CHAPTER

# LOWER LIMB ACTIVE PROSTHETIC SYSTEMS—OVERVIEW

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## 23.1 INTRODUCTION

Changes in lower limb mechanics, sensory feedback, and power output associated with lower limb loss have substantial impacts on the gait biomechanics and energetics of individuals with amputations. During unimpaired level ground walking, the ankle produces significant net positive ankle power throughout the stride [1,2], with other locomotor tasks, such as walking up slopes, relying on additional net positive power output at the knee [3]. As a result, lower limb amputation and the associated loss of ankle and knee power production is linked to slower self-selected walking speeds and increased energy expenditure [4]. People with transtibial and transfemoral amputation expend up to 30% and 60% more metabolic energy, respectively [4,5], when compared to unimpaired individuals walking at the same speed. Similarly, the preferred walking speed of individuals with amputation can be 10%–65% slower than the average walking speed of unaffected individuals, depending on the level of amputation and the walking surface [4,6].

Individuals with lower limb amputation often adapt compensatory gait strategies that can lead to significant changes in gait dynamics, joint loading and work, and muscle activity in the affected and unaffected limbs. For example, unilateral below-knee prosthesis users tend to favor their unaffected limb [7,8], which often endures greater joint forces, moments, and stress during daily activity [9–11]. The hip joint on the contralateral limb can produce up to three times more work than the hip joint of an unimpaired individual, likely as compensation for lack of power production of conventional passive prostheses [1,12]. The affected limb generally exhibits significantly lower knee moments when compared to the contralateral limb or to individuals without impairment [11]. In addition, affected limb knee flexor and extensor muscles can exhibit higher co-contraction levels [13], a strategy commonly adapted to stabilize the joint in uncertain conditions (e.g., [14–16]). Such marked gait asymmetry and joint loading can lead to chronic health issues, such as premature joint degeneration, early-onset osteoarthritis, and joint pain [7].

Slower walking speeds, increases in muscle co-contraction, and changes in gait symmetry can also be indicators of compromised balance [15,17,18]. About half of all individuals with amputation experience at least one fall a year, usually during walking. Of these individuals, 75% fall at

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least twice, and 10% experience at least one fall that results in serious injury requiring medical treatment [19]. About half of individuals with lower limb amputation report a fear of falling and 60% list inability to walk on natural surfaces, such as wooded areas or fields, as a major limitation. As the causes of these limitations, about 50% of affected individuals cite an inability to adequately sense the walking surface, 40% an inability to walk without a stabilizing gait aid, and 30% difficulty in controlling their prosthesis [20]. Gait asymmetry, changes in joint loading often leading to pain, and elevated risk of falling and other injuries severely reduce mobility and overall quality of life of individuals with amputation.

## 23.2 BACKGROUND

Most currently available commercial prostheses rely on elastic elements, such as carbon-fiber plates or mechanical springs, to absorb and return energy passively [21,22]. Such energy storage and return devices cannot actively generate power, unlike biological lower limb muscles which produce up to 80% of the mechanical work per step [23−25]. More so, these devices can only be mechanically adjusted and lack the ability to actively adapt to the user or changes in the environment. Even partial actuation in prosthetic devices appears to benefit user performance. Several semiactive devices, which do not generate net positive power but allow for controlled damping or transition between gait modes, have led to improved user balance and stability by actively adjusting joint angles based on gait type [26,27]. Effective control of fully actuated devices could address the current limitations of prosthetic devices even further, potentially improving balance, comfort, and gait symmetry in individuals with lower limb amputation.

Development of active lower limb prosthetic devices began to gain traction in the late 1970s, with the introduction of a tethered, hydraulically powered knee prosthesis [28] (Fig. 23.1A). This system demonstrated the feasibility of active prostheses to generate prescribed movement at the desired times in the gait cycle. Although no biomechanical user data were collected, individuals with amputation tended to prefer the active device with simple control paradigms to their prescribed passive knees. Evaluation of devices that followed (e.g., Fig. 23.1F) revealed that an active knee prosthesis could produce kinematics and power output similar to intact limbs and lead to a more kinematically symmetric gait [33], with one such device produced commercially (Fig. 23.1I). In turn, different control architectures could accommodate walking on sloped terrain (implemented in devices similar to Fig. 23.1C,D), while still maintaining the trajectories of an unaffected joint [38].

Previous research has also suggested that active ankle power modulation could have significant gait benefits. For example, active inversion/eversion assistance implemented on an experimental ankle prosthesis (similar to Fig. 23.1E) reduced the metabolic cost of walking without significantly affecting gait mechanics or the perception of comfort [39]. This suggests that energy economy is correlated with the active need to correct body dynamics on a step-to-step basis, and that active devices can improve performance even if overall mechanics remain unchanged. The effects of active ankle push-off work on energy expenditure are less clear. One commercially available device (based on Fig. 23.1G) has been shown to lead to energetic reductions of up to 8% [34,40], as well as increased walking speeds [41,42], in users with transtibial amputation during walking on level

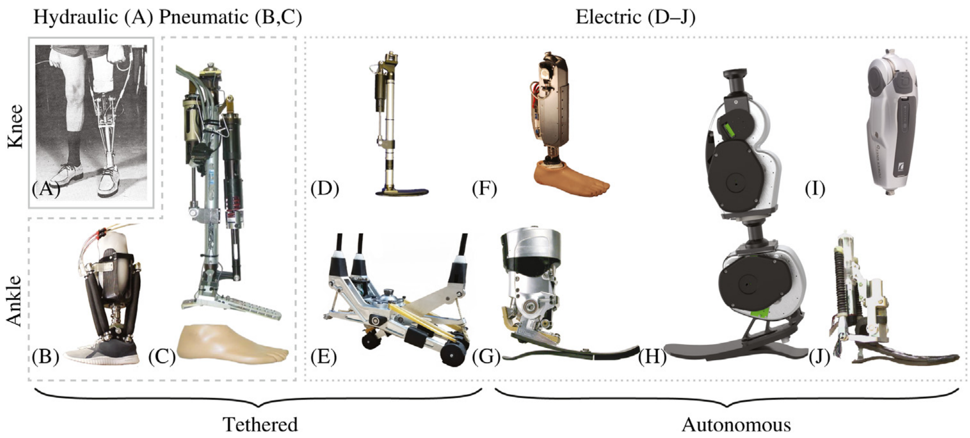


**FIGURE 23.1**

Selected autonomous and tethered active ankle and knee prosthetic devices. With the exception of (E) and (J), all devices are actuated only in the sagittal plane. Solid, dashed, and dotted outlines indicate hydraulic, pneumatic, and electrical devices, respectively. (A) The first active knee prosthesis with echo control [28], (B) ankle prosthesis with pneumatic artificial muscles and myoelectric control from North Carolina State University, based on [29] (image courtesy of S. Huang), (C) Pneumatic cylinder actuated ankle and knee prosthesis controlled by an impedance-based strategy, similar to [30] (image courtesy of M. Goldfarb), (D) Prosthetic knee joint with passive ankle and impedance-based control [31] (image courtesy of M. Goldfarb), (E) Ankle prosthesis emulator with three individually actuated digits for frontal and sagittal plane actuation [32] (image courtesy of V. Chiu), (F) Variable impedance knee prosthesis [33] (image courtesy of E. Martinez-Villalpando), (G) Bionic ankle foot prosthesis that served as the basis for the BiOM [34], and consequently the Ottobock Empower (image courtesy of A. Grabowski), (H) The Open-source Robotic Leg, a scalable and customizable knee and ankle prosthesis system intended as an easily accessible platform for research testing [35] (image courtesy of E. Rouse), (I) The Össur Power Knee, the only commercially available powered knee prosthesis to date [36] (image courtesy of Össur), (J) The SPARKy 2 ankle prosthesis that allows for fore-aft and medio-lateral control of the joint [37] (image courtesy of T. Sugar).

ground, when compared to their prescribed device. In contrast, using a tethered prosthesis to increase prosthesis mechanical work in isolation has not been found to affect energy expenditure [43]. Differences in outcomes could be due to variation in other prosthesis characteristics, rather than the addition of active power itself, including control features that might affect balance. To facilitate well-controlled comparisons of prosthesis function outside the laboratory, the Open-source Robotic Leg Prosthesis (Fig. 23.1H), an active ankle and/or knee prosthetic device, is currently being developed for laboratory applications [35]. The standalone device can be configured both mechanically and in software, allowing for evaluation of a range of configurations with just one device.

Differences in user response to active assistance may also be due to the difficulties of hand tuning control parameters for each individual. For example, even minor changes in timing and magnitude can have substantial effects on energy expenditure [44–46]. More accurate control optimization strategies could allow devices to be more effectively tuned to each user and have already shown promising results in improving performance of exoskeleton assistive devices (e.g., [47,48]). Similarly, devices relying on more direct control from the user, for example by using muscle activation signals to control the device (as in Fig. 23.1B), could lead to reductions in energetic cost [29].

The biomechanical response of people with amputation to active devices is promising, but very few active prostheses are currently on the market. Further advances in energetically efficient actuators (like those used for Fig. 23.1J), controllers capable of adapting to the user and the environment, and accessible platforms for prosthetic research, could significantly increase the availability of powered prostheses to affected individuals. This could improve balance, reduce falls and other injuries, redistribute joint loading, and decrease the severity of premature joint generation and other mobility restrictions.

## 23.3 SYSTEMS

Building lower limb prosthetic devices that can emulate the behavior of natural joints is a long-standing goal of researchers aiming to enhance mobility for people with amputations. Effective prosthetic devices would need to generate sufficient torque while still being relatively lightweight, adapt to a variety of terrains and locomotion tasks, and effectively interface with the user. For comparison, for an 86-kg individual, the ankle can provide up to 450 W peak power and 150 Nm of peak torque [23], with an ankle/shank mass of less than 5 kg [1]. Prosthetic devices must be even lighter, depending on the weight of the residual limb post amputation, while still providing comparable energetic outputs. Effective device controllers would provide intuitive integration with the user and an ability to conform to day-to-day activities. Finally, proper socket fit is critical for user comfort and effective attachment of the device. Although not discussed in this chapter, socket fit and design is one of the most valued features of a prosthesis [49], and requires further development. Active prosthetic devices have only been a research focus since the 1970s, yet significant progress in advancing this technology has been made. These advances, in both autonomous and research-oriented devices, are discussed in the following sections.

### 23.3.1 MECHANICAL CONFIGURATION AND ACTUATION APPROACHES

To be most effective, active lower limb prosthetic devices may need to provide torque and power outputs similar to the capabilities of the biological leg. Electric actuators are well suited for these purposes and have been previously implemented in research and commercial prosthetic devices. Pneumatic actuators can also generate a range of behaviors necessary for human gait but generally require off-board air compressors, thus limiting their use primarily to research. Several hydraulically powered devices have also been proposed, although hydraulic actuators are mainly used for controlled damping and are not discussed in detail in this chapter. Finally, tethered systems that place actuation components off-board are crucial in advancing the development of prosthetic devices, as they provide the ability to test a wide range of control ideas without the need for building new hardware. The trade-offs in benefits and limitations of several mechanical configurations and actuation approaches are discussed further.

#### 23.3.1.1 Remote actuation

As lower limb prosthetic device designs move towards smart, actuated systems, there are still many aspects of device control and human–robot interaction that remain unclear. Testing new control and actuation approaches often requires building new hardware for evaluation, which is both expensive and time consuming. For this reason, several testbed, tethered prosthesis systems move control and/or power components from the wearable device and place them off-board. This allows for the worn system to remain lightweight and low-profile, while still maintaining the ability to produce forces and torques similar to those seen in biological joints. In addition, tethered devices are generally tested within a laboratory or clinical setting, which provides access to lab-based equipment and a means to test approaches which would otherwise be infeasible. For example, one tethered ankle prosthesis used ground reaction forces from an instrumented treadmill to identify center of mass sway and to control inversion/eversion torques generated by the device [39]. Implementing such a control architecture would be more challenging in a standalone device, which would need a separate, independent system of detecting ground reaction forces. Tethered devices are more conducive to long-term studies in a controlled environment, which lead to insights into gait mechanics and control of individuals with lower limb amputation.

Tethered prostheses have been powered by hydraulic, pneumatic, and electric actuators, with the transmission often relying on flexible tubes or cables so as to not restrict user movements. Hydraulic and pneumatic approaches are well-suited for off-board actuation, as they can be connected to external pumps or compressors via flexible tubing. The first active prosthesis systems relied on an off-board hydraulic pump and computing hardware to drive an artificial knee joint with an on-board hydraulic cylinder [28] (Fig. 23.1A). The setup served as a proof of concept that active devices could lead to gait patterns similar to unimpaired gait. It was later adapted to test other control architectures, such as using myoelectric activity to dictate knee damping magnitude [50], or enforcing unimpaired gait kinematics to remove vaulting during stance [51]. Similar tethered pneumatic systems also place cylinders directly on the device, while an air compressor and computing hardware remain off-board. Such research prototypes have helped demonstrate, for example, that direct proportional myoelectric control can effectively control ankle torque [29], and that adaptable ankle stiffness may improve gait in individuals with lower limb amputation [52].

Recent advances in electrically powered tethered testbeds provide a way of exploring an even broader range of prosthesis design and control approaches. A typical system consists of powerful off-board motors and control hardware, connected to a robotic ankle prosthesis, or end-effector, by means of Bowden cables and flexible tethers [53,54]. High-torque, low-inertia electrical motors generate device peak torques and powers up to 50% higher than seen in the biological ankle, while more easily tracking fast-changing torque profiles when compared to pneumatic and hydraulic systems of comparable weight. This approach also significantly reduces the worn mass and profile of the prosthesis since, unlike in tethered pneumatic and hydraulic systems, no part of the actuator is placed on the leg. In turn, this allows for the development of tethered prostheses with more degrees of freedom. To date, end-effectors developed for this testbed include devices with one, two, or three degrees of freedom (Fig. 23.1E), which allow for plantarflexion, inversion/eversion, and center of pressure control by actuating various components of the device [32,53,54].

#### 23.3.1.2 Pneumatic actuators

Pneumatic artificial muscles, a type of compliant pneumatic actuator, have been used for actuation of prosthetic devices. One of the more common pneumatic muscles, the McKibben artificial muscle, consists of a cylindrical inner rubber bladder, enclosed in a braided sheath [55]. As the inner bladder inflates, it shortens, expands radially, and produces a pulling force between the two actuator end-points. Adding a damper in parallel with the pneumatic actuator drives actuator performance to more closely emulate the force–velocity dynamics of biological muscle–tendon units [56,57]. Furthermore, folding the membrane of the actuator in on itself in a cylindrical shape along its central axis, akin to accordion bellows, reduces hysteresis by mitigating material strain during inflation [55,58,59]. This leads to more predictable dynamics of the actuator, reduces actuator failure rates, allows for simpler control strategies that do not need to take into account actuator deformation, and yields a position error of less than 2% [60]. In all, pneumatic artificial muscles have the advantage of being extremely lightweight and compliant, with the capability of generating high peak torques and variable stiffness spring-like behavior similar to biological muscle [55,61].

Like biological muscle, pneumatic actuator muscles only provide pulling forces and often need to be arranged in an antagonistic configuration for prosthetic devices. This is exemplified by several designs of prosthetic ankles, which consist of a one degree of freedom ankle joint attached to a commercially available foot. The muscles are then attached to the front and back of the foot and to protrusions either on the socket or the pylon of the device (e.g., [29,62], Fig. 23.1B). A similar arrangement was also demonstrated in a pneumatic knee prosthesis, where two artificial muscles produced bidirectional motion about the knee joint while the ankle joint remained passive [63].

Pneumatic cylinder actuators offer certain trade-offs compared to soft pneumatic muscles. One of the main benefits of pneumatic cylinders is their ability to generate force in both directions. This force is independent of actuator displacement and simplifies control by mitigating the need to compensate for actuator dynamics [55]. Pneumatic cylinders can also operate at much higher pressure and thereby produce larger forces than soft pneumatic actuators. However, all pneumatic devices need to consider the trade-off between pressure, area, and flow rate. Greater force production can be achieved by increasing pressure or actuator cross area, but both approaches introduce some drawbacks. For example, increasing pressure is more likely to lead to mechanical failures, while increasing the cylinder size results in higher flow rates and power loss across valves. The importance of these trade-offs is apparent in a powered ankle and knee prosthesis that relied on two double-acting pneumatic cylinders to actuate the joints [30]. The particular configuration limited the maximum size of the actuators, in turn reducing the maximum torque that could be produced as well. As a result, only 76% of the maximum biological ankle torque was achieved, since a larger actuator would impede the range of motion of the prosthesis. Nonetheless, the device was able to produce ankle and knee kinematics comparable to nonimpaired gait, and provided significant assistance to the user.

Pneumatic actuators offer certain benefits but several drawbacks exist. For one, air compressors tend to be heavy and inefficient, so pneumatic systems are rarely untethered. Efforts to develop portable pneumatic actuation systems, for example by using monopropellants [64,65], could address this limitation in the future. Another drawback is that pneumatic artificial muscles expand significantly when inflated and are not suited for low-profile applications. Finally, pneumatic actuators are often loud, making them impractical for standalone prostheses.

#### 23.3.1.3 Electric actuators

Electric motors are the most common actuators in active prosthetic devices, especially for autonomous devices. However, heavier, larger motors are generally required to produce the peak and average power demonstrated by human joints during gait [66]. In addition, most electric motors achieve peak power at high speeds, which is not always conducive to generating appropriate prosthesis behavior [66]. Transmissions with high gear reductions can address this issue, but also introduce high impedance when the system is unpowered. More complex transmissions involving springs or clutches can improve torque control or allow energy capture and return, at the cost of added mass and size.

Several knee prostheses have been developed that rely on ball screw systems to translate rotary motion into linear motion. One such setup was able to demonstrate power profiles comparable to biological knees during slow walking [31] (Fig. 23.1D). Although tethered, it was estimated that the device could provide power for an 85-kg user to walk for up to 5 km on a small battery pack. The knee joint of another autonomous knee and ankle device relied on a similar setup, but with the ball screw assembly attached to a slider crank mechanism that actuated the joint (based on Ref. [30]) [67]. Although such configurations can generate the required power outputs for level walking, they cannot take advantage of the passive dynamics of the leg or other energy storage and return mechanisms, leading to considerable power consumption [68]. The only commercially available powered knee device to date, the Össur Power Knee, is similarly actuated [36].

Adding passive compliance to the transmission system can provide additional benefits. Springs placed in parallel to the actuator can reduce actuator torque and power requirements, particularly when the desired behavior of the motor is spring-like and not antagonistic to the elastic element. A mechanical spring in parallel with a ball screw mechanism of a joint, for example, can supplement motor torque and reduce actuator energy use [67]. Springs instead placed in series with the actuator can help regulate and maintain joint torque and protect against damage due to impact (for example, at heel-strike) [69], although this makes the control of joint angle and speed more difficult. A range of devices utilize this series elastic actuator (SEA) approach to improve performance. One such autonomous knee device used two antagonistically placed SEAs to emulate the elasticity and damping characteristics of a biological knee [33,68] (Fig. 23.1F). The only commercially available ankle prosthesis to provide active power, the BiOM (now the Ottobock Empower), incorporates both series and parallel elastic elements [70–72]. A rotary SEA helps with the control of joint torque, while the parallel spring allows for a lower gear ratio and, consequently, faster joint movement. The prosthesis is similar in weight to an intact limb (Fig. 23.1G), provides adequate power and torque output to the user, and was even demonstrated to lead to reductions in user energy expenditure when compared to passive devices [34,70].

However, SEA components have drawbacks, such as increased prosthesis weight and reduced control bandwidth. Elastic components can also be used with locking mechanisms to improve energy storage and release during the gait cycle [73], or with systems that transfer energy from the knee to the ankle joint [74], both of which can improve device performance. The previously mentioned Open-source Robotic Leg Prosthesis can be configured to include series elasticity of variable stiffness to facilitate research on a wide range of prosthesis designs [35]. Series elasticity can be added to both the ankle and knee joints of the device, which rely on an electric motor coupled to a multistage belt-drive transmission for torque generation. The configuration results in a relatively low transmission ratio, usually associated with smaller mass and improved force control.

## 23.3.2 CONTROL APPROACHES

One of the main challenges of developing active prostheses is identifying controllers that can reliably improve performance during a given locomotor task. Walking follows a cyclical pattern and imitating this behavior mathematically is an attractive option. Feedforward control may be sufficient for certain cyclic tasks but does not allow for human input or significant variability in the gait cycle. To provide reactive control and handle a greater variety of tasks, some prosthesis controllers incorporate varying levels of online feedback from the user in order to drive the device or direct state transitions.

### *23.3.2.1 Gait pattern generator control*

Prosthesis joint torque or angle dynamics are often prescribed as predefined functions of some state of the system. Such gait pattern generators can be time-dependent or based on measures such as joint angle and velocity, user muscle activity, or gait phase. One of the earlier active knee prostheses relied on this approach by "echoing" leg dynamics of the contralateral side. Angle trajectories from the intact side were recorded and scaled, and played back on the prosthesis half a gait cycle later [75]. Participants adapted to the system quickly during steady-state walking, although such enforcement of biological knee trajectories did not improve user hip dynamics [76]. Furthermore, since the angle trajectory profile was defined through instrumentation of the sound-leg, this control approach enforces a control lag and can only be used to control devices for individuals with unilateral amputation, and limits human–device interaction.

Referencing predefined joint dynamics with respect to time or gait phase can mitigate time delay concerns and allow for prostheses to behave independently from the contralateral limb. The Spring Ankle with Regenerative Kinetics (SPARKy) prosthesis, for example, was controlled by expressing able-bodied ankle moments as a time-dependent function [37,77–79]. The configuration of the device was such that a low-power motor could regulate the loading of an in-series spring that then recoiled to provide positive power later in the step [80,81]. Torque generation started at heel-contact, with the motor moving the spring through a predetermined pattern configured to assist in steady-state gait. Alternatively, the phase of one or more variables can be used to define the desired prosthesis torque. One such approach utilized thigh angle as the phase variable to effectively generate more diverse cyclic, as well as discrete motions in a knee prosthesis [82]. The gait cycle was divided into several segments, as determined by foot contact, with the phase of the thigh angle mapping to various desired torque profiles for each segment. This control approach naturally scaled to gaits at different speeds, nonsteady-state tasks such as crossing obstacles, and even backwards walking. Furthermore, the control approach could then be adapted for more intuitive tuning of the desired knee trajectories by using a control interface that allowed clinicians to adjust certain sections of the ankle and knee angle trajectories [83]. A similar approach mapped the states of the residual limb to reference states of the missing knee joint using statistical regression of data from unimpaired gait. In other words, the prosthetic knee was driven relative to residual hip motion, based on the hip–knee relationship seen in unimpaired gait. This allowed for successful generation of locomotion with near physiological gait patterns, as well as for stair ascent and descent.

Gait patterns can also be successfully mimicked using mathematical models, rather than mapping to predefined states. For example, a model can characterize joint behavior as a simulated spring and damper, with different parameterizations for different phases of the gait cycle (e.g.,

[30]). Such impedance-based control approaches only generate net power when switching between the gait phases, but can produce joint dynamics similar to unimpaired walking and generate the torques and power required for even, sloped, and uneven terrain walking [30,38,84]. Alternatively, some models generate gait patterns using biologically inspired approaches. One such neuromuscular model, comprising a human ankle-foot complex with Hill-type muscle dynamics, received prosthesis angle and angular velocity as inputs and generated the desired ankle torque command [85]. Without additional sensing of the walking surface, the system was able to successfully adapt to changes in ground slope. When modified to include muscle length and velocity terms, it could also demonstrate adaptation to walking speed [86]. A similar approach has been demonstrated in a powered prosthetic knee [87]. Adding history-dependent muscle properties, such as a winding-spring system mimicking the role of titin in muscle [88], can produce variable speed walking and stair ascent without explicit changes to model parameters [89]. It is important to note that the presented controllers rely on an estimation of gait dynamics, which may be limiting in real-world, unpredictable environments.

#### *23.3.2.2 EMG-based control*

Active prosthesis control generated from electromyographic (EMG) signals can potentially provide intuitive, volitional control of the device. However, electromyography signals obtained from muscles in the residual limb vary in quality and availability between users [90], and EMG electrodes are either invasive or prone to noise and movement artifacts. This makes EMG control approaches challenging to develop, with early myoelectric controllers primarily relying on EMG signals to produce knee-lock during stance (e.g., [91,92]). Some adaptations have used EMG to modulate joint impedance, for example, by employing flexor and extensor thigh muscle activity to determine the damping magnitude of a knee prosthesis [50,93], without generating any active work.

Although residual limb muscle recruitment patterns tend to be highly variable between individuals, they tend to be consistent from stride to stride and suitable for certain types of feedforward control of prosthetic devices [90]. As a result, several active devices have utilized direct myoelectric control. One knee prosthesis relied on a weighted summation of residual thigh flexor and extensor muscles to define prosthesis joint stiffness and set point [94]. With this approach, a participant with a transfemoral amputation could control the device during steady-state walking, although knee dynamics significantly differed from unimpaired gait. Further addition of a state-dependent damping term to the myoelectric torque controller allowed participants to ascend and descend stairs [95]. Ankle prostheses controlled directly with myoelectric signals have also been developed. In one pneumatically powered device, artificial plantarflexor muscle pressure was directly regulated by filtering and rectifying residual limb gastrocnemius muscle activity [29]. With less than an hour of training, a participant with a transtibial amputation was able to control the device and produce functional gait. Muscle pattern adaptation was further improved by adding visual feedback, leading to increased ankle power and ankle positive work during walking [96].

More often, EMG-based control for active prostheses is used alongside additional control paradigms, such as gait pattern generators (see Section 23.3.2.1), in order to avoid relying on direct EMG control. To control a virtual ankle joint angle, one approach used a flexor and extensor muscle model that used EMG signals from the residual limb gastrocnemius, soleus, and tibialis anterior muscles to determine the model's force−velocity characteristics [97]. Similarly, EMG signals from the residual limb have been used as input to a Hill-type muscle model of ankle behavior [98].

A powered ankle prosthesis was controlled using a finite-state machine, with ankle torque gain at late stance selectively driven by the model. This control architecture allowed one participant with a unilateral transtibial amputation to modulate net ankle work and peak ankle power.

#### 23.3.2.3 Motion intent detection control

Natural transitions are seamless and intuitive, without requiring distinct actions (e.g., pressing a button) from the user. Since prosthetic devices are often governed by multiple finite-state controllers designed for different gait phases, intent recognition algorithms are necessary to differentiate and transition between these controllers. One such algorithm relied on a probabilistic model to successfully differentiate between standing, sitting, and walking on a powered ankle and knee prosthesis, based on measurements of knee, ankle, and socket kinematics and forces experienced at the foot [99]. After training the model using approximately 60 trials of prerecorded walking and standing data from an individual with a transfemoral amputation, the model generated a database that could classify a range of activity modes during prosthesis use. The classification approach resulted in reliable mode-recognition and could be adapted to include other locomotion modes, although longer training periods would likely be required. Electromyography has also been successfully used for motion intent recognition [100], in contrast to being used to directly control the device as described in Section 23.3.2.1. In combination with ground reaction force data, electromyography is more informative in identifying user intent than just prosthesis kinematics. In fact, pattern recognition algorithms applied to muscle activation signals from lower limb muscles have been shown to identify up to seven gait modes for users with transfemoral amputation, with classification error rates between 4% and 15% [101]. In practice, a feedforward neural network was able to accurately dictate transitions between walking on level ground and stair ascent/descent for an ankle prosthesis, using residual limb gastrocnemius and tibialis anterior muscle signals [72]. Similarly, a quadratic discriminant analysis classifier, using residual limb hamstring and quadriceps muscle signals, reliably identified flexion and extension modes with less than 5 minutes of training data [102]. Residual limb thigh muscle activity can even be used to identify intent of both knee and ankle flexion and extension during nonweight-bearing activities in individuals with transfemoral amputation [103]. A linear discriminant analysis classifier recognized knee and ankle motion with approximately 90% accuracy using signals from nine residual limb muscles located in the thigh. Finally, developments in targeted muscle reinnervation in individuals with lower limb amputation are showing promise in helping identify user intent [104]. During the amputation surgery of one patient, nerve branches previously used to control the ankle joint were redirected to innervate two hamstring muscles. After recovery, the patient was able to volitionally contract the hamstring muscles when intending to move the ankle joint. Using pattern recognition algorithms, EMG signals from the hamstrings could provide robust, intuitive transitions between multiple gait modes in a prosthetic ankle with less than a 2% error rate. However, people usually take several thousand steps per day, meaning that even a 99.9% success rate in identifying gait modes could lead to several falls a day, depending on the consequences of a classification error. Near-zero error rates or robust error recovery would be required to develop a reliable, usable device.

#### 23.3.2.4 Tuning control parameters

Currently available active prostheses show only modest improvements in energy expenditure and balance during walking on level ground, when compared to passive systems (e.g., [34,42]). This is possibly due to active prosthetic devices typically being controlled using actuation profiles similar to those of unaffected individuals, although optimal assistance likely varies between users and from unimpaired mechanics [105]. Tuning the device to each user could address this issue but conventional prosthetic device fittings are typically based on subjective clinical evaluations [106,107], which may be ineffective for tuning more complex active devices. In addition, user dynamics fluctuate with time in response to adaptation and increased exposure to an active device [108,109], making effective control parameters for each individual difficult to identify. Methods for automatically identifying optimal user-specific characteristics could improve the effectiveness of active prosthetic limbs, especially for devices with balance-enhancing control.

Several control optimization algorithms have been effective in improving human performance by systematically adapting assistive exoskeleton control parameters. One such approach relied on a gradient descent technique [110] to successfully optimize the onset time of assistive ankle torque, with the goal of minimizing user energy expenditure [109]. Unlike gradient descent optimization, which is susceptible to noise and does not scale well with more complex controllers, strategies such as the Covariance Matrix Adaptation Evolution Strategy (CMA-ES) and Bayesian optimizations are potentially better suited for optimizing a larger number of control parameters in noisy conditions. CMA-ES, in particular, was tested on individuals walking with an ankle exoskeleton providing unilateral assistive plantarflexion torque as defined by four parameters, and led to optimized assistance that reduced energy expenditure between 14.2% and 41.5%, and 24% on average [47]. Such reductions in metabolic cost due to an assistive device are the largest reported to date, and are a several-fold improvement over hand-tuned controllers [111], even outperforming devices that provide assistance to multiple joints (e.g., [112]). Bayesian optimization does not require multiple parameter evaluations like CMA-ES and is specifically well suited for optimizing noisy and changing systems, but becomes computationally impractical when attempting to tune many parameters simultaneously. When used to optimize step frequency with the goal of minimizing energetic cost, Bayesian optimization required significantly less time compared to the gradient descent approach [113].

Although the above-described optimization strategies have been tested on unassisted walking or walking with an exoskeleton assistive device, it is possible that they will also be effective in optimizing prosthesis control. Assistance profiles would need to be redefined for prosthetic devices, however, since common torque profiles may no longer be suitable for individuals with amputation. Nonetheless, with some changes to the optimization strategy and device controllers, human-in-the-loop optimization strategies have the potential to advance the development of user-specific active prosthesis control.

## 23.4 CONCLUSIONS AND FUTURE DIRECTIONS

Significant progress has been made in the last several decades to advance active prosthetic devices, yet many engineering challenges still remain. For example, autonomous devices are often cumbersome, with restricted maximum torques and battery life. Further development of efficient, energy-dense power sources, as well as light and powerful actuators, is crucial for effective active prostheses. Current prosthetic controllers are also rarely individualized to each user, and struggle with effectively adapting to changes in user biomechanics or the environment. Developments in online control optimization and intent recognition are promising for addressing these limitations, especially given that persons with amputation may benefit from assistance different from the biological profiles of unimpaired gait. Still, full human–prosthesis integration is limited by the difficulty of closing the feedback loop between user and device. For example, introducing agonist–antagonist muscle interaction can improve controllability of a two degrees-of-freedom electrically powered ankle prosthesis [114]. During transtibial amputation surgery for several individuals, lateral gastrocnemius and tibialis anterior residual muscles were surgically connected. When one muscle contracted, the other elongated, thus providing sensory feedback to the nervous system. This setup, tested on one participant, showed promise in generating more natural and intuitive behaviors when compared to conventional devices. Furthermore, interfacing with the peripheral nerves in the residual limb through electrode implant is promising for controlling upper limb prostheses (e.g., [115]). Still, methods for administering feedback to the user from lower limb prosthetic devices are underdeveloped. This likely restricts user ability to take full advantage of an active prosthesis, making it challenging to determine optimal assistance approaches. As a result, proper feedback pathways, to both human and device, need to be explored in future research in order to improve our understanding of the needs of prosthesis users.

Another factor that significantly affects the usability and efficacy of prosthetic devices is the mechanical interface between the prosthesis and user. Most prostheses connect to the residual limb of the user through a socket, which can significantly affect the load transmission, stability, and control of the device [116]. However, proper socket fit is difficult to achieve and is one of the primary concerns of prosthesis users, many of whom frequently experience discomfort and chronic skin problems [49]. Osseointegration, or the direct attachment of a prosthetic device to the residual bone of the patient, is one way to circumvent the use of prosthetic sockets. Long-term studies show that individuals with an osseointegrated prosthesis demonstrate significant improvements in walking ability and overall quality of life [117]. However, this surgical approach is still uncommon and can cause complications, meaning that further development of comfortable and reliable prosthetic sockets is likely necessary.

With the number of people with amputations expected to quadruple in the US alone by the year 2050 due to increasing rates of obesity and diabetes [118], active prosthesis development is becoming increasingly prevalent in research and industry. Given the technological advances made in a relatively short time, active prosthetic devices hold great promise for restoring natural function to individuals with lower limb amputation in the near future.

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