Wearable Lower-Limb Exoskeleton for Children With Cerebral Palsy: A Systematic Review of Mechanical Design, Actuation Type, Control Strategy, and Clinical Evaluation

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***Abstract*—Children with a neurological disorder such as cerebral palsy (CP) severely suffer from a reduced quality of life because of decreasing independence and mobility. Although there is no cure yet, a lower-limb exoskeleton (LLE) has considerable potential to help these children experience better mobility during overground walking. The research in wearable exoskeletons for children with CP is still at an early stage. This paper shows that the number of published papers on LLEs assisting children with CP has significantly increased in recent years; however, no research has been carried out to review these studies systematically. To fill up this research gap, a systematic review from a technical and clinical perspective has been conducted, based on the PRISMA guidelines, under three extended topics associated with "lower limb", "exoskeleton", and "cerebral palsy" in the databases Scopus and Web of Science. After applying several exclusion criteria, seventeen articles focused on fifteen LLEs were included for careful consideration. These studies address some consistent positive evidence on the efficacy of LLEs in improving gait patterns in children with CP. Statistical findings show that knee exoskeletons, brushless DC motors, the hierarchy control architecture, and CP children with spastic diplegia are, respectively, the most common mechanical design, actuator type, control strategy, and clinical characteristics for these LLEs. Clinical studies suggest ankle-foot orthosis as the primary medical solution for most CP gait patterns; nevertheless, only one motorized ankle exoskeleton has been developed. This paper shows that more research and contribution are needed to deal with open challenges in these LLEs.**

***Index Terms*—Assistive robotics, cerebral palsy, lower-limb exoskeleton, systematic review, wearable robot.**

I. Introduction

CEREBRAL Palsy (CP), Spinal Cord Injury (SCI), Spina Bifida (SB), and Traumatic Brain Injury (TBI) are the primary causes of mobility disorder in children [1]. CP is the most common mobility disorder in children [2], [3], with 8,000 to 10,000 children being diagnosed with cerebral palsy worldwide every year [4]. CP results from injury or damage to the brain before birth or in early childhood [3]. Common to most individuals with CP is major trouble controlling movement, posture, and balance [5]. The energy cost of walking for children with CP can be more than two to three times higher in comparison to their unimpaired peers [6]. Unfortunately, there is no cure for CP. Routine medical care, such as surgery [7], physical therapy [8], and muscle injections [9], can enhance walking ability in children affected by CP, but decreased knee extension generally persists or recurs after treatment. In spite of treatments to help the individuals who can walk, some children with CP worsen and lose their capacity to walk when they are grown-ups [10], [11].

Since CP has no cure, CP therapies emphasize how best to support people to enhance their life quality [12]. In developed countries, children with CP are referred for rehabilitation therapy and constitute the biggest diagnostic community treated in pediatric rehabilitation [13]. A need for complementary and additional technology for gait pathology in children with CP has resulted in the establishment of new strategies for gait rehabilitation. Treadmill-based gait training in the presence of Immersive Virtual Reality [14] and Cable-Driven Parallel Robot [15] have shown some positive initial findings in improving gait patterns of children with CP. A systematic review by Lefmann *et al.* [16] and Carvalho *et al.* [17] reported that robotic-assisted gait training (RAGT) devices (e.g., Lokomat, Gait Trainer GT1) might positively affect children with CP on activity parameters such as walking speed and standing ability, emphasizing further research is needed in this area. Whereas treadmill-based solutions are generally limited to clinics and research facilities, wearable technologies that can be implemented regularly in community settings have the expected added advantage of improved treatment access and dosage [11]. Repeated exposure to assistance leading to modified posture may augment more desirable walking patterns while also making possible increased physical activity [11].

A considerable demand for improved rehabilitation outcomes when treating neurological gait abnormalities has resulted in the creation of a lower-limb exoskeleton

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(LLE) [18], [19]. Nevertheless, only several exoskeletons so far have been designed for children [20], [21], particularly for children with CP [22]. Yilmaz and Dehghani-Sanij reviewed assistive robotic exoskeletons for children to establish requirements of such devices for the pediatric population [23]. Gonzalez *et al*. [24] reviewed different design features and the treated conditions focusing on robotic devices for pediatric rehabilitation. They could identify that safety, weight, and operability are critical parameters in the successful design of robotic devices for children, particularly for children with neurological disorders [24]. Specifically developed for a pediatric population, an exoskeleton should address some special considerations in the design [21]. First, a child's growth rate, which is approximately 6–7 cm and 3–3.5 kg per year [25], must be taken into account in a pediatric exoskeleton design. To this end, the wearable robot should support a broad range of heights and weights either by adjusting to a child with extendable elements or by creating a customizable design that is refitted or adjusted easily as the child grows. Second, the exoskeleton must be light, compact, and silent. Ideally, it should be much lighter than the child wearing the robot, where 24 kg is the average weight of children who are six years old. Third, since children have a developing cognition resulting in weaker concentration skills, the designed exoskeleton should be easy to learn and use. Fourth, the exoskeleton should generally recognize any abnormality in the physiology or anatomy of patients. Last of all, as gait therapy in CP and other neurological disorders begins typically at a young age, as early as five [26], the exoskeleton must be safe and comfortable.

An exoskeleton for children with CP not only should address the issues mentioned above, but it should address the following concern as well. It should be suitable for CP gait patterns (see section II.D); however, a very limited number of exoskeletons have been developed in this regard. Given crouch gait, characterized by excessive knee flexion, is the most prevalent gait disorder in CP [27], most of these devices are knee exoskeletons [28]–[32].

Since there is a close interaction between a human and a robot in wearable exoskeleton systems, safety is the utmost priority [33], [34]. In this regard, the type and location of actuators play critical roles in increasing user safety and comfort [35]. Three types of actuator systems, including pneumatic, hydraulic, and electric actuators, have been employed to provide mechanical power for exoskeletons [36]. In the literature for powered exoskeletons, electric actuators are the most popular type because they are lightweight, compact, reliable, and low-noise [37]–[39], albeit they have a lower power-to-weight ratio than the other types [40], [39]. Additionally, the location of actuators is an important issue because it is directly related to the mechanical design and the control architecture [41].

The mechanical structure of LLEs is a crucial factor that directly influences the effectiveness and efficacy of their user interaction [42], [43]. Wearable LLEs need a mechanical design matching human lower limbs to enable them to imitate a normal gait [44]. In this regard, choosing the correct number of degrees of freedom (DOF) for these robotic exoskeletons can guarantee kinematic compatibility with the user's limbs, resulting in maximum safety, minimum collision, and comfortable walking [45]–[48]. The exoskeleton's weight and mass distribution significantly affect both functional performance and metabolic consumption, as studies have shown that metabolic energy consumption during walking increases with additional load mass and more distal location [49]–[51]. However, children with CP consume more metabolic energy during walking compared with their typically-developing peers [52]–[54]. Therefore, it is important to pay special attention to the weight of exoskeletons tailored for children with CP as a critical mechanical factor.

Human gait in interaction with an LLE has extremely unstable dynamics (in the simplest form, similar to the inverted pendulum [55]), and it is susceptible to any input variation [56]. Even a small variation or disturbance in the input can lead to a total failure in the coordination of the complex human-exoskeleton system [56]–[59]. Walking assistance requires a strong alignment between human intentions and exoskeleton movement [60], [61]. Hence, a control structure is highly desirable to make stable human walking possible [56], [62]. The primary purposes in control systems of LLEs are assisting users in following normal gait patterns, improving their walking ability, and facilitating their recovery [63].

In recent years, multiple lower limb orthosis and exoskeleton devices have been designed and developed for children with CP, and clinical results show that wearable LLEs have a considerable potential to improve CP children's walking capability [64]–[68], [20], [21], [29]–[31]. Although there is a growing body of review studies on the effectiveness of LLEs in SCI [69]–[73], stroke [74], [75], and neurological disorders broadly [76], no literature has been conducted to systematically review the efficacy and different characteristics of these devices for children with CP. These devices can be broadly categorized according to their joint mechanisms, including single-joint and multi-joint exoskeletons. Along with the mechanical structures, various actuator types, assistive control strategies, and clinical characteristics are discussed and reviewed in this paper. To understand some clinical and technical terminologies addressed in this paper, popular Cerebral Palsy classifications and common control strategies in this paper have been presented in sections II and III, respectively.

II. Cerebral Palsy

CP is described as an umbrella term addressing a range of non-progressive motor disorder conditions beginning in early childhood and continuing through the lifetime [77]. In addition to motor disorders, symptoms of CP include secondary musculoskeletal disorders, epilepsy, and disturbances of cognition, perception, sensation, communication, and behavior [2]. Originally reported by Little in 1861 (and initially known as 'cerebral paresis'), CP is the result of a lesion or defect in an immature brain [77].

The prevalence, incidence, and the most usual causes of CP have changed over time due to improvements in prenatal and pediatric treatment [3]. Population-oriented researches from all across the world reveal CP prevalence estimates, based on age, sex, race, time interval, and geographical location, range from 1 to more than 4 per 1,000 live births [78]–[83].

Since CP is a heterogeneous condition rather than a single disease, and to shed light on prognosis and guide selection of the best interventions, a group of classification systems is required; motor-type, topography, gross motor function, and gait patterns [84], [85]. Additionally, such a classification system could be useful in evaluating whether a child with a particular class of CP would benefit from specific treatments or not [86].

A. Motor-Types

The motor-type classification system includes five sub-groups: spasticity, dyskinesia, ataxia, hypotonia, and mixed motor types [87]. The best available evidence reports that spasticity is the most prevalent type of CP, where approximately 90 percent of children with CP have a predominantly spastic motor type [88], [89].

Spasticity is the term employed to define situations where muscles become overactive due to a velocity-dependent resistance to stretch [84]. Spasticity can result in secondary impairments, such as joint dislocation, loss of muscle length, and pain [84]. Dyskinesia involves uncontrolled and involuntary muscle contractions resulting in either repetitive movements and twisting or unusual postures or both [87]. Ataxia leads to loss of muscular coordination and balance problems where motions suffer from abnormal rhythm, force, and accuracy [84]. Hypotonia is when muscle tension is reduced, which is the least frequent CP motor-type [84]. A mixed motor type contains clinical symptoms of more than one type, mostly spasticity and dyskinesia [87]. Results show that 30% of children affected by CP have a mixed pattern of motor-type disorders [90]. While classical sub-groups are easily identified, mixed or varying motor-types are far more complicated to be identified [87]. It is clear that, in terms of reliability, the motor-type classification system is poor [87].

B. Topography

Classifications based on the topographical distribution are commonly used [87]. According to the Surveillance of Cerebral Palsy in Europe (SCPE) definitions, CP can be divided into two definable topographies [2], [83]: Unilateral (one side of the body is affected totally or partially) and Bilateral (both sides of the body are affected totally or partially). Unilateral CP includes monoplegia (one limb is affected, mostly a lower limb) and hemiplegia (only one side of the body is involved, and an upper limb is affected more frequently than a lower limb). Bilateral CP comprises diplegia (all limbs are affected, although lower limbs are usually more impaired than upper limbs), triplegia (three limbs are involved, usually one upper limb and both lower limbs), and quadriplegia (all four limbs and trunk are impaired, with upper limbs being affected more than lower limbs) [91]–[94] (Fig. 1). In most reports, diplegia is the most usual type (30% - 40%), with rates for hemiplegia around 20% - 30%, and quadriplegia around 10% - 15% [90], and monoplegia and triplegia being fairly rare [94]. The topographical classification system is thus also very generic and does not satisfy specific requirements.

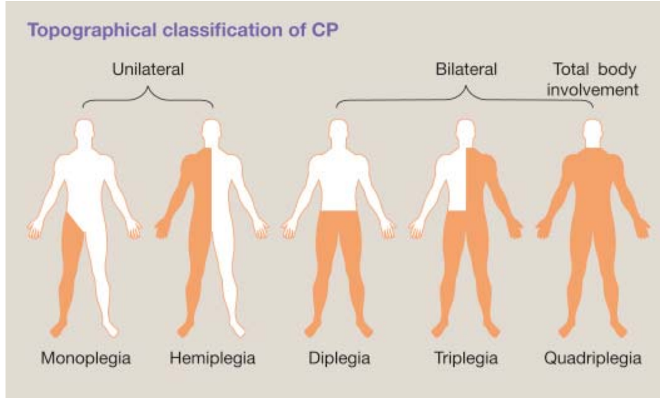

Fig. 1. Topographical distribution, according to SCPE identification [85].

Moreover, motor-type and topographical classifications have low reliability, particularly when investigators have received specific training in CP classification [87]. To address the need for an easy-to-use, quick, valid, and reliable CP classification system, scientists have established gross motor function as a five-level classification system similar to the grading and staging structures used in medicine [86].

C. Gross Motor Function

The Gross Motor Function Classification System (GMFCS) is a standard and reliable age-based classification system dividing children affected by CP into five levels according to their gross motor abilities [86]. Before developing the GMFCS, the severity of the children's gross motor disability was classified only as Mild, Moderate, or Severe [95], [96]. The GMFCS is based on the evaluation of self-initiated movement with a particular focus on function during sitting and walking [86]. Distinctions between different levels are according to functional limitations, the need for assistive technology or wheeled mobility, and movement quality [86].

These levels can be generally described according to the following [86]: Individuals in Level I walk without restriction with some limitations in more advanced gross motor skills. Individuals in Level II walk without assistive mobility devices, while they have some restrictions on walking outdoors and in the community. Individuals in Level III can walk with assistive devices, while they have some limitations on walking outdoors and in the community. Individuals in Level IV have self-mobility with some restrictions, while children are transported or use power mobility outdoors and in the community. Finally, individuals with Level V have extremity reduced self-mobility even with the use of assistive mobility devices.

The GMFCS is an aged-based classification, where each level has a different definition for the various age bands [86]: before the second birthday, from age 2 to the fourth birthday, from age 4 to the sixth birthday, from age 6 to 12. The available literature on Robotic-assisted gait training (RAGT) in pediatric participants shows that gait therapy generally starts after age 5 [26]; as a result, the definition of each level of GMFCS is presented for ages 6-12 in Table I.

TABLE I
GROSS MOTOR FUNCTION CLASSIFICATION SYSTEM (GMFCS) LEVELS, FOR CHILDREN AGED 6–12 YEARS [86]

GMFCS Level	Description
Level I	Children can walk outdoors and indoors and climb stairs without restrictions. Children can perform gross motor skills such as jumping and running, but coordination, balance, and speed are limited.
Level II	Children can walk outdoors and indoors and climb stairs holding onto a railing but may find it difficult to walk on uneven surfaces, inclines, crowds, or confined spaces. At best, children have only a limited capability to perform gross motor skills, including jumping and running.
Level III	Children can walk outdoors and indoors using assistive mobility devices. Children can walk up or down stairs holding onto a railing. Children can use wheelchairs to move on uneven terrain or travel for long distances.
Level IV	Children need physical assistance to move at home, school, outdoors, and elsewhere. Children can achieve self-mobility using power mobility.
Level V	Children are transported, and there is a severe lack of independence even in voluntary antigravity postural control of head or trunk. All motor function areas are restricted, and self-mobility is extremely limited, even with assistive mobility devices.

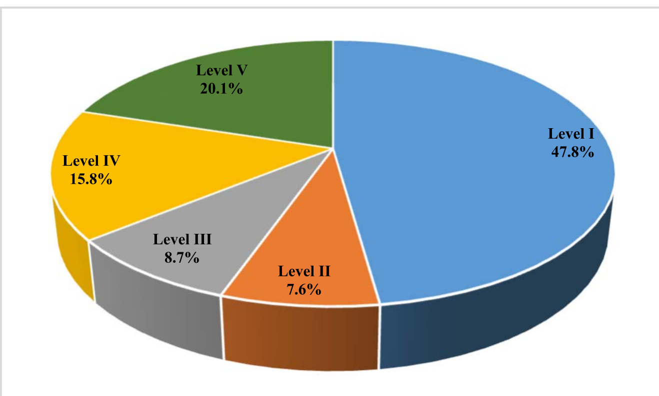

Fig. 2. GMFCS Level distribution for 8-year-old children with CP across the US in 2010 [89].

In 2010, an investigation of 8-year-old children with CP across the US disclosed distribution of GMFCS Levels for 375 CP participants: 47.8% Level I; 7.6% Level II; 8.7% Level III; 15.8% Level IV; 20.1% Level V [89] (Fig. 2).

Moreover, this investigation reveals that 58.9% of participants could walk independently, 7.8% walked utilizing a hand-held assistive device, and 33.3% had limited or no walking ability [89], [97] (Fig. 3).

D. Gait Patterns

Typical gait patterns in CP can be divided into the spastic hemiplegia (drop foot, equinus with different knee positions) and spastic diplegia (true equinus, jump, apparent equinus and crouch) [98], [99].

In the spastic hemiplegia, four groups of gait patterns based on kinematic data in the sagittal plane were identified (Fig. 4). The key feature of Group I is foot drop throughout the swing phase, which results in a lack of the first rocker at initial contact. The gait pattern of Group II is characterized by a drop foot in the swing phase and a permanent plantarflexion in the stance phase. This pattern is associated with knee hyperextension.

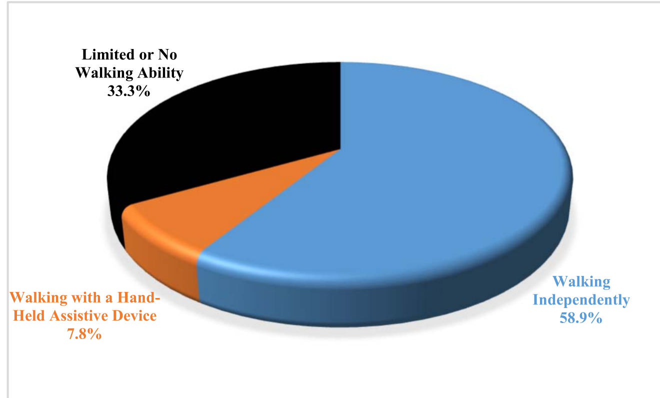

Fig. 3. Walking ability distribution for 8-year-old children with CP across the US in 2010 [89].

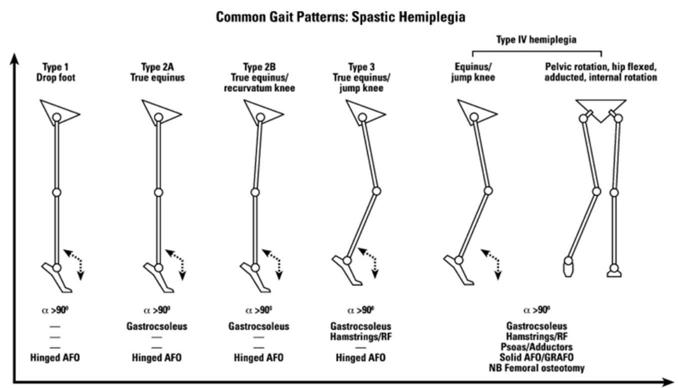

Fig. 4. Gait patterns and orthotic management for CP spastic hemiplegia [99].

Group III has all the deviations of groups I and II, plus reduced knee flexion during the swing phase, an increased lumbar lordosis, and a hyperflexion of the hip. In addition to deviations of groups I to III, patients in group IV have limited motion at the knee and hip [98], [99].

In the spastic diplegia, based on the kinematics of the ankle, knee, hip, and pelvis in the sagittal plane, four main groups were addressed: true equinus; jump gait; apparent equinus; and crouch gait (Fig. 5).

True equinus is characterized by the ankle in plantarflexion during the stance phase and the hips and knees extended. The jump gait pattern is defined by ankle equinus, knee and hip flexion, anterior tilt, and increased lumbar lordosis. The apparent equinus pattern addresses a regular range of dorsiflexion at the ankle, but the knee and hip are in excessive flexion across the stance phase, causing walking on the toes and giving a sense of equinus. Crouch gait is characterized by excessive dorsiflexion at the ankle, as well as excessive flexion at the hip and knee joints [98], [99].

III. CONTROL ARCHITECTURE

One of the most popular control system categories for exoskeletons, in particular LLEs, is a hierarchy-based control

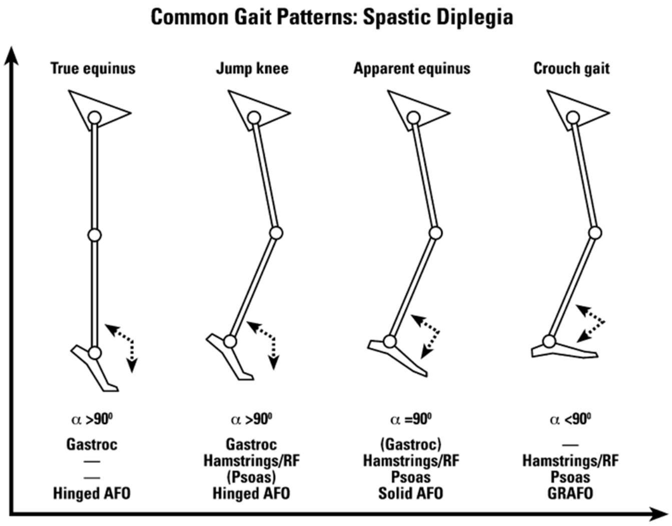

Fig. 5. Gait patterns and orthotic management for CP spastic diplegia [99].

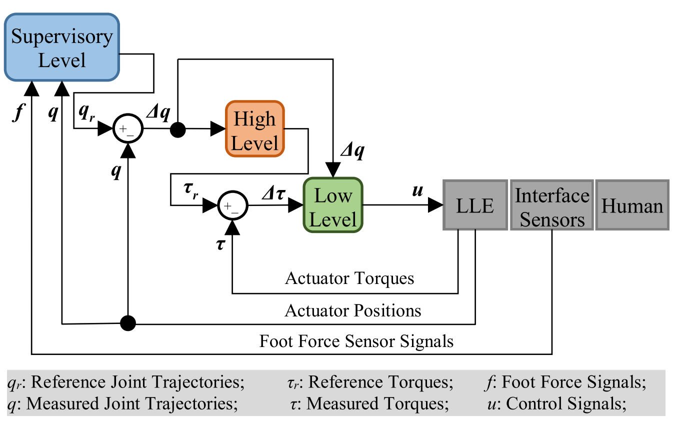

Fig. 6. General control architecture for LLEs [60], [103].

system divided into three levels of supervisory level, high-level, and low-level [35], [60], [100]–[102]. In LLEs, the supervisory level controller (e.g., Finite State Machine (FSM)) detects and, in some applications, predicts human gait phases to provide the high-level of a control system with reference joint trajectory. The high-level controller (e.g., impedance control) is responsible for torque control of human-exoskeleton interaction according to the signals received from the supervisory level to generate a reference torque for the low-level of a control system. The low-level controller (e.g., position/torque control) intends to control the position or torque of actuators based on the reference torque of the high-level or reference joint trajectory of supervisory-level [35], [60], [100] (Fig. 6). In LLEs, this hierarchy-based control system is also known as gait pattern control [100]. This section describes the most popular control strategies of different levels of a hierarchy control structure for LLEs designed for children with CP. These control strategies and signals received from different sensors distributed across the LLEs enable various rehabilitation therapies and gait walking improvement.

A. Supervisory-Level Controller: Finite State Machine (FSM)

Due to the transitional nature of the gait cycle, especially the stance and swing phases, it is usually helpful to divide the controller into separate control states based on the phase of the gait cycle [38]. In LLEs, FSMs are frequently used to define different controller states, such as between stance and swing phase. More divisions of the gait cycle are sometimes employed, such as specifying a late-stance phase for powered plantarflexion or for splitting up swing into several separate phases for swing flexion (early) or swing extension (mid/late). Although not all controller architectures have an FSM, many do [38]. Since FSM can combine force and position control and separate the controller into different states based on the gait cycle phase, it is a common supervisory-level controller [35], [104]. Many of the states of FSMs are highly dependent on the interactions with the environment. For instance, foot contact with the ground indicates the stance and swing phases of the gait cycle [35]. In this study, the states of the FSM are closely related to the contact situations with the environment.

B. High-Level Controller: Impedance Control

It is critical to make sure that patients actively participate in the therapy and never resist the applied movement. To this end, impedance control is commonly used to implement rehabilitation therapies and takes advantage of patients' residual movement under the philosophy "Assist-As-Needed" [67], [105], [106]. A robot manipulator with impedance control is presented by an equivalent mass–spring–damper system with tunable parameters [107]. Impedance control was introduced by Hogan in 1985. The impedance of a system, Z(s), is specified as the relationship between the force of the system, F(s), and its movement, θ(s), (Eqs. (1) and (2)) [108], [109], [110].

$$Z(s) = \frac{F(s)}{\theta(s)} = Is^2 + Bs + k \tag{1}$$

$$f = I\ddot{\theta} + B\dot{\theta} + k\theta \tag{2}$$

where f is force, I is inertia, B is damping, and k is stiffness of the system. θ, $\dot{\theta}$, and $\ddot{\theta}$ are position, velocity, and acceleration of a robot respectively. Adaptive impedance control is employed to overcome uncertainties in the dynamic parameters of robotic exoskeletons [111].

C. Low-Level Controller: Position/Torque Control

The low-level control is the closest level to the actuators; hence, it is necessarily device-dependent [112]. Position and torque controls are the most common control strategies for low-level controllers.

1) Position Control: Position control, or trajectory tracking, guides the patient's lower limb to fixed reference gait trajectories based on the joint angles as feedback [113], [114]. It comprises an internal control loop using the error between the reference joint angle trajectories and measured angles by sensors on each LLE joint [105], [106]. For position control, defining the reference trajectory is critical, while pre-recorded gait trajectories from healthy subjects and mathematical models of normal gait trajectories are commonly employed [113], [114]. A positive point of this control strategy is the imposition of a predefined joint angle trajectory resulting in limited kinematic error, an influential factor in driving human motor learning [113]. Position control is especially appropriate for individuals with neurological disorders who are unable to move their lower limbs due to a lack of muscle strength [38], [113]. Nevertheless, it gives patients the least control and interaction with LLEs, limiting its overall applicability [38].

2) Torque Control: Although position control is common in robotics, it is not appropriate for all tasks, particularly when robots need to interact physically with the environment [115]. Torque control can be beneficial in achieving versatile and robust behavior as well as dependable and safe tasks in the presence of humans [115]. Therefore, torque control is widely employed in exoskeletons [116]. The low-level torque control tries to effectively track a reference torque using the actuator's electric current as the system input [117]. This control method uses the error between the reference torques and measured torques by sensors on each actuator. Since torque control has a close interaction with actuators, it delivers a simple way of controlling the flow of energy from the exoskeleton to the user, which is helpful in biomechanics research [118]. Moreover, this control method plays a critical role in the implementation of rigid-body inverse-dynamics control strategy [119].

IV. Search Methodology

Two popular databases of Scopus and Web of Science with the following topics are used to collect publications on LLE for children with CP. Topic1 = (Ankle OR Foot OR Knee OR Hip OR Leg OR (Lower AND (Body OR Extremity OR Limb))) AND Topic2 = (Exoskeleton OR (Assistive Robot*) OR (Wearable Robot*) OR (Robot* Suit) OR (Portable Robot*)) AND Topic3 = (Cerebral Palsy). All topics were designed to be searched for titles, abstracts, and keywords, where all of the results are in the English language. It should be mentioned that the asterisk (*) is a wildcard symbol that extends a search by finding words starting with the same letters.

With the above topics, the following results were found (research results were updated until the end of 2020): 94 papers from Scopus and 80 papers from Web of Science, with 55 papers in common. After a screening of the Abstracts, 44 papers were discarded because they did not focus on the review topics. The full texts of the remaining 75 papers were reviewed in further detail. Sixty papers did not meet the eligibility criteria and were excluded from the reviewing process due to the following reasons: 11 papers had no prototype and only had focused on simulation results; 13 papers developed stationary exoskeletons, while this study only reviews portable LLEs; 6 papers had developed passive exoskeletons; one paper had presented a very old review of LLE; finally, 29 papers had repetitive information and significant overlap with other papers, where only the main papers were included. The main papers have fully described engineering aspects of prototyped LLEs, referenced by other related papers. Moreover, after studying the full-text of the papers, two additional relevant papers were found and added to the review process. In total, 17 papers were included for the full-text review, of which seven studies are addressed in Table III, and ten studies are cited in Table VI. The process by which the articles were selected in this systematic review is represented in a flow diagram based on PRISMA (Fig. 7) [120].

For years, the number of publications in this area was extremely limited. Fig. 8 shows the temporal distribution of published papers in Topics 1-3 in Scopus and Web of Science after removing duplications during 5-year intervals from 1996 to 2020. By the year 2016 only 28 papers in total had been published, and by the year 2021 the number was 120. It is evident that the number of published papers, the interest, and the amount of investment in this area have exponentially increased over the last five years. It is predicted that between 197 and 341 papers will be published in this area during 2021-2025 (APPENDIX).

Based on the covered joints, multi-joint LLEs can be generally classified as trunk–hip–knee–ankle–foot (THKAF), hip–knee–ankle–foot (HKAF), trunk–hip–knee (THK), hip–knee (HK), and knee–ankle–foot (KAF) orthoses and exoskeletons, as presented in Fig. 9, where lower-limb single-joint

TABLE II
Clinical Study Details for Ankle Exoskeleton by Lerner *et al.*

Study Ref.	# CP Participants	# Male	Age (years) Range (years)	GMFCS Level (#)	CP Type (#)
[65]	5	4	15.2 ± 10.57 (5–30)	I(3)/II(1)/III(1)	—
[123]	3	2	21.33 ± 9.02 (12–30)	I(2)/III(1)	—
[124]	1	0	22	III(1)	—
[11]	5	4	15.2 ± 10.57 (5–30)	I(3)/II(1)/III(1)	—
[125]	8	7	14 ± 2.62 (9–17)	I(4)/II(4)	S[1](8)
[126]	6	5	15.83 ± 9.13 (9–31)	I(3)/II(1)/III(2)	—
[127]	6	5	15.83 ± 9.13 (9–31)	I(3)/II(1)/III(2)	—
[128]	6	5	14.94 ± 1.94 (11–16)	I(2)/II(4)	SD[2](3)/SH[3](3)
[129]	5	4	14.75 ± 2.10 (11–17)	I(2)/II(3)	SD(2)/SH(3)
[130]	7	6	14.83 ± 9.13 (6–31)	I(4)/II(2)/III(1)	S(7)
[131]	7	6	14.00 ± 9.49 (6–31)	I(4)/II(2)/III(1)	—

[1]Spastic.
[2]Spastic Diplegia.
[3]Spastic Hemiplegia.

TABLE III
AVAILABLE SINGLE-JOINT LLES FOR CHILDREN WITH CP

Reference		Mechanical Design			Actuation Type	Clinical Characteristics			Control Strategy		
Author (year)	Device Name	Single-Joint	Active DOF	Weight	Actuator	Age Band	CP Type	GMFCS	Supervisory-Level	High-Level	Low-Level
Lerner *et al.* (2018) [65, 123]	—	Ankle	2	1.8 kg (small) / 2.2 kg (big)	Brushless DC Motor	5–35	SD[1] SH[2]	I–III	FSM[3]	Sigmoid Function / PJMC[4]	Torque Control (PID)
Lerner *et al.* (2016) [22]	—	Knee	2	3.2 kg	Brushless DC Motor	5–19	SD	I–III	FSM	—	Torque Control (PID)
Yamada *et al.* (2018) [29]	—	Knee	2	3.52 kg	Electromagnetic Brake	—	—	—	—	—	—
Chen *et al.* (2018) [30]	P.REX	Knee	2	—	Brushless DC Motor	—	—	—	FSM	Impedance Control / Adaptive Control	Torque Control (PID)
Washabaugh *et al.* (2018) [31]	—	Knee	2	1.6 kg	Electromagnetic Brake	—	—	—	—	—	—
Mohd Adib *et al.* (2019) [32]	ExoRobo Walker	Knee	2	8 kg	Linear Actuator	—	—	II–III	—	—	—

[1] SD: Spastic Diplegia.
[2] SH: Spastic Hemiplegia.
[3] FSM: Finite State Machine.
[4] PJMC: Proportional Joint-Moment Control.

TABLE IV
CLINICAL STUDY DETAILS FOR KNEE EXOSKELETON BY LERNER *et al.*

Study Ref.	# CP Participants	# Male	Age (years) Range (years)	GMFCS Level (#)	CP Type (#)
[133]	4	1	11.5 ± 5.45 (6–19)	II(3)/III(1)	—
[136]	6	4	10.67 ± 5.01 (5–19)	I(1)/II(5)	SD[1](1)
[134]	7	4	10.57 ± 4.58 (5–19)	I(1)/II(6)	SD(7)
[64]	7	4	10.57 ± 4.58 (5–19)	I(1)/II(6)	SD(7)
[137]	4	—	10.0 ± 2.71 (6–12)	I(1)/II(3)	—
[22]	1	1	6	II(1)	SD(1)
[135]	7	4	10.5 ± 4.6	I(1)/II(6)	—
[18]	6	3	11.33 ± 4.5 (5–19)	I(1)/II(5)	SD(6)
[138]	2	—	11.50 ± 0.71 (11–12)	—	—
[28]	10	—	—	—	SD(10)

[1]Spastic Diplegia.

exoskeletons are easily categorized into hip, knee, and ankle systems [121].

The joint-based distribution of LLE prototypes for children with CP, according to the above classification, has been displayed in Fig. 10.

This Fig. shows fifteen LLE prototypes for children with CP are available, eleven of which are multi-joint, and only four single-joint. This Fig. clarifies that the multi-joint LLEs have received more attention than the single-joint ones for children with CP. Furthermore, HKAF exoskeletons are the most popular type of multi-joint LLEs for children with this neurological disorder.

V. LLE PROTOTYPES FOR CHILDREN WITH CP

As mentioned earlier, researchers on robotic exoskeletons so far have mainly focused on adults rather than children. As a result, the existing literature on exoskeletons for children, particularly for children with CP, is extremely limited. According to Fig. 10, fifteen LLEs for children with CP in this section have been categorized into single-joint and multi-joint exoskeletons based on their actuated joints. For each exoskeleton, available information regarding the mechanical design, actuator type, clinical characteristics, and control strategy has been delivered. In mechanical design, some important considerations, including actuated joints, degree of freedom (DOF), and weight of exoskeletons, have been addressed. In clinical characteristics, available data related to the age range of eligible participants, type of CP, and GMFCS have been explained, where control strategy addresses supervisory-level, high-level, and low-level controllers. Moreover, available information regarding sensors, power transmission, power supply, microcontrollers, and communication protocols are presented to support the topic. Needless to say, although the information was collected as uniformly as possible, some details regarding several exoskeletons are not available.

A. Single-Joint LLEs for Children With CP

Fig. 10 shows that six single-joint exoskeletons have been designed for children with CP, of which one is an ankle exoskeleton and five are knee exoskeletons. Table III presents

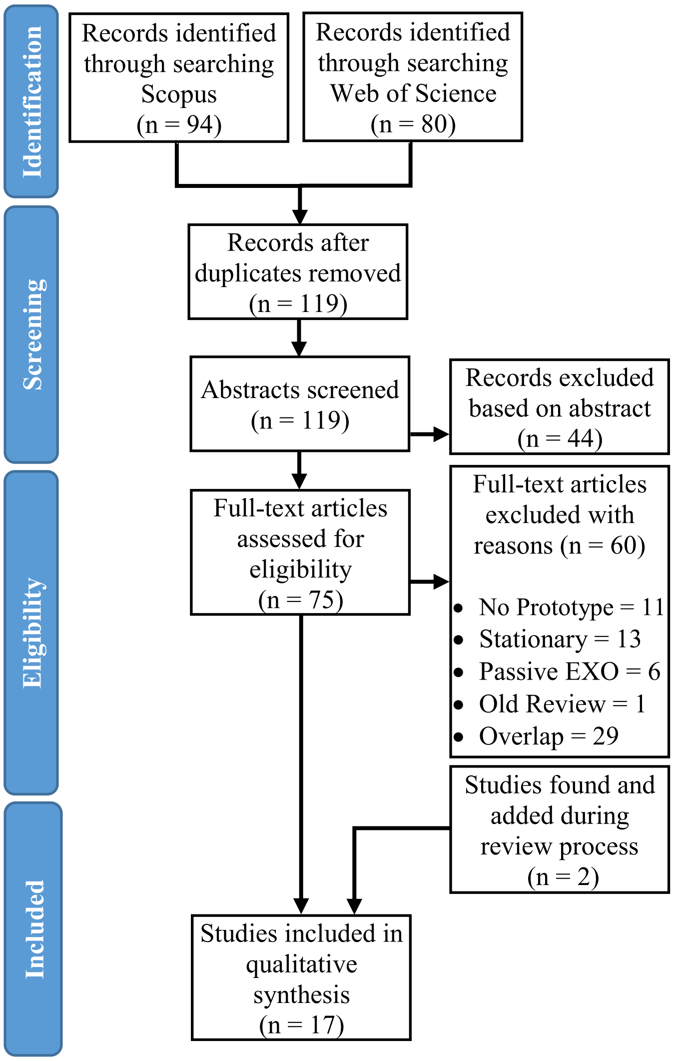

Fig. 7. Four-phase flow diagram of the systematic review based on PRISMA.

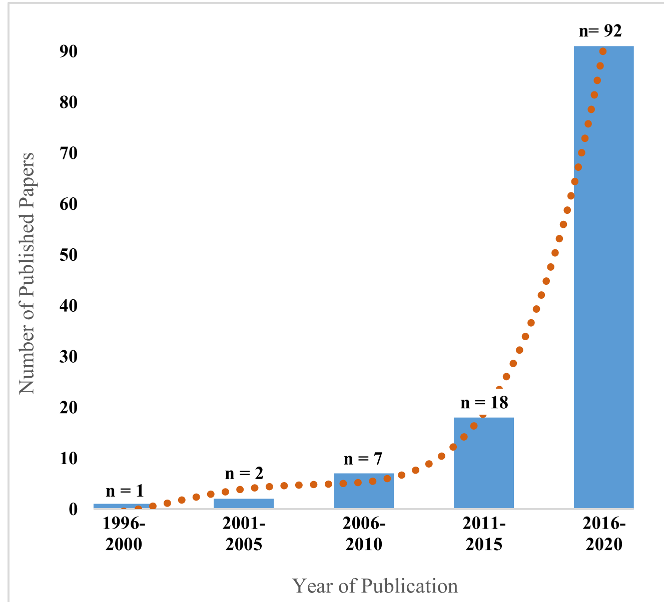

Fig. 8. Publication distribution of LLE for children with CP.

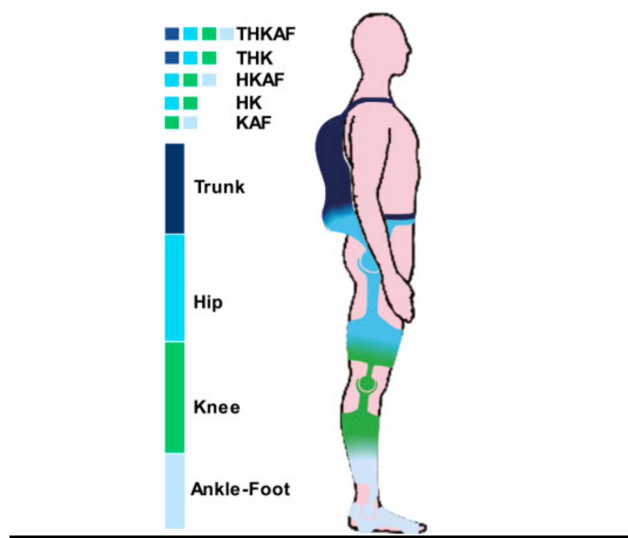

Fig. 9. Joint-based classification of multi-joint LLEs [121].

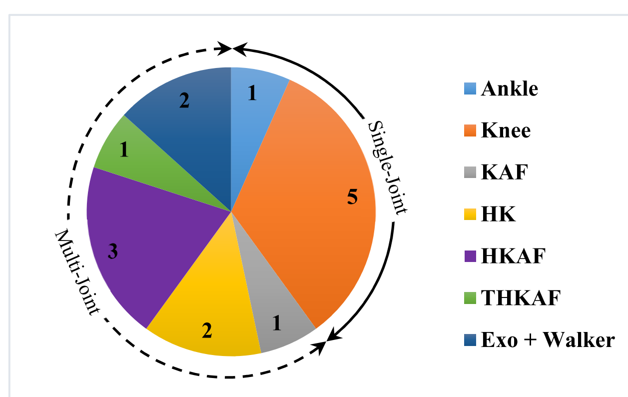

Fig. 10. Joint-based distribution of LLE prototypes for children with CP.

the available information about these exoskeletons in terms of mechanical design, actuator type, clinical characteristics, and control algorithm. As seen in this table, the lightest single-joint exoskeleton is 1.6 kg [31], and the heaviest is 4 kg [32], [122]. All of them have 2 DOF, where the DC motor is the common actuator among them. Most of these exoskeletons support children with GMFCS levels I-III, and only one of them is suitable for GMFCS level I-V [122]. These exoskeletons are adequate for children with spasticity, especially Spastic Diplegia (SD) and Spastic Hemiplegia (SH) [65], [123]. In terms of control algorithms, a FSM is used as a supervisory-level controller, and a torque control based on PID has been employed as a low-level controller. Sigmoid function [65], Proportional Joint-Moment Control (PJMC) [123], and Impedance Control [30] have been used as a high-level controller across the different exoskeletons.

1) Ankle-Foot Exoskeleton: Only one ankle-foot exoskeleton has been recently developed for children with CP. This exoskeleton has been described in the following.

a) Ankle-Foot exoskeleton by Lerner et al.: Lerner *et al.* designed an untethered ankle-foot exoskeleton in 2018, shown in Fig. 12 (a), to reduce the metabolic cost of transport during walking in children and young adults affected by CP. To this end, a low-profile, lightweight, and battery-powered ankle

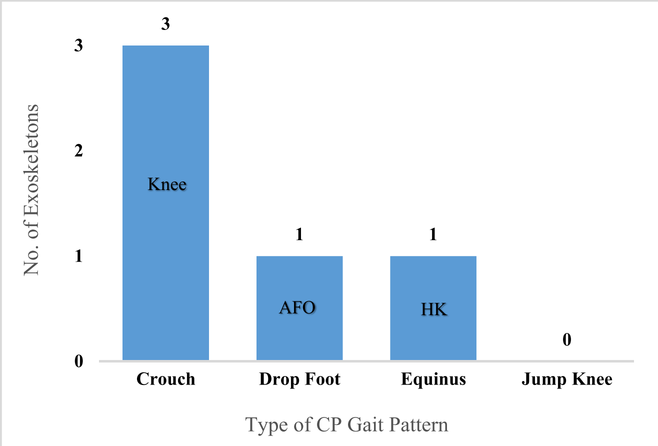

Fig. 11. Distribution of exoskeletons based on the CP gait patterns.

exoskeleton was developed that can be quickly adjusted for a broad range of heights, encompassing adults, adolescents, and young children [65], [124].

This exoskeleton has been developed according to the ankle-foot orthosis structure, involving one actuated DOF per leg at the ankle joint to assist a CP user. Since the target community includes a broad range of weights, two sizes of assembled exoskeleton were created. Including a battery, the assembled exoskeleton has a mass of 2.20 kg for a larger assembly and 1.85 kg for a smaller assembly. This exoskeleton has been constructed from aluminum for ease of fabrication [65].

For the smaller size of the assembly, a 24 V, 90 W motor with 89:1 integrated planetary gearbox (EC-4pole, Maxon) has been used. The larger assembly has a 24 V, 120 W motor with 111:1 integrated planetary gearbox (EC-4pole, Maxon). The gear reduction of the small assembly is 267:1, supporting up to 12 Nm, and for the large assembly, it is 333:1, supporting up to 20 Nm [65]. Torque has been transferred via the Bowden cables from pulleys installed on the DC motors to pulleys installed on the ankle assemblies. The ankle joint of the exoskeleton works as a simple revolute joint in the sagittal-plane, enabling both dorsiflexion and plantar flexion [65].

The exoskeleton has used force sensitive resistors (FSRs) embedded under the heel and ball of the foot to detect the gait phases. An FSM, the supervisory-level controller, receives these signals to command the on-off timing of exoskeleton assistance in the gait cycle. The low-level controller in this exoskeleton is a torque control based on a PID controller equipped with a torque sensor (TRT-500, Transducer Techniques) located in line with each exoskeleton ankle joint [65].

Two strategies have been presented for the high-level controller of this exoskeleton to generate the desired torque signal for the low-level controller. The first is a simple sigmoid function [65], where the second is a Proportional Joint-Moment Control (PJMC) [123]. A 32-bit ARM microprocessor (Teensy 3.6, TJRC) at 1 kHz has been employed for the real-time control of the exoskeleton. The exoskeleton is powered by a 0.14 kg 22.2V 910 mAh Lithium Polymer (Li-Po) battery (E-flite 6S), and a graphical user interface (GUI) in Matlab remotely controls the wearable robot through Bluetooth [65].

To evaluate the effectiveness of the exoskeleton for individuals with CP, some clinical studies have been developed. The details have been presented in Table II. These clinical studies showed that the lightweight untethered ankle exoskeleton could lead to a significant improvement in the metabolic cost of transport across walking with untethered exoskeleton assistance compared to how participants with a diverse range of age, body mass, and mobility disorder walked normally.

2) Knee Exoskeleton: According to the human gait data, knee exoskeletons have been employed to assist patients with mobility disorders by achieving an improved gait pattern [132]. Recently, five knee exoskeletons have been designed for children with CP and have been described in the following.

a) Knee exoskeleton by Lerner et al.: Lerner *et al.* designed a knee exoskeleton in 2016, shown in Fig. 12 (b), for children with crouch gait caused by CP to increase knee extension and supplement existing muscle function during overground walking [133], [134]. Instead of directing the limbs into a trajectory, this knee exoskeleton provides accurately timed knee extension assistance during walking [135].

In terms of mechanical structure, the exoskeleton is designed according to the knee ankle foot orthosis architecture, including one actuated DOF per leg at the knee joint to deliver extension assistance [22]. The ankle of the exoskeleton is a passive joint that can be set to assist or restrict ankle motion or can be adjusted to permit free rotation as a simple revolute joint [135].

Since classification in this paper is based on the actuated joint, this wearable robot is categorized as a knee exoskeleton. The exoskeleton contains custom-molded thermoplastic braces for the foot, shank, and thigh, which affix to children using Velcro straps [22], [135]. The exoskeleton weight is approximately 3.2 kg (1.58 kg per leg) without considering the control box weight (1.96 kg) [22].

A low-profile, lightweight, and appropriate motor assembly has been employed to deliver the desired level of assistance at the knee joint in most individuals with a crouch gait [22]. The motor assembly includes a back-drivable 24 V, 90 W brushless motor with a 3-stage, 89:1 reduction planetary gear head, and an embedded quadrature encoder (EC-powermax, Maxon Motors, Fall River, MA, USA). A second stage, chain sprocket transmission with a 3.5:1 reduction rate, delivers torque from the motor shaft to the knee rotation center. The motor assembly has a maximum output torque of 16.1 Nm, with a maximum continuous current draw of 3.82 A. The motor assembly has been attached to the upper (thigh) and the lower (shank) uprights by customized brackets [22].

The exoskeleton is equipped with encoders to pursue knee angular position and velocity, a reaction torque sensor at the knee, and FSRs embedded in the feet to chase ground contact [22]. Utilizing data from encoders and FSRs, an FSM has been employed in the supervisory-level controller to divide the gait cycle into three discrete phases of stance, early swing (knee flexion), and late swing (knee extension). A proportional-integral-derivative (PID) control scheme, as a low-level controller, uses the experimentally measured torque from a reaction torque sensor to obtain the desired torque output during each phase. In other words, the low-level controller

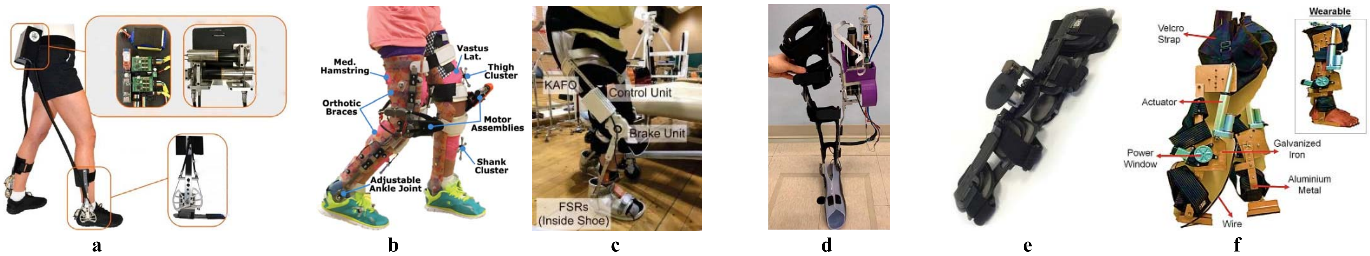

Fig. 12. Single-Joint LLEs for Children with CP: (a) Ankle Exo. by Lerner *et al.* [65], (b) Knee Exo. by Lerner *et al.* [133], (c) Knee Exo. by Yamada *et al.* [29], (d) Knee Exo. by Chen *et al.* (P.REX) [140], (e) Knee Exo. by Washabaugh *et al.* [31], (f) Knee Exo. by Mohd Adib *et al.* (ExRoLEG) [32].

TABLE V
DETAILS OF CLINICAL STUDY FOR HAL

Study Ref.	# CP Participants	# Male	Age	GMFCS Level (#)	CP Type (#)
[151]	12	8	16.2 ± 7.3 years (9–37 years)	I(2)/II(3)/III(5)/IV(2)	SD[1](9)/SH[2](2)/SQ[3](1)
[152]	6	4	16.8 ± 3.5 years (13–24 years)	II(1)/III(4)/IV(1)	SD(5)/SQ(1)
[153]	1	1	15 years	IV(1)	SD(1)
[154]	20	15	15.0 ± 6.3 years (8–37 years)	I(4)/II(3)/III(9)/IV(4)	Spastic(19)
[155]	19	13	15.4 ± 5.2 years (9–29 years)	I(3)/II(2)/III(9)/IV(5)	SD(13)/SH(3)/SQ(3)

[1]Spastic Diplegia.
[2]Spastic Hemiplegia.
[3]Spastic Quadriplegia.

is a torque control based on the PID controller. Knee extension assistance can be enforced during all phases, and when no extension assistance is needed, the PID controller applies a 0 Nm torque (zero torque) across the knee. A 22.2 V lithium-ion battery can supply the required DC power source of the exoskeleton [22].

To validate the preliminary efficacy and functional performance of the exoskeleton, some clinical trials have been conducted. In ten different clinical studies [18], [22], [28], [64], [133], [134], [136]–[138], inclusion criteria for recruiting participants generally included a diagnosis of crouch gait from CP between 5-19 years old with knee flexion contracture less than 5°, GMFCS Level I-III, and the lack of any health condition, apart from CP, that may have an effect on safety. In total, 60 participants could meet the criteria, 54 of whom could successfully complete the clinical trials. The details have been presented in Table IV.

To evaluate the effectiveness of the powered exoskeleton compared to the baseline during overground walking, the following criteria were measured for each participant: (i) kinematics (hip/knee/ankle angle) [18], [22], [133], [134], [136]–[138], (ii) muscle activity (vastus lateralis, medial hamstring, semitendinosus, gastrocnemius, rectus femoris) [18], [22], [64], [133]–[137], (iii) exoskeleton torque [18], [28], [64], [133], [134], [137], [138] and (iv) exoskeleton power [134].

These clinical trials showed that the exoskeleton effectively reduced crouch (increasing knee extension) when assistance was provided during the overground walking [22], [64], [133], [136], [137]. Results showed strong positive linear relationships between exoskeleton assistance and knee extension and strong negative quadratic relationships between spasticity and knee extension [18].

In a different study, a 15-year-old male with GMFCS Level III and spastic CP was recruited [139]. The ultimate goal of this study was to synchronize Neuromuscular Electrical Stimulation (NMES) with the knee exoskeleton to develop a hybrid NMES-exoskeleton device to assist CP children with crouch gait. Results show NMES to the rectus femoris and vastus lateralis during stance instantly improved mean peak knee extension within mid-stance and total knee excursion in the more affected leg [139].

Notwithstanding, in these clinical trials, the participants never carried the control box (1.96 kg) during the experiments. Moreover, a tethered supply was used to power the exoskeleton, while in practice, the battery should be carried by users. If users carry both the battery and the control box, the total weight of the exoskeleton will increase; as a result, this changes the inertia of movements. Additionally, the exoskeleton's weight is a highly challenging characteristic of wearable robots and plays a critical role in their stability. Hence, this issue can directly affect the results of the clinical trials.

b) Knee exoskeleton by Yamada et al.: A knee exoskeleton was developed by Yamada *et al.* in 2018 for children with crouch gait from CP (Fig. 12 (c)). To this end, a modular electromagnetic brake unit has been attached to the knee joint of the orthosis to improve the knee extension and mitigate the abnormal gait [29].

This exoskeleton has a knee ankle foot mechanical structure which has been outfitted with an electromagnetic brake per leg at the knee joint, a control unit, and FSRs for gait phase detection. The exoskeleton has one active DOF per limb at the knee joint to support extension assistance with a passive ankle mechanism designed to facilitate free rotation at the ankle joint. Since only the knee joint is actuated; therefore, this device is categorized as a knee exoskeleton. The mass of the exoskeleton

TABLE VI
AVAILABLE MULTI-JOINT LLES FOR CHILDREN WITH CP

Reference		Mechanical Design			Actuation Type	Clinical Characteristics			Control Strategy		
Author (year)	Device Name	Multi-Joint	Active DOF	Weight	Actuator	Age Band	CP Type	GMFCS	Supervisory-Level	High-Level	Low-Level
Patané *et al.* (2017) [66]	WAKE-up	KAF	4	2.5 kg	RSEA[1]	5–13	SH	—	FSM [2]	—	Position Control (PID)
Laubscher *et al.* (2017) [21]	—	HK	4	5.1 kg	Brushless DC Motor	6–11	—	—	—	—	Position Control (PD)
Kawamoto *et al.* (2009) [148, 149]	HAL	HK	4	—	DC Motor	8–37	SD[3], SH[4], SQ[5]	I–IV	—	—	—
Eguren *et al.* (2019) [67]	P-LEGS	HKAF	6	8 kg	Brushless DC Motor	4–8	—	—	—	Adaptive Impedance Control	Position Control (PD)
Andrade *et al.* (2019) [20]	ExoRobo Walker	HKAF	6	6.57 kg	Brushless DC Motor	5–24	—	—	FSM	Impedance Control	Torque Control (PD)
Canela *et al.* (2013) [156]	—	HKAF	6	—	Brushless DC Motor	7–17	Spasticity	—	—	—	—
Marsi-Bionics (2016) [157]	ATLAS	THKAF	12	12 kg [157] ATLAS2020 14 kg [145] ATLAS2030	Brushless DC Motor	3–14	—	—	—	Impedance Control	Position Control (PID)
Maggu *et al.* (2018) [158]	Trexo	HK + Walker	4	—	—	3–6	—	III–V	—	—	—
Bayón *et al.* (2018) [106]	CPWalker	HKAF + Walker	10	—	Brushless DC Motor	11–18	SD	I–IV	—	Impedance Control	Position Control (PID)

[1]RSEA: Rotary Series Elastic Actuator.
[2]FSM: Finite State Machine.
[3]SD: Spastic Diplegia.
[4]SH: Spastic Hemiplegia.
[5]SQ: Spastic Quadriplegia.

is 3.52 kg, where the mass of each brake unit is only 0.19 kg [29].

According to the weight of the brake unit (0.19 kg) and magnitude of required torque, a 12 V electromagnetic-actuated brake (Miki Pulley, BS-06-A-56-V1) with 7 Nm (static friction torque) and 15 ms (armature pull-in time) was employed for this exoskeleton. This brake comprises two rotary disks aligned on the axis with a passive spring mounted between the disks to separate them by a certain distance. As the voltage increases in the coil of the electromagnetic brake, these discs are magnetically absorbed into each other; therefore, the torque of friction is produced resulted in resisting limb motion. When the brake unit is not connected to the power supply, the passive spring keeps the discs' distance; thus, no friction torque is generated led to assistive limb movement. As a result, the brake can provide both resisting and assisting in the movement. Two 3D-printed links have connected the discs to the lower and upper braces. These links have been developed to deliver the brake torque to the Knee-Ankle-Foot-Orthosis (KAFO) without using any gear [29].

Two FSRs with a square measurement area (SparkFun, FSE-SEN-09376) are installed on the insole of each shoe to track the changes of pressure on the heel and forefoot of each foot. The first one is installed on the rear central part of the insole to detect the ground-contact state of the heel. The second one is installed on the forefoot to detect the ground-contact state of the toe. A microcomputer (Adafruit, HUZZAH32-ESP32 Feather Board) collects the signal from FSRs to detect the gait phase of the user in real-time, while a lithium-ion polymer battery (Data Power Technology, DTP603450, 3.7 V, 1000 mAh) supplies its required power [29].

Two adjacent lithium-ion batteries (KEEPPOWER, 18650, 3.7 V, 3500 mAh), which have been boosted to 12 V, have been utilized as the power supply. To avoid disturbance of the user during walking, the control unit, containing the microcontroller and the batteries, has been mounted over the knee joint [29].

In order to evaluate the effectiveness of the exoskeleton, two participants were recruited for clinical studies. The first participant was a healthy boy aged ten years, who wore the exoskeleton only on his left leg to walk 6 m straight forward in two conditions of with and without assistance. The second participant was an 18-year-old male individual with a crouch gait caused by CP. He could walk for 10 m in the following conditions: 1) without any assistance, 2) with the assistance only for the left leg, 3) with the assistance only for the right leg. The clinical results showed that the brake unit has the ability to improve knee extension during stance and facilitate the knee joint motion during the swing [29].

The downside of this study was the low number of participants in the clinical trials (only two participants), of which only one applicant was affected by CP. In addition, this exoskeleton requires a more accurate gait phase detection method with a more developed control architecture.

c) Knee exoskeleton by Chen et al. (P.REX): In 2018, Chen *et al.* designed and developed a pediatric knee exoskeleton (P.REX) to assist individuals with CP outside of a clinical environment (Fig. 12 (d)) [30], [140].

This exoskeleton has a KAFO mechanical architecture with one actuated DOF per leg at the knee joint. A customized bevel gear set, which includes a 3:1 reduction ratio, planetary gearhead GP 32C (51:1 reduction ratio), and a Maxon motor, was utilized to deliver the desired torque assistance. This exoskeleton can generate a maximum continuous assistive torque of 15.7 Nm at a joint velocity above 350°/sec. The motor assembly unit is linked to the upper (thigh) and lower (shank) uprights by customized designed brackets [30].

The exoskeleton has been equipped with encoders with the capacity of 500 counts per turn for angular position and speed, while FSRs mounted under the foot are used for gait phase detection. Moreover, an inline reaction torque sensor has been installed on the knee shaft between the gear and lower leg attachment. A multi-layered hierarchy control architecture and a microcontroller, according to the data acquisition system, were recruited to generate the appropriate control signals. The supervisory-level controller is based on an FSM receiving signals of FSRs and encoders, where the high-level controller is based on the impedance controller to generate the desired torque. The low-level controller is a torque control based on a local PID controller and feedforward control scheme to create an adequate control signal for the actuator [30].

The exoskeleton can be powered by a DC power supply in a laboratory setting or supplied by a 24V lithium-ion battery for out-of-laboratory uses [30].

To evaluate the effectiveness of the exoskeleton for individuals with CP, a 15-year-old male with GMFCS level III and crouch gait from bilateral spastic CP was recruited [141]. The findings show that the exoskeleton could synchronize assistance during overground walking with forearm crutches [141].

In [140], one Typical Development (TD) 25-year-old man and one CP 15-year-old male with GMFCS Level III and mild spasticity in the right limb were recruited. Results from the participant with TD were employed to validate control system performance, while data from the participant with CP were used to assess and compare the effectiveness of two exoskeleton assistance modes (Constant and Adaptive) for the device during overground walking. Gait speed for the healthy participant was significantly slowed for all assistance modes while walking with the exoskeleton was disrupting compared to walking without it. For children with CP, the Adaptive mode could show the best results in peak knee angle with no undesired EMG effects and a minor impact on gait speed [140].

d) Knee exoskeleton by Washabaugh et al.: In 2016, Washabaugh *et al.* developed a lightweight and low-cost wearable robotic brace that provides different levels of resistive torque across the knee joint during walking for individuals with neurological injuries, such as stroke, CP, SCI (Fig. 12 (e)) [31], [142], [143].

This wearable device has one actuated DOF per leg at the knee joint, and the exoskeleton mass is 1.6 kg [142]. This exoskeleton uses an electromagnetic brake at each joint to deliver about 56 Nm of torque during normal gait, while two pairs of permanent magnets (DX08B-N52, KJ Magnetics, Pipersville, Pa) create the required magnetic field [142].

In a clinical study, kinematics and electromyography of seven healthy participants were collected to evaluate the biomechanical influences of the exoskeleton on them [142]. In another pilot study, six individuals who were chronic stroke survivors (3 males; age: 58.0 ± 8.5 years (range: 18–75 years)) were recruited [31]. The results of these studies provide preliminary evidence that resisted walking with this exoskeleton induces meaningful biomechanical after-effects resulting in improvement of overground walking [31], [142].

e) Knee exoskeleton by Mohd Adib et al. (ExRoLEG): In 2018, Mohd Adib *et al.* developed a low-cost exoskeleton robotic leg (ExRoLEG) device for individuals affected by CP with GMFCS Level II and III to improve their walking pattern and speed up their rehabilitation process (Fig. 12 (f)) [32].

This exoskeleton has one actuated DOF per leg at the knee joint. ExRoLEG mainly comprises metals; therefore, the device is heavy, and the total weight of the assembly is about 4 kg per leg. A power window motor (2.9 Nm) and a linear actuator (500 N linear force) are installed at each knee joint to generate the desired torque. ExRoLEG uses an Arduino system to control the device. Ten participants have been recruited to evaluate the effectiveness of ExRoLEG. It should be noted that details of clinical characteristics of the participants in terms of CP type, age, GMFCS level, etc., are not available to the best of the authors' knowledge. Most of them have indicated that ExRoLEG is comfortable and has good handling. However, ExRoLEG is too heavy for a child to carry [32].

B. Multi-Joint LLEs for Children With CP

As Fig. 10 shows, this subsection articulates different multi-joint LLEs for children with CP. This Fig. shows nine multi-joint LLEs are available for children with CP, of which one is KAF, two are HK, three are HKAF, one is THKAF, and two are exoskeleton-plus-walker devices. Details of these exoskeletons in terms of mechanical design, actuator type, clinical characteristics, and control algorithm have been presented in Table VI. This table shows that the minimum actuated DOF is 2 [28], and the maximum DOF is 12 [144], where the lightest multi-joint exoskeleton is 2.5 kg [66], and the heaviest is 12 kg [144]. Furthermore, while most of these exoskeletons have used brushless DC motors as their actuators, servo motors have rarely been employed. As can be seen in the table, the youngest age for the use of multi-joint exoskeletons is three years old, as used in ATLAS 2030 [145]. Moreover, in terms of CP motor-type, all individuals recruited for clinical studies of these exoskeletons had spasticity, not other CP motor-types.

In terms of control strategy, FSM is utilized for supervisory-level, impedance control has been used for high-level, and position/torque control based on PD/PID is employed for low-level.

1) Knee-Ankle-Foot (KAF) Exoskeleton: One KAF exoskeletons have been designed and developed in recent years for children with CP, and it has four actuated DOF at the knee and ankle joints. These three exoskeletons are explained below.

a) KAF exoskeleton by Patané et al. (WAKE-up): In 2017, Patané developed a Wearable Ankle Knee Exoskeleton, the so-called WAKE-up, to support gait in children affected by neurological disorders, such as CP (Fig. 13 (a)). This exoskeleton is the updated version of the alpha-prototype [146].

The WAKE-up comprises two actuated joint modules, a knee joint module as well as an ankle joint module, to rehabilitate children with CP. Since each module has one active DOF, the exoskeleton has two active DOF per limb. In terms of weight, the WAKE-up is 2.5 kg, and it is the lightest exoskeleton compared to other types of multi-joint exoskeletons designed for children with CP. To guarantee the user's safety, the range of motion (ROM) of the actuated joints is mechanically restricted to 100 ° and 45 ° at the knee joint and ankle joint, respectively [66].

To generate torque for the WAKE-up and improve the user's safety, each module is outfitted with a Rotary Series Elastic Actuator (RSEA) and a belt/pulley stage with a reduction ratio of 2/3. The RSEA includes a servomotor with an embedded PID controller (Dynamixel EX-106+, Trossen Robotics, IL, USA) and an ad-hoc designed torsion spring. An absolute 14-bit magnetic encoder (MA3, US Digital, WA, USA) records the spring rotation [66].

Depending on the modularity of the WAKE-up, the exoskeleton control architecture has only one WAKE-up supervisor (WU-S) to configure and communicate WAKE-up nodes (WU-Ns). The WU-S is the main supervisory-level controller and comprises a MyRio controller (National Instruments, TX, USA) outfitted with an ARM Cortex-A9 in the programming environment of LabVIEW Real-Time (National Instruments, TX, USA) grounded on a Linux Real-Time kernel. Not only is the WU-S responsible for the overall supervision of the exoskeleton, but it also should manage the high-level GUI, where the user can access an internet browser using a WiFi connection to select a scenario, to monitor the system status, to send commands to the nodes, and to download data logging [66].

WU-N contains a 32-bit microcontroller (PIC32MX440F512H, Microchip, USA), a 9-axis inertial measurement unit (IMU) (MPU-9350, InvenSense, USA), a foot insole with six footswitches, a 14-bit magnetic absolute encoder, and high torque digital servomotor. Relying on FSM, the local supervisory-level controller uses the collected sensory data and the microcontroller for gait phase detection. Moreover, an RSEA-embedded PID position controller in each module works as the local low-level controller [66].

The WAKE-up has been developed for the rehabilitation of CP children aged from 5 to 13 years old. To analyze the exoskeleton clinically, four healthy children (10.25 ± 2.63 years) and three CP children (11.00 ± 2.65 years) affected by hemiplegia on the right side have participated in clinical trials [66]. Clinical results derived from different walking trials confirmed that the exoskeleton was successful in the assistance of the examined children with CP in improving the physiological gait patterns, particularly at the ankle joint. Nevertheless, the following concerns can be raised. First of all, the provided torque assistance at the knee level should be increased [66]. Furthermore, the WAKE-up control system should support a rehabilitative scenario [66].

2) Hip-Knee (HK) Exoskeleton: Only one HK exoskeleton has been developed for children with CP. This exoskeleton was developed in 2017 with four activated DOF.

a) HK exoskeleton by Laubscher et al.: In 2017, an HK robotic exoskeleton was developed by Laubscher *et al.* in Fig. 13 (b) to assist and rehabilitate children with gait disorders, such as those affected by CP or SB [21]. This 5.1 kg device has a hip-knee exoskeleton structure, including one active DOF for each joint to provide the required torque assistance. A frame has been designed only to hold the exoskeleton in the preliminary stages. The frame is not a part of the exoskeleton, and it is separated from the exoskeleton for the pilot studies. Each actuator is driven by a 70 W brushless DC motor using a 3-stage toothed-belt transmission with a total speed-reduction ratio of 40.6:1. This provides a nominal output speed of 375 deg/s, stall torque of 35.7 Nm, and a continuous torque of 5.4 Nm [21], [147]. Belts are employed in the transmission system to enable compliant behavior under load while also being lightweight and quiet. The actuators have a modular structure, while they are lightweight and thin, with a weight of 0.6 kg and a height of 46 mm.

A magnetic angle sensor has been employed to directly measure the angle of each joint, where the joint velocity is estimated according to the motor velocity measured by Hall effect sensors. Powered by a 38 V power supply, a dSPACE MicroLabBox collects the measured signals to control the device. To this end, a decentralized proportional-derivative (PD) position controller, as a low-level controller, was employed to empower the exoskeleton to chase the desired gait trajectory with a gait cycle period of 1.1 s [21].

Until now, no clinical trial has been conducted on this device to evaluate its effectiveness. However, this exoskeleton has been designed for a target user age range of 6–11 years. The age six has been chosen as the lower boundary to guarantee that participants have adequate communication skills to follow a series of commands, sufficient attention span to carry out specific tasks, and proper motor skills. The age of eleven is chosen as the upper age boundary because children older than this age are tall enough to use an adult exoskeleton [21].

Although some experimental mechanical results are promising, there are a number of issues that were raised during testing and should be addressed. First, the exoskeleton has been powered externally at this development stage; hence, the weight of 5.1 kg does not include the battery's weight as a portable power supply [21]. This issue negatively affects exoskeleton effectiveness, particularly for children at the lower age boundary. Second, no clinical study has been conducted on the exoskeleton to assess the device's effectiveness directly, to the best of authors' knowledge. Third, since the simulation model may be inaccurate, it may not exactly show the mechanical device behavior [147].

b) HK exoskeleton by Kawamoto (HAL): Hybrid Assistive Limb (HAL) was developed by Kawamoto *et al.* as a walking assistive device to support individuals with hemiplegia [148]. In 2012, Taketomi and Sankai developed a control strategy for HAL to assist individuals with CP to ascend stairs

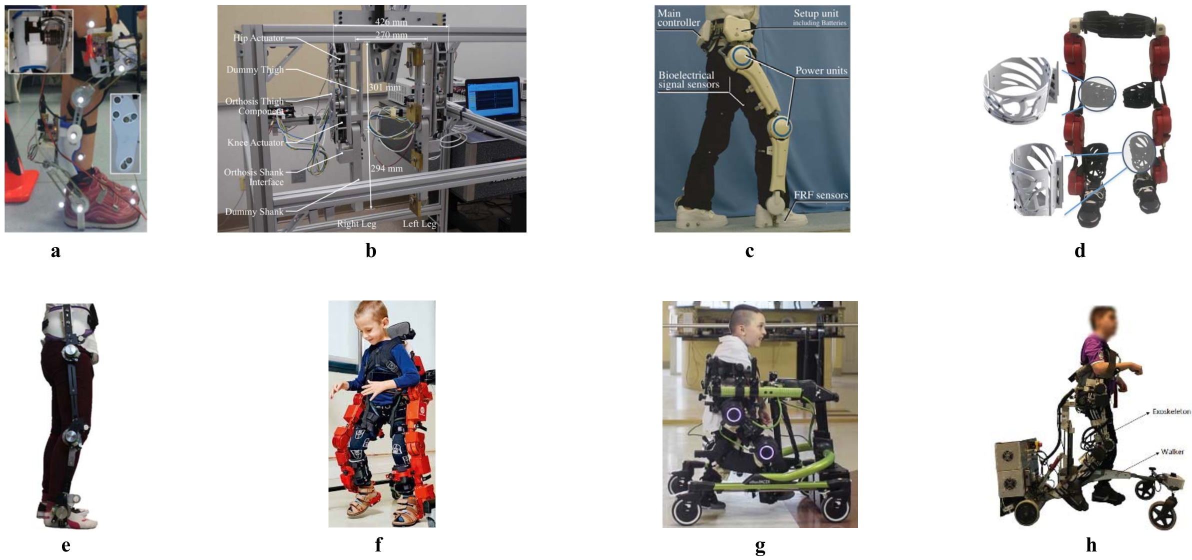

Fig. 13. Multi-Joint LLEs for Children with CP: (a) KAF Exo. by Patané *et al.* (WAKE-up) [66], (b) HK Exo. by Laubscher *et al.* [21], (c) HK Exo. by Kawamoto *et al.* (HAL) [148], (d) HKAF Exo. by Eguren *et al.* (P-LEGS) [67], (e) HKAF Exo. by Andrade *et al.* (ExoRoboWalker) [20], (f) THKAF Exo. by Marsi-Bionics *et al.* (ATLAS) [161], (g) HK Exoskeleton-Plus-Walker by Maggu *et al.* (Trexo) [158], (h) HKAF Exoskeleton-Plus-Walker by Bayón *et al.* (CPWalker) [68].

(Fig. 13 (c)) [149]. HAL has two actuated DOF per leg at hip and knee joints and one passive DOF at ankle joint [150]. The HAL frame comprises steel and aluminum alloy materials to guarantee the lightness of the exoskeleton. In order to produce the torque of each actuated joint, the actuator has a DC motor and harmonic drive with a considerable reduction gear ratio [150].

A rotary encoder is employed in each actuator to measure the joint angle, while force sensors are mounted in the foot sole to measure the floor reaction force (FRF). HAL uses the center of ground reaction force (CoGRF) for phase detection [149]. However, more information about the supervisory-level and low-level controllers has not been provided in this study. In this regard, a healthy male was recruited for a pilot study. Results showed that their strategy for control of HAL is effective for walking and stair ascent assistance.

In order to assess the effectiveness of HAL in gait improvement for individuals with CP, the following clinical studies have been conducted. Details of these studies are presented in Table V. These clinical studies revealed that HAL could improve the walking function, stability and safety, and gross motor capabilities of individuals with CP. The effectiveness of the exoskeleton has been evaluated by (i) measuring the transition between commanded torque and calculated torque, (ii) tracking the trajectory of center of ground reaction force (CoGRF) during walking.

3) Hip-Knee-Ankle-Foot (HKAF) Exoskeleton: HKAF exoskeletons similar to KAF orthosis are used. The HKAF exoskeletons are bilaterally attached to a hip unit via a pelvic band called the lumbar-sacral orthosis. These exoskeletons are developed for the control of flexion/extension (f/e) and abduction/adduction (a/a) with locking or free motion in the hip joint [132]. As shown in Fig.5, four HKAF exoskeletons are designed to assist children with CP. These four wearable robots are discussed in the following.

a) HKAF exoskeleton by Eguren et al. (P-LEGS): The paediatric lower-extremity gait system (P-LEGS) exoskeleton was developed by Eguren *et al.* in 2019 to assist young children with gait disorders such as those found in the CP, SB, and SCI populations (Fig. 13 (d)) [67].

The 8 kg P-LEGS involves six actuated DOF (hip, knee, and ankle joints) in the sagittal plane and two passive DOF at the hip in the frontal plane allowing external rotation and abduction/adduction motion resulting in the weight balance. A 24V Maxon motor, a 161:1 ratio gearbox, and crossed roller bearings are included in the actuator housing at each joint. The maximum momentary peak torque at each joint is 76 Nm, while the nominal torque is 13.5 Nm [67].

The exoskeleton has a joint control module at each joint. The joint control module comprises an ARM Cortex M-4 MCU as the microcontroller, an actuator with a motor driver, a 9-axis IMU, FSRs, a rotary encoder, a Wheatstone bridge-based torque sensor to enable Assist-As-Needed control, and system monitoring sensors. Each modular joint can work separately from the other joints while connected, based on the CAN bus communication protocol [67].

Relying on the hierarchical control architecture, an Assist-As-Needed (AAN) control has been employed in the exoskeleton. The microcontroller collects the signals from the sensors for gait phase detection in the supervisory-level controller. The high-level controller is adaptive impedance control, where a PD position controller is used for the low-level controller. For each joint control module, the control parameters of the PD controller have been separately tuned [67].

In terms of clinical characteristics, children between 4-8 years old, weight (16-28 kg), and height (1-1.23 m) with a type of neurological disorder are eligible to participate in clinical studies of P-LEGS. However, some clinical trials with different participants in terms of CP type, age, CMFCS, etc. are needed to prove the effectiveness of the exoskeleton in practice.

b) HKAF exoskeleton by Andrade et al. (ExoRoboWalker): In 2019, ExoRoboWalker was developed by Andrade *et al.* to assist the motion of the hip, knee, and ankle joints of both legs in children and young adults affected by CP (Fig. 13 (e)) [20].

The ExoRoboWalker includes six actuated DOF (hip, knee, and ankle joints) in the sagittal plane. Six actuators, including EC 45 flat (70 watts, brushless, Maxon Motors) and a CSD-20-160-2a harmonic drive (reduction ratio of 160, Harmonic Drive LLC), are employed in this exoskeleton to assist the motion of the joints. The masses of each hip, knee, and ankle joint are 1.1 kg, 0.85 kg, and 1.34 kg, respectively, and the total weight of the exoskeleton is 6.57 kg. The full length of the exoskeleton is 0.95 m, with the thigh and shank parts measuring 0.31 m and 0.34 m, respectively [20].

Each actuator includes Hall effect sensors to measure the rotor speed and a potentiometer to measure each joint angle. Additionally, strain gauges in the ExoRoboWalker can monitor the interaction forces between the exoskeleton and a user. Collecting signals of pressure sensors in the insole for the gait phase prediction, the supervisory-level controller, based on the state machine, addresses three distinct states: Idle, Standing, and Walking. The high-level controller is designed based on the finite state impedance controller, where the PD torque controller is the low-level controller [20]. Unfortunately, no clinical study has been developed to show the exoskeleton effectiveness.

c) HKAF exoskeleton by Canela et al.: In 2013, Canela *et al.* developed a pediatric exoskeleton to rehabilitate the physical disabilities resulting from CP [156]. This pediatric exoskeleton has been inspired by a former adult exoskeleton [159]. Each leg of the exoskeleton has three actuated DOF to facilitate each joint movement. The ankle, knee, and hip joints have a similar actuator (EC-45 70W 24V) with a 160/1 reducer (Brushless DC motor also known as EC Motor). This exoskeleton has been developed for children with CP from 7 to 17 years old. Some pilot studies are required to confirm the usefulness of the exoskeleton [156]. To the best of the authors' knowledge, more information about the exoskeleton in terms of mechanical design, actuator type, controller structure, and clinical characteristics are not available. Moreover, the authors could not find any picture for the prototype of the exoskeleton.

4) Trunk-Hip-Knee-Ankle-Foot (THKAF) Exoskeleton: Patients who need more stability in the trunk and hip mainly use THKAF exoskeleton devices. These exoskeletons are also employed to amplify and strengthen the human muscle in paraplegic people [132]. Only one exoskeleton has been developed in this category.

a) THKAF exoskeleton by Marsi-Bionics (ATLAS): Marsi-Bionics designed the ATLAS exoskeleton to improve rehabilitation and increase the life expectancy of children affected by neuromuscular diseases (NMDs), particularly Spinal Muscular Atrophy (SMA) and CP (Fig. 13 (f)) [144], [160], [161].

The ATLAS project developed several prototypes with similar mechanical structures and small differences between each updated version [162]. The ATLAS 2020 is a THKAF exoskeleton with ten active DOF and two passive DOF. The ATLAS 2020 includes two legs joined together and a thoracic junction. The joints are interconnected with titanium tubes, and the assembly consists essentially of aluminum and titanium. Each leg includes five actuated DOF, two on the hip (flexion-extension, abduction-adduction), one on the knee (flexion-extension), and two on the ankle (flexion-extension and eversion-inversion), and one passive DOF on the hip for rotation. The three flexion joints (hip, knee, and ankle) allow movement in the sagittal plane, while the two abduction joints (hip and ankle) allow displacement in the frontal plane needed for the exoskeleton stability [144], [157]. Using lightweight materials such as aluminum and titanium, the ATLAS exoskeletons are designed with the least possible weight, while the ATLAS 2020 is 12 kg [157], and the ATLAS 2030 is 14 kg [145].

ATLAS 2020 employs two types of actuators; one for the movement in the sagittal plane (rotation drive) and the second for the motion in the frontal plane (linear drive) [144], [157]. The rotation drives are composed of a gear assembly, including a 70W brushless DC motor with a quadrature encoder, a Hall Effect sensor, and a 160:1 gear reduction ratio. The DC motor has a maximum torque of 60 Nm and a maximum angular speed of 30 rpm. A similar 70W brushless DC motor has been used for the linear drive, with pre-stage reduction 3:1 and linked to the output spindle 10×3 ball [157].

The supervisory-level controller in ATLAS 2020 is based on a National Instruments board (MyRIO) using FPGA Xilinx and a real-time processor. This master microcontroller collects the position, force, and pressure sensor signals to generate speeds, angles, and trajectories needed for each joint. The microcontroller transmits these signals via an I2C serial line to the local drivers. The low-level controller is based on a microcontroller ATMEL, and this module can support only two joints. Hence, three low-level control modules have been used per leg, and six batteries distributed in both legs of the ATLAS 2020 supply the required power of the exoskeleton [144].

ATLAS 2020 includes four operation modes: self-test mode operation, standby mode, walking mode, and assistance-as-needed mode. The exoskeleton uses a PID controller in the low-level controller for the walking mode. However, the ATLAS 2020 suffers from the lack of self-balance control; as a result, an updated version, ATLAS 2030, has been launched on the market with a self-balance control and removing the frame [144].

In a clinical study for ATLAS 2020 [163], seven children, three boys and four girls, aged between 3 to 9, were recruited. Among them, four children, two boys and two girls, aged between 3 to 6, were selected for eight clinical sessions. Unfortunately, further clinical details about these participants have not been provided in this study [163]. In order to evaluate the effectiveness of this assistive device [163]: (i) The pressure at the tibia and feet of participants have been monitored during the exercise; (ii) The number of meters walked without any need for the assistant is counted; (iii) After each session, the level of fatigue is observed; (iv) Blood pressure and respiratory rate are recorded. The clinical results show that this exoskeleton can be used at the beginning level of neuromuscular illness in order to delay the onset of complications.

ATLAS 2020 and ATLAS 2030 are designed for children from 3 to 14 years old [145], although the weight of the 12 kg ATLAS 2020 or the 14 kg ATLAS 2030 is heavy for young children, particularly a three-year-old child with an average weight of around 14 kg [164].

5) Exoskeleton-Plus-Walker: This device has an architecture similar to an exoskeleton, while a smart walker has been added to it. Since the walker results in a high degree of stability, this type of exoskeleton is appropriate for CP children with a high-level of GMFCS (IV-V) who have limited or no walking ability.

a) HK Exoskeleton-Plus-Walker by Maggu et al. (Trexo): In 2018, Trexo Robotics was developed by Maggu *et al.* [158] in Trexo Robotics (Mississauga, ON, Canada) as a safe, comfortable, and easy-to-use assistive mobile robot for children with different conditions, such as CP, TBI, SCI, etc. (Fig. 13 (g)). Due to the success of some clinical trials and research studies, the Trexo Plus will be launched on the market in 2021 [165], [158].

Trexo Plus is a combination of a walker and LLE in pediatric sizes, with four actuated DOF in hip and knee joints to support and assist users as they mobilize independently. This device has the flexibility to provide the exact level of support needed by adding or removing accessories, such as arm support, trunk support, and weight-bearing assistance. Using a tablet for remote control, an operator can manage speed, gait angles, and the amount of torque assistance of the exoskeleton. This device can be modified in height and width to suit a broad spectrum of ages from two years upwards [165], [166]. Unfortunately, more information about the mechanical design, actuator type, and control algorithm is not available to the best of the authors' knowledge.

In a clinical study, children aged from three to six were recruited to ascertain the advantages and disadvantages of training with Trexo Plus in comparison with routine clinical care, particularly for children with CP [166]. This device has been designed for CP children with moderate and severe mobility disorder (GMFCS III-V) [167]. Moreover, a seven-year-old female affected by CP with CMFCS Level (V) was recruited [168]. In order to evaluate the effectiveness of the exoskeleton, the lower extremity range of motion (ROM), postural control, and spasticity were assessed before the Trexo use and weekly to biweekly thereafter. The clinical results reveal that regular use of the Trexo plus has a positive effect on the quality and frequency of bowel movements (BMs) and may improve knee flexor spasticity and head control [168].

b) HKAF Exoskeleton-Plus-Walker by Bayón et al. (CPWalker): In 2016, a robotic platform called CPWalker was developed for children with CP to reduce their rehabilitation period after surgery, integrating Peripheral Nervous System (PNS) and Central Nervous System (CNS) into robotic-based therapies (Fig. 13 (h)) [106]. The CPWalker rehabilitation device is composed of an exoskeleton attached to a smart walker that provides balance and assistance to the user across overground walking [68].

The conceptual design of CPWalker is according to the commercially available NFWalker (Made for Movement, Norway) device with some mechanical modifications to develop a powered rehabilitation robot further than the passive NFWalker [106]. In this regard, other robotic platforms such as Active Reciprocated Gait Orthosis (ARGO) [169] and a robotic walking aid [170] have been designed based on the NFWalker. These devices have a smart walker and passive orthosis for each joint, while CPWalker enjoys not only an intelligent walker but also an actuated exoskeleton.

CPWalker has ten actuated DOF, six of which are for the HKAF exoskeleton, and four of those are for the walker. The actuator of each exoskeleton joint includes a harmonic drive coupled to a brushless flat DC motor (Maxon EC-60 flat 408057) with a high gear reduction ratio of 1:160. The exoskeleton device is outfitted with the following sensors: potentiometers, force sensors, and an insole pressure sensor. The smart walker has been equipped with the following actuators: (i) Two gear motors (Kelvin K80 63.105) with encoders coupled to each rear wheel to drive the platform. (ii) An electric linear actuator (CAHB-10-B5A-050192-AAAP0A-000) for control of a user's weight. (iii) A linear actuator (Bansbach E21BX300-U-001) is composed of a hydraulic pump and two cylinder-piston for control of hip height [105], [106].

The interaction between a user and the device takes place via a Multimodal Human-Robot Interface (MHRI), including (i) an Electroencephalographic (EEG) acquisition unit; (ii) Electromyography (EMG) system; (iii) IMU; (iv) a Laser Range Finder (LRF) to measure gait cycle and to control the robotic platform accordingly [105], [106].

Two PC-104 in MATLAB Real-Time Workshop are responsible for the control of CPWalker. The first controls the smart walker, while the second supports the exoskeleton and collects the sensor signals for the gait detection by the supervisory-level controller. A high-level controller is based on the impedance control, and a position controller using a PID controller is the low-level controller [105], [106].

To assess the effectiveness of CPWalker, some clinical trials have been conducted [68], [171]. The inclusion criteria for recruiting the patients were: (i) individuals aged 11 to 18 years with spastic diplegia; (ii) GMFCS Level (I-IV); (iii) maximum weight 75 kg; (iv) anthropometric measures of lower limbs based on the exoskeleton of CPWalker; (v) ability to understand the exercises; (vi) capability of signaling pain or discomfort. The exclusion criteria followed: (i) children who had concomitant treatments 3-months before the study; (ii) children with unhealed skin lesions in the lower limbs or muscle-skeletal deformities that could prevent the use of the exoskeleton; (iii) children with critical alterations of motor control, for instance, due to ataxia or dystonia; (iv) self-harming or aggressive behaviors; (v) severe cognitive impairment. In [68], 4 CP patients with spastic diplegia: (2 males; GMFCS Level: II(2)/III(2); age: 14.50 ± 2.38 years (range: 12-17 years)) were recruited. In order to evaluate the effectiveness of the exoskeleton during experiments, parameters such as ROM, partial body weight support (PBWS), gait velocity, and levels of assistance are evaluated for each participant in each session. Clinical results show improvement in different aspects including strength, mean velocity, and gait performance [68].

In [171], 3 CP participants with spastic diplegia (2 males; GMFCS Level: II(1)/III(1); age: 13 ± 1 years (range: 12-14 years)) were recruited. In this pilot study, ROM, PBWS, and gait velocity are assessed for each participant. After walking with the assistive device, these three children could improve their mean velocity, cadence (step/minute), and step length [171]. These clinical studies show the potential of CPWalker to serve as a rehabilitation tool.

VI. Discussion

The design and development in lower-limb robotic exoskeletons for children with CP have been systematically reviewed from a technical and clinical perspective in this paper, highlighting the lack of an in-depth review study in this area. This systematic review identified seventeen eligible studies focused on fifteen exoskeletons designed for children with CP. These studies show some consistent positive results on the efficacy of LLEs in improving gait patterns in children affected by CP. CP participants have been recruited only for eight of these fifteen LLEs. These exoskeletons provide their users with musculoskeletal rehabilitation and motor assistance. Based on the actuated joint, these wearable devices have been categorized into single-joint and multi-joint robotic exoskeleton systems. Different actuator types, including motors and gearboxes, have been explained for these assistive devices. To deliver a better understanding of the control of these robotic exoskeletons, various control approaches have been investigated. In terms of the clinical study, clinical characteristics of patients involving gender, severity of mobility disorder, type of CP, and age have been presented. Moreover, other important information, such as sensors, power transmission, power supply, etc., have been described to the best of the authors' knowledge.

A. Mechanical Design

This systematic review addresses fifteen LLEs designed for children with CP (Fig. 10), of which six exoskeletons are single-joint (Table III), seven exoskeletons are multi-joint, and two devices are multi-joint exoskeletons added to intelligent walkers (Table VI). Given crouch gait, characterized by excessive knee flexion, is the most prevalent gait disorder in CP [27], knee exoskeletons are the most popular type of exoskeletons for individuals with CP. Fig. 10 confirms this finding, where five knee exoskeletons [28]–[32] and one KAFO [66] constitute the most common type (40%) of all the exoskeletons.

Rodda and Graham [99] showed that hinged AFO is the most common clinical solution for most gait patterns in individuals with CP (Fig. 4, Fig. 5) [99], [172]. Moreover, clinical results in this review study have revealed that powered LLEs have a significant potential to improve CP children's walking capability [64], [65], [30], [68]. For individuals with a neuromotor disorder, AFO can be a replacement for inappropriate muscle function during the gait cycle. AFO supports the handling of abnormal gait patterns to optimize leg alignment [173]. However, only one motorized AFO has been developed so far [65]; as a result, more focus and research on actuated AFO is required. In addition, hip displacement is widespread in children with CP, and progressive hip displacement can cause severe pain for patients [174], [175]. Nevertheless, this paper reveals that no hip exoskeleton has been developed for children with CP, and more contribution is needed in this area.

TABLE VII
Mechanical Characteristics for Prototyped LLEs With CP Participants

Study	Device Name	Actuated Joint	Active DOF	Weight (kg)	No. of CP Participant
Lerner [65]	–	Ankle	2	1.8 (small size) 2.2 (big size)	59
Lerner [22]	–	Knee	2	3.2	54
Yamada [29]	–	Knee	2	3.52	1
Chen [30]	P.REX	Knee	2	–	1
Patané [66]	WAKE-up	KAF	4	2.5	3
Kawamoto [148]	HAL	HK	4	–	58
Maggu [158]	Trexo	HK +Walker	4	–	1
Bayón [106]	CP-Walker	HKAF +Walker	10	–	7

Furthermore, a considerable portion of children with CP has limited or no walking ability [89], [97] (Fig. 3); however, only two exoskeleton-plus-walker devices have been developed for children with CP so far (Fig. 10). Exoskeleton-plus-walker devices, which allow a high degree of stability, are suitable for users with limited or no walking ability. Therefore, more assistive mobility devices, including exoskeleton and walker, are needed to be designed and prototyped in order to support CP children with severe mobility disorders in overground walking.

The exoskeleton's weight and mass distribution significantly affect both functional performance and metabolic consumption, as studies have shown that metabolic energy consumption during walking increases with additional load mass and more distal location [49], [50], [51]. However, children with CP consume more metabolic energy during walking compared with their healthy peers [52], [53], [54]. Hence, it is critical to pay special attention to the weight of LLEs designed for children with CP as an effective mechanical parameter. The average weight of a six-year old child is 24 kg [21], and LLEs for children with CP should be much lighter than the child wearing the robot.

Table VII shows mechanical characteristics of prototyped LLEs with CP participants. The table shows that in single-joint LLEs, the lightest device is AFO by Lerner *et al.* at 1.8 kg for the small size and 2.2 kg for the large LLE, whereas the heaviest device is the knee exoskeleton by Yamada *et al.*

TABLE VIII
ACTUATORS DETAILS FOR PROTOTYPED LLEs

Actuator Type	Study	Device Name	Actuated Joint	Voltage (V)	Power (W)	Gear Ratio	Max Torque (Nm)	Manufacture Company
BLDC[1] Motor	Lerner [65]	–	Ankle	24	90* 120**	267:1 333:1	12 20	Maxon Motor
BLDC Motor	Lerner [22]	–	Knee	24	90	89:1	16.1	Maxon Motor
BLDC Motor	Chen [30]	P.REX	Knee	–	–	51:1	15.7	Maxon Motor
BLDC Motor	Laubscher [21]	–	HK	–	70	40.6:1	35.7	–
BLDC Motor	Eguren [67]	P-LEGS	HKAF	24	–	161:1	76	Maxon Motor
BLDC Motor	Andrade [20]	ExoRobo Walker	HKAF	–	70	160:1	–	Maxon Motor
BLDC Motor	Canela [156]	–	HKAF	24	70	160:1	34.3	Maxon Motor
BLDC Motor	Marsi-Bionics [157]	ATLAS	THKAF	–	70	160:1	60	Maxon Motor
BLDC Motor	Bayón [106]	CPWalker	HKAF+ Walker	–	–	160:1	–	Maxon Motor
DC Motor	Kawamoto [148]	HAL	HK	–	–	–	–	–
EM[2] Brake	Yamada [29]	–	Knee	12	–	–	7	Miki Pulley
EM Brake	Washabau-gh [142]	–	Knee	–	–	26:1	56	K&J Magnetic
Linear Actuator	Adib [32]	ExRoLEG	Knee	–	–	–	500	–
RSEA[3]	Patané [66]	WAKE-up	KAF	–	–	–	–	–
–	Maggu [158]	Trexo	HK+ Walker	–	–	–	–	–

[1]Brushless DC.
[2]Electromagnetic.
[3]Rotary Series Elastic Actuator.
*Small Size AFO.
**Large Size AFO.

at 3.52 kg. In addition, this table shows that a large number of participants were recruited for AFO and knee exoskeleton by Lerner *et al.*, while there is no data on any CP child using the heavy 8 kg knee exoskeleton by Mohd Adib *et al.* [32]. Hence, it can be concluded that 2.2 kg for AFOs and 3.2 kg for knee exoskeletons seems appropriate for children with CP.

Exoskeleton-plus-walker devices seem heavier than multi-joint LLEs without walkers; however, exoskeleton-plus-walker devices have been partly successful in recruiting CP participants. One reason is that walkers in these devices carry the weight of exoskeletons, so there is no need for users to sustain the exoskeleton's weight. For heavy exoskeletons with high DOF, using a walker is recommended to carry the exoskeleton's weight and make the device stable.

B. Actuator Type

DC motors, particularly brushless DC (BLDC) motors, are the most popular type of actuator in the literature for powered LLEs because they are lightweight, compact, reliable, and silent [37]–[39], although they have a lower power-to-weight ratio than the other types of actuators such as AC motors, pneumatic motors, and hydraulic actuators [39], [40]. In the fifteen included LLEs, nine BLDC motors manufactured by Maxon Motor, one DC motor, two electromagnetic brakes, one linear actuator, one RSEA, and one unspecific actuator have been employed to assist the mobility of devices (Table VIII). This table shows that BLDC motors are compatible with different types of joints, including ankle, knee, and hip, while electromagnetic brakes and the linear actuator have only been used for knee joints. This table shows that the minimum power of 24 V BLDC motors for these LLEs is 70 W, while the maximum power is 120 W. Of the 24V BLDC motors, four motors have 70 W, one 90 W, and one 120 W power. The minimum gear ratio is 26:1, and the maximum is 333:1, while 160:1 is the most common. The maximum torque of actuators ranges from 7 to 500 Nm, whereas for BLDC motors, it is between 12 and 76 Nm. The maximum torque exerted at the ankle, knee, and hip by typically developing (able-bodied) children in overground walking is 1.5 Nm/kg, 0.7 Nm/kg, and 0.9 Nm/kg, respectively [176]–[179]. This means that for a 30 kg child, the total toque provided by the actuator and the user should be approximately 45 Nm in the ankle joint, 21 Nm in the knee joint, and 27 Nm in the hip joint. Table VIII shows that a 24 V BLDC Maxon motor with at least 70 W and a gear ratio of 160:1 seems a reliable actuator for LLEs designed for children with CP.

C. Control Strategy

The control architecture for LLEs reviewed in this paper is based on the hierarchy control strategy. Table IX shows that Finite State Machine and impedance control are the most common supervisory-level and high-level controllers, respectively, while torque control and position control based on PID or PD are the most popular low-level controllers.

Of the fifteen LLEs included, five use FSM as the supervisory-level controller for gait phase detection, while this has not been determined for ten devices. FSM is a popular supervisory-level controller because it can integrate torque and position control as well as split the controller into multiple states based on the different phases of the gait cycle [35], [104]. Another advantage of FSMs is that they are adaptable enough to enable LLEs to work on uneven terrain and execute duties such as sitting and standing [35], [104]. However, abnormal gait patterns of children with CP make it difficult for the supervisory-level controller to detect or predict gait phases.

In the high-level controller, four impedance controllers, one adaptive impedance controller, one sigmoid function, and one Proportional Joint-Moment Control (PJMC) are used to generate the desired torque for the low-level controller (Table IX). Impedance control is commonly used in high-level control of LLEs to take advantage of patients' residual movement under

TABLE IX
CONTROL STRATEGIES AND SENSOR TYPES FOR PROTOTYPED LLES

Study	Device Name	Supervisory-Level Controller	Supervisory-Level Sensor	High-Level Controller	Low-Level Controller	Low-Level Sensor	Actuated Joint
Lerner [65]	–	FSM[1]	FSR[2]	Sigmoid Function / PJMC[3]	Torque Control (PID)	Torque Sensor	Ankle
Lerner [22]	–	FSM	FSR, Encoder	–	Torque Control (PID)	Torque Sensor	Knee
Patané [66]	WAKE-up	FSM	FSR, IMU, Encoder	–	Position Control (PID)	–	KAF
Chen [30]	P.REX	FSM	FSR, Encoder	Impedance Control / Adaptive Control	Torque Control (PID)	Torque Sensor	Knee
Andrade [20]	ExoRobo Walker	FSM	FSR, Potentio-meter	Impedance Control	Torque Control (PD)	Strain Gauge	HKAF
Marsi-Bionics [157]	ATLAS	–	FSR, IMU	Impedance Control	Position Control (PID)	Encoder	THKAF
Bayón [106]	CPWalker	–	FSR, EEG, EMG, IMU, LRF[4]	Impedance Control	Position Control (PID)	Potentio -meter	HKAF + Walker
Eguren [67]	P-LEGS	–	FSR, IMU, Encoder,	Adaptive Impedance Control	Position Control (PD)	Torque Sensor, Encoder	HKAF
Laubscher [21]	–	–	Angle Sensor, Hall Effect Sensor	–	Position Control (PD)	–	HK
Yamada [29]	–	–	FSR	–	–	–	Knee
Washabaugh [31]	–	–	EMG	–	–	–	Knee
Adib [32]	ExRoLEG	–	EMG	–	–	–	Knee
Kawamoto [148]	HAL	–	FSR, EMG, Encoder	–	–	–	HK
Canela [156]	–	–	–	–	–	–	HKAF
Maggu [158]	Trexo	–	–	–	–	–	HK+ Walker

[1]Finite State Machine.
[2]Force Sensitive Resistor.
[3]Proportional Joint-Moment Control.
[4]Laser Range Finder.

the philosophy “Assist-As-Needed” [105], [106]. Impedance control relies on the Assist-As-Needed control strategy to conveniently adjust assistance to the user’s needs, actively involve the user in the function, and increase neural plasticity [73].

Five position-control (three based on PID and two based on PD) and four torque control (three based on PID and one based on PD) have been utilized for the low-level controller to control the actuator (Table IX). Position control is commonly used in assistive robotic LLEs for clinical applications because it can provide a level of safety as the reference motion trajectory is tracked strictly [104]. This controller is especially appropriate for individuals with neurological disorders who are unable to move their lower limbs due to a lack of muscle strength [38], [113]. The low-level torque control effectively tracks a reference torque using the actuator’s electric current as the system input [117]. Torque control has a close interaction with actuators; therefore, it delivers a simple way of controlling the flow of energy from the exoskeleton to the user, which is helpful in biomechanics research [118].

Furthermore, Table IX shows sensor types for supervisory-level and low-level controllers used for gait phase detection and actuator control, respectively. At the supervisory-level, the following sensors are included: ten Force Sensitive Resistors (FSRs), five encoders, four IMU, four EMG, one EEG, one potentiometer, one LRF, one angle sensor, and one hall effect sensor. These results reveal that FSR is the most common supervisory-level sensor followed by other sensors, particularly encoder and IMU. At the low-level, three torque sensors and one strain gauge for torque controllers, and two encoders and one potentiometer for position controllers are employed. At this level, torque sensors and encoders are the most popular sensors for torque controllers and position controllers, respectively.

D. Clinical Characteristics

Although fifteen LLEs have been developed for individuals with CP (Fig. 10), only eight of them have recruited CP participants in pilot studies (Table X). Of these eight LLEs, four are single-joint, two are multi-joint, and two are exoskeleton-plus-walker devices (Table X). Half these devices, including one KAF and three knee LLEs, are designed for crouch gait. Of these, three LLEs, including Ankle by Lerner *et al.* [65], HAL [148], and Knee by Lerner *et al.* [28], have been trialed with considerably more CP participants in several pilot studies.

Table X shows that of these eight LLEs, four have two DOF, three have four DOF, and one has ten DOF. This means that LLEs with low DOF (two or four) have been more successful than devices with high DOF. Most of these devices are lightweight (1.8 kg – 3.52 kg), enabling children with CP to carry these devices. The most common actuator in these LLEs is the BLDC motor, where four BLCD motors, one DC motor, one RSEA, and one Electromagnetic brake are employed to actuate these devices. In the control structure, FSM, impedance control, and torque/position control based on PID are the most common supervisory-level, high-level, and low-level controllers in these assistive devices with CP participants.

Collectively, 184 participations were recorded for clinical trials of these wearable robots, of which 32.06% were recorded for Ankle by Lerner *et al.*, 31.52% for HAL, and 29.35% for Knee by Lerner *et al.* It should be noted that since some CP individuals may have attended several clinical trials, this study

TABLE X
PROTOTYPED LLEs WITH CP PARTICIPANTS

Reference		Clinical Characteristics					Mechanical Design			Actuation Type	Control Strategy		
Author (year)	Device Name	# CP Participants	# Male	Age (years) Range (years)	GMFCS Level (#)	CP Type (#)	Actuated Joint	Active DOF	Weight (kg)	Actuator	Supervisory-Level	High-Level	Low-Level
Lerner *et al.* (2018) [65, 123]	—	59	48	15.3 ± 7.4 (5–31)	I(30)/II(19)/ III(10)	SD[1](5)/SH[2](6)/ S[3](15)	Ankle	2	1.8[*] / 2.2[**]	BLDC[4] Motor	FSM[5]	Sigmoid Function / PJMC[6]	Torque Control (PID)
Kawamoto *et al.* (2009) [148, 149]	HAL	58	41	15.5 ± 5.8 (8–37)	I(9)/II(9)/ III(27)/IV(13)	SD(28)/SH(5)/ SQ[7](5)/S(19)	HK	4	—	DC Motor	—	—	—
Lerner *et al.* (2016) [22]	—	54	21	10.7 ± 4.1 (5–19)	I(6)/II(35)/ III(1)	SD(32)	Knee	2	3.2	BLDC Motor	FSM	—	Torque Control (PID)
Bayón *et al.* (2018) [106]	CP-Walker	7	4	13.9 ± 1.9 (12–17)	II(3)/III(3)	SD(7)	HKAF + Walker	10	—	BLDC Motor	—	Impedance Control	Position Control (PID)
Patané *et al.* (2017) [66]	WAKE-up	3	—	11.0 ± 2.6 (8–13)	—	SH(3)	KAF	4	2.5	RSEA[8]	FSM	—	Position Control (PID)
Yamada *et al.* (2018) [29]	—	1	1	18	—	—	Knee	2	3.52	EM[9] Brake	—	—	—
Chen *et al.* (2018) [30]	P.REX	1	1	15	—	—	Knee	2	—	BLDC Motor	FSM	Impedance Control	Torque Control (PID)
Maggu *et al.* (2018) [158]	Trexo	1	0	7	V(1)	SQ(1)	HK + Walker	4	—	—	—	—	—

[1]SD: Spastic Diplegia.
[2]SH: Spastic Hemiplegia.
[3]S: Spastic.
[4]BLDC: Brushless.
[5]FSM: Finite State Machine.
[6]PJMC: Proportional Joint-Moment Control.
[7]SQ: Spastic Quadriplegia.
[8]RSEA: Rotary Series Elastic Actuator.
[9]EM Brake: Electromagnetic Brake.
[*]Small Size.
[**]Big Size.

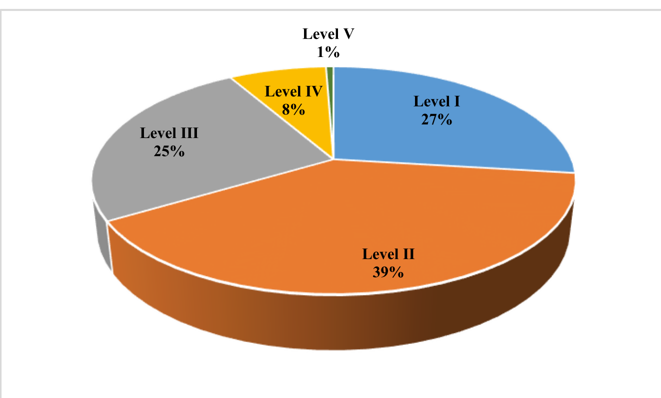

Fig. 14. Distribution of GMFCS Levels of participations in clinical trials for prototyped LLEs.

has only focused on the number of participations in the clinical trials regardless of the number of participants.

Of these 184 participations, if 17 participations with unspecific GMFCS Level are ignored, 27% participations were recorded with GMFCS Level (I), 39% with GMFCS Level (II), 25% with GMFCS Level (III), 8% with GMFCS Level (IV), and 1% with GMFCS Level (V) (Fig. 14). Generally, 20.1% of individuals with CP are categorized in GMFCS Level (V) (Fig. 2); however, only one participant with CP in all these clinical trials had GMFCS Level (V) (Fig. 14). This shows more LLEs are needed to fill open research gaps for CP individuals with severe mobility disorder, particularly GMFCS Level (V).

Of these 184 participations, if 57 participations with unspecified motor-type and topography are ignored, 57% of participations were recorded with spastic diplegia, 13% with spastic hemiplegia, 4% with spastic quadriplegia, and 26% with spasticity and unspecified topography. It seems that most of prototyped LLEs are compatible with spasticity, particularly spastic diplegia.

Fig. 15 shows the age distribution of participations for prototyped LLEs. This Fig. shows that the participation age range was from 5 to 37 years with an average age of 14.19 ± 6.28, although 11 and 12 years were the most common ages.

E. Limitation

The studies included in this systematic review were limited to Scopus and Web of Science databases. Grey literature databases were not searched, and the exclusion of non-English

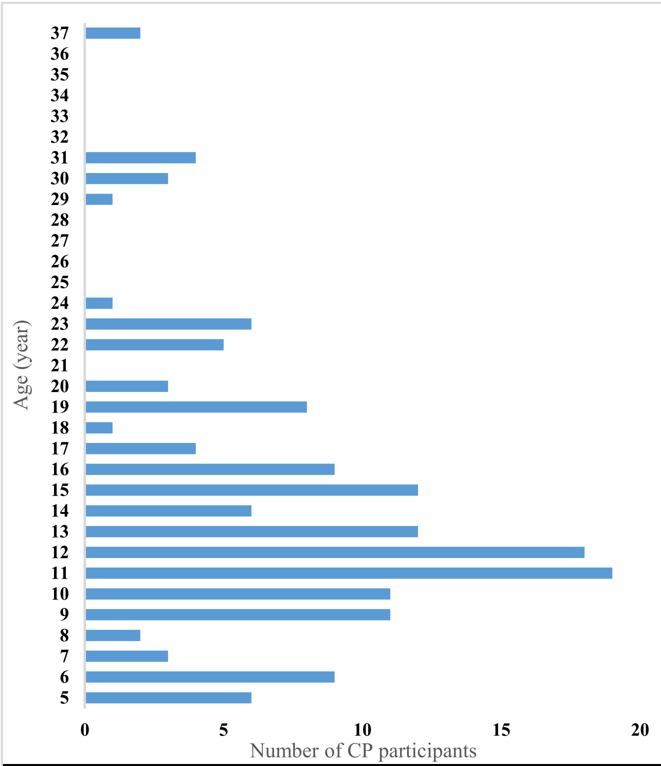

Fig. 15. Age Distribution of participations in clinical studies for prototyped LLEs.

publications further limited the scope of the literature. This paper is only limited to the prototyped LLEs for the target population of children with CP. It has not reviewed these assistive wearable devices for other neurological disorders, such as SCI, TBI, etc., which can be helpful to find the potential areas of growth for LLEs tailored for children with CP. Another limitation of the research structure employed in this review is that it excluded commercialized devices not written about in any peer-reviewed published paper explaining clinical evidence associated with functional mobility of CP participants. Finally, this interdisciplinary research is only limited to a technical and clinical perspective, while social and psychological factors have not been considered. Indeed, none of the reviewed studies in this paper reported data on psychological reaction and potential acceptance of this new technology from CP children's perspective. As a result, further multi-disciplinary investigations are required to understand more issues on this modern technology for children with CP and to address open research challenges.

VII. Conclusion

This paper provides a systematic review of the existing literature on lower limb robotic exoskeletons for children with CP to fill an open research gap in this area. After several inclusion and exclusion criteria, this systematic review included seventeen studies for further investigation. These seventeen studies addressed fifteen LLEs, of which only eight have recruited CP participants. These studies established some consistent promising outcomes on the effectiveness of LLEs in improving gait patterns in children with CP. This paper has focused on the mechanical design, actuator type, control strategy, and clinical characteristics of selected LLEs from a technical and clinical perspective.

In terms of mechanical design, results show that the most popular type of LLEs is a knee exoskeleton, generally used for crouch gait, the most common gait pattern in children with CP. Clinical studies show that hinged AFO is the most popular clinical solution for most gait patterns in individuals with CP; however, only one powered ankle exoskeleton has been prototyped for children with CP so far. Moreover, hip displacement is prevalent in children with CP, but no single-joint hip exoskeleton has been prototyped, addressing potential research and development in the future. The exoskeleton's weight and mass distribution have a considerable impact on both metabolic consumption and functional performance. When compared to their healthy peers, children with CP consume more metabolic energy in walking. Hence, it is critical to pay close attention to the weight of LLEs for children with CP. This paper shows that a large number of participants were recruited for AFO and knee exoskeleton by Lerner *et al.*, while there is no data on any CP child using the heavy 8 kg knee exoskeleton by Mohd Adib *et al*. Hence, it can be concluded that 2.2 kg for AFOs and 3.2 kg for knee exoskeletons seems appropriate for children with CP.

Since DC motors are lightweight, silent, reliable, and compact, they are broadly employed in LLEs. Statistical results show that a 24 V BLDC Maxon motor with at least 70 W and a gear ratio of 160:1 seems an appropriate actuator for LLEs prototyped for children with CP. Furthermore, Finite State Machine and impedance control are the most used supervisory-level and high-level controllers, respectively, while position control and torque control based on PID or PD are the most deployed low-level controllers. Finite State Machine is widely used for gait phase detection, while impedance control is employed to actively involves patients in walking function under the philosophy "Assist-As-Needed." Position control is broadly deployed in assistive robotic LLEs for clinical applications because it can provide a level of safety by closely tracking the reference motion trajectory. Torque control, interacting directly with actuators, is widely used for LLEs as it provides a simple way to control the energy flow from exoskeletons to users. This study showed that FSR is the most used sensor in the supervisory-level controller, followed by other sensors, particularly encoders and IMUs. In the low-level controller, encoders and torque sensors are commonly used for position and torque controllers, respectively.

Collectively, 184 participations with CP subjects were recorded in the clinical trials of these wearable robots, of whom most were recruited for Ankle by Lerner *et al.*, HAL, and Knee by Lerner *et al.* In terms of GMFCS, the majority of participations were recorded by mild and moderate mobility disorders (GMFCS Level I-III), and only a few participations had severe mobility disorders (GMFCS Level IV-V). In these clinical trials, only one child with CP had GMFCS Level V recruited by an exoskeleton-plus-walker device. However, previous clinical studies have shown that a considerable number of children with CP have severe mobility disorders. To this

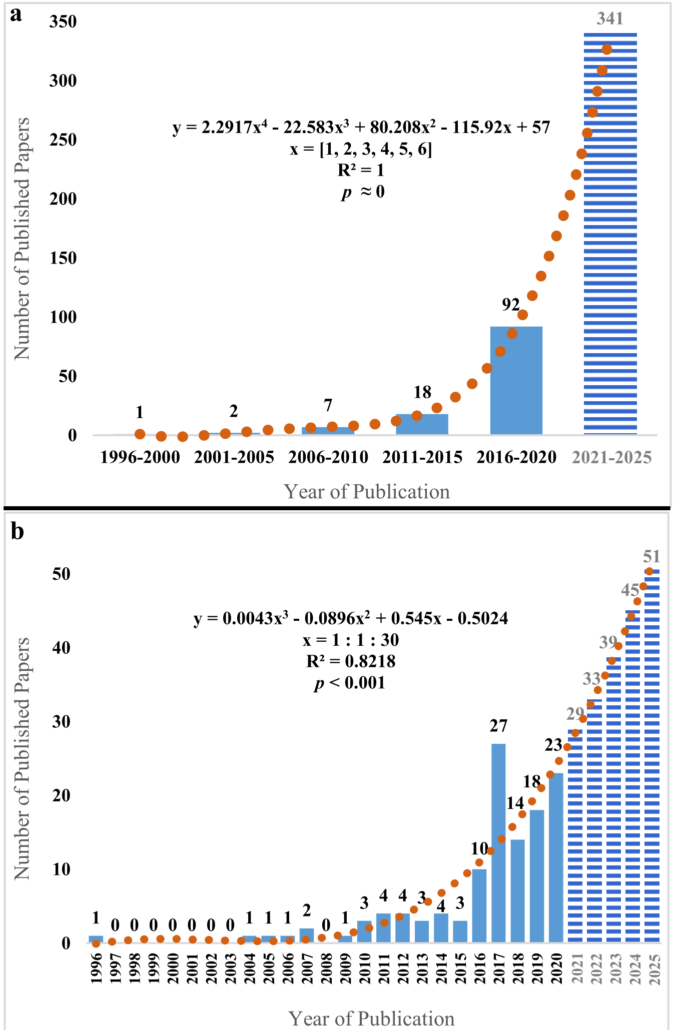

Fig. 16. Publication distribution of LLEs for children with CP in (a) 5-year intervals (4^{th} order polynomial regression; $n = 5$, $R^2 = 1$, $p \approx 0$), and (b) each year (3^{rd} order polynomial regression; $n = 25$, $R^2 = 0.8218$, $p < 0.001$).

end, exoskeleton-plus-walker assistive devices have shown a potential to support CP children with severe gait dysfunction, which can be an open research area for the future. In terms of CP motor type, all participations were recorded by spasticity, and in terms of topography, diplegia was the most common, and quadriplegia was the rarest. This suggests that prototyped LLEs are compatible with spasticity, particularly spastic diplegia. The majority of participations were recorded at 11 and 12 years old, although the participations' age range was from 5 to 37 years with an average age of 14.19 ± 6.28 years.

Although this new assistive technology is still at an early stage, the number of published papers, clinical trials, and participations in clinical trials show that it has reached relative maturity; therefore, a review study was needed to overview state of the art and recent development in this area. Despite this relative maturity, many open challenges and research gaps have remained, needed to be addressed in the future. Since the literature on LLEs for other neurological disorders, particularly SCI, is much richer than CP, researchers and designers can broadly and initially focus on all neurological conditions; then narrow their topic to focus on CP. This rich literature may help them find solutions for current challenges and identify further gaps in this area for the future.

Appendix

To predict the number of published papers in Topics 1-3 (Section III) in Scopus and Web of Science after removing duplications in the 5-year interval of 2021 to 2025, two regressions have been employed. The first is a 4^{th} order polynomial regression for 5-year intervals from 1996 to 2020 ($R^2 = 1$, $p \approx 0$) (Fig. 16 (a)) which predicts 341 papers will be published in the 2021-2025 interval. The second is a 3^{rd} order polynomial regression for each year from 1996 to 2020 ($R^2 = 0.8218$, $p < 0.001$) (Fig. 16 (b)). This regression predicts that 29, 33, 39, 45, and 51 papers will be published in 2021, 2022, 2023, 2024, and 2025, respectively. Collectively, the second regression predicts 197 papers will be published during 2021-2025. The first regression has included the extraordinary number of published papers in 2017, while the second has excluded it. Consequently, it is anticipated that from 2021 to 2025, between 197 and 341 papers will be published in this area.

Author Contributions

All authors contributed to the conception and design of the study and methodology selection. Mohammadhadi Sarajchi did the literature search, collected data, selected studies, analyzed and interpreted data, drafted the article, and made critical revisions to the article for important intellectual content. Mohammadhadi Sarajchi and Mohamad Kenan Al-Hares critically reviewed and edited the article. Konstantinos Sirlantzis was responsible for the study supervision and project funding. All authors read and approved the final manuscript.

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