Let Them Walk! Current Prosthesis Options for Leg and Foot Amputees

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Amputation has been long used to alleviate both congenital and acquired ailments of the extremities. The Indian poem "Rig-Vieda" is the first written record of an amputation (between 3500 and 1800 BC); it describes a warrior queen who lost her leg in battle and was fitted with an iron prosthesis.[1] The majority of amputations in early history were performed for war injuries, legal punishment, and religious sacrifices.[2]

Early prostheses were rudimentary devices such as the peg leg or hooks.[3] Advances in prosthetic design occurred during the Renaissance period, the American Civil War, and the two world wars, with an emphasis on recovering limb function. The predecessor of the SACH (solid-ankle, cushioned heel) foot was initially devised by JE Hanger, a Southern soldier during the American Civil War. Interest and awareness of the importance of prostheses during World War I prompted the organization of the American Prosthetics and Orthotics Association. The Artificial Limb Program was developed with sponsorship from the Veterans Administration, the Department of Health, Education, and Welfare, and the armed forces. Public support for these research institutions led to the rapid development of modern prosthetic technologies using innovative designs and materials.

The purpose of this article is to review current information regarding indications and techniques of lower extremity amputations, with an emphasis on current prosthetic designs to enable a functional limb.

Hip disarticulation

Hip disarticulation is a radical lower extremity amputation and has been performed with minimal alteration in technique since it was first performed by Kirk in 1943. Boyd[4] and Slocum subsequently recommended a more anatomic dissection designed to minimize blood loss and to aid prosthetic use.[5] Sugarbaker and Chretien[6] emphasized optimal suture line placement, with muscular and soft tissue covering the acetabulum, to improve prosthetic fitting and minimize complications.

Mortality of the procedure corresponds to the disease etiology and associated comorbidities. One study noted a 44% combined mortality rate for hip disarticulation.[7] If no limb infection was present preoperatively, hip disarticulation for arteriosclerosis carried a 20% mortality rate. But if preoperative limb infection was present, the mortality rate increased greatly, to 60%.[7] Preoperative limb infection in the setting of trauma with associated ischemia resulted in a 100% mortality rate.[7] This group of patients also had a very high (60%) postoperative infection rate. To minimize wound infections, recent studies have recommended aggressive wound care and even transfemoral guillotine amputation before definitive hip disarticulation.[7]

Higher incidences of wound complications and mortality for hip disarticulation have been found in operations performed for ischemic indications, and have been attributed to the occurrence of nonviable skin flaps.[8,9] A high rate of wound infection was also found in individuals with earlier ipsilateral above-knee amputation (AKA). But this may be related to a delay in obtaining a definitive hip disarticulation procedure. So, early operation for revascularization or AKA for extremity ischemia, or both, have been recommended to minimize the need for hip disarticulation. If revascularization in a compromised AKA cannot be performed, initial hip disarticulation may be indicated.[9]

Hip disarticulation in patients with cardiovascular disease has a 37% early postoperative mortality rate. This may be from the poor preoperative state of the patients or the time consuming and traumatizing nature of the operation.[10] There is a higher mortality rate in patients undergoing multiple operations when compared with primary operations.[11] The complicated character of the surgical problems seen in these patients needing hip disarticulation can be illustrated by the fact that an average of 2.9 salvage operations and 2.3 major operations per patient are performed on this population.[10]

Prosthetics

There are a number of prosthesis fitting protocols after hip disarticulation. One common approach is to use a tempo-

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Abbreviations and Acronyms

AKA	= Above-knee amputation
BKA	= Below-knee amputation
ICEROSS	= Icelandic Roll-On Silicone Socket
IPOP	= Immediate postoperative prosthetic placement
SACH	= Solid ankle cushion heel
TSB	= Total surface bearing

rary prosthesis in the first 6 months after discharge. Although most patients who undergo hip disarticulations for tumors are discharged with a prosthetic ambulatory status, only 0% to 10% of vascular patients are able to achieve prosthetic ambulation. These patients were mobile only with walkers or chairs; others are essentially bed bound.[8]

Because hip disarticulation amputees do not have a functional hip, they rely on increasing and decreasing spine lordosis for propulsion during prosthetic ambulation. The weight of the prosthesis exerts an ambulatory load that restricts movement and often results in easy fatigue.[12] Increased energy expenditure in hip prosthesis mobilizers are evidenced by an 82% increase in energy use and a 40% slower ambulation speed than in able-bodied individuals[13] (Table 1). Patients using the hip disarticulation prosthesis are able to achieve 61% of comfortable walking speed and 70% of fast walking speeds of able-bodied persons.[14,15]

The Canadian hip disarticulation prosthesis (Fig. 1), introduced by McLaurimin in 1957, was the most commonly used prosthesis. This traditional design has been modified through the years into modern prosthetic systems such as the Otto Bock Modular Endoskeletal hip prosthesis (Table 2), which has developed into the prosthesis of choice in hip disarticulation. Advantages of the newer design include the use of lighter weight materials such as titanium and carbon fiber composites, improved cosmesis, and flexibility of use with interchangeable and adjustable components. The widely used socket consists of a basket-shaped socket enclosing almost the entire hemipelvis with a Silesian belt for suspension. Recently, a total suction socket and custom made pelvic belt has been introduced.[14] The suction hip prosthesis socket has been reported to enhance proprioception and decrease piston-like action in the prosthesis.[14,15] It should be noted that the hip disarticulation prosthesis is aligned with the load axis passing anterior to the knee joint during the stance phase. This helps the knee joint to remain in extended position during load bearing and stress.[16]

Patients using the Canadian hip disarticulation prosthesis exert, on average, 2.3 times more energy than their average resting metabolic rate and have a comfortable ambulation speed of 13 yards per minute. This is approximately 1.7 times more energy expenditure than patients with AKA.[17] Other developments include the introduction of residual limb lengthening using the modular prosthesis. This involves implantation of a modular proximal femoral replacement prosthesis with a special rounded end piece and a large weight bearing surface (Mutars, Implantcast Corp). The device is made of titanium-aluminum-vanadium to minimize weight. The joint capsule is reconstructed from a trivira tube. Refixation of the ileopsoas and gluteal muscles is necessary for good functional outcomes. The prosthesis is then covered with a well-perfused flap generally consisting of adductor muscles, rectus femoris, and biceps femoris muscles, which are supplied by the femoral vessels and femoral and obturator nerves. Preliminary studies have shown good functional outcomes in patients ambulating with and without crutches. Other strategies to preserve residual limb length with hip disarticulation include endoprosthesis implantation into the distal residual limb and selective resection of the proximal femur and hip joint, with preservation of the distal femur.[18] Recent studies have suggested that the intelligent knee prosthesis for can be used in conjunction with the hip disarticulation prosthesis to decrease energy expenditure in place of the constant friction knee joint with stance control.[12]

Table 1. Energy Expenditure and Ambulation Rate at Various Amputation Levels

Amputation level	Energy expenditure above normal, %	Ambulation rate, %
Syme amputation	43	
Below-knee amputation		
Long stump	10	
Short stump	40	80
Knee disarticulation	71.5	31 (prosthesis fitting rate)
Above-knee amputation	63	38–50
Hip disarticulation	82	0–10 (vascular patients)

Above-knee amputation

Above-knee amputation was the most commonly performed lower extremity amputation for vascular disease before the 1960s. This procedure has the advantage of achieving a healing rate that approaches 100%. The relative energy cost of ambulation in AKA amputees is 63%, compared with 40% in able-bodied individuals, 43% in patients with a Syme amputation, and 42% in patients with below-knee amputation (BKA; Table 1). The maximum aerobic capacity and walking velocity are higher for patients with a BKA than for those with an AKA, with younger individuals doing better than elderly patients.[19]

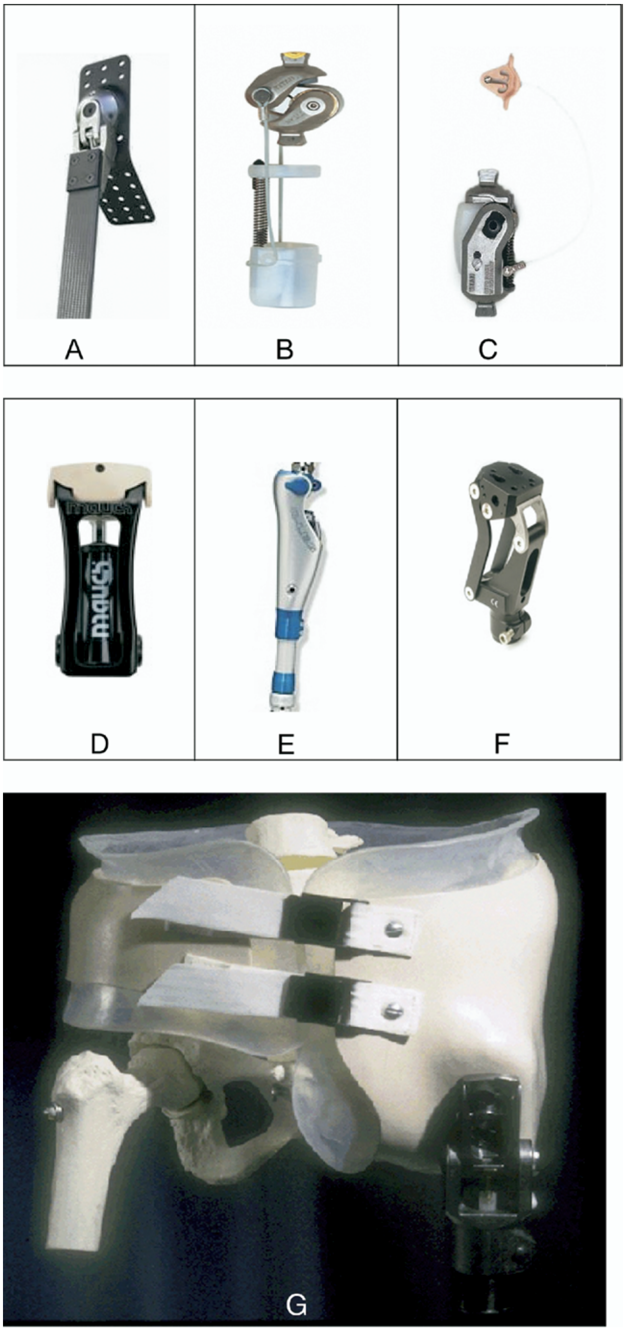

Figure 1. (A) Otto Bock modular endoskeletal hip prosthesis, \$5,350-\$7,350. (B) Single axis friction control knee prothesis, \$500-\$740. (C) Fluid swing control locked knee mechanism, \$650-\$980. (D) Fluid swing control open knee mechanism, \$850-\$1,300. (E) "Intelligent" transfemoral microprocessor-controlled prosthesis, \$5,000-\$7,300. (F) Four bar linkage polycentric knee joint, \$25,500-\$36,700. (G) Hip disarticulation socket with hip joint.

Among all amputees, only those with an AKA can ambulate faster using crutches compared with prosthesis alone. Crutch-only ambulation in this patient population produce respiratory quotients (0.97 versus 0.96) and heart rates (130 versus 126 beats per minute) comparable to those with prosthesis usage.[19]

Knee prosthesis design varies from the most rudimentary single axis knee to the state-of-the-art high tech "intelligent" prostheses such as the "C-leg" (Table 2). The appropriateness of each prosthetic option is determined by patient characteristics such as age, preamputation functional status, comorbidities, and rehabilitation goals. Careful consideration is necessary to ensure that reasonable patient expectations are met. From the simple articulated knee, we have progressed to the fluid control knee prosthesis, which allows swing control with either pneumatic or hydraulic systems. Although knee disarticulation has been recommended as a generally superior operation, acceptance has been slow. This has been at least partly because of the difficulty of designing a prosthesis that would allow both the native and prosthetic knees to be at the same level, and issues with healing. Currently, novel technologies such as the polycentric knee with an adjustable center of rotation have made knee disarticulation a much more attractive option than it was previously. The "intelligent" prosthesis has allowed amputees with good ambulatory function to maximize their mobile potential by allowing activities at a higher speed and achieving a more complex level of function. In addition, software algorithms have been used to optimize gait dynamics and performance through data collected from sensors in the prosthesis.

Basic knee prosthetic designs

The most common type of knee is the single axis knee, in which flexion and extension occur around a single axis. Advantages are its reliability, simplicity, low maintenance, and low cost. This is a simple prosthetic knee that is rarely used except in poorer developing countries. One of the most important design advancements for the prosthesis is control of the swing phase. The initial prototype was a friction based system in which an adjustable friction cell was pressed against the knee axle, resulting in a degree of swing control. An elastic or spring loaded mechanism is also frequently provided to promote full leg extension before heel strike (Table 2). Unfortunately, the friction swing control mechanism functions well in only one cadence and is difficult to use on irregular surfaces. Despite the simplicity of design and limitations, the single axis knee is a surprisingly reliable and inexpensive design ideally suited to individuals with restricted access to regular health care and prosthetic maintenance.[20]

Table 2. Overview of Lower Extremity Prostheses

Model	Advantages	Disadvantages	Indications
Otto Bock Modular Endoskeletal hip prosthesis	Light weight Improved cosinesis Flexibility of use with interchangeable components	High energy expenditure Slow ambulation	Able bodied patients post hip disarticulation
Single axis friction control knee prosthesis	Inexpensive Very reliable	One cadence Difficult to use on irregular surfaces	Restricted access to regular health care
Fluid swing control locked knee mechanism	More stable knee mechanism Lower energy expenditure Lower heart rate	Lower ambulation speeds	Weaker, less steady amputees such as elderly or stroke patients
Fluid swing control open knee mechanism (Hans Mauch S-N-S cylinder)	Faster ambulation Wide range of walking speeds	Higher energy expenditure Higher heart rate	Patients in good physical condition with higher exercise demands
"Intelligent" transfemoral microprocessor-controlled prosthesis (C-Leg)	Able to adjust swing speed according to "perceived" ambulation speed Higher ambulation speeds	Expensive Patients with average physique derive little additional benefit	Patients in excellent physical condition desiring a high exercise capacity
Four bar linkage polycentric knee joint	Prosthetic knee bends on sitting at same level as contra lateral native knee Stable center of rotation Anterior weight bearing axis	Higher cost	Ambulatory patients post knee disarticulation

Fluid phase swing control mechanisms

More advanced fluid swing-phase control mechanisms were developed to overcome some of the limitations posed by the friction-based swing-phase control. Fluid phase control prostheses with either pneumatic or hydraulic mechanisms are more able to function over a range of ambulation cadences and speeds. Pneumatic units are based on compressible air, and are recommended for the slow or moderate ambulator because vigorous walkers may "out-walk" this design. The advantage of this air-based mechanism, however, is that it is much less sensitive to changes in ambient temperature.[20] The pneumatic swing control provides nonlinear resistance for shin swing control. This results from a progressive increase in air pressure with compression and transfer of air from one side of the piston to the other.[21]

Compared with the friction-control mechanism, a fluid-control mechanism allows faster walkers to attain higher ambulation speeds, and unequal excessive swing phases are more pronounced with the constant friction mechanism. Overall, fluid-control mechanisms allow more optimal ambulation performance than the constant friction mechanism.[22]

Hydraulic dampers are the most commonly used form of fluid swing control design. They function by restricting flow of incompressible fluid (commonly silicone oil) through a fixed orifice. This results in a strong dampening force for shin swing motion. Basic designs respond to a narrow range of walking speeds with a laminar fluid flow. More recent designs allow turbulent flow at higher ambulation speeds and exert a correspondingly higher force to dampen shin swing. The patient can walk comfortably at a much wider range of walking speeds, from slow to race walking. As a rule, less active individuals will do well with the basic and less expensive hydraulic units; vigorously active patients will benefit from the more advanced hydraulic knee design using turbulent flow.[20]

A number of additional features are incorporated into the prosthetic knee to meet the needs of specific patient populations. The locked knee mechanism (Table 2) is a feature more suited for weaker and less steady amputees. This includes the elderly amputee population, in whom the locked knee would actually enable higher walking speed and lower cardiac effort despite an awkward stiff-legged gait.[20] Amputees with cardiac dysfunction also benefit from the locked knee mechanism, resulting in a lower heart rate and faster walking speed when compared with patients using the open knee mechanism.[23] Patients who are totally unable to control the knee, such as the stroke patient, would also be suitable candidates for the locked knee mechanism.[20] But younger and healthier amputees often prefer the open knee mechanism, which allows greater ambulation speed. Despite greater energy costs and heart rate, the open knee mechanism functions well in those with greater physical and energy reserves.[24]

There are a number of commercially available hydraulic knee mechanisms using similar fluid control principles. The Hans Mauch S-N-S cylinder, (Table 2) is a widely used hydraulic knee mechanism. This is a high performance design that allows patients with good muscular strength, coordination, and reflexes to participate in challenging activities such as confident descent of inclines, walking down stairs, and performing step-overstep. The Otto Bock 3R80 rotary hydraulic knee prosthesis is a recent development in hydraulic stance that incorporates a weight loading sensitive mechanism. The knee stance mechanism is engaged with normal weight bearing and automatically disengages as weight is transferred to the leading limb during normal walking. This design is more suited for slow and limited ambulation because the brake-stance mechanism may interfere with preswing knee flexion during the early phases of walking.[20]

In general, individuals of smaller stature, such as women and children who have a slow to moderate cadence, are more suited to the smaller, simpler, and lighter hydraulic swing control units. Taller, more active ambulators, especially those involved in competitive sports, will derive benefit from the larger, more powerful hydraulic swing control mechanisms.[20]

Polycentric knee mechanisms

A polycentric knee mechanism consists of four points of rotation, each connected by a linkage bar (Table 2). This allows optimal control of swing and stance phases. One design demonstrates minimal extrusion of the knee mechanism during knee flexion to 90°. This permits better cosmesis during activities such as sitting, which is especially important in patients with long femoral stumps. Another feature of the polycentric knee is better stance stabilization while maximizing ease of preswing knee flexion. With a center of rotation being more proximal and posterior compared with the native knee, the device results in greater leg extension and prosthesis stability. When the knee flexes only a few degrees, the center of rotation is shifted anteriorly, with subsequent ease of continued flexion accompanied by a mild decrease in prosthesis length. This ensures knee stability while the patient is walking at a moderate to brisk pace. In addition, toe clearance can be increased by up to 1 to 2 cm during midswing, leading to less perceived risk of stumbling. This is preferred by patients who are in relatively good physical condition and desire stance phase knee stability.[20]

Advanced prosthetic knee technologies

In conventional pneumatic swing control devices, swing adjustment occurs in a nonlinear manner dependent on the size of the orifice through which air passes. "Intelligent" transfemoral microprocessor-controlled prostheses can change the orifice size according to varying walking speeds to allow appropriate shin swing time. A knee joint sensor detects the swing speed and sends a signal to the stepper motor, which adjusts the valve size in the pneumatic cylinder.[21]

An energy expenditure benefit has been shown at walking speeds that are slower or faster than normal, with a decrease in oxygen consumption of 5% to 9% when compared with conventional pneumatic mechanisms. Ambulating at speeds outside of those designed for conventional hydraulic mechanisms can lead to inefficient gait characteristics such as vaulting, circumduction, and excessive lateral sway.[24]

There are several available types of microprocessor-controlled prosthetic knees including the Endolite Intelligent Prosthesis Plus (Chas Blatchford & Sons), the Seattle Limb Systems Power Knee (Seattle Limb Systems), and the recently released C-Leg (Otto Bock). The C-Leg (Table 2) is an advanced processor-controlled prosthesis that uses a hydraulic cylinder to provide greater swing control and variable hydraulic stance phase control. The shin includes numerous sensors that accumulate biomechanical data such as vertical loading amplitude and sagittal knee movement, and can also determine the direction and angular acceleration of the knee joint. A software analysis system optimizes prosthetic characteristics through a process of data sampling and calculations of up to 60 times in a 1.2-second gait cycle. Studies to date have demonstrated significant benefit only in relatively healthy individuals who are otherwise good ambulators using conventional knee prosthetic technology.[20] When examined using the swing-phase treadmill test, the C-Leg has clearly demonstrated superiority at higher walking speeds when compared with mechanical hydraulic knees such as the Otto Bock 3R45 and 3R80. The C-Leg has demonstrated greatest efficiency in the areas of flexion angle, flexion speed, and extension speed. It was more beneficial at higher ambulation speed in physically fit patients, with one report documenting a walking speed of up to 6.67 km per hour with the C-Leg.[25]

Knee socket–residual limb interface

There are generally two types of transfemoral sockets. The quadrilateral socket allows direct ischial weight bearing, with a relatively horizontal posterior brim exerting a vertical upward pressure on the ischium. On the other hand, the ischial containment socket has a sloping contour in the posteromedial brim designed to rest the ischial tuberosity 3 to 4 cm below the brim edge. This type of socket is usually recommended for patients engaged in high activity sport and running. The ischial containment socket is usually recommended for short, fleshy, unstable stumps and is preferred by bilateral transfemoral amputees. A flexible brim is preferred for this particular socket. The quadrilateral socket is suited for longer firm residual stumps with intact adductor musculature. It is also useful in geriatric and disabled persons who can use canes or walkers to assist in ambulation.[26]

Rehabilitation in above- and below-knee amputations

The proportion of below-knee amputations increased in the late 1960s and early 1970s because of a number of advancements. Incorporation of a posterior flap technique performed with a better vascular supply led to an improved healing rate.[27] Clinicians were also beginning to assess the appropriate amputation level based on Doppler evaluation[28] and measurement of transcutaneous pO_2.[29] At the same time, revascularization techniques were being developed that resulted in salvage of threatened limbs or allowed a more distal level of amputation. Increased awareness of the benefits of distal amputation, relating decreased energy expenditure[30,31] and improved rehabilitation potential,[32] promoted the trend toward more distal amputations. Other significant developments during this period include advances in the care of the diabetic amputee, and improved understanding of nutrition in stump healing.[33]

Recent major advances in the field of prosthetic technology have had a major impact on surgical decision making and technique regarding lower extremity amputation level. There are major distinctions between AKA and BKA, including rates of healing and rehabilitation potential. There is often a fine balance or compromise between increased rates of healing in AKA versus more optimal rehabilitation results and better energy efficiency from BKA.

Of the patients who attained a healed stump, approximately 80% of those with BKA were eventually able to achieve ambulation. This is in contrast to only 38% to 50% of healed AKA patients achieving regular prosthetic ambulation.[34,35] But these outcomes are counterbalanced by a lower 65% to 89% healing rate with BKA as compared with a greater than 90% healing rate with AKA.[28] Not surprising, an even lower prosthetic ambulation rate of 19% is seen in bilateral lower extremity amputees.[36] The presence of coronary artery disease is also a strong negative predictor of successful ambulatory rehabilitation. Of amputees who are able to successfully achieve ambulation, 78% of those with BKA have coronary artery disease as compared with 7% of AKA patients.[37] This must be taken into consideration when determining the most suitable amputation level for a given patient.

Knee disarticulation

Knee disarticulation has significant theoretic advantages over conventional AKA including greater end-weight-bearing along normal proprioceptive pathways, simplicity of technique, minimal blood loss, residual limb balanced by strong muscles, and less energy expenditure with a longer femoral stump. This technique is not often used because of poor residual limb coverage by thin skin and subcutaneous tissue, leading to a higher risk of tissue ischemia and necrosis.[37] It is interesting to note that the metaphyseal cancellous bone and cartilage of the femoral condyles in knee disarticulation are more compliant than the diaphyseal cortical bone and have better shock absorption.[38] In patients with poor rehabilitation potential, a knee disarticulation provides more optimal sitting balance, bed mobility, and transfer compared with an AKA.[39]

Energy expenditure

When compared with AKA amputees, through-knee amputees are able to achieve higher normal and maximum walking speeds (34.5 versus 24.5 m/min and 47.8 versus 29 m/min, respectively), with lower relative (a ratio of energy expenditure rate [mL O_2 consumed/kg/min] at normal versus maximum walking speed) and functional energy costs (a ratio of energy expenditure rate at normal walking speed versus at rest; 71.5 versus 87.7 and 0.092 versus 0.101, respectively). But more distal leg amputations have better ambulatory velocities and energy efficiency.[30] Even a short BKA is more advantageous than a knee disarticulation given that it is well tolerated by the patient and free of complications.[40]

Although many studies advocate limb length preservation and knee joint preservation when possible in amputees with peripheral vascular disease, it has been suggested that in elderly patients, knee disarticulation may become a favored option.[37,38] Elderly amputees with peripheral vascular disease do not seem to derive the theoretic benefit of BKA as compared with more healthy amputees. So it was found that knee disarticulation in patients who would otherwise heal a BKA allowed earlier weight bearing, with a lower risk of the wound complications seen in BKA patients. In addition, the ease of fitting, allowance for direct load transfer, and the stability of a polycentric knee may outweigh the benefits of preserving a compromised native quadriceps-knee mechanism.[39]

Prosthetics

The socket design for a knee disarticulation prosthesis places the biomechanical support at the femoral condyles, a weight bearing surface. The proximal trimlines of the socket do not need to contain the ischium with a high medial trimline, but can be lowered to approximately to one-third of the total limb length. Suspension can be obtained superior to the condyles through a variety of methods that include pads, straps, cut out windows, and inflatable bladders made in the socket. Knee disarticulation prostheses have become much more cosmetically acceptable for patients with the arrival of the four bar linkage polycentric knee joint (Fig. 1). With the older knee prosthesis being simply applied to the end of the residual limb, an inefficient gait pattern occurs, with the prosthetic knee rotational axis being much too distal in comparison with the contralateral native knee. The polycentric knee provides an instantaneous stable center of rotation, an anterior weight bearing axis, and allows ease of prosthetic knee flexion on unloading. With the patient seated, the prosthetic knee folds under the socket, allowing seating with both knees on the floor.[38]

Below-knee amputation

Energy expenditure

Patients with BKA have more efficient energy expenditure during ambulation and can achieve higher ambulation speeds when compared with patients with more proximal amputations.[19,30] As perceived by the patient, the length of the BKA stump does not correlate with comfortable walking speed. But objective measurements indicate that long stumps had only a 10% increase in energy expenditure compared with short stumps, which had a 40% increase in energy expenditure above normal.[31]

Interestingly, in able-bodied individuals,[41] unilateral amputees,[39] bilateral amputees,[42] and patients with hip disarticulation,[15] the chosen walking speed is commonly that with the most energy efficiency or demand.

Knee socket-stump interface

The socket for a prosthetic limb is an important interface governing parameters such as energy transfer to the prosthesis and residual limb viability. There are a number of factors to be considered before fitting a definitive prosthesis. First, the residual limb should be "matured" by using a compressive stump shrinker. This process results in subcutaneous tissue shrinkage, residual muscle atrophy and hypertrophy, and thigh and residual limb lengthening with return of collateral circulation.[43]

Although a BKA length can vary from 2 inches below the knee to just above the ankle, the usual recommended length is approximately 6 inches, which allows for near normal ambulation. In general, the ideal distal residual limb circumference should be equal to or less than that at the level of the knee joint. An excessively bulbous residual limb may be difficult to fit.[44] For residual limbs less than or equal to 4 inches, a patellar-tendon bearing prosthesis would be needed.[43]

The standard socket design for below-the-knee amputees is the patellar-tendon bearing prosthesis, which was developed in the University of California at Berkeley in 1957. But not all patients tolerate this design, and some report excessive pressure on the patellar tendon. Knee flexion may be limited, and adventitial bursae, skin abrasions, and dermatitis can develop.[45] The total surface bearing socket (TSB) uses suction for prosthetic securement. Development of a silicone-lined socket allows improved pressure seal during high socket seal activities such as athletics, bending, and sitting. In a case series of 32 patients studied by Hachisuka and colleagues,[46] the TSB socket was well tolerated, with overall satisfaction reaching 75%. Patients were most satisfied with the comfort, ease of swing, and less piston movement. It was believed that the weight was well distributed over the entire stump, with less pressure against the patellar tendon area. But donning the socket may be difficult in patients with visual, sensory, and motor deficits. The TSB socket is sensitive to changes in stump circumference from edema or gradual decreases in stump diameter with progressive stump maturity. Negative issues associated with the TSB include distal stump pain, discomfort during knee flexion, and excessive perspiration inside the socket.[46]

Another socket that was developed in the mid 1980s was the Icelandic Roll-On Silicone Socket (ICEROSS). A prefabricated roll-on sleeve, closed-end liner is turned inside out, then placed on the residual limb, and proximally rolled up over the knee. The socket is then secured onto the liner, most commonly by means of a shuttle lock, in which a pin on the end of the liner engages automatically with a spring loaded clasping mechanism located in the bottom of the prosthetic socket. The ICEROSS uses a silicone sleeve that is flexible and elastic, conforming very closely to the amputee's residual limb so that air is excluded and effective retention is maintained throughout gait despite minor volume changes. Friction is transferred to the outer portion of the liner and socket interface with the skin, with the inner portion of the liner remaining relatively friction free. So, the ICEROSS socket allows the stump to conform to a noncompliant socket. Care should be taken in prescribing the ICEROSS socket to the elderly because of their compromised ability to remain independent, and it should be done only if adequate handling and maintenance are assured. In addition, the ICEROSS socket should not be recommended if the patient is unable to don the prosthesis and maintain proper hygiene.[38]

The below-knee bypass prosthesis (Fig. 2) is generally indicated for delayed healing amputations and has the advantage of preventing the sheer forces applied by the conventional total contact socket. It is also used in amputees with residual limb pain, stump dermatolopathology, recent skin grafting, or bone overgrowth. The advantage of this prosthetic design is that the rehabilitation process can continue despite stump complications. This is particularly important for bilateral amputees, in whom prolonged bed rest secondary to stump complications can lead to severe muscle atrophy and can hamper or delay later rehabilitation attempts.[47]

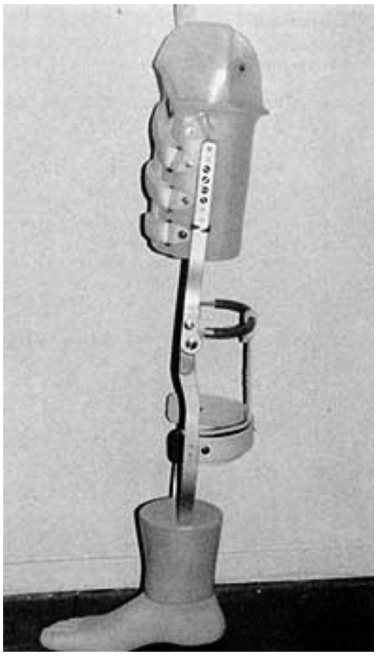

Figure 2. Below-knee bypass prosthesis.

Immediate postoperative prosthetic placement

The idea of immediate postoperative prosthetic placement (IPOP) was first proposed by Berlemont in the late 1950s, and the procedure was subsequently modified by Weiss and associates.[48] Rigid plaster IPOPs were used in the United States starting in the 1960s and required significant skill to apply. It was not usually removed for up to 7 days.[49] Early results from IPOP were mixed, with one study in 1968 showing a beneficial outcome in patients with BKA, with a 90% primary healing rate and only 10% requiring a higher amputation because of suboptimal healing. But other studies at the time noted a deleterious effect of IPOP, with a 10 times greater reamputation rate in BKA performed for peripheral vascular disease, perhaps from further compromise of marginal vascular supply.[50] Another report found no difference in wound healing, wound infection, stump necrosis, need for reamputation, and postoperative mortality when compared with soft dressing.[51]

In recent years, after lower extremity amputation, patients are often fitted with a soft stump dressing that follows the stump contour. The stump wound is allowed to heal, with daily dressing and wound assessment. The patient remains nonambulatory until the fitting of a permanent prosthesis 4 to 6 weeks after surgery and may delay physical therapy and rehabilitation.[52] Recent studies support the use of IPOP to facilitate operative recovery, rehabilitation, and achieve independent ambulation.[53,54] There appears to be significantly less wound infection, fewer nonhealing wounds, and less surgical wound dehiscence requiring surgical debridement or reamputation at a higher level. Overall, patients using IPOP experienced fewer wound complications, with improved wound healing and earlier prosthetic fitting and training.

Conventional IPOPs use inflatable air cells placed inside a universally sized rigid shell that can be inflated to fit the limb stump (Air-limb [Aircast]) and is secured with straps. The Air-limb is inflated to 20 to 30 mmHg, and a knee suspension sleeve is applied.[54]

Prosthetic foot

The prosthetic foot serves as an important, multifaceted component of all lower extremity prostheses. The primary purpose of the prosthetic foot is to serve as the anatomic ankle and foot. A prosthetic foot should provide the functions of joint simulation, shock absorption, a stable weight bearing base of support, muscle simulation, and cosmesis. There are essentially four different designs of prosthetic feet (Fig. 3). The SACH (solid ankle cushion heel) foot is the most frequently used foot (Fig. 3A). It consists of a rigid keel covered by a semi noncompressible foam, and a synthetic rubber heel wedge. The cushion heel compresses when weight is applied, allowing the forefoot to approach the floor. The amount of this simulated plantar flexion depends on the relative softness of the heel material and weight of the amputee. Because the keel is rigid, the SACH foot does not provide dorsiflexion. SACH feet are relatively inexpensive and are manufactured in a wide range of sizes.

Articulated designs allow motion between the foot and shank that occurs around an articulation in the region corresponding to the anatomic ankle. There are two variations of articulated design: single axis and multiple axis. Single axis (Fig. 3B) includes a transverse ankle axis that permits the foot to go into plantar flexion and dorsiflexion, with all motion taking place around a single axis. As the foot plantar flexes, a posterior bumper (a synthetic rubber in a cylindrical shape posterior to the ankle axis) is compressed, resisting the motion. The action of the compressed bumper simulates the action of the dorsiflexors of the native foot. In the opposite direction, the motion is resisted by the dorsiflexion bumper, which is positioned anterior to the ankle axis. Multiple-axis prosthetic feet (Fig. 3C) permit movements in any direction: plantar flexion, dorsiflexion, inversion, eversion, and a slight amount of rotation around a vertical axis. This type of prosthetic feet accommodates to uneven walking surfaces and absorbs some of the torsional forces created in walking, reducing torque to the limb by the socket. Although there is greater latitude, the increased motion may create instability in patients with marginal coordination. An increase in function comes at the cost of an increase in size and weight, and may require more main-

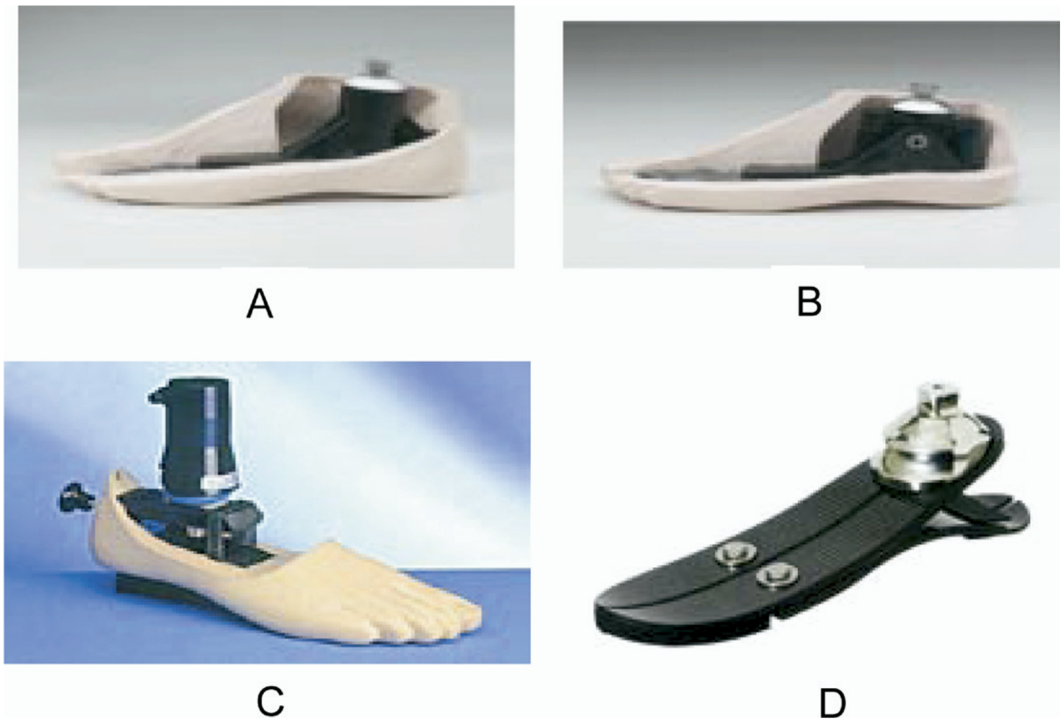

Figure 3. (A) SACH (solid ankle cushion heel) foot (Courtesy of Ohio Willow Wood). (B) Single axis foot (courtesy of Ohio Willow Wood). (C) Poly axis foot. (D) LP Vari Flex Foot (courtesy of Ossur).

tenance. Newer versions are now fabricated of carbon composite structures to reduce weight.

Prosthetic feet are predominantly designed for walking, but many young and active lower limb amputees have expressed the need to be more active, resulting in a new generation of prosthetic feet (energy storing feet, Fig. 3D). But there are vast differences between force vectors for walking and running. During running, there is a period of free flight and a downward force at heel contact that exceeds the body weight two to three times. In walking, this force barely exceeds body weight.[55] The energy storing feet incorporate a shock absorption mechanism in the form of a flexible keel that stores energy to be used during toe off. As an amputee's cadence increases, the amount of time on the heel will decrease, and the amount of time on the forefoot will increase. More time on the forefoot will increase the force exerted to the forefoot, causing an increase in the dorsiflexion moment. Since the design of new materials, such as graphite composite, Delrin (DuPont), Kevlar (DuPont), polyurethane elastomers, and flexible rubbers, the dorsiflexion moment allows the keel to compress, absorbing energy, which, during push-off, is released and aids in propelling the amputee's limb forward. In addition to the energy storing and releasing capabilities, the feet allow a more fluid motion, which produces a more normal gait.

Foot amputations

Midfoot and transmetatarsal amputations are typically performed for digital gangrene, osteomyelitis of the forefoot, or nonhealing ulcerations of the forefoot. Acute infection is usually stabilized before primary closure, so midfoot or transmetatarsal amputations are often performed as staged procedures. The technique of maintaining a smooth metatarsal or midfoot parabola and beveling the distal osseus structure lessens the likelihood of ulceration postoperatively from excessive pressure once the patient begins rehabilitation and ambulating. Custom molded shoes and inlays, which off-load and support excessive pressure areas, are often prescribed postoperatively once the patient has completely healed the amputation. Often, recurrent problems are avoided with a well constructed shoe and inlay. The more proximal the amputation, the more tendon insertions are at risk, which could result in an acquired equinas or varus deformity as a result of a tendonous imbalance if the tendons are sacrificed. These acquired deformities and any osseous prominence often result in pressure points leading to ulceration. Acquired equinas deformities with limited ankle dorsiflexion and subsequent plantar ulceration of the forefoot respond well to an Achilles tendon lengthening, reducing the forefoot pressure associated with ulceration. Partial foot amputations are limb salvage procedures that avoid more disabling energy inefficient higher leg amputation. Patients with foot amputations consume less energy in ambulation than individuals with more proximal leg amputations.[56] Cardiac function and oxygen consumption after midfoot, Syme's, below-, through-, and above-knee amputations are directly related to the level of amputation. In addition, the ability to walk long or short distances is decreased in patients with more proximal amputations.[30] Transmetatarsal and midfoot amputations are limb salvage procedures, preserving the extremity and allowing the patient to expend less energy throughout the gait cycle.[57,58]

Prosthetics

In the management of partial foot amputations, the device is, at times, referred to as a "prosthosis," which is a combination of a prosthesis and an orthosis. Many prosthosis designs include principles that are used in the fabrication and design of a foot orthosis, ankle foot orthosis, and lower limb prosthesis. With all levels of amputation, a number of factors are assessed in determining the proper prosthosis. First, the amputee's ability to control the joints and muscle function of the foot and ankle must be evaluated. Muscle imbalance and joint instability or deformity, either fixed or correctable, should also be assessed. Skin integrity and sensitivity and the level of scar adhesion must be evaluated and taken into account when determining the prosthosis design and material composition. The patient's weight bearing ability is assessed as well. Finally, the patient's goals and aspirations with regard to activity level and cosmetics of the prosthosis must be considered.

A transmetatarsal amputation eliminates the entire normal forefoot load-bearing capacity. This means that it is no longer practical to transfer the total forefoot ground reaction force used during midstance to terminal stance onto the plantar surface of the residual foot.[59] Prosthetic designs for this level of amputation attempt to correct this and are similar to the molded inner sole (foot orthotic), with the addition of a forefoot filler. This type of design serves to maintain the correct relation between the residual foot and the patient's shoe. To assist in reducing the increased forefoot ground reaction force, a full length carbon graphite plate is placed under the toe filler. Shoe modifications, such as the addition of a rocker sole, with its apex proximal to the amputation site, can further reduce the plantar surface pressure onto the residual limb.

Tarsometatarsal and midtarsal amputations have functional limitations similar to those of transmetatarsal amputation. The reduced surface area makes the task of interfacing a prosthosis to the residual limb even more difficult with the absence of push-off at the end of stance phase, which compromises the quality of gait. The prosthosis for

Figure 4. (A) Ossur AFO Dynamic (courtesy of Ossur). (B) Partial foot prosthesis (courtesy of Custom Composite).

this more proximal level of amputation requires a higher supportive design, typically proximal to the ankle joint, which allows a ground reaction force to imitate the sense of push-off for the patient. One model incorporates an ankle foot orthosis incorporated with a forefoot filler. This design may also be used for nonamputees with inadequate plantar flexion strength, allowing dorsiflexion resistance at the end of the stance phase, leading to a sense of push-off. A prosthosis based with the rigid ankle foot orthosis design prevents dorsiflexion by blocking the ankle joint motion at the appropriate angle. The socket interface resists dorsiflexion anteriorly, effectively blocking this motion, which allows a more energy efficient gait cycle.[56] A prosthosis constructed with an anterior shell and a posterior opening is more functional than the posterior shell ankle foot orthosis design. Some modern designs incorporate a full length energy storing footplate of carbon composite with a strut extending to an anterior shell (Fig. 4A). Other designs use a high anterior shell and a carbon composite footplate with a foam material creating the desired foot shape (Fig. 4B).

Silicone prosthetics were originally introduced to provide superior cosmetic appearance (Fig. 5A). Silicone prostheses have also proved successful for amputees with fragile skin and adherent scar tissue. A silicone prosthesis allows for successful restoration of balance and a more normal gait, if sufficiently reinforced to obtain a degree of rigidity required for the patient's activity level. In all patients, a comfortable socket, a balanced foot, and an optimal gait pattern are the objectives for users of partial foot prosthetics.

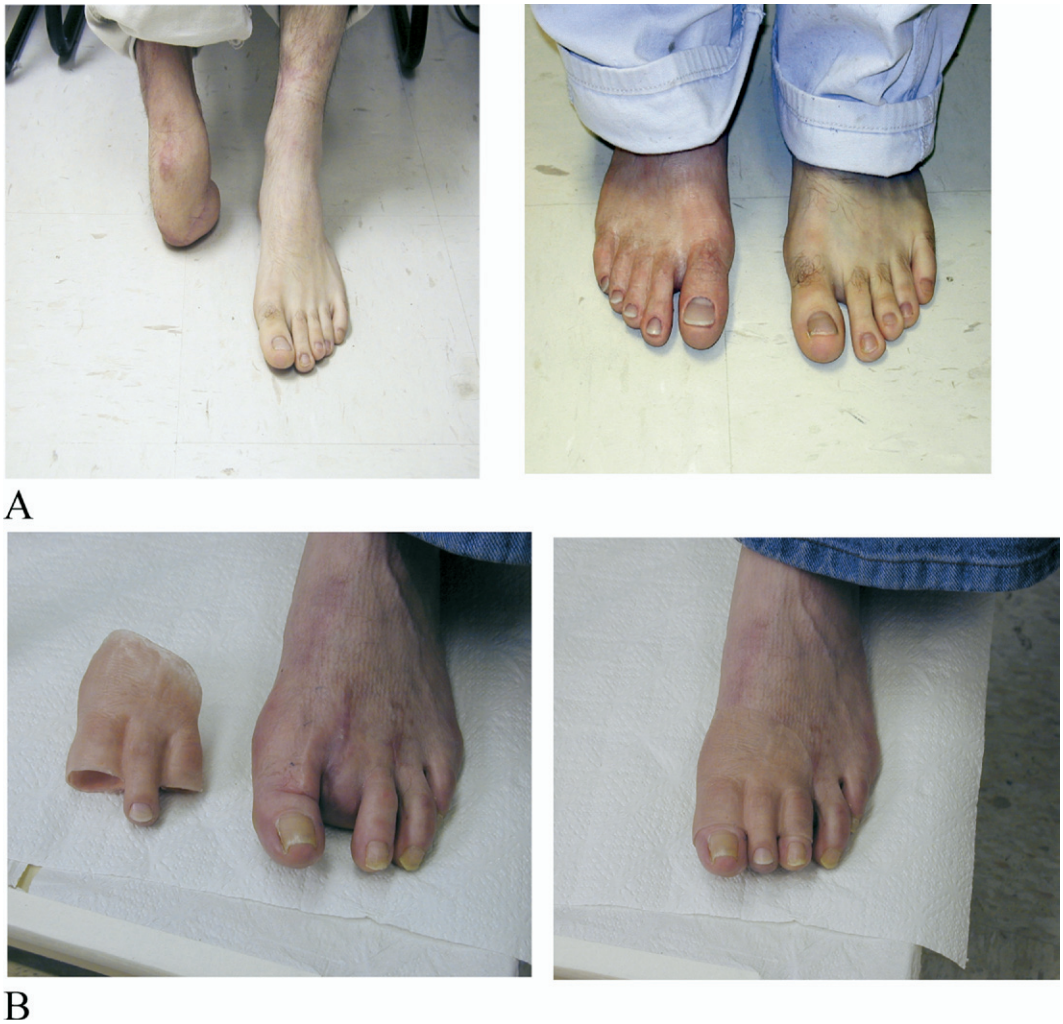

Figure 5. (A) Partial foot silicone prosthesis (courtesy of Alternative Prosthetics). (B) Silicone "filler" prosthesis provided by Alternative Prosthetics.

Digital and ray amputation

As with partial foot amputations, the goal of digital or ray amputation is to maintain a functional foot that allows the patient to ambulate with as little energy as possible. The digital amputation can be performed at three different levels: the distal interphalangeal joint (distal Syme's), proximal interphalangeal joint, or metatarsalphalangeal joint. The ray amputation, either partial or complete, is performed to the level of uninvolved bone, which may require a resection of the entire ray. Resection of the entire first or fifth ray often leads to tendon imbalances and acquired deformities with new pressure points as opposed to central ray amputations. The digital and ray amputations are extremely functional, with few complications, in patients with adequate healing potential.

Prosthetics

The consequence of amputation of one or more toes is a functional reduction in the forefoot load-bearing areas. Digital amputation can lead to increased pressure at the metatarsal head level from transverse deviation of the adjacent digits, leading to an instability at the metatarsalphalangeal joint and resulting in a more prominent metatarsal head. The increased pressure is greatest when the hallux is amputated. In normal gait, loss of a digit is not a functional problem. But at higher activities levels, the loss of the hallux creates difficulties that result from the loss of active push-off. Most prosthetics for toe amputations consist of a toe filler. The toe filler's function is to maintain proper shape and prevent deformation in the shoe. Toe fillers range from simply packing material placed in the shoe (ie, lamb's wool) to a modified foot orthotic made of silicone or foam

(Fig. 5B). Toe spacers also prevent deviation of the other toes in the transverse plane and offer excellent cosmesis (Fig. 5), which can have a psychological benefit for the patient.

Perspective

Approximately 75% of AKAs and BKAs occur in patients aged 65 or older. In this elderly population, arteriosclerosis accounts for 90% of amputations.[60] Amputees with peripheral vascular disease are often already operating at approximately 80% of maximal capacity during "normal" walking pace. So there is a greatly decreased reserve for increasing their level of activity. At successively higher levels of amputation, the walking speed decreases, with an accompanying increase in oxygen and energy consumption. The result is a severe limitation of ambulation and activity level, translating into decreased walking speed and distance.[30]

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