Chapter 20

LOWER LIMB PROSTHETICS

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INTRODUCTION

Lower extremity amputation is the most common level of amputation in both civilian and military populations. These two groups, although outwardly similar, have vastly differing etiologies that lead to amputation and have dramatically different rehabilitation and prosthetic concerns that exist throughout their lifetimes. A service member with lower extremity amputation will rely on a variety of prosthetic devices and components to maximize rehabilitation and integration to independence. The proper prescription, fitting, and training of lower extremity prosthetic devices are critical to successful usage and reintegration.

Military medical treatment facility (MTF) personnel, who specialize in amputee care, have designed programs based on a multidisciplinary team approach. The team is comprised of both medical professionals (surgeons, physiatrists, prosthetists, physical therapists, occupational therapists, nurses, physician assistants, psychiatrists, and physiologists) and nonmedical specialists (case managers, benefits counselors, and volunteers) who provide comprehensive services throughout the service members' rehabilitation.

The goals of providing lower extremity prosthetics devices are to allow the service member to achieve basic ambulation, to progress to a variety of advanced activities (eg, running, skiing), and, if warranted, to perform warrior tasks required to meet return-to-duty standards. These goals are best accomplished by using the comprehensive team approach to rehabilitation and prescribing multiple prosthetic limbs that include the use of technically advanced prosthetic devices. Use of advanced prosthetic limb components in this population is associated with successful rehabilitation outcomes. Experience has revealed that this group consistently outperforms the most advanced systems.

The purpose of this chapter is to identify the role of the prosthetist as a member of the comprehensive rehabilitation team, provide an overview of prosthetic components used at MTFs, and present the treatment protocol.

ROLE OF THE PROSTHETIST

Ideally, the rehabilitation team works in unison and complements each other's roles. Prescription formulation usually occurs during an interdisciplinary specialized clinic and is centered on the service member's needs and desires for prosthetic rehabilitation. Each member of the clinical team plays an important role. However, during prosthetic limb prescription formulation, the prosthetist takes center stage. The increasing complexity involved in prosthetics technology and the development of unique materials, computer programs, and fabrication techniques/applicability require the specialized knowledge of the prosthetist. The American Board for Certification in Orthotics, Prosthetics & Pedorthics (ABC) and the Board for Orthotic/Prosthetic Certification (BOC) provide benchmark certification levels for competency in the field. There are several pathways to certification, and it is important for each MTF to ensure that its prosthetists are appropriately certified. ABC, formed in 1946, requires that prosthetic education be obtained from institutions accredited by the Commission on Accreditation of Allied Health Education Programs and that prosthetists successfully complete a 1-year residency requirement per discipline (prosthetics and orthotics), as well as take subsequent board examinations. BOC requires 2 years of education, training, and/or work experience and 2 years of experience in providing direct patient care services.

Nearly all of the clinical specialties that comprise this interdisciplinary team are uniformed service members bridging both officer and enlisted ranks. The military occupational specialty classification for orthotist (originally code 42C) was eliminated in the mid-1990s. However, because of the global war on terror and the ensuing combat operations in Iraq and Afghanistan, the need for military prosthetists and orthotists is on the rise. To fill this gap, civilian orthotists and prosthetists have been employed at MTFs either through service contracts with companies or through hiring individual prosthetists as government service employees. It is important that prosthetists have a clear position within the military organizational system. This position has typically been classifed under the Department of Orthopaedics and Rehabilitation. In addition, it has not been uncommon for the Department of Defense to request that prosthetists help provide strategic and operational knowledge of prosthetic programs to allied governments, which has called into question the need for uniformed military prosthetists.

For example, during Operation Iraqi Freedom (specifically from January 2006 to June 2006), a five-man team of rehabilitation providers from the US Armed Forces Amputee Patient Care Program worked with Iraqi prosthetic and rehabilitation specialists to educate them on best practices in caring for patients with limb loss and complex trauma. The team instructed, assisted, and supervised Iraqi clinicians in the delivery of prosthetic and rehabilitation services to 124 patients (totaling more than 350 patient clinical visits). The clinic that they developed was transferred to the Iraqi Veterans Agency (in Baghdad, Iraq) and is now considered the premier prosthetics and rehabilitation clinic in Iraq.

The prosthetic mission was to educate Iraqi providers on state-of-the-art prosthetic care, with emphasis on enhancing provider clinical decision-making skills, improving tradecraft and bench skills, and enhancing the use of available technology. The primary goal was to shift Iraqi practice patterns from those of a semiskilled worker model to a more professional paradigm. Typically, Iraqi prosthetists are locally trained by the International Committee of the Red Cross or by nongovernmental organizations with support from the Iraqi Ministry of Health. Practitioners are often characterized as technicians working in "limb factories," as opposed to the Western model of an allied health professional performing clinical services. The assistance team trained local Iraqi prosthetists how to adopt an internationally recognized role as allied health providers in a community of rehabilitation science experts. Clinically, this required a change in paradigm from fabricating, fitting, and delivering fair-to-adequate lower extremity prosthetic limbs to one of consistently producing reliable and comfortable prosthetic arms and legs in a reasonable and timely manner while optimizing the use of available technology [eg, computer-aided design (CAD)/computer-aided manufacturing (CAM) systems]. Additionally, their Iraqi counterparts were instructed in how to refine their clinical evaluation skills, clinical documentation techniques, and prosthetic fabrication abilities. On completion of formal training, the Iraqi prosthetic providers returned to full-time direct patient care.

The art and science of prosthetic care and rehabilitation of persons with limb loss and polytrauma by necessity accelerates during times of war. Clinicians have a strong moral imperative to share with our allies the lessons learned in caring for seriously wounded service members. Thus, the outcome of this mission was to demonstrate the importance of appropriate technology, emphasize a multidisciplinary approach to care, and describe the standard of care in a subsegment of the Iraqi population of war wounded.

OVERVIEW OF COMPONENTS

Prosthetic Feet

Function

Prosthetic feet are classified according to function and features. An understanding of prosthetic foot function and design will assist clinicians in selecting feet best suited to a patient's activities. The prosthetic foot is the interface between the patient and the ground, and ideally would emulate the anatomical foot perfectly. Achieving this, however, is difficult. Prosthetic feet range from simple to complex as they attempt to mimic anatomical function. Anatomical articulation at the ankle and midfoot greatly affects gait efficiency and smoothness. It can increase knee stability, and a multiaxial foot can increase the base of support by accommodating uneven terrain. Prosthetic joint simulation can be accomplished by true articulation of the prosthetic foot, as well as compression of the foot itself. Related to joint movement is shock absorption, another critical feature that the prosthetic foot must emulate. The foot must dampen the impact at loading response during gait. This is necessary to diminish forces transmitted to the residual limb. An example of this is the compressible heel of the foot. At loading response, the heel of the prosthetic foot compresses, emulating the eccentric contraction of the dorsiflexors. This is referred to as simulated or relative plantar flexion. Another less functional, yet important, feature of prosthetic feet is appearance. This is more important to some individuals than others. If the patient intends to wear sandals, it is obvious that the cosmetic appearance of the foot must be good. With the exception of specialty feet designed for running or other activity, all feet have a cosmetic foot shell as a minimum. This shell protects the internal components from unnecessary wear and tear. The overall design of the prosthetic foot, its ability to articulate, and its material composition can have a significant effect on the foot's ability to emulate anatomical function. Generally, the greater the ability and activity of the patient, the more features the patient may be able to take advantage of during use.

Dynamic Response

The dynamic response foot has a deformable spring-type keel that provides a lively, responsive feel while the patient is walking. As the keel deforms under loading, it absorbs shock and then quickly returns to its original position once the load is removed. This type of design is often referred to as an energy storage and return (ESAR) design.[1] Suggested benefits of ESAR feet include the following: increased self-selected walking velocity, increased stride length, decreased sound side weight acceptance force, and increased prosthetic side propulsive force.[2,3] The spring-like keels are fabricated from carbon fiber, Delrin (E. I. du Pont de Nemours and Company, Wilmington, Del), or other materials. They deform as body weight is applied (Figure 20-1). Designs vary in configuration. Most have a forefoot keel and a hind foot keel that function as springs and deform primarily in the sagittal plane. The hind foot keel that projects posteriorly (a heel spring) deforms at loading response and simulates plantar flexion. From late midstance to toe off, the forefoot keel deforms under the increased load. As the patient begins to push off, the spring resists this motion while it yields slightly. As the load is being removed from the foot, the toe spring pushes back against the patient until it returns to its original shape. This yields a livelier feel for the patient. Some carbon keels are split longitudinally, which allows the toe spring to scissor slightly, thus simulating inversion and eversion. Dynamic response feet are designed to yield upon load application. Keel thickness and strength are based on the patient's weight and activity level. If either one increases, it is possible that the keel may fracture. A decrease in activity or weight may make the keel seem too rigid. Therefore, it is crucial that body weight and activity level be accurately assessed when selecting a dynamic response foot, given that keels are designed with a relatively narrow tolerance. With many carbon dynamic response feet, the keel continues to the distal socket that acts as both the foot and shank of the prosthesis. This increased carbon strut length provides more stored energy than those that terminate at the ankle. Patient height and residual limb length often will determine which style of foot can be used. Dynamic response feet provide excellent stability. The composition may vary, but, in general, these feet are typically lightweight, durable, and covered with a cosmetic foot shell.

Dynamic Response With Vertical, Shock-Absorbing Pylons

Vertical, shock-absorbing pylons help to dampen the axial loading of the prosthesis and thereby decrease the shock transmitted to the residual limb.[4] Several designs are available; some are integrated into the foot using a side carbon strut (Figure 20-2) that supports

Figure 20-1. Renegade Prosthetic Foot (Freedom Innovations, Irvine, Calif).
Photograph: Used with permission from Freedom Innovations.

Figure 20-2. Re-Flex VSP prosthetic foot (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

the patient's weight and bows outwardly on weight bearing as the piston slides freely into and out of the cylinder. Other designs provide the user with the ability to adjust the resistance to vertical load by changing the psi in the unit (Figure 20-3), whereas other designs are separate components added to the standard pylon. These feet are available in standard and low profile heights to accommodate most limb lengths. Their ability to provide shock absorption can increase the patient's comfort significantly when stepping off a curb, going down stairs, or performing other high-impact activities. These feet are designed with a split keel and also offer simulated inversion and eversion.

Articulated. Commonly used articulated feet are multiaxis in the sagittal, coronal, and transverse planes. Articulating dynamic response feet possess the advantages of the multiaxis foot in terms of terrain accommodation and the liveliness of the dynamic response foot. These feet are intended to provide accommodation with the ground without sacrificing their ability to provide dynamic response. They articulate through either a joint or a deformable wedge. If the prosthetic foot has an actual joint axis about which it articulates (Figure 20-4), it is referred to as a true articulation. Almost any prosthetic foot with an axis of rotation will require regular maintenance to lubricate the joint, which prevents wear and tear. Compressible bumpers are available with varying firmness (or durometer), allowing the prosthetist to adjust the plantar flexion resistance to accommodate the patient. Some designs also allow for dorsiflexion bumper exchange so that resistance to dorsiflexion may be adjusted until the foot hits the anterior stop.

Figure 20-4. TruStep prosthetic foot (College Park Industries, Fraser, Mich).
Photograph: Courtesy of College Park Industries.

Another method of providing articulation without moving parts, such as an axis, is to use an elastic bond of some sort between two dynamic response keels to provide ground conformance (Figure 20-5). A strap on the posterior aspect of the foot prevents excessive

Figure 20-3. Ceterus prosthetic foot (Ossur Americas, Aliso Viejo, Calif).
Photograph: Used with permission from Ossur Americas, www.ossur.com.

Figure 20-5. Talux prosthetic foot (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

elongation of the posterior section of the elastic bonding material as the patient rolls over onto the toe of the foot. Other dynamic response feet use a thick elastomeric material between carbon fiber keel plates. In this case, the bottommost keel readily accommodates to uneven terrain, while the more proximal keel may provide the typical energy-storing function. These types of feet are referred to as simulated articulation, because there is no true articulation (Figure 20-6).

Increased knee stability on the prosthetic side can be achieved by an articulated ankle. The knee flexion moment is reduced as the foot plantar flexes at loading response. The articulation can also increase knee stability as the patient walks down an incline. Plantar flexion resistance in the foot can be adjusted to accommodate the patient with compressible bumpers or wedges and through prosthetic alignment.

Additional Features

Heel Height Adjustability. Prosthetic feet are usually manufactured with a contour on the plantar surface. This is intended to match the patient's preferred heel height. Heel height is defined as the net difference between the thickness of the shoe sole at the ball of the foot (eg, ¼ inch) and the thickness of the heel at the posterior aspect of the shoe (eg, 1 inch). In this instance, the net difference is of an inch; a prosthetic foot with a -inch heel rise would be appropriate. When placed in the shoe, the proper orientation of the prosthetic foot aligns the proximal surface of the foot horizontally, parallel to the floor. Feet are available that permit patients to adjust the foot's dorsiflexion/plantar flexion position, providing them with the freedom to wear shoes of varying heel heights.

Inversion/Eversion. Many carbon keels are split longitudinally (Figure 20-7), which allows the forefoot keel to scissor slightly, simulating inversion and eversion as the foot traverses uneven terrain. This increased ground conformance provides greater stability. Another way to accomplish inversion and eversion is with an articulated foot, which provides coronal plane motion about the joint and is dampened by stiff rubber bumpers.

Microprocessor Ankle/Foot. One foot incorporates microprocessor-controlled and motor-powered dorsiflexion and plantar flexion during the swing phase. It is designed to make navigation of inclines, stair ascent and descent, and sitting onto or rising from a chair easier. A combination of sensors acts together to replace the mechanoreceptor function and provide artificial proprioception via gait pattern recognition algorithms (Figure 20-8). It is especially useful when there is a range of motion limitation of the contralateral knee and/or ankle.

Knee Units

Axis

Single Axis. In addition to their other features, prosthetic knees can be categorized as single axis or polycentric. The single axis knee is a simple, low-maintenance hinge design generally lightweight and

Figure 20-6. Axtion prosthetic foot (Otto Bock HealthCare, Minneapolis, Minn).
Photograph: Courtesy of Otto Bock HealthCare.

Figure 20-7. Vari-Flex prosthetic foot (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

Figure 20-8. PROPRIO FOOT (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

low profile. Prosthetic knee stability with a single axis knee relies heavily on alignment of the knee in relation to the socket and foot, and on the volitional control of the patient. The single axis design is coupled with constant friction, fluid, and microprocessor or powered stride control features. It is used for any level of patient ability.

Polycentric. The polycentric design is more complex and yields multiple centers of rotation, much like the anatomical knee. Prosthetic polycentric knees simulate the rocking and gliding motions of the anatomical knee either by curved-bearing surfaces or by linkages. Location of the axis changes as the prosthetic knee moves through its available range of motion. This is referred to as an instantaneous center of rotation (ICR) and results in an inherently stable knee. The way that the ground reaction forces move about the ICR makes this design more stable early in stance when the patient needs it most. In addition, the polycentric design shortens the shank during the swing phase, thus helping the foot clear the ground. Because of reorientation of the linkages during flexion of the polycentric knee, the shin segment translates posteriorly, and the segment rotates differently than it would if it acted about a single axis. This action results in a relative dorsiflexion of the foot as the knee flexes, which in turn creates increased toe clearance. The polycentric design also provides a more cosmetic appearance while sitting for those patients with long transfemoral amputations or knee disarticulations. The knee is designed so the shin folds back behind the socket at 90 degrees of flexion. This design minimizes the length discrepancy between the knees while sitting. The polycentric design is also ideal for patients with short residual limbs because the linkages place the axis of rotation proximal (closer to the workforce of the hip), requiring less hip extensor action, and the linkages also place the axis of rotation posterior to the weight line.[5]

Cadence Control

Constant Friction. Prosthetic knee joints should provide stability during stance and smooth, controlled knee extension during the swing phase. The most basic knee joint operates via a feature termed constant friction. A constant friction, single axis knee has a feature that provides continuous pressure / friction around the knee axis to control the velocity at which the shank and foot can swing to prevent excessive heel rise during flexion and terminal impact at full extension. The resistance to swing is adjustable, but not self-adjusting as velocity changes. For patients who do not alter the speed of their cadence, a constant friction knee is indicated because of its simplicity, durability, and ease of use. However, most individuals walk at varying speeds and will require a more gait-responsive knee.

Pneumatic/Hydraulic/Magnetorheologic. A fluid (or hydraulically)-controlled knee provides frictional resistance about the knee axis that will increase proportionally with speed. There are two basic types of fluid-controlled knees: (1) hydraulic or (2) pneumatic. Fluids include liquids, vapors, and gases. The major difference with fluids in prosthetic knees is that air is more easily compressible, and hydraulic fluids are not. Oil or air is forced through a small orifice or tube. Adjustment screws allow the size of the orifice to be changed to control the rate in which fluid flows through these ports. This permits fine-tuning of swing-phase resistance to individual needs. If air is used, the air under compression within the pneumatic cylinder also acts as an extension assist. The ability of air to compress gives pneumatic knees a springier feel to the patient. One knee functions with a magnetorheologic fluid (Figure 20-9). This fluid contains metallic particles that respond to an electrical charge affecting the viscosity of the fluid. Instead of flowing through valve-controlled ports, the fluid acts on the knee axis to provide braking friction by becoming more or less viscous. Here, too, sensors

Figure 20-9. RHEO KNEE (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

send knee angle, velocity, and force data to the knee's microprocessor, which in turn adjusts a magnetic field around the fluid.

Microprocessor Knee

Microprocessor knees (Figure 20-10) are a class of knees that sense the conditions acting on the knee joint and can quickly make internal adjustments to safely meet those conditions. Joint angles and forces on the pylon are measured via sensors and are then sent to the microprocessor for rapid adjustment. Valves open or close electronically to increase or decrease fluid flow through the knee's internal ports, or the viscosity of the magnetorheologic fluid changes to vary resistance to knee flexion or extension. The advantage that microprocessor knees have over mechanical knees is that rapid input from the microprocessor makes them significantly more responsive to the patient's activity, whether walking, running, descending stairs, or stumbling.[6] These knees allow for manipulation of the computer program to vary the stability and safety of the knee as the patient progresses through rehabilitation. Stumble recovery is a significant feature of these knees.

Motorized Knee

Building on microprocessor technology, the motorized knee (Figure 20-11) provides flexion and extension to an externally powered motor-driven knee. The knee replaces lost muscle function. It is therefore possible for the user to climb stairs step by step, ascend inclines, and walk longer distances on level ground. Sensors are positioned on the sound side, which accurately measure motion, load, and position. This information is transmitted to the knee via Bluetooth technology, where the microprocessor analyzes the data and determines the response of the knee to the activity and the amount of power or force needed from the knee to generate the appropriate knee flexion or extension.

Additional Knee Features

Stance Flexion. Controlled prosthetic knee flexion under weight bearing is referred to as stance flexion. As the patient transfers weight onto the prosthesis in the early stance phase, the knee gradually yields to approximately 15 degrees, thereby cushioning the impact of weight bearing to the wearer. Stance flexion is adjustable and may be completely eliminated if the patient finds it disconcerting and fears that the knee may buckle, or it can be adjusted to allow up to 15 degrees of flexion. Although it requires some getting used to, this can be a very smooth knee mechanism for the patient to ambulate on because the stance flexion absorbs some of the shock transmitted to the socket during loading response.

Geometric Lock. The geometric lock feature, also

Figure 20-10. C-Leg (Otto Bock HealthCare, Minneapolis, Minn).
Photograph: Used with permission from Otto Bock HealthCare.

referred to as a mechanical stance phase lock, locks the knee on full extension (initial contact) and does not release it until the weight line passes over the forefoot and hyperextension of the knee unit occurs. This clever geometric design is engineered into a polycentric knee and provides excellent stability. The geometric lock can be convenient because it provides reliable stance control.

User-Activated Lock. A manual lock is available on some constant friction and fluid knees. For those activities requiring maximum stability, this mechanism is ideal. Prolonged standing (ie, at a workbench) is one such example, as is traversing exceptionally rugged terrain. The patient simply flips a lever to engage the locking feature. A patient with bilateral amputations may occasionally need the stability of a locked knee. Ambulation with a locked knee compromises normal gait mechanics and should be carefully considered before being used for the purpose of daily ambulation. Locked knees are intended for activities beyond normal ambulation. Whereas mechanical knees are locked at full extension, one microprocessor knee can be adjusted through programming to lock the knee at any degree of knee flexion.

Other Components

Torsion

The rotation/torsion that occurs about the limb during gait can result in shear pressures if too much torsion is translated onto the limb.[7] Gel liners, feet with integrated ankles, and torsion adaptors can absorb

Figure 20-11. POWER KNEE (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

some of this torsion to decrease the rotational shear preventing skin abrasions and reducing stress on the knee joint. The torsion adaptor should be placed as proximal on the transtibial or transfemoral prosthesis as possible to minimize the perceived weight of the prosthesis by the patient. This feature is particularly helpful with activities that include a lot of pivotal movements (eg, golfing) in which patients experience a greater level of comfort. Axial motion occurs in the adaptor instead of between the socket and the skin. Some torsion adaptors have independent internal and external rotation adjustments.

Combined Torsion and Vertical Shock

Independent torsion and shock-absorbing units are available and can be paired with any foot. Interchangeable elastomer rods or adjustable springs provide dampening of vertical and rotational forces, thus reducing the strain on joints and soft tissues. Both torsion absorption and vertical shock dampening are inherent to some prosthetic feet (Figure 20-12).

Positional Rotators

A positional rotator (Figure 20-13) permits the patient to rotate the flexed knee and shank of the prosthesis out of the way for ease of movement. It enables patients to sit cross-legged and swing the prosthesis away from the pedals while driving. This is an added convenience for patients with transfemoral or higher level amputations. Designed for non–weight-bearing activities only, the rotation is activated by pressing a button, and snaps back into place. Changing shoes and socks while wearing the prosthesis also becomes a much easier task.

Quick Release Couplings

Knees and feet can be easily interchanged by the patient under a well-fitting socket. A quick release coupler is available when varying activities necessitate such a change. Examples include the following: exchanging a microprocessor knee for a water-resistant knee and foot or temporarily removing components below the socket to maneuver about tight spaces.

Figure 20-12. Delta Twist Shock Absorber (Otto Bock HealthCare, Minneapolis, Minn).
Photograph: Courtesy of Otto Bock HealthCare.

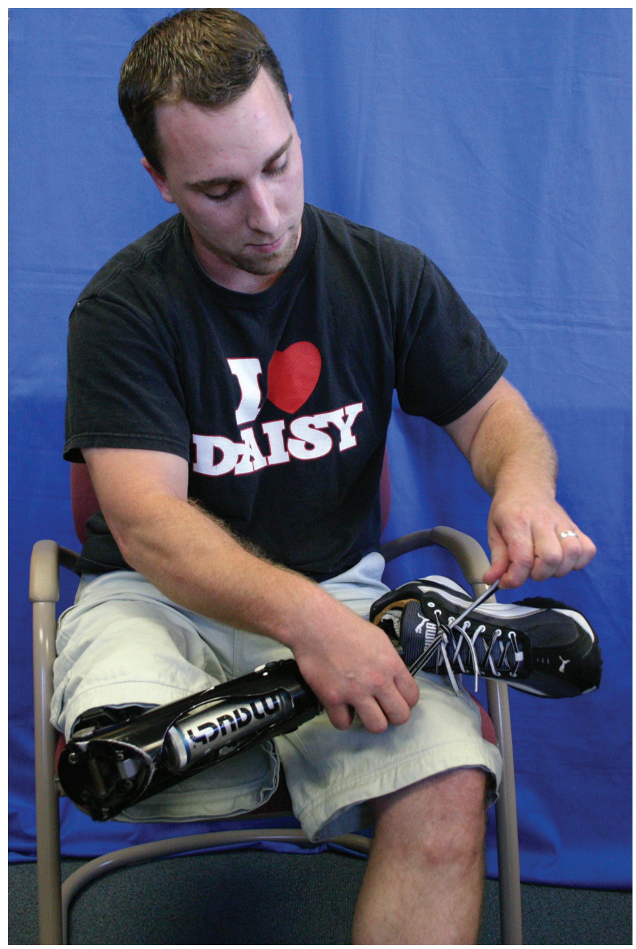

Figure 20-13. Positional Locking Rotator (Otto Bock Healthcare, Minneapolis, Minn).
Photograph: Used with permission from Otto Bock HealthCare.

Socket Interfaces

Gel Liners

Cushioning and shear-reducing properties of gel liners make them ideal for all patients at any level of amputation. In addition, their compressive properties make them ideal as a shrinker during the early phase of rehabilitation. Gel liners can be used as a means of suspension, given their friction properties against the skin that prevents distal migration of the liner (Figure 20-14). They can be used for comfort in conjunction with any type of suspension. Patients exhibiting scar tissue will particularly benefit from the gel. Not only is shear greatly reduced, but also some liners slowly diffuse a medical-grade mineral oil or other emollient onto the skin. This aids in scar management by keeping the scar tissue soft and supple. Protection from socket pressures and shear has historically been treated with custom-made soft liners of high-density foams or with a combination of leather and silicone. Additionally, lack of protective sensation is another condition commonly managed with gel liners. Gel liners are available in 3-mm, 6-mm, and 9-mm thicknesses, with the latter being the most protective. They are available with or without fabric covering. Choosing a liner with a matrix embedded within the gel minimizes vertical displacement. The liners are available off the shelf in many sizes, durometers, and thicknesses. They are made of silicone, urethane, mineral oil, or a thermoplastic elastomer. Custom liners can be designed for atypical limb contours with varying durometers. Some liners are manufactured with a unique distal fabric weave to minimize elongation, and most are thinner posteriorly or rippled, thus facilitating comfortable knee flexion. Although the liners are relatively durable, they are not puncture-resistant or tear-resistant. Proper caution must be exercised to minimize damage. Longevity of the liners is very much dependent on the activity level of the patient and the care given to the liner. Heavy duty use, improper donning, lack of routine cleansing, or careless storage can shorten the life of a liner. Worn liners may abrade the skin and should be replaced. Given direct contact with the skin and the sealed environment, it is imperative that the liners be cleansed daily. They are designed to be hypoallergenic, and the occurrence of dermatitis or other skin irritations is rare. If a skin reaction does occur, it can be dealt with by switching to a gel liner made of another material and/or consulting with a physician. If the problem persists, a physician should be notified. Applying the liner correctly requires some degree of dexterity and cognitive function. Donning is achieved by turning the liner inside out, placing the end firmly against the limb, and rolling the liner up the limb. The inherent frictional (suspension) properties of the liners preclude them from being pulled on like a sock. Prosthetic socks can be worn over the liner to accommodate limb volume fluctuation.

Prosthetic Socks

Prosthetic socks are the interface materials between the residual limb and the socket. Even with the use of a gel liner, socks are still required to adjust the fit of the prosthesis. Generally, socks are

Figure 20-14. Original Alpha Liner (or gel liner; Ohio Willow Wood, Mt. Sterling, Ohio).
Photograph: Used with permission from Ohio Willow Wood.

provided in one-ply, three-ply, and five-ply thicknesses, and are combined to the desired thickness. They cushion the limb and wick away moisture when worn against the skin. In suction or vacuum suspension prostheses, they are necessary to wick air out of the system. Socks are knitted with wool, cotton, or any number of synthetic fibers. Some synthetic blends offer stretch, increased wicking ability of moisture away from the skin, and ease of care. When using a hard socket, oftentimes placing a prosthetic nylon sheath directly over the skin, followed by thicker prosthetic socks, reduces shear and increases comfort. As the patient's limb volume changes throughout the day, the addition or reduction of a few plies will maintain a comfortable fit. Temperature, activity, weight change, and diet all have an effect on socket fit.

Prosthetic socks for fit adjustment are placed over the liner and do not affect the function of the pin. A hole is required in the distal end of the sock to allow the pin to pass through to the locking mechanism. Sock ply can be adjusted without affecting pin suspension. Patients should always be reminded to pull up the sock completely onto the liner and not allow any sock material to remain over the area of the pin. If the fabric is pulled into the locking mechanism along with the pin, it may jam and prevent the disengagement of the pin. This action prevents removal of the prosthesis. If this happens, it may be necessary to dismantle the locking mechanism to free the pin.

Hard Sockets

Transtibial hard socket designs have the advantage of simplicity, less bulk, durability, and easy cleanability. The classic description indicates that only a sock interface is present. The popularity of gel liners makes the hard socket with socks only a less prescribed option. However, patients who choose not to wear a gel liner and who desire the low maintenance and simple socket style can use the hard socket with a sock interface. Because the socket is not as forgiving as one with a gel liner, a well-fitting socket is crucial, and the patient must be diligent in sock management to remain comfortable. Scar tissue and sharp bony prominences are contraindications for this socket style.

Transfemoral Flexible Inner Liner With Rigid Frame

The rigid frame provides socket stability, and the inner flexible socket provides comfort proximally in the transfemoral socket and also space for muscle expansion in both the transtibial and transfemoral sockets via specifically placed frame cutouts. If designed with large frame cutouts, the flexible inner socket might even provide added proprioceptive feedback through the socket wall when in contact with, for example, a chair edge while sitting. Postfabrication adjustments of socket contours are also more easily accomplished when there is a flexible inner liner. In the instance of heterotopic ossification, flexible inner liners can be removed, heat-expanded to provide relief, and placed back into the rigid frame.

SYME PROSTHESIS

Socket Design and Suspension

Although the Syme amputation is considered end bearing, the socket design and trimlines remain similar to that of the transtibial socket. The high socket trimline provides the same loading characteristics as that of the transtibial socket. This trimline also reduces the likelihood of edge pressure along the tibial crest during late stance. The prosthesis should spread the load over the entire limb. The Syme prosthesis generally resists axial rotation because of the elliptical contour at the level of the malleoli. Shock absorption is limited because of length constraints because no vertical shock units can be incorporated into the prosthesis. The calcaneal heel pad, however, provides padding and protects the distal limb.

Commonly used suspension designs are window designs placed medially, cut to the smallest size necessary to allow the malleoli to pass through, and are of tubular construction with no opening to weaken the design. The latter offers the greatest socket strength against compressive forces during gait and is therefore most desirable. The medial window design is most appropriate when the distal limb width is much greater than the width just proximal to the bulbous end. This commonly occurs when the flares of the malleoli are left intact and prominent. The tubular-styled prosthesis is donned by applying a thin pad placed just proximal to the widest part of the distal limb. Once the limb is seated firmly into the prosthesis, it is wedged into the socket. Secure suspension is achieved via pad thickness and sock ply. Occasionally, this pad may also be incorporated into a full polyethylene closed-cell foam liner. If the malleoli have been completely tapered and no flare at the distal limb exists, then the prosthesis cannot be self-suspended. In this case, a gel liner, outer sleeve, and one-way expulsion valve can be used to provide suction suspension.

Foot Options

The Syme amputation retains the talus and calcaneus, resulting in limited clearance between the limb and floor. Most prosthetic feet are too tall to fit under a Syme prosthesis without creating a limb length discrepancy. Low profile feet specifically designed for use with the Syme amputation are available. Many are of a carbon plate design. Bolted directly to the distal socket, they can also be affixed with epoxies or other adhesives.

Syme Biomechanics and Socket Strength

At loading response, the patient will feel pressure on the posterior proximal and anterior distal residual limb. At late-stance phase, the pressures occur in exactly the opposite locations: the anterior proximal and posterior distal residual limb. Pressures are greatest on the residual limb during this late-stance phase as the patient pushes off to propel forward. To decrease forces on the residual limb at this time, the foot can be slightly dorsiflexed or moved more posterior. This is also the rationale for keeping the socket trimline to the level of the patellar ligament, thereby preventing edge pressure along the tibial crest. Similarly, increasing toe-out to functionally reduce the toe lever will lower anterior proximal pressure on the limb. Increased toe-out can provide additional lateral stability.

The greatest forces on the socket occur in the sagittal plane. The anterior distal aspect of the socket is subject to high compressive loads during late stance and should be strengthened to prevent socket failure. This is most important when using a socket with a medial window. Extra layers of carbon fiber or fiberglass should be incorporated into the laminated socket in this area. Decreasing the anterior lever (toe lever) during alignment reduces the compressive loads on the anterior distal socket.

TRANSTIBIAL PROSTHESIS

Socket Design

Patellar Tendon Bearing

The patellar tendon-bearing (PTB)[8] socket consists of a laminated or thermoplastic socket that provides an intimate, total contact fit over the entire surface of the residual limb. The anterior wall of the socket extends proximally to encapsulate the distal third of the patella. Just below the patella, located at the middle of the patellar ligament, is an inward contour or bar that (via other biomechanical forces) converts the patellar tendon (ligament) of the residual limb into a weight-bearing surface. PTB is a misnomer, however, because the patellar tendon is not the only weight-bearing surface used by the PTB socket. The medial and lateral walls extend proximally to about the level of the adductor tubercle of the femur. Together, they serve to control mediolateral/rotary forces applied to the residual limb. The medial wall is modified with an undercut in the area of the medial flare of the tibia, a pressure-tolerant surface and primary weight-bearing area. The lateral wall provides relief for the head of the fibula and the cut end of the fibula. This wall applies pressure along the fibular shaft to enhance medial-lateral stability. In addition, the lateral wall acts as a counter pressure to the medial wall. The posterior wall is slightly higher than the patellar bar and is designed to apply an anteriorly directed force to maintain the patellar tendon on the bar. The posterior wall is flared proximally to allow comfortable knee flexion and is contoured to prevent pressure on the hamstring tendons. The total contact fit provides relief over nonpressure-tolerant areas and supports the body's weight over the pressure-tolerant areas of the limb. Total contact is necessary to prevent limb edema, but does not imply equal pressures throughout the socket. PTB socket design is appropriate for nearly all transtibial amputations, but is seen less often with introduction of the total surface-bearing socket. The PTB prosthesis is designed to maintain the residual limb in slight initial flexion (from 5 degrees to 10 degrees) to convert the patellar bar to a more horizontal supporting force. However, because the patellar bar is not completely horizontal, the residual limb still has a tendency to slide down posteriorly. This tendency must be counteracted by the posterior wall that is contoured to maintain the patellar tendon's position on the patellar bar.

Total Surface Bearing

The total surface-bearing (TSB) socket design is commonly used when a gel liner is prescribed. The gel dissipates pressures throughout the socket and relieves bony prominences. Different from the PTB socket, pressure is intended for global distribution over the entire limb, as opposed to pressure-tolerant areas only. Bone and other sensitive regions are cushioned by the gel liner. Socket contours are smooth and without any obvious reliefs or undercuts lowering peak pressures.[9] Similar to the PTB socket, this design also provides a total contact fit for comfort. The TSB socket significantly differs from PTB concepts in modification technique. In the PTB socket, plaster is removed over the patellar tendon, lateral pretibial group, the medial flare, and the popliteal region. Areas of plaster addition are the tibial tubercle, along the tibial crest, and the proximal and distal fibulae. TSB modification does not require buildups. The gel material dissipates pressures throughout the socket and relieves bony prominences. Patients with fragile soft tissues, burns, skin grafts, scarring, or bony residual limbs can benefit significantly from the extra protection offered by the gel liner and TSB design.[10,11]

Suspension

Suspension Sleeves

Suspension sleeves (Figure 20-15) are available in neoprene, gel, or a breathable elasticized fabric. A neoprene suspension sleeve provides simple, low profile, relatively effective suspension. Its ease of application makes it an ideal option as an auxiliary form of suspension for periods of increased activity. The sleeve is fitted to the proximal third of the prosthesis and onto the midthigh several inches past the prosthetic socks. A small puncture or tear in the neoprene or gel sleeve can significantly impact its effectiveness because it provides suspension through a snug fit, resulting in friction and negative pressure. Activities such as kneeling or other heavy duty use may easily cut through a sleeve, thus hindering the ability to provide partial suction. On occasion, a patient may be allergic to the material. Because sleeves fit snuggly onto the thigh, the patient must have the strength and dexterity for proper application. If this is not the case, a more suitable suspension option should be sought. Heat buildup and perspiration are also concerns. Patients with upper extremity involvement may find the sleeve difficult to pull up onto the thigh. Sleeves do not provide sagittal or coronal stability.

Supracondylar/Supracondylar Suprapatellar Design

The supracondylar socket design affords the patient two primary benefits: (1) its high trimlines offer inherent suspension, and (2) it provides some control to a mildly unstable knee joint. The medial wall encompasses the femoral condyle and contours closely to the femur to purchase just past the level of the adductor tubercle, therefore creating a reduced proximal mediolateral dimension. This tight proximal configuration suspends the prosthesis over the femoral condyles. It is indicated primarily for individuals with short residual limbs to provide increased medial-lateral stability. It is

Figure 20-15. Neoprene suspension sleeve (Southern Prosthetic Supply, Alpharetta, Ga).
Photograph: Used with permission from Southern Prosthetic Supply.

contraindicated when knee instability requires more support. When a high anterior trimline—just past the proximal aspect of the patella—is added, then the socket becomes a supracondylar suprapatellar design. This area is referred to as the quadriceps bar. Encompassing the patella provides additional suspension, a kinesthetic reminder to limit knee extension, and prevents the medial and lateral brims from spreading.

Mechanical Lock Mechanisms

Mechanical locks are a positive locking system for suspension of the prosthesis. They are coupled with gel liners. The gel liner is rolled directly onto the skin and provides suspension through the mechanical lock. A variety of locks are available, some providing an audible clicking sound (ratchet style) that lets the patient know that the pin is engaged. Another lock style incorporates a clutch mechanism that allows the patient to manually draw the limb even deeper into the socket by turning a knob to eliminate any vertical displacement. A lanyard system is available when pin engagement is difficult because of mobile distal tissue or when a low profile lock is required to avoid making the prosthesis too long and creating a length discrepancy (Figure 20-16). Lock mechanisms can be very secure and less cumbersome forms of suspension, avoiding straps, belts, etc. To remove the prosthesis, a release pin is depressed that disengages the lock.

Figure 20-16. Lanyard-style suspension (KISS Technologies LLC, Baltimore, Md).
Photograph: Used with permission from KISS Technologies LLC.

Suction

Suction suspension can be created by using a gel liner without the locking pin and by adding an airtight outer sealing sleeve. A one-way expulsion valve located at the distal socket removes air from the socket, thus creating a negative vacuum environment. With such a system, the patient can enjoy the benefits of suction suspension, minimal vertical displacement, and increased proprioception. Maintenance of an airtight seal is important for the system to function properly.

Sealing Liners

Similar to suction suspension with a knee sleeve, sealing liners have the added benefit of eliminating the knee sleeve by providing a hypobaric seal (Figure 20-17) between the liner and the socket. The seal is pressed against the socket wall, and suction is maintained below the level of the sealing ring. A thin sheath is worn below the seal to wick air out through a one-way, push-button release expulsion valve. Knee flexion is unrestricted because no additional sleeve is required. These liners are available prefabricated or can be customized to place the seal at any desired level.

Vacuum Assisted

Suction can be enhanced with a vacuum. The hypobaric environment is achieved and maintained via a telescoping mechanical vacuum pump (Figure 20-18) or an electric pump. With each step, air is actively drawn out of the socket, maintaining a consistent and

Figure 20-17. Iceross Transfemoral Seal-In Liner with DermoSil (Ossur Americas, Aliso Viejo, Calif).
Photograph: Courtesy of Ossur Americas, www.ossur.com.

Figure 20-18. Harmony mechanical vacuum pump (Otto Bock HealthCare, Minneapolis, Minn).
Photograph: Used with permission from Otto Bock HealthCare.

comfortable level of vacuum. Additionally, the mechanical pump acts as a vertical shock absorber. The system consists of a gel liner covered by a sheath with a sealing sleeve placed over the prosthesis and rolled onto the thigh to create a sealed chamber. A tube exits the socket distally and routes into the vacuum pump. As with other suction systems, patients typically report increased proprioception.

Transtibial Biomechanics and Alignment (Endoskeletal Systems)

Biomechanics

The biomechanics of socket fit, alignment, foot function, and suspension are interrelated and highly dependent on each other. Socket biomechanics has already been partially covered in the section on socket design. To complete the discussion, dynamic forces and their relationship to the socket should also be explained. By placing the foot just medial (12 mm) to the socket midline, the medial flare and the shaft of the fibula are loaded during single limb support. These are both excellent, weight-tolerant areas, and foot placement is fine-tuned during dynamic alignment to maximize these regions of the limb. Contours to the pretibial region and the medial aspect of the tibia protect the distal end and crest of the tibia during loading response. At this time, the quadriceps are active, and distal tibial pressures can occur. Foot selection can alter gait and pressures within the socket. Articulated feet, feet with firm hind foot keels, or feet with shorter internal keels all have a different effect on a patient's comfort and gait efficiency. Inadequate suspension will result in a prosthesis that appears too long and will alter swing phase gait. The patient may vault on the sound side to clear the prosthesis. Shear within the socket may cause skin breakdown.

Prosthetic alignment refers to the fixed spatial relationship between the prosthetic socket and foot. With appropriate attachments, it is possible to reposition the foot in all planes. The foot can be moved in a linear fashion anteriorly or posteriorly and medially or laterally in relation to the socket. Prosthetic foot inversion or eversion and foot dorsiflexion or plantar flexion (socket flexion and extension) are also possible. All of these adjustments directly affect pressures within the socket as the patient ambulates.

Overall length of the prosthesis is determined by the length of the pylon used. Length of a transtibial prosthesis is correct when the medial tibial plateau of the residual limb is maintained at the same height as the medial tibial plateau on the sound side (as the patient stands with weight borne equally on both feet). The most convenient indication of correct height is through a comparison of the iliac crests (or the anterior superior iliac spines). If they are not the same level, then the height of the prosthesis should be corrected. This is a general rule, of course, and may not apply if the patient has scoliosis, pelvic obliquity, or congenital leg length discrepancy. Such cases must be taken individually, and often the best indicator of correct height is through gait analysis and patient comfort.

Proper foot rotation is important both cosmetically and functionally. Toe-out refers to the angle between the line of progression and the medial border of the prosthetic foot. A transtibial prosthesis is initially aligned with the medial border of the foot parallel to the line of progression. This position may need to be altered during static and dynamic alignment so that foot rotation on the prosthesis visibly matches that of the sound limb. In addition to appearance, foot rotation affects prosthetic function. How this occurs may be understood if the keel of the foot is viewed as a lever arm. During the stance phase, the tendency of the body to fall over the foot is resisted by the counter force of this lever arm. Therefore, rotation of the foot directly affects the length and direction of force exerted by the lever arm. The net effect of externally rotating the foot is to increase stability by widening the base of support. There is a cosmetic trade-off, of course, because the toe-out attitude of the prosthesis will not match that of the contralateral limb. Although it is rarely necessary to alter toe-out in this way, an awareness of the biomechanical effects of foot rotation is important.

Socket flexion was briefly discussed in the section on PTB design. To reiterate, pressures on the residual limb can be greatly influenced by the inclination of the supporting surfaces in contact with the limb. For example, if a prosthesis was fabricated with the patient's residual limb in 90 degrees of flexion, the shaft of the tibia would be horizontal in relation to the supporting surface of the socket, and the vertical force exerted by the socket would be equal to the weight of the body. The socket is placed in 5 degrees to 10 degrees of initial flexion to increase its supporting surface.

Alignment

Prosthetic alignment will change as the patient's range of motion, muscle strength, and balance improves, and as the patient gains confidence and becomes more physically comfortable in the prosthesis. Intervention by physical therapy is crucial during this early period to aid the patient in developing appropriate gait habits. Prosthetic alignment during early fitting, just after amputation, can be significantly different on the same patient one year later when the limb is well healed and the patient returns to high-level activity. As confidence and ability increase and limb contours change, the socket and/or foot will be replaced, resulting in a need for prosthetic alignment change. Initial static and dynamic alignment goals consider the patient's level of ability, limb length, stage of healing, range of motion, and muscle strength. Use of instrumented gait evaluation has also been beneficial in optimizing prosthetic alignment. The goals in each plane during the first fitting/alignment visit are as follows:

Coronal

- Iliac crests are at the same level. Patient's gait is smooth and symmetrical with no excessive trunk lean to either side.
- Foot is inset loading the proximal-medial and distal-lateral aspects of the residual limb and encouraging an energy-efficient, narrow-based gait.
- Socket adduction or abduction matches the residual limb, resulting in a vertical pylon or a foot that is flat on the floor at midstance. On average, sockets are in 5 degrees of adduction, but some patients may actually require an abducted socket.
- There is less than 6 mm of socket displacement during the swing phase.

Sagittal

- No forced knee flexion or extension when standing. Shoe has even contact with floor and is bearing weight evenly.
- Presence of a smooth, energy-efficient gait pattern—including controlled knee flexion after initial contact-loading response—smooth rollover with no recurvatum tendency, or early dropoff at late stance.
- As previously described, an attitude of initial flexion increases the weight-bearing capacity of all load-tolerant areas. In addition to improving the weight-bearing characteristics, appropriate socket flexion also allows a smooth gait pattern by placing the quadriceps muscles on stretch. Because the knee is maintained in slight flexion, the quadriceps have a mechanical advantage to control the prosthesis. The constant attitude of knee flexion also lessens the possibility of hyperextension of the knee during midstance and terminal stance.

Transverse

- The degree of toe-out on the prosthesis should approximate that of the sound limb. The degree of toe-out should not decrease the patient's stance phase stability.

KNEE DISARTICULATION

Although the anatomical knee joint has been lost, the patient with a knee disarticulation amputation will be somewhat more functional than those with a higher level transfemoral amputation. The intact femur provides a long lever arm with good muscle control and strength because muscle insertions are not compromised. The asymmetrical shape of the distal end prevents socket rotation about the limb and provides a broad, partial weight-bearing surface. Another benefit of the knee disarticulation is that outward flaring of the femoral condyles can often be used to suspend the prosthesis. Control of the prosthetic knee is analogous to transfemoral principles. However, the full-length lever-arm, retention of thigh muscles, and center of rotation close to the knee center all enable the patient with a knee disarticulation amputation to have better control of the knee joint. The stable femur minimizes lateral socket shift and allows for abbreviated socket trimlines, resulting in increased comfort and range of motion. It is unlikely that hip flexion or hip abduction contractures will exist because the musculature remains balanced across the hip. When the prosthesis is removed, it is still possible for the patient to kneel because the residual limb can commonly withstand end bearing.

Socket Design

Socket design for knee disarticulation is often dictated by the degree to which distal weight bearing can be tolerated by the patient, as well as size of the femoral condyles in relationship to the thigh. Although knee disarticulation amputation is able to tolerate partial or full end bearing, the socket should provide total contact over the entire thigh to dissipate pressures over the greatest amount of surface area as possible. Ischial containment socket concepts remain the same, but trimlines and dimensions can be less aggressive than those of the transfemoral prosthetic design, even to the point of totally eliminating ischial containment. The proximal brim of the socket will be near round, terminating just distal to the ischium and pubic ramus to prevent impingement (similar to the old plug-fit transfemoral socket). Weight bearing will be distributed over the entire surface area of the residual limb contained in the socket, as well as over the distal end. Sockets are either thermoplastic or laminated and can include a gel liner or a flexible inner liner with a rigid outer frame. In weight bearing, a thin, firm, resilient pad provides the patient with a distal end pad to increase comfort under the bony condyles. During the early stance phase, the patient stabilizes the prosthetic knee in much the same way as a person with a transfemoral amputation: by actively contracting the hip extensors. Therefore, the rotary forces occurring in the sagittal plane during this gait phase will be greatest over the anterior proximal aspect of the thigh and over the distal posterior end of the residual limb. Consequently, the proximal anterior brim of the socket should provide even, comfortable, counter pressure, and the distal posterior socket should provide total contact and comfortable loading. During midstance, the socket will rotate about the distal end of the knee disarticulation, thus generating counter pressure against the proximal tissues of the thigh. Although the long residual limb provides a greater surface area to disperse these forces, the direction of socket rotation concentrates pressures around the distal femur laterally and the proximal thigh medially. In light of these forces, socket design for this level should facilitate comfortable distal loading, the proximal brim should stabilize laterally, and the medial-proximal brim should flare generously to minimize tissue rolls and avoid painful pressure.

Suspension

The spared femoral condyles offer similar suspension options as the malleoli do in the Syme prosthesis. Supracondylar suspension is obtained with either a pad placed over the flares of the condyles to wedge the limb into the socket or a medial cutout in the distal socket that allows the widest points of the condyles to pass. The cutout is closed to hold the socket in place. A removable medial plate is locked in place over the opening with Velcro (Velcro USA, Inc, Manchester, NH) straps after the prosthesis is donned to securely hold the residual limb in place. A disadvantage of this design is that the opening weakens the structural strength of the socket, and care in fabrication to reinforce this area is necessary. This suspension method works well when the femoral condyles are prominent, but suspension can be compromised by excessive redundant soft tissue or postoperative edema. In such cases, some form of auxiliary suspension is needed. The medial window design is simple and easy to don. The pad or partial sleeve suspension design results in a strong, continuous socket and eliminates circumferential straps about the distal aspect of the socket. It consists of a split cylindrical pad that the patient slips over the lower femur proximal to the condyles, converting the residual limb to a cylindrical shape. Prosthetic socks are used to hold the pad in place. The patient then pushes into the socket until it is firmly seated. Compression of the pad and friction suspend the prosthesis. Because this method does not require a complete liner, no additional bulk is added around the bulbous distal end. Both designs require a socket interface, such as a gel liner, socks, or both. Suspension during swing is maintained by ensuring a tight fit via added sock plies or by increasing the thickness of the suspension pad. If the distal residual limb contours are captured well enough, this can provide suspension and rotational control because the rounded brim cannot be relied on to prevent rotation.

Knee Selection Considerations

The full length of the femur poses a challenge in that virtually any prosthetic knee joint will result in a lower knee center on the prosthetic side. Historically, outside knee joints were incorporated to minimize knee center discrepancy, but their many disadvantages—most importantly lack of swing phase control—make them rarely used today. The knees of choice at this level are those of the polycentric design, and some knees are designed specifically for the knee disarticulation amputation. They decrease the overall prosthetic length of the thigh section, while decreasing the length of the shin section. In addition, some knees incorporate a linkage that, when flexed to 90 degrees, translates posteriorly under the socket, thereby reducing the thigh section length. As the linkage folds up underneath the prosthetic socket during sitting, the foot will often rise off the floor. The thigh segment will still appear somewhat longer when compared with the sound side. These are common issues, and the patient should be made aware of them during the fitting process. The polycentric design also reduces the length of the shank section during swing, thus facilitating toe clearance.[12]

TRANSFEMORAL PROSTHESIS

Socket Design

Ischial Containment

In 1955, Radcliffe[13] described the biomechanics affecting lateral stabilization of the femur and pelvis in the transfemoral prosthesis. These principles are summarized as follows:

- During single limb support on the prosthetic side, the weight of the amputee's body, acting through the center of gravity, causes the pelvis to dip toward the sound side.
- Because of this tendency for the pelvis to rotate toward the sound side, the pelvis can be described as a lever with a fulcrum, or supporting point, located lateral of the ischium.
- The tendency for the pelvis to dip or rotate toward the sound side is resisted by contraction of the hip abductor (gluteus medius), which exerts a counteracting moment to the pelvic lever.
- For the gluteus medius to have maximum effectiveness, it should be maintained at its normal resting length.
- This can be achieved if the femur is kept in a normal position of adduction.
- For the person with a transfemoral amputation, the lateral wall of the socket must be shaped to maintain the femur in adduction because contraction of the gluteus medius causes the femur to stabilize against the lateral wall. Pressures thus generated must be distributed evenly, and excess pressure over the distal lateral end of the femoral shaft must be avoided.
- As a result of these forces acting against the lateral shaft of the femur, counter pressure is generated by the medial wall against the proximal medial tissues of the limb.

These principles have not changed, although the ongoing challenge of managing them has produced a gradual evolution toward ischial containment socket designs and changing theories of prosthetic alignment.

One of the primary goals of the ischial containment socket is to provide medial-lateral stability by controlling the lateral shift of the femur during stance. This is accomplished by the narrow medial-lateral design and closely fitting socket that encases the medial aspect of the ischial tuberosity and ramus.[14] Per its name, the ischial containment socket has the ischium within the socket itself. The socket is just wide enough at the level of the ischial tuberosity to allow the ischium to drop down inside the socket. The portion of the socket that extends proximally along the medial aspect of the ischial tuberosity and the ischial ramus is called the medial containment wall. This wall is angled to match the ischial ramus angle and does not follow the line of progression. Because the medial containment wall extends above the ischium, it will not allow the socket to migrate laterally. A tight dimension spanning the area just below the trochanter and at the medial aspect of the ischium maintains the bony structures snugly against the medial containment wall. In the transverse plane, the ischial containment socket tends to follow the anatomical shape of the proximal thigh.

It has been proposed that the ischial containment socket can provide a bony lock via a three-point force system and increase femoral stabilization in the socket during weight bearing. It is also believed that this three-point force system provides additional means for holding the femur in adduction. In 1989, a study by Gottschalk et al[15] demonstrated that socket design is not necessarily a primary determinant of femoral adduction. This work showed that there was no significant difference in femoral adduction in the socket when comparing quadrilateral or ischial containment designs. The postsurgical length of the adductor musculature had a much greater bearing on proper femoral adduction.

When the socket trimlines in the perineum are aggressively high to capture the ischium and posterior aspect of the ramus, using a flexible inner liner with a rigid frame is recommended. The frame needs only to support or control the bony structures, thus leaving the flexible plastic to contain the anterior ramus and soft tissues. The added feature of the frame cutouts provides comfort, room for expanding musculature, and some proprioceptive feedback.

Ischial Containment Variants

Once the concept of ischial containment was established, there have been varied iterations of this design. Some sockets accentuate and clearly define the functioning muscle bellies, allowing space for muscle contraction. Others lower the posterior aspect of the socket for sitting comfort. This minimizes gluteal support, but renders a better cosmetic result. Each of these designs follows the basic ischial containment principles with additional specific goals. Furthermore, each prosthetist will custom design any socket a bit differently, given the many contours and trimlines possible with the transfemoral socket.

Quadrilateral

If the ischial containment design is not successful, then the more historical quadrilateral socket may be considered. Quadrilateral refers to the four-sided shape of the socket when viewed transversely. As Radcliffe[13] states: "The four walls of the socket are designed to apply pressures and counter pressures to facilitate comfortable load bearing through soft tissue and underlying structures." The ischial tuberosity and gluteal musculature are used as primary weight-bearing structures and are supported by a posterior shelf. The ischium is maintained on the shelf or seat by the counter force of an inward contour (Scarpa's bulge) in the anteromedial socket located over the femoral triangle. Although the femoral triangle contains the femoral artery, vein, and nerve, years of clinical trials have demonstrated that correctly distributed pressure over these structures is well tolerated. A convex contour in the proximal anterolateral socket accommodates the bulk and the contraction of the quadriceps muscle group. The quadrilateral socket may be suspended in a number of ways, including use of suction, a Silesian belt, or a hip joint with a pelvic belt.

Suspension

Suction by Skin Contact

A suction socket offers intrinsic suspension and is held in place by a combination of negative air pressure, surface tension between the socket and the patient's skin, and muscular contraction against the socket walls. For the socket walls to be in direct contact with the patient's skin, no socks are worn. A valve, usually located at the distal-medial aspect of the socket, allows air to escape from the socket during weight bearing while preventing air entrance during the swing phase. The patient generally dons a suction socket by removing the valve, applying a donning sleeve over the residual limb, and feeding the loose end of the material through the valve hole. The material breaks the surface tension between the socket and the patient's skin, allowing the patient to pull the limb comfortably into the socket. By using the donning sleeve to help pull skin and soft tissue into the socket, the patient can usually minimize problems caused by tissue overhanging the socket brim. Patients with a long, muscular limb may choose to push directly into the socket with a small amount of skin lubricant. Either way, it is critical that no distal air pockets remain. If there is lack of total contact distally in a suction socket, the high negative pressures created in the air space can result in circulatory congestion and edema. There are many advantages to selecting suction suspension. The most apparent advantage is elimination of belts and liners. Thus, the patient enjoys an unencumbered, lightweight, and relatively simple-to-apply suspension. The socket is a total contact fit that minimizes the possibility of edema. Suction is a solid form of suspension because once the skin is drawn down into the socket, the residual limb is effectively trapped in the socket, thereby minimizing vertical displacement. Because the socket is firmly attached to the patient's residual limb, and rotational control is provided by the brim design, patients usually report that they have much better control of the prosthesis when using this form of suspension, especially during the swing phase.

One noted contraindication to a suction socket is that limb volume must remain stable to maintain a negative pressure environment. It is common for a patient's residual limb to change volume over time, especially within the first year postsurgery. Even the residual limb of a longtime wearer can decrease in volume over the course of the day. Slight fluctuations are normal and generally not problematic, but those seen just after surgery make this suspension option inappropriate. If the patient presents with scarring, which extends across the proximal tissues of the residual limb, it is possible that those scars will break the air seal around the proximal socket and allow air to enter. As a result, suction is not maintained. Donning the suction socket requires strength and balance. If the patient does not possess sufficient ability and endurance, or has upper extremity involvement, another suspension mechanism must be used.

Sealing Liners

Suction is the suspension of choice because it eliminates belts and buckles. To address some of the aforementioned contraindications, a sealing liner as presented in the transtibial section is a good choice. A gel liner with a hypobaric seal is rolled onto the limb and then pushed into the socket. The seal is pressed against the socket wall, and suction is maintained below the level of the sealing ring. A thin sheath is worn below the level of the ring to wick air out through a one-way, push-button release expulsion valve.

Locking Gel Liners

When the two previous suspensions cannot be used, gel locking liners may be considered. The locks can incorporate pins, lanyards, or buckles. The goals of the gel locking liner for the transfemoral design are similar to that of the transtibial socket, with a primary goal to suspend the prosthesis and distribute pressure over the entire surface of the residual limb. As in the transtibial prosthesis, the silicone or gel type liner can provide a significant amount of protection for the residual limb. Locking liners can be applied to virtually any length of transfemoral residual limb. If a locking liner is planned for a long transfemoral or knee disarticulation amputation level, then a lanyard-type suspension may be required. This does not require the use of a long locking mechanism and keeps the prosthetic knee center at or just slightly below the anatomical knee.

The most common advantage for the gel locking liner is an added measure of protection for the residual limb. The use of a liner can provide reduced shear on the residual limb and thus increased protection for fragile tissues. The liner can accommodate varying circumferences of the residual limb, thus ensuring total contact in the socket. As a result, the patient is much less likely to experience the failure of suspension because of suction loss. As an added benefit, donning the locking liner can be much easier than the skin suction socket because the patient is not required to lean over and pull into the socket. Typically, this locking liner suspension only requires that the patient engage the pin or pull up on a lanyard while stepping into the socket. Another significant advantage for the transfemoral patient, who tends to experience volumetric changes of the residual limb, is that the patient can apply prosthetic socks over the locking liner. The socks must have a hole at the distal end, which allows the pin to pass through and engage into the locking mechanism of the distal socket. As the patient's residual limb volume decreases, the sock ply can be increased accordingly and vice versa.

Silesian Belt

The Silesian belt is a simple, low-profile, belt-type suspension mechanism. Indication for its use is primarily the need for rotational control of the prosthesis. The Silesian belt can be used as an auxiliary to suction suspension and, the in the case of an individual with weak adductors, can even assist the adductor muscles during swing phase. The belt sits firmly over the sound side ilium and may not be as effective with the obese patient. Scarring about the pelvis can also prevent its use because the belt creates friction in this area. If the patient presents with severe mediolateral instability, the hip joint and belt are better choices. The Silesian belt can be easily used with prosthetic socks to suspend the prosthesis while also accommodating residual limb volume fluctuation. The Silesian bandage is an excellent auxiliary suspension for a suction socket fitting for shorter levels. Support of the belt helps to prevent the socket from slipping off when the patient is seated. It also provides additional security during any activity.

Elastic Belt

An elastic suspension belt is fabricated of Neoprene (or polychloroprene; DuPont Performance Elastomers LLC, Wilmington, Del) or similar elastic material. Typically, this kind of belt is fastened anteriorly by Velcro (Velcro USA, Inc, Manchester, NH). Because the belt is made of elastic material, it is not ideal as a primary suspensor. The belt can allow rotation and elongation of the prosthesis. Because of its simplicity, however, it is not uncommon for an elastic belt to be used as an intermediate form of suspension or as an auxiliary suspension (along with suction). If the patient is involved in high-level activities, there might be a concern, on occasion, of losing suction. For those high-level activities, the patient may choose to temporarily apply a belt to the prosthesis in the event that suction suspension is lost. With its simple Velcro closure, it is very easy to don.

Hip Joint and Pelvic Band

Whereas the Silesian belt is generally considered an auxiliary suspension, the pelvic belt is a primary suspension. This system includes a hip joint attached to the proximal lateral socket, a metal or plastic pelvic band contoured to the patient's pelvis just below the iliac crest, and a leather belt that encircles the patient's pelvis. The hip joint and pelvic band are typically considered a suspension means of last resort. Accurate placement is critical to proper function. If any other suspension can be successfully used, it should be explored first. A short transfemoral residual limb may be too short or may not generate enough abductor force during single limb support on the prosthetic side to provide a level pelvis. The added leverage offered by the hip joint and extension up to the pelvic band can usually provide adequate medial-lateral stability to the hip. This suspension system is very simple to use. If the patient can use an ordinary belt, then usually this form of suspension can be adequately donned, thus allowing the amputee to wear socks so that volumetric changes can be easily accommodated. The hip joint and pelvic band are obviously bulky and heavy. This can restrict hip range of motion and make sitting less comfortable.

Transfemoral Biomechanics and Alignment (Endoskeletal Systems)

The smooth and safe function of a transfemoral prosthesis is dependent on the relation between socket, suspension, knee, and foot. The socket is positioned in all planes for optimum function. Generally, 5 degrees of socket flexion will place the hip extensors at a mechanical advantage, ready to resist a knee flexion moment at heel contact. If a hip flexion contracture is present, additional socket flexion will be necessary. A flexion contracture of up to 25 degrees may be accommodated by prosthetic alignment. Anything beyond 25 degrees will cause excessive lumbar lordosis, leading to possible low back pain. Knee stability is determined by volitional control, the mechanics of the knee itself, and, most importantly, alignment. The knee is positioned at or just behind a line connecting the trochanter and ankle (also known as the TA line). Moving the knee fore or aft, even a few millimeters, can have significant impact on the overall safety of the prosthesis. In the coronal plane, the adduction angle of the sound side is matched, unless there is a limitation at the hip preventing full adduction. The posterior brim of the socket is divided into thirds, and the foot is placed under the medial third mark. This provides good lateral loading of the femoral shaft during stance. A shorter limb or weak hip abductors might require that the foot be placed a bit more laterally under the midsocket to reduce the forces along the lateral aspect of the socket. In the transverse plane, the medial containment wall is rotated to follow the angle of the ramus. The prosthetic knee is set into 5 degrees of external rotation to the line of progression, and the foot is positioned to match the patient's sound side foot rotation. Alignment of prosthetic components can be refined by gait lab analysis using center of mass instead of the TA line.

Alignment

As described in the section on transtibial prostheses, the prosthetic alignment will change as the patient progresses through the first year postsurgery, especially during early therapy treatment. Improvements in strength, balance, comfort, and confidence will periodically make reassessment of alignment necessary. Much attention is given to knee stability. Not only does moving the knee in relation to the ground reaction line affect stability, but also internal knee adjustments to resistance can reduce the likelihood of unexpected knee collapse and yield a smooth and efficient gait pattern.

Initial static and dynamic alignment goals take into consideration the patient's level of ability, limb length, stage of healing, range of motion, and muscle strength. The goals in each plane during the first fitting/alignment visit are as follows:

Coronal

- Iliac crests are at the same level.
- Patient's gait is smooth and symmetrical with no excessive trunk lean to either side.
- Foot is inset-loading the lateral femur during stance, encouraging a narrow-based and energy-efficient gait.
- The foot/shoe has even contact with the floor and is bearing weight evenly.
- The socket is well suspended and does not drop away from the residual limb.

- There is no vaulting on the sound side.
- There is no abducted or circumducted gait.
- Steps are equal.
- Lumbar lordosis is not excessive.

Sagittal

- The prosthetic knee is stable at initial contact/loading response.
- During terminal stance, the foot rolls without dropping off or causing pelvic rise.
- The plantar flexion bumper of the articulated foot is firm and does not allow foot slap.
- Heel rise is not excessive.
- There is no terminal impact.

Transverse

- The degree of toe-out on the prosthesis should approximate that of the sound limb. The degree of toe-out should not decrease the patient's stance phase stability.
- There is no medial or lateral whip at terminal stance.
- There is no rotation of foot at initial contact/loading response.

HIP DISARTICULATION AND TRANSPELVIC AMPUTATION

These levels of amputation are almost always treated with the lighter weight endoskeletal prosthesis, given their lightweight properties, ease of fabrication, and postfabrication adjustability. Knee components, feet, and alignment are selected to enhance stability and ease of use. Energy required to ambulate with these prostheses is 200% times greater than normal.[16]

Hip Disarticulation Socket Design

Originally developed as the Canadian hip disarticulation prosthesis, this prosthesis incorporates both the sound and affected side iliac crest contours into the socket design. The socket is intended to bear weight on the ischium; however, this weight is shared along the anterior torso and the posterior gluteal muscle area by sloped anterior and posterior walls lifting the torso off the ischium to some degree. Primary suspension is provided over the affected side ilium. The contralateral ilium provides only minimal support during swing phase. Trimlines extend proximally above the ilium, and socket contours accentuate the narrower waist groove. During swing phase, the weight of the prosthesis is borne here. One socket feature that facilitates donning and doffing the prosthesis is a flexible lamination located about the posterior midline. Flexibility at this location allows the patient to spread the rigid, supporting socket open and wrap it around the pelvis. Because of the asymmetric contours of the pelvis and the intimate socket fit, rotational stability of the socket is good.

A two-piece socket design was developed at the University of California, Los Angeles.[17] By splitting the socket into two parts and fastening them together with a crisscrossing strap design posteriorly, this design can provide support and suspension during stance, as well as swing phase. During midstance, as the weight is borne into the socket, the strap system draws the inferior aspect of the sound side portion of the socket medially. This holds the patient firmly into the prosthetic socket while easing up on the iliac crest. During swing phase, the prosthesis attempts to move inferiorly on the patient's pelvis. The contralateral portion of the socket provides increased pressure over the iliac crest, increasing swing phase stability of the socket on the patient. This floating sound side design is more comfortable for the patient and is easier to fabricate for the prosthetist.

Transpelvic Socket Design

Transpelvic amputation is completed at the junction of the sacrum and ilium. The primary consideration with the transpelvic socket design is that the patient's weight is borne against the socket wall through soft tissues alone. The transpelvic socket extends to the body midline and encompasses the lower rib cage bilaterally, with the abdomen and the pelvis on the sound side. The socket may extend superiorly as high as the tenth rib. Inferiorly, the socket must clear the sound side limb and genitalia. On the sound side, the trimlines are brought inferior just proximal to the trochanter. In general, by encompassing more of the rib cage, the stability of the socket on the lower trunk is improved. There is generally less rotation of the pelvis and the socket because no ischial tuberosity exists that forms a fulcrum, as in the case of hip disarticulation. Raising the trimlines also increases the available surface area over which weight can be borne. The concave contour of the socket matches the convex contour of the residual limb. The socket wall generates forces oriented obliquely upward on soft tissue and the sound side pelvis.

Hip and Knee Joint Considerations

There are numerous endoskeletal components available to be used in the hip disarticulation prosthesis. Hip joints are single axis requiring an extension stop and extension assist. Single axis or polycentric knee units can be incorporated into the prosthetic design. A polycentric knee with a mechanical stance phase lock with extension aid in reaching full knee extension is a good choice for this prosthesis. The polycentric feature shortens the shank during swing phase, making it easier to clear the ground. In the polycentric design, the knee joint center resides more proximally than a single axis design. This provides the patient with an added biomechanical advantage in both initiating and resisting flexion because of the proximity of the instantaneous center of rotation to the anatomical hip joint center. Also ideal is a microprocessor knee. Both designs offer excellent stance phase stability. Careful consideration of overall prosthetic weight versus function and stability is necessary for optimal outcome.

Hip Disarticulation/Transpelvic Biomechanics and Alignment

The hip joint is placed far anterior on the socket relative to the normal hip joint location. The knee axis also falls considerably posterior to the anatomical knee. A combination of these two joint positions creates a significant knee extension moment. The knee hyperextends until an extension stop is contacted. This particular alignment provides an extremely secure knee, making it difficult for the knee to buckle inadvertently. If the center of gravity remains within boundaries formed by vertical reference lines placed at both the hip and knee centers, the hip and knee joints will remain stable. If the patient's center of gravity falls behind the posterior line located at the knee joint, this creates a knee flexion moment and causes the knee to buckle. If the center of gravity falls anterior to the hip joint, this creates a hip flexion moment and causes the socket to flex at the hip joint.

Alignment

A line projected through the hip joint axis and the knee joint axis should intersect with the ground approximately 1½ inches posterior to the heel. The theoretical weight line begins at the point in the socket 1 inch anterior to the ischial tuberosity. It extends vertically to the ground passing ½ inch anterior to the knee center and intersects with the ground at approximately $^{7}/_{16}$ of an inch anterior to the center of the prosthetic foot. In proper coronal plane alignment, one proposed method places the hip joint 10 mm lateral to the lateral ¼ mark. In the transverse plane, the hip joint should be externally rotated 5 degrees to 10 degrees for a more anatomical swing phase action of the thigh and shank. The length of the prosthesis should be approximately ½ inch shorter than the sound side to provide adequate clearance of the toe during gait.

Gait

The gait pattern for these individuals to properly ambulate requires that the patient use atypical motions to generate appropriate forces to advance the prosthetic limb. These forces include vaulting on the sound side, hip hiking on the prosthetic side during swing, and posterior tilting of the pelvis at late stance on the prosthetic side to initiate hip and knee flexion. As patients become more proficient, these motions are minimized, but nevertheless are present during gait. The gait of these individuals requires a great deal of energy expenditure to facilitate balance and advance the prosthetic limb. The energy costs can be twice that of normal human gait. Comfort is another issue for these individuals. Whether sitting, standing, or ambulating, it is important that socket contours and trimlines provide the greatest comfort to the patient.

COMORBIDITIES AFFECTING THE REHABILITATION PRESCRIPTION

Multiple Limb Loss

Bilateral lower extremity amputations require special consideration related to prosthetic design and alignment. Socket design remains the same; however, accommodative gel liners are recommended because the patient is not able to shift weight to a sound side foot. This is particularly important for the more bony transtibial amputation. Torsion adaptors reduce anatomical knee and hip joint torque and shear at the skin/socket interface. Any additional components placed as close to the sockets as possible reduce the perceived weight of the prostheses. Constructing the prosthesis as light in weight as possible helps the patient keep energy costs down. An alignment that moves the feet slightly more lateral or increasing toe-out gives a wider base of support. This decreases the coronal plane forces placed on the limb during weight bearing. A wider base also results in additional stability. Foot selection can influence prosthetic knee stability and ground conformance. For bilateral transfemoral patients, articulated feet—or those with a softer heel or hind foot keel—move the weight line anterior to the knee joint more quickly, stabilizing the knee in early stance. Regardless of the level of amputations, identical foot components should be used for consistent gait mechanics. The use of stubby prostheses, sockets without knees, keeps the patient's center of gravity low to ground and helps the patient gain balance, strength, and confidence early on. Incremental height increases and the addition of knees, added one at a time if necessary, bring the patient to, or near to, preinjury height. If only one knee at a time is added, leaving the other side with a straight pylon, then the longer limb or the limb with better muscle control is fitted with a knee first. Progress requires commitment by the patient and an intense physical therapy regimen. To facilitate good posture and a well-placed base of support, feet may be articulated and perhaps mounted backward. Foot blocks with rocker design are also common. Knees for the bilateral transfemoral patient should offer good stability. Polycentric design, fluid control, and microprocessor control all offer stability and can be aligned for maximal safety. The polycentric knee shortens through swing phase, a useful feature, when the opposite limb is a transtibial amputation. Marking liners, socks, valves, etc, as right and left is helpful when donning.

When upper limb amputations are also present, the most pressing issue is one of independent donning. Donning devices can be designed for individual needs, and easy suspension options (eg, gel liners with locking pins or sealing liners) can be used.

Bone

Heterotopic ossification is a concern for amputations resulting from blast injuries.[18] Socket modifications, a soft gel interface, and/or socket replacement help to manage and accommodate the changing limb contours. External fixators on the amputated side or the contralateral limb do not preclude prosthetic fitting. The prosthetic management goal is to get the patient ambulatory quickly. Sockets can be modified to distribute weight away from the fixator, and/or cutouts can be made around the device. Prosthetic knees offering maximum stability will provide safer ambulation.

Soft Tissue

Shrapnel fragments, foreign bodies, and other detritus may migrate until they are just under the skin surface. Often they are expelled. This process requires close monitoring and ongoing socket modification. Neuromas are treated prosthetically in much the same way because they are painful and limit prosthetic use.

Skin

Grafted skin or excessive scar tissue can be a challenge, and a comfortable fitting socket can be difficult to achieve. Liners made of softer rather than firmer gel protect fragile skin. Alignment intended to reduce forces on the limb is also helpful. Pain as a result of fragments in the soft tissue and under the skin may be managed with soft gel liners as well. Occasionally, shrapnel can work its way through the skin.

Joint Function and Condition

Limited range of motion, contractures, and/or loss of muscle function require alignments and component selections that accommodate these situations. Fall prevention and socket comfort are the foremost considerations. If alignment alone does not produce the desired results, then components providing either additional range of motion (eg, articulated feet) or knee components with added stance control can be selected.

Blindness

Donning is a consideration when the patient is blind. Tactile markers on the prosthesis and its components help the patient to orient the gel liner, socks, or belt for proper donning. Ratcheted locking pins emit audible clicks as they engage into a shuttle lock confirming positive suspension.

THE PROSTHETIC REHABILITATIVE PROCESS

The prosthetic phases (preprosthetic, initial prosthetic, basic prosthetic, and advanced prosthetic) mirror the timelines set by the Physical Therapy Service for treatment protocols for amputees.

Phase I: Preprosthetic (Weeks 1–4)

Care typically includes prosthetic plan formulation, patient education, wound management, and volume control through the use of shrinker socks, silicone or equal liners, or elastic wraps.

Phase II: Initial Prosthetic (Weeks 2–10)

Care typically includes the provision of a prosthetic limb with multiple replacement sockets as warranted**.** The use of early postoperative prosthesis (EPOP) is common at military MTFs. EPOPs comprise custom-designed sockets fabricated from ThermoLyn PETG (Otto Bock HealthCare, Minneapolis, Minn) material attached to advanced prosthetic components. The components may consist of a C-Leg (Otto Bock HealthCare, Minneapolis, Minn) microprocessor knee unit for knee disarticulation level and higher, and a dynamic response or stored energy foot. EPOP sockets are replaced at an extraordinary rate. Standard practice is to maintain an appropriately fitting socket at all times. As service members progress through the rehabilitation process, their limbs undergo changes in size and shape. These changes are atypical when compared with the geriatric dysvacular amputee, in which shape and size reductions are typically uniform. For the service member who has sustained a blast injury, limb size and shape reduction are not uniform, in particular for those with heterotopic ossification.

CAD/CAM is an excellent tool to use in an acute care setting for a variety of reasons. It does not replace the skill and experience of the prosthetist; rather, it enhances prosthetic services by decreasing delivery times and increasing efficiency. It provides a way to objectively document limb morphology and volume and to track limb change. Residual limb shape capture is accomplished by noncontact scanners, digitized passive casts, measurements, or photos dependent on the particular CAD system used. The software to manipulate the shapes is intuitive and relatively easy to use. Output is to a milling machine for fabrication of a positive model. Standard fabrication procedures of the prosthetic socket are followed at that point.

Use of microprocessor knee units as first-fit systems are not uniform at all military MTFs. Safety is paramount, and the facilities rely on a team approach for prescription rationale of such devices. In general, all of the military MTFs are prescribing microprocessor units for those with bilateral lower extremity amputation and/or those with lower and upper extremity amputations.

Phase III: Basic Prosthetic (Weeks 11–20)

Care includes continued socket replacements, also the introduction of dynamic response or stored energy prosthetic feet as indicated. Progression to more dynamic feet should coincide with the patient's progress in physical therapy. Sockets may be fabricated from definitive materials (eg, carbon graphite and acrylic resin) as indicated.

Phase IV: Advanced Prosthetic (Months 3–6)

Care includes the design and delivery of specialty prostheses to include, but are not limited to, sport use prostheses. These prosthetic limbs should be fabricated from definitive materials.

ACTIVE DUTY MILITARY, LOWER LIMB PROSTHETIC PROTOCOLS

Recommendations for Daily Use Prosthesis

1. Sockets
    a. ThermoLyn or its equal until ready for definitive socket.
    b. Definitive socket materials (eg, carbon graphite and acrylic or epoxy resin) prior to high-level sporting/recreational activities.
    c. Definitive socket for transfemoral amputees: flexible socket with rigid frame.
    d. Hip disarticulation: start with socket; add custom liner/suspension system, depending on individual limbs; consider using three-dimensional modeling to help guide design as indicated, especially with secondary problems (eg, heterotopic ossification).
2. Suspension
    a. Suction suspension for all amputees unless meets criteria for pin lock or lanyard.
    b. Pin lock or lanyard for amputees with anticipated large-volume changes or when suction is contraindicated.
    c. Auxiliary suspension added as needed for higher level activities.
3. Feet
    In general, limit three different feet for each amputee during the first year. Integrate gait lab to augment clinical decisions.
    a. Transtibial
        i. Use soft, multiaxial, easy, rollover, and shock-absorbing properties to promote early weight bearing and weight

shifting (eg, Talux [Ossur Americas, Aliso Viejo, Calif] or TruStep [College Park Industries, Fraser, Mich]).

ii. Progress to a higher energy returning foot, such as Ceterus (Ossur Americas, Aliso Viejo, Calif), Renegade (Freedom Innovations, Irvine, Calif), Re-Flex VSP (Ossur Americas, Aliso Viejo, Calif), Modular III (Ossur Americas, Aliso Viejo, Calif), or low-profile equivalents. Can have a second, high-energy returning foot if properties of new foot are different from existing foot.

iii. Consider microprocessor foot or other advanced power prosthesis, depending on individual goals and activity levels after they have mastered above feet for a minimum of 3 months or as otherwise decided by the rehabilitation team.

b. Transfemoral
Same as transtibial level.

4. Specialty Prosthetic Limbs
Lower extremity amputees may also receive a water/swimming leg and running leg.
 a. The water leg may be used for bathing, as well as for other water activities. Modifications to be used for swimming may be added if the amputee expresses such interest.
 b. Running-specific legs will be prescribed when clinically appropriate for amputees who express the desire and motivation to return to more frequent and longer distance running than that accommodated through the use of the dynamic response foot. The primary goal will be to help the service member return to Army physical fitness standards. Providers must be aware that conditions such as osteoporosis, nerve injury, fractures, etc, might be contraindicated to aggressive running with a running specialty leg. Use of a dual-energy X-ray absorptiometry scan (or DEXA scan) is warranted to determine osteopenia. Furthermore, appropriate training with an experienced therapist is essential to successful and safe running on a running specialty leg. All running legs will be fabricated with a definite socket. Guidance is to start with a distance foot (eg, FlexRun [Ossur Americas, Aliso Viejo, Calif] or Cheetah [Ossur Americas, Aliso Viejo, Calif]) and progress to a sprint foot if the patient demonstrates proficiency.

5. Knees
One microprocessor; need to have gait assessment first with the C-Leg, then gait analysis after 4 weeks of training with the RHEO KNEE (Ossur Americas, Aliso Viejo, Calif) and/or the POWER KNEE (Ossur Americas, Aliso Viejo, Calif).

SUMMARY

This chapter focused on the prosthetist as a member of a comprehensive, clinical rehabilitation team at military MTFs. Also reviewed were the unique role of prosthetists during combat deployments, as well as the lower limb prosthetic technology provided to combat-injured service members who sustained an amputation. The prosthetist is an integral part of the multidisciplinary rehabilitation team, often working in unison with other team members, yet also taking center stage during formulation of the prosthetic limb prescription. During the global war on terror, prosthetists played a unique role in worldwide deployments. They provided strategic and operational knowledge to allied governments in the areas of prosthetic clinical education, clinical evaluation, and prosthetic fabrication techniques. The increasing complexity involved in prosthetics technology and the development of unique materials, computer programs, and fabrication techniques applicability require the specialized knowledge of the prosthetist. This chapter provided an overview of lower limb prosthetic technologies, ranging from classification of prosthetic feet, knee units, socket design, socket interfaces, and components to how these technologies related to the amputation levels of the lower limb. The prosthetic rehabilitation process used at military MTFs and the comorbidities affecting the rehabilitation prescription were also examined.

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