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Review

Review of control algorithms for robotic ankle systems in lower-limb orthoses, prostheses, and exoskeletons☆

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A B S T R A C T

This review focuses on control strategies for robotic ankle systems in active and semiactive lower-limb orthoses, prostheses, and exoskeletons. Special attention is paid to algorithms for gait phase identification, adaptation to different walking conditions, and motion intention recognition. The relevant aspects of hardware configuration and hardware-level controllers are discussed as well. Control algorithms proposed for other actuated lower-limb joints (knee and/or hip), with potential applicability to the development of ankle devices, are also included.

Contents

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1. Introduction

The ankle–foot complex plays an important role in human locomotion. The stance phase of walking involves an intricate dynamical behavior of the lower limbs and its interaction with the floor. The ankle flexor and extensor muscles are crucial to provide vertical support and forward progression of the body [1]. The lack of the ankle functionality represents a very important limitation for walking and many other human activities.

For low-speed walking, the commercial passive orthoses and prostheses can mimic the behavior of a healthy ankle in a satisfactory way. However, for normal and fast walking speeds, the ankle provides additional energy for propulsion at the plantar flexion phase [2,3]. The lack of a source of energy in passive orthoses and prostheses is commonly associated to gait deficiencies and a higher metabolic energy consumption.

During the last five decades, researchers have been developing active and semiactive ankle devices to help impaired individuals to walk in a more natural way. Many important results have already been achieved. However, there are several factors that limit the general use and commercialization of these devices. Portable power supplies, lightweight actuators, and high-efficiency transmissions are some of the most important issues that still need improvement [4]. In the case of active orthoses, the specific nature of the ankle disabilities, that widely varies among patients, makes very difficult the development of general design methodologies. In particular, the development of general control strategies is challenging.

Active ankle devices can be classified in three categories: (1) prostheses and orthoses, (2) rehabilitation robots, and (3) powered exoskeletons. A thorough review on the past and current research on exoskeletons and active orthoses is presented in [4] and also in [5], where a general framework for the study and classification of these devices can be found. Versluys et al. [6] give an overview of the evolution of the prosthetic feet, going from completely passive models to active prototypes. Marchal-Crespo and Reinkensmeyer [7] present a review of supervisory strategies for rehabilitation robots, emphasizing the aspects that induce plasticity on the patients.

From the mechatronics perspective, exoskeletons, active orthoses/prostheses, and rehabilitation robots are very similar at the functional level. However, control objectives and human interfaces are essentially different for each device category.

Active orthoses and prostheses might involve user motion intent recognition and adaptation to different working conditions (e.g., different walking speeds and walking surfaces, in the case of a lower-limb device).

The main difference between prostheses and orthoses resides in the hardware configuration. Orthoses are worn on the existing but impaired limb. Thus the control system has to deal with a possible lack of coordination between the user and the orthosis (involuntary muscle reaction) and with the intrinsic limb dynamics. Besides, the hardware weight is added to the limb weight. One of the main problems still encountered in the design of active orthoses is the weight of the components (specially the actuator) which constitutes an important limitation for the development of autonomous and efficient devices.

Prostheses replace lost limbs and, in the case of lower limb devices, the total weight in some recent prototypes is close to that of the replaced limb.

The primary objective in control schemes for rehabilitation devices is the reproduction of motion sequences to facilitate the patient's recovery. Trajectory-tracking controllers are commonly employed in these systems to impose repetitive motion patterns (see, for example [8]).

The control objective of an exoskeleton is to follow the movements of a healthy user, augmenting his/her physical capabilities for specific tasks in a relatively safe and transparent way. Although orthoses and exoskeletons share many functionalities, the term orthosis usually designates an assistive device for a patient with a motor pathology [5]. The large variety of ankle robots that has been proposed to date is also accompanied by a large variety of control strategies. Nevertheless, some features are common to many control schemes as, for example, the use of finite-state machines for gait phase detection. At the hardware level, the mechanical components and sensors used in the prototypes usually confine the nature of the low-level controllers to particular configurations.

The generation of control signals for active prostheses and orthoses can be addressed using different sources of information. The main categories are: (1) biomechanical signals, (2) electromyographic (EMG) signals, (3) peripheral nervous system signals, and (4) central nervous system signals [9]. However, mechanical and, in a less extent, EMG interfaces predominate in the reviewed works.

The level and rate of human adaptation to the use of active devices are strongly related to the user interface and to the control method. Cain et al. [10] compared the level of user's adaptation to an active orthosis using a foot-switch "bang-bang" controller and a proportional EMG-based controller. According to their results, the nervous system can adapt more easily to the continuous response of the proportional control than to the discontinuous response of the foot-switch control.

The present work reviews recent control strategies proposed for active and semiactive ankle devices. Special attention is paid to the algorithms for motion intention recognition, adaptation to different walking conditions, gait phase identification, and generation of walking patterns. Control algorithms for knee and hip devices are also included to extend the panorama of ideas that can be applied to control ankle devices. When available, experimental and numerical results are mentioned to illustrate the behavior of the algorithms.

In the following sections, the term control system is used to denote the assemblage of control elements (hardware and software). The term control scheme refers to the set of control algorithms in all levels of the control hierarchy. The term controller is used to designate the physical implementation of a dedicated algorithm, usually in charge of a particular task. For instance, gait events are used as inputs to finite-state controllers that switch between different low-level controllers for the actuators.

The manuscript is arranged as follows: Section 2 gives a short description of human gait to introduce the basic elements of ankle biomechanics. Section 3 presents algorithms based on gait pattern generators. Algorithms that implement different control strategies according to the gait events are discussed in Section 4. More sophisticated algorithms that try to determine the user's motion intent are presented in Section 5. Section 6 provides a description of some algorithms whose general structure do not fit in the previous categories. Final remarks are given in Section 7. A summary of the key features of the control schemes and their hierarchical structure is presented in Table 1.

2. Human ankle biomechanics

For the discussion of the different control algorithms it is convenient to have in mind the main aspects of human locomotion. This brief review of the ankle biomechanics will be restricted to the case of ground-level walking, as most of the proposed algorithms are related to it.

The ankle motion during ground-level walking is quasi-periodic and is usually divided into two main phases: the stance phase and the swing phase. The ideal gait cycle is typically defined as starting with the heel strike of one foot and ending at the next heel strike of the same foot. The stance phase begins when the heel strikes the

Table 1
Summary of control algorithms.

Control scheme	Control modes	Coordination level (high-level control)		Hardware level (low-level control)	Measured signals
		Motion intent recognition	Gait control		
Control using gait patterns (ankle–foot orthoses/prostheses) [13–17,20,21]	GLW[a]	–	GPG[b]	Trajectory tracking control	MEC[c]
Impedance control for an ankle orthosis [22,23]	GLW	–	FSM[d]	Variable impedance (stance), PD position control (swing)	MEC
Impedance control for an orthosis with a robotic tendon [24,25]	GLW	–	FSM	Variable impedance	MEC
Impedance control for knee–ankle prostheses [27,28]	GLW	–	FSM	Variable impedance	MEC
Control for an ankle–foot prosthesis [29]	GLW with slope adaptation	–	FSM	Torque control (stance), position control (swing)	MEC
Control for semiactive orthoses [30,31]	GLW	-	FSM	Variable friction torque	MEC
Control algorithms for orthoses with artificial pneumatic muscles [32–35,37,10,38]	–	EMG signal conditioning	–	Proportional EMG control	EMG
Position control for an ankle–foot prosthesis [39]	–	EMG signals processing using a model/neural network	–	–	EMG
Gait pattern recognition for a knee–ankle prosthesis [40]	Standing; slow, normal and fast GLW	*k*-Nearest neighbor algorithm	–	–	MEC
Supervisory control for knee–ankle prostheses [41,27,42,43]	Standing, sitting, GLW	Gaussian mixture model and pattern voting scheme	FSMs	Variable impedance	MEC
Motion intent recognition in an ankle–foot prosthesis [44]	GLW, stair descent	EMG signals processing using a neural network	FSMs	Torque, impedance, and position control	EMG, MEC
Control algorithms for the HAL [45–49]	Standing up, standing, GLW	EMG signals processing	FSM	EMG torque generation, variable impedance	EMG, MEC
Model-based control schemes for a leg orthosis [50–53]	–	EMG signals conversion to forces/torques	–	Kinematic control/torque control at knee joint	EMG, MEC
High-sensitivity control for the BLEEX exoskeleton [54–57]	–	High-sensitivity control	FSM	Model-based torque control	MEC
Manual control/On-off control (ankle–foot prosthesis) [58,34]	GLW	–	–	Manual actuator positioning/two-state controller	MEC
Decomposition-based control for a knee–ankle prosthesis [9]	GLW	Prosthesis socket force controlled by the user	FSM	Torque control	MEC
Hybrid assistive systems [59–61]	–	–	SS[e]	Model-based controllers	–

[a] Ground-level walking.
[b] Gait pattern generators.
[c] Mechanical signals.
[d] Finite state machine.
[e] Stimulation sequences.

floor and ends at toe-off, when the same foot rises from the ground surface. The swing phase corresponds to the part of the cycle when the foot is off the ground.

The stance phase can be divided into three subphases: controlled plantar flexion, controlled dorsiflexion and powered plantar flexion [2]. Controlled plantar flexion begins at heel strike and ends at foot flat. Controlled dorsiflexion begins at foot flat and continues until dorsiflexion reaches a maximum point. Powered plantar flexion initiates after controlled plantar flexion and ends when the foot leaves the floor. In general, from moderate to fast walking, additional energy is provided besides the energy stored during the previous subphase. In the swing phase, the ankle position is controlled until it reaches an adequate angle for heel strike. The shape and duration of the gait cycle varies from step to step and it strongly depends on the gait speed, subject morphology, subject weight, and terrain conditions.

For the development of active orthoses it is also important to take into account the intrinsic dynamics of the ankle. The ankle–foot complex consists of different biological elements such as skin, muscles, tendons, ligaments, bones, cartilage, and connective tissue. The mechanical properties of these elements depend on several factors as the deformation rate, position, and motion speed [11]. Their behavior is inherently nonlinear. Moreover, the mechanical impedance of the ankle can be modified by the reaction of the muscles to the different electrical signals from the nervous system. This behavior also varies among different subjects and the wide variety of dysfunctional conditions that can affect the ankle motion [12].

3. Control based on gait-pattern generators

Several control schemes developed for active ankle devices rely on the fact that gait is essentially a periodic motion. Preprogrammed patterns, that may be adjusted as a function of the stride time and information about the current kinematic/kinetic state, have been proposed to mimic the ankle behavior. A simplified block diagram of a control system with this characteristics is shown in Fig. 1.

Oymagil et al. [13,14] used a trajectory tracking controller for an active orthosis actuated by a linear motor with a series elasticity element. This low-level controller has a proportional-derivative (PD) structure and uses an adjustable gait pattern as a reference for the motor position. The reference pattern is generated by a polynomial fit of a normal-gait pattern and is a function of the stride time [15]. The system determines the duration between two heel strikes and adjusts the polynomial function accordingly. Ward et al. [16] reported some performance improvements in terms of gait speed, maximum ankle moment, and peak power in a series of tests with three stroke survivors.

Hitt et al. [17] briefly describe the control scheme employed in an ankle prosthesis with regenerative kinetics (the SPARKy project [18,19], Fig. 2). The prosthesis has an electric motor with a series

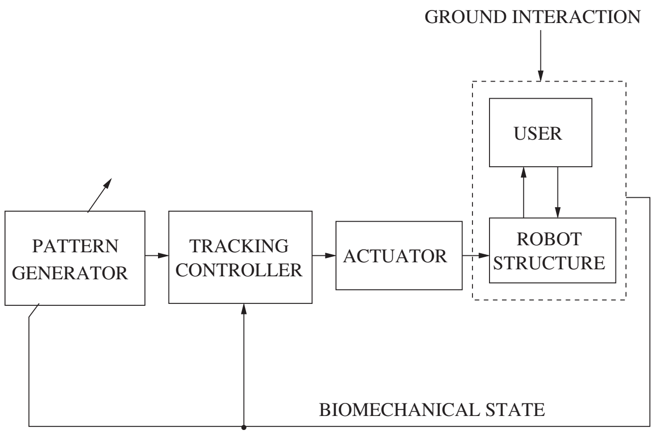

Fig. 1. Block diagram showing the main elements in a control system using a gait pattern generator.

stiffness and a keel that acts as a torsional spring. The prosthesis controller has a preprogrammed gait pattern that is used as a reference to control the actuator position (based on the same algorithm proposed in [15]). Detection of heel strike is used as a trigger to initiate the gait sequence.

Holgate et al. [20] describe two control methods for ankle robots. The first method employs the tibia angle and angular velocity to fit a mathematical function that relates the gait percent, the stride length, and the ankle angle (see also [21]). To cope with variations in walking speed, the second method adjusts the actuator motion profile in both, duration and amplitude. This method requires the Fourier coefficients of different actuator trajectories that correspond to different stride times. With these coefficients and the stride time of the last gait cycle, the current cycle is adjusted.

Despite some difficulties in the numerical estimation of the tibia velocity, in tests with an able-bodied subject the first method showed a smooth operation for different stride lengths and a fast response in updating the ankle angle, even if the user walked backward. The second method presents some drawbacks as the necessity of computing different actuator trajectories for persons with different weights and different walking conditions, and the one-gait-cycle delay required to estimate the current stride time. Nevertheless, this method has proven effective in tests with an able-bodied subject wearing an ankle foot orthosis.

4. Hierarchical control systems with gait-phase identification

Along the three stance subphases during ground-level walking, the ankle behavior can be described by a combination of passive elements (stiffness and damping) and power sources [2]. Different configurations of these elements have also been proposed to describe stair climbing [3]. During the swing phase, the ankle works as a position control system that prepares the foot for the next contact with the walk surface.

The biomechanical behavior of the ankle has inspired the design of many active devices. Some control schemes try to emulate this behavior by regulating the effective mechanical impedance of the device or by implementing virtual mechanical elements. Stiffness and damping parameters are adjusted as a function of the gait phase, the gait speed, and other criteria.

The structure of this kind of systems usually involves a mid-level algorithm to detect the current gait subphase and different low-level controllers for each gait subphase. In several cases, finite-state machines are used for switching between the different dedicated or low-level controllers. Biomechanical signals from the orthosis

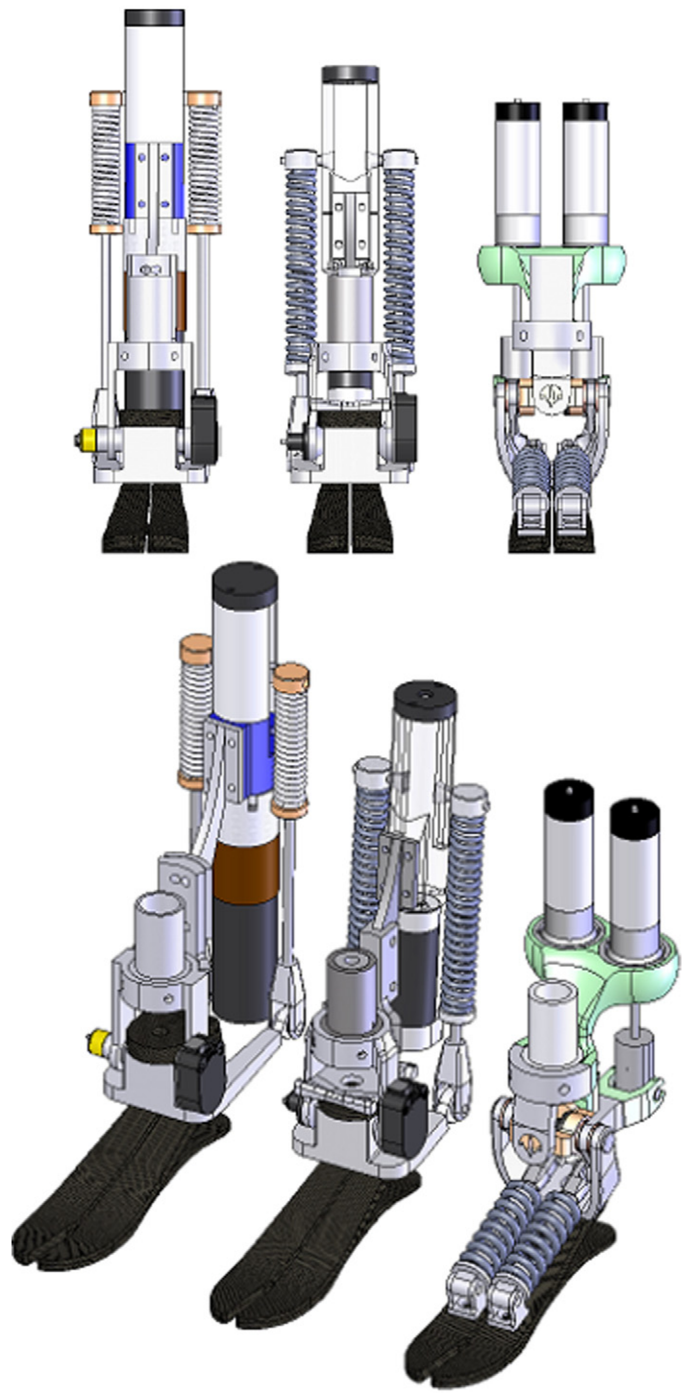

Fig. 2. Initial designs of SPARKy 1, 2, and 3.
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(and the user's body, as it will be seen in the following sections) are fed back to both control levels to estimate the gait phase (or other decision elements) and to perform specific control tasks. Fig. 3 presents a general block diagram for these systems.

4.1. Variable impedance control for an ankle orthosis

In [22,23] an active ankle–foot orthosis is presented for patients with drop foot (Fig. 4). A commercial polypropylene ankle orthosis is actuated by a series-elastic linear motor. A rotary potentiometer is used to determine the angle between the shank and the foot. Ground reaction forces (GRFs) are estimated using capacitive force

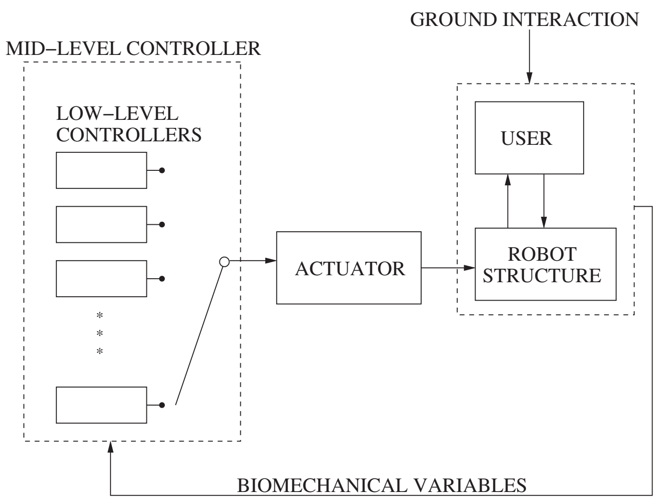

Fig. 3. General architecture for control systems with gait phase identification.

transducers on the bottom of the foot. An additional foot switch placed on the shoe heel is employed to detect heel strike events. The control algorithm is implemented on an analog input–output electronic board installed on a Linux computer.

A finite-state machine is proposed to regulate the mechanical impedance of the orthosis. The first state covers the period between heel strike and midstance. In this state the associated controller simulates a linear rotary spring at the ankle with variable stiffness. The control algorithm, based on the analysis of the GRF at the moment of forefoot impact, modifies the stiffness of the simulated rotary spring to prevent foot slap. The second state goes from midstance to toe-off. As drop foot does not affect the stance period, the actuator is set to zero force for not disturbing the gait of the user during this phase. The last state corresponds to the swing phase of walking. For this state, the ankle was modeled as a linear second-order under-damped mechanical system with a PD position controller. A pair of stiffness and damping parameters is selected from a look-up table, according to the user's gait speed.

In tests conducted on drop-foot patients, the system was able to reduce the occurrence of foot slap and to improve the swing phase when compared to the zero and constant-impedance control schemes. According to the authors, further improvements are required in both, the actuation system (too heavy and power consuming for practical purposes) and the control scheme (to include other patterns for different gait conditions).

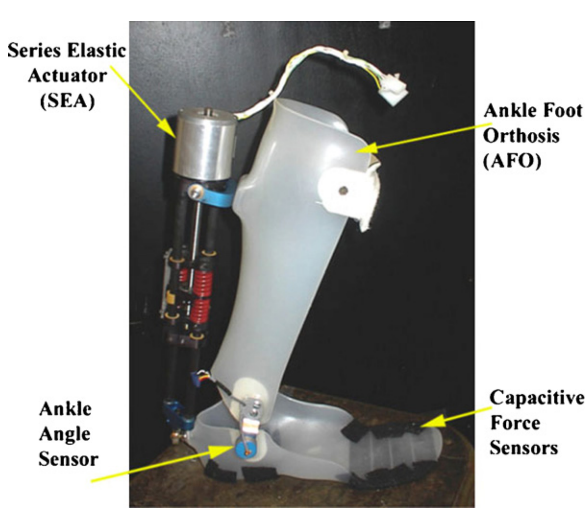

Fig. 4. Active ankle–foot orthosis.
Adapted from [23], © 2004 IEEE, with permission from IEEE.

4.2. Control algorithms for orthoses with robotic tendons

Hollander et al. [24,25] developed a control algorithm to emulate a variable stiffness behavior in an active ankle orthosis. The proposed algorithm is based on a dynamical model of a robotic tendon (a linear DC motor in series with a spring, see also [26]). At the supervisory level the control scheme has a finite-state machine with five fundamental states. Each state is associated to a different part of the stance phase in a typical ground-level gait cycle. The dedicated controllers can work on stiffness or velocity control.

The first state initiates at heel strike and its corresponding controller keeps the motor speed at a constant level, proportional to the speed of the previous swing phase. State two is activated when the ankle angular velocity crosses through zero (due to the interaction with the floor) or when the ankle angle reaches a predefined threshold. In this case the controller maintains a constant stiffness 1.35 times the actual spring stiffness. A similar behavior is assigned to the third state but with 3 times the stiffness of the actual spring. The third state starts at foot flat. The fourth state starts when the heel leaves the ground. During this state the controller maintains a constant speed equal to the speed achieved in the previous state. The last state starts when the peak plantar flexion occurs and ends when the swing phase begins. During this phase the motor remains in a constant position to allow the spring to release the energy accumulated in the previous phases. Additional states are included in the implementation of the algorithm to manage the initialization and the swing phase.

The authors report a kinematic behavior comparable to that of a healthy ankle in tests performed on an able-bodied subject. Along one gait cycle, the peak mechanical power developed by the motor in conjunction with the spring is around 1.89 times the peak mechanical power developed only by the motor, reflecting the energy storage/release contribution of the spring. Unfortunately, there is no evaluation on subjects with limb pathologies and no other performance indexes are provided.

4.3. Impedance control for knee–ankle prostheses

Sup et al. [27] introduced a finite-state machine for a tethered knee–ankle prosthesis with pneumatic actuators. The stiffness and damping parameters of the impedance controller can take different values according to four gait states: stance flexion/extension, preswing, swing flexion, and swing extension. Transition between states is determined in a cyclic fashion using thresholds for the ankle angle, the ankle torque, the knee velocity, and the axial load (associated to heel strike), respectively. With this approach, the prosthesis is guaranteed to stabilize around the equilibrium points associated to each of the gait states and to produce net work only when it switches between states.

A similar scheme is presented in [28] for a variant of the previous prosthesis equipped with electric actuators and an unidirectional series spring. In this system, the force is controlled through a load cell in series with each actuator. The prosthesis foot was specially designed to include strain gauges to measure the GRF on the ball of the foot and on the heel. The main difference in the control algorithm, with respect to the previous work, is that the transition from the preswing state to swing flexion is triggered by the load at the ball of the foot. The performance of both systems was evaluated in able-bodied tests on a treadmill. In both cases, the gait patterns generated with the prototypes were similar to those found in normal gait tests, except for some discrepancies in the knee trajectories reported in the first work.

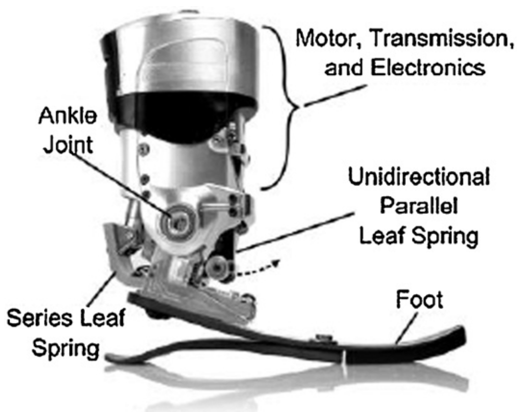

Fig. 5. Active ankle–foot prosthesis.
Adapted from [29], © 2010 IEEE, with permission from IEEE.

4.4. Control scheme for an ankle–foot prosthesis

Eilenberg et al. [29] presented a control system for an ankle–foot prosthesis with series and unidirectional parallel stiffness. The algorithm uses a nonlinear dynamic model of the foot–ankle complex to determine the torque that needs to be applied at the ankle joint. An angle sensor at the ankle provides the main control input to the algorithm. Strain gauges inside the prosthesis are also used to estimate the torque (Fig. 5).

The algorithm determines the required torque using a neuromuscular model that comprises two antagonistic virtual muscles: one unidirectional plantar flexor Hill-type muscle and one dorsiflexor muscle that can act as a bidirectional position regulator or as an unidirectional spring-damper system, depending on the gait phase. The algorithm relies on the estimation of the gait phase (stance and swing) using the available sensor information and a finite-state machine. The finite-state machine manages the participation of one or the other, or both virtual actuators. The net virtual torque is used as the ankle torque command for the prosthesis. The use of these virtual muscles to produce the torque command can be interpreted as a sort of mechanical impedance control employing a nonlinear model of the ankle.

In a clinical evaluation on a transtibial amputee, the torque and ankle angle profiles generated by the controller were qualitatively close to those developed by an able-bodied subject with the same weight and height. However, there are no measurements or quantitative indicators of the performance of the proposed device as to have a clear idea of its advantages. Nevertheless, an interesting feature observed in this evaluation is the ability of the system to adapt to different walking conditions. The prosthesis was able to increase the net work developed at the ankle when the user ascended an inclined platform, and to decrease it when descending it, a behavior similar to that observed in healthy subjects.

4.5. Control algorithms for semiactive orthoses

A semiactive ankle foot orthosis is presented in [30]. The orthosis includes commercial ankle joints; heel and forefoot switches, an angle sensor to detect the different gait phases; and a magneto-rheological rotary damper to provide a variable viscous damping torque at the ankle joint. The term semiactive refers to the fact that the device is able to manipulate the foot position not by injecting energy into the system but by modifying its energy dissipation properties. The angular position of the foot depends on the external forces acting on it and the friction force developed in the damper.

The walking cycle is divided into three different phases: (1) initial stance, from heel strike until the activation of the forefoot switch; (2) mid-stance, when both foot switches are activated; and (3) terminal stance and swing, when the shank angle reaches a predefined threshold. The control algorithm has a simple structure, for the initial stance phase it applies a moderated level of friction torque to allow smooth plantar flexion motion. For the mid-stance phase the algorithm uses the lowest level of friction torque for not disturbing the natural leg motion. For the last walking phase, foot clearance is achieved by holding the foot in a dorsiflexed position before the swing phase initiates. The maximum level of friction torque is used for this purpose.

The proposed orthosis and the control algorithm were evaluated on a hemiplegic subject. Heel contact was improved at the initial stance phase and enough foot clearance was achieved for the swing phase. Some of the advantages that can be found in this approach, when compared to active orthoses, are a simpler control algorithm, a relatively light weight and a lower power consumption.

Another semiactive orthosis is reported in [31]. The orthosis consists of a compact magneto-rheological rotary brake, an amplification linkage mechanism that connects the brake to the ankle joint, a rotary potentiometer to measure the ankle angle, an accelerometer, and an unidirectional parallel spring to assist during plantar flexion. In order to avoid typical problems associated to foot switches (premature failure due to excessive bending and rubbing on the sensor surface, inaccurate readings due to walking surface unevenness and footwear sole deformation) the authors proposed the use of an accelerometer to detect the initial contact of the foot with the ground. The total mass of the system is 990 g while the average of active ankle devices is around 2 kg.

The algorithm is intended for ground-level walking and uses a finite-state machine to regulate the friction torque in the brake. State One is defined from initial contact of the foot with the ground to foot flat. A threshold in the vertical acceleration is used to determine the initiation of this state. The control system applies a proportional-integral law to regulate the ankle angular velocity and to allow an adequate plantar flexion after heel contact. State Two goes from foot flat to heel off. The ankle angle is used to trigger this state when the rotational direction of the ankle changes from dorsiflexion to plantar flexion. A zero brake torque is assigned to State Two to allow a smooth rotation of the ankle. State Three utilizes the same control method as in State One but the angular velocity is set to zero to prevent drop foot. This state goes from heel off to the next foot contact with the ground. State Three is triggered when the ankle angle passes through a predetermined value and the rotational direction changes from dorsiflexion to plantar flexion. In experiments with a post-Guillain–Barré syndrome patient, the system successfully worked as a velocity regulator during initial stance and as an angle limiter during swing.

5. Control systems with motion intent recognition

Nowadays, the available computational power and the level of electronics miniaturization do not pose important problems in controlling assistive devices. The major challenges remain at the supervisory and coordination levels. One of the most difficult problems that still requires attention is the development of robust and efficient algorithms to determine the user's motion intention and to generate the correct trajectories with the wearable robots.

5.1. Control algorithms for orthoses with artificial pneumatic muscles

Ferris et al. [32–35] have been developing lightweight powered ankle orthoses using artificial pneumatic muscles. The orthoses are intended for basic locomotion studies or for rehabilitation purposes (see, for example [36]). Attaining a high level of portability is not

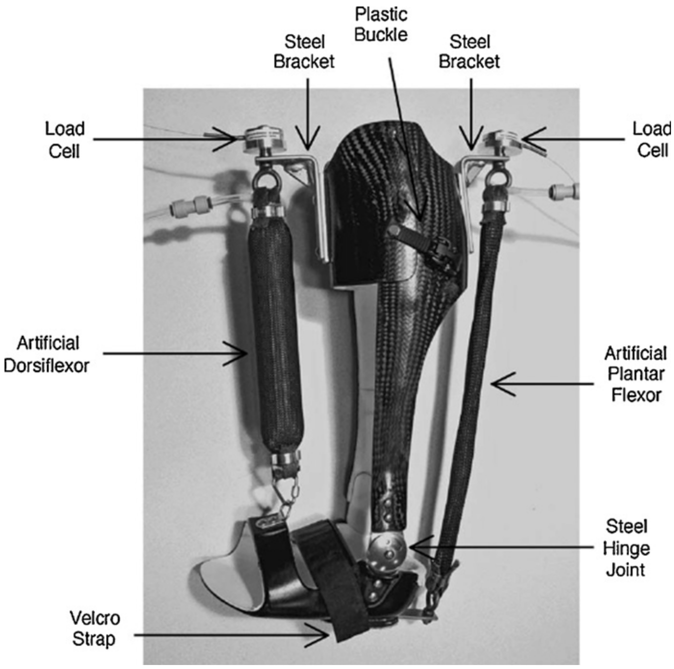

Fig. 6. Active ankle orthosis with artificial penumatic muscles. Reprinted from [33], © 2005 Elsevier, with permission from Elsevier.

a major issue in their designs. Nevertheless, pneumatic artificial actuators have a very good force-to-weight ratio that allows to keep the total weight of the orthoses around 1.5 kg.

Their orthoses consist, in general, of a custom-fit carbon-fiber shell, a hinge joint, and pneumatic muscles (Fig. 6). An active knee–ankle–foot orthosis using similar actuators is presented in [37].

The controller generates a command signal proportional to the muscle activation pattern. EMG signals from the soleus and/or tibialis anterior are processed in real time and converted to an analog signal to control the pressure in the artificial muscles. Fig. 7 shows a schematic representation of the fundamental structure of this system.

In [32] the authors used four walking conditions to evaluate one of the powered orthosis on a healthy subject: without orthosis, with the orthosis in passive mode, with only soleus EMG plantar flexor control, and with only tibialis anterior EMG dorsiflexor control. The results corresponding to the powered tests qualitatively resemble those of the passive test but important discrepancies can be observed at the ankle angle measured at the beginning (1 min of walking) of the tests. The results showed also an EMG signals modification during the assisted-plantar-flexion walking tests indicating a sort of adaptation of the user nervous system to the active device.

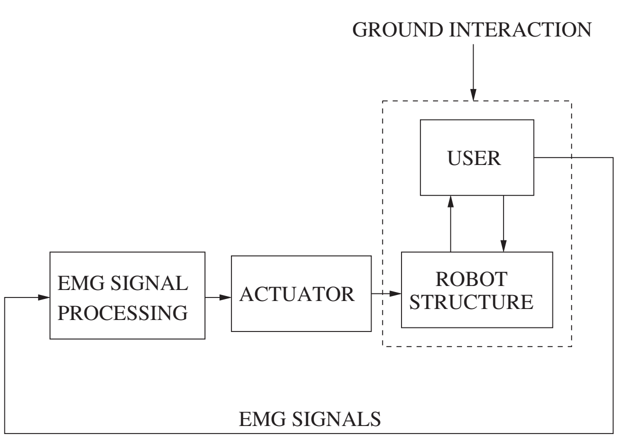

Fig. 7. Structure of the control system presented in [33].

A comparison of the performance of an on-off control algorithm (described in Section 6.1) and the proportional myoelectric control (using signals from the soleus) is presented in [10]. The authors tested the two control methods on healthy subjects walking on a treadmill. The results suggest that for the nervous system it might be easier to adapt to a control algorithm that provides a smoother dynamic response. For the nervous system, wearing the orthosis with the EMG control might represent a minor change in the relationship between the ankle movement and the biological control signals. In the case of the footswitch control, this relationship might be completely different. A lower metabolic cost of transport and a smaller learning effort might be expected from the orthosis user when using the EMG control system. Gordon and Ferris [38] also showed that, after some practice, users of a powered orthosis controlled by EMG signals learnt to reduce soleus recruitment by approximately 35% and to develop almost exclusively positive work at the ankle.

5.2. Position control for an ankle–foot prosthesis

Two control methods, one based on a muscle model and another based on a neural-network, are proposed in [39] for an active ankle–foot prosthesis. Both methods employ EMG signals to predict the user motion intent. The first method is based on a neuromuscular model of the ankle joint that describes motion in the sagittal plane. The second method uses a standard multi-layer feedforward neural network trained with a back-propagation algorithm. EMG signals from the gastrocnemius and soleus muscles are recorded for plantar flexion control, and from the tibialis anterior for dorsiflexion control.

In a series of off-line experiments it was found that both control methods were able to qualitatively reproduce the desired trajectories. The model-based control trajectory was smoother and more natural than that generated by the neural network. Authors suggest the implementation of real-time learning algorithms to compensate for variations in muscle response.

5.3. Gait pattern recognition for a knee–ankle prosthesis

A k-nearest-neighbor algorithm with four gait modes (standing and three different walking speeds) to estimate the user motion intent is presented in [40]. This algorithm utilizes the measurement of the force and moment produced by the interaction between the user and the prosthesis. The gait intent is stated as a pattern recognition problem where the input is a combination of the socket interface forces and the dynamic state of the prosthesis. At each time step, a k-nearest-neighbor recognizer classifies the input into one of the four gait modes according to the smallest distance with respect to a training data set. The algorithm is susceptible to chattering and presents a delay in the classification process (0.17 s). Nevertheless, in off-line experiments using an experimental database it was able to adequately differentiate standing, slow walking, normal walking, and fast walking.

5.4. Supervisory control for knee–ankle prostheses

A high-level control algorithm is proposed by Varol et al. [41] to recognize the user motion intent. The control scheme consists of a supervisory intent recognizer and a set of mid-level (or gait) controllers based on the algorithm presented in [27] (cf. Section 4.3). The intent recognizer switches between the mid-level controllers to accommodate different gait modes (in this preliminary work the gait modes were standing and walking). A Gaussian mixture model

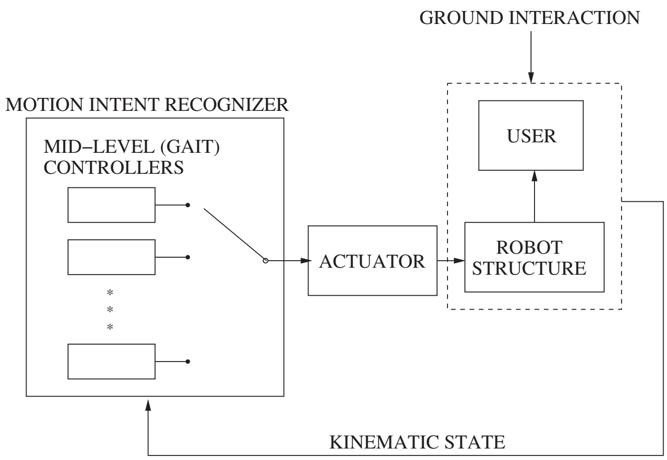

Fig. 8. Control architecture for the knee–ankle prosthesis presented in [41].

and a voting scheme are used to process the information coming from the prosthesis sensors. A comparison of the sensor information against walking patterns in a database is used to determine when the user is switching between gait modes, and to choose the appropriate gait controller. Fig. 8 shows a schematic representation of this control system.

An extension of the latter work can be found in [42] where the authors present also a three-level control architecture. An intent recognizer at the supervisory level infers the user motion intent, as described above, and switches between the different mid-level controllers. Walking, standing, sitting, and stair ascent/descent constitute possible gait modes, each of which would have a corresponding mid-level controller. The mid-level controllers are based on finite-state machines that modulate the joint impedance according to the subphase of the current gait mode. In this work, the authors present finite-state machines to manage a sitting mode and the transition between sitting and standing modes. The low-level controllers regulate the joint torque in a closed-loop fashion, using the torque reference signal generated by the mid-level controllers. The algorithm was tested on an unilateral amputee using the prosthesis shown in Fig. 9. The real-time implementation and tests of this scheme are presented in [43].

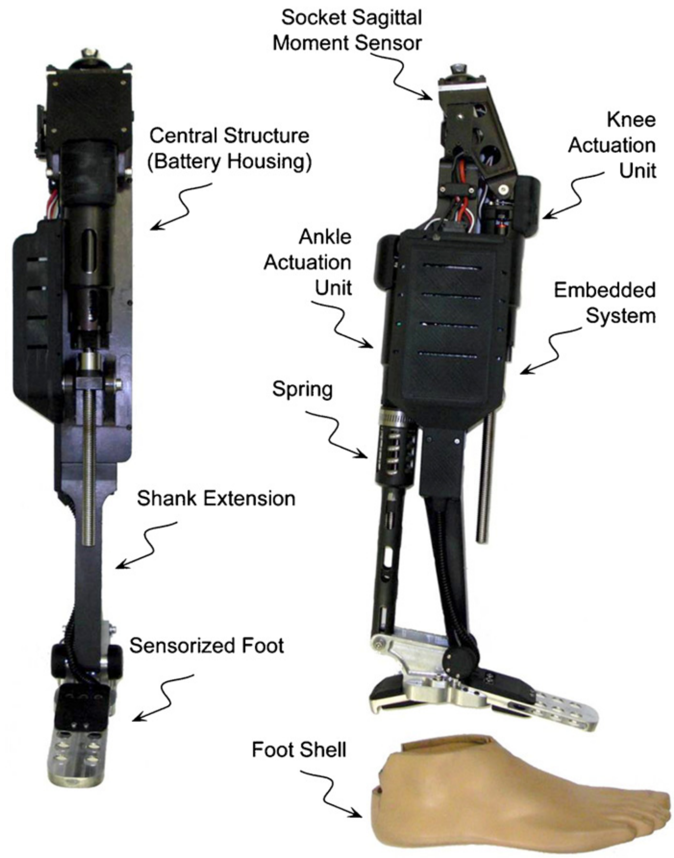

Fig. 9. Active knee–ankle prosthesis.
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5.5. Control algorithms for level-ground and stair-descent gaits

An EMG control with finite states is proposed in [44] for a powered ankle–foot prosthesis. The prosthesis has a similar configuration to that described in Section 4.4 and also includes an unidirectional parallel stiffness and a series-elasticity actuator (Fig. 10). The force developed by the actuator is controlled by measuring the deflection of the series spring. The control system comprises three low-level controllers: a torque controller, a mechanical impedance controller, and a position controller. Two separated finite-state machines are implemented, one for level-ground walking and another for stair descent. Ankle angle, torque, and foot contact pressure are monitored to determine the gait phase and to manage the transitions between states.

EMG signals from the residual limb muscles (gastrocnemius and tibialis anterior) are used to switch from one finite-state machine to the other. To change from level-ground walking to stair descent, the user needs to contract the gastrocnemius muscle during the swing phase. To switch back to level-ground walking, the user is required to contract the tibialis anterior muscle during the swing phase of stair descent. The raw EMG signals are filtered, amplified, and digitized. The processed EMG signals are fed into a neural network trained to classify the user's motor intent according to three possible discrete states: plantar flexed, relaxed and dorsiflexed (in this work the authors only used the first two states, for stair descent and level-ground walking, respectively). A simplified representation of this control scheme is shown in Fig. 11.

In a clinical study with a bilateral transtibial amputee the prosthesis was able to deliver net positive work and high mechanical

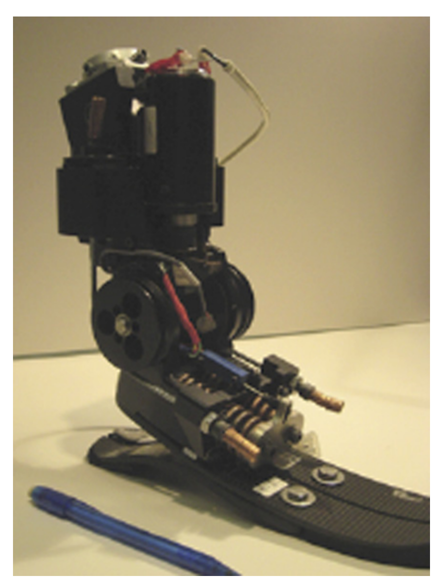

Fig. 10. Active ankle–foot prosthesis for testing level-ground and stair-descent gait controllers.
Reprinted from [44], © 2008 Elsevier, with permission from Elsevier.

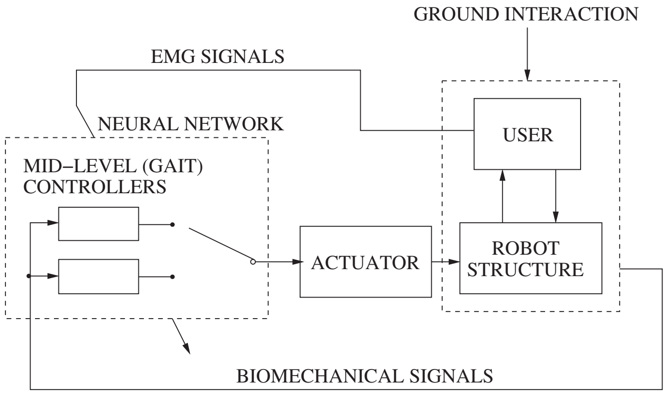

Fig. 11. Control architecture for the orthosis presented in [44].

power during stance. However, marked amplitude discrepancies where found at the end of the stance phase when compared to normal power profiles. On the other hand, the system was able to predict the user motor intent in a successful manner. Also, for the stair-descent condition, the prosthesis dissipated energy in a similar way as that of a healthy ankle.

5.6. Control algorithms for the Hybrid Assistive Leg

The Hybrid Assistive Leg (HAL) is a multipurpose exoskeleton that has been under development at the University of Tsukuba, Japan. Full-body, one- and two-leg versions have been proposed for a wide range of applications, going from rehabilitation and health care to human capacity augmentation in heavy-duty work [45].

The lower-limb versions consist of an articulated frame with single-degree-of-freedom joints at the hip, knee, and ankle. Only the hip and knee joints are actuated while the ankle remains passive (see, for example, [46] for a detailed description of the hardware). Kawamoto and Sankai [47] utilize a proportional EMG control algorithm for the HAL-3, one of the lower-limb versions of the system. The joint torque is estimated through the sum of a virtual torque and a variable mechanical impedance term. The virtual torque is the additive contribution of the corresponding extensor and flexor muscles of the joint expressed by a linear combination of the magnitudes of the EMG signals (pre-processed by an analog circuit). The mass, stiffness, and damping coefficients of the variable impedance term are estimated through the kinematic state of the user and the intrinsic mechanical impedance of the joint [48].

Kawamoto et al. [46,49] describe some of the methods developed to recognize the user motion intent using the HAL. At the supervisory level, a finite-state algorithm is proposed for detecting sitting, standing up, standing, and walking. Walking is divided into the support phase and the swing phase. The algorithm uses thresholds in the GRF detected in the rear and front parts of both feet to manage the transition between phases. In each walking phase the control system tries to emulate the behavior of a healthy leg, considering the typical direction of motion (flexion or extension) and the level of muscle activation (reflected by the EMG signals). Standing up from a chair is divided into four phases: sitting, upper body bent forward, upper body lifted (associated to the maximum positions of hip angles), and standing position. GRF at the rear part of the feet and the hip and knee angles are used to determine the transition between the phases in this particular motion. The GRF at the supporting leg is used to detect start and stop walking intent.

5.7. Model-based control schemes for a leg orthosis

An interesting approach to control an active thigh–shank–foot orthosis is presented in [50]. The orthosis is actuated at the knee

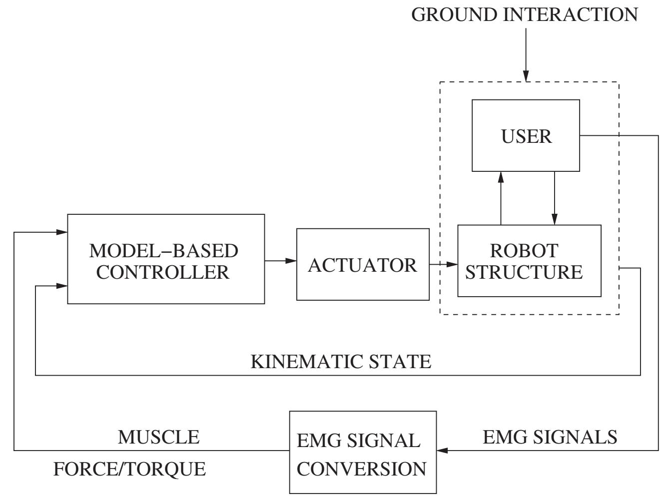

Fig. 12. Control system for the orthosis presented in [52].

joint and uses EMG sensors to acquire signals from the semimembranosus and the vastus medialis that directly contribute to the knee flexion–extension. The kinematic state of the user is calculated using angle sensors at the ankle and the knee joints and accelerometers at the thigh and trunk. Pressure sensors are used to detect the contact of the feet with the ground. Both of the user legs are equipped with sensors.

The sensor information is used as the input to a biomechanical model that consists of two legs with feet, shanks, thighs and torso (also described in [51]). The EMG signals are converted to muscle forces and serve also as inputs to the biomechanical model. With this information, the model is used to calculate the knee torque and the corresponding knee acceleration. The knee acceleration is interpreted as the input reference for the actuator controller (and, in a certain way, as the user intent). Fig. 12 shows a simplified block diagram with the essential elements of this control system.

The process of conversion of EMG signals into muscle forces is managed by an optimization algorithm that updates in real time some critical parameters associated to the knee torque calculation. The ability of the control algorithm to reproduce the torque curves was evaluated with healthy subjects in free-motion tests (without the actuator). The EMG signals were applied to the body model and the predicted torque was similar in shape and magnitude to the torque calculated by inverse dynamics. However, no measurement of the error between the curves or other performance indicators are provided. In stair-climbing tests the scheme has some problems to evaluate the knee torque during the double support phase when both feet are in contact with the stair.

A similar control scheme based on the same algorithm to convert EMG signals into torques is proposed in [52]. The main difference with respect to the previous work is the use of a torque control loop instead of a body model. The control loop takes as input the torque estimated by the conversion process of the EMG signals. Then, it implements a proportional controller to generate the knee torque command based on the difference between the current torque, estimated using the sensor information, and the torque derived from the EMG signals. A review of the work done by these authors can be found in [53].

5.8. High-sensitivity control for the BLEEX exoskeleton

The Berkeley Lower Extremity Exoskeleton (BLEEX) is an energetically autonomous underactuated exoskeleton aimed to augment human load-carrying capabilities during locomotion [54]. The mechanical system has a total of fourteen degrees of freedom of which only four per leg are actuated (flexion–extension and abduction–adduction at the hip joint, flexion–extension at the knee joint, and flexion–extension at the ankle joint).

The main objective of the control algorithm is to allow the system to move in concert with the user with a minimal interaction force between the two. Also, to facilitate implementation and reliability during field operation, the control scheme is based only on measurements taken from the exoskeleton itself. An interesting feature of the exoskeleton is that no external inputs are required (as EMG signals or force/torque information at the interface with the user). To accomplish this objective, a high-sensitivity control algorithm has been proposed and successfully tested [55,56].

The control scheme has two feedback loops: one negative feedback loop that represents the influence of the user over the exoskeleton and has a stabilizing effect on the system; and one positive feedback loop with a high sensitivity to the forces acting on the exoskeleton. The high-sensitivity loop allows it to quickly follow the voluntary and involuntary movements of the user. In the overall system architecture, the wearer has the role of guiding the exoskeleton but also has to undertake its stability. Thus he or she needs to develop the ability of moving quickly and create a stable condition for him- herself and the exoskeleton [55].

Using information from force sensors in the feet of the exoskeleton, the control algorithm can distinguish between three different situations during ground-level locomotion: (1) single support, with only one leg in contact with the ground; (2) double support, with both feet flat on the ground; and (3) double support with redundancy, with both legs in contact with the ground but only one flat. For each of these situations, the control system uses different dynamic models of the exoskeleton to produce the desired control signals. The main disadvantage of this methodology is that together with a high force sensitivity also comes a high sensitivity to parameter variations. As a consequence, the algorithm needs an accurate dynamic model of the exoskeleton. Ghan et al. [57] describe a procedure to experimentally identify the dynamic parameters of each of the models used by the control system.

6. Other control schemes

6.1. Manual and on-off controllers

Sawicki et al. [58] implemented hand-held pushbuttons to control an orthosis with pneumatic artificial muscles. Air pressure control inside the artificial muscles is based on a signal proportional to the plunger displacement. When the plunger is fully depressed the system commands the maximum air pressure to the artificial muscles and when it is not depressed at all no air pressure is supplied. The main disadvantage of the control scheme is that it requires a high level of external supervision from the therapist or the user itself.

Gordon et al. [34] tested a simple on-off controller in a similar powered orthosis using a footswitch on the forefoot. When the forefoot was in contact with the ground (the signal from the footswitch was above certain threshold), the controller supplied the maximum air pressure to the artificial muscle. When the forefoot was not on the ground, zero pressure was applied. Marked differences on the ankle angle profiles were found in experiments using the orthosis in passive mode (no action from the artificial muscle) and in active mode (using one or two artificial muscles). Nevertheless, the orthosis was able to produce around 70% of the positive plantar flexor torque developed during normal walk.

6.2. Decomposition-based control for a knee–ankle prosthesis

An active-passive torque decomposition method for controlling an active knee–ankle prosthesis with monopropellant actuators is presented in [9]. The total torque at each joint is split into a passive and an active component. The passive component corresponds to the part of the joint torque that can be represented by linear stiffness and damping elements. This component is assumed to be a passive mapping of the angular displacement and angular velocity. The active torque is represented as a function of the force that acts on the prosthesis socket and is assumed to be the user's input. The closed loop of the control algorithm is designed to exploit the passivity properties of the passive torque component and the leg dynamics in order to guarantee the stability of the system.

The phase portrait (angle and angular velocity at each joint) is segmented in order to preserve the passive behavior. To find the mechanical parameters of the passive part, each segment of the phase-plane trajectories is converted to an optimization problem. The active component of the torque is estimated through force measurements at the prosthesis socket. A finite-state machine is used to reconstruct the control torque signal as the sum of the passive and active components. The effectiveness of the approach to reconstruct the joint torque is demonstrated using experimental data corresponding to the knee joint at different walking speeds.

6.3. Hybrid assistive systems

Under certain circumstances, paralyzed muscles that remain innervated can be stimulated by an electric current applied to the neuromuscular structures. The resulting muscle contraction can be used to perform limited ambulation with the assistance of the upper extremities and supporting structures or crutches. This strategy of external muscle activation is named Functional Electrical Stimulation (FES). The use of FES in combination with mechanical orthoses, called hybrid assistive systems, has been proposed for stabilization of the lower limbs using the biological muscles as actuators. In terms of weight reduction the proposal seems convenient. However, FES usually implies a high energy cost to develop walking. In addition, muscle fatigue, spasticity, joint contractures, and bone fragility can limit its effectiveness. A brief survey on FES systems and active prostheses can be found in [59]. In [60] a hierarchical three-level structure for FES systems is presented. The top level of the hierarchy corresponds to the decision level and is completely controlled by the user. The user acts as a supervisory controller and triggers different preprogrammed sequences of stimulation. The middle level is the coordination level and is implemented with a set of situation-action rules that deal with different gait conditions. The bottom level of the hierarchy executes the commands of the coordination level and interacts with the muscles with customized, possibly model-based, controllers (see, for example, [61]).

7. Discussion

A review of the most relevant control algorithms for ankle devices was presented. Special consideration was given to the algorithms with a supervisory role as those dedicated to human motion intention recognition and adaptation to different gait conditions. Table 1 presents a summary with the key features of the reviewed algorithms, their hierarchical structure, and the nature of the measured signals used in their implementation.

The human ankle is able to vary its mechanical behavior to adapt to a wide spectrum of walking conditions and it plays an important role in the powered phases of locomotion. A common denominator of most of the algorithms suggested to date is the use of finite-state machines, exploiting the fact that gait is predominantly cyclic and that the gait cycles can be divided into different subphases for which a particular mechanical behavior of the ankle can be distinguished.

At this moment, the main problems are the correct identification of the transitions between phases and between different gait modes (e.g., ground-level walking, stair ascent/descent motion, varying gait pace, etc.), and the adaptation to the different situations that can be encountered in daily activities. The kinetic and kinematic information available at the device and its interface with both, the user and the environment, has been useful to detect these transitions in many of the schemes studied in this work. Foot switches and angle sensors dominate the instrumentation for this purpose. However, acceleration-based schemes can also bring new perspectives to surpass the limitations of force detectors as, in principle, they do not require direct contact with moving surfaces [62].

An important advantage of the EMG control is that the user can regulate the device in a more natural way using his/her own muscles. However, EMG signals can be affected by many factors and it may be difficult to find a correlation between the EMG signals and the output of the actuators. Also, the availability and quality of EMG signals can vary from patient to patient. Fragility and installation requirements of electrodes can also be restrictive outside the laboratory.

In the case of assistive and rehabilitation devices, the marked variation of the character of disabilities from patient to patient makes very difficult the development of general control methodologies. Control schemes are customized to the pathology characteristics, the available biological signals, morphology of the remaining limbs, and the mechanical configuration of the device. In this respect, the development of parametric and nonparametric adaptive schemes can offer new alternatives to solve problems that may be induced by the variation of the ankle mechanical behavior. Another approach that can be explored is the use of energy-based control techniques to design controllers that mimic the energy exchange functions performed by a healthy ankle.

In many of the reviewed works there is a strong lack of quantitative indicators about the performance of the proposed algorithms. Measurements of the error between the expected and the real system trajectories as well as measurements of the user energy consumption [4], walking endurance, or any other indication of a real improvement or advantage of the new schemes are, in most cases, missing. Tests under realistic conditions, using patients with limb pathologies, is also an important issue that needs more attention.

In general, it is difficult to assess the dynamic behavior of the proposed systems or to make comparisons among them. The information provided in most cases is not enough to determine what are the real advantages or disadvantages of a given control scheme with respect to other proposals. In the case of active prostheses/orthoses, performance comparisons against other similar algorithms or against the performance obtained by commercial passive devices worn by the same subject could provide a hint about the convenience of a proposed control system.

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Conflict of interest

There are no conflicts of interest to report.

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