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Soft Ankle-Foot Exoskeleton for Rehabilitation: A Systematic Review of Actuation, Sensing, Mechanical Design, and Control Strategy

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***Abstract*—Robot-assisted rehabilitation therapy has become a mainstream trend for the treatment of stroke patients. It can not only relieve physiotherapists from heavy physical duties, but also provide patients with effective ankle-foot rehabilitation and walking assistance. Soft ankle-foot exoskeletons have rapidly evolved in the last decade. This article presents a compressive review of soft ankle-foot exoskeletons in terms of robot actuation, wearable sensor, mechanical design, and control strategy. Representative commercial and laboratory ankle-foot exoskeletons are demonstrated. Special attention is paid to the emerging soft actuators, wearable sensing techniques, and human-in-the-loop and hierarchical control methods. Finally, essential challenges and possible future directions are also analyzed and highlighted in this paper, which can provide reliable guidance on the development of next-generation soft ankle-foot exoskeletons.**

***Index Terms*—Soft ankle-foot exoskeleton, actuation, mechanical design, control strategy**

I. Introduction

According to World Health Organization (WHO), the proportion of the world's population over 60 can rise from 12% to 22% between 2015 to 2050. Meanwhile, about 15% of the world's population experience some form of disability, of whom 2-4% suffer significant functional difficulties [1]. Due to the aged society and increasing disabled population, ankle joint injury can be easily caused by external forces or nervous system diseases, and long-term abnormal gait patterns may lead to increasing energy consumption and secondary injury [2]. Therefore, it is imperative for patients to improve motor function and mobility [3], [4]. Rehabilitation through active exploration and movement can promote the immediate and dramatic increase in new neural connections associated with the injured area, which effectively promotes neural circuit remodeling [5], [6]. Traditional rehabilitation is heavily dependent on the physiotherapist's experiences and physical capability, which makes it difficult to meet the demands of high-intensity and repetitive training. Robotic exoskeletons have the potential to maximize neural recovery through repetitive passive training, resistance training, and isometric training [7], [8], which play a positive role in nerve remodeling through function-oriented exercise [9].

Ankle-foot orthosis (AFO) can prevent foot drop and inversion by locking the ankle in a neutral position, which may result in instability in stance, elevation of energy consumption during walking, and lack of adaptability in varying conditions [10]. Compared with AFO, ankle-foot exoskeletons can provide auxiliary torques for dorsiflexion and plantarflexion movements to ultimately improve pathological gait, which can compensate for ankle function and promote functional recovery after stroke [11], [12]. Moreover, ankle-foot exoskeletons possess the capacity to interact with the external environment by adopting the multimodal human-robot interaction (HRI) method, which includes physical HRI and cognitive HRI [13], [14]. The COVID-19 pandemic causes a shortage of medical resources, whereas stroke patients require constant monitoring and nursing [15]. Robotic exoskeletons can be integrated into home-based rehabilitation systems to achieve tele-rehabilitation and tele-supervision [16]-[18].

Several exoskeletons have been successfully commercialized and approved for marketing [19], [20], such as Rewalk [21], [22], Ekso [23], [24], HAL [25], [26], Lokomat [27], [28], and BEAR-H [29], as shown in Table I. These exoskeletons are predominantly powered by rigid motors, which cannot provide intrinsic mechanical safety. Multi-sensor fusion combines information from different sensors to overcome the shortcomings of the single sensor, which has the potential to decode motion intention more effectively [30]. However, patients slipping on the sensor surface may result in sensor slippage and detachment, thus disrupting the measurements [30], [31]. Due to the uneven surface of the human body, wearable sensors that can be attached to the skin surface have significant advantages [32]-[34]. Most commercial exoskeletons weigh over 20kg [21]-[24], [29], making exoskeletons cumbersome and inconvenient to operate, and the ankle part of most exoskeletons is unactuated [21]-[28], resulting in uncoordinated and laborious walking. Furthermore, some ill-fitting mechanisms may cause skin abrasions, and 3D printing can realize lightweight and compact designs with the ergonomically comfortable interface [12]. In the process of rehabilitation training, an effective HRI strategy can improve the effectiveness and comfort of rehabilitation [35]. The commercial exoskeletons lack patient-dominated control to decode the patient's motion intention, which makes it difficult to promote sustained motivation for rehabilitation [27]-[29].

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TABLE I
COMMERCIAL WEARABLE EXOSKELETONS WITH ANKLE JOINT

Device	Actuator	Sensor type	DOFs	W	User H/W	Manufacturing method	Low-level	Mid-level	High-level
ReWalk ReStore [21], [22]	DC motors	Tilt sensor, foot sensors, angular sensors	Hip FE(actuated); Knee FE(actuated); Ankle DP(passive)	23.3	160–190 /100	3D printing /Rigid metallic structure	Closed-loop control	Reference-trajectory generator	Voluntary control
EksoNR [23], [24]	DC motors	Crutch sensors, foot sensors, IMU, potentiometers, and encoders	Hip FE(actuated); Knee FE(actuated); Ankle DP(passive)	23	158–188 /100	3D printing of individual components/ Metallic structure [48]	PID control	Trajectory generator	State machine (using button or HMI)
HAL [25], [26]	DC motors	Bioelectrical signal sensors, foot sensors, angular sensors, acceleration sensors	Hip FE(actuated); Knee FE(actuated); Ankle DP(passive)	14	150–190 /100	3D printing/ Metallic frames with molded plastic bands	PD control	FSM	Voluntary and autonomous control
Lokomat [27], [28]	Linear drives	Force sensor, angular position sensor	Hip FE(actuated); Knee FE(actuated); Ankle DP(passive)	NA	35-47 (Thigh length) /135	CNC machining/ Metallic structure	PD control	Trajectory generator	Impedance control
BEAR-H [29]	Compliant actuators (referring to SEA)	Foot sensors, IMU, encoders	Hip FE(actuated), AA(passive); Knee FE(actuated); Ankle DP(actuated), IE(passive)	24	150-190 /85	CNC machining/ Metallic structure	PD control	Individualized trajectory generator	Adaptive impedance control

FE = flexion/extension, DP = dorsiflexion/plantarflexion, AA = adduction/abduction, IE = inversion/eversion, SEA = series elastic actuator, IMU = inertial measurement units, W = weight (kg), H = height (cm), HMI = human-machine interface, PID control = proportional-integral-derivative control, PD control = proportional and derivative control, FSM = finite-state machine, CNC = computer numerical control.

Soft exoskeletons are actuated by soft actuators or are made of soft materials, including textiles, velcro, and straps. However, most exoskeletons can be a hybrid of soft and rigid exoskeletons, and not technically are fully soft exoskeletons. Currently, most soft exoskeletons are rigid-flexible-soft hybrid structures, such as the exoskeleton with soft actuators adopting rigid load-bearing structure and the exoskeleton with rigid actuators mounted on the lower limb adopting the soft structure. 3D printing can be used to fabricate key components for lightweight and high-performance exoskeletons, as well as to enhance the strength of soft structures at specific locations. Wearable sensors combined with hierarchical control method can make HRI more intelligent. The above characteristics make the soft exoskeleton can be used as a supplement to the rigid exoskeleton in the middle and late stages of rehabilitation.

Several reviews have been published that focused on addressing the key problems of exoskeletons. Some reviews presented the soft pneumatic artificial muscles (PAMs) and sensors in terms of the design, modeling, and control, in which linear PAMs and bending PAMs could be embedded in the robotic system [36], [37]. Kian et al. [30] and Tiboni et al. [31] provided a comprehensive review of wearable sensors in ankle-foot exoskeletons, and other reviews highlighted wearable e-skin in health monitoring and multifunctional applications [38], [39]. Tang et al. [40] divided exoskeletons into passive exoskeletons, quasi-passive exoskeletons, and powered exoskeletons. Meng et al. [41] grouped them as treadmill-based exoskeletons and orthosis-based exoskeletons, and Moeller et al. [42] mainly considered stationary and mobile exoskeletons. Additionally, other reviews kept detailed focus on cable-driven wearable robots [43] and lower-limb exosuits [44]. As the aspect of control, Tucker et al. [45] and Li et al. [46] categorized control strategies according to similarities within the whole strategy, and Sawicki et al. [47] mainly focused on control strategies reducing energy consumption. In recent years, novel key technologies have been widely deveoped for ankle-foot exoskeletons, but they have not been systematically summarized in previous papers.

This paper keeps a detailed focus on state-of-the-art technologies, current research gaps, and future directions. The main contributions of this article can be listed as follows: First, we present a compressive review on robot actuation, wearable sensor, mechanical design, control strategy, and clinical trial of soft ankle-foot exoskeleton, and these aspects have summarized the key and state-of-the-art technologies that need to be taken into account throughout the design process. Moreover, special attention is paid to novel pneumatic actuators, multifunctional e-skin sensors, rigid-flexible-soft hybrid mechanisms, and human-in-the-loop control. Finally, the research gaps and possible future directions are analyzed in detail. The remaining of this paper is organized as follows: Section II presents the research method. Section III concludes the soft actuator, wearable sensor, and mechanical structure. Section IV summarizes the hierarchical control method and clinical research. Section V analyses and concludes the research limitations and future directions. Finally, conclusions are drawn in Section VI.

II. RESEARCH METHOD

In the review paper, literature related to "soft ankle-foot exoskeletons" published from 2003 to 2023 was retrieved through the Scopus and Web of Science database with four retrieval formulas: Formula 1: TS= (commercial exoskeleton OR ReWalk OR Ekso OR Indego OR ATLAS OR Erigo OR Lokomat OR Andago OR BEAR-H OR HAL); Formula 2: TS= ((ankle OR lower limb) AND (exoskeleton OR exosuit) AND (rehabilitation OR soft OR wearable) NOT hip NOT upper limb); Formula 3: (TS= soft actuator OR Bowden cable OR cable-driven OR (pneumatic AND (actuator OR muscle)); Formula 4: (TS= wearable sensor OR soft sensor OR electronic skin) [49].

Formula 1 mainly focuses on searching for successfully commercialized exoskeletons, prioritizing those exoskeletons proposed in journals that support clinical verification. However, all exoskeletons are designed for the rehabilitation of lower limbs. We ultimately decided on five commercial exoskeletons with ankle joints. Formula 2 mainly focuses on searching for soft ankle-foot exoskeleton prototypes in the laboratory stage. Soft ankle-foot exoskeletons mainly provide mechanical power to the wearer's ankle for dorsiflexion and plantarflexion movements. The mechanical structure of soft ankle-foot exoskeletons is usually designed from the shank to the ankle joint or the sole, which can prevent foot drop, eversion, and inversion. Papers that meet the above inclusion criteria and standards are defined as papers related to the topic. Formula 3 and Formula 4 focus more on searching for soft actuators and wearable sensors, respectively, which are expected to apply to soft ankle-foot exoskeletons.

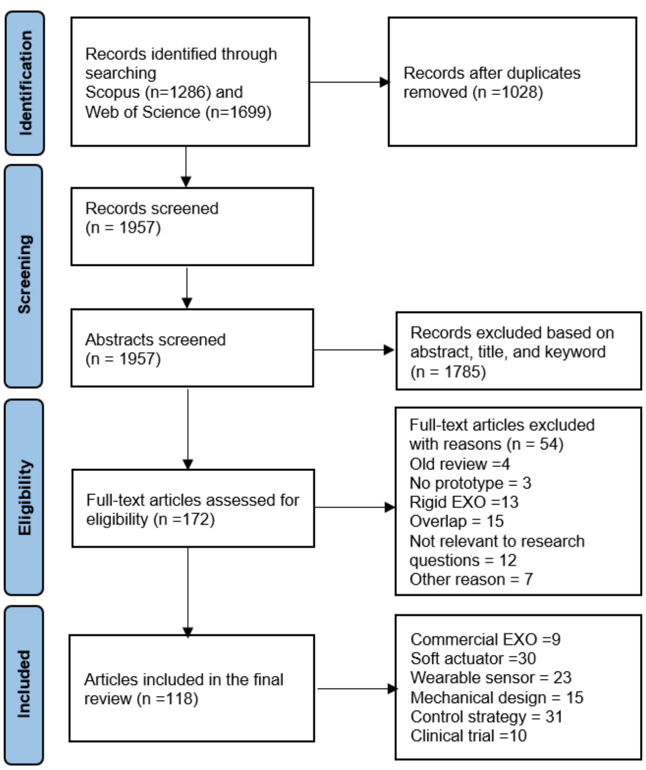

Fig. 1. PRISMA diagram on search results.

The literature downloaded from the database was exported in PDF format and then imported into Endnote software. 2985 articles related to soft ankle-foot exoskeletons were initially searched from two databases (Scopus (n=1286) and Web of Science (n=1699)), as shown in Fig. 1. 1957 articles were screened after removing 1028 duplicates. After screening the abstract, title, and keyword, 172 articles were assessed for eligibility based on full text. Finally, 118 studies were selected.

III. Exoskeleton Components

A. Soft Actuators

Compared with rigid actuators, soft actuators possess better flexibility, intrinsic mechanical safety, and HRI performance [30], [31], [49], which mainly include elastic actuators, cable-driven actuators, and PAMs. Elastic actuators are specific type of soft actuators. Elastic actuators are mainly composed of elastic elements and rigid motors, which include series elastic actuators (SEA) [50], [51] and parallel elastic actuators (PEA) [52-54], as shown in Fig. 2 (a, b), and the energy consumption is reduced by 20% (PEA) and 78% (SEA) compared to rigid actuators [52]. SEA-driven exoskeletons are mostly rigid-soft hybrid structures. SEAs consist of a linear helical spring with constant stiffness in series with a stiff actuator, which own the advantages of lower impedance, higher energy efficiency, and shock isolation [51]. However, the adaptability is limited by the constant spring stiffness. Variable stiffness actuators (VSAs) can overcome the limitations by dynamically adjusting the intrinsic compliance [55], [56].

The components of cable-driven actuators are similar to SEAs, mainly including Bowden cables and other rigid actuators. The Bowden cable is composed of an inner cable and an outer sheath [57], which is mainly made of polyurethane and nylon [58-60]. When the motor exerts tension on the cable, even if the outer sheath is bent, the cable can slide inside outer sheath and transfer power. Bowden cables allow for the relocation of actuators to more advantageous positions [61]-[60]. The mobile exoskeletons usually transfer the actuators into the backpacks to reduce the inertia on the lower limb [59], [61], as shown in Fig. 2 (c, d). The treadmill-based exoskeletons lighten the weight of the exoskeleton by using Bowden cables to transfer power from external motors [58], [60], as shown in Fig. 2 (e, f). The motor, exoskeleton framework, and controller are the main sources of weight for cable-driven exoskeletons.

The motor can only produce tension, therefore, two cables are required for dorsiflexion and plantarflexion movement of the ankle [57]. The combination of pulley and Bowden cable reduces the issue of friction. The cable can be moved by rotating the pulley, and it is crucial to maintain tension on the cable to prevent it from slipping off the pulley [62]. The exoskeleton framework and Bowden cables are connected through anchor points, and Bowden cables are laid out along the load paths. The Bowden cable is stretched by the motor through the cable transmission mechanism, which imitates muscle movements to activate lower limb joints [57]. However, Bowden cables produce instantaneous relaxation intermittently, which places high demands on the control system, and Bowden cables should maintain constant tension while determining the optimal cable routing to maximize the range of motion [43]. The human-oriented control method, the load paths, the anchor points, and the cable transmission mechanism are the key considerations of the cable-driven structure.

PAMs are designed based on bionic principles, which can mimic the natural movements of skeletal muscles. PAMs are lightweight but can produce large contractile force, and PAMs are well-suited for HRI due to the intrinsic and adjustable compliance [63]-[69]. PAMs with different parameters can simulate the movement of antagonistic muscles [70]. Several

TABLE II.
EMERGING SOFT PNEUMATIC ARTIFICIAL MUSCLES

Study	Actuator	Size	Contraction ratio	Performance	Model	Feature
Fang et al. [63]	FPBAs	Rectangular sheets:120mm×110 mm;weight:91.84 g.	Maximum bending angle of 360°	25.74N·m at 40 kPa	Based on the virtual work principle	No airflow restriction occurs at any bending angle.
Paez et al. [64]	SPAs	Air chamber size：2mm×4mm×8mm; weight：16.07g	Maximum bending angle of 90°	600mN·m at 25kPa	Based on the mechanical behavior of the paper-folded vertex	Provide specific bending resilience throughout the actuator’s range of motion
Wirekoh et al. [65]	FPAMs	60mm×20mm×4 mm	Contraction of 22.4% ± 0.3%	Force of 38.1N ± 1N	A theoretical model assuming the actuator is composed of two flat rectangular composite sheets	Embed fibers and sensor elements to provide efficient control while minimizing the bulkiness
Yang et al. [66]	High-displacement PAMs	4cm×5cm×3cm	Contraction ratios of 60% at 34.5kPa	Force of 120N at 34.5kPa	Based on the geometry of the actuator	Integrate electronics to sense its pressure and displacement
Yang et al. [67]	VAMPs	34mm×28mm×46.5 mm	≈40% in vertical length, ≈5% in horizontal width	Power density peaks at ≈18.5W·kg^{-1}	Characterize two mechanical relationships under quasi-static conditions	Allow repair after damage and demonstrate a reasonable thermodynamic efficiency
Wirekoh et al. [68]	sFPAMs	55mm×26mm×5.5mm; weight:17 g	Maximum contraction of 19.8%± 0.2%	Maximum force of 24 ± 1.37N	A theoretical model using a modified von Karman formulation for large deformation of membranes and the energy method	Incorporate embedded sensors to provide real-time position and force feedback for muscles
Veale et al. [69]	Peano muscles	3 mm thick foam core wrapped in 0.5 mm silicone line	Contraction ratios of 15% at 30kPa	Force of 160N	Sarosi empirical dynamic model	Provide high force and compliance with the low threshold pressure

FPBAs = foldable pneumatic bending actuators, SPAs = soft pneumatic actuators, FPAMs = flat PAMs, VAMPs = vacuum-actuated muscle-inspired pneumatic muscles, sFPAMs = sensorized, flat, PAMs.

PAMs have been successfully commercialized, such as Fluidic Muscle from Festo (Germany), Air Muscle from Shadow (UK), and Rubbertuator from Bridgestone (Japan). However, when air pressure increases, McKibben muscles harden and expand radially, which cannot be mounted densely in exoskeletons and obtain redundant drive mechanisms [71]. The exoskeletons require flexible, compact, and lightweight PAMs to perform different functions. Bending PAMs such as FPBAs [63] and SPAs [64], can be incorporated into soft exoskeletons to assist ankle dorsiflexion and plantarflexion. Sheet-like muscles can fit tightly into the body to drive the exoskeletons. FPAMs [65] and Peano muscles [69], [72] can provide assistance from the soles to assist ankle dorsiflexion. Current designs fall short of multimodal proprioceptive sensing and effective real-time control. Self-sensing PAMs such as sFPAMs [68], integrate electronics to sense pressure and displacement, which provide exoskeletons with a continuous stream of sensory data. Perforations of PAMs are often caused by overpressure and wear, and pleated PAMs (PPAMs) [73] and VAMPs [67] can allow self-heal after damage. As summarized in TABLE II, PAMs are lightweight and flexible with high power-to-weight ratio, which can be mounted densely in exoskeletons [63]-[69].

Highly nonlinear and time-varying behaviors of PAMs make the modeling and control challenging, and analytical modeling methods can realize precise motion control [74], [75]. PAMs can be mathematically described by theoretical models or phenomenological models [76]. Theoretical models describe the relationship between the characteristics of PAMs and the geometry and material properties of PAMs [64]-[68]. The finite element method (FEM) can explore the various combinations of geometric features and material choices, which can create high-performing PAMs configurations and novel actuator shapes. However, the complex structure and numerous parameters required are hard to obtain, which restrict the application in real-time control [77]. The phenomenological models illustrate the relationship between the input and output of PAMs without considering the physical properties [63], [69]. The three-element model considers PAMs as a parallel arrangement of the spring, damping, and contractile elements, which can compensate for the uncertainties in the dynamics model [78], and artificial neural network (ANN) can also effectively solve the difficulties related to nonlinearity [79]. Muscle hysteresis can cause oscillation or degradation of the system, and an independent model for the hysteresis of PAMs is necessary [80].

Elastic actuators can realize lower impedance, higher energy efficiency, and shock isolation, but poor in adaptability, which can be used for patients in the early and middle stages of rehabilitation. Cable-driven actuators can lighten the device weight and achieve a close-fitting design, and PAMs can be embedded in exosuit and fiber material. Cable-driven actuators and PAMs place high demands on the control system, which are suitable for patients in the late stage of rehabilitation. Flexible elements and rigid motors can be combined to realize the rigid-flexible hybrid actuator to acquire better performance [50]-[60]. It is essential to design specific structures for different rehabilitation tasks, and different types of soft actuators can be combined to achieve more complete and complex functions.

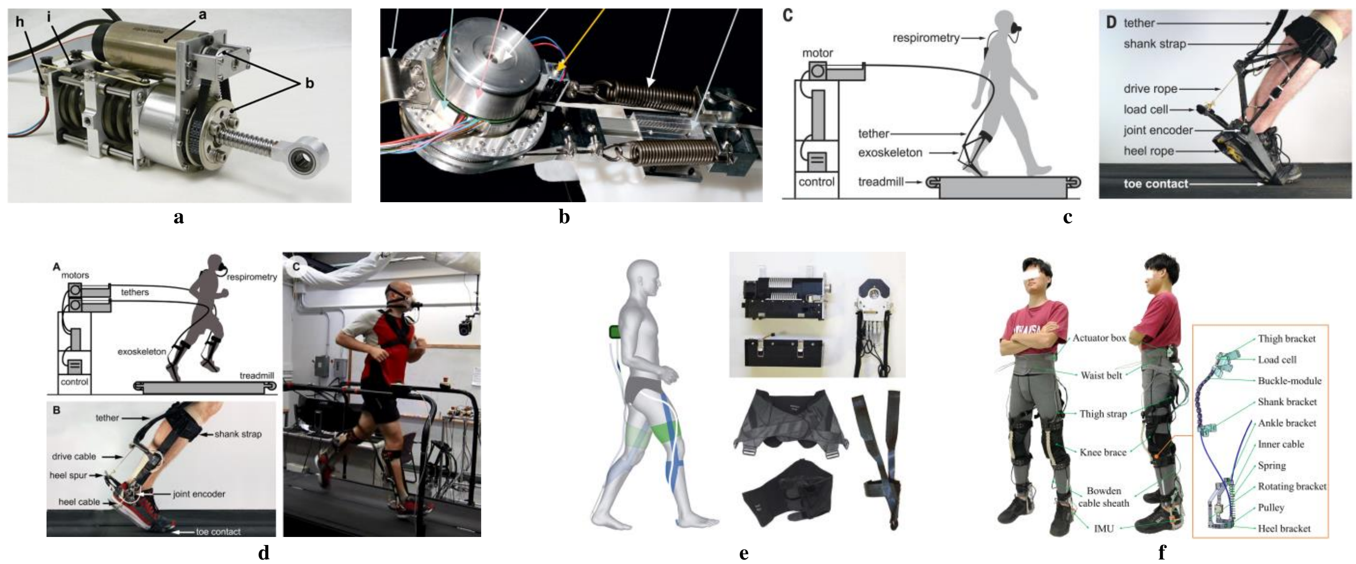

Fig. 2. Representative soft actuators: (a) SEA by Paine *et al.* [51], with permission from IEEE. (b) PEA by Penzlin *et al.* [53], an Open Access article with unrestricted use permission. (c) cable-driven actuator by Zhang *et al.* [60], with permission from Science. (d) cable-driven actuator by Witte *et al.* [58], with permission from Science Robotics. (e) cable-driven actuator by Panizzolo *et al.* [59], an Open Access article with unrestricted use permission. (f) cable-driven actuator by Ma *et al.* [81], with permission from IEEE.

B. Wearable Sensors

As a human-centered intelligent rehabilitation system, ankle-foot exoskeletons need to perceive the state of the human-robot coupling system and decode motion intention through the sensing system to realize human-robot coordinated control [13]. Therefore, the perception system needs to obtain human-robot state information and human-robot interaction information. Human-robot state information mainly includes kinematic information, kinetic information, and physiological information, which can be obtained from the pressure-sensitive insole, FSR [82]-[84], encoder, IMU [85], [86], load cell [87], [88], and EMG sensor [89]-[92]. Some representative commercial sensors are shown in Fig. 3 (a-f). HRI information mainly emphasizes the force and reaction relationship between the human and the exoskeleton, which can be obtained from force/torque sensors.

IMU is a sensor that combines accelerometer, gyroscope, and magnetometer [85], [86]. Multiple possibilities for sensor placement exist using IMUs, and appropriate sensor placement usually predigests or even eliminates calibration required for gait detection algorithms and obtains better intra-stride temporal gait features. Prasanth et al. reported the shank was the most widely preferred lower-limb placement for gait analysis, closely followed by the foot [85], and Niswander et al. suggested the algorithm would perform well when sensors were placed on the sacrum, lower anterior thigh, lower lateral shank, and heel [86]. Gait detection algorithm based on foot angular velocity or linear acceleration could be more accurate in detecting end contact (EC), toe-off, and initial contact (IC) [93]. In summary, the shank and foot can be the preferred locations for IMU placement.

Pressure-sensitive insoles and FSRs measure the pressure distribution of the foot, which are widely used to monitor foot-ground interaction events [82]-[84]. Different from IMUs, the sensor placement is relatively fixed for pressure-sensitive insoles and FSRs, which are always placed on the sole, and the toe, heel, first, and fifth metatarsals are the main mechanical monitoring points [82], [83]. Pressure-sensitive insoles allow for indoor and outdoor measurements, which is considered a substitute for force plates. Compared with real-time estimation of force magnitude, FSR is more used in detecting temporal information, such as the timing of force application. However, if the pressure-sensitive insole is arranged with too few mechanical monitoring points, sensor slippage and detachment can disrupt the measurements. The pressure-sensitive insole is turned to place as many sensors as possible in the insole to improve the resolution [84].

Surface electromyography (sEMG) signals are bioelectrical signals generated during muscle contraction and can be obtained on the skin surface in a non-invasive way, which can reflect the patient's physical conditions and motion intention [89]-[92]. sEMG has higher signal-noise ratio (SNR) and is well applied to the recognition and prediction of human motion intention [89]-[91]. Meanwhile, sEMG-based recognition performance is also limited by users' differences, electrode shifts/displacements, and muscle fatigue [92].

Load cells are often integrated into the joints or other load-bearing components of the exoskeletons, which can offer real-time feedback on the forces during exoskeleton assistance. Load cells primarily measure Bowden cable force [94], pneumatic muscle force [74], actuation force [95], and forces transmitted to the wearer [94]. Load cells are typically positioned between the motor and the end-effector, in series with the force transmission [94], [96], For example, load cells can be attached to the end of the PAMs, in line with the calf wrap and the Bowden cable, and connected to the actuator via the series elastic element. The sensitivity and accuracy of the load cell are determined according to the minimum and maximum force to be measured, usually in the 25kg or 50kg range.

E-skins inspired by human skin have stretchable, flexible, and soft properties, which have emerged as wearable sensors for real-time and continuous monitoring of physiological status

TABLE Ⅲ
WEARABLE SENSORS USED IN THE EXOSKELETON

Authors	Sensor type	Axes×Sensors	Location	Feature	Algorithm	Speed	Purpose	Performance
Lee et al. [105]	IMU	3(A)×2 (200Hz)	Thigh(1), torso(1)	Thigh position, thigh velocity, torso position, torso velocity	Stacked network architecture	0.5, 1.0, 1.5, 2.0 m/s	Gait phase classification (4 phases)	Average error is 1.67±1.36%(slow speed), 1.45 ± 1.47%(fast speed).
	FSR	1×1(200Hz)	The heel	Heel force				
Wu et al. [83]	Goniometer	1×2(1000Hz)	Hip and knee joints	Joint angle	GCNM	1.8km/h	Gait phase classification (4 phases)	Maximum accuracy is 97.43%
	FSR	1×1(1000Hz)	The heel	Heel force				
Gautam et al. [89]	EMG	1×4(1000 Hz)	VM, ST, BF, RF	sEMG signals	LRCN	NP	Locomotion mode recognition (3 modes), Joint trajectory prediction	Classification accuracy 98.1%(healthy), 92.4 %(pathology);
Parri et al. [84]	Sensorized insoles	64×1(100Hz)	The heel	GRF	Fuzzy-logic recognition method	Self-selected speed	Locomotion mode recognition (4 static modes, 3 dynamic modes)	Averaged accuracy is 99.4%
Pergolini et al. [82]	Encoders	1×1(100Hz)	Hip joint	Hip angles	LDA classifier	NP	Locomotion mode recognition (4 quasi-static modes, 4 dynamic modes)	Recognition accuracy is 94.8%
	Pressure-sensitive insole	16×1(100Hz)	The heel	GRF				
Ding et al. [106]	IMU	9(A, G, α) ×5(100 Hz)	Ankles(2), hips(2), chest(1)	(A, G, α) +linear A]	LSTM with attention mechanism	NP	Joint trajectory prediction	NRMSE is 9.06%(ankle) and 7.64%(hip)
Zhu et al. [90]	EMG	1×16(200 Hz)	Right leg	sEMG signals	CNN-LSTM with improved PCA	3km/h	Joint trajectory prediction	Prediction error is 1.34±0.25 deg
Xiong et al. [91]	EMG	1×10	Right leg	sEMG signals	ANN with Elastic Net	2, 3, 4, 5m/s	Joint Moment Prediction	NRMSE<7.89%, ρ>0.9633
	Angle sensor	1×5		Hip, knee, ankle angles				

NP = not provided, IMU = inertial measurement unit, EMG = electromyography, 9(A, G, α)×5 = 5 IMUs were used and for each IMU the 3D accelerations, the 3D rotational speed, and the 3D orientation were extracted, GCNM = graph convolutional network model, LRCN = long-term recurrent convolution network, PCA = principal components analysis, FSR = force sensitive resistors, GRF = ground reaction force, VM = vastus medialis, ST = semitendinosus, BF = biceps femoris, RF = rectus femoris, NRMSE = normalized root mean square error.

[97], [98]. The e-skin expected to be used in ankle-foot exoskeletons mainly includes pressure sensor [99], strain sensor [100]-[102], EMG sensor [32], [103], metabolite sensor [88], [104], and multifunctional sensor [34], as shown in Fig. 3 (g-i). The sensitive pressure sensor can map spatial interaction pressure distribution. Gong et al. developed a wearable sensitive pressure sensor, which could detect various forces including pressing, bending, and acoustic vibrations [99]. Strain sensors with high sensitivity and stretchability are important components in wearable electronics [100], [101]. Liu et al. designed a strain sensor to detect bending and stretching deformations with a wide sensing range, which was promising for the detection of human motion [102]. The conformal contact between the EMG sensor and the skin surface effectively enhances the sensor performance. Kwon et al. reported an EMG sensor made of biocompatible nanomaterial, which could realize high-fidelity EMG recording and recognize seven gestures through three-channel EMG signals with 98.6% accuracy [32]. Won et al. realized the classification of wrist flexion and extension through the EMG sensor for HMI [103]. Metabolite sensors monitor metabolism by detecting humidity [104] or glucose [88] in the skin. The multifunctional sensor is composed of the highly stretchable and conformable matrix network (SCMN) that can sense multi-functionalities, including but not limited to pressure, strain, humidity, and sEMG, which has broader applications in ankle-foot exoskeletons [34].

Sensor fusion can compensate for the shortcomings of the single sensor and support robust algorithms [86], [107]. Sensor placement, sensor resolution, and specific combinations of different sensors are important factors for wearable systems. E-skin can be incorporated into wearable electronics or electronic textiles to form conformal contact with irregular and deformable skin surfaces, which is suitable for personal information collection and health monitoring.

C. Mechanical Design

Ankle-foot exoskeletons can help patients in gait rehabilitation or power enhancement, which can be used as rehabilitation devices after stroke [108]-[110]. The soft ankle-foot exoskeleton can be mainly divided into calf module, foot module, and actuator module, which can be tightly fixed to the lower limb by separately fixing the calf and foot modules, as shown in Fig. 4. The actuator module is generally placed between the two modules, and the calf and foot modules are ergonomically designed to fit the human body tightly [54], [111]. Rigid exoskeletons adopt cumbersome rigid metal structures, which increase the metabolic cost of movement [21]-[29]. Meanwhile, rigid exoskeletons inevitably have joint misalignment problems, and the problem can be solved by using the soft structure instead of the rigid structure as the interface between the wearer and the robot. However, the level of assistance is limited due to the lack of load-bearing structures in most soft exoskeletons [75], [109]-[112]. Hence,

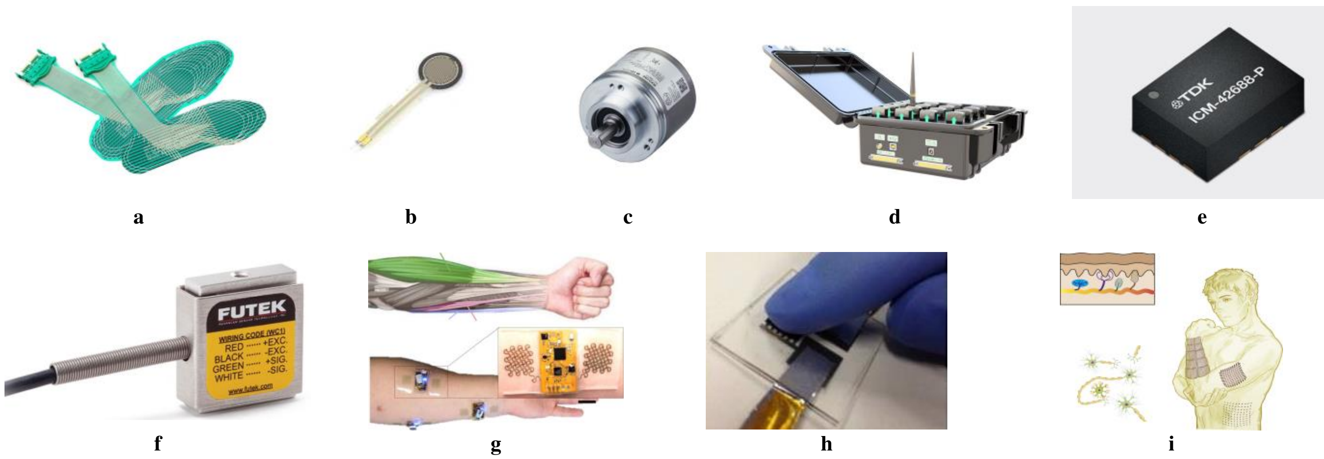

Fig. 3. Representative wearable sensors: (a) Tekscan by Tekscan, (b) FSR by Robodo, (c) Encoder by IFM, (d) EMG by Delsys, (e) IMU by TDK, (f) Load cell by FUTEK, (a), (b), (c), (d), (e), and (f) are reprinted from webpage with unrestricted use permission. (g) EMG sensor by Kwon *et al.* [32], an Open Access article with unrestricted use permission.(h) Pressure sensor Gong *et al.* [99], an Open Access article with unrestricted use permission. (i) multifunctional sensor by Hua *et al.* [34], an Open Access article with unrestricted use permission.

soft exoskeletons are more utilized in less demanding situations, such as power enhancement of healthy subjects or gait rehabilitation of patients with significant motor abilities. This paper focuses on soft unpowered exoskeleton, soft powered exoskeleton, and soft exosuit, as shown in Table IV, and the difference between soft exoskeleton and soft exosuit lies in the load-bearing structures [44].

Passive exoskeletons have orthopedic functions and can help the wearer by storing mechanical energy and subsequently returning it to the wearer without positive augmentation power [40]. Passive exoskeletons mainly rely on elastic elements to absorb and release energy. Elastic elements store the impact energy on IC, and release mechanical energy on EC to reduce energy consumption [108]. However, due to the fixed mechanical structure, passive exoskeletons have insufficient adaptability and are at best only switchable between locomotion modes [47].

Soft exoskeletons integrate soft actuators and lightweight load-bearing mechanisms, which have relatively better compliance and portability [12], [75], as shown in Fig.4 (a, c, f). Soft exoskeletons typically weigh less than 10kg, which are much lighter than rigid exoskeletons. The load-bearing structure of the soft exoskeleton is mainly made of lightweight materials, such as carbon fiber [54], polypropylene [74], and thermoplastic [111]. 3D printing technology can provide ergonomically comfortable interfaces and reduce the weight of exoskeletons while increasing the strength. The exoskeletons can be customized for each patient to ensure comfort [54], [74]. Park et al. designed an active soft orthotic which mimicked the morphology and functionality of muscle–tendon–ligament structures, and it can provide auxiliary force without restricting the natural DOF of the ankle. The auxiliary force could be transmitted to the ankle through the soft actuator, while the orthotic could be adjusted to provide a more natural gait pattern [75]. Kim et al. reported an AFO that could help drop-foot patients with DF assistance, and the DF angle was optimized by an average improvement of 12.3° [111]. The combination of AFO and elastic elements can reduce actuator torque and energy consumption. Orekhov et al made the exoskeleton closely match the desired torque in the early stance phase, and the actuator would generate momentum to counter against the torque produced by the elastomer spring, increasing 20–50% more possible auxiliary steps for the same battery capacity [54].

Exosuit structures are mainly composed of non-rigid components, such as PAMs and textiles, as shown in Fig.4 (b, d, e, g-i). Custom orthotic fabrication and malleable textile technologies are key techniques for soft exosuits, which determine the compatibility, functionality, and comfort of exosuit structures [59], [109]-[112]. Without considering the external power supply, the weight of soft exosuit is extremely low, even less than 1kg [59], [75], [108]. PAMs and cable-driven actuators are the main actuators of soft exosuit. Wehner et al. developed a soft exosuit made of custom McKibben PAMs, and the PAMs were attached to the exosuit via inextensible webbing. The virtual anchor technique was designed to transfer auxiliary forces to locations which could absorb load, and the exosuit could be tuned truly to be the transparent suit when not in use [109]. Soft robotic boots such as ExoBoot [110] and smart anklet [112] can be applied to exoskeleton assistance. Chung et al. integrated the soft textile-based actuator and wearable sensor into ExoBoot without compromising movement [110].

Drop-foot patients vary widely, thus exoskeletons should be customized for individuals. On the other hand, patients with different conditions should be able to select rehabilitation robots with corresponding DOFs and configurations. The principle of modularization is to adopt appropriate modules according to different stages of patients to achieve structural restructuring [113], [114]. The symbiotron exoskeleton developed by Meineke et al. could be personalized to accommodate differences in injuries between individuals, and the personalization was achieved through modular mechanism that allowed the reconfiguration of the exoskeleton [115].

Rigid-flexible-soft hybrid structure, customized personalized structure, and modular structure are important technologies for the mechanism design of the ankle-foot exoskeleton. The ergonomically comfortable interface made by 3D printing and lightweight materials can reduce resistance and skin abrasion. The focus of HIL optimization lies in the selection of cost function, and the establishment of the

TABLE IV
MECHANICAL STRUCTURES OF TYPICAL ANKLE-FOOT EXOSKELETONS

Research	Actuator	Sensor	Component	DOF/Weight	Output force/torque	Metabolic power	Feature
Collins et al. [108]—Unpowered exoskeleton	NA (elastomer spring)	No electronics	Comprise rigid sections attached to the human shank and foot	P/ 0.057 kg	Provide a maximum auxiliary torque of about 0.4N·m·kg^{-1}	Reduce the metabolic cost of walking by 7.2±2.6%	Lightweight composite frame with a lever about the ankle
Sawicki et al. [74]—KAFO	PAM	Load cells	A polypropylene foot section, a carbon fiber shank, and thigh	DP/ 2.9±1.3kg	A nearly constant force over the stride, peaking at 472 ± 147 N	The orthosis peak positive power is 1.88 ± 0.28 W/kg	Custom molded from a cast unique to each subject
Kim et al. [111]—AFO	PAM	Magnetic encoder, GRF	A metallic slider crank mechanism, custom thermoplastic braces	D(20°) P(45°)/ 2.9kg	Provide a maximum assistive torque of 9.8 N·m at a functional frequency of 1 Hz	The current mechanical output is approximately 8W	User safety can be guaranteed despite possible failures
Orekhov et al. [54]—Parallel-elastic exoskeleton	PEA	FSR, torque sensor	Carbon fiber leaf spring, aluminum pulley bridge	DP/ 2.4-2.6 kg	Produced 10–15 N·m peak assistive torque at all walking speeds	The difference between the no spring and spring controller conditions is −0.04 ± 0.13 W/kg	Customized for each person
Park et al. [75]—Active soft orthotic	PAM	IMU,	Anchors on the foot brace, thin non-slip silicone pads	D(14°) P(13°), IE/ 500g	110N·m, 53N·m, 20 N·m, and 21N·mfor dorsiflexion, plantarflexion, inversion, and eversion,	NA	Agonist–antagonist muscle architecture
Kwon et al. [12]—AFO	Bi-directional tendon-driven actuation	GRF, IMU, soft strain sensor	3D-printed flexible brace and an ankle support	DP/ 1540g	The maximum assistive force of 70N in plantarflexion	NA	Provide support in vertical direction preventing the structure from buckling
Panizzolo et al. [59]—Soft exosuit	Cable-driven actuators	EMG, load cells	Mainly use textiles	P/ 450g	272 ± 43 N of ankle plantarflexion 204 ± 32 N of hip flexion, and 68 ± 24 N of hip extension	EXO_ON (7.5±0.6W·kg^{-1}) is 7.3±5.0% and 14.2±6.1% lower than EXO_OFF_EMR and EXO_OFF	The exosuit is truly transparent when the cables go slack.
Wehner et al. [109]—Soft Exosuit	Custom McKibben PAM	Force sensor	A network of soft, inextensible webbing	DP/ 9121g	The maximum output force of 235N	43.9 W or 10.2% reduction in average metabolic power	Suit can be tuned to the passive unpowered suit and virtual anchor technique
Siviy et al. [96]—Soft Exosuit	Cable-driven actuators	Integrated sensor	Textile components, custom fabricated waist belt, and calf wrap	DP/ 4932.2g	PF assistance is 8.29±1.75 N in P+ and 6.55±1.22 N in P−	P− deliver 3.77±2.13× 10^{-3} Wkg^{-1}, P+ deliver −3.27±1.18×10^{-3} Wkg^{-1}	Offline assistance optimization can isolate P+ and P−
Chung et al. [110]—ExoBoot	PAM	IMU	Soft textile	DP/ 255g	Maximum torque applied on the ankle is 23N·m.	Generate about 45% of the power during gait	Actuator and IMU are integrated into ExoBoot

AFO=ankle foot orthoses, KAFO=knee-ankle-foot orthosis, AFE=ankle-foot exoskeleton, DOF=degrees of freedom, EXO_ON=with the device powered, EXO_OFF=with the device unpowered, EXO_OFF_EMR=with the device unpowered with equivalent mass removed, P−=negative augmentation power, P+=positive augmentation power, PF=plantarflexion, DF=dorsiflexion, DP=dorsiflexion/plantarflexion, IE=inversion/eversion

rigid–flexible coupling model that conforms to human musculoskeletal system and human–robot coupling model that conforms to robot dynamics model can effectively improve the time efficiency of HIL optimization.

IV. CONTROL OF SOFT EXOSKELETONS FOR REHABILITATION

A. Rehabilitation Strategies and Training Modes

The appropriate training modes should be formulated based on the therapist's judgement and the patient's level of recovery [116]. In the early stage of rehabilitation, the main purpose is to stimulate muscle activity, promote blood circulation, and reduce spasm. Passive mode tends to be adopted at this stage, and patients can follow robot to perform periodic reciprocating movement [117], [118]. In the middle stage, the affected extremity can be trained to improve the coordination of the joints and gradually restore motor ability. This stage adopts training mode from passive to partially assisted, and then to active [119]. The assistive mode is to provide appropriate robotic assistance when the patient intends to move but without sufficient strength. Resistive mode means that when the patient voluntarily exercises, the robot provides appropriate resistance to help the patient improve muscle strength. In the late stage, patients mainly improve various activities of daily living (ADL) capabilities [8], and this stage is dominated by resistive mode.

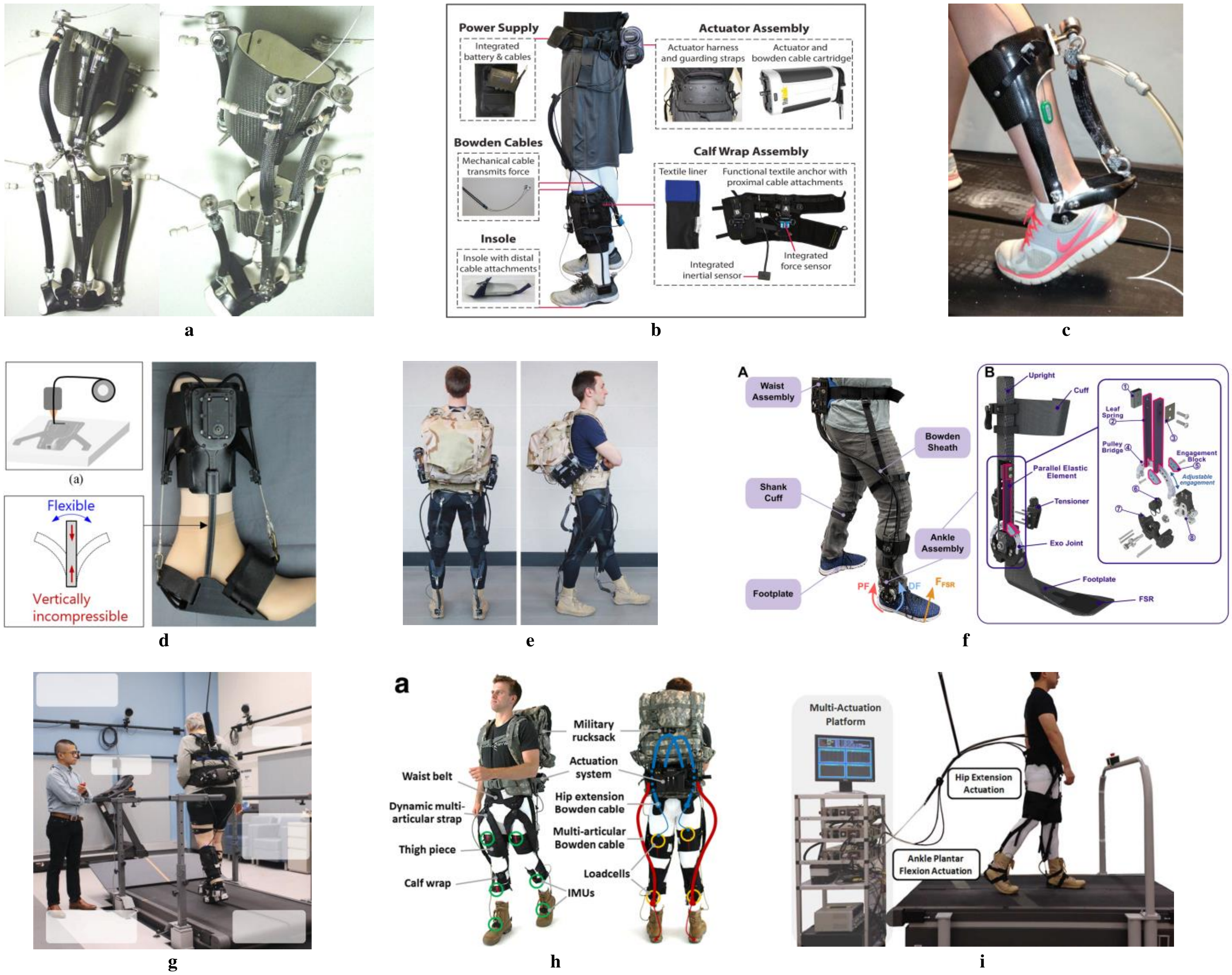

Fig. 4. Representative soft ankle-foot exoskeletons: (a) KAFO by Sawicki *et al.* [74], an Open Access article with unrestricted use permission.(b) AFO by Awad *et al.* [145], an Open Access article with unrestricted use permission. (c) PAM-driven exoskeleton by Takahashi *et al.* [146], an Open Access article with unrestricted use permission. (d) ankle-foot-orthosis by Kwon *et al.* [12], with permission from IEEE. (e) cable-driven exosuit by Panizzolo et al [59], an Open Access article with unrestricted use permission. (f) parallel-elastic exoskeleton by Orekhov *et al.* [54], with permission from IEEE. (g) cable-driven exosuit by Siviy *et al.* [96], with permission from IEEE. (h) cable-driven exosuit by Lee et al. [94], an Open Access article with unrestricted use permission. (i) cable-driven exosuit by Ding *et al.* [147], with permission from IEEE.

Leia robot is a typical example of applying passive mode and active mode. In the passive mode, the therapist sets the exercise time, number of repetitions, and exercise intensity, and the robot performs a repetitive flexion/extension movement. In active mode, the therapist selects two specific sEMG thresholds: the minimum threshold for basic exercise and the maximum threshold for intensive exercise, and the back-and-forth training mode can help the patient to recover effectively and avoid secondary injuries [120].

B. Low-level control

Hierarchical control enables compliant HRI, which includes high-level [35], [121]-[128], mid-level [129]-[131], and low-level control [132]-[142], as shown in TABLE V. The high-level controller perceives the patient's motion intention, consisting of locomotion mode recognition or human-robot interaction control. The mid-level controller translates motion intention to the desired trajectory. The low-level controller tracks the desired trajectory with high accuracy [143], [144]. For example, wang et al. [124] proposed an adaptive interaction torque-based assist-as-needed (AITAAN) control, which included assist-as-needed control (high-level control), computed torque control (mid-level control), and prescribed performance control (low-level control), which provides necessary torque to promote the wearer's active engagement.

The low-level controller calculates the error between the current and desired state of the robot and adjusts the actuator to reduce trajectory tracking error, and trajectory parameters such as velocity, amplitude, and number of repetitions can be set by therapists. Low-level control mainly includes PID control [132]-[134], fuzzy control [135], [136], adaptive control [137], [138], sliding-mode control [139], [140], iterative learning control [141], [142]. PID control is commonly used, but not suitable for robots with changing parameters. The PID method is usually used in conjunction with other methods [133]. Due to its ability to adapt to object changes and external disturbances,

TABLE V
TYPICAL EXOSKELETON ROBOT MOTION CONTROL METHODS

Control Strategy		Features	Advantage	Disadvantage	Research
Low-level control	PID control	Controlled by linear combination of proportional, integral and derivative.	Simple structure, no modeling is required	Difficult to adapt to complex system	Liu et al. [132], Amiri et al. [133], Bai et al. [134]
	Adaptive control	Adapt to the changes of controlled object and external disturbances.	Real-time identification of system parameter changes	Identifying model parameters online requires huge amount of calculations.	Liu et al. [132], Victor et al. [138], Guerrero et al. [137]
	Fuzzy Control	Control the system with fuzzy control theory	Simplify system design without over-reliance on mathematical models.	Difficult to control complex systems.	Yin et al. [135], Huo et al. [136]
	Sliding mode control	Forcing system to follow the predefined reference trajectories	Fast response, insensitive to disturbances, no online system identification required	Easy to generate shake when the trajectory reaches the sliding mode surface.	Ai et al. [139], Taheri et al. [140]
	Iterative learning control	Continuously iterate, repeat and learn to improve system performance	Don't depend on the precise mathematical model and track predefined trajectory with high accuracy	Unable to provide perfect tracking under any circumstances, noise and non-repeating interference hinder performance.	Kirby et al.[141], Meng et al. [142]
Mid-level control	Finite-state control	Periodic activities are described by a series of different states	The exploding dimensionality of the tunable parameter space	Cannot handle dynamic changes that are not included in the preset states.	Hayami et al. [129] Alberto et al. [130]
	Virtual constraint control	Use output feedback linearization to track the optimized time-invariant trajectory.	Guarantee stability along the trajectory, self-balance without crutches	Use only generic, normalized shape parameters	Gregg et al. [131]
	Impedance /admittance control	Achieve stable position and force control by tracking the target impedance model	Strong robustness against disturbances, few offline task planning	Difficult to set appropriate impedance. If the setting is too rigid, the patient feels passive movement.	Veneman et al[126], Nagarajana et al.[127], Baser et al. [128]
High-level control	EMG-based control	Collect the sEMG signals to decode the movement status or motion intention	Non-intrusive, easy to obtain, highly operable, and highly secure.	Difficult to use in stroke patients with severe muscle atrophy.	Zhuang et al. [35], Kai et al. [121]
	EEG-based control	Obtain the patient's consciousness signal and convert it into control signal	Direct volitional control, patients can express motion intention through motor imagination	Low signal-to-noise ratio, strong non-stationary randomness, the use of BCI requires concentration and causes fatigue.	Ai et al. [122]
	Assist-as-needed control	Provide robotic assistance based on patient needs to maximize patient motivation	Maximize voluntary patient contributions	Need to cooperate with other control strategies	Tatsuya et al. [123], Wang et al. [124]. Dao et al. [125]
	Human-in-the-loop control	The device control is systematically adjusted during use to enhance human performance.	Automatic discovery, personalization, and continuous adaptation of assistance.	Lengthy evaluation periods, strong history dependence, and slow components of adaptation	Kim et al. [148] Witte et al. [149] Han et al. [150]

BCI = Brain-Computer Interface, EEG = Electroencephalogram,

adaptive control (AC) is widely applied to exoskeletons [137], [138]. Liu et al. proposed an adaptive PID controller for the lower extremity exoskeleton, which could suppress the undesired chattering of PID gains, and adaptively adjust PID coefficients when internal or external disturbances were encountered [132]. External disturbances such as abnormal muscular torque and ground impact may limit AC, since errors caused by interference may affect the adaptive performance and system parameter identification [137]. Fuzzy control without over-reliance on mathematical models has good adaptability to exoskeletons driven by PAMs [135]. Huo et al. proposed a fast gait detection method, which could dynamically estimate gait patterns in real time. The method could obtain the auxiliary torque required for human movement in a natural way [136]. The combination of adaptive control and fuzzy control can provide better personalized assistance. Yin et al. adopted an adaptive fuzzy strategy to control ankle-foot exoskeleton, which could adjust the rule base based on feedback to meet the needs of personalized motion assistance [135].

Sliding mode control (SMC) has the fine ability to suppress parameter disturbance and external interference. However, the discontinuous switching of the controller can easily cause high-frequency vibration [137]. SMC can also solve the non-linear characteristics of the actuator and the uncertainty in HRI. Ai et al. proposed a disturbance-estimated adaptive backstepping sliding mode control (ABS-SMC) to adjust the output of the robot to adapt to external changes, and the external interference of the robot could be estimated by the observer [139]. Iterative learning control (ILC) can learn and adjust controller parameters from repeated training scenarios without knowledge of the actual system [141]. Meng et al. proposed a multiple degrees-of-freedom normalized robust iterative feedback tuning (IFT) technique which improved the robustness by obtaining the optimal value of the weighting factor and provided the strategy with the learning ability to obtain the optimal controller parameters [142]. To support complex robot motions, a combination of control methods can be implemented, in which multiple methods can eliminate the disadvantages brought by the single method.

TABLE VI
TYPICAL CLINICAL TRIALS FOR EXOSKELETON

Research	Goal	Subjects	Time	Clinical scales	Evaluation index
Louie et al. [156]	Determine the efficacy of exoskeleton-based therapy training	Patients with subacute stroke (<3 months)	6 months 3 times/week	Distance walked on the 6-MWT, days to achieve unassisted gait, FMA, BBS	Gait speed, endurance, balance
Wall et al. [157]	Evaluate self-perceived functioning, disability and recovery after rehabilitation	Stroke patients aged 18–67 years	6 months 4 days/week	SIS	Walking speed /endurance and balance, ADL ability
Calabrò et al. [158]	Evaluate the effects of Ekso™	Stroke patients aged ≥55 years	45-min/session, 5 times/week	Distance walked on the 10-MWT, MAS	Gait cycle, EMG, EEG, CSE, SMI

MWT = minute walk test, FMA = fugl-meyer assessment, BBS = berg balance scale, ADL = activities of daily living, SIS = stroke impact scale, CSE = cortico-spinal excitability, SMI = sensory motor integration (SMI), MAS = modified ashworth scale.

C. Mid-level control

The mid-level controller translates motion intention obtained by the high-level controller into the desired trajectory required by the lower-level controller [143]. Finite-state controllers (FSCs) are the overwhelming choice of mid-level controllers, mainly due to conceptual operability and ease of implementation [129], [130]. FSCs are commonly used for state transition, and FSCs implement a set of discrete parametric control laws, which can cycle through each gait phase. However, the parameter value of FSCs is fixed, and the device may not be able to accommodate the patient's gait changes optimally due to fatigue, sudden changes in weight or gait patterns [143]. More importantly, FSCs cannot handle dynamic changes that are not included in the preset states. A four-phase model is most commonly adopted, including the initial foot contact (IC) or heel strike (HS), flat foot (FF), heel-off (HO), the initial swing phase (SP), or toe-off (TO). The ankle-foot exoskeleton can prevent foot slap in HS, minimize the exerted impedance during FF, and lift the patient's foot in SP. Similar to the energy shaping control method, the virtual constraint-based control applied to exoskeletons can effectively support patient training [131]. The hybrid dynamics model is established by applying a switching surface to continuous dynamics. When the pre-collision state encounters the switching surface, it can transit to the post-collision state through a pre-calculated reset mapping [151].

Impedance or admittance control can alter the dynamic response of the exoskeleton and encourage voluntary muscle recruitment [128], [152]-[154]. Veneman et al. designed an impedance controller including both low impedance and high impedance control modes, also known as "patient-in-charge" and "robot-in-charge" modes [126]. Nagarajana et al. presented the integral admittance shaping framework, which could determine the dynamic response of the human-robot coupling system and synthesize with fast response [127]. Joint impedance may increase due to the mass and friction of the exoskeleton, the impedance reduction control methods are necessary [152]. It is necessary to adjust the parameters of the impedance model online according to the change of the contact force, which can compensate for the dynamic change of the environment in real time [153], [154]. Liu et al. proposed the cascade impedance control (CIC) method, which adopted the exoskeleton torque and sEMG signals as input, and the variable impedance as output. the proposed algorithm can effectively decreased the impedance of the exoskeleton when detecting the spasticity from patients [153].

D. High-level control

The high-level controller decodes the patient's motion intention, consisting of locomotion mode recognition and HRI control. Effective HRI can avoid secondary injuries and achieve compliant control [121]-[128]. sEMG signals can be decoded through feature extraction techniques including signal processing, dimensionality reduction, pattern classification, and other techniques [155]. sEMG can be also mapped to joint angle or voluntary joint torque without overcoming gravity and inertia of the lower limb. Zhuang et al. proposed an EMG-based admittance controller (EAC) to improve human-robot synchronization, which could reduce the delay between the patient's voluntary torque and the exoskeleton's assistive torque [35]. However, sEMG signals are difficult to apply to patients who suffer severe muscular dystrophy, such as patients with amyotrophic lateral sclerosis (ALS), multiple sclerosis (MS), and stroke [121]. Electrode placement and muscle contraction movements can be adjusted under the guidance of a professional rehabilitation physician, and EMG signals can be collected from the mirror limb or the neighbouring limb for muscles that are almost impossible to collect EMG signals. EEG-based control can solve these problems. Most EEG-based exoskeletons resort to pattern recognition to decipher the EEG signals into discrete classes. Ai et al. proposed a collaborative control strategy based on motor imagery to control an ankle-foot robot for patients with impaired nerve conduction [122]. Research in EEG-based control mainly focuses on offline classification and regression analysis, but the practical applications and experimental results of interactive control based on EEG signals are scarce.

Assist-as-needed (AAN) control can assist patients on demand, which can stimulate patients' maximum voluntary contribution by minimizing the robot assistance to complete specific tasks [19], [125]. Tatsuya et al. proposed a novel AAN controller based on a model predictive control (MPC) method, in which the exoskeleton only provided the deficient torque to follow the target trajectories based on the patient's voluntary movements [123]. Wang et al. proposed an adaptive interaction torque-based assist-as-needed (AITAAN) control in which the exoskeleton provided residual torque through interaction, and the interaction torque tracking problem could be transformed into the trajectory tracking problem to achieve precise torque tracking [124]. Compared to biomedical signals, force signals are more deterministic and intuitive.

Motion pattern recognition enables the high-level controller to switch between mid-level controllers appropriate for various gait phases [83], [105] or locomotion modes [82], [84], [89], as shown in TABLE III. The motion intention can be obtained by decoding physiological signals or predicting joint motion information [90], [91], [106]. Decoding physiological signals can endow the exoskeleton with the intelligence to understand the motion intention, thus avoiding the control delay of the exoskeleton. However, it is worth noting that the sensors are usually placed at the joint connection, and the measured parameters can vary with the wearing state and body reaction, which may lead to prediction error [92]. Gautam et al. proposed the LRCN framework to classify locomotion mode through sEMG signals, including walking, standing, and sitting, and the average classification accuracy of healthy subjects and patients was 8.1 % and 9.2 % respectively [89]. Natural gait planning aims to simulate the movement of healthy people. Predefined trajectories have some limitations when applied to exoskeletons, such as limited adaptability to different individuals and environments and limited system stability [159]. The deep learning methods have been applied to customized and personalized gait trajectory generation to realize motion integration between patients and robots [160]-[162]. Common learning-based methods include back propagation neural network (BPNN) [163], convolutional neural network (CNN) [164], and recurrent neural network (RNN) [165]. Several algorithms are integrated to overcome the limitations of each algorithm, such as LSTM-CNN [166] and CNN-LSTM [167]. Zhu et al. proposed CNN-LSTM with an improved PCA algorithm, which could extract the principal components more effectively, and the CNN-LSTM algorithm possessed the best prediction accuracy compared to the single algorithm [90].

In recent years, research on reducing energy consumption and cost is growing. Panizzolo et al. designed a soft lower extremity exosuit, which transmitted controlled force to the wearer through Bowden cable. Force-based position control adopted the predefined gait trajectory as the function of the gait period to achieve specific force distribution, and the experimental result demonstrated that net metabolic power in the EXO_ON condition was 7.3±5.0% lower than in the EXO_OFF_EMR condition [59]. Wehner et al. found that the effect on kinematics was greatest when the actuator was turned on at 10% and 60% of the gait period, whereas the effect was minimal at 30%, and average metabolic power could be reduced by 10.2% [109]. As shown in Table IV, the soft ankle-foot exoskeleton can realize about 10% reduction in average metabolic power. The human-in-the-loop (HIL) control can automatically and iteratively generate assistance patterns of the exoskeleton, which significantly improves the walking economy. The auxiliary curve of the ankle is parameterized, and the assist torque is modified through energy consumption while patients use exoskeletons [148]-[149]. Appropriate assistance patterns and cost functions can substantially improve the time efficiency of HIL optimization. Zhang et al. adopted the covariance matrix adaptive evolution method to optimize the parameters of the auxiliary curve, and the auxiliary curve iteratively approached the optimum [169]. Han et al. constructed the cost function based on muscle activity and found the appropriate initial assistance mode in HIL optimization [150].

Ankle-foot exoskeletons need to continuously perceive patient's requirement and intention. EMG and EEG signals can respond quickly to the intentions, and machine learning methods combined with wearable sensors can extract more features and achieve higher accuracy. HRI control can significantly improve the level of active participation and rehabilitation, and multiple feedback can be used to achieve patient-in-charge rehabilitation. Safety and stability are critical for mid- and low-level controls. Customized and personalized gait trajectory generation can adapt to different patients, which can achieve coordinated motions between patient and robot. Furthermore, the integrated control strategy can support complex movements and dynamic human behavior.

E. Clinical Trials

To verify the effectiveness of exoskeleton-based physiotherapy training, it is necessary to complete clinical verification of ankle-foot exoskeletons [170], as shown in TABLE VI. Commercial exoskeletons such as Rewalk or Ekso™, have completed many clinical trials, such as safety, reliability, feasibility, and tolerability tests [145], posture and balance tests [171], gait training tests [172], energy efficiency and patient satisfaction tests [173]. Clinical scales can better evaluate the rehabilitation level of patients, including FMA, BBS, SIS, and MAS [156]-[158]. Some prototype exoskeletons in the laboratory stage have also been tested on patients. These tests are usually set up according to therapist guidance, which typically includes the metabolic cost of walking, lower limb muscle activation, joint kinematics, and gait parameters measured during different assisted walking trials [174], Compared with commercial exoskeletons, clinical trials on laboratory exoskeletons are inadequate. The ankle-foot exoskeletons should recruit healthy subjects for preliminary validation [175]. Although the experimental results are in line with expectations, there are some physiological differences between patients and healthy subjects. Importantly, incorporating participant feedback into the design and refinement of the exoskeleton can make the device more suitable for patients [176].

V. Discussion

Uncontrolled foot drop and foot inversion are among the most severe symptoms in stroke patients. Ankle-foot exoskeletons have shown the rehabilitation efficacy [177], [178]. This review focuses on the key and state-of-the-art technologies of the soft ankle-foot exoskeleton. The robot actuation, wearable sensor, mechanical design, control strategy, and clinical trial of the soft ankle-foot exoskeleton have been analyzed and discussed in detail, each of which is presented with the tabulated information of representative technologies. Aspects such as the size, contraction ratio, performance, and analytical modeling method of the emerging soft pneumatic artificial muscles are explained, and current developments in soft actuators include the novel materials, actuator-sensor integration, multi-function integration, and innovative designs based on bionic principles. The sensor type, sensor placement, adopted algorithm, and performance are also investigated. Multi-functional skin-like and textile-based sensors have the potential to enhance the comfort and wearability of soft ankle-foot exoskeletons. Materials, manufacturing methods, output force/torque, and energy consumption of exoskeletons are the main factors in the design of exoskeletons. Soft exoskeletons are evolving rapidly towards being lighter and modularized, more adaptable, and more intelligent. Appropriate rehabilitation strategies combined with control methods should be chosen for patients at different stages of rehabilitation to help them recover more efficiently [179]. Control algorithms are evolving towards data-driven methods, intelligent control, and human-oriented control to achieve better human-robot interaction.

Research on soft actuators is primarily based on cable-driven actuators and PAMs [180]. Bowden cables can transfer the actuator from the lower limb to the backpack or external device. The main goals of cabled actuators are firm attachment and steady force transmission, thus the force transmission path and timing of assistance should be carefully considered. As bio-inspired actuators, the control accuracy and response characteristics of PAMs still require significant improvement to mimic skeletal muscles, and the utilisation of high-performance materials and optimised design of pneumatic muscles can reduce energy loss and enhance response time. Innovative PAMs have been developed to remedy the shortcomings of McKibben muscles, such as bending PAMs [63], self-sensing PAMs [68], and Peano muscles [69], [72]. The innovative PAMs based on flexible materials can fit tightly to the body and provide auxiliary force from the soles or other specific positions. Soft actuators require structures designed for specific tasks, and different types of soft actuators can be integrated to achieve more complete and complex functions. However, most PAMs cannot support hours of training due to energy limitations, which pose challenges for clinical rehabilitation. Therefore, it is inevitable to exploit cost-effective power sources with high storage capacity and light weight, such as solid-state hydrogen storage [181] and liquid carbon dioxide containers [182]. The sensor placement may affect the completeness and quality of collected data. IMUs are commonly placed on the shank and foot, while the toe, heel, first, and fifth metatarsals can be the preferred location for pressure-sensitive insoles and FSRs [82], [83]. Environmental sensing sensors can be integrated into the exoskeleton to sense changes in terrain and obstacles. Many traditional sensors are not compatible with wearable robots. Fabric and flexible materials are well-suited to the irregular structure of the exoskeleton and human skin, and can be incorporated into textile-based and skin-like sensors [88]. The e-skins are compact and customizable, and can be mounted on the uneven skin surface. Lightweight, flexible, integrated, multi-functional, and multi-array sensors, while improving the resistance to sensor slippage and interface interference may become the future trend [84].

Soft ankle-foot exoskeleton should be compact in structure, light in weight, and high in efficiency of force transmission [12]. There is a human-robot matching problem between rigid robots and the human body, which may cause skin abrasions. The soft exoskeletons are a viable solution. However, due to the lack of load-bearing structure of the soft exoskeleton, the auxiliary force/torque is limited. The rigid-flexible-soft hybrid mechanism can well integrate the advantages of each structure, and can be realized by adopting advanced materials and manufacturing technologies, such as soft textiles [59], [96], [110], carbon fibers [54], polypropylene [74], and other 3D printed materials [12], [108]. Modularization can select corresponding modules in accordance with the recovery stage and rehabilitation demands to realize structural restructuring. Meanwhile, 3D printing can enable ergonomic design to achieve humanoid structures [12], and the combination of the two technologies can improve the applicability and utilization of the equipment. Optimizing the walking economy has been a research hotspot. Passive exoskeletons can store mechanical energy and subsequently return it to the user when assistance is needed, but they lack adaptability [108]. Powered exoskeletons rapidly measure the metabolic profile, and regulate the torque profile to reduce energy consumption [96], [109]. The combination of active and passive components can provide assistance and minimize the energy consumption. Appropriate assistance patterns and cost functions can substantially improve the time efficiency of HIL optimization [150]. On the other hand, the magnitude of peak torque and the timing of torque onset varied across patients in terms of HIL control parameters. Generically optimized assistance has a wider range of applications, and individually optimized assistance can lead to greater improvements in energy consumption. Thus, the optimization method needs to be determined according to specific tasks [148].

The three-layer structure of hierarchical control is similar to the structure of the central nervous system, which is an effective way to achieve controlled normal locomotion [45]. Physiological signals combined with wearable sensors can quickly identify the motion intention. The mid-level and low-level control enable safe and coordinated movement for the patient. Decoding the patient's motion intention, user-friendly HRI, and personalized gait trajectory generation are important aspects of providing optimal assistance. In terms of improving the accuracy of intention decoding, sensor fusion and deep learning methods may be of interest. Patients vary greatly, and AAN control can provide the corresponding auxiliary force on demand [184]. The important problem of AAN control is determining the patient's motion state and intention accurately and providing appropriate assistance force accordingly. Machine learning algorithms bring intelligence, but also raise privacy concerns, and it is important to obtain approval from the ethics committee. Additionally, high-level decisions in intelligent control should be determined by the physician to avoid algorithm's misjudgements. Furthermore, customized and personalized gait patterns can be combined to realize motion integration between the exoskeleton and the user.

Patients require gradual muscle strength recovery through long-term rehabilitation training, and prolonged interaction is likely to cause skin abrasions. Therefore, it is necessary to reduce the transparency of robots and enable friendly human-robot interaction, and soft exosuits have natural advantage in this aspect. The compliance of the robot is another important factor to consider, and high compliance can prevent greater damage to patients during spasms and unplanned movements. Serious games offer avenues and opportunities for customized and contextualized gameplay, which can provide sustained motivation for rehabilitation [185]. Interactive technologies such as virtual reality (VR) and augmented reality (AR) can be used for serious games, delivering feedback in multiple forms, such as voice, vision, and touch, which can enhance the experience and participation of the patient [186]. More importantly, user-centered design principles of serious games can contribute to better acceptance and outcomes. The patient's active participation can enhance the rehabilitation outcomes. Friendly human-robot interaction, the transparency and compliance of the robots, attractive rehabilitation games and personalized training can contribute to the active willingness. The absence of uniform evaluation criteria for the experimental results creates challenges when comparing performance across various articles. Additionally, the limitations and potential improvements of the method are not thoroughly discussed. Assessing the patient's rehabilitation progress through professional rehabilitation assessment standards, modifying the difficulty level of serious games, and incorporating sensory feedback can further encourage the user’s voluntary contribution [187].

VI. Conclusion

Ankle-foot exoskeletons have attracted increasing attention over the last decade. It can be predicted that the application potential of soft ankle-foot exoskeletons and intelligent robotic assistant systems is huge. The commercial exoskeleton still needs to be optimized in actuators, structures, and control methods, and the soft ankle-foot exoskeleton can partially compensate for its deficiency. This article reviews the ankle-foot exoskeleton robot from soft actuator, mechanical structure, wearable sensor, hierarchical control, and clinical trial, which has realized the expansion from point to area and the transition from laboratory to practical application. Flexible elements and rigid motors can be integrated to achieve the rigid-flexible hybrid actuator. Soft actuators can be shaped for specific tasks, and different actuators can cooperate to complete more complex functions. The rigid-flexible-soft hybrid structure, customized and personalized structure, and modular structure can improve adaptability and comfort for different patients. Ergonomically comfortable interface and energy consumption reduction are also important factors for the soft ankle-foot exoskeleton. Sensor fusion and the integrated control strategy result in better performance, which can be widely used in human-robot environments. Serious games may provide sustained motivation for rehabilitation. More and more evidence shows the effectiveness of robotic rehabilitation, and clinical trials can provide scientific proof, which is an indispensable part. The next-generation exoskeletons to facilitate patient’s daily lives can require further collaboration between engineers, clinicians, and patients.

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