of walking (cal/m walked/kg body weight). Table 11-1 gives the normal values.[41] In hemiplegic ambulation, the walking speed is markedly reduced. Therefore, the rate of energy consumption is close to the normal values and there is no excessive load placed on the cardiovascular and respiratory systems. On the other hand, the efficiency in terms of calories per meter walked per kilogram body weight is markedly reduced. The values are therefore increased (Table 11-2). As an example, a hemiparetic patient was studied walking without an orthosis, with a plastic orthosis, and with a double metal upright orthosis. Both orthoses had the same biomechanical functions. Therefore, the rate of energy consumption was the same in both cases. In addition, functional values, like the self-selected walking speed and the maximum walking speed, were the same but improved over the performance without an orthosis.

The energy consumption for any achievable walking speed was higher as compared with the normal curve, but slightly less than walking without orthoses. The patient had to be spotted continuously because of falling and twisting the ankle without an orthosis. With either orthosis, the patient could walk safely.

Assessing Different Designs for Patients' Specialized Needs

The biomechanical function of any orthotic design can be easily evaluated initially by looking at it. As an example, the Teufel orthosis, shown in Figure 11-57, is obviously a posterior plastic leaf spring orthosis, which provides toe clearance during swing, but is not likely to produce significant me-

TABLE 11-1

ENERGY REQUIREMENTS IN NORMAL AMBULATION*

Researcher and Date	N	Type of Disability	Speed (m/min)	Energy Expenditure (kcal/min/kg)	(kcal • 10^{-3}/m/kg)
Ralston, 1958[1]	19	Normals (M&F)	74†	0.058*	0.78
McDonald, 1961[2]	583	Normals (F)	80	0.067*	0.83
		Normals (M)	80	0.061*	0.76
Peizer, 1969[3]	?	Normals (?)	80‡	0.043‡	0.57
Corcoran, 1970[4]	32	Normals (M&F)	83†§	0.063*	0.76
Waters, 1976[5]	25	Normals (M&F)	82§	0.063*	0.77

*Calculated knowing kcal/meter, m/min, and weight.
†Most efficient speed of ambulation.
‡Approximated from a graph.
§Speed chosen by the subjects.
M: male
F: female
Adapted with permission from Fisher SV, Gullickson G: Energy cost of ambulation in health and disability: A literature review. *Arch Phys Med Rehabil*. 1978;59:125. Sources: (1) Ralston HJ. Energy-speed relation and optimal speed during level walking. *Int Z Angew Physiol einschl Arbeitsphysiol*. 1958;17:277-283. (2) McDonald I. Statistical studies of recorded energy expenditure of man. Part II, Expenditure on walking related to weight, sex, age, height, speed and gradient. *Nutr Abstr Rev*. 1961;31:739-762. (3) Peizer E, Wright DW, Mason C. Human locomotion. *Bull Prosthet Res*. 1969;10-12:48-105. (4) Corcoran PJ, Brengelmann GL. Oxygen uptake in normal and handicapped subjects, in relation to speed of walking beside velocity-controlled cart. *Arch Phys Med Rehabil.* 1970;51:78-87. (5) Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking of amputees: Influence of level of amputation. *J Bone Joint Surg Am*. 1976;58:42-46.

TABLE 11-2

HEMIPLEGIC AMBULATION*

Researcher and Date	N	Type of Disability and Appliance	Speed (m/min)	Energy Expenditure kcal/min/kg	kcal • 10^{-3}/m/kg
Bard, 1963[1]	15	Hemiplegics	41*	0.044†	1.06
Corcoran, 1970[2]	15	Hemiplegics - no brace	42*	0.062	1.49‡
	15	Hemiplegics with plastic brace	49*	0.067	1.37‡
	15	Hemiplegics with metal base	49*	0.067	1.37‡

*Speed chosen by the subjects.
†Calculated knowing kcal/meter and m/min.
‡Calculated knowing kcal/min and m/min.
Reprinted with permission from Fisher SD, Gullickson G. Energy cost of ambulation in health and disability: A literature review. *Arch Phys Med Rehabil*. 1978;59:130. Sources: (1) Bard B. Energy expenditure of hemiplegic subjects during walking. *Arch Phys Med Rehabil*. 1963;44:368-370. (2) Corcoran PJ, Jebsen RH, Brengelmann GL, Simons BC. Effects of plastic and metal leg braces on speed and energy cost of hemiparetic ambulation. *Arch Phys Med Rehabil.* 1970;51:69-77.

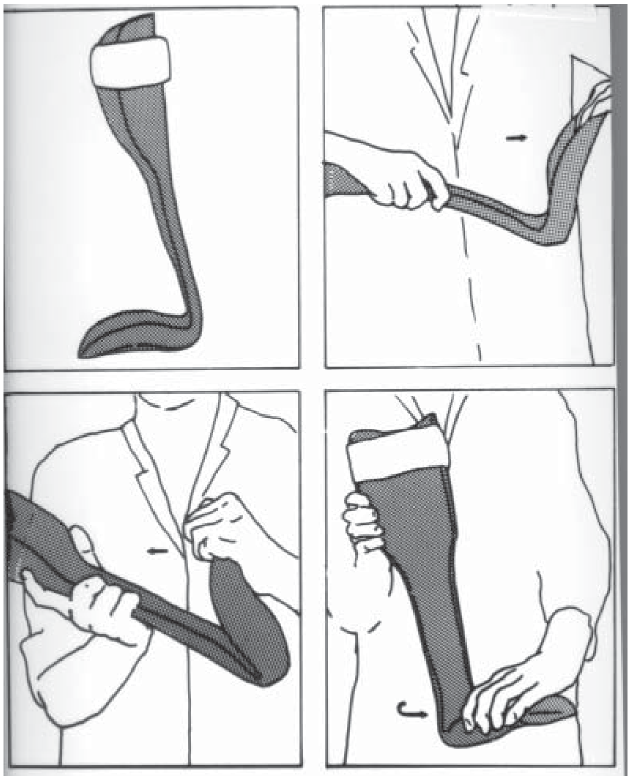


Fig. 11-57. The Teufel orthosis (top left); showing mediolateral twisting (top right); showing manual deformation into dorsiflexion (bottom left); showing manual deformation into plantar flexion (bottom right). Reprinted with permission from Lehmann JF. Biomechanics of ankle-foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

diolateral stability because it does not enclose the malleoli. This can be verified by twisting the orthosis to simulate the inversion or eversion of the foot, which the orthosis does not resist (see Figure 11-57, bottom right). Tests can measure the amount of force the orthosis provides against a calf that is driving the foot into equinus, as shown in Figure 11-57, top right. The lack of resistance to dorsiflexion demonstrates it is not a pushoff substitute (see Figure 11-57, bottom left). Figure 11-57 also shows that the axis of motion occurs in the orthosis at the posterior aspect of the heel and therefore some pistoning is unavoidable. A similar leaf spring design, the VAPC shoe clasp orthosis (see Figures 11-55 and 11-56), has essentially the same limitations, but it absorbs the pistoning motion by the insertion of a posterior plastic spring into a sleeve of the calf band. By pushing the orthosis into plantar flexion, an assessment can also be made of the force with which it will resist a spastic calf that is driving the foot into equinus position. Other orthoses can be tested as described in these examples, and based on this information, a customized orthosis can be prescribed for the individual.

Functional Electrical Stimulation of Peroneal Nerve

FES has been used in hemiparetic patients in an attempt to replace the AFO. Functional electrical stimulation was introduced by Liberson and coworkers[42] and was applied to the peroneal nerve in the area of the fibular neck to produce dorsiflexion and eversion during the swing phase of gait. Initially, skin electrodes were used, and the phasing of stimulation was controlled by switches incorporated in the shoe. Pulse duration of the current was between 20 and 33 microseconds with a repetition rate of 24 to 100 Hz, and intensity between 90 and 200 mA.

While this type of FES is rarely prescribed for permanent use, it is an extremely helpful tool when used during the recovery phase, for reeducation of the patient with hemiparesis resulting from brain injuries and stroke, and especially with patients who have paresis associated with decreased sensory feedback. While it produces a more normal gait pattern, it also produces a significant enhancement of sensory and proprioceptive feedback, including feedback from joint position sense and muscle contraction. If the proper level of stimulation at the peroneal nerve is used, it may produce not only dorsiflexion and eversion of the foot but also knee flexion and hip flexion during swing. For this purpose the stimulation level usually has to be increased to the nociceptive level. The whole system could be modified by implanting a receiving antenna under the skin of the anteromedial aspect of the thigh (Figure 11-58); a stimulus would be conducted to a surgically implanted electrode placed around the peroneal nerve. Stimulation would be activated by a transmitter antenna placed over the same area outside the skin where the receptor antenna was implanted. Through telemetry, the transmitter antenna receives the properly timed stimulatory signal, which is controlled by a transmitter, that in turn receives the signal from shoe switches. The transmitter and the battery pack are carried on a waist belt. The FES system was intended for long-term use. However, from the beginning, the indications for successful use of the system were fairly

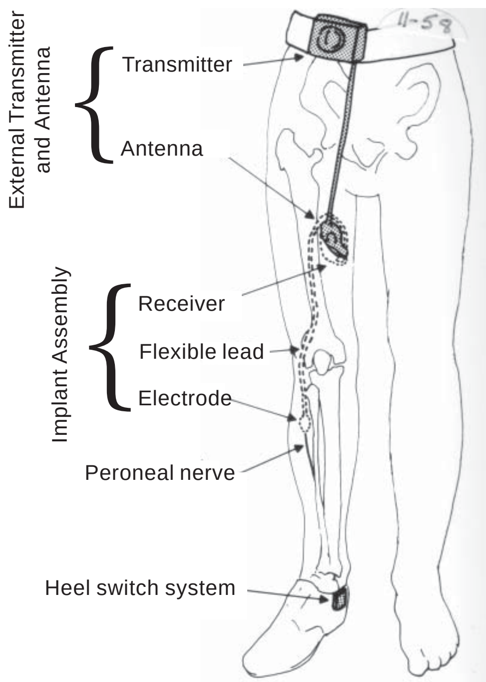


Fig. 11-58. Relative location of neuromuscular assist equipment on a patient with right side hemiplegia. Reprinted with permission from Lehmann JF. Lower limb orthotics. In: Redford JB, ed. *Orthotics Etcetera*. 3rd ed. Baltimore, Md: Williams & Wilkins; 1986; Redrawn with permission from Waters RL, McNeal D, Perry J. Experimental correction of foot drop by electrical stimulation of the peroneal nerve. *Arch Phys Med Rehabil.* 1975;53:276-281.

restrictive. Waters and Miller[43] found in their successful patients using FES that the patients had to walk faster than 25 m/min without any orthosis and needed good balance; and the system worked best for gait problems caused by foot drop.

One advantage of any electrical stimulation is that reciprocal inhibition of the gastrocnemius and soleus is produced when the peroneal nerve is stimulated. Significant problems with the FES system are that it must be surgically implanted, and if the wires break, repeat surgery will be necessary. Also, most patients prefer a simple PAFO that serves well and has fewer problems.

Knee-Ankle-Foot Orthoses

Knee-ankle-foot orthoses are commonly used for conditions caused by upper and lower motor neuron lesions.[44] Bilateral application is common in SCIs that produce upper motor neuron lesion, which cause spasticity, and lower motor neuron lesions of the cauda equina, which cause flaccidity. Unilateral use is more common when weakness of the quadriceps is combined with weakness around the ankle. The main function these orthoses provide, beyond those the AFOs provide, is unequivocal knee stability.[33,45] The reasons for application of these orthoses are to provide mediolateral stability at the ankle (as in AFOs); simulate pushoff by provision of an anterior dorsiflexion pinstop in combination with the sole plate to the metatarsal head area; and provide, during the swing phase, toe pickup by use of a plantar flexion stop at the ankle (as in AFOs).

Standard Components

The standard KAFO is constructed with double metal uprights, a knee joint (usually lockable with a bail), and a double-stopped ankle joint connected to a stirrup and sole plate at the metatarsal head area. The uprights are held together by padded rigid upper and lower posterior thigh bands and the calf band (Figure 11-59).[44] The relative depth of the upper thigh band to the lower thigh band will determine the position of the thigh and, therefore, the position of the knee in the orthosis. A deep upper thigh band with a shallow lower thigh band positions the thigh and knee in slight flexion, while a relatively shallow upper thigh band with a deep lower thigh band produces an extension of the knee. Tight and shallow refer to the distance of the posterior aspect of the rigid bands from the uprights.

Six common variations of the KAFO are used.[44] The orthoses vary significantly as to where the knee stabilizing force in front is applied. The corresponding counter forces are at the upper thigh band above and at the shoe below. Three brace configurations (see Figure 11-59) apply the stabilizing force through two straps to prevent the knee from bending in the brace. The first one uses the combination of a suprapatellar and patellar tendon strap, where both straps apply forces to very tolerant areas: quadriceps above the patella and the patellar tendon below. The second orthosis uses a soft closure of the lower thigh band in combination with the soft closure of the calf band. And, the third orthosis uses a combination of the lower thigh band closure and the patellar tendon strap. These three models have the advantage of distributing the knee stabilizing force over two straps, reducing the force per square centimeter on the skin.

The remaining three configurations use only one strap to apply the knee stabilization force: (1) with a suprapatellar strap, (2) with the patellar tendon

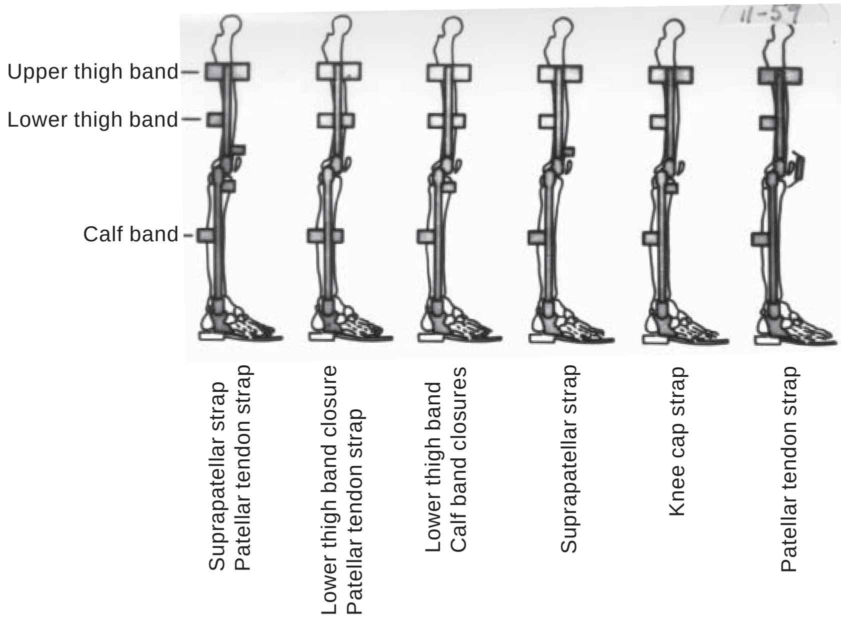


Fig. 11-59. Knee-ankle-foot orthoses: six common configurations. Reprinted with permission from Lehmann JF, Warren CG. Restraining forces in various designs of knee-ankle orthoses: Their placement and effect on the anatomical knee joint. *Arch Phys Med Rehabil.* 1976;57:430-442.

strap, and (3) with the knee cap strap, or spider. Note that the suprapatellar and the knee cap straps apply a single stabilization force above the knee joint. In the knee cap strap the force is applied to the patella, and the patella, in turn, applies the force through the patellarfemoral joint to the femur above the knee. All modifications of this orthosis have a soft closure of the upper thigh band in front.

These six configurations have been evaluated biomechanically, as to their effectiveness; a major objective of this evaluation was the reduction of the force applied to the skin by the knee stabilization bands. If the total stabilization force was measured irrespective as to whether it was applied by one or two straps, it was found that the highest force application was produced by the use of the lower thigh band and calf band closures because of the poor leverage of the forces applied far away from the knee center. This demonstrates that the stabilization forces should be applied close to the knee center.

Measurements of the stabilization force that keeps the knee from buckling in the brace reveal that if a few degrees of knee flexion are allowed in the orthosis, the force is doubled. The knee position can be controlled by the proper depth of the upper and lower thigh bands. It is also important that the brace be properly applied. Often the brace is applied while the patient is sitting, so that the stabilization straps are cinched too loosely; as the patient stands up, the knee bends in the orthosis and increases the forces against the straps. Proper recinching of the straps is necessary when the patient returns to a seated position. This clearly demonstrates that the brace must be designed and applied to keep the knee straight.

If the force required to stabilize the knee is measured at each strap, it will be found that the lowest force across the surface area of skin is produced by the combination of the suprapatellar and patellar tendon straps. These straps are close to the knee, distribute the forces between two straps, and apply the forces to very tolerant areas. Therefore, the minimal force across surface area is provided by a combination of the suprapatellar strap and patellar tendon strap. Since the patellar tendon strap applies the force to an extremely tolerant area, it also can be used alone, provided the strap is not placed over the shin.

A study of the shear forces acting on the knee ligaments in the orthosis in paraplegic swing-through ambulation showed that if an improper placement of the knee stabilizing straps was used these shear forces were excessive in amplitude and duration, with the femur shearing back on the tibia during the later part of the stance (Figures 11-60 through 11-62).[44] This shows that to avoid excessive knee shear, at least a major portion of the knee stabilization force should be applied below the knee.

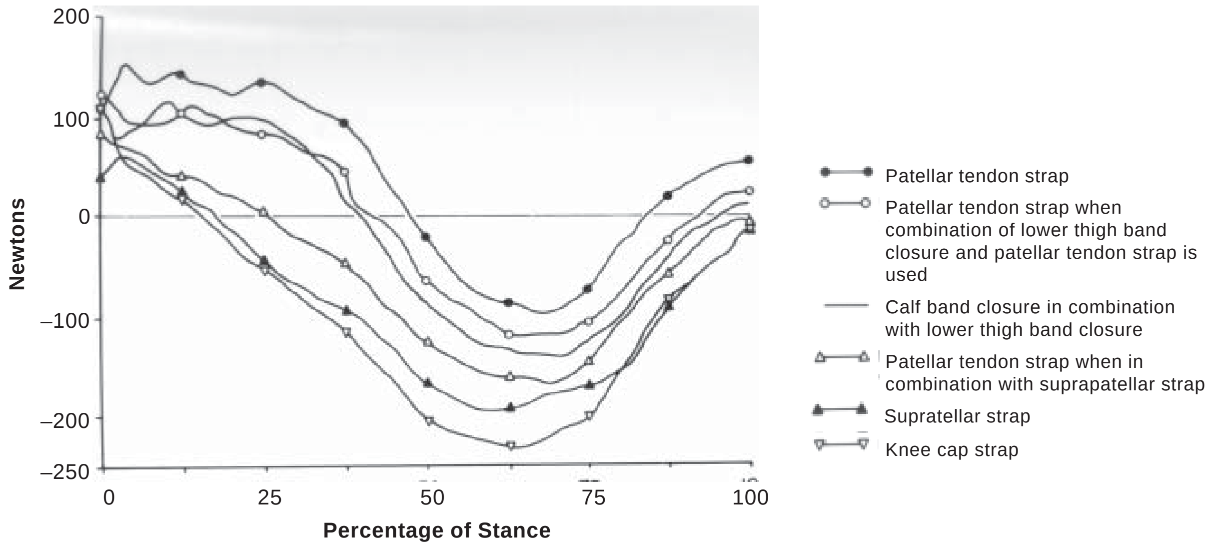


Fig. 11-60. Anatomical knee shear showing force interaction between femur and tibia. Positive values indicate the femur shearing forward on the tibia. Negative values indicate the femur shearing backward on the tibia. Reprinted with permission from Lehmann JF, Warren CG. Restraining forces in various designs of knee ankle orthoses: Their placement and effect on the anatomical knee joint. *Arch Phys Med Rehabil.* 1976;57:430-437.

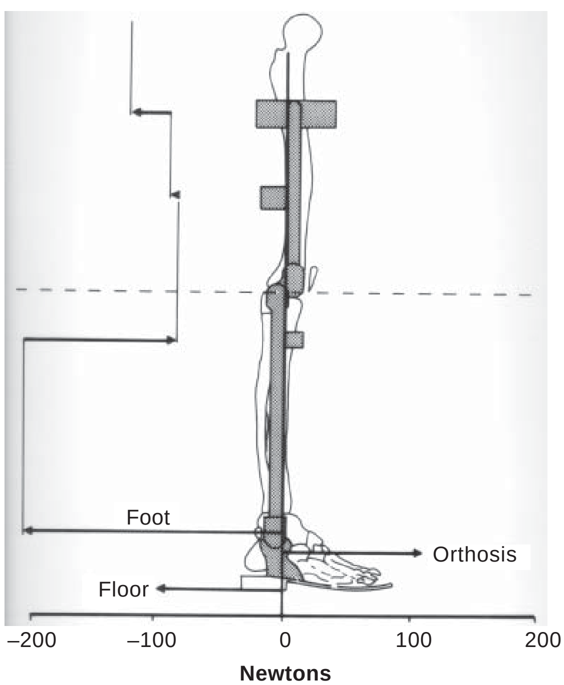


Fig. 11-61. Schematic representation of shear in the limb and orthosis during pushoff, the stabilizing strap below the knee. Reprinted with permission from Lehmann JF, Warren CG. Restraining forces in various designs of knee ankle orthoses: Their placement and effect on the anatomical knee joint. *Arch Phys Med Rehabil.* 1976;57:430-437.

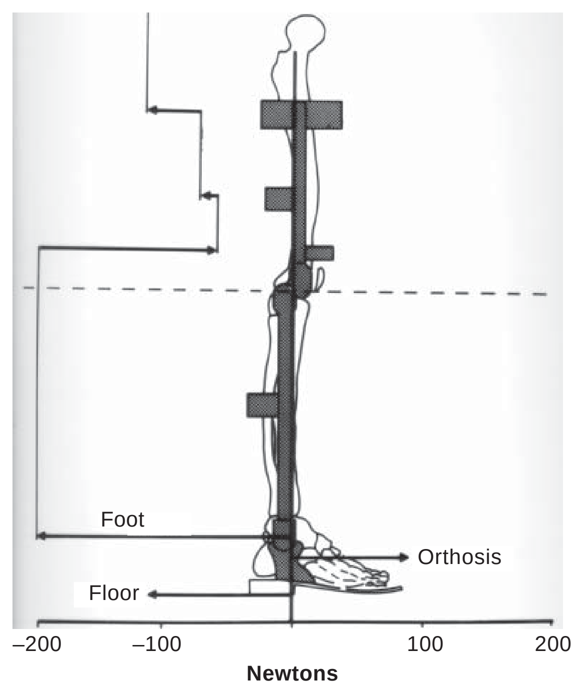


Fig. 11-62. Schematic representation of shear in limb and orthosis during pushoff with stabilizing force above the knee. Reprinted with permission from Lehmann JF, Warren CG. Restraining forces in various designs of knee ankle orthoses: Their placement and effect on the anatomical knee joint. *Arch Phys Med Rehabil.* 1976;57:430-437.

If the forces are measured against the edge of the upper thigh band when the orthosis is too long and does not allow enough clearance between the ischium and the upper thigh band with the hip extended, the forces during paraplegic swing-through ambulation are approximately 10 to 15 N/cm^3 (approximately 2,000 mm Hg). Therefore, abrasion and sores are produced because of this brace fault.[44] In conclusion, when the hip is fully extended, there should be at least two fingers width of clearance between the edge of the upper posterior thigh band and the ischium.

In summary, the orthotic designs of the biomechanically optimal KAFO should place knee restraining forces close to the knee, hold the patient's knee straight, distribute forces between straps, and apply the primary force below the knee. On hip extension, the upper edge of the posterior thigh band should adequately clear the ischium. In addition, the orthosis should be designed so that weight is reduced, donning and doffing are easy, free standing balance is possible even for a paraplegic patient with hands free, and energy expenditure is reduced.

Scott-Craig Orthosis

The Scott-Craig orthosis was designed to reduce the weight of the orthosis and to ease donning and doffing.[46,47] In this design, a posterior thigh band remains as the upper rigid connection between the two uprights of the orthosis. The lower posterior thigh band is eliminated. The bail lock at the knee is retained. The posterior calf band has been eliminated, but a rigid anterior shin piece not only connects the metal uprights below the knee but also supplies the knee stabilization force. The remainder of the orthotic design at the ankle and foot is the same as the standard orthosis. In evaluation studies[46] to determine if the rigidity of this design was adequate to handle stresses produced by paraplegic swing-through ambulation, it was found that a minimum of three rigid connections are required to hold the brace uprights together and prevent permanent deformation: the posterior upper thigh band above, the stirrup below, and one more rigid cross connection in the middle.

The Scott-Craig orthosis more than adequately fulfills the rigidity requirements; it also applies stabilization forces below the knee and, therefore, reduces knee shear. However, the original design of the knee stabilizing force should be modified to produce a plastic molded or padded piece that applies most of the force to the patellar tendon area in addition to the tibial condyles. This avoids development of skin sores in a vigorous walker, when only a rigid padded band over the shin is applied. Another modification required is the replacement of the eliminated posterior calf band by a soft closure to keep the orthosis from slipping from the conically shaped thigh, forward on the knee, when the patient sits down. With these modifications, the Scott-Craig orthosis improves donning and doffing and moderately reduces the weight of the orthosis.[46]

Metabolic Requirements

It is essential that metabolic requirements for patients using KAFOs, specifically for paraplegic or SCI persons, be reduced to a minimum. Paraplegic ambulation is limited because of the high rate of energy consumption when walking at a functional speed, compared to the mechanical ability of patients using a swing-through gait.[48-50] The features, which reduce energy consumption in any one of the orthoses already discussed, depend largely on the amount of substitution of pushoff the orthosis provides (see Figure 11-43).[51] The incorporation of a rigid anterior dorsiflexion pinstop at the ankle, in combination with the sole plate riveted to the stirrup and extending to the metatarsal head area, substitutes for pushoff. The standard orthosis and the Scott-Craig orthosis incorporate these features. During a swing-through gait, at the time when the two crutches are in front of the feet and the center of gravity moves forward, pivoting will occur at the end of the sole plate as dorsiflexion is stopped. Therefore, the heel rises and the center of gravity pathway curve is elevated at the lower end. As a result, the lift required for the clearance during swing is reduced.[46] Reduction of the amplitude of the center of gravity motion in the vertical direction saves approximately 30% of the mechanical work and metabolic requirements for ambulation. An orthosis not equipped with a dorsiflexion stop and sole plate does not provide substitution for pushoff and, therefore, does not save the 30% of energy consumption.

In conclusion, the addition to the KAFO of an anterior dorsiflexion pinstop in combination with the rigid sole plate to the metatarsal head area simulates pushoff, reduces vertical center of gravity excursions, and reduces mechanical work done and energy expenditure when the patient walks. This may prevent the patient from entering the anaerobic phase of metabolism, and, thus, increases walking endurance. Both the standard orthosis and the Scott-Craig orthosis provide this benefit.

TABLE 11-3

COMPARISON OF STANDING BALANCE

			Average Standing Balance/s		
Patient no.	**Gender**	**Level of lesion**	**Post. Stop (n=5)**	**Post. and Ant. Stop (n=5)**	**t Value (*p*)**
1	F	T12	1.55	300.0	1039.171 (0.0001)
2	F	T7-8	3.5	300.0	1228.081 (0.0001)
3	M	L2-3	3.6	246.0	8.602 (0.0001)
4	M	T4	1.58	8.9	1.218 (0.4872)
5	M	L3	40.25	80.15	1.640 (0.072)

Reprinted with permission from Lehmann JF, de Lateur BJ, Warren BS, Simons BC, Guy AW. Biomechanical evaluation of braces for paraplegics. *Arch Phys Med Rehabil.* 1969;50:187.

Standing Balance

The same KAFO features that produce substitution for the pushoff influence the standing balance. These are the anterior dorsiflexion pinstop in combination with the sole plate to the metatarsal head area, and the incorporation of a posterior pinstop at the ankle. In this brace configuration, an area of stable support exists within which the center of pressure may sway without imbalance of the patient. This support extends from the end of the sole plate in front to the posterior aspect of the heel, and from the side of one foot to the side of the other foot. In contrast, without the anterior dorsiflexion pinstop, the area of stable support extends only from the level of the ankle joint to the posterior aspect of the heels, which makes standing balance possible for only a few seconds rather than for minutes (Table 11-3).

The Use of Pelvic Bands with KAFOs

A pelvic band may be connected to the KAFO by a hip joint that may be locked through a drop lock. The pelvic band encircles the pelvis between the greater trochanter and the iliac crest. The front part of the pelvic band is rigid and padded, and the posterior part has a soft closure (Figure 11-63). Pelvic bands were expected to reduce excessive lumbar spine motion in ambulation that used a swing-through gait pattern. The band was also supposed to produce more standing stability, but a biomechanical functional evaluation has shown greater lumbar excursions with the pelvic band than without because of the restriction of motion at the hip for which the lumbar spine motion compensated. For the same reason, the mean stride length was reduced.[52] Due to the immobilization of the hip, a greater lift of the center of gravity of the body was required during swing. The time required for donning and doffing was almost doubled as compared with the standard KAFO. On the other hand, standing balance was slightly improved. Also, the uneven forward swing of the leg in paraplegic ambulation, which is caused by spasticity, was slightly improved. In most cases this could be overcome by training.

The pelvic band offers the slight advantage of controlling spastic lower extremities and improves standing balance. However, lumbar excursions and vertical excursions of the center of gravity are both increased. The step length is decreased. All these factors lead to an increased metabolic demand. Therefore, the pelvic band should be used only if there is a stringent reason for its application. A normal, swing-through gait pattern, where the patient has intact hip ligaments and no flexion contractures will allow the hip to be stable when the crutches are in front of the feet. In this position, gravitational forces extend the hip against the iliofemoral, pubofemoral, and ischiofemoral ligaments, around extension check ligaments, to stabilize the hip in extension. During the phase of the gait cycle where the crutches are behind the feet, arching of the back would place the center of gravity of the body above

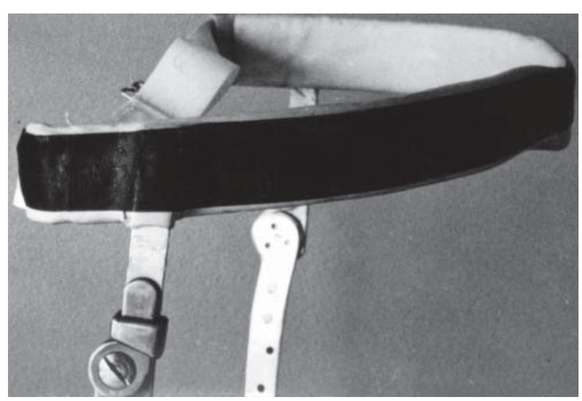


Fig. 11-63. The pelvic band. Reprinted with permission from Lehmann JF, de Lateur BJ, Price R. Knee-ankle-foot orthoses paresis and paralysis. *Phys Med Rehabil Clin North Am.* 1992;3:161-183.

TABLE 11-4

PARAPLEGIC AMBULATION

Researcher and Date	N	Type of Disability and Appliances	Speed (m/min)	Energy Expenditure kcal/min/kg	Energy Expenditure kcal • 10^{-3}/m/kg
Clinkingbeard, 1964[1]	4	Thoracic paraplegics	4	0.043	9.05
	3	Lumbar paraplegics	20	0.048	2.37
Gordon, 1956[2]	3	Thoracic paraplegics	27	0.090	2.44
	10	All paraplegics	27	0.086	2.32

Adapted with permission from Fisher SV, Gullickson G Jr. Energy cost of ambulation in health and disability: A literature review. *Arch Phys Med Rehabil.* 1978;59:130. Data sources: (1) Clinkingbeard JR, Gertsen JW, Hoehn D. Energy cost of ambulation in the traumatic paraplegic. *Am J Phys Med.* 1964;43:157-165. (2) Gordon EE, Vanderwalde H. Energy requirements in paraplegic ambulation. *Arch Phys Med Rehabil*. 1956;37:276-285.

and, therefore, the ground reactive force line behind the hip joint, thus extending it. This process locks the hip stably in extension against the ligaments. Therefore, there is no need for the pelvic band to stabilize the hip during the swing-through or swing-to gait patterns. In conclusion, the only advantages of the pelvic band would be some slightly better control of spastic lower extremities and improved standing balance.

Functional Use of Bilateral KAFOs in Paraplegia

The metabolic efficiency of paraplegic ambulation is poor (Table 11-4). To keep the rate of energy expenditure low enough to avoid entering the anaerobic phase of metabolism, which would reduce endurance, the paraplegic person considerably decreases walking speed.[43] Because the metabolic efficiency of walking also depends on the SCI level, it has been generally recommended that only persons with lower lumbar level lesions be provided with orthoses and trained for ambulation. A more functional way to look at this problem follows.

The demands for distance walked and speed of ambulation placed on the patient in his or her environment should be assessed irrespective of the level of the spinal cord lesion. If the paraplegic patient can cover the distance required and provide the speed needed, braces should be prescribed and training should be provided. For a long distance ambulator, the wheelchair should be considered since its metabolic requirements (Table 11-5) are less

TABLE 11-5

WHEELCHAIR LOCOMOTION*

Researcher and Date	N	Type of Disability and Appliances	Speed (m/min)	Energy Expenditure kcal/min/kg	Energy Expenditure kcal • 10^{-3}/m/kg
Hildebrandt, 1970[1]	30	Wheelchair-bound	67	0.037	0.47*
Glaser, 1975[2]	9	Normals	53	0.047†	0.89†
	9	Normals	70	0.056†	0.79†
	9	Normals	83	0.071†	0.84†

*Calculated knowing kcal/min and m/min
†Calculated from O_2 consumption
Reprinted with permission from Fisher SV, Gullickson G Jr. Energy cost of ambulation in health and disability: A literature review. *Arch Phys Med Rehabil.* 1978;59:131. Sources: (1) Hildebrandt G, Voigt ED, Bahn D, Berendes B, Kröger J. Energy costs of propelling wheelchair at various speeds: Cardiac response and effect on steering accuracy. *Arch Phys Med Rehabil.* 1970;51:131-136. (2) Glaser RM, Edwards M, Barr SA, Wilson GH. Energy cost and cardiorespiratory response to wheelchair ambulation and walking (abstract). *Fed Proc*. 1975;34:461.

than that of normal ambulation. For a high level of SCI, such as quadriplegia, even the use of a manual wheelchair may force the patient to use the anaerobic metabolism because of the small muscle mass involved in this activity. In those cases, an electric wheelchair should be considered.

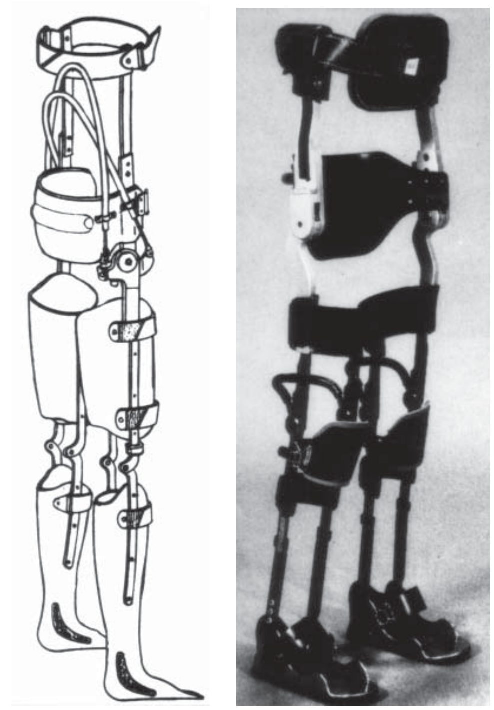
In conclusion, depending on the demands for ambulation placed on the SCI person by his or her environment and the ability to fulfill these requirements, the patient should be fitted and trained with orthoses or given a manual or electric wheelchair. This approach is in contrast to training for the use of braces and crutches solely based on the level of the spinal cord lesion.

Reciprocating Gait Orthosis

Usually, walking speed is significantly limited with the reciprocating gait orthosis (RGO). In a study by Merritt[53] using speeds from 17 to 41 m/min, it was found that the oxygen consumption per kilogram of body weight per meter walked rose rapidly at speeds above 27 m/min when using the alternating gait pattern. By comparison, the Scott-Craig orthosis was more energy efficient when used with the swing-through gait at the same speed as the RGO. If the RGO was used like the Scott-Craig with a swing-through gait pattern, the energy consumption was the same in both cases. Thus, subjects using RGOs with reciprocating gaits have higher energy expenditures (VO_2/min) and poorer energy efficiencies (VO_2/m) than when using standard orthoses with a swing-through gait. Therefore, long sustained ambulation at functional speeds is more difficult than with the standard or Scott-Craig orthosis.

The RGO is commonly used for patients with SCIs, or those with cauda equina lesions. The orthotic design for the RGO is shown in Figure 11-64. To keep the weight of this orthosis within tolerable limits, a greater amount of plastic material is used compared to, for example, the AFO and the thigh cuff. In addition, one of the main features is that low friction Bowden cables are used, which transfer power from hip extension on one side to hip flexion on the other side, or power from hip flexion into hip extension on the opposite side. Thus, a reciprocating gait pattern can be achieved provided the patient has some hip flexion or hip extension power available at hip musculature. The orthosis braces not only the hip but also the trunk. Locking mechanisms for hip and knee can be added.

The development of the RGO has been based on the design of the hip guidance orthosis, which allows limitations to be set for hip flexion-extension



Figs. 11-64 and 11-65. (**11-64**)The reciprocating gait orthosis. Photograph: Courtesy of Durr-Fillauer Medical, Inc., Chatanooga, Tennessee. (**11-65**) The ParaWalker. Photograph: Courtesy of ORLAU publishing. Oswestry, Shropshire, Great Britain.

and uses a rocking motion to get one leg in front of the other, thus producing a reciprocating gait pattern.[54,55] Originally, it was primarily designed for lower motor neuron lesions, such as meningomyelocoele and low level SCIs, that is, cauda equina lesions (Figure 11-65). Its modification for paraplegic persons is also known as the ParaWalker.[56]

Functional Electrical Stimulation for Bracing

FES of key muscle groups in proper sequence for walking has now been used in major SCI treatment centers.[57,58] The sequence is controlled either by the patient with switches, or by a microprocessor. For balance and stability, walkers or crutches are also used. The gait pattern is a reciprocal one. Common muscles stimulated are gluteus medius, gluteus minimus, gluteus maximus, iliacus, rectus femoris, vastus lateralis, vastus intermedius, and vastus medialis.[59] Both surface electrodes as well as transcutaneous wire electrodes are used. The lack of sensation in the SCI patient is an advantage, as the level of current used for stimulation may be unpleasant. Further research in this area is required because the speed of ambulation with this type of brace is very

TABLE 11-6

ENERGY COSTS OF AMBULATION WITH FUNCTIONAL NEUROMUSCULAR STIMULATION

Subject	Trial Number	Distance (m)	Speed (m/s)	O_2 Consumption (L/min)	Energy Expenditure (kcal/kg/min)	Energy Expenditure (kcal/kg/m)	O_2 Debt (L)
1	1	58.5	0.32	1.36	0.089	4.74×10^{-3}	1.89
1	2	30.5	0.35	1.43	0.094	4.43×10^{-3}	1.62
1	3	29.6	0.17	1.57	0.099	9.84×10^{-3}	2.30
1	4	30.5	0.39	1.42	0.094	4.05×10^{-3}	1.33
1*	5	28.7	0.11	1.38	0.091	13.40×10^{-3}	1.62
2	1	58.5	0.16	1.64	0.103	10.98×10^{-3}	1.59
2	2	30.2	0.15	1.39	0.088	9.53×10^{-3}	1.73
2	3	39.6	0.13	1.52	0.101	13.13×10^{-3}	1.79
2	4	30.5	0.16	1.49	0.099	10.13×10^{-3}	1.59
2*	5	30.2	0.16	1.46	0.093	9.66×10^{-3}	1.80
3	1	58.5	0.56	1.48	0.093	2.77×10^{-3}	2.90
			$\bar{x} = 0.24$ m/s		$\bar{x} = 0.095$ SD = 0.005		

*No hamstrings or gluteals; subject 1: $\bar{x}$= 0.093, SD = 0.003; subject 2; $\bar{x}$= 0.097, SD = 0.007
Reprinted with permission from Marsolais EB, Edwards BG. Energy costs of walking and standing with functional neuromuscular stimulation and long leg braces. *Arch Phys Med Rehabil*. 1988;69:243-249.

slow and the metabolic efficiency is very low (Table 11-6 and Figure 11-66).[60,61]

An effort is presently made in research centers to overcome the slow walking speed and the high energy consumption of FES by combining this method with some of the commonly used orthoses, including the RGO. In conclusion, because of the low walking speed and the high energy cost of ambulation by electrical stimulation, the use of this method of ambulation is currently limited.

Orthoses for Skeletal Insufficiency

Orthoses for skeletal insufficiency can be used for early mobilization of patients with fractures and to allow conservative management of common joint deformities that result from ligamentous injuries. They should, however, be considered as a supplement or an alternative to a surgical approach.

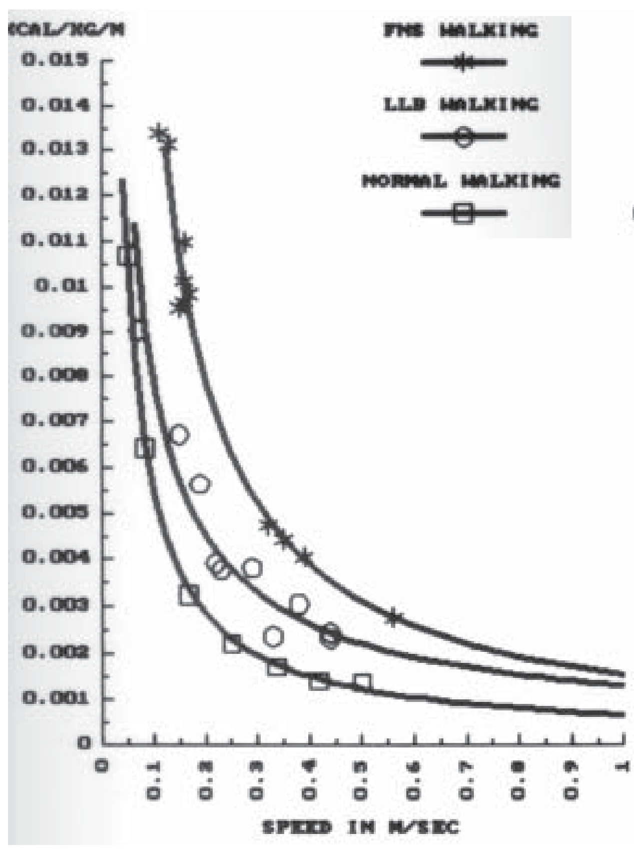


Fig. 11-66. Energy efficiency of paraplegic persons walking with functional nerve stimulation and with long leg braces (LLB) compared with normal walking. Functional nerve stimulation data from Marsolais EB, Edwards BG. *Arch Phys Med Rehabil*. 1988;69:243-249; long leg brace data from the literature; normal subject data from Inman VT, *Human Walking*. Baltimore, Md: Williams & Wilkins; 1981.

These orthoses incorporate two main functional features: (1) reduction of weight bearing on the lower extremity by channeling a major amount of the weight-bearing forces through the ground via the orthosis and, thus, bypassing the skeletal system; and (2) maintenance of bone and joint alignment.

Ischial Weight-Bearing Orthoses

The ischial weight-bearing orthosis uses a cuff design essentially identical to that of the socket for the above-knee amputee (Figure 11-67). The bulge over the femoral triangle pushes the limb backward so the ischium rests on the ischial seat of the orthotic cuff. The plastic cuff is made using lamination techniques or using heat-forming plastics. It can extend all the way to the level of the knee and, therefore, maintain full alignment of the femoral shaft. At the knee, the orthosis usually incorporates a lockable knee joint, as in the KAFO. Below the knee, it may incorporate the equivalent of a PAFO with adequate front closure to preserve proper alignment of tibia and fibula. As an alternative, a metal double-stopped ankle joint can be used in combination with a sole plate to the metatarsal head area. The orthosis is purposely fitted too long, that is, with a clearance of up to one inch between the heel of the foot and the sole of the shoe, to force weight bearing through the orthosis. The transmission of force into the orthosis occurs by way of the ischium into the ischial seat and through the noncompressible soft tissues into the thigh cuff.

Biomechanical studies[62] have compared the effectiveness of the weight-bearing function of this orthosis with different designs, with and without the patient having been trained in their use. The ground reactive force was compared with the amount of force transmitted through the instrumented orthosis. The difference between the two represents the force transmitted through the skeletal system. In general, the weight-bearing function of this orthosis depends on the design and training as follows:

1. The orthosis with a locked knee and a patten bottom produces 100% weight bearing through the orthosis (Figure 11-68).
2. The orthosis with the locked knee, the fixed ankle, and rocker bottom, with training to avoid the pushoff, produces 90% weight bearing through the orthosis. Training consists of teaching the patients to avoid plantar flexion and pushoff within the orthosis, which would load the skeletal system and unload the brace.

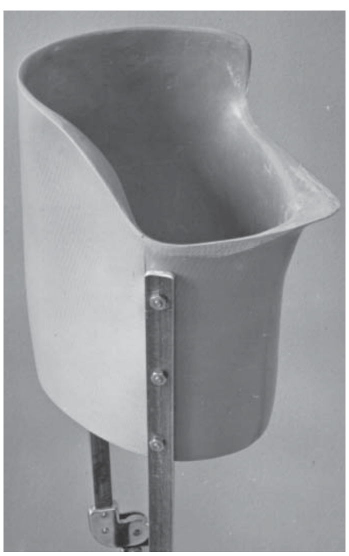


Fig. 11-67. Rigid, quadrilateral cuff for ischial weight-bearing orthoses. Reprinted with permission from Lehmann JF, Warren CG, de Lateur BJ, Simons BC, Kirkpatrick G. Biomechanical evaluation of axial loading in ischial weight-bearing braces of various designs. *Arch Phys Med Rehabil.* 1970;51:331-337.

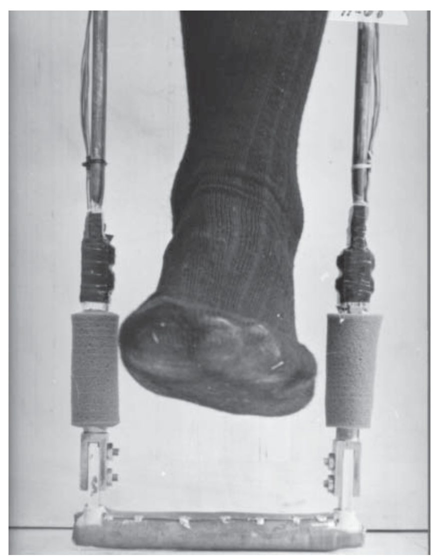


Fig. 11-68. Orthosis with locked knee and patten bottom. Reprinted with permission from Lehmann JF, Warren CG, de Lateur BJ, Simons BC, Kirkpatrick G. Biomechanical evaluation of axial loading in ischial weight-bearing braces of various designs. *Arch Phys Med Rehabil.* 1970;51:331-337.

3. The orthosis with the locked knee and fixed ankle, without a rocker bottom but with training, produces weight bearing through the orthosis at approximately 80% of body weight.
4. The orthosis with the locked knee and fixed ankle, with no training, produces weight bearing through the orthosis at approximately 50% of body weight. Due to the lack of training to avoid pushoff, the patient loads the skeletal system and unloads the orthosis by plantar flexing the foot, pushing against the ground with the forefoot during the latter part of the stance. This reduces the weight bearing from 80% to 50%.
5. The orthosis with the locked knee and free ankle joint, with no training, produces approximately a maximum of 50% weight bearing throughout the orthosis, but only during the heelstrike phase. During pushoff, the lead foot contacts the ground and loads the skeletal system, thereby reducing the weight-bearing function of the orthosis.

In conclusion, the ischial weight-bearing orthosis can reduce weight bearing through the skeletal system (ie, the femoral shaft and below) to variable degrees; the amount of reproduction depends on the modifications of the orthosis and the training of the patient. However, these orthoses cannot be effectively used for protection of the hip joint because only about one third of the transmission of weight occurs at the ischium, which bypasses the hip joint. The rest goes through the hip joint and then through the soft tissues into the cuff of the orthosis.

The weight-bearing orthosis has commonly been used in the form of fracture cast bracing for early mobilization after femoral fractures, and has also been successfully used in cases of delayed union or nonunion fractures. This is presumably because the orthosis maintains the fracture alignment, while at the same time allows some force transmission through the fracture site during functional ambulation, thus stimulating healing.[63,64]

Patellar Tendon Bearing Orthosis

The basic design of the weight-bearing cuff is the same design as that of the patellar tendon bearing socket used for the below knee amputee. For easier donning and doffing, the cuff is halved and the two halves are connected by buckles similar to those of a rigid ski boot. The cuff may be extended to incorporate the ankle joint. The cuff positions the knee into 10° of flexion to allow proper weight bearing through the patellar tendon area, which pushes against the indentation of the cuff. Correspondingly, the fixed ankle joint (Figure 11-69) should be adjusted in 7° of dorsiflexion to accommodate the flexed knee. Again, clearance between the heel and the sole of the shoe can be up to one inch. With a fixed ankle, and training to avoid plantar flexion (pushoff) of the foot, weight bearing can be maximally 50% to 60% of body weight. To avoid weight bearing during pushoff, a rocker bottom may be added.

According to Fry and associates,[65] indications for the short-term use of the orthosis are fractures of the os calcis, postoperative fractures of the ankle, painful conditions of the heel that have been refractory to conservative management, and where surgery is contraindicated. Long-term use of this orthosis has been recommended for delayed union or

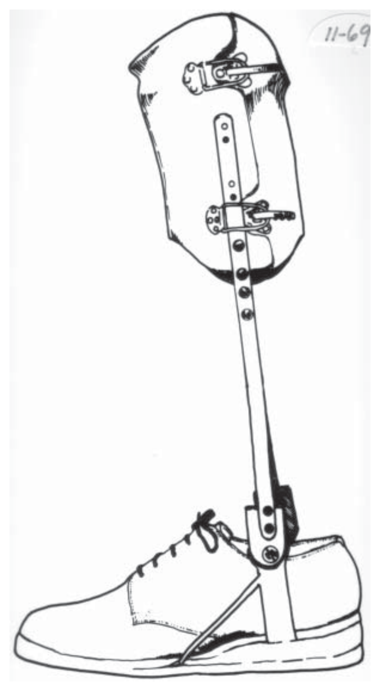


Fig. 11-69. Patellar tendon-bearing brace for limiting weight-bearing, incorporating bivalved patellar tendon-bearing cuff closed by ski boot buckles, standard uprights, double-stopped ankle joint, and sole plate extending to the metatarsal head area. Reprinted with permission from Lehmann JF. Lower limb orthotics. In: Redford JB, ed. *Orthotics Etcetera.* 3rd ed. Baltimore, Md: Williams & Wilkins; 1986.

nonunion of fractures and fusions, avascular necrosis of the talar joint, degenerative arthritis of the talar or ankle joint, osteomyelitis of the os calcis, and sciatic nerve injury with secondary anesthesia that involves the sole of the foot.[65] The only contraindication identified was interference of the orthosis with circulation in the limb. This may be the result of the pressure on the popliteal space which pushes the knee with the patellar tendon area against the front of the cuff, thus interfering with arterial circulation in the limb.

Depending on the design, the functional leg length of any one of the weight-bearing orthoses may be increased. Compensation should be made with shoe lifts on the opposite side.

Orthoses for Control of Knee Alignment

Orthoses can be used to control common knee alignment problems, such as genu recurvatum, or valgus or varus deformity of the knee.

Many knee cage designs are unsuitable for use because they slide up and down on the limb. When this happens, the axis of motion of the cage does not remain aligned with the axis of motion through the anatomical knee joint. As a result, forces are created between the orthoses and the knee, potentially damaging an already injured knee and its ligaments. An effective knee cage (Figure 11-70) consists of a plastic thigh cuff [66] that fits tightly above the femoral condyles and, therefore, prevents downward sliding of the orthosis. Correspondingly, the leg cuff is tightly pulled in below the tibial condyles to prevent upward riding. The two components, upper and lower, are connected by a knee joint that should be properly aligned to approximate the location of the anatomical knee axis of motion and may be polycentric to correct for any slight deviation of an orthotic axis from the instantaneous anatomical knee axis. An extension stop could be added to the joint to prevent hyperextension of the knee. Therefore, this design could be used for correction of genu recurvatum. It also provides a fairly good alignment of the knee and prevents valgus or varus deformities. Complete immobilization of the knee could be achieved if the knee joint is locked. However, this could be achieved less expensively by a cylinder cast or by commercially available braces.

If the cause for the hyperextension of the knee is located at the ankle, a knee cage should not be used to correct a genu recurvatum. Genu recurvatum is often due to the gastrocnemius and soleus contracture with hemiplegia fixing the foot in plantar flexion and, therefore, creating an excessive knee extension moment during the latter part of the stance.[31,32] In this case, physical therapy should be used to stretch out the contracture or, if necessary, heel cord lengthening surgery should be performed. Alternatively, or in addition, a metal double upright AFO, with an adjustable posterior plantar flexion pinstop in combination with a sole plate, should be used. As the patient continues to ambulate, the gastrocnemius and soleus are gradually stretched by adjusting the pinstop so that the foot is fixed more and more in dorsiflexion.

More extensive valgus deformity of the knee, as it is seen in patients with rheumatoid arthritis or other medial collateral ligament destruction, requires the use of a KAFO. The alignment of the knee is maintained by adding a padded dial to the medial upright of the orthosis or extending a plastic cuff from below the knee medially to the knee level. Smith and associates[67] found that a single upright orthosis can be used when the corrective forces do not exceed 18 to 20 lb. With greater forces the double upright orthosis should be used. The maximal knee flexion contracture that can be successfully fitted with this orthosis is 15° to 20°. If any flexion-extension movement is allowed at the knee, it is important that the location of the brace axis coincides as closely as possible with the location of the anatomical axis of the knee joint, or else forces between the

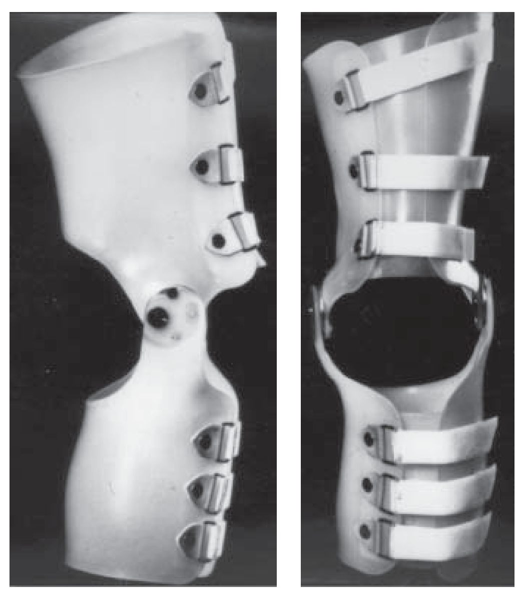


Fig. 11-70. Anterior (left) and medial (right) views of the genucentric knee orthosis. Reprintedwith pemission from Foster R, Milani J. The genucentric knee orthosis—a new concept. *Orthot Prosthet.* 1979;33:31-44.

orthosis and the knee will further injure an already damaged knee joint. Smith and colleagues[67] found that a well-fitted orthosis was well tolerated for periods of 7 to 48 months, thus providing protection against further deformities of the knee and allowing time for healing.

CONCLUSION

In conclusion, the successful application of the orthoses for the upper and lower extremities requires a full understanding of the function of muscles and joints and the biomechanics of orthotic design. Therefore, it is necessary to examine the patient carefully before prescribing a customized orthosis. A "shotgun" approach to orthosis prescription may lead to development of bad habits in the patient, for example, as in prescribing a foot drop orthosis for a stroke patient while disregarding the biomechanical influence of the orthosis on the knee.

REFERENCES

1. Bennett RL. Orthotics for function: Prescription. *Phys Ther Rev.* 1956;36:721-730.
2. Long C. Upper limb bracing. In: Licht S, ed. *Orthotics Etcetera. 3rd ed.* Baltimore, Md: Williams & Wilkins; 1966.
3. Schutt AH. Upper extremity and hand orthotics. *Phys Med Rehabil Clin North Am.* 1992;3:223-241.
4. Anderson M. *Upper Extremity Orthotics.* Springfield, Ill: Charles C Thomas; 1958.
5. Bender LF. Prevention of deformities through orthotics. *JAMA.* 1963;183:946-948.
6. Jain AS, McDougall D. A versatile hand splint. *Prosthet Orthot Int.* 1987;11:21-24.
7. Stenehjeu J, Swenson J, Sprague C. Wrist driven flexor hinge orthosis: Linkage design improvements. *Arch Phys Med Rehabil.* 1983;64:566-568.
8. Engen TJ. Development of externally powered upper extremity orthotic systems. *J Bone Joint Surg.* 1965;47B:465-468.
9. Peckham OH, Mortimer JT, Marsolais EB. Controlled prehension and release in the C-5 quadriplegic elicited by functional electrical stimulation of the paralyzed forearm. *Ann Biomed Eng.* 1980;8:369-388.
10. Stolov WC. The concept of normal muscle tone, hypotonia and hypertonia. *Arch Phys Med Rehabil.* 1966;47:156-168.
11. Bunnell S. *Bunnell's Surgery of the Hand.* 4th ed. Revised by JH Boyes. Philadelphia, Pa: JB Lippincott; 1964: 170.
12. DeLisa JA, Greenberg S. Basic upper extremity orthotics. *Am Fam Physician.* 1981;24:169-175.
13. McPherson JJ, Kreimeyer D, Aalderks M, Gallagher T. A comparison of dorsal and volar resting hand splints in the reduction of hypertonus. *Am J Occup Ther.* 1982;36:664-670.
14. Wilson DJ, McKenzie MW, Barber LM. *Spinal Cord Injury: A Treatment Guide for Occupational Therapists.* Thorofare, NJ: Charles B. Slack; 1974.
15. Smith EM, Juvinall RC. Theory of "feeder" mechanics. *Am J Phys Med.* 1963;42:113-139.
16. Dillingham TR, Spellman NT, Braverman SE, et al. Analysis of casualties referred to army physical medicine services during the Persian Gulf conflict. *Am J Phys Med Rehabil.* 1993;72(4):214-218.
17. Claus BS, Godfrey KJ. Brief or new: A distal support sling for the hemiplegic patient. *Am J Occup Ther.* 1985;39:536-537.
18. Hurd MM, Farrell KH, Waylonis GW. Shoulder sling for hemiplegia: Friend or foe? *Arch Phys Med Rehabil.* 1974;55:519-522.

19. Prévost R. Bobath axillary support for adults with hemiplegia: A biomechanical analysis. *Phys Ther.* 1988;68:228-232.

20. Smith RO, Okamoto GA. Checklist for the prescription of slings for the hemiplegic patient. *Am J Occup Ther.* 1981;35:91-95.

21. Brooke MM, de Lateur BJ, Diana-Rigby GC, Questad KA. Shoulder subluxation in hemiplegia: Effects of three different supports. *Arch Phys Med Rehabil.* 1991;72:582-586.

22. Wynn Parry CB. Orthotics and rehabilitation after extensive upper limb paralysis. *Hand Clin.* 1989;5:97-105.

23. Wynn Parry CB. *Rehabilitation of the Hand.* 4th ed. London, England: Butterworths; 1981.

24. Williams R, Taffs L, Minuk T. Evaluation of two support methods for the subluxated shoulder of hemiplegic patients. *Phys Ther.* 1988;68:1209-1214; erratum, *Phys Ther.* 1988;68:1969.

25. Villanueva R. Orthosis to correct shoulder pain and deformity after trapezius palsy. *Arch Phys Med Rehabil.* 1977;58:30-34.

26. Truong XT, Rippel DV. Orthotic devices for serratus anterior palsy: Some biomechanical considerations. *Arch Phys Med Rehabil.* 1979;60:66-69.

27. Schottstaedt ER, Robinson GB. Functional bracing of the arm: Part I. *J Bone Joint Surg.* 1956;38A:477-499.

28. Rudin LN, Cronin DJ, Croucher JS. Corrective brace for the upper extremity in hemiplegia. *JAMA.* 1953;153:479.

29. Inman VT, Saunders JB de CM, Abbott LC. Observations on functions of shoulder joint. *J Bone Joint Surg.* 1944;26A:1-30.

30. Lehmann JF, Ko MJ, de Lateur BJ. Double stopped ankle-foot orthosis in flaccid peroneal and tibial paralysis: Evaluation of function. *Arch Phys Med Rehabil.* 1980;61:536-541.

31. Lehmann JF, Warren CG, de Lateur BJ. Biomechanical evaluation of knee stability in below-knee braces. *Arch Phys Med Rehabil.* 1970;51:688-695.

32. Lehmann JF, Ko MJ, de Lateur BJ. Knee moments: Origin in normal ambulation and their modification by double-stopped ankle-foot orthoses. *Arch Phys Med Rehabil.* 1982;63:345-351.

33. Lehmann JF. Lower extremity orthotics. In: Kottke FJ, Lehmann JF, eds. *Krusen's Handbook of Physical Medicine and Rehabilitation.* 4th ed. Philadelphia, Pa: WB Saunders; 1990.

34. Lehmann JF. Biomechanics of ankle-foot orthoses: Prescription and design. *Arch Phys Med Rehabil.* 1979;60:200-207.

35. Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Ankle-foot orthoses: Effect on gait abnormalities in tibial nerve paralysis. *Arch Phys Med Rehabil.* 1985;66:212-218.

36. Wong AMK, Tang F-T, Wu S-H, Chen C-M. Clinical trial of a low-temperature plastic anterior ankle foot orthosis. *Am J Phys Med Rehabil.* 1992;71:41-43.

37. Lehmann JF, Esselman PC, Ko MJ, de Lateur BJ, Dralle AJ. Plastic ankle-foot orthoses: Evaluation of function. *Arch Phys Med Rehabil.* 1983;64:402-407.

38. Fillauer C. A new ankle-foot orthosis with a moldable carbon composite insert. *Orthot Prosthet.* 1981;35:13-16.

39. Lehmann JF, de Lateur BJ, Price R. Ankle-foot orthoses for paresis and paralysis. *Phys Med Rehabil Clin North Am.* 1992;3:139-159.

40. Inman VT. *The Joints of the Ankle.* Baltimore, Md: Williams & Wilkins; 1976.

41. Fisher SD, Gullickson G. Energy cost of ambulation in health and disability: A literature review. *Arch Phys Med Rehabil.* 1978;59:124-133.

42. Liberson WT, Holmquest HF, Scott D, Dow M. Functional electrotherapy: Stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemiplegic patients. *Arch Phys Med Rehabil.* 1961;42:101-105.

43. Waters RL, Miller L. A physiologic rationale for orthotic prescription in paraplegic. *Clin Prosthet Orthot.* 1987;11:66-73.

44. Lehmann JF, Warren CG. Restraining forces in various designs of knee-ankle orthoses: Their placement and effect on the anatomical knee joint. *Arch Phys Med Rehabil.* 1976;57:430-437.

45. Lehmann JF, Stonebridge JB. Knee lock device for knee ankle orthoses for spinal cord injured patients: An evaluation. *Arch Phys Med Rehabil.* 1978;59:207-211.

46. Lehmann JF, Warren CG, Hertling D, McGee M, Simons BC, Dralle A. Scott-Craig orthosis: A biomechanical and functional evaluation. *Arch Phys Med Rehabil.* 1976;57:438-442.

47. Scott BA. Engineering principles and fabrication techniques for Scott-Craig long leg brace for paraplegics. *Orthot Prosthet.* 1971;25:14-19.

48. Clinkingbeard JR, Gersten JW, Hoehn D. Energy cost of ambulation in the traumatic paraplegic. *Am J Phys Med.* 1964;43:157-165.

49. Gordon EE, Vanderwalde H. Energy requirements in paraplegic ambulation. *Arch Phys Med Rehabil.* 1956;33:276-285.

50. Merkel KD, Miller NE, Westbrook PR, Merritt JL. Energy expenditure of paraplegic patients standing and walking with two knee-ankle-foot orthoses. *Arch Phys Med Rehabil.* 1984;65:121-124.

51. Lehmann JF, de Lateur BJ, Warren CG, Simons BC, Guy AW. Biomechanical evaluation of braces for paraplegics. *Arch Phys Med Rehabil.* 1969;50:179-188.

52. Warren CG, Lehmann JF, de Lateur BJ. Use of the pelvic band in orthotics for adult paraplegic patients. *Arch Phys Med Rehabil.* 1975;56:221-223.

53. Merritt JL. Knee-ankle-foot orthotics: Long leg braces and their practical applications. *Phys Med Rehabil State Art Rev.* 1987;1:67-82.

54. Jefferson RJ, Whittle MW. Performance of three walking orthoses for the paralysed: A case study using gait analysis. *Prosthet Orthot Int.* 1990;14:103-110.

55. Mazur JM, Sienko-Thomas S, Wright N, Cummings RJ. Swing-through vs reciprocating gait patterns in patients with thoracic-level spina bifida. *Z Kinderchir.* 1990;45(Suppl 1):23-25.

56. Meadows CB, Stallard J, Wright D, Gordon L, Major RE, Jones N. The Edinburgh-ORLAU prosthetic system to provide reciprocal locomotion in children and adults with complete transverse lower limb deficiency. *Prosthet Orthot Int.* 1990;14:111-116.

57. Phillips CA, Koubek RJ, Hendershot DM. Walking while using a sensory tactile feedback system: Potential use with a functional electrical stimulation orthosis. *J Biomed Eng.* 1991;13:91-96.

58. Mulder AJ, Veltink PH, Boom HBK, Zilvold G. Low-level finite state control of knee joint in paraplegic standing. *J Biomed Eng.* 1992;14:3-8.

59. Marsolais EB, Kobetic R. Functional walking in paralyzed patients by means of electrical stimulation. *Clin Orthop.* 1983;175:30-36.

60. Durfee WK, Hausdorff JM. Regulating knee joint position by combining electrical stimulation with a controllable friction brake. *Ann Biomed Eng.* 1990;18:575-596.

61. Marsolais EB, Edwards BG. Energy costs of walking and standing with functional neuromuscular stimulation and long leg braces. *Arch Phys Med Rehabil.* 1988;69:243-249.

62. Lehmann JF, Warren CG, de Lateur BJ, Simons BC, Kirkpatrick GS. Biomechanical evaluation of axial loading in ischial weight-bearing braces of various designs. *Arch Phys Med Rehabil.* 1970;51:331-337.

63. Kohn D, Wirth C-J, John H. The function of the Thomas splint. *Arch Orthop Trauma Surg.* 1991;111:26-28.

64. Mooney V, Nickel VL, Harvey JP, Snelson R. Cast brace treatment for fractures for the distal part of the femur: A prospective controlled study of one hundred and fifty-three patients. *J Bone Joint Surg.* 1970;52A:1563-1578.

65. Fry LR, Davis FJ, Lippert FG, Simons BC, Remington J. The patellar tendon bearing brace: Report of 16 patients. *J Trauma.* 1974;14:216-221.

66. Foster R, Milani J. The genucentric knee orthosis: A new concept. *Orthot Prosthet.* 1979;33:31-44.

67. Smith EM, Juvinall RC, Corell EB, Nyboer VJ. Bracing the unstable arthritis knee. *Arch Phys Med Rehabil.* 1970;51:28-36.