CHAPTER 33

DESIGN OF ARTIFICIAL LIMBS FOR LOWER EXTREMITY AMPUTEES

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33.1 OVERVIEW

Artificial limbs for lower extremity amputees are designed and fabricated by prosthetists. These limbs include custom interfaces between the residual limb and prosthesis (prosthetic socket) and commercial components specifically selected for the individual. Biomedical engineers have been involved in the design of some of these commercial components and in the quantitative evaluation of these prostheses and the performance (e.g., walking and running) of the amputee with his or her prosthesis. Biomedical engineers have also been involved in the development of computer-aided design (CAD) and computer-aided manufacture (CAM) of these limbs. Future opportunities for biomedical engineers include continued development and incorporation of strong, lightweight materials in lower extremity prosthetic limbs, high-technology prosthetic components that improve lower extremity amputee performance, and prosthetic sockets and interface materials that minimize risk of dermatological breakdown of residual limb tissues. Additional opportunities involve the enhancement of CAD-CAM systems and technology, assessment of lower extremity prosthetic fit, evaluation of lower extremity amputee function, and development of sensors and technology to more reliably produce comfortable sockets and optimally align prostheses.

33.2 HISTORY OF LIMB PROSTHETICS

The earliest limb amputations generally resulted in death due to blood loss or infection. The amputation was a “guillotine” operation such that all tissues were divided at the same level in one stroke of the blade. Bleeding vessels were immediately cauterized with heated irons or boiling oil. The first recorded successful amputation dates back to 484 B.C., when Hegesistratus reportedly escaped from...
prison by cutting off one of his feet. (He built himself a wooden foot to compensate for the limb loss.)

The use of anesthesia (ether—Long, 1842; chloroform—Flourens, 1847) and antiseptics (Lister, 1860) significantly improved amputation surgery. Other medical advances contributing to the reduced morbidity of amputation included the use of ligatures or sutures (originally described by Hypocrates, reintroduced by Pare in 1529) to cauterize blood vessels and the tourniquet (von Gersdorff, 1517; Morel, 1674; Faby, 1853).

In addition to resourceful amputees, early prosthetists were blacksmiths, armor makers, and other skilled artisans. Artificial limbs in Europe, and later in America, used metal, wood, and leather. Notable lower extremity prostheses include the first transtibial prosthesis with an unlocked knee, a thigh corset and external hinges introduced in 1696 by Verduyn, a Dutch surgeon and the first transfemoral prosthesis with an articulating ankle and knee, the Anglesey leg (Fig. 33.1), introduced by Potts in 1816. This latter leg was introduced in the United States in 1839. The first artificial limb shop in the United States was opened by Hanger in Richmond, Virginia, during the Civil War. Hanger is also credited with replacing the plantar-/dorsiflexion cords in the Anglesey leg with bumpers. In 1863, Parmalee introduced a transfemoral prosthesis with a suction socket, polycentric knee, and multiarticulated foot (Fig. 33.2). Other historical prosthetic developments include the introduction of the Canadian hip disarticulation prosthesis in the United States (1954, McLaurin), the solid-ankle, cushioned-heel (SACH) foot (1956), and the patellar-tendon-bearing (PTB) prosthesis (1958, University of California at Berkley). The first hydraulic knee, the Hydra-Cadence leg, was introduced by Stewart-Vickers in 1960.
As alluded to earlier, war resulted in many amputees and motivated investment in prosthetic development and training. In 1870 (after the Civil War), Congress passed a law providing artificial limbs to all persons honorably discharged from the U.S. military or navy who had lost a limb in service. This law also entitled these individuals to a new prosthesis every 5 years. In 1945 (after World War II), in response to veterans’ demands for more functional prostheses, the National Academy of Sciences (NAS) initiated a study to develop design criteria to improve function for artificial limbs. This Committee on Artificial Limbs (CAL) contracted with universities, industry, and healthcare providers in this initiative. The Veterans Administration established prosthetic laboratories in New York, the University of California at Berkley (lower extremity), and the University of California at Los Angeles (upper extremity) in 1947. In 1949, the American Orthotics and Prosthetics Association developed educational criteria and examinations to certify prosthetists and orthotists. From 1947 to 1976, NAS sponsorship and support from the Veterans Administration, CAL, the Committee on Prosthetics Research and Development (CPRD), and the Committee on Prosthetic-Orthotic Education (CPOE) influenced the development of prosthetics and orthotics. (A detailed history of lower extremity amputation and prosthetics is summarized in Chapter 2 of Sanders’ text.1)

33.3 AMPUTATION SURGERY

Amputation is defined as the removal, usually by surgery, of a limb, part, or organ. Its purpose is to remove dead or diseased tissue, to relieve pain, to obtain healing, and/or to rehabilitate the individual. If the amputation level is properly selected and the surgery is performed well, amputation should be considered not as a salvage procedure but as a rescue procedure that is the first step in rehabilitation.

Causes of amputation include trauma (loss of arterial supply, avulsion, and thermal injury), disease, and congenital deformity. Disease categories leading to amputation include vascular or circulatory disease (e.g., arteriosclerosis, diabetes mellitus, Buerger’s disease), cancer, and infection. Since the United States does not have national healthcare, accurate records regarding the incidence and level of amputation are unavailable. However, random polls by the U.S. Public Health Service indicate that there are approximately 1.53 amputees per 1000 population, not including patients in institutions. A 1991 U.S. National Health Survey indicates that the total number of amputees in the United States is 1.5 million, with 75 percent of these amputations (lower and upper extremity) attributed to disease, 23 percent to trauma, and 3 percent to birth defects. Most lower extremity amputations are performed secondary to peripheral vascular disease (Fig. 33.3), with primary incidence among the 61- to 70-year age group. Lower extremity amputation is often preceded by attempts at limb salvage through revascularization procedures. Although the data are dated (pre-1975), such data indicate that 10 percent of lower extremity amputees lose their other leg within 1 year, 20 percent within 2 years, and 33 percent within 5 years.

The level of amputation is selected based on the potential for healing and future function. Vascular surgeons, who may or may not have special training in prosthetic rehabilitation, perform most amputations. Orthopedic surgeons, who complete a course in prosthetics as part of their residency, perform amputations necessitated by trauma, malignancy, and other nonvascular causes. Factors influencing the level of vascular amputation include the local blood supply, the nutritional state of the patient, and the probability of successful fitting. Level selection for traumatic amputations or major tumors is based on the nature of the injury and the viability of the remnant tissues.

In the management of acute and chronic ischemia, the vascular status and potential need for amputation are evaluated, followed by assessment of the level of amputation. The primary determinant in assessing the level of vascular amputation is the adequacy of skin blood flow. Assessment measures include segmental blood pressure using Doppler flowmetry, transcutaneous oxygen tension, quantitative skin fluorescence, isotope clearance, and/or laser Doppler and maintenance of normal skin temperature. (Moore, Chaps. 4 through 9, reviews relevant...
methodological details and the diagnostic utility of these techniques and measures.) Amputation levels of lower extremity (Fig. 33.4) include

- Partial foot: transmetatarsal, metatarsal disarticulation (Lisfranc), disarticulation between the talus and the rest of the foot (Chopart), and transcalcaneal amputations (Pirogoff, Boyd)

FIGURE 33.3 Trends in lower extremity amputation. These data are from Denmark, but comparable trends are expected in other developed countries. (Adapted from Ref. 16, Table 1.1 and Figs. 1.3 and 1.4.)

FIGURE 33.4 Lower extremity amputation levels of the foot (left), leg (center), and pelvis and femur (right). (Adapted from Ref. 12.)
- Ankle disarticulation or Symes amputation
- Transtibial or below-knee amputation
- Knee disarticulation
- Transfemoral or above-knee amputation
- Hip disarticulation
- Hemipelvectomy

The two primary goals of amputation surgery are the ablation of the diseased or traumatized tissues and reconstruction of the remnant or residual limb. Generally, surgeons want to save as much length as possible while providing a residual limb that can tolerate stresses induced by the prosthesis and ambulation. To minimize the incidence of large, painful neuromas, the major nerves are pulled firmly distally, resected sharply, and allowed to retract into the soft tissue. Amputation is typically a closed technique in which skin flaps are used for primary closure. In cases of infection or when all devitalized tissue has not been removed by debridement, open amputation may be performed. Skin traction is then used to pull the skin and muscle distally over the end of the bone until the wound heals.

When a muscle is severed, it loses its distal attachment. Without attachment at both ends, a muscle is unable to function. If left loose, the muscle will retract, atrophy, and scar against adjacent structures. To improve future muscle function, surgeons have investigated both myodesis and myoplasty to secure the distal muscle. Myodesis involves attaching the distal muscle to the bone. While myodesis attempts to prevent excessive muscle shortening and resulting muscle weakness, it is not commonly practiced because it may “tether” the stump and contribute to bone spur formation. Instead, myoplasty is often performed. The muscles are regrouped about the bone to create distal padding. This procedure secures the distal muscle and improves future muscle function. It also serves to counter the tendency for development of joint contractures.

In addition to the loss of motor function of the amputated joints, the amputee is also deprived of sensory information present in the intact limb. This lack of sensory information is important in the lower extremity but is likely more important in the upper extremity, where lack of sensation is a major factor limiting the effective use of artificial hands and hooks.

Because the artificial limb is attached to the residual limb by a prosthetic socket that encompasses the remnant limb tissues, there is a false joint between the skeletal system and the prosthesis. This insecure attachment often results in perceived instability and uncertainty in control of the prosthesis. The unstable attachment to the skeletal system contributes to the perception that the artificial limb is heavier than the intact limb, even though it may, in fact, be considerably lighter.

The prosthetic socket is fitted over tissues that do not normally bear weight. As such, secondary problems are often observed, including edema from a socket that is too tight proximally, osteoporosis due to reduced skeletal weight bearing, allergic reactions to the socket (or socks, inserts), reduced blood flow, and development of bone spurs, cysts, and/or infections. In addition, many amputees experience phantom sensation (awareness of missing limb, often described as tingling, pressure sensation, or numbness) and/or phantom pain (cramping or squeezing sensation, shooting or burning pain in the missing extremity). Phantom sensation is frequently experienced, whereas phantom pain is reported less commonly.

### 33.4 Prosthetic Clinic Team

Following amputation surgery, the patient is often referred to physical therapy for education regarding proper limb positioning and residual limb care and to a prosthetic/amputee clinic. The clinic team typically includes a physician, a physical therapist, and a prosthetist. It may also include a social worker and/or vocational counselor. These multidisciplinary teams were first established after World War II to provide evaluation, prescription, delivery, and follow-up prosthetic services.
33.6 REHABILITATION ENGINEERING

33.4.1 Rehabilitation

The earlier the onset of rehabilitation, the greater is the potential for prosthetic success. The longer the delay, the more likely is the development of complications such as joint contractures, general debilitation, and depressed psychological state. The postoperative rehabilitation program includes preprosthetic and prosthetic phases. The preprosthetic or early postoperative residual limb management includes the time between surgery and fitting with a prosthesis. As such, it involves minimizing edema, enhanced healing of the residual limb, prevention of joint contractures and other secondary complications, maintaining or regaining strength in the affected lower extremity, initiating adjustment to the loss of a body part, regaining independence in mobility and self-care, and learning proper care of the intact extremity. The prosthetic phase begins with prosthetic prescription, proper prosthetic fitting and training and follow-up in a prosthetic or amputee clinic.

Preprosthetic Treatment. A postoperative dressing is often used to protect the incision and residual limb and control edema. It may involve an immediate postoperative fitting, a rigid or semirigid dressing, a controlled environment, or a soft dressing. The use of an immediate postoperative dressing greatly limits postoperative edema, thereby reducing postoperative pain and enhancing wound healing. It allows early bipedal ambulation with the attachment of a pylon and foot and allows early fitting of a definitive prosthesis by reducing the length of time to shrink the limb. However, it requires careful application and close supervision during early healing and does not allow daily wound inspection and dressing changes. Soft dressings such as an elastic wrap and/or an elastic shrinker (a socklike garment of knitted rubber-reinforced cotton) are relatively inexpensive, readily available, and washable. Elastic wraps require frequent rewrapping. Shrinkers may be used after the sutures have been removed and drainage has stopped. Semirigid dressings provide better control of edema than the elastic wrap and shrinker but may loosen with time.

The dimensions of the residual limb vary with time (Fig. 33.5). The decrease in stump size results from the reduction of edema, wasting of soft tissues from prosthetic stresses, disuse atrophy of the residual limb musculature, and decrease in fatty tissue with overall weight loss. Initial prosthetic fitting often begins 1 to 2 weeks after surgery, following early compression wrapping. The time

![Graph](image-url)
between amputation and definitive prosthetic fitting is at least 6 weeks because definitive fitting requires an approximately stable limb volume for 2 to 3 weeks.5

**Prosthetic Prescription.** The director of the prosthetic clinic (e.g., orthopedist or physiatrist) typically writes the prosthetic prescription, with input from the physical therapist and prosthetist team members. *Prosthetic prescription* refers to specification of the prosthetic socket design, commercial componentry (foot, knee units), suspension, and interface materials (insert, stump socks). Prosthetic selection is influenced by the age and general condition of the patient, his or her skin and vascular status, the presence or absence of disease, and any limitations imposed by such disease.

There are two primary types of prostheses: preparatory and definitive prostheses. The purpose of a *preparatory prosthesis* (Fig. 33.6) is to allow early ambulation at an efficient and safe level and yet allow for rapid changes in limb volume that often occur in the days and weeks postoperatively. Preparatory prostheses may also include temporary prostheses or permanent prostheses that lack the cosmetic cover. Preparatory prostheses are typically endoskeletal, facilitating the interchange of components as may be necessary before finalizing the limb design.

*Permanent prostheses* may be either endoskeletal or exoskeletal (external support, often used synonymously with crustacean or an outer-shell design). Most permanent endoskeletal prostheses are covered by a soft cosmetic outer covering. Exoskeletal prostheses are less versatile in terms of interchangeability of components and alignment variations but may be indicated for clients with a large build or excessive loading, prior experience with such designs, or use in environments in which the cosmetic cover (foam and stocking) of an endoskeletal prosthesis would be torn or stained.
Due to the limited success of direct skeletal attachment, prosthetic sockets form the interface between the residual limb and prosthesis. The socket fits over and encloses the residual limb tissues. The more intimate the fit, the less need there is for supplemental suspension, and the more control the amputee has over the prosthesis. However, a tightly fitting socket often makes greater demands on the skin, which must be tough enough to tolerate pressure and some slippage and healthy enough to withstand the confining environment. The intimately fitting socket does not merely mirror the residual limb. Rather, its contours are designed so as to provide a comfortable and functional connection between the residual limb and the prosthesis under dynamic loading.

33.5 PROSTHESIS DESIGN

Prosthetic design involves making a replacement of a missing body part of the appropriate shape and size. The prosthesis must be comfortable, functional, and cosmetic (appearance of the prosthetic device, including visual appearance, smell, sound). The prosthesis should take into account the client’s general health, weight, activity level, and motivation so as to set realistic goals. The patient’s residual limb length, shape, skin condition, circulation, range of motion, and maturation should also be taken into account. Since prostheses are expensive and many insurance companies provide limited reimbursement or a single artificial limb, cost may also be a factor.

The principal function of the residual limb is to serve as the lever to power and control the prosthesis. The prosthetic socket must support the patient’s body weight and hold the residual limb firmly and comfortably during all activities. As such, the prosthetic socket should be designed to support the residual limb tissues, facilitate control of the prosthesis during stance and swing, provide suspension during swing, and facilitate alignment of the artificial limb. Near-total contact between the distal limb and socket is required to aid proprioceptive feedback and prevent edema and skin problems.

Since the design of the prosthesis varies with amputation level, prosthetic design will be reviewed for the aforementioned lower extremity amputation levels. Since the primary levels of lower extremity amputation are transtibial (54 percent) and transfemoral (33 percent), the prostheses for these amputation levels will be presented in greater detail.

Partial Foot Amputation Protheses. The significance of partial foot amputation includes (1) the loss of the anterior lever arm of the foot, thereby affecting the base of support and stability, (2) the functional loss of ankle dorsiflexion, (3) the tendency for the ankle to be fixed in equinus or plantarflexed, (4) a bony anterior residual limb that has poor skin coverage and is therefore difficult to fit and prone to tissue breakdown, and (5) potentially poor cosmetic replacement for prostheses that extend above the shoe. For the more distal levels of partial foot amputations (e.g., transmetatarsal, Lisfranc), prosthetic replacement is minimal, involving the use of a toe filler and arch support. The toe/shoe fillers prevent anterior migration of the foot, provide resistance to creasing of the shoe vamp, and provide shoe-size symmetry. These fillers are typically fabricated from wool, cotton, Silastic foam, synthetic rubber, Plastazote, or Kemblo. More proximal partial foot amputations may have prosthetic replacements that include a slipper or boot, extending to an ankle-foot orthosis (AFO) for Chopart, Boyd, and Pirogoff amputations (Fig. 33.7).

Symes Amputation Protheses. A Symes prosthesis must compensate for the loss of foot and ankle motion and the loss of leg length (approximately 2 in), as well as provide adequate support during stance and suspension of the prosthesis during swing. The primary difficulty with Symes prostheses is the design of a socket that accommodates the relatively bulbous distal residual limb and yet remains durable and facilitates donning. Symes prostheses include the leather-socket Symes, the posterior-opening Symes, the medial-opening Symes, and the hidden-panel Symes (Fig. 33.8).
Transtibial (Below-Knee) Amputation Prostheses. By definition, the residual limb of a functional transtibial amputee includes the tibial tubercle into which the quadriceps tendon inserts so as to retain knee extension capability. The prosthesis for a transtibial amputee, in general, consists of a socket with an optional insert, adapter hardware to attach the socket to the shank, and an artificial foot. In addition, the prosthesis often includes some means of auxiliary suspension.

There are two primary transtibial prosthetic designs: the historic design (dating back to 1696), which incorporates a thigh corset, side joints, and an open-ended wood socket, and the patellar-tendon-bearing (PTB) design (Fig. 33.9). In the historic design, the thigh corset takes load off the residual limb, the side joints prevent knee hyperextension and provide medial-lateral stability, and the
open-ended socket may provide a cooler environment for the residual limb. However, this historic prosthetic design is bulky, heavy, and noncosmetic and may cause distal edema due to the lack of total contact. In addition, this design contributes to thigh atrophy and may result in limb pistoning within the socket. In contrast, the PTB design is nearly total contact. As such, the PTB design may increase proprioception and decrease edema.

The soft tissues of the lower extremity residual limb are not well suited for load bearing. The load-bearing potentials of the residual limb soft tissues are not uniform and vary between individuals. Some tissues, such as the patellar tendon that is relatively avascular, can accept higher loads for a longer period of time without tissue degradation. As such, the patellar tendon is well suited to handling compressive load. The tissues over both the medial and lateral flares of the tibial condyles and shaft are also well suited for load bearing. However, the tissues covering the anterior crest of the tibia cannot assume load without tissue breakdown. Similarly, the tissues over the distal ends of the tibia and fibula cannot tolerate compressive stresses. Finally, due to the presence of the peroneal nerve below the head of the fibula, very low stresses are tolerated in this region.

The PTB socket design, initially developed at the University of California at Berkeley in the late 1950s, accommodates the nonuniform load-bearing tolerances of the residual limb. The basic concept of the PTB socket is to distribute the load over areas of the residual limb in proportion to their ability to tolerate load. Therefore, the majority of the load is to be borne on the patellar tendon (hence the name), medial and lateral flares of the tibia, and the popliteal area. The PTB socket precompresses the residual limb tissues in these load areas so that forces are transmitted comfortably and movement of the socket relative to the skeleton is minimized. The socket is thus a replica of the residual limb,

![FIGURE 33.9 Transtibial prostheses: historic design (left) and patellar tendon bearing (right). (Adapted from Ref. 10, Fig. 4.4.)](image-url)
with appropriate shape modifications or rectifications such that the pressure-tolerant areas bear the majority of the load and the pressure-sensitive areas are largely relieved of load.

While the PTB socket design remains the default socket design today, several variants exist that enhance medial-lateral stability and/or socket suspension. The nominal suspension for a PTB prosthesis is the supracondylar cuff. (This design can be augmented by a waist belt and modified fork strap for improved suspension.) The PTB prosthesis is therefore indicated for individuals with good ligamentous structure or medial-lateral stability of the knee. Alternative PTB designs include suspension sleeves, the PTB-supracondylar (PTB-SC) socket, the PTB-suprapatellar (PTB/SP) socket, and the PTB-supracondylar-suprapatellar (PTB-SC/SP) socket (Fig. 33.10).

The suspension sleeve may be silicone, latex, neoprene, or elastic. Since this sleeve does not provide medial-lateral stability, it requires inherent knee stability. The sleeve provides excellent suspension and helps mask the prosthetic socket trimlines. However, the sleeve is warm and may induce excessive perspiration and contribute to dermatological problems. As such, suspension sleeves may not be indicated for vascular patients.

The medial and lateral walls of the PTB-SC design are extended proximally so as to enhance medial-lateral stability and provide suspension. This design is indicated for individuals with transtibial amputation who require increased medial-lateral stability and indicate dissatisfaction with

FIGURE 33.10 Alternative means of suspending transtibial prostheses: PTB, PTB-SC, PTB-SC/SP, joints and corset, PTB-SP, removable medial wedge, and suspension sleeve. (Adapted from Ref. 49, Fig. 5.11, and Ref. 4, Fig. 4.11).
the supracondylar strap. Since the anterior socket brimline is lowered, this socket does not provide a rigid hyperextension stop. A variation of the PTB-SC design incorporates a medial supracondylar wedge or detachable medial brim. These variations facilitate donning and suspension for individuals whose femoral dimensions do not otherwise permit donning and suspension via the supracondylar brim.

The PTB socket may also be modified to extend the anterior brimline so as to enclose the patella. This PTB-SP design does not provide inherent medial-lateral stability but provides a hyperextension stop. Finally, the PTB-SC/SP design extends the proximal socket brimlines anteriorly and medially-laterally, providing the hyperextension stop as for the PTB-SP design and the medial-lateral stability offered by the PTB-SC design.

An alternative means of suspension that has become increasingly common for transtibial amputees is suction. Suction has been used routinely for individuals with transfemoral amputation for some time. Its use in transtibial prostheses has increased through the use of a shuttle locking pin, such as incorporated in the Silicone Suction Socket (3S, Durr-Fillauer, Chattanooga, Tenn.) (Fig. 33.11).

While transtibial prostheses may be worn without an insert, such practice is less common. Silicone or foam (e.g., Pelite, Spenco, Plastazote) inserts are often used to cushion the residual limb and potentially absorb and redistribute some of the compressive and shear forces generated during ambulation. In addition, the transtibial amputee typically wears cotton, wool, or synthetic socks, 1 to 15 ply, to absorb or wick perspiration and accommodate residual limb volume fluctuations often observed throughout the day.

**Knee Disarticulation Prostheses.** The knee disarticulation amputation, while relatively rare in the United States (more common in Europe), is generally fitted as a long transfemoral amputation. Since this long residual limb has a bulbous end due to the retention of the femoral condyles, the socket is similar to that for a Symes amputee. The prosthesis may include a laced leather socket or anterior-opening socket with side joints and a posterior check strap.
Transfemoral Amputation Prostheses. The femur of functional transfemoral amputees must extend distal to the ischial public ramus, providing some lever arm for hip extension and flexion. The prosthesis for a transfemoral amputee includes a prosthetic foot, a shank that may be endo- or exoskeletal in design, a knee unit, a socket, and a means of suspension (Fig. 33.12).

General concepts of transfemoral socket design include (1) proper contouring to facilitate remnant muscle function, (2) stabilized force application so as to apply load to the skeletal structures as much as possible, (3) stretching the hip muscles for improved functionality (i.e., length-tension relationship for muscle), (4) maximized contact area so as to minimize soft tissue pressures, and (5) adequate contact between the distal limb and socket walls so as to prevent edema.

There are three common types of sockets for transfemoral amputees. Plug fit (wood) is the more historic design in which the conical socket interior mirrors that of the residual limb. This design was replaced by the quadrilateral socket, designed by Radcliffe and Foort at University of California at Berkeley in the early 1960s (Fig. 33.13). This socket design provides total contact but not end bearing. This socket design reportedly allows better muscle control and limb comfort by contouring the socket walls and adducting and flexing the femur within the socket. The adducted lateral wall provides support for the femoral shaft and allows the hip abductors to contract effectively during gait. The anterior wall pushes the residual limb onto the posterior ischial seat, whereas the posterior wall provides a major area for weight bearing on the ischial tuberosity and gluteus maximus. The socket also serves to support the femoral shaft so that the hip extensors can contract effectively. The quadrilateral socket design has since evolved to the ischial-containment socket design.

FIGURE 33.12 Prosthetic prescription options for individuals with transfemoral amputation. (Adapted from Ref. 4, Fig. 4.14.)
The ischial-containment socket [e.g., normal shape, normal alignment (NSNA), contoured adducted trochanter controlled-alignment method (CAT-CAM), skeletal CAT-CAM (SCAT-CAM)] contains the ischium within the socket and attempts to provide a bony lock to maintain femoral geometry within the socket. This socket design reportedly provides better mechanical support of the pelvis by restoring the normal pelvic/femoral angle. The ischial lock acts to prevent pelvic shift. In contrast to the quadrilateral socket, the ischial-containment socket is narrower in the medial-lateral dimension, wider in the anterior-posterior dimension, and of course, contains the ischium within the socket. However, this socket design has increased cost, and the fitting and fabrication procedures are not as well documented.

Transfemoral sockets are typically rigid. Flexible sockets fabricated using a malleable thermoplastic [low-density polyethylene and Surlyn (Thermovac)] have recently been incorporated within a rigid or semirigid frame\(^\text{16}\) (Fig. 33.14). The potential advantages of these flexible sockets include increased perceived comfort due to the flexible walls, improved proprioception, accommodation of minor volume changes, and enhanced suction suspension.

Suspension of the transfemoral socket and prosthesis is typically achieved through suction (expulsion valve...
or roll-on gel liner with shuttle lock), a total elastic suspension (TES) belt, a Silesian belt, or a hip joint/pelvic band/waist belt. In rare instances, suspenders may also be used. Suction is believed to reduce pistoning of the residual limb within the socket, improve proprioception, provide total contact fit, and result in a lighter prosthesis. However, suction requires that the residual limb maintain constant limb volume and shape. Suction sockets are also hot and, for expulsion systems, may be difficult to don. (As seen in Fig. 33.15, socket donning for an expulsion-valve system requires that a stockinet or nylon stocking be pulled over the residual limb. The residual limb is inserted into the socket. The soft tissues are drawn into the socket by pulling the stockinet out through the valve opening before closing the valve. No sock is worn between the skin and the socket.) The TES belt is a wide neoprene band lined with nylon tricot that can be used with suction or as a primary suspension mechanism. The Silesian belt is typically an auxiliary means of suspension, often augmenting suction. The Silesian belt provides rotational stability but no medial-lateral stability. In contrast, the hip joint/pelvic band/waist belt provides medial-lateral stability and is easy to don, although it is quite cumbersome.

**Hip Disarticulation Amputation Prostheses.** The Canadian hip disarticulation prosthesis, introduced in the United States in 1954, is still used almost universally today. It consists of a foot, a shank, a prosthetic knee, a “thigh,” a hip joint/bumper/control strap, and a socket (Fig. 33.16). The hip disarticulation socket is essentially a bucket, providing a seat for the ischial tuberosity, medial-lateral stability, suspension, and support for weight bearing.
Hemipelvectomy Prostheses. The prosthesis for an individual with a hemipelvectomy is similar to that of the hip disarticulation prosthesis. However, the residuum of an individual with a hemipelvectomy no longer contains the ischial tuberosity. The socket must therefore provide distal support solely through the soft tissues. To minimize soft tissue pressures, the socket may extend more proximally using the iliac crest, distal ribs, and gluteal muscles for support. Suspension may be augmented by shoulder straps. This prosthesis is typically endoskeletal so as to minimize the weight of the prosthesis.

Although the modified Canadian-type hip disarticulation prosthesis can be successfully fitted, many hemipelvectomy amputees prefer crutches for speed and agility in ambulation and use the prosthesis for cosmetic purposes on special occasions.3

33.5.1 Prosthetic Socket Fabrication

Lower extremity prostheses are custom devices. While the connective componentry, joints (knees, feet/ankles), and suspension are commercially available, the socket that encases the residual limb is custom-designed. The socket design may or may not involve a computer, as in CAD-CAM.

Many prosthetic facilities now use central fabrication to allow use of complex and expensive CAD-CAM technology without the need for each facility to purchase and maintain such equipment. With central fabrication, patient evaluation, casting, fitting, and delivery continue to be conducted in the prosthetist’s office. The fabrication of the socket itself is done off-site and minimizes noise, dust, odors, and potential zoning law difficulties. As such, there is less need to hire technicians and theoretically more efficient use of the practitioner’s time. Disadvantages of central fabrication include potential communication problems between the prosthetic and central fabrication facilities, shipping delays, and quality control.3,7

PTB Socket for Transtibial Amputees. Prosthetic socket fabrication for transtibial amputees includes some means of residual limb shape sensing (casting or digitizing), socket design (plaster positive or digital representation), and manufacture (vacuum forming or lamination). Specific methodologies for both hand-rectification and CAD-CAM techniques are detailed below.
Sockets for transtibial amputees were originally carved from wooden blocks. However, very few prosthetists use this medium today, and exceedingly few amputees request wooden sockets. Most transtibial sockets are fabricated using thermoplastics (e.g., polypropylene, copolymer) or laminated resins (e.g., acrylic, epoxy, polyester).

Hand-rectified PTB sockets for transtibial amputees involve donning a stockinet over the residual limb, identifying anatomic landmarks (patellar ligament, tibial flares, fibular head) with an ink pen, and casting the residual limb with plaster bandages. This cast is then removed and filled with plaster to create a positive model of the residual limb. The inked bony landmarks are thus transferred to this plaster model. The plaster model is then modified or rectified such that plaster is built up over areas of intended socket relief. A rasp is used to remove plaster over regions of preferential loading. The socket is then vacuum-formed or laminated over this modified plaster positive.

The plaster casting and rectification procedures in current prosthetic practice require skill and experience, as well as considerable time. In an attempt to eliminate factors relating to manual dexterity both in the casting and rectification processes, Murdoch developed the Dundee socket in which hydrostatic pressure is used during casting. Unlike the PTB socket design, the entire surface of the residual limb theoretically bears the same load with this design. A similar procedure was recently documented by Lee et al. for use in developing countries.

Successful hand-rectified socket designs cannot be easily duplicated. In addition, such designs do not facilitate documentation of residual limb shape over time. The introduction of CAD-CAM into prosthetics was motivated by the desire to improve the accuracy and quality of socket design, as well as reduce the amount of time for socket manufacture.

CAD-CAM for transtibial prosthetic sockets generally emulates the procedures used in conventional hand-rectified PTB socket fabrication. Starting with a replica of the residual limb, changes are made such that the majority of weight bearing will be borne on the pressure-tolerant areas of the residual limb. As such, the first step of CAD involves obtaining a digital representation of the residual limb, a process known as shape sensing. Computer software is then used to modify the digital socket, generally mimicking the rectification procedure of the plaster positive. The final stage, CAM, involves transfer of the proposed socket design to a computerized milling machine so that either a positive mold suitable for vacuum forming is produced or a socket is directly fabricated.

Both contact and noncontact methods of shape sensing of the residual limb have been developed. In general, contact methods are susceptible to errors due to soft tissue deformation. Current commercial shape-sensing technologies include contact techniques such as casting with subsequent cast digitization [Provel d1L ShapeMaker, Seattle Limb Systems (Model+Instrument Development Corporation), Poulsbo, Wash.; CMM and Compression Probe, Active Life Sciences, Troy, Mich.; CANFIT-PLUS, Vorum Research Corporation, Vancouver, British Columbia, Canada], and direct limb shape sensing (TracerCAD, Tracer Corporation, Miami, Fla.; Virtual Casting, BioSculptor Corporation, Coral Gables, Fla.). Noncontact methods typically involve optical or laser scanning techniques [ShapeMaker 3000, Seattle Limb Systems (Model+Instrument Development Corporation), Poulsbo, Wash.; Delta Systems II, Clynch Technologies, Inc., Calgary, Alberta, Canada; CAPOD Systems, CAPOD Systems, South Lyon, Mich.].

The design of the socket is implemented using CAD. In CAD, the sculpting tools of conventional socket rectification are replaced by a computer, graphics terminal, mouse, and on-screen cursor. Modifications typically include scaling, addition or removal of ply, patching or local rectification, and smoothing. Quantification of the rectifications allows direct comparison between various socket designs. In addition, the permanent record of the residual limb geometry provides the capability of monitoring residual limb shape over time (local atrophy or edema, volume changes), as well as the capability of producing duplicate sockets.

As stated previously, CAM of transtibial sockets typically involves use of a computer numerically controlled (CNC) milling machine to carve the equivalent of the positive plaster socket mold. Fabrication of the socket is again performed by vacuum forming or laminating the socket over this mold. A recent development in CAM of prosthetic sockets that obviates the plaster positive is direct fabrication of the prosthetic socket using rapid prototyping techniques (e.g., SQUIRT Shape, Northwestern University Prosthetics Research Laboratory and Rehabilitation Engineering Research...
Such methods have resulted in successful fitting but are limited to some extent by the materials available for deposition and stereolithography.

Regardless of the rectification and fabrication technique, the prosthetic socket is then trimmed proximally by hand to either allow complete or limited motion of the knee.

**Socket Fabrication for Transfemoral Amputees.** Fabrication of the transfemoral socket is similar to that for a transtibial socket. Information regarding the residual limb shape is obtained using casts. A Berkeley or Hosmer brim frame is used to shape the proximal socket and provide the ischial seat for quadrilateral sockets. If the patient is unable to stand during casting, CAD-CAM techniques may be used. For transfemoral amputee sockets, CAD involves residual limb shape sensing based on discrete measures (e.g., limb length, limb circumference at specific levels, select anterior-posterior and medial-lateral dimensions). The socket design is then based on scaling of an appropriate socket template. As before, the final socket is vacuum formed or laminated over a plaster positive.

### 33.5.2 Prosthetic Componentry

The transtibial prosthesis incorporates one of the aforementioned prosthetic socket designs and suspension methods, a spacer (shank), and an artificial foot. The transfemoral prosthesis is similar, with the inclusion of a prosthetic knee unit and an additional spacer for the thigh. The spacer is usually made of wood, plastic, or metal depending on whether the design is exo- or endoskeletal.

**Prosthetic Feet.** With the exception of partial foot amputees, the prosthesis for all lower extremity amputees require a prosthetic foot. The prescription criteria for these feet take into consideration the amputation level, residual limb length, subject activity level, cosmetic needs, and the weight of the individual. Prosthetic feet range from the SACH (solid ankle cushioned heel) foot, which is relatively simple and inexpensive, to dynamic-response or energy-storing feet that are more complicated and considerably more costly. Note that prosthetic feet are often foot and ankle complexes. As such, prosthetic feet may replace plantarflexion/dorsiflexion, pronation/supination, and inversion/eversion. Prosthetic feet are typically categorized in terms of the function(s) they provide or replace and whether or not they are articulated.

Nonarticulated feet are typically the simplest and least expensive. The foot and ankle are combined in a single component, and shock absorption and ankle motion are provided by the materials and structure of the foot. Since these feet are nonarticulated, they are quiet and typically require little maintenance. These feet are also cosmetic, lightweight, and provide good shock absorption and limited inversion/eversion on uneven terrain. Disadvantages of nonarticulated feet are the limited range of plantarflexion/dorsiflexion, difficulty with inclines due to heel compression, lack of adjustability for different heel heights, and little torque-absorption capability.

The SACH foot is the most common nonarticulating foot. As the name implies, this foot contains a rigid wooden keel with a compliant or flexible heel and forefoot (Fig. 33.17). The heel wedge of the SACH foot compresses to emulate plantarflexion; the forefoot flexes to emulate dorsiflexion. This foot is contraindicated for active amputees and amputees who require torque-absorption and/or inversion/eversion capabilities.

Single-axis feet/ankles are articulated to allow plantarflexion and dorsiflexion (see Fig. 33.17). The range of motion is maintained by bumpers or stops and typically ranges from 15 degrees of plantarflexion to 6 degrees of dorsiflexion. The amount of plantarflexion and dorsiflexion can be adjusted by changing the durometer of the bumpers. These feet do not permit inversion/eversion or transverse rotation. As such, ambulation on uneven terrain with these feet results in transmission of torques and shears to the residual limb. This type of articulated foot is typically heavier and less cosmetic than the SACH foot, and the moving parts may become noisy with use and may necessitate bumper replacement.

Multiaxis feet incorporate a “full” functional ankle (e.g., Greissinger, Otto Bock, Minneapolis, Minn.; College Park, College Park Industries, Inc., Fraser, Mich.; Genesis II, MICA Corp., Kelso,
Degrees of freedom include plantarflexion/dorsiflexion, inversion/eversion, and transverse rotation and may be invoked with mechanical joints and/or composite structures (see Fig. 33.17). While such feet may be bulky, heavy, noisy, and expensive and may require frequent repair, they accommodate uneven terrain and therefore reduce shear and torsional forces that might otherwise be transferred to the residual limb.

FIGURE 33.17 Prosthetic feet and ankle units: SACH, single-axis, Carbon Copy HP, STEN flexible keel foot, 1D25 dynamic plus foot, TruStep, Modular III, Re-Flex VSP, and Pathfinder. Also shown is a torsion adapter.
While rotators are not explicitly a foot or an ankle, they act to reduce the torque/shear forces on the residual limb due to ambulation over uneven terrain by allowing the socket to rotate independent of foot position. This component may be positioned anywhere in the prosthesis, not just at the ankle, where inertial effects may be problematic (see Fig. 33.17). Since this component adds mass and complexity to the prosthesis, its inclusion may necessitate improved or auxiliary suspension. Another component that is again not a foot or an ankle is shock absorbers. As for the rotators, shock absorbers, which decrease the effective length of the shank during loading, are typically positioned in the shank.

One of the most common areas of prosthetic development is recent years involves dynamic-response or energy-storing feet (e.g., Flex Foot, OSSUR/Flex Foot, Inc., Aliso Viejo, Calif.; STEN, Kingsley Manufacturing Co., Costa Mesa, Calif.; Carbon Copy II, Ohio Willow Wood, Mt. Sterling, Ohio) (see Fig. 33.17). This research and the exceptional performance of disabled athletes using such technology have motivated product development and research assessing the utility (energy-storing capabilities, optimality of energy release) of such technology. These feet store energy via heel/forefoot compression during stance, with partial energy release during heel-off and/or toe-off. Some of these feet are multiaxial, accommodating inversion/eversion and rotation. The advantage of these feet include potentially reduced energy expenditure and decreased mass. However, such feet are expensive and may not accommodate children and large adults.

**Prosthetic Knees.** During gait, the knee acts as a shock absorber and a support structure and shortens the limb during swing. The purpose of prosthetic knee units is to restore normal function (i.e., quadriceps and hamstring function) and the appearance of gait with minimal energy expenditure. Factors influencing prosthetic prescription include the client’s ability to voluntarily control the knee during stance based on their hip musculature and residual limb length and the inherent stability of the knee unit itself. Prosthetic knee units may be classified in terms of their control, such as no mechanical knee control (locked knee), stance phase control only, swing phase control only, and swing and stance phase control.

Stance-phase-control knees include (1) manual locking knees that prevent knee motion until the locking mechanism is manually disengaged, (2) knee units that are alignment controlled such that knee axis is positioned posterior to the weight line from heel strike to midstance such that an inherently stable knee extension moment is provided, (3) friction brakes that “lock” the knee on weight bearing, (4) polycentric linkages, and (5) fluid-resistive devices. A knee unit that is inherently stable requires little voluntary control of the hip musculature for function. Manual-locking knees provide the most inherent stability, followed by many polycentric knees (e.g., Stability Knee, DAW Industries, San Diego, Calif.), weight-activated friction knees, constant-friction knees, and lastly, outside hinges. The manual-locking knee, which can be unlocked for sitting, provides maximum stability during stance but results in an unnatural gait. Polycentric knees (typically

![FIGURE 33.18 Four-bar polycentric knee in a transfemoral prosthesis. (Adapted from Ref. 50, Fig. 36.)](image-url)
by definition, have a moving center of rotation that may result in an extremelystable knee. This design is relatively compact, thereby minimizing leg-length discrepancies for knee disarticulation and long transfemoral amputees. Weight-activated stance-control knees (e.g., Stabilized Knee, Ohio Willow Wood, Mt. Sterling, Ohio) function as a single-axis knee during swing and a braked knee during stance. Most weight-activated knees lock as axial load is applied following foot contact and remain locked until the weight has shifted to the sound limb after heel-off, just prior to toe-off. Such knees can lock at 20 to 30 degrees of flexion and may be used in conjunction with a knee-extension assist to initiate swing. Problems such as increased noise, frequent maintenance, and difficulty in jackknifing on stairs have been noted with these knee units. Conventional constant-friction knees are durable, easy to maintain and repair, and relatively inexpensive, but they provide little inherent knee stability. Thus these knees require good voluntary control of the hip musculature. Outside hinges may be used for individuals with knee disarticulation amputation because they do not add length to the femoral section of the prosthesis. They provide little inherent knee stability (but the very long residual limb of a knee disarticulation amputee typically has good hip musculature), with any stability provided via alignment.

Swing-phase-control knees may have constant resistance, variable resistance, or cadence-responsive resistance during swing. The resistance of a constant-resistance swing-phase-control knee, as the name implies, does not vary during swing, regardless of knee angle or cadence. The amount of resistance can be preset by the prosthetist. The resistance of variable-resistance swing-phase-control knees varies as a function of knee flexion angle. In contrast, the resistance of cadence-responsive resistance swing-phase-control knees varies (multiple preset options) as a function of knee angular velocity.

Knee-resistance mechanisms include sliding mechanical friction and fluid mechanisms. The sliding mechanical friction is contact friction, usually applied by a clamp around a knee bolt. It is relatively simple and inexpensive, but since the friction does not vary with cadence, an amputee wearing a prosthesis incorporating this type of knee unit is able to walk at only one cadence with optimal security and ease. The fluid mechanisms consist of a fluid-filled cylinder joined to the knee bolt by a piston, typically located in the posterior aspect of the shank. The resistance to knee flexion during swing is produced as the piston in the cylinder pushes against air or oil. The resistance to swing triggers a knee extension bias that then assists the prosthetic knee into extension. All stance phase control is achieved by alignment and muscle contraction of the hip extensors during stance.

The fluid in these cadence-responsive knee units may be oil (hydraulic) or air (pneumatic). For hydraulic knees, the fluid is incompressible. The resistance to piston motion results from fluid flow through one or more orifices. As such, the resistance is dependent on the fluid viscosity and density, the size and smoothness of the channel, and the speed of movement. In contrast, for pneumatic knees, the fluid is compressible. The resistance is again due to fluid flow through the orifice(s) but is also influenced by fluid compression. Since air is gas, potential leaks in pneumatic knee units will not result in soiled clothing, unlike what may occur with hydraulic knees. In addition, since air is less dense than oil, pneumatic units tend to be lighter than hydraulic units. However, since air is less dense and less viscous than oil, pneumatic units provide less cadence control than hydraulic units. Note that since viscosity is influenced by temperature, hydraulic (and pneumatic) knee units may perform differently inside and outside in cold weather climates. An example of a hydraulic cadence-responsive knee unit is the Black Max (USMC, Pasadena, Calif.). Additional examples include the Spectrum Ex (pneumatic, Hosmer, Campbell, Calif.), Pendulum (pneumatic, Ohio Willow Wood, Mt. Sterling, Ohio), and Total Knee (hydraulic, Model 2000, Century XXII Innovations, Jackson, Mich.), which combine a cadence-responsive resistance swing-phase-control knee with a four-bar polycentric stance control knee.

### 33.5.3 Prosthetic Fit

The prosthetist also evaluates prosthetic fit, or the comfort (pressure distribution), stability, suspension, alignment, and function of the prosthesis. The fitting of a prosthesis is an empirical process. The prosthetist has no quantitative information regarding the load distribution of the soft tissues and must
rely on experience, feedback from the amputee, and indirect indications of tissue loading to assess socket/prosthesis fit. To assist this process, the prosthetist often fits a test or check socket. This test socket is transparent and is typically fabricated by vacuum forming a polycarbonate sheet over the plaster positive model. After the amputee dons the test socket, the prosthetist provides distal resistance, or the amputee stands in an alignment rig. Skin blanching during loading and subsequent redness (reactive hyperemia) on removal of the prosthesis are used to identify areas of excessive pressure. If a prosthetic sock is worn, areas of excessive pressure are indicated by the compression of the prosthetic sock or sock weave impressions on the residual limb. Other means of estimating tissue loading include the use of chalk, talcum powder, clay, or lipstick to assess distal end bearing and the presence of calluses on the residual limb. These methods provide some basis on which to qualitatively assess comfort and the weight-bearing pressure distribution.

Fitting also includes assessment of prosthetic suspension, the adequacy of the proximal trim lines in terms of comfort and the associated mobility of the proximal joint, and the alignment of the prosthesis itself (see Sec. 33.5.4).

As alluded to earlier, knowledge of the interface stress distribution between the residual limb and the prosthetic socket enables objective evaluation of prosthetic fit. It is this desire for quantitative description of the prosthetic interface stress distribution that has motivated many experimental and numerical investigations of prosthetic interface stress.

Several groups have investigated the stress distribution between the residual limb and prosthetic socket for both transtibial and transfemoral amputees in laboratory and clinical settings. Investigation of the effects of prosthetic alignment, relative weightbearing, muscle contraction, socket liners, and suspension mechanisms on the interface pressure distribution have been conducted (for review, see ref. 38). These experimental stress measures have been limited to specific sites on the limb, since measurements can only be obtained at transducer locations. Comparison of the results of these investigations is difficult because the interface pressure measures are highly dependent on both the means of measurement (transducer type) and the transducer calibration method employed. Ferguson-Pell39,40 has reviewed several key factors in transducer use and selection relevant to stress measures between human soft tissues and a support structure.

Many of the interface pressure-measurement techniques involve research applications and methodologies that are not appropriate for clinical use. However, the relatively recent development of commercial systems using force-sensitive resistors and capacitive sensors to measure interface loading for seating systems and prosthetic sockets provides a diagnostic tool that can be incorporated into prosthetic fitting. These systems have clinical potential and may facilitate creation of prosthetic databases such that interface pressures, whether measured in research settings or clinical settings or estimated with computer models, may be properly interpreted.

Polliack et al.41 recently compared the accuracy, hysteresis, drift, and effect of curvature on sensor performance for these commercial prosthetic systems (force sensitive resistors: F-Socket Measurement System, Tekscan, Inc., Boston, Mass.; and Socket Fitting System, Rincoe, Golden, Colo; prototype capacitive sensor, Novel Electronic, Minneapolis, Minn.). These authors concluded that the current systems were more appropriate for static use because the hysteresis, drift, and large standard deviations become more problematic during dynamic and long-term loading.

In contrast to the experimental techniques, computer models of the residual limb and prosthetic socket have the potential to estimate interface pressures for the entire residuum and indeed are not limited to the interface but can also provide information regarding the subcutaneous tissue stresses. Nola and Vistnes42 and Daniel et al.43 have found that initial pathological changes in pressure sore formation occur in the muscle directly overlying the bone and then spread outward toward the skin. Therefore, the subcutaneous stresses may be important when evaluating long-term prosthetic success. These subcutaneous stresses are particularly difficult to measure in vivo.

Several groups have used computer models of the residual limb to investigate the residual limb-prosthetic socket interface.44,45 Many investigators have also used finite-element modeling of the residual limb and the prosthetic socket of lower extremity amputees to investigate residual limb-prosthetic socket biomechanics and to estimate the interface stress distribution (for review, see refs. 38, 46, and 47).
Two primary limitations of these modeling efforts involve the representation of tissue properties across the entire limb and the interface condition between the residual limb and prosthesis. The ability of current finite-element models to estimate prosthetic interface stresses, while performing reasonably well in some cases, has not been highly accurate. Nevertheless, the methodology has potential. Advances in finite-element software enabling nonlinear elastomeric formulations of bulk soft tissue, contact analysis, and dynamic analysis may help address some of the current model limitations. Corresponding advances in pressure-transducer technology will help validate the computer models and facilitate interpretation of the analyses.

Finally, finite-element models have potential applicability in CAD of prosthetic sockets. Current prosthetic CAD systems emulate the hand-rectification process, whereby the socket geometry is manipulated to control the force distribution on the residual limb. Incorporation of the finite-element technique into future CAD would enable prescription of the desired interface stress distribution (i.e., based on tissue tolerance). The CAD would then compute the shape of the new socket that would theoretically yield this optimal load distribution. In this manner, prosthetic design would be directly based on the residual limb-prosthetic socket interface stresses.

33.5.4 Prosthetic Alignment

Prosthetic alignment refers to the orientation of the various components of a lower extremity prosthesis such that the user has optimal physical security, the best possible gait, minimum energy expenditure during walking, and a generally comfortable leg, resulting in good posture without injury to the residual limb even when used for comparatively long periods of time. Good alignment begins with proper fit of the residual limb in the socket (see Sec. 33.5.3). The adapter hardware used to connect the socket to the endoskeletal components, and to the foot and knee, facilitates alignment changes.

Alignment of the prosthesis consists of static and dynamic stages. For transtibial amputees, static alignment places the knee in approximately 5 degrees of flexion to prevent hyperextension and increase pressures on the anterior surface of the limb. The pylon is vertical, and alignment in the sagittal plane is such that a plumb line at the center of the greater trochanter intersects the mediolateral axis of the knee and aligns with the anterior surface of the pylon so that there is no tendency for the knee to buckle during stance. Additional modifications may be invoked during dynamic alignment based on observation of amputee gait in the frontal and sagittal planes. (Sanders, Chap. 20, reviews static and dynamic alignment of the transtibial prosthesis and identifies malalignment symptoms.)

Zahedi and Spence found that a transtibial amputee can adapt to several alignments, ranging from as much as 148-mm shifts and 17-degree tilts. This tolerance of alignment variability was attributed to the degree of control that an amputee has over the prosthesis. For a transtibial amputee, for example, the retention of the knee joint allows the body to compensate more readily to malalignments. In addition, Zahedi and Spence hypothesized that the nonuniform distribution of alignment data about the mean was indicative of an optimum alignment configuration not readily achieved by current alignment procedures.

For transfemoral amputees, the socket is mounted on an alignment rig. The desired height of the prosthesis is determined, as is the relative position of the knee and foot with respect to the socket/limb. This bench alignment is such that the center of the heel falls just under the ischial seat, with the trochanter-knee-ankle (TKA) line passing through the knee and ankle axes of rotation. As for alignment of transtibial prostheses, this bench alignment is modified based on static and dynamic observations of amputee gait in the frontal and sagittal planes.

In general, prosthetic socket fit and alignment are dependent. Prosthetic alignment depends on the length of the residual limb (lever arm), strength of the remnant musculature, and the amputee’s balance and control. For the hip disarticulation prosthesis, alignment defines prosthetic stability because there are no knee extensors to prevent the knee from buckling during stance. For stability, the weight line during stance must go through the base of support, be posterior to the hip so as to provide an inherently stable hip extension moment, and be anterior to the knee so as to provide a
knee extension moment. As such, the prosthetic hip joint is located distal and anterior to the axis of
the anatomic hip (see Fig. 33.16). (The position of the hip joint also determines the length of the
thigh when sitting.) The posterior bumper, which controls the amount of hip extension, is mounted
well forward of the weight line so as to produce a hip extension moment during stance. The stride-
length control strap is positioned posterior to the hip and anterior to the knee. This strap is a critical
feature of the hip disarticulation prosthesis, acting as the quadriceps muscle—preventing excess hip
extension and excess knee flexion while assisting in knee extension.

33.6 AMPUTEE GAIT

Since one of the goals of lower extremity prosthetics is to replace function, much attention is given
to the restoration of gait or ambulation. While some argue that optimal gait for a lower extremity
amputee need not be symmetric, symmetry in ambulation is cosmetic. As such, the prosthetist and
physical therapists attempt to restore normal, symmetric gait—given constraints such as joint
contractures, weak hip/knee musculature, poor balance, and the potential need for an assistive device.

Another primary factor that influences ambulation and prosthetic use is tissue pain. The residual
limb soft tissues are routinely loaded during ambulation. As such, these tissues are stressed, unlike
prior to amputation. These stresses are the direct result of load transfer from the prosthetic foot
through the residual limb soft tissues and subsequently through the skeleton. Thus, while symmetric
gait may be desired, tissue sensitivity and pain due to the presence of neuromas and local stress
concentrations may mandate altered gait.

The residual limb-prosthetic socket stresses are influenced by the fit of the socket and the
alignment of the prosthesis. For transtibial amputees, medial-lateral stability is influenced by foot
placement. Foot inset (or outset) may result in varus (or valgus) moments applied to the limb.
Similarly, anterior-posterior stability is influenced by the fore-aft position (extension-flexion
moment) of the foot, foot plantarflexion/dorsiflexion (extension-flexion moment), and heel
durometer (soft heel increases foot plantarflexion). For transtibial amputees with normal knee
extensors, knee flexion moments on heel strike are desired, as for individuals without amputation.

Since most prosthetists do not have motion-analysis systems, they cannot quantitatively evaluate
lower limb kinematics nor measure lower extremity kinetics. Therefore, prosthetists rely on visual
assessment of limb kinematics and indirect measures of tissue loading and joint moments. Joint
moments are inferred from the relative position of a weight line with respect to the ankle, knee, and
hip axes of rotation. Muscle activity that may be required to oppose such joint moments for stability
is similarly inferred. While such moment estimation ignores inertial factors, it provides a visual
estimate for evaluation of prosthetic fit and alignment. These visual estimates are only be applied
during stance, since stance phase dictates stability and tissue loading. Analysis of swing is restricted to
kinematic analysis in the frontal and sagittal planes.

Gait analysis of the transtibial amputee is reviewed in Refs. 7 (Chaps. 14 and 18) and 1. These
references highlight common gait deviations, their potential prosthetic and/or physiological causes,
and possible corrective measures.

For transfemoral and knee disarticulation amputees, the remnant hip musculature is vital to normal
gait and prosthetic stability. As noted previously in the discussion of transfemoral socket design, the
thigh is adducted in the socket so as to position the hip abductors at an appropriate functional length.
The hip abductors play an important role in stabilizing the pelvis in the frontal plane during single-
limb support. Since individuals with a short transfemoral residual limb may have weak hip abductors,
their pelvis may dip excessively to the swing side during single-limb support. The hip adductors are
important during weight transfer. The hamstrings (hip extensors) act to decelerate the limb during
swing in anticipation of heel strike. If the hip extensors are weak, lumbar lordosis may be observed
as a common compensatory mechanism. To maximize the efficiency of the hip extensors, the residual
femur is routinely flexed within the socket. The amount of hip flexion is dictated by the length of the
residual limb, the lever arm for hip extension.
Gait analysis of the transfemoral amputee is reviewed in Ref. 7 (Chap. 14). This reference highlights common gait deviations, their potential prosthetic and/or physiological causes, and possible corrective measures. Note that many individuals with transfemoral amputation choose not to wear a prosthesis and walk with crutches. The use of these assistive devices may facilitate faster and more efficient (per distance traveled) ambulation.

Gait analysis of the hip disarticulation amputee will not be discussed due to the relative rarity of this level of amputation and the fact that the excessive cost of ambulation often dictates that such amputees do not routinely use a prosthesis for ambulation. Many individuals with hip disarticulation prefer to use forearm crutches for ambulation and don their prosthesis for cosmetic purposes only. However, as discussed previously in the design of the prosthesis for individuals with this level of amputation, the stride-length control strap and hip bumper are critical to prosthetic limb stability and function during ambulation.

As alluded to earlier, energy consumption during ambulation is higher for amputees than for nonamputees. While energy cost (energy consumption per distance traveled) is comparable for unilateral traumatic transtibial amputees and normal individuals, energy cost increases substantially as the level of amputation becomes more proximal (Fig. 33.19). These energy costs tend to be higher for vascular amputees than for individuals who have lost their limbs due to trauma. As one might expect, the energy cost of ambulation for individuals with bilateral lower extremity amputation is higher than for unilateral amputees. Finally, the natural velocity of individuals with amputation is typically less than for normal individuals, with the velocity decreasing for the more proximal levels of amputation (see Fig. 33.19). Assessment of amputee performance is further confounded by age, since the majority of lower extremity amputations are due to vascular complications, prevalent in older individuals.

33.7 RECENT DEVELOPMENTS

As indicated throughout this chapter, many advances in lower extremity prosthetics are due to technological advances in materials. This has facilitated fabrication of markedly lighter prostheses, stronger prosthetics sockets, and composite feet with energy-storing capabilities. Prosthetic liners for transtibial amputees have also benefited from such technology. The OrthoGel liners (Otto Bock, Minneapolis, Minn.) incorporate polyurethane gel and provide enhanced comfort for individuals whose residual limb tissues are very sensitive and/or prone to breakdown.

In terms of recent innovations, it is worth acknowledging advances in lower extremity prosthetics with respect to suspension. Historic suspension mechanisms such as hip and waist belts and fork straps and supracondylar cuffs for transfemoral and transtibial amputees, respectively, have been largely replaced by suction. Suction has been induced using neoprene suspension sleeves, TES belts, and shuttle-locking mechanisms.

Sabolich Research and Development (Oklahoma City, Okla.) recently has developed commercial systems for the upper and lower extremities that attempt to supplement sensation lost by amputation. Their Sense of Feel leg provides sensory feedback to the residual limb. Output from force transducers in the sole of the foot is transmitted to the residual limb via electrodes in the prosthetic socket. The amplitude of the “tingling” sensation is proportional to the force measured at the foot, thereby providing direct feedback to the amputee regarding the relative loading of the forefoot versus the heel. Such information is hypothesized to improve lower extremity balance. Clinical trials are currently underway.

Another new commercial product is the 3C100 C-Leg (Otto Bock, Minneapolis, Minn.) (Fig. 33.20). This transfemoral leg incorporates a microprocessor-controlled hydraulic knee with swing and stance phase control. The controls are adjusted for the individual subject. The knee angles and moments are measured at 50 Hz, and the prosthesis facilitates ambulation at various speeds on inclines, declines, stairs (step over step), and uneven terrain. The rechargeable lithium-ion battery provides sufficient power for a full day (25–30 hours). Similar microprocessor-controlled pneumatic
**FIGURE 33.19** Rate of oxygen consumption and walking speed for surgical (gray), traumatic (black), and vascular (white) amputees for various levels of amputation (HP = hemipelvectomy, HD = hip disarticulation; TF = transfemoral; KD = knee disarticulation; TT = transtibial). (From Ref. 6.)
knees are also available from Endolite (Intelligent Prosthesis Plus, Centerville, Ohio) and Seattle Limb Systems (Power Knee, Poulsbo, Wash.)

One final noteworthy lower extremity development in that of shock-absorbing pylons and shock-absorbing feet (Re-Flex VSP, OSSUR/Flex-Foot, Inc., Aliso Viejo, Calif.) (see Fig. 33.17). These pylons and feet alter the effective length of the shank during loading and introduce new fitting issues and potential problems. The shock absorption or vertical compliance serves to spare the residual limb tissues during high-impact activities.

Some of the latest technology with respect to artificial limbs was recently highlighted in a medical device journal. Many of these developments involve the use of robotics and/or smart materials in lower (and upper) extremity prostheses. A transfemoral prosthesis is under development at Biedermann Motech (Schweinzen, Germany) that incorporates sensors in the prosthetic knee to measure the force and moment exerted on the prosthesis, as well as the angular orientation of the knee. The knee of this prosthesis incorporates a magnetorheological fluid in its damper that reportedly results in improved response times over conventional hydraulic knee units. The BioRobotics Laboratory at the University of Washington in Seattle is investigating use of the McKibben artificial muscle (pneumatically operated actuators) to power a lower extremity prosthesis. After development of the functional prototype of the powered prosthesis, these researchers will compare their active design with conventional limbs in terms of function (gait) and energy expenditure.

Prosthetics research is also being conducted in a collaborative effort by researchers at Sandia National Laboratories (Albuquerque, N.M.), the Seattle Orthopedic Group (Poulsbo, Wash.) and the Russian nuclear weapons laboratory at Chelyabinsk-70. A recent project involves the design of a lower extremity prosthetic limb that can adjust to an amputee’s gait/environment (incline, decline, and irregular terrain) and compensate for changes in residual limb shape. The first initiative involves the use of microprocessor controls for the hydraulic joints and piezoelectric motors governing the ankle and knee mechanisms. The latter initiative is based on sensors in the socket that monitor the diameter of the residual limb over the course of the day. Difficulties reportedly involve the development of a robust, lightweight power source for the entire system.

As prostheses improve, amputee function improves. Such improved function is often accompanied by increased desire to participate in additional activities and/or further improvements in performance, especially with respect to athletics and recreation. As such, there have been continued developments regarding task-specific prostheses and adaptive equipment. Amputee athletes’ prowess with respect to running has received considerable press. These amputees post amazing times, supported by energy-storage prosthetic feet—with dorsiflexed alignment specific for running. Many lower extremity amputees also ski. Individuals with transfemoral amputations also ski. Individuals with transfemoral amputations often ski without a prosthesis and may use outriggers (adapted ski poles that are a cross between a crutch and a miniski). Transtibial amputees, on the other hand, typically ski with a prosthesis that has been modified (e.g., alignment that accommodates skiing; step-in, rear-entry boots). Individuals with lower extremity amputations may also swim and again may opt to use or not use a prosthesis. Modifications for swimming include a prosthetic socket with a distal fin, peg legs for the beach, and/or a swim leg that does not float but provides some functionality. As for snow skiing, individuals with lower extremity amputation may also waterski, typically using a single ski; they may or may not use a prosthesis.

![FIGURE 33.20 C-leg system with microprocessor-controlled hydraulic knee with stance and swing phase control.](image-url)
Finally, amputee golfing has gained in popularity. For transfemoral amputees, shear on the residual limb is minimized by the inclusion of a rotator.

Finally, future developments in CAD of prosthetic sockets are also likely to be influenced by alternative shape-sensing methodology and finite-element model development that will enable timely evaluation of residual limb geometry and/or material properties.

REFERENCES


