Lower Limb Wearable Robots for Assistance and Rehabilitation: A State of the Art

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***Abstract*—Neurologic injuries, such as stroke, spinal cord injuries, and weaknesses of skeletal muscles with elderly people, may considerably limit the ability of this population to achieve the main daily living activities. Recently, there has been an increasing interest in the development of wearable devices, the so-called exoskeletons, to assist elderly as well as patients with limb pathologies, for movement assistance and rehabilitation. In this paper, we review and discuss the state of the art of the lower limb exoskeletons that are mainly used for physical movement assistance and rehabilitation. An overview of the commonly used actuation systems is presented. According to different case studies, a classification and comparison between different types of actuators is conducted, such as hydraulic actuators, electrical motors, series elastic actuators, and artificial pneumatic muscles. Additionally, the mainly used control strategies in lower limb exoskeletons are classified and reviewed, based on three types of human–robot interfaces: the signals collected from the human body, the interaction forces between the exoskeleton and the wearer, and the signals collected from exoskeletons. Furthermore, the performances of several typical lower limb exoskeletons are discussed, and some assessment methods and performance criteria are reviewed. Finally, a discussion of the major advances that have been made, some research directions, and future challenges are presented.**

***Index Terms*—Actuation design, assessment methods, control strategies, lower limb exoskeletons.**

I. Introduction

Along with the worldwide population aging, there is an increasing interest in the use of robotic devices to assist the elderly people to accomplish their main daily living activities as well as rehabilitation exercise training. A survey from the United Nations shows that people aged 60 and over represent almost 11.5% of the global population in 2012, and this percentage is projected to nearly double to 22% by 2050 [1]. Moreover, in some regions and countries, the problem of aging population is particularly prominent. For example, in Japan, the population aged 60 and over had reached more than 30% by 2012, and 34% of the European population will be aged 60 and over by 2050 [1]. Some common and frequently encountered neurologic injuries such as stroke, spinal cord injuries, and weaknesses of the skeletal muscles seriously limit the ability of elderly and patients to achieve their daily living activities [2]. Current research challenges concern chiefly the development of new therapeutic methods and assistance modes that help the elderly/patients to improve their daily living activity performances and to restore lost or impaired motion control. Recently, robotics has shown a natural superiority in helping the elderly and the handicapped people to ensure rehabilitation training programs and physical movement assistance [3]. Indeed, they can cover many limits of the traditional therapeutic devices and physical capabilities of the therapists, such as the great labor intensity and bad repeatability [4].

In particular, wearable robots have been a focus of the robot researches in the last decade. A wearable robot is usually defined as a mechanical device that is designed around the shape and the function of the human body and can be worn by the operator, with segments and joints corresponding to those of the person it is externally coupled with [5]. Wearable robots can effectively integrate the cognitive ability of human being and the advantage of robotic devices to assist the users to accomplish their desired activities. Initially conceived for mainly military and therapeutic applications [6], [7], nowadays, the use of wearable robots is rapidly increasing toward assistive purposes, where the robot is designed to promote the functional activities at home, community, and society [8], [9].

In this paper, the state of the art of the lower limb wearable robots that are mainly used for physical movement assistance and rehabilitation is reviewed. Exoskeletons and active orthoses are two typical examples of wearable robots. Exoskeleton generally refers to a mechanical device that is used to assist the able-bodied wearer by augmenting his/her strength and endurance, etc. [6]. Active orthoses are exactly what their name indicates, i.e., orthoses equipped with powered actuation, and usually used to assist the patients to modify or recover the motor function of the neuromuscular and skeletal system. However, the distinction between exoskeletons and active orthoses is not clear-cut. In the following, we mainly focus on these two types of wearable robots. The term exoskeleton or active orthoses will be used as it is referred in the associated reference.

In order to develop effective, portable, and safe robotic exoskeletons, three main aspects usually need to be taken into account: the actuator design, the control strategy, and the assessment methods for performance evaluation [8]. Issues related to these three aspects are thus reviewed and discussed in this paper, with respect to stand-alone and portable lower

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limb exoskeletons/active orthoses operating in parallel with human lower limbs. For each aspect, we classify the associated exoskeletons/active orthoses and provide some typical examples. The characteristics, the advantages/disadvantages, and the evaluation of the effectiveness of each device, which are described in related literatures, are provided and discussed. Furthermore, different types of actuator designs and/or control strategies are compared and discussed upon their potential applications and characteristics.

The design and the choice of the actuation modes usually have the crucial significance for the wearable robots, since they generally determine the performances of these robots, such as the output force/torque, efficiency, and portability. Hence, the review of the active actuators used in exoskeletons and active orthoses is a main part of this paper. Meanwhile, although exoskeletons and orthoses with passive joints are also important types for movement assistance and rehabilitation purposes, such as MIT exoskeleton [10], they are not taken into account in this review. Moreover, exoskeletons that are not portable and stand-alone are not included either, e.g., the treadmill-based rehabilitation devices [12]–[14]. The review of the different control strategies is conducted according to the human–robot interface that is used. The estimation of the human motion intention is a key component for the development of efficient control strategies for exoskeletons and active orthoses that are used for movement assistance and rehabilitation. Wearable robots using control strategies without a human–robot interface [10] or using the functional electrical stimulation [15]–[18] are not taken into account in this study. Furthermore, only the "high-level" control algorithms are included. By definition, "high-level" control algorithms are those which infer the human intention and determine the assistance strategies rather than control the actuators to achieve a desired position or force, etc. (the "low-level").

In terms of assessment methods, although a great progress has been made in the study of wearable robots during the last decade, only a few published literature provided the kinesiology performance assessment of the used exoskeletons/active orthoses. Hence, we review some major physiological assessment methods that have been chiefly used and discuss the related effectiveness of the robotic devices.

The rest of this paper is organized as follows. Section II presents a review of the main actuation designs of the lower limb exoskeletons/active orthoses according to the actuation mode that is used. Section III presents chiefly the main control strategies used for controlling the wearable robots according to the human–robot interaction schemes. Section IV addresses the main assessment methods used in the literature to evaluate the performance of the exoskeleton/orthoses. Section V discusses current research, future directions, and challenges. Section VI concludes this paper.

II. Actuation Design of Lower Limb Exoskeletons

In this section, the development of the actuation design for lower limb exoskeletons is reviewed and classified according to the actuation mode used in each device. Other issues related to power supply and portability are also discussed and compared.

Since the early 1960s, a variety of exoskeletons has been designed, and many associated researches have been conducted in this field [6], [19]. The first walking active exoskeleton was developed by Prof. M. Vukobratovic and his team at Mihailo Pupin Institute, Belgrade, Serbia, in 1969 [19]. This device was designed to produce near-anthropomorphic gait and was pneumatically actuated. Almost at the same time, General Electric Research [20], [21] and some researchers of Cornell University, Ithaca, NY, USA, developed an enormous full-body powered exoskeleton prototype (about 680 kg) actuated using a hydraulic unit. Later, in 1974, the first known example of active exoskeleton actuated using electrical motors was designed [22]. Considering some drawbacks of the nonbackdrivable actuators for robots (e.g., low power density), researchers started, since the 1990s, to develop series elastic actuators (SEAs) [23], [24]. Due to their special characteristics, such as high fidelity, extremely low impedance, low friction, etc., SEA-based actuators were then used as an important actuation mode for wearable robots. Meanwhile, many researchers considered using the pneumatic muscle actuators to control the lower limb exoskeletons or orthoses as well.

Recently, novel types of actuations are still constantly being developed. For example, Haines *et al.* [25] proposed a novel inexpensive high-strength artificial muscle from fishing lines and sewing thread. The output force changes in response to temperature, and the performance of the artificial muscle even matches or exceeds that of the skeletal muscle in certain aspects, such as nonhysteretic, long life, and output power. However, the main actuation modes that have been used recently in wearable devices are shown as follows: hydraulic actuators, electrical motors, SEA, and pneumatic muscle actuators. In general, during the development process of a given exoskeleton or active orthosis, one of the aforementioned modes of actuation is chosen according to the intended application and/or the potential users. Meanwhile, the other parts of the exoskeleton mechanisms are designed around the actuator, and then, the associated sensors and power supply are developed based on the actuation mode as well. Hence, one classifies the lower limb exoskeletons upon the intended actuation mode.

A. Lower Limb Exoskeletons Using Hydraulic or Pneumatic Actuators

Because of the specific characteristics of hydraulic or pneumatic actuators, high ratio of actuator power to actuator weight, they are usually considered as important choices for the exoskeletons that are designed for augmenting the human performance significantly. The Berkeley's lower extremity exoskeleton [BLEEX; Fig. 1 (left)] and the Sarcos exoskeleton [Fig. 1 (right)] are two typical examples of this type of exoskeletons, which were both sponsored by the Defense Advanced Research Projects Agency. The goal of this program is to "increase the capabilities of ground soldiers beyond that of a human" [26]. For BLEEX, it also has other potential applications which include helping disaster relief workers and wildfire fighters and helping other emergency personnel to carry major loads without the strain typically associated with demanding labor [27]–[29]. For these purposes, the BLEEX

Fig. 1. (Left) BLEEX exoskeleton (IEEE image credit: [28]). (Right) Sarcos exoskeleton (XOS 2; IEEE image credit: [30]).

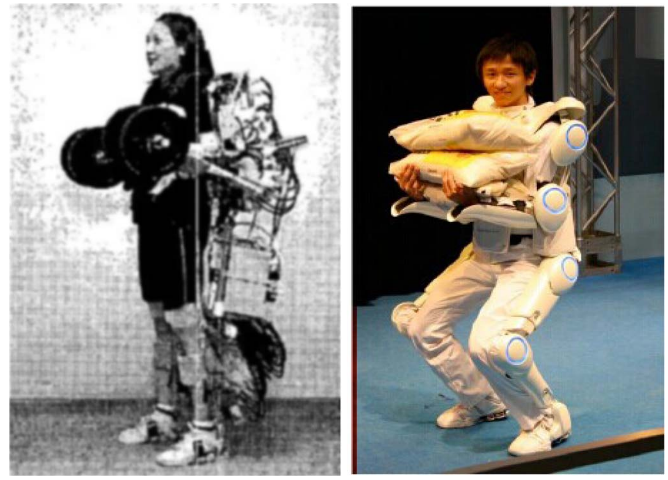

Fig. 2. (Left) Nurse-assisting exoskeleton (IEEE image credit: [91]). (Right) HAL-5 exoskeleton (IEEE image credit: [92]).

and Sarcos exoskeletons are used either to support their weight or to provide ability to carry extra loads.

There are seven degrees of freedom (DOF) in BLEEX, which coincide with the associated DOF of the human lower limbs. Four DOF (flexion/extension at hip, knee, and ankle as well as abduction/adduction at the hip) that require relatively significant torques are actuated by hydraulic actuators. The required torques to actuate these joints are computed based on the measurement of typical human walking gait cycle data (subject: 75-kg person) and clinical gait analysis (CGA). Due to their compact size, low weight, and high-force capabilities, linear hydraulic actuators are chosen in BLEEX. Unlike the BLEEX exoskeleton, the Sarcos second-generation exoskeleton (XOS 2) has been developed into a full-body exoskeleton as shown in Fig. 1 (right). A significant difference between these two exoskeletons is that the Sarcos exoskeleton uses rotary hydraulic actuators that are located directly at the powered joint level of the exoskeleton [26], [30], [31].

In terms of performances, BLEEX has been reported to support up to 75 kg of load (exoskeleton weight plus payload), with a walking velocity of 0.9 m/s and a speed velocity of up to 1.3 m/s without load [27]. XOS 2 has been demonstrated to help a person hold a 68-kg payload without feeling the load [30]. These performances clearly show that hydraulic actuators can supply "high specific power (i.e., high ratio of actuator power to actuator weight)" [11], [28]. Among these two types, the rotary ones have bigger internal leakage or considerable friction, compared to linear hydraulic-based actuators [27].

Regarding the power supply, a hydraulic–electrical power unit (HEPU) was developed for the BLEEX exoskeleton [32]. It can satisfy both the requirements of the hydraulic and electrical power (2.3 kW at 6.9 MPa and 220 W at 15 V dc, respectively) and has, at the same time, a relative small noise output (87 dBA). However, the HEPU's total net mass is about 30 kg, which is significantly over the requirements (23 kg) of the BLEEX. Regarding XOS 2, it uses a tethered power source, as shown in Fig. 1 (right).

The nurse-assisting exoskeleton is another exoskeleton prototype that uses pneumatic rotary actuators as actuation mode, as shown in Fig. 2 (left). This exoskeleton was developed at Kanagawa Institute of Technology, Atsugi, Japan [33], [34], and was used for assisting a nurse to carry a patient in her arms. It is a full-bodied suit and consists of three power-assisting units (arm unit, wrist unit, and knee unit). Regarding the lower limb, a direct-drive pneumatic rotary actuator is used for the flexion and extension of each knee. During the development of this exoskeleton, many efforts were conducted to improve the efficiency of the actuators. At the beginning, a rubber tube actuator was developed, and then, a pneumatic slide-box rotary actuator was developed [34]. Although these two actuators are small and powerful, they have both their respective drawbacks: The former has remarkable hysteresis in its performance characteristics, while the later has weak points of solid friction and its overweight. At last, an improved pneumatic rotary actuator was developed by utilizing pressure cuffs, which overcome the drawback mentioned previously. Furthermore, the weight of the actuator for the knee joint is reduced to only 1.5 kg [33]. Experiments showed that the leg actuator could produce a torque of about 15 $N \cdot m$ at the beginning of the stretching, which was about half of the necessary torque (30 $N \cdot m$) for stretching the knee joint. In terms of power supply, micro air pumps with portable battery are used to satisfy the power requirements.

B. *Lower Limb Exoskeletons With Electrical Motors*

Electric actuators were also developed for the BLEEX exoskeleton; hence, we can compare the characteristics and performances between the hydraulic actuators and the electric ones. Mechanical–structural, weight, size, generated torque, and gearing of these electric actuators were also designed based on CGA data similarly to the design of the hydraulic actuators [35]. By comparing the two actuation modes, it has been shown that the weight of the electric actuators is about twice of that of the hydraulic ones; however, the electric actuators are 92% more power efficient than the hydraulic ones during walking. At the same time, unlike the hydraulic actuators that can have part of their weight located away from the joints' axis of rotation, the weight of electrical joints is all centered at the actual joint. This may have some drawbacks for the mechanical design of exoskeletons. This comparison shows that the electric actuators are more power efficient but have a bigger size and are heavier than the hydraulic ones for the same purpose.

However, if the purpose of the lower limb exoskeleton is not to assist the potential wearers to carry relatively heavy load, electric actuators are generally more suitable to be used in exoskeletons. Indeed, the advantages of electric actuators can be highlighted with the decrease of the required torque output (i.e., decrease of actuators' size and weight). Additionally, since the required torques for lower limb exoskeletons and orthoses are relatively high and the speed is relatively low, it is usually hard for direct-drive electric actuators to satisfy the requirements of high torque output, low speed, small size, and lightweight simultaneously. Hence, geared drive and/or cable drive electric actuators are usually used to satisfy such requirements.

The hybrid assistive leg (HAL)-5 is a full-body exoskeleton using electric motors. It was developed by Sankai *et al.* [36]–[41], [92] for both performance augmenting and rehabilitative purposes [Fig. 2 (right)]. Each electric actuator consists of a dc servomotor and a harmonic drive gear. These electric actuators can supply the required torques at the hip, knee joints, and upper body joints (elbow and shoulder joints). Due to the use of advanced aeronautical materials and by reducing the bulky backpack that was present in the previous prototype (HAL-3, etc.), the weight of the current version of HAL-5 has been reduced to 21 kg. Furthermore, since HAL-5 can support itself, the wearer is even not aware of its weight.

In terms of performances, HAL-5 can support healthy wearers during standing up, walking, climbing stairs, and a range of other daily living activities. Additionally, it is claimed that HAL-5 can help a healthy adult lift up to 80 kg of load, which represents roughly twice his normal capability (30–40 kg) without assistance [26]. This means that HAL-5 can support about 40 kg of payloads. It should also be noted that the endurance of the wearers could not be improved significantly using HAL-5. They would become tired quickly when they hold a heavy load.

Ohta *et al.* [42] presented a 2-DOF motor-powered gait orthosis for spinal cord injury patients. An important feature of this orthosis is related to the fact of using linear motor actuators, at the knee and hip joints, as a combination between dc motors and ball screws. The increase of the moment arms will result in the enhancement of the output force, while the size of the actuator is smaller. The masses of the knee and hip joints are only 0.8 and 1.4 kg, respectively. Consequently, the required power supply can be reduced.

C. Lower Limb Exoskeletons/Active Orthoses Using SEAs

Using traditional nonbackdrivable actuators (e.g., electric motors) in some wearable robots showed certain inherent drawbacks, such as the poor torque density of the motors at low speed as well as the friction, backlash, torque ripple, and noise of the gears. In recent studies, researchers have tried to develop a novel type of actuator to remedy these problems. In this case, SEAs are presented.

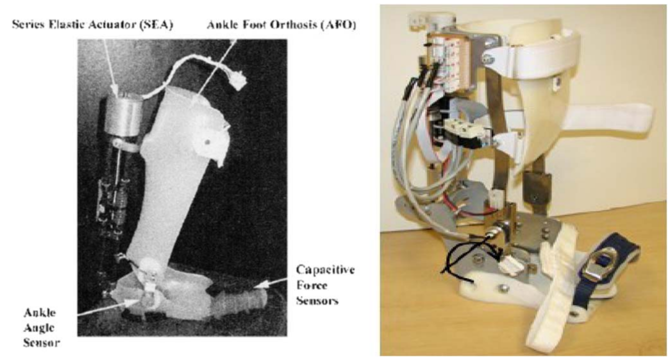

Fig. 3. (Left) MIT AAFO (IEEE image credit: [44]). (Right) Powered ankle-foot orthosis developed by researchers in Arizona State University (IEEE image credit: [45]).

Compared with the nonbackdrivable actuators, the SEA includes the following advantages: shock tolerance, lower reflected inertia, more accurate stable force control in unconstrained environments, and energy storage [23], [24]. Hence, recently, the SEA has been employed in many exoskeletons and orthoses. In this section, only the active SEA is reviewed; therefore, some passive SEAs are not included, e.g., the MIT exoskeleton [10] and the ankle exoskeleton developed by Sawicki *et al.* [43] (they only control the energy storage and release without actuation).

MIT active ankle-foot orthosis (AAFO) [44] is one of the main devices that use the SEA as the actuation mode. AAFO is designed for treating a gait pathology known as drop foot. Regarding the structure of the AAFO, a SEA and sensors are attached to a conventional ankle-foot orthosis [Fig. 3 (left)]. The total weight of the AAFO is 2.6 kg, excluding the weight of an off-board power supply. The SEA consists of a dc motor-powered ball screw mechanism in series with helical springs. It can vary the impedance of the ankle in plantar flexion during stance and assists dorsiflexion during the swing phase of the walking gait cycle based on the measurement of the ground force and joint kinematics [6]. Additionally, the SEA is powered through an off-board power supply by electrical cables. In terms of performances, AAFO showed a significant effectiveness for assisting the drop-foot patients. The experimental results showed that the AAFO has three benefits: reducing the occurrence of slap foot, allowing greater plantar flexion, and proving for less kinematic difference during walking when compared to the walking of healthy subjects [44].

Another powered ankle-foot orthosis, actuated by a SEA-based robotic tendon, was developed by researchers in Arizona State University, Phoenix, AZ, USA, [Fig. 3 (right)] [45]–[47]. Like the AAFO, this orthosis is also a single-joint device designed for supporting the flexion\extension movements of the ankle joint. Although there are some differences between the design of SEA in AAFO and this orthosis, e.g., the robotic tendon used a custom threaded lead screw and a single spring in series with a dc motor, these two types of SEA-based orthoses both show the benefits of using SEAs. In this paper, a comparison between the effectiveness of the robotic tendon and the traditional direct-drive system is done. The results show that, in an ideal example, the peak power required at the ankle joint level when using the robotic tendon is 77 W, while it is equal to 250 W when using the direct motor. This result can be explained by the fact that the robotic tendon can store and release energy during cyclic repetitive tasks. Meanwhile, the weight of the robotic tendon is just 0.95 kg (the latest version is only 0.5 kg), which is seven times less than its equivalent motor/gearbox system. Additionally, ideal energy requirements are reduced from nearly 36 to 21 J. This means that orthosis using the SEA can be more portable, lightweight, and highly efficient.

With the development of wearable robots, different types of SEAs have been designed. For example, Veneman *et al.* [48] developed a series elastic and Bowden-cable-based actuation system. Although the main purpose of this actuator is to support the power source for a stationary robotic trainer LOPES, it might have wider application in wearable exoskeletons. The highlighted characteristic of this actuator is that it introduces Bowden cables to the design of SEA to detach the actual motor from the exoskeleton frame. This can make the exoskeleton frame more portable. In terms of performances, this actuator shows its advantage in three important aspects of wearable robots: bandwidth of force training, reduction of the output impedance, and force fidelity. The bandwidths for full force range and smaller range are 11 and 20 Hz, respectively, while the corresponding required frequencies, which are obtained by studying the human gait cycle and the motion control range of a human therapist, are 4 and 12 Hz, respectively. The impedance output can be reduced to a hardly perceptible level. The fidelities for tracking 2- and 20-Hz sinusoidal trajectories are 98.7% and 94.7%, respectively.

Recently, novel types of SEA, such as those developed in [49]–[51], are continually developed to cover some drawbacks of SEA such as friction to improve the performances such as reducing the weight, improving the compliance, and highlighting the benefits of SEA mentioned previously.

D. Lower Limb Exoskeletons With Pneumatic Muscle Actuators

Artificial pneumatic muscle can be considered as one type of pneumatic actuation. However, it has some specific characteristics, such as high power/weight ratio, relatively lightweight, and inherent compliance. The original concept of the artificial pneumatic muscle was presented by McKibben for prosthetic applications in the 1950s [52]. Due to the complexity of controlling such systems, it was abandoned for a time. However, recently, with the development of new control strategies, more researchers have adopted the pneumatic muscles as exoskeleton actuators. Pneumatic muscles are very simple to manufacture and can be used in antagonistic form like the natural skeletal muscles. In addition, they are very safe and suitable for the rehabilitation applications because of their inherent compliance and their limited maximum contraction.

Caldwell *et al.* [53] developed a lower body 10-DOF exoskeleton for active assistance of the human walking with the use of artificial pneumatic muscles as actuators [Fig. 4 (left)]. There are 3 DOF at each hip joint level, 1 DOF at each knee joint level, and 1 DOF at each ankle joint. Artificial pneumatic muscles are used to provide the flexion and extension torque at each active joint. Due to the use of pneumatic muscles, the resulting exoskeleton is very portable, and the total weight (excluding the power source) is less than 12 kg.

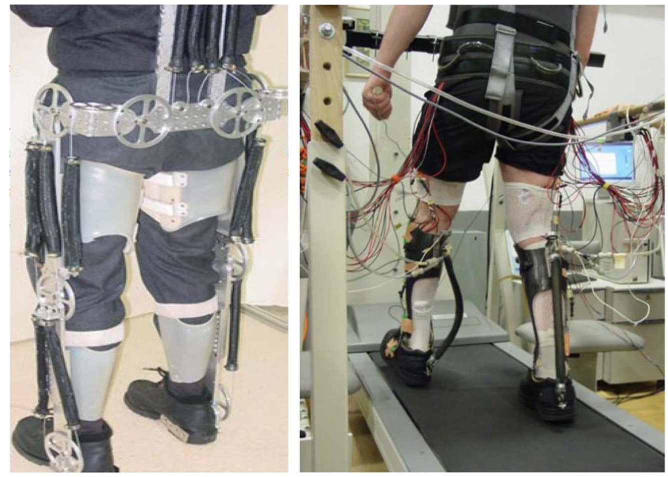

Fig. 4. (Left) Lower body 10-DOF exoskeleton (IEEE image credit: [53]). (Right) Powered lower limb orthosis designed by Sawicki et al. (IEEE image credit: [55]).

Sawicki *et al.* [54], [55] developed a powered lower limb orthosis used for motor adaptation and rehabilitation purposes as shown in Fig. 4 (right). This orthosis also uses the artificial pneumatic muscles to provide flexion and extension torque at the associated joints. Experiments show that this lower limb orthosis can supply the substantial plantar flexor torque: $\sim$57% of the peak ankle plantar flexor torque during stance and $\sim$70% of the plantar flexor work done during normal walking [55]. By comparing the change of the electromyography (EMG) signal of the soleus muscle from initial use to 30 min of walking by healthy wearers, it was shown that the soleus EMG amplitude was reduced by 50% and tended to the normal level [55].

However, there are some disadvantages of using artificial pneumatic muscles. For instance, the control methods become relatively complex when using artificial pneumatic muscles. Moreover, the control bandwidth of artificial pneumatic muscles is relatively low, compared to the hydraulic actuation one. This is due to the fact that the hydraulic fluid is generally incompressible while the pneumatic muscles use the compressible air.

Additionally, there are also some novel actuation designs which integrate the characteristics of two or more types of actuators mentioned previously. For example, Saito *et al.* [56] developed an externally powered lower limb orthosis, which is actuated using bilateral-servo actuator. This actuator consists of two cylinders that can simulate the characteristics of the human muscles. Meanwhile, the actuator is operated stably by the antagonistic action of sealed oil, which is controlled by a servomotor. Hence, this actuator has both the characteristics of the hydraulic and electrical actuators.

III. Human–Robot Interaction and Control Algorithms

In recent years, many studies have been conducted to improve the performance of the lower limb exoskeletons for assistance and rehabilitation training [19], [57]. Alongside with the development of sophisticated and portable exoskeleton designs, researchers also focus on improving the control strategies to increase the precision, efficiency, and comfort of the exoskeletons. Since exoskeletons are worn by humans, the interaction between the wearable robot and the human is a critical factor for ensuring smooth and efficient control strategies that will be based on the estimation of the wearer's motion intention. Additionally, recent evidences about the outcomes of rehabilitation training paradigms have shown that the motion intention plays an important role in rehabilitation training. Hence, accurate prediction of the human motion intention by the human–robot interaction is essential for both movement assistance and rehabilitation purposes. Therefore, in the following, the control algorithms are classified according to the human–robot interaction mode: the signals collected from the human body, the interaction force signals measured between the human and the exoskeleton, and the signals that are only collected from the exoskeletons or active orthoses.

A. Control Strategies Based on Signals Measured From the Human Body

This type of human–robot interaction is designed based on the signals measured from the human body that reflect the human motion intention directly. Hence, the motion intention can be fully estimated without information lose and delay, compared to the other case studies [58]. Two types of signals are usually used: the skin surface EMG [41], [59]–[61] and the electroencephalogram (EEG) [62], [63]. Based on these two types of signals, corresponding control strategies have also been developed to assist the users to ensure daily living activities and rehabilitation exercises.

The brain–machine interface can usually be divided into two main categories: the invasive brain–machine interface and the noninvasive one. Compared with the former one, the surface and the noninvasive brain–machine interface based on the field potentials are more commonly used in the practical control of the rehabilitation robot. The EEG-based interface is considered as a primary mean of noninvasive brain–machine interface.

In the last decade, a great progress has been made on the use of EEG to control exoskeletons [64]–[69]. A lot of researches have been conducted on utilizing EEG-based interface in the exoskeletons and active orthoses, e.g., the BETTER project (Brain–Neural Computer Interaction for Evaluation and Testing of Physical Therapies in Stroke Rehabilitation of Gait Disorders) aimed to improve the physical rehabilitation of patients with gait disorders suffering from spinal cord injury, stroke, or amyotrophic lateral sclerosis by using exoskeletons or gait orthoses and brain signal measurements [58], [70]. In this project, EEG signals are used to discrete commands, such as the intended action [71], [72], while EMG signals are also used for controlling the lower limb exoskeleton.

In [64], an EEG-based generic control (command) interface was presented to enable the user to realize social interactions. An extending P300 speller with characters and symbols was used to help the user control the complex scenarios. First, EEG signals measured at eight electrode positions (mostly over occipital and parietal regions) were recorded to train a linear discriminant analysis classifier to distinguish between the EEG signals of correct commands (characters and symbols) and the other EEG signals during training. Then, the EEG-based interface was tested by helping a user respond to messages and distinguish the symbols for home control, such as "light," "move," etc. Results showed that the user's performance for spelling messages improved with increase of sessions.

Do *et al.* [69] have recently presented an EEG-based brain–computer interface to control a commercial robotic gait orthosis (RoGO). An important aspect of this study is that it was successfully operated by both able-bodied and SCI subjects. It was claimed that it was the first demonstration of "a person with paraplegia due to SCI re-gaining brain-driven basic ambulation and completing a goal-oriented walking task" [69]. In this paper, EEG data were recorded from subjects while they engaged in alternating epochs of idling and walking kinesthetic motor imagery. These data were then analyzed offline to generate an EEG prediction model by using a combination of principal component analysis and approximate information discriminant analysis. In terms of the control strategy, a nonlinear autoregression model was used to compute the posterior probabilities. A series of five 5-min-long online experiments was conducted. During the experiments, the subjects were tasked to ambulate with the brain–computer interface RoGO system by following computerized cues. The results showed that the offline accuracy averaged 86.30% across both subjects (chance: 50%); the cross-correlation between the computerized cues and RoGO walking epochs was 0.812 ± 0.048 (p-value $< 10^{-4}$) across both subjects and all sessions. This study also addressed some challenges that need to be covered, such as elimination of the false alarms and reduction of the response lag delay.

Although many convincing evidences show that it is feasible to use EEG-based interface to control wearable robots, there are still some limitations for widely used EEG-based interfaces to estimate human motion intention. The limitations mainly include the relatively high sensitivity, the shortcomings of the wearable EEG measurement equipment, the overlapping of different electrical activities generated by different cortical areas, and the balance between the training time and accuracy [68], [73].

On the other hand, the EMG signals have been extensively studied by many researchers to analyze human disabilities and to monitor the progress of the rehabilitation processes [60]. Some studies have shown that there are important advantages of using the EMG signals for predicting the human motion intention. For example, the relatively weak EMG signals (i.e., EMG signals from some patients or elderlies) can even be used to estimate the intention efficiently; the relationship between the joint torque and the corresponding EMG signals in isometric contractions is linear [41], [60].

For the HAL-3 and HAL-5 systems, EMG signals are used to measure the level of human–robot interaction [26], [40],

[41]. The bipolar skin surface electrodes are placed over the selected muscles (biceps femoris and medial vastus associated with the knee joint, and gluteus maximus and rectus femoris associated with the hip joint). The joint torque is, respectively, estimated by different antagonistic EMG signals in the extensor and flexor muscles. In addition, many works were conducted for the calibration of the EMG signals, which is one of the major challenges that need to be solved when using the EMG signals for human intention prediction. In the power assist control methods of HAL-3, the required assistive torques for each motion, such as walking and standing up, are calculated according to the EMG signals. However, by using the HAL-5 system, researchers added a new control system besides the EMG system [26]. This new control system is used to store the walking patterns, which are recorded when users wear HAL-5 for the first time. This control system can match the wearer's gait and can be used for the category of wearers whose EMG signals are difficult to collect. The EMG system is then used to predict the wearer's intention to move and triggers the associated actuators into action.

Researchers from Berlin University of Technology, Berlin, Germany (Fleischer *et al.* [59]–[61]), also presented a method to calculate the human walking intention for an exoskeleton by combining EMG signals and a human musculoskeletal model. The human body model consists of two legs with feet, shanks, thighs, and torso. The EMG signals are measured at the selected muscles responsible for the flexion and extension of knee joint (semimembranosus and vastus medialis). Besides the EMG sensors, pose sensors are used to obtain the postures of the wearer. Regarding the control strategy, EMG data are processed by an EMG-to-force function to estimate the intended forces. The forces are then used to calculate the desired angular accelerations of the knee joints by using the forward dynamics of the human body model. These angular accelerations are set as the reference acceleration of the orthosis. An online calibration method was also proposed by iteratively optimizing the desired torque and the torque calculated using the inverse dynamics of the human body model. In terms of performances, healthy subjects have conducted the experiments. By comparing the calculated torque based on the use of EMG data and the one computed using the inverse dynamics during the free daily living activities such as walking, climbing stairs, etc., it was shown that these two torque curves matched very well.

Although EMG signals have been widely used to estimate the human intention prior to the system control, there are still some inherent limitations during practical application that need to be solved in the future. For example, the EMG signals are easily influenced by the electrode placement neighboring muscle signals and the noises that affect the original EMG recordings. Also, novel methods need to be proposed to simplify the calibration procedure of the EMG signals, since they may change considerably from wearer to wearer and day to day.

Besides the EMG and EEG, muscle hardness is able to infer the motion intention of the wearer. A novel muscle sensor [based on force sensor resistor (FSR)] was designed for the nurse-assisting exoskeleton [33]. This muscle sensor consists of a contact projection made of crude rubber and an FSR, which are fixed on the muscle by a rubber band. Hardness of the muscle is detected by the contact projection between the muscle and the FSR. When muscles are generating forces, hardness of muscles is also changing correspondingly. Based on this principle, the contact projection can transfer hardness of muscles to the pressure on the FSR. Then, pressure can be used to estimate the motion intention. At the same time, some optimum points for detecting muscle hardness at the user's limb were chosen. The sensors were used to predict the motion intention of the elbow joint, knee joint, and wrist joint. The experiments showed that the muscle hardness sensors were almost proportional and sensitive to the weight of the load. Hence, the signals could be considered as the input of the control system and reflect the motion intention.

B. Control Strategies Based on Interaction Force Measurement

In this case, human–robot interfaces are designed according to the interaction forces between the user and the exoskeleton. Some of them measure the interaction forces directly from the attachment points between the users and the exoskeletons; others may measure the interaction forces through the deformation of an elastic transmission element or structure placed at the robot link [10], [36].

De Rossi *et al.* [74] proposed a novel distributed method to measure the pressure interaction between the human and the robot. This new human–robot interface was tested with LOPES lower limb exoskeleton worn by healthy subjects. Note that only one sensor system was tested at the wearer's thigh. Three types of measurements were conducted to evaluate this sensor system: static test, dynamic test, and gait training test. During the static test, the subject was asked to generate an incremental torque at the hip joint without movement. In the dynamic test, a given chirp torque was applied at the hip joint of the LOPES exoskeleton while the subjects were asked to keep standing. Additionally, a six-axis load cell was used in the interface at the same time. By comparing the curve of the measured torques using load cell and this interface, respectively, it was shown that the human–robot interface can accurately reflect the interaction force between the human limb and the exoskeleton. Moreover, this distributed design can improve the comfort and safety of the users and detect the distributed pressure in real time. In addition, one of the interesting aspects of this human–robot interface lies in the fact that developing a general framework, flexible and adaptable distributed interaction measurement system, applicable to different kinds of exoskeleton devices.

A human–robot interface based on the forces was also developed for HAL-5 [37], which uses the floor reaction force (FRF) to estimate the motion intention. The FRF is used to calculate the position of the center of gravity that can be a reliable information for the intention estimation. When assisting a given patient, the walking motion is divided into three phases: swing phase, landing phase, and support phase. Reference patterns can be calculated for each phase. The control strategy measures the patient's walking intention through the use of FRF and assists him or her to walk; however, it cannot guarantee patient's balance and stability.

In [75], a force-sensing design for extracting information regarding physical interaction through a force sensor is presented. The latter is placed at the kinesis exoskeleton structure to perform online assessment of human–robot interaction. The online physical interaction assessment procedure also allows the estimation of the muscle performance during dynamic walking control. In general, the human–robot interface is based on the interaction forces and facilitates the design of the control strategy, since it can directly reflect the human intention at every joint without building complex models of the exoskeleton or human body. However, the accuracy of the interaction model and the comfort of the user are two important points which should be taken into account when designing any exoskeleton prototype and measuring the human intention.

Based on the interaction-force interfaces, impedance control has also been widely used in the control of exoskeletons and powered orthoses, especially for rehabilitation purposes. When the wearer's movement deviates from the normal gait, the exoskeleton or orthosis provides an interaction force (mechanical impedance) to assist the wearer to return to rational gait; however, it does not intervene if the wearer's movement moves along the normal gait. This strategy is so far called "assist-as-needed." Many advances have been made in this field. For example, Agrawal *et al.* [76] presented a force field controller to help the patients accomplish the neuromotor gait training. A virtual tunnel was designed along the desired trajectory, which keeps the foot in the desired range of the tunnel and allows the foot to deviate from the desired trajectory within certain limits. By adjusting the width of the virtual tunnel, the deviation area and the impedance force can be adapted. The experiments proved that the proposed control strategy is able to alter the gait pattern of healthy subjects in short-term training. There are also many other types of impedance-control-based methods. For more details about impedance control methods, the reader is invited to refer to a recent review done by Marchal-Crespo *et al.* [57].

C. Control Strategies Based on Signals Measured From Exoskeletons or Active Orthoses

There are some control strategies that do not collect any signals from the human body or any interactive forces between the user and the exoskeleton. The advantage of this type of control strategies is that they do not need to design an additional interface to predict the wearer's motion intention, whereas they are able to estimate or follow the wearer's intention based on the information obtained from the exoskeletons. This kind of control system is usually designed based on the models of the exoskeletons and/or the human body.

It has been applied to the BLEEX exoskeleton and only uses information from the exoskeleton to estimate the human motion intention [27]. To obtain this information, force and torque sensors are installed at the hip, knee, and ankle joints to measure the forces and the torques imposed by the hydraulic actuators and the wearer. In particular, the control system needs a high level of sensitivity to respond to the forces and the torques imposed by the wearer. The control algorithm consists of two closed loops: one presents how the wearer affects the exoskeleton, and the other one shows how the actuators affect the exoskeleton. Unlike the classical control method that results in relatively low sensitivity, this control algorithm needs to maximize the sensitivity of the closed-loop system to forces and torques. To achieve this goal, the dynamics of the exoskeleton needs to be calculated, which usually requires a realistic model of the exoskeleton and the wearer load. Due to the sensitivity amplification method and the gravity compensation (the exoskeleton is able to guarantee the balance by itself), the wearer can use a relatively small force to move the exoskeleton legs.

As mentioned previously, this type of human–robot interactions or control strategies usually needs an accurate model of the exoskeleton, which is generally difficult to calculate and sometimes impossible to obtain for some exoskeletons. Hence, in the later research of the control method of the BLEEX exoskeleton, a new hybrid control method was developed. The walking gait cycle is divided into two phases: stance phase and swing phase. According to these two phases, the position control and the previous sensitivity amplification method are used in each phase, respectively. Due to the characteristics of the swing phase, the control algorithm does not need the computation of the exoskeleton model. However, as a tradeoff, the exoskeleton needs the use of some wearable sensors to measure the human limb and torso angles.

IV. Performance Assessment Methods

The performance of lower limb exoskeletons determines the acceptability of such devices by the potential users. Reduction of the metabolic cost and the average muscular generated torques by the wearers are the common goals of the lower limb exoskeletons for locomotion assistance. Most of the related researches mainly focus on the performance of the mechanical design or control algorithm from the engineering perspective. A few research studies have treated the effect of human kinesiology on the assistive devices, particularly metabolic cost and gait analysis when the users are wearing the exoskeletons. Although many improvements had been made on the mechanical design of lower limb exoskeletons, many devices were abandoned by the users/patients because of the enormous physical energy requirement [8], [19]. Hence, recently, more researchers and physiologists have started working in close collaboration trying to tackle these issues. So far, varieties of assessment methods have been presented in relevant literatures. According to the measuring means, most of these assessment methods fall into three main categories: metabolic cost, gait biomechanics, and muscular activity analysis. In this section, these assessment types and the performance of some typical exoskeletons based on these evaluation methods are reviewed and discussed. Note that, although some exoskeletons were not included in the aforementioned sections, their assessment methods are meaningful for the types that are included.

A. Metabolic Cost

Regarding the augmented-performance exoskeletons, an efficient assessment performance method is to compare the metabolic energy expenditure of the user while walking with the use of an exoskeleton and without the use of an exoskeleton. The oxygen consumption, the carbon dioxide production, and the urinary nitrogen excretion are the basic measurement means. Meanwhile, a number of measurement methods and models have been presented [77]–[79].

The MIT exoskeleton [13] is considered to be the first reported study that compared the metabolic cost while the user was walking with the assistance of the exoskeleton and without it. However, the result was undesirable (the metabolic expenditure was higher by 10% when the exoskeleton was worn). Similar conditions also occurred in other researches; Gregorczyk *et al.* [80], [81] studied the efforts on metabolic cost when carrying loads using a lower limb exoskeleton prototype. Nine U.S. Army participants were asked to walk at 1.34 m/s on a 0% grade for 8 min while carrying the loads of 20, 40, and 55 kg with and without the exoskeleton. The results showed that the mean oxygen consumption was significantly higher by 60% (body mass) and 40% (body mass and exoskeleton mass) when the participants wore the exoskeleton, compared with the normal condition. Meanwhile, this study also analyzed the reasons: the change of the users' gaits and walking patterns were thought to take the main responsibility of this situation.

On the other hand, there are also some lower limb orthoses that are reported to reduce the metabolic cost of the user while walking. For instance, Sawicki *et al.* [82] used an ankle exoskeleton, powered by artificial pneumatic muscles and controlled through the EMG signals, to study the metabolic cost of plantar flexor mechanical efforts exerted during walking with different step lengths at constant frequency. In this paper, nine healthy subjects participated in the experiments, and the results show that the exoskeleton slightly altered the user's ankle joint kinematics during walking for all of the speed/step-length conditions. In this case, the net metabolic power $(6.19 \pm 0.29\ \mathrm{W \cdot kg^{-1}})$ with the assistance of the exoskeleton is significantly lower for walking without the exoskeleton $(7.18 \pm 0.50\,\mathrm{W \cdot kg^{-1}})$ at 1.75-m/s walking speed. However, as the speed became slower (at the same stride frequency, only reducing the stride length), the effectiveness decreased. Although some single-joint powered orthoses have demonstrated that they can reduce the metabolic cost in certain conditions, most other studies show that the metabolic cost is still relatively higher when the users are walking with the assistance of the lower limb exoskeleton and/or the load. This is a common problem for all lower limb exoskeleton studies, and further explorations need to be conducted in the future.

B. Gait Biomechanical Analysis

Gait biomechanical analysis is also an important assessment method, especially for rehabilitation purposes. Variables that can be classified as gait kinematic variables, temporal–spatial gait variables, physiological cost variables, and other variables related to the gait function are frequently used as assessment indicators. Gait kinematics is direct and meaningful for the joint control and performance evaluation. Temporal–spatial variables such as speed, walking distance, step length, and cadence are also frequently used. This assessment method is the most common one used to evaluate the effects of lower limb exoskeletons and orthoses on gait functions.

Ohta *et al.* [42] developed a 2-DOF motor-powered gait orthosis for spinal cord injury patients. The gait biomechanics (stability, gait speed, step length, etc.) were used to assess the performance of the orthosis. Five male patients with complete spinal cord injury took part in this study. After a period of training with the gait orthosis, their gait speed and length showed a certain increase (from 3% to 23% and 0% to 7%, respectively). At the same time, the stability and dynamics of walking were also improved, while the lateral and vertical compensatory motions showed a relatively low decrease.

Clinical experiments were conducted to evaluate the performance of the AAFO developed by Herr *et al.* [44]. Three drop-foot participants and three normal subjects took part in this study. The kinematic and kinetic data were collected to analyze and compare the six subjects' occurrences of slap foot, swing dorsiflexion angular range/power plantar flexion angle, and gait symmetry. In their research, three different control conditions were evaluated: zero, constant, and variable impedances, where the zero impedance was meant to approximate unassisted drop-foot gait and constant impedance was meant to imitate the conventional ankle-foot orthosis. The results showed that, by comparison with the zero- and constant-impedance control schemes, the occurrence of the slap foot of the drop-foot participants was reduced, and the swing dorsiflexion angular range and maximum plantar flexion angle were increased by actively adjusting the joint impedance. Meanwhile, the swing phase ankle kinematics was closed to the physiological pattern. In terms of gait symmetry, it was found that the variable-impedance controller improved the spatial and temporal gait symmetry in comparison to the zero-impedance controller, while there were no significant differences between the effectiveness of the variable-impedance and constant controllers.

Furthermore, gait biomechanical analyses were also done by other researches on exoskeletons/powered orthoses, such as the HAL exoskeletons [36], the powered gait orthosis developed by Kang *et al.* [83], and the Lokomat gait orthosis [84]. Most of these researches have shown that exoskeleton/active orthosis have the advantages of improving the patients' gait abnormality or stability. On the other hand, there are also some studies on lower limb exoskeletons, which demonstrated that the gait biomechanics of the users could be negatively altered when walking with some exoskeletal devices. For example, Gregorczyk *et al.* [81] found that the kinematics and the kinetics of the users' walking patterns (healthy subjects) were all negatively altered. Furthermore, many studies have shown that the gait biomechanics can be improved when using rehabilitation exoskeletons with elderlies or patients, while it cannot show a desirable result when using exoskeletons for performance augmentation.

C. Muscle Activity Analysis

Averages of the muscle's activation levels can be used to assess the performance of the lower limb exoskeleton, as the level of the muscle's activation depends on the torque imposed by the corresponding joint and/or consequently the assistance performance. This assessment method was used for the evaluation of the HAL-3 exoskeleton for assisting healthy users during walking, standing up, ascending stairs, etc. [36]. The results show that the activation levels of user's muscle decreased with each type of activities when the user was assisted by HAL-3. This means that the user employed lower force/torque to accomplish the same activities.

In addition, Akahira *et al.* [85] used the muscle EMG activities to assess the effectiveness of a motor-assisted knee motion device for orthotic gait. Six paraplegic persons with traumatic SCI have participated in this study. All subjects had injuries at thoracic level and had complete motor paralysis in the lower limb muscles. For comparison purposes, two reciprocal gait orthoses (an advanced reciprocating gait orthosis (ARGO, Steeper Inc., U.K.) and the improved powered orthosis) were tested. For the ARGO, it does not have any powered actuators. However, the torque exerted by the right or left hip joint can be mechanically transmitted to each other by a reciprocal cable, which connects both sides of the leg frame. By this means, each leg is propelled forward reciprocally. For the powered orthosis, it consists of the hip joint and knee joint, which are both actuated by motor-based actuators. All experiments were conducted on a treadmill system to guarantee the same walking situation as well as the safety of the patients. By measuring the muscle EMG activities of the soleus, medial head of the gastrocnemius, tibialis anterior, rectus femoris, and long head of the biceps femoris, it was found that the amplitudes and phases of the EMG activities of gastrocnemius and rectus femoris muscles were enhanced and prolonged significantly, as functions of the knee motions generated by the powered orthosis. This clearly means that the active orthosis has a potential to activate the neuromuscular function with paralyzed lower limbs.

Furthermore, the capacity of robotic force guidance (and gait speed) in affecting neuromuscular control is essential for designing the correct reference and control systems to develop assist-as-needed robotic rehabilitation protocols. The dependence of muscular activation on the level of robotic exoskeleton control during assisted walking (guidance force and speed), for lower limb muscles, has been characterized by means of EMG analysis in [86].

Except the methods mentioned previously, there are also other assessment methods for lower limb exoskeletons/active orthoses based on different special purposes. For instance, some military exoskeletons are designed to help soldiers carry heavier loads and walk faster, and this usually exceeds the capability of a normal person. Some rehabilitation exoskeletons or active orthoses are developed to assist the users with completely motor disability to accomplish walking. In these cases, the assessment method of these lower limb wearable devices can be considered as a limit-challenge method, and some assessment methods like metabolic cost would not be the primary methods.

V. Discussion

After the early lower limb exoskeletons and orthoses that were initiated in the 1960s, many related researches have been done during the last decades. In particular, in the last five years, an exponential increase of researches on the exoskeletons and orthoses has been shown, compared with the last 40 years [19]. A number of advances have been made to improve the wearability, portability, and performance of robotic exoskeletons and orthoses. Some small and light actuators, comfortable and relative accurate human–robot interfaces, and efficient and long-lasting power supplies have been developed. At the same time, human body models and gait biomechanics have also been developed to provide a solid scientific background for the mechanical design and the control strategies [5]. More clinical theories and physiology theories have been used to assess the performance of the wearable robots as well. However, these developments still have not met the requirements of the wearable robots to be used in daily practices. There are still many works that need to be done.

A. Factors Limiting the Lower Limb Wearable Robots

In terms of actuators and mechanical designs, portability and efficiency need to be further improved. For the hydraulic actuators, the efficiency is still relatively lower in comparison to the electrical motors. This means that more power consumption is required. On the other hand, novel lightweight electrical actuators need to be developed (relatively high weight is usually one of their inherent disadvantages). SEA has been an important trend in the development of actuators for lower limb exoskeletons and orthoses. Developing compact and highly efficient series elastic components will still be a critical challenge in the future. The artificial pneumatic muscle may represent a good choice for a wearable robot because of its relatively low weight, high efficiency, and intrinsic safety. However, relatively low control bandwidth and complex control strategy are the usual challenges for its application. In addition, the noises resulting from the actuator use are still an important issue to take into account. Hence, developing new lightweight, highly efficient, and stable actuators still remain a challenge for the researchers in the near future.

Regarding the mechanical design, many of the current prototypes affect and alter the gait biomechanics of the normal human walking pattern. This results in a big metabolic cost. In this case, human body biomechanics need to be extensively considered during the mechanism design of any lower limb or full-body exoskeleton. Additionally, associated aspects such as materials and power supply need to be improved to lighten the exoskeletons and support long-lasting power.

In terms of human–robot interfaces and control strategies, the approaches based on the information measured from the human body, such as the EEG and EMG signals, have shown to allow relatively quick user adaptation to the exoskeleton use [89]. Future trends of the human interface will be to use the neural signals to estimate the user's intention that naturally exists when the wearer executes a given task [90]. However, the sensitivity of the EEG/EMG electrodes, the accuracy of the model between the motion intention and the EEG/EMG signal measurements, etc., are still challenging problems. Hence, developing the sensor technology, the EEG/EMG electrodes, the wearable force sensors, and the kinematic measurement sensors will constitute important research directions. In addition, another great challenge of human–robot interfaces is to ensure the safety and comfort of the human limb while wearing the exoskeletons. Regarding the control strategies, more quantitative assessment methods (metabolic cost and kinesiologic-based methods) need to be considered to assess the performance of the control strategies rather than considering mainly the control perspective.

B. Future Research Directions

In the future, the enabling technologies will constitute a key research direction for wearable robots. The components of the wearable robots, such as the sensors, the actuators, and the power sources, will be further developed to make the wearable robots more portable and highly efficient. In terms of actuator design, Herr *et al.* [11], [87], [88] have shown that the artificial muscle actuators will play an important role in the research direction of the wearable robots. Furthermore, the recent research conducted by Haines *et al.* [25] also shows that artificial muscle actuators can present many characteristics similar to those of skeletal muscles, such as noise-free operation, nonhysteretic, etc.

The human–robot interface and associated control algorithms are another parallel important future direction. The improvement of the accuracy of estimation of the human motion intention, as well as ensuring the wearer's comfort, will guarantee smooth assistance and better rehabilitation effects. Since the inherent advantages for estimating human motion intention, the human–robot interfaces that measure information from the human body will become a crucial research direction to push the wearable robots into wider practical application. Researches on novel control techniques to implement efficient assist-as-needed strategies will also be required in the future. In addition, the understanding of the underlying physiological mechanisms that govern healthy and pathological gait, the reduction of the metabolic energy expenditure, and the study of the effects of exoskeletons on the neuromuscular control during an intervention will chiefly guide the mechanical design and the development of the control methods [89].

Safety issue is still a crucial challenge for the lower limb wearable devices, particularly devices designed for the elderly and patients with pathologies. On one hand, wearable devices should have redundant mechanisms for its electromechanical system. Wearable devices should guarantee users' safety in case of exceptional situations. Hence, some important parameters, such as operational velocities and interactive pressure, should be detected in real time, and given the safety constraints, at the same time, special mechanisms and adapted control strategies should be taken into account to guarantee user's stability and safety in emergency. On the other hand, building the biomechanics assessment system for the design of wearable devices and desired trajectories will also be essential in the future. The wearable robots should be chosen carefully according to the characteristics of impaired users and the associated biomechanics assessment system. Unsuitable gait biomechanics for the patients with lack of motor function may cause a chronic secondary damage.

VI. Conclusion

This study has addressed the latest advanced research in the lower limb wearable robots for assistance and rehabilitation. The mechanical design of the reviewed robots was classified based on the actuation modes. The advantages and disadvantages of the use of different actuator modes have been analyzed according to different purposes. Three types of common human–robot interfaces and the associated control methods were also reviewed. Additionally, three major assessment methods used with lower limb exoskeletons were reviewed and classified. Some robots' performances have been discussed based on these three assessment methods.

One can notice that it is hard to find an exoskeleton or active orthosis that can really reduce the metabolic cost during daily activities. Relatively heavy weights of the exoskeletons and associated power supply as well as the efficiency of actuators have an important adverse impact on exoskeletons' performances. Developing more portable and highly efficient wearable robots will still be a crucial challenge in the future. Additionally, the deviation between the wearers' real motion intention and that estimated by the human–robot interface also increases the wearers' metabolic cost and introduces certain constraints on the motion of wearers. Hence, the accuracy of current human–robot interfaces needs to be further optimized. Finally, more investments need be devoted in human biomechanics and mechanism designs of the wearable robots. This will have considerable effects on the wearer's motion performance and human metabolic cost.

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