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Rehabilitation of People with Lower-Limb Amputations

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Abstract Rehabilitation of persons with lower-limb amputation is a complex endeavor that requires the consideration of a multitude of factors. This article provides an overview of the current practice of prosthesis prescription, mobility training, and the utilization of wheeled mobility options in the clinical care for this population. Recent technological advancements have helped fit persons with lower-limb amputation with more functional, better fitting, and less activity-limiting artificial limbs and wheelchairs. This is exemplified in modern computer-controlled prosthetic components and biomechanically optimized socket-fitting methods, as well as light weight and versatile wheelchairs to supplement or replace prosthetic devices. In the research setting, technology has enabled new approaches to the kinematic and kinetic assessment of prosthetic interventions, and the development of more accurate fitting and evaluation methods. Despite the noted progress in the field, there is still a considerable gap between the functionality of a sound leg and even the most advanced prosthesis. It can be predicted that continued research efforts will be undertaken to further close this gap.

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Introduction

Between amputations and congenital deficiencies as mechanism of lower-limb loss, amputations are by far most prevalent, representing more than 99 % of all cases in the US [1]. In the population of under 15 year olds, congenital defects represent the majority of limb loss cases [2], as the accumulated risk for amputation is comparably low [3, 4]. The incidence of the major causes for amputation, namely traumata, tumors, and most of all, vascular diseases, increases with the time at risk and as a direct result of physical aging [5–7]. Currently, more than 80 % of amputations in the United States are secondary to dysvascular disease, with an equal division between peripheral vascular disease and diabetes. Less than 10 % result from trauma [8].

Rehabilitation of persons with lower-limb amputation poses, therefore, generally slightly different challenges than the rehabilitation of persons with congenital limb loss. Members of the latter population begin coping with their impairment early in life by developing habits and motion patterns that can be quite different from able-bodied individuals. Persons who have experienced lower-limb amputation as an adult, however, are faced with the task of modifying and adapting their established habits and motion patterns in order to cope with their acquired impairment. This adaptation taxes both mental and physical resources, and is, therefore, generally more successful in younger, healthier, and physically more active persons [9, 10].

Defining rehabilitation goals and selecting appropriate interventions requires careful consideration of a person's capabilities, functional demands, and goals. Ideally, the function level prior to the amputation would be restored. However, it is not always simple to define the appropriate point in time prior to surgery at which to allocate that baseline. Depending on the diagnosis, a patient may have been immobile for many years before eventually requiring limb amputation, which can be the case in persons with joint degeneration and vascular problems [11, 12]. A prosthetic limb may then help to restore a level of function that exceeds the immediate pre-operative level of function. On the other hand, it is possible that the severity of the limb loss, e.g., level and number of amputations, prohibit a rehabilitation goal that is oriented on the pre-surgery status. This is often the case in traumatic amputations [13, 14].

Prosthetics are a necessity if the rehabilitation goal is restoration of locomotor abilities to some extent. If prosthetics are not indicated because of the severity of the limb loss and/or reduced capabilities of the person, wheeled mobility aids are prescribed [15••, 16]. In many cases, both prosthetics and wheeled mobility aids are used to complement each other in different situations and activities of daily life. This paper reviews recent findings and developments in the fields of mobility training, wheelchair equipment and skills, prosthetic componentry and fitting, as well as advanced assessment tools for the rehabilitation of lower-limb amputees.

Mobility Training

A predominant concern of many amputees and a detriment to prosthetic mobility is the loss of limb stability and control [17–19]. Users of lower-limb prosthetics experience an increased risk of stumbling and falling that is associated with the level of amputation, age, and the severity of comorbidities, such as vision or hearing loss, joint contractures, impaired sensitivity, and muscle atrophies [20–23].

Exercises to increase the voluntary control of the prosthetic limb as well as the stability of standing and walking are essential components of mobility training [24–26]. With increased perceived safety grows the confidence in the prosthesis and thus the likelihood of fully utilizing the functions of the artificial limb [27, 28]. Particularly in post-operative rehabilitation, when convalescents have not yet accumulated much prosthesis experience, it is important to balance the safety and dynamic characteristics of a prosthesis—that are generally on opposing sides of the same equation—in correspondence to the successively changing capabilities of the user.

Individuals with lower-limb amputation must learn about the expected behavior of their new artificial limb that is inevitably different from the previously known behavior of their original limb. The lost function of muscles and joints, as well as afferent nerves cannot be adequately replaced, thus necessitating compensatory strategies and limitations. In terms of stability, a prosthesis that includes a foot/ankle component, e.g., any leg prosthesis for amputations proximal to the level of the (partial) foot, requires the user's CoG to be in a comparably small area vertically of the foot during stance phase. Any static or dynamic load that deviates from that area (and that would be effortlessly accommodated with some slight corrections of ankle angle in a sound leg) cannot be accommodated by the prosthetic ankle joint due the lack of muscular control. For example, during initial ground contact in a prosthetic step, when the foot is placed in front of the body's center of gravity, the resulting plantar-flexing moment may—depending on the components used to construct the prosthetic limb—either cause the prosthetic foot to (passively) deform in the sense of plantar-flexion, or translate into a knee flexing moment at the proximal end of the shank segment. The first is a safer option but comes at the expense of dynamic efficiency, as the impact energy during initial ground contact is not used to propel the body forward but is lost in the elastic deformation of the compliant foot. Similar considerations are made in the question of prosthetic knee joint selection, where again a range of options exist that emphasize one feature of able-bodied gait or another. It requires experience to utilize the respective prosthetic components in the intended fashion and to understand the fall risks in different situations.

A secondary objective of mobility training, after ensuring static and dynamic stability, is to reduce the burden of walking with a prosthetic limb. Generally, users of lower-limb prostheses expend more metabolic energy than able-bodied controls in walking and other activities of daily life [29, 30, 31•]. They are also more susceptible to acute or chronic degeneration of their bodily structure, such as joints in the residual limb, the contralateral limb, or the spine [32–35, 36••]. Therapeutic training can help reduce these undesirable effects by emphasizing the importance of symmetrical motion patterns. Symmetrical gait is believed to be least taxing in terms of energy consumption and overuse wear [37–39]. However, it has been debated that some degree of asymmetry is more functional, given that the physical condition after amputation is signified by considerable asymmetry [40, 41].

In addition, rehabilitation goals of lower priority that may nonetheless be of increased importance for individual users of prostheses include factors such as cosmetic inconspicuousness of prosthesis use and the ability to perform certain vocational or leisurely activities [42–44].

Initial prosthetic fitting is an iterative process, corresponding with the physical and psychological changes of the convalescent. In the same sense, the strategy for mobility training must be adapted over the course of the rehabilitation regimen. Parameters that tend to change over time are for instance the condition of the residual limb muscles and skin.

Immediately after surgery, sutures cannot be subjected to high stresses or loads [45, 46]. There is swelling of some extent and often a great amount of pain. In this phase, it was long assumed that the best course of action was bed rest to allow the healing process to take its course [47–49]. However, the idea of reducing the recommended time span until early prosthetic fitting has become more popular [50, 51] and has led to the practice of IPOP (immediate post-operative prosthesis) prescription. These prostheses are designed to be swiftly fitted to a residual limb that still has limited tolerance to contact pressure and is still changing in volume and shape. Their main purpose is to allow the patient to leave the bed even if for short periods of time, thereby reducing considerably the comorbidities that are secondary to prolonged bed rest, such as decubiti, thrombosis, muscle atrophies, but also depression and fatigue [52–54]. It is also believed, that early verticalization helps patients in their subsequent stages of prosthesis training, by preparing body and mind for the challenges ahead [9, 55].

As wound healing progresses and the residual limb approaches its eventual volume, customized sockets are fitted, usually in a sequence of several check-sockets for temporary use. Ideally, during this time there is also an opportunity to try out different prosthetic componentry options. As mentioned before, a compliant foot might be the option of choice in the early stages as it facilitates a more stable ground contact. Later on, when priorities change, the same user may be better served with an Energy Storage and Return (ESAR) foot [56, 57]. Electronically controlled or even powered components [58–60] are often only prescribed after trials with conventional parts have confirmed a respective indication. The main reasons for that are certainly of economic nature, but it may also be helpful to have experienced different prosthetic technologies in order to utilize all the benefits of the latest high-tech feet and knees and to be prepared for possible emergencies, e.g., malfunctions or empty batteries.

Prosthetic knees support a range of functionalities, differing widely between models [61, 62]. Important for the successful mobility training is, therefore, the understanding and practicing of the, respectively, prescribed knee system. Polycentric knees have stance safety built in by interconnected multiple axes, which in extension effectively moves the instantaneous center of rotation of the system outside of the actual joint structure, much like is the case in a human knee [63]. In any other but the extended position, there is no resistance to the flexion motion, which helps with the swing phase of the step. Users of such knee systems have learned to only apply their body weight on the prosthesis when the knee is fully extended. If they are subsequently fitted with a stance phase controlled hydraulic knee, they need to abandon that concept [64]. The advantage of such a knee, using for instance the SNS (swing 'n stance) hydraulic unit, is that it can be adjusted to provide a flexion resistance in the stance phase, which allows a more natural and comfortable gait especially on downward slopes and stairs. Training is required to overcome the previously valid fear of falling when the knee starts buckling and to control the new prosthesis reliably.

Even after a prosthesis user is well versed in the use of their artificial limbs, it may be recommendable to monitor the gait pattern at regular intervals [65]. Gait asymmetries and relief postures can become habitual and progressive and should be rectified as possible by continued instruction and training.

Wheelchair Equipment and Skills

The more proximal the level of amputation [16, 66] and the more severe the impairment due to age [67] or comorbidities [68] in a given individual is, the greater the benefit of supplemental wheelchair usage [69]. This may be best illustrated by the following clinical example.

A 35-year-old young male with amputation of both legs above the knees saw the need to supplement the mobility with his prosthetic legs, which he felt were limiting his functional ambulation to fulfill his role as a full-time employed civil engineer, husband, and father. He needed to be able to maneuver over uneven recreational outdoor terrain such as playgrounds, sports fields, family theme parks, and rougher terrain at construction sites [70, 71]. Therefore, he wished to explore usage of a manual wheelchair in conjunction with his prosthetic limbs. He was not interested in a power wheelchair, as he preferred active self-propulsion of a manual wheelchair, and a power wheelchair would also limit his transportation and vehicle options needed for his work. He decided that an ultralight manual wheelchair equipped with durable lightweight carbon-fiber wheels and quick release axles would work best for him, as he would have to transfer in and out of his car multiple times; the lighter the components that he would have to manipulate, the less strain on his upper extremities [72, 73]. The wheels were equipped with ergonomic hand-rims designed to reduce the risk of repetitive strain injury to wrist joints and protect the hand from injuries and lacerations [74, 75]. He was provided with wheelchair mobility training to become familiar with sensing the center of gravity (CoG) and its effect on the chair's responsiveness and stability. This was important as his CoG would be different when using the chair with and without his prosthetic legs. Instead of recommending an "amputee axel plate" that would allow the rear wheels to be set further back to increase chair stability, yet in turn, would shift the weight onto the casters and thereby compromising maneuverability as well as significantly increased risk for repetitive strain injuries to the upper extremity joints during harmful upper biomechanics during active self-propulsion [72, 76–78], he chose a solid backrest with multiple back-to-seat angle adjustments. The backrest support worked very well, as he could use a 90–100° back-to-seat angle when wearing his prosthetic legs and "crunch" the back-to-seat angle to 85°, when using his wheelchair without the legs, reminding him to keep his shoulder slightly forward to prevent his chair from tipping backward [79••]. The adjustable features and light weight components of the ultralight manual wheelchair provided him with an ideal supplemental mobility device option to let him accomplish the high-energy consuming activities for raising his young son and for field work use at construction sites.

Prosthetic Components and Fitting

Several 100 prosthetic feet and prosthetic knee models are estimated to be available in the market today, a circumstance that illustrates the perpetual efforts of the orthopedic industry to provide ever more sophisticated and innovative solutions for the prosthetic fitting of people with amputations. The catalog of one manufacturer alone lists a portfolio of 30 foot and 35 knee models [80, 81]. Most interesting among the recent developments are probably powered foot/ankle units and similarly powered knee units that strive to replace lost muscle function [31•, 82, 83]. Active motion is not provided with more conventional microprocessor-controlled knee systems that, however, facilitate safe and energy-efficient ambulation by automatically adjusting flexion and extension resistances most accurately to the, at any given point in time, required values [84, 85]. Other approaches of improving technology have consisted of reducing weight and complexity [43, 86], enhancing adaptability [87, 88], and optimizing biomechanical properties of passive prosthetic knees and feet [89, 90].

Socket-fitting options have by trend increased in number since the introduction of silicon liner-based suspension in the 1980s [91] which marked the first big paradigm shift in socket fitting. The previously used approach of suspending (trans-tibial) prostheses by epicondylar containment, thigh cuffs, or derivations thereof has continually lost importance in the subsequent decades. Today, most trans-tibial prostheses are equipped with a liner suspension system of some kind [92] and it is likely that this trend will persist, given the noted advantages over traditional approaches to include reliable suspension, ease of donning and doffing, and applicability even for unfavorable residual limb shapes [91]. The concept of elevated vacuum suspension that has proven beneficial on a number of accounts [93–95] has contributed to the superiority of liner suspension.

In trans-femoral prosthetics, liner suspensions are only slowly becoming more prevalent, which is owed to the less readily accommodated properties of typical residual limbs, namely their length, shape, and tissue composition [96]. Nonetheless, remarkable improvements in terms of socket compliance, hip range of motion, appearance (damage to clothing), and muscle utilization have been achieved with modern socket designs and materials [97–100].

A different approach, that eliminates the need for a socket entirely, is available with the method of osseointegration [101–103]. Here, the prosthesis is attached directly to the bone via a titanium fixture, inserted to the bone and connected to an abutment which penetrates the skin. Despite some remarkable early results, the inevitable contraindications and complications of the procedure have so far prevented wide-spread use in the US. Even less prevalent is the procedure of allotransplantation of lower limbs that aims at a full restoration of the pre-amputation status, albeit at the expense of possibly severe complications [104].

Advanced Tools

Recent advances in motion and gait analysis have benefited the practice of assessing and optimizing prosthetic care in research and clinic.

The simplest form of a gait analysis is an observational analysis conducted by a clinician. For example, the test person ambulates up and down a hallway and is evaluated. While simple, a qualified clinician with extensive training and experience is necessary to identify gait deviations. Observational analyses are also limited due to the inability to simultaneously evaluate motion at different joints (e.g., hips, knee, and ankle) and planes of motion (sagittal, coronal, and transverse). Conventional gait analysis is conducted in a laboratory and uses a motion capture system to simultaneously measure Ground reaction forces (GRFs), joint kinematics, joint kinetics, and muscle activation. GRFs are measured with force platforms embedded in the laboratory floor or with a treadmill. Joint kinematics and kinetics are measured using multiple synchronized cameras to record positions of markers placed on anatomical landmarks of the test participant, combined with force plate data. Marker positions and GRFs are used with anthropometric measures and biomechanical models to obtain joint kinematics and kinetics. Muscle activations are measured with electromyography using surface or indwelling electrodes. Gait is measured as a test person walks through the motion capture space. Multiple trials may be required to ensure people accurately step on the force platform without altering their gait to target the force platforms. After multiple gait cycles are recorded, data are processed and analyzed off-line and a report is generated.

In recent decades, force platforms have been embedded in treadmills to allow GRFs to be collected continuously during consecutive gait cycles. Most treadmills used for gait analyses have two belts with a force platform under each belt, called dual-belt or split-belt instrumented treadmills. The dual-belt system allows each limb to be evaluated independently, medial–lateral GRFs to be measured, and belts to be controlled independently or together. Numerous studies have compared overground and treadmill gait resulting in altered kinematics [105, 106], kinetics [105, 107], and energy costs [108, 109]. Causes of these differences are speculated to be multifactorial, including altered visual feedback [110, 111]. Treadmill walking lacks the relative motion between an observer and the environment (optic flow), thus altering visual feedback [110, 111].

Novel technologies such as the computer-assisted rehabilitation environment (CAREN) provide optic flow during treadmill walking which has resulted in temporal–spatial parameters and joint kinematics similar to overground walking [112, 113]. The CAREN system (Fig. 1; Motek Medical, Amsterdam, Netherlands) is a virtual reality environment combined with an optoelectronic motion analysis system and motion base. Virtual reality provides a computer simulation of a real-world environment that is experienced by the user through a human–machine interface [114]. Test persons are immersed into the virtual environment via surround sound and images projected on one of the three available options: a flat screen, 180° cylindrical screen, or 360° dome enclosure. The motion analysis system allows real-time tracking of person's motion and GRFs throughout the gait analysis. The motion base is 6° of freedom platform controlled with hydraulic actuators and is embedded with a dual-belt instrumented treadmill. D-flow software (Motek Medical) provides real-time data streaming and control of these systems. Test participants are immersed in this real-time feedback loop where their motions and behaviors are considered inputs. Output devices return motor-sensory, visual, and auditory feedback to the participant. For example, joint kinetics can be measured. Outputs include real-time visual feedback of joint kinetics, physical movement of the motion base, visual movement of a projected cursor, and auditory feedback. Physical and visual perturbations can also be applied to the person to simulate environmental conditions during clinically important events. For example, a trip can be simulated with a unilateral treadmill belt acceleration or deceleration applied at specific gait events (heel strike, mid-stance, toe-off, etc.) [115]. Biomechanical gait analyses and training using the CAREN system have benefitted lower-limb amputees with demonstrated clinical improvements in pelvis and hip kinematics [116], reduced oxygen consumption [116], and improved vertical GRF symmetry [117].

Mobile gait analysis methods have been proposed to overcome some of the limitations of conventional approaches, namely the limitations on capture volumes and environments. While virtual reality systems, as described above, provide a most versatile and accurate technical approach to that end, their use is limited to large research laboratories as they are prohibitively expensive for smaller scale clinical applications. Wearable electronics, however, are more accessible and have been adapted for prosthetic gait analysis.

The option of easily inserting measurement equipment directly into the weight bearing structure of the limb is unique to prosthetics gait analysis. This poses a considerable advantage over able-bodied gait analysis methods, as most forces and moments of interest can be recorded directly where they occur and do not have to be derived from force plate and motion analysis data using a set of more or less valid assumptions [118, 119]. It should be noted that such direct measurements have also been conducted in non-amputee subjects, which, however, required the development of implantable wireless sensors to be inserted in the body as a part of endo-prostheses [120, 121].

Researchers have long utilized specially prepared load cells in experimental prostheses studies [122, 123], and the advent of microprocessor-controlled prosthetic components about 20 years ago has brought about the inclusion of sensor technology in commercially available products. But only in recent years have stand-alone sensor units been marketed with the primary purpose of collecting gait data for interpretation by a clinician or researcher, rather than by an integrated microprocessor (Fig. 2). There is yet a limited amount of published research that investigated or utilized the capabilities of such products, but early findings suggest that they are useful for a range of applications [124].

Interpretation of load cell-based gait data has proven somewhat challenging, which is mostly owed to the unilateral nature of the available information. Since only kinematics from the prosthetic leg is recorded, many popular outcome variables—most notably bilateral symmetry—cannot be computed. It is, therefore, necessary to define unilateral variables that can be used as predictor of prosthetic performance. Various approaches have been proposed, including the use of force-moment curves [125], GRF segments by step cycle compartment [126], peak forces during stair gait [127], and step-by-step variability [128]. The limitation was circumvented entirely in a study on gait symmetry in persons with bilateral trans-tibial amputation [129].

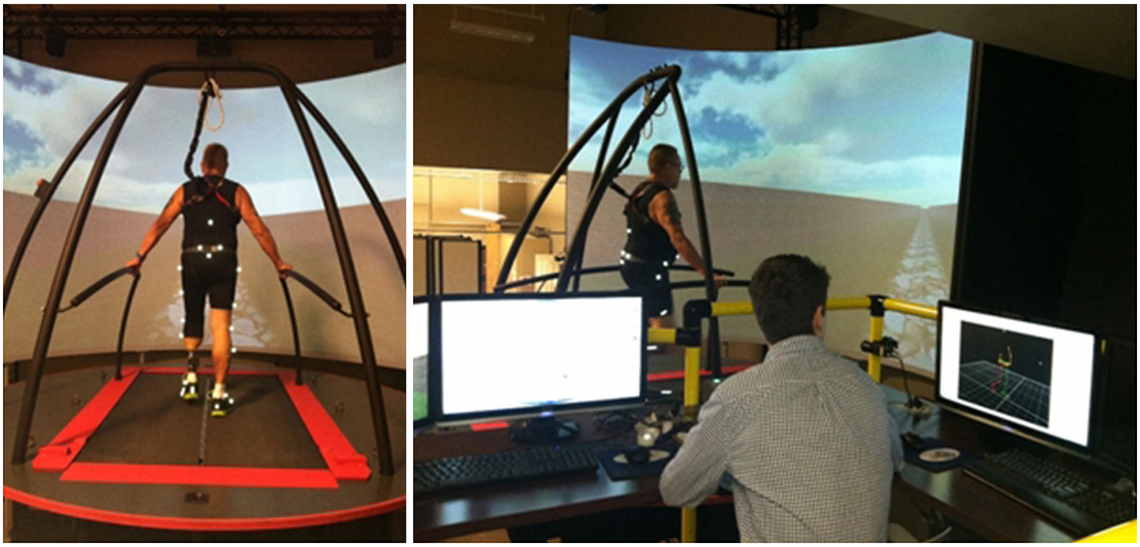

Fig. 1 Gait analysis of a lower-limb amputee using the CAREN extended system at the VA Human Engineering Research Laboratory in Pittsburgh, PA

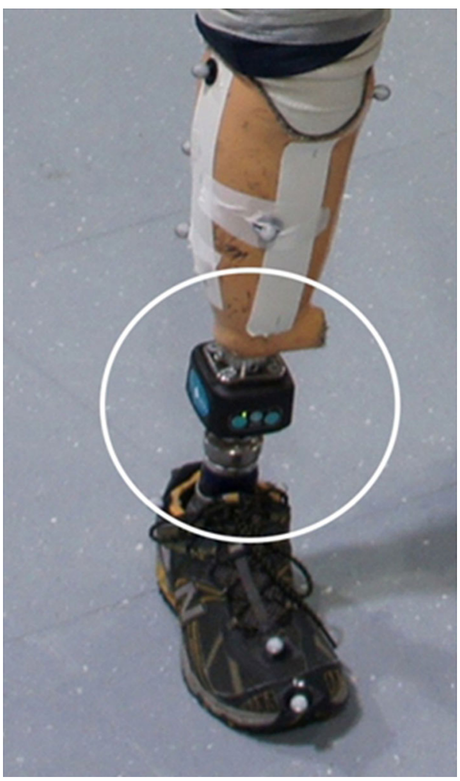

Fig. 2 Load cell for temporary inclusion in the prosthesis structure (here iPecs mobile gait lab, RTC Electronics, Ann Arbor, MI)

In clinical applications, mobile sensors have primarily been used for activity assessment purposes. The Galileo system (Orthocare, Tacoma, WA) is discussed as an accurate and reliable tool for the definition of K-levels in lower-limb prosthetics users [130]. Load cell data supported methods of prosthesis alignment optimization [126, 131] are still of restricted practical utility and will have to be further investigated to improve their accuracy and adaptability.

Conclusions

Rehabilitation of persons with lower-limb loss after amputation is an endeavor that has benefitted greatly from technological advances over the last decades. Major improvements have become possible by modern prosthetic componentry, wheelchair design, therapy regimens, and outcome assessment methods. At the same time, it must be acknowledged that rehabilitation efforts often still fall short of the ideal of restoring a functional level identical to the one from before the amputation surgery. Many persons with lower-limb amputations are still limited in their mobility and not fully satisfied with their artificial limbs or wheeled mobility solutions. Continued efforts are, therefore, indicated to further improve rehabilitation care and community reintegration for persons with lower-limb amputations.

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Compliance with Ethics Guidelines

Conflict of Interest G. Fiedler, J. Akins, R. Cooper, S. Munoz, and R.A. Cooper all declare no conflicts of interest.

Human and Animal Rights and Informed Consent This article does not contain any studies with human or animal subjects performed by any of the authors.

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- •• Of major importance

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