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REVIEW ARTICLE

THE ADDED VALUE OF AN ACTUATED ANKLE-FOOT ORTHOSIS TO RESTORE NORMAL GAIT FUNCTION IN PATIENTS WITH SPINAL CORD INJURY: A SYSTEMATIC REVIEW

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Objective: **To provide an overview of robot-assisted rehabilitation devices developed for actuation of the ankle-foot complex and their ability to influence the attributes of normal gait in patients with spinal cord injury.**
Methods: **A search was conducted in MEDLINE, Web of Knowledge, National Academic Research and Collaborations Information System, and Physiotherapy Evidence Database (1985–2011), using, "ankle", "foot", "robotics", "orthotics" and "spinal cord injury" as most relevant keywords. Article inclusion was performed in 3 stages; at the level of: (*i*) title, (*ii*) abstract and (*iii*) full text.**
Results: **The actuated ankle-foot orthoses currently available are characterized by several combinations of an actuator and a control mechanism. Both the actuator and the control strategy substantially influence human-machine interaction and therefore the potential of the device to assist in modifying locomotor function and potentially modify the underlying motor control mechanisms.**
Conclusion: **Due to small sample sizes, limited studies in patients with spinal cord injury, and limitations in study design, it is difficult to draw firm conclusions on the effect of different types of actuated ankle-foot orthoses. Based on the limited data available, pneumatic artificial muscles in combination with proportional myoelectric control are suggested to have the potential to meet most of the preconditions to restore the attributes of normal gait and therefore facilitate neuroplasticity.**

Key words: actuated ankle-foot orthosis; spinal cord injury; rehabilitation.

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INTRODUCTION

The global annual incidence of traumatic spinal cord injury (SCI) is estimated to range from 10.4 to 83 new patients per million individuals, not including individuals dying before hospital admission (1, 2). Spinal lesion is characterized by a partial or complete transection of the ascending and descending pathways that ensure communication between spinal and supraspinal locomotor centres (3). This communication deficit may lead to systemic problems and severe and long-term deficits, including abnormal posture and locomotor dysfunction. A major component leading to motor impairment in SCI is the decrease in muscle function (4) due to muscle weakness and slowness in voluntary torque development and, to a lesser extent, the deficit in dexterity (the ability to coordinate muscle activity to meet environmental demands) (4–8). However, following SCI, and especially following incomplete spinal lesions, patients do show considerable recovery in muscle strength (8–10), locomotor independence (8, 11) and even gait function (8, 12).

Under normal circumstances, neuromotor control of gait is based on a hierarchical system in the central nervous system (13). At the level of the spinal cord, spinal central pattern generators (CPG) are defined as networks of nerve cells that generate movements. They contain all the information necessary to activate motoneurons of flexor and extensor muscles in the appropriate sequence and intensity to generate human gait (14–16). Although the CPGs are capable of generating movement independently of sensory input, the basic locomotor pattern is under the constant influence of central, supraspinal and peripheral input (13, 14, 17). In patients with SCI, input from the cerebral cortex is partly or completely deprived. In spite of the absence of supraspinal input, the timing and sequence of motoneuron activation provided by the CPGs is preserved at the spinal levels below the lesion site. Therefore, peripheral afferent input plays an even more crucial role in activating and modulating the remaining CPG activity (13, 17–19). To facilitate CPG activity through appropriate afferent peripheral input, repetitive execution of the specific task at hand and minor step-to-step variability is essential (3, 20–24). Crucial peripheral afferent input for achieving normal human gait, and thus task-specific training, was previously identified to relate to hip joint position and proprioceptive input by load receptors located in the extensor muscles and by mechanosensors in the sole of the foot, similar to normal overground walking (13, 17, 19, 25–30). Afferent input from the hip joint plays a key role in modulating the muscle activation pattern for initiating stance to swing transition during normal human gait (29, 31). The significance

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of loading for regulation of stance and gait has been confirmed in healthy subjects and patients with SCI. Through spinal reflex pathways, the load-related afferent input provided by the extensor muscles and mechanosensors in the sole of the foot contribute to the adaptation of the locomotor pattern to the ground conditions (29, 31, 32). This task-specific training is often achieved during treadmill training (33–35). Two forms can be distinguished: (*i*) body-weight support treadmill training (33–35), and (*ii*) robot-assisted gait training (RAGT) (36–39). Currently, the common approach for RAGT is the application of a mechanically driven gait orthosis (DGO) to provide active guidance of the hip and knee joint according to a predetermined kinematic trajectory. However, most of these devices largely neglect the ankle-foot complex (AFC), despite its crucial role in contributing to normal human gait function. Based on the documented importance of afferent input of load receptors located at the level of the AFC in the process of gait recovery, the extension of a DGO with ankle-foot actuation might further facilitate motor recovery in patients with SCI. This review focuses on the potential added value of isolated ankle-foot actuation in restoring the attributes of normal walking, and therefore on their role in facilitating motor recovery in patients with SCI. Furthermore, this review critically evaluates the added value in a DGO and discusses the feasibility of this approach in optimally synchronizing the actuation of multiple joints of the lower limb orthosis.

METHODS

A comprehensive search of a selection of the English, German, French and Dutch literature was conducted through multiple databases (MEDLINE, ISI Web of Knowledge, Physiotherapy Evidence Database (PEDro), National Academic Research and Collaborations Information System (NARCIS) and Cochrane Controlled Trails Register), using keywords combined through Boolean operators (“AND”, “OR” and “NOT”) (Table I).

The selection procedure used to compile a list of appropriate surveys comprised 3 stages. The first stage was a selection based on the presence of predetermined keywords and/or keyword combinations in the title (Table II). In the second stage the abstracts of the relevant publications retained from stage one were evaluated by two independent researchers based on contents, using appropriate inclusion and exclusion criteria (Table III). The third stage involved a methodological quality assessment of the relevant studies by two independent researchers (Table IV). The quality of the individual studies was assessed through a set of generic core items for quality assessment derived from “The evaluation of descriptive research” by Ball et al. (40). The quality assessment scale was modified for descriptive studies and appended with a list of self-afflicted criteria. Two researchers scored the studies independently. In case of conflicting opinions, a consensus was reached by negotiation. If no consensus was achieved, a third independent observer made the final decision. Cohen’s kappa was used to test inter-rater reliability between the two evaluators. The consecutive stages of the selection procedure are summarized in Fig. 1.

RESULTS

Methodological quality assessment

Initially there was disagreement between the two raters about inclusion or exclusion of the studies based on abstracts for 6 of the 36 items, resulting in a Cohen’s kappa score of 0.80.

Table I. *Keywords and combinations of keywords used in the search. The terms within the columns are allied with “OR”, words from different columns are combined using the Boolean operator “AND”*

Key words				
Actuation	Foot	Orthosis	Gait	Rehabilitation
Assisted	Ankle	Orthotics	Locomotion	SCI
Driven	Drop foot	Exoskeleton		
	Lower limb	Robotics		
	Lower extremity			

SCI: spinal cord injury.

Methodological quality assessment by two independent raters resulted in an initial disagreement over 43 of the 456 items in the quasi- and pre-experimental trials, resulting in a Cohen’s kappa of 0.79 (Table IV). After using the consensus method, the mean score on the quality evaluation of the intervention studies equalled 22/40 (standard deviation (SD) 3.14).

Descriptive assessment

An overview of the most important parameters described in the studies included in this review is presented in Table IV.

1) Actuated ankle-foot orthosis design: description of actuator and control system

- *Actuator*. In the actuated ankle-foot orthoses (AAFO) design, two types of actuators are used: pneumatic artificial muscles (PAM) and series-elastic actuators (SEAs) (Table V).

  PAM produce joint torque by (de)pressurizing the pneumatic muscle (41). PAMs are built up by an inflatable inner bladder sheathed with a double helical weave (42). When filled with pressurized air the artificial muscle expands and shortens, producing force (43). The SEAs are an actuator built up by a spring in series with a motor (electrical, hydraulic, pneumatic or other traditional servo system) and produce torque by activating the motor, resulting in a linear movement (44).

- *Control systems*. Control systems manage the actuation of the AAFO based on real-time data collected during gait.

Table II. *Key words for inclusion and exclusion based on the title*

Inclusion criteria	Exclusion criteria
Actuation	Surgery
Assisted	Passive orthosis
Drive	Electrical stimulation
Orthotic	
Exoskeleton	
Robotics	
Foot	
Ankle	
Lower extremity	
Gait	
Locomotion	
Rehabilitation	
Drop foot	
Spinal cord injury	

Table III. *Inclusion and exclusion criteria for assessment based on contents*

Inclusion criteria	Exclusion criteria
Adult (≥ 18 years) population of healthy subjects	Child population
Adult (≥ 18 years) population of acute or chronic complete or incomplete SCI patients (cervical, dorsal and lumbar)	Population of neurological patients other than patients with SCI
Assessment of the influence of ankle-foot actuation on gait performance based on biomechanical and physiological outcome measures	Application of body weight support systems with or without RAGT
The design of an active AFO (isolated as well as integrated in a driven gait orthosis)	Assessment of the influence of ankle-foot actuation on gait performance exclusively based on clinical outcome measures
Studies comparing the influence of different control methods used to actuate the ankle-foot	Animal population
	Application of functional electrical stimulation
	Application of a passive AFO

SCI: spinal cord injury; AFO: ankle-foot orthosis; RAGT: robot assisted gait training.

This data can relate to gait parameters, kinematics, kinetics or myoelectric signals. Using this information as input, the controller provides an appropriate output signal to the actuator driving the orthosis. To date, 5 types of control strategies have been predominantly used: (*i*) on/off control, (*ii*) proportional myoelectric control (PMC), (*iii*) position control, (*iv*) explicit force/torque control, and (*v*) impedance control (Table IV).

An on/off control system generates a real-time signal to control the torque provided by the actuator (41, 45). An on/off control system manages the torque production of the actuator based on the signal sent by a footswitch, a force sensor or a push button. When the analogue signal coming from a footswitch or a force sensor exceeds a predefined threshold the pressure regulator provides for maximal torque (46). The push-button controller exhibits linear behaviour, proportionally to the displacement of the plunger (47).

Proportional myoelectric control (PMC) is a physiologically-inspired control system, using the subject's own surface electromyography (EMG) to control the timing and magnitude of the force produced by the PAM (48–51).

Position control uses the patient's instantaneous position and velocity as a feedback signal in order to achieve a desired/ideal position profile in time by adapting the actuator force/torque (i.e. the position profile of a healthy subject, of the patient during unactuated walking with the AAFO or the mean position profile of a healthy population).

Explicit force/torque control aims to apply a desired force/torque by means of the actuators applying the patient's own joint force/torque as a feedback signal.

Impedance control is an implicit force/torque control, in which the applied force/torque is related to the deviation from a target trajectory, through a desired, adjustable mechanical impedance. The force/torque resulting from this impedance drives the orthosis towards the target trajectory.

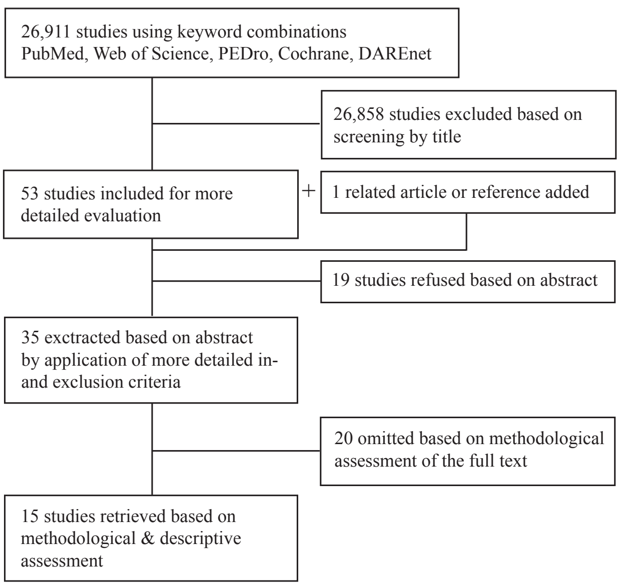

Fig. 1. Selection procedure for studies included in the review. PEDro: Physiotherapy Evidence Database; DAREnet: Digital Academic Respositories.

2) Effect of the different designs in actuated ankle-foot orthoses on gait. The orthotic designs published in the literature can be subdivided into 4 categories

- *Ankle-foot actuation through PAM and SEA controlled by an "on/off" controller.* Gordon et al. (45) and Sawicki et al. (47) combined PAMs with an "on/off" control system, in the form of a footswitch and a handheld push-button, respectively. In healthy subjects, walking with an AAFO based on footswitch control system in combination with PAM results in hip, knee and ankle joint kinematics similar to normal overground gait kinematics following an adaptation period (45). In patients with SCI, plantarflexion assistance improves the ankle kinematics at push-off (increased ankle plantarflexion), but decreases the hip range of motion at low walking speed. These changes in joint kinematics are accompanied by a decrease in activation amplitude of the *M. soleus* and the *M. rectus* femoris (50). Plantarflexion assistance provided by a therapist-controlled push-button increases the ankle range of motion, but decreases the hip range of motion during push off in patients with SCI. In case of push-button control handled by the patient, the overshoot in plantarflexion torque generated by the AAFO decreases compared with therapist control (47).

When the on/off control (footswitch) is combined with a SEA the AAFO principally induces joint kinematics similar to normal overground walking in healthy subjects (52).

Table IV. *Descriptive assessment of the literature*

Reference	Methods quality	Population, *n*	Intervention: Protocol	Intervention: Treadmill speed	Outcome: Outcome measures	Outcome: Results
Kao et al. 2010 (53)	25/52	11 (6♀, 5♂) HS	2 sessions (72 h): 1) Pass. orthosis (10 min) 2) Act. orthosis: no DF actuation (30 min) 3) Pass. orthosis (15 min)	1.25 m/s	3D kinematics 3D joint angles EMG Muscle activity Mechanical performance Amount of assistance	3D kinematics ↑ PF ankle (SwP & initial HS) (D1 & D2) Kinetics A, K & H kinetics (D1 & D2) EMG ↓ EMG ampl. M sol. Adaptation Steady-state faster (D2 vs D1)
Kao & Ferris 2009 (55)	29/52	10 (5♀, 5♂) HS	2 sessions (72 h): 1) Pass. orthosis (10 min) 2) Act. orthosis: no DF actuation (30 min) 3) Pass. orthosis (15 min)	1.25 m/s	3D kinematics 3D joint angles EMG Muscle activity Mechanical performance Amount of assistance	3D kinematics ↓ PF ankle (PO) (D1) (CC) ↑ DF ankle (mid-late SwP) & = PF ankle (PO) (D1) (SC) EMG ↓ EMG ampl. M tib. ant. & ↑ EMG ampl. 4 M (D1) (CC) response EMG ampl. M tib. & 4M (SC) Mechanical performance Peak DF torque = 0.22 ± 0.14 Nm/kg (initial HS) & 0.12 ± 0.09 Nm/kg (SwP) (CC) Peak DF torque = 0.11 ± 0.06 Nm/kg (SwP) (SC) Adaptation Steady-state D1 & D2 (ankle angle correlation common variance) (CCvsSC) Steady-state faster D2 Swing VS Continuous control (M tib. ant. EMG BURST 1) (CCvsSC)
Kinnaird & Ferris 2009 (43)	29/52	10 (6♀, 4♂) HS	2 sessions: 1) Pass. orthosis (10 min) 2) Act. orthosis: no DF actuation (30 min) 3) Pass. orthosis (15 min)	1.25 m/s	3D kinematics 3D joint angles Gait parameters EMG Muscle activity	Kinematics ↑ peak PF angle (init.) Time till baseline = 6.2 ± 5 min (30 min) EMG ↑ M activity & ↓ EMG RMS M gastr. med. (init) ↓ M gastr. med. activity (12% vs] baseline) & ↓ M sol. activity (26% vs baseline) (30 min)
Sawicki & Ferris 2009 (50)	26/52	3 (♂) HS	1) Pass. AKAFO (10 min) 2) Pass. AKAFO (7 min) 3) Act. AKAFO PMC (7 min) 4) Act. AKAFO PMIC (7 min)	1.25 m/s	Kinematics Joint angles Kinetics GRF EMG Muscle activity Exoskeleton mechanics	3D kinematics Ankle DF PMC > PMIC EMG = M sol. & M tib. ant. (PMC vs PMIC) Exoskeleton mechanics Peak PF/DF force & torque PMC > PMIC Mechanical power PMC < PMIC
Sawicki & Ferris 2009 (58)	32/52	9 (4♀, 5 ♂) HS	D1 & D2 1) Practice 90 min Step length (0,3; 1,0;1,2;1,4 X PSL) 2) Pass. AAFO (7 min.) Step length (0,3; 1,0;1,2;1,4 X PSL) 3) Rust (3 min.)	1.25 m/s	Kinematics Joint angles Kinetics GRF EMG Muscle activity	3D kinematics ↑ PF ankle (early stance & PO) (> unpowered) Gait kinematics (↑ step length) ↑ step width & ↓ double support time EMG ↑ act. M triceps surae & M tib ant (= unpowered)

Table IV. *Contd.*

Reference	Methods. quality	Population, *n*	Intervention: Protocol	Intervention: Treadmill speed	Outcome: Outcome measures	Outcome: Results
			4) Act. AAFO (7 min.) Step length (0,3; 1,0;1,2;1,4 X PSL)		Metabolic cost VO_2 & VCO_2 Exoskeleton mechanics	Exoskeleton mechanics ↑ peak ankle joint angular velocity & ↑ peak exoskeleton mechanical power (PO) ↑ absolute (+) mechanical power
Sawicki & Ferris 2008 (48)	30/52	9 (5♀, 4♂) HS	1) Pass. orthosis (7 min) 4 ≠ surface incline (0%, 5%, 10%, 15%) 2) Act. orthosis (7 min) 4 ≠ surface incline (0%, 5%, 10%, 15%)	1.25 m/s	3D kinematics 3D joint angles Angular velocity Gait parameters EMG Muscle activity Exoskeleton mechanics	3D Kinematics ↑ ankle, knee en hip extension angle (early stance phase) Gait parameters =step width, step length, step period & double support period EMG ↓ act M sol & M gastrocnemius med. & lat. Mechanics exoskeleton Peak exoskeleton torque ≈ 33% peak ankle joint moment (level unpowered walking)
Sawicki & Ferris 2008 (49)	25/52	9 (5♀, 4♂) HS	1) Pass. orthosis 2) Act. orthosis 3) Pass. orthosis	1.25 m/s	3D kinematics 3D joint angles Gait parameters Kinetics GRF EMG Muscle activity	3D kinematics Ankle, knee & hip kinematics ≈ unpowered condition (SPh & S) Gait parameters = EMG ↓ amplitude M soleus (SPh)
Cain et al. 2007 (41)	30/52	12 (6♀, 6♂) HS	1) Protocol controller 1/ controller 2 (72 h) a) Pass. orthosis (10 min) b) Act. orthosis (30 min) c) Pass. orthosis (15 min)	1.25 m/s	3D kinematics 3D joint angles Kinetics GRF EMG Activation patterns	3D ankle kinematics Sign. Δ ankle, knee & hip & ↑ PF ankle (toe off) (SC & CC) EMG SC & CC: ↑ muscular act. (SPh) & Δ muscle activation pattern SC & CC: adaptation period= SC: ↓ soleus & gastrocnemius activation
Gordon & Ferris 2007 (51)	24/52	10 (5♀, 5♂) HS	1) No orthosis 2) 2 sessions (72 h apart): a) Pass. orthosis (10 min) b) Act. orthosis: no DF actuation (30 min) c) Pass. orthosis (15 min)	1.25 m/s	3D kinematics 3D joint angles Gait parameters Kinetics GRF EMG Activation patterns	3D ankle kinematics ↑ peak PF (D1) =norm. ankle kinematics (D2) EMG ↑activation M7 &=M sol act (D1) ↓ Δ EMG activity (M soleus) (D2)
Norris et al. 2007 (56)	30/52	16 (8♀, 8♂) HS	D1 Preferred walking velocity (overground) D2/D3 Preferred walking velocity (overground/ treadmill) a) Standard athletic shoes b) AAFO inactive c) AAFO active	Self	PWS 1) Overground 2) Treadmill Gait performance Metabolic cost AAFO performance Power & energy AAFO	PWS ↑ PWS Gait performance =metabolic cost of transport & energy per stride AAFO performance Peak PF power=0.63±0.13 W/kg – 0.46±0.12 W/kg

Table IV. *Contd.*

Reference	Methods. quality	Population, *n*	Intervention: Protocol	Intervention: Treadmill speed	Outcome: Outcome measures	Outcome: Results
Gordon et al. 2006 (45)	26/52	3 ♂ HS	1 or 2 pneumatic muscles for PF 1) 4 different speeds (0,5; 1,0; 1,5 & 2,0 m/s) 2) 5 different situations a) no orthosis b) 1 passive PM c) 1 active PM + 1 passive PM d) 2 passive PM e) 2 active PM	0,5 m/s 1,0 m/s 1,5 m/s 2,0 m/s	3D Kinematics 3D joint angles Gait parameters Kinetics GRF Gait performance Metabolic cost	Properties PM Peak force (1PM) = 1700 N (100% length) & 0 N (0% length) 3D Kinematics ↑PF angle ankle & = hip & knee angle Kinetics Peak PF moment = 57% (net ankle PF force during stance phase) Mech. performance Work orthosis < work PM & = work PM (≠ situations)
Hwang et al. 2006 (52)	20/52	5♂ HS	1) No orthosis 2) Unilateral plas AFO 3) Unilateral act. AFO		3D kinematics 3D joint angles	3D kinematics DF & PF ankle ≈ norm. Kinetics ↑ PF torque (midstance – terminal stance) & = pelvic obliquity/ rotation Peak PF torque = 20 Nm/kg (pre-swing)
Sawicki et al. 2006 (47)	28/52	5 (3♀, 2♂) SCI	1) 2 different levels of BWS (T & P) (30–45 min) 2) 4 different speeds a) No orthosis b) Bilateral pass AAFO c) Bilateral act AAFO (T) d) Bilateral act AAFO (P)	0.36 0.54 0.72 0.89	3D kinematics 3D joint angles Gait parameters Kinetics GRF, Mech. properties EMG Muscle activity	3D kinematics ↑ ankle angle (PO) (vs no AAFO (9º) & pass AAFO (7º)) & ↑ PF ankle (PO) > high speed (T vs P) Gait parameters ↑ time double support (T vs no AAFO & pass AAFO) EMG ↓ activation M sol (T) & ↑ activation M tib ant (P) Power AAFO Peak PF torque 0.38 ± 0.03 Nm/kg (T) & 0.33 ± 0.02 Nm/kg (P)
Ferris et al. 2005 (54)	20/52	1 HS	1) No orthosis (6 min) 2) Pass. orthosis (6 min) 3) Act. orthosis (DF actuator loss (30 min)) 4) Act. orthosis (PF actuator loss (30 min))	1.2 m/s	3D kinematics Joint angles EMG Muscle activity	3D ankle kinematics ↓ peak PF & Peak DF torque = (init vs 30 min) EMG ↑ ampl. M tib. Ant. (80% irt pass. orthosis) (init vs 30 min) = ampl. M soleus (110% irt pass. orthosis) (init vs 30 min)
Blaya & Herr 2004 (57)	25/52	2 DF 3 HS	Zero, constant & variable impedance 1) Self-selected speed 2) 25% ↑ 3) 25% ↓	1) Self 2) 25% ↑ 3) 25% ↓	3D kinematics 3D joint angles Gait parameters Kinetics GRF	3D kinematics ↓ swing DF angular range (VI < CI) Gait parameters Asymmetry length & time (VI < CI)

HS: healthy subjects; pass.: passive; act.: active; 3D: 3-dimensional; EMG: electromyography; PF: plantarflexion; DF: dorsiflexion; D1: day 1; D2: day 2; CC: continuous control; SC: swing control; M: muscle; tib.ant.: tibialis anterior; vs: versus; init.: initial; gastr. med.: gastrocnemius medialis; sol: soleus; PMC: proportional myoelectric control; PMIC: proportional myoelectric inhibition control; act.: activation; lat.: lateral; GRF: ground reaction force; mech.: mechanical; T: therapy; P: patient; SCI: spinal cord injuries; VI: variable impedance; CI: continuous impedance; norm: normal; AAFO: actuated ankle-foot orthosis; PSL: step length; self: self-selected; AKAFO: actuated knee-ankle-foot orthosis; BWS: body weight support; VO_2: oxygen consumption; VCO_2: carbondyoxide production; SwP: swing phase; sol: soleus; ampl.: amplitude; PO: push off; BURST: summated electrical response; RMS: root mean square; irt: in relation to; ≈: similar to; plas: plastic; A: ankle; K: knee; H: hip; ≠: different.

- *Ankle-foot actuation through PAM controlled by PMC.* Plantarflexion- and dorsiflexion-assisted walking results in an initial adaptation of the gait pattern in healthy subjects unfamiliar with AAFO. Ankle-foot actuation induces an overshoot of the actuated joint angle and decreases the joint angle excursion in the opposite direction (41, 43, 49, 51, 53–55). Furthermore, a decrease in agonistic and an increase in antagonistic muscle activation amplitude is observed (41, 43, 49, 51, 53–55). Only a limited number of studies reported the time necessary to fully adapt to external actuation of the AFC in healthy subjects. The period necessary to adapt to plantarflexion- and/or dorsiflexion actuation through pneumatic muscles in combination with PMC, differed significantly between sessions, varying from 14.1 to 25.0 minutes (41, 43, 49, 51, 53, 54) and decreased significantly for consecutive sessions (41, 43, 51, 54).

  Despite the fact that the powered AFO replaces part of the ankle torque, healthy subjects show ankle, knee and hip kinematics similar to normal gait following this adaptation period (41, 43, 45, 51, 53–56). When plantarflexion is applied bilaterally, assistance results in an increased plantarflexion angle during early stance and at push-off compared with normal walking (49, 51). These differences in ankle joint angle can be attributed to the increased mechanical performance achieved through bilateral plantarflexion actuation instead of unilateral assistance (45, 49, 53). Following the adaptation period, the muscle activation amplitudes return to values comparable with normal overground walking, except for the muscle providing the control signal for the ankle-foot actuation. These muscles remain below their normal overground gait value (41, 43, 51, 53–55). AAFO providing dorsiflexion actuation are scarce (55). Dorsiflexion actuation of the AFC using the EMG signal of the *M. tibialis* anterior in healthy subjects was applied through two types of PMC: (*i*) continuous control (CC) and (*ii*) swing control (SC). CC provides active dorsiflexion assistance both at heelstrike and during swing, whereas SC supplies dorsiflexion assistance during swing (55). Both control strategies resulted in gait adaptation, though joint kinematics and muscle activation patterns generated by the SC correspond best to normal gait (55).

- *Ankle-foot actuation through SEAs controlled by impedance control.* Blaya & Herr (57) used SEAs in combination with impedance control for RAGT in patients with drop foot. SEAs with constant impedance control, better control excessive plantarflexion angle in patients with drop foot. However, SEAs, with variable impedance control, resulted in a better temporal and spatial symmetry between the affected and unaffected side in patients with drop foot during dorsiflexion-assisted walking (57). No studies describe the initial response to this type of AAFO during the adaptation period, nor is there information on the influence of AAFO on gait performance in healthy subjects.

3) Integration of ankle-foot actuation in a DGO

Extending the ankle-foot orthosis to a knee-ankle-foot orthosis presents additional challenges. Despite the potential value of PMC to achieve joint kinematics similar to normal overground walking in an ankle-foot orthosis, its application in a knee-ankle-foot orthosis results in an increased flexion pattern at the knee and ankle joint, in combination with excessive *M. soleus* and *M. tibialis* anterior activation amplitudes (45, 48–51, 53, 58). The addition of a flexor inhibitor algorithm to the standard PMC reduces this artificial muscle co-activation and produces ankle and foot joint kinematics and kinetics similar to normal overground walking (50). There have not yet been any publications on the implementation of an AAFO actuated by SEAs into a DGO.

DISCUSSION

Rehabilitation devices targeting gait re-education in patients with incomplete SCI should act mainly on the attributes of human walking by providing afferent input at the hip position and load receptors of the foot resembling normal treadmill walking. In addition, they allow the minor step-to-step variability necessary to achieve plasticity of the CPGs. Despite progress in the field of rehabilitation robotics, recent research is not conclusive on the additional value of RAGT for patients with SCI (36). This might be attributed to the fact that currently designed DGOs solely actuate the knee and hip joint, thereby neglecting the need to restore afferent input from the load receptors (36). The purpose of this review is to focus on the potential added value of isolated ankle-foot actuation on the attributes of normal walking known to facilitate motor recovery in patients with SCI. Furthermore, we will critically assess its added value in a DGO.

Characteristics of actuator and control types in AAFO

Two different actuator types (PAM and SEAs) and 5 types of control mechanism (PMC, on/off control, position control, explicit force/torque control, and impedance control) for the different AAFO are reported in the literature. Both actuator types present inherent compliance, attributable to the compressibility of air in the PAM and to the presence of an intentional spring, in series with a motor, in SEAs. Despite the fact that both actuator types are characterized by a low-weight high-power output and provide for inherent compliance, they present specific differences: in view of the autonomy of the driven gait orthosis, SEAs is preferred over PAM for overground walking. The exoskeletons using PAM as an actuator are limited to laboratory use as they require a large source of compressed air. SEAs, on the other hand, use batteries integrated in the exoskeleton for power supply, allowing autonomous overground walking.

When comparing the different control mechanisms applied in AAFO, on/off control, position and force/torque control have a clinical advantage as they use the patient's own gait parameters (i.e. heelstrike), joint kinematics and force pattern as an input signal to determine the orthosis output. When the patient deviates from the "ideal joint position/force output" or is unable to initiate the movement, the actuators are activated and guide the AFC to the "ideal" joint position or force/torque. As these control types do not require gait initiation by

Table V. *Design of actuated ankle-foot orthosis*

Design	Authors	Weight, kg	Uni/Bilateral	Controller	Actuator	DoF
PMC+PM	Kao et al. (2010) (53)	1.08 ± 0.09 kg	Unilateral	PMC (M. SOL)	AFO: PM	PF
	Kao & Ferris 2009 (55)		Unilateral	PMC (M. TA) + FS	AFO: PM	DF
	Kinnaird & Ferris 2009 (43)	1.23 kg	Bilateral	PMC (M. GM)	AFO: PM	PF
	Sawicki & Ferris 2009 (50)	2.9±1.3 kg	Unilateral	PMC (M. SOL) & PMIC	AFO: PM	PF
	Sawicki & Ferris 2009 (48)	1.18±0.11 kg	Uilateral	PMC (M. SOL)	AFO: PM	PF
	Sawicki & Ferris 2009 (58)	1.18 kg	Uilateral	PMC (M. SOL)	AFO: PM	PF
	Sawicki & Ferris 2008 (49)	1.21 kg	Uilateral	PMC (M. SOL)	AFO: PM	PF
	Gordon & Ferris 2007 (51)	1.2 kg	Unilateral	PMC (M. SOL)	AFO: PM	PF
	Ferris et al. 2005 (54)	1.6 kg	Unilateral	PMC (M. SOL & M. TA)	AFO: PM	DF/PF
PMC/FS+PM	Cain et al. 2007 (41)	1.1 kg	Unilateral	PMC (M. SOL) (C1) + FS (C2)	AFO: PM	PF
FS+PM	Sawicki et al. 2006 (47)	1.09±0.15 kg	Bilateral	Pushbutton (P/T)	AFO: PM + Elastic cord (DF)	PF
	Gordon et al. 2006 (45)	1.3–1.7 kg	Unilateral	FS	AFO: 1 of 2 PM	PF
PC+PM	Norris et al. 2007 (56)		Bilateral	Angular velocity control	AFO: 1 PM	PF
PC++SEAs	Hwang et al. 2006 (52)		Unilateral	Position control	AFO: SEAs	DF/PF
			Unilateral	4 FSR sensors		
FC+SEAs	Blaya & Herr 2004 (57)	2,6 kg	Unilateral	Force control	AFO: SEAs	PF

DoF: degrees of freedom; AFO: ankle-foot orthosis; PC: position control; FC: force control; PMC: proportional myoelectric control; FS: footswitch control; PM: pneumatic artificial muscle; M. TA: M. tibialis anterior; M. SOL: *M. soleus*; M. GM: *M. gastrocnemius medialis*; DF: dorsiflexion; PF: plantar flexion; EMG: electromyography; SC: swing control; CC: continuous control; SEAs: series of elastic actuators; C1: controller1; C2: controller2; P/T: patient & therapist.

the patient him/herself and are able to provide assistance-as-needed, these types of control systems lend themselves excellently for application in complete (no EMG activity) as well as incomplete SCI patients (minor or distorted EMG activity). As PMC uses the patient's own EMG signal as a feedback signal for orthotic control, PMC is limited to application in a patient population showing some recuperation in muscle activity following SCI.

When evaluating the different types of control systems from a clinical perspective, it is obvious that PMC, as well as force/torque and position control allow for a more natural response in orthosis dynamics compared with on/off control. Firstly, the abrupt transition from no/minimal torque to maximal torque in the on/off control results in a non-human orthosis output. Secondly, the input signal to the control system is not related to the patient's own joint kinematics or muscle activation. Thirdly, on/off control is a non-compliant control type, excluding the integration of step-to-step variability in the orthosis output and reducing human-robot interaction. Torque/force control and position control, on the other hand, are non-compliant control systems; though, by adding an additional controller, variability can be introduced. PMC closely approaches the physiological functioning of the nervous system and provides for a graded response in orthosis dynamics. Finally, PMC allows for step-to-step variability enhancing motor learning.

Influence of actuator and control type on primary factors enhancing neuroplasticity

The primary factor that contributes to gait recovery in patients with SCI, and thus to appropriate activation and modulation of the CPGs, is appropriate peripheral afferent input related to hip joint position and load receptors, providing proprioceptive input from the leg extensor muscles and exteroceptive input from the mechanoreceptors at the level of the foot (23, 25–29, 59). Results indicate that walking with an AAFO provides hip and knee joint kinematics closely resembling normal overground walking for healthy subjects, regardless of the control-actuator combination used to drive the ankle-foot. This means that, irrespective of the AAFO design, actuation of the ankle-foot does not influence hip joint kinematics (41, 43, 45, 51, 53–56). During the adaptation period, ankle-foot actuation results in a temporary change in joint kinematics and muscle activation (41, 43, 49, 51, 53–55). In subsequent training sessions the duration of this adaptation period decreases. The shorter adaptation time for subsequent sessions suggests healthy subjects formed and stored internal models of system dynamics (i.e. a lasting representation of limb dynamics when wearing the AAFO) for locomotion. Since the purpose of an AAFO is to provide adequate afferent input related to hip joint position and load receptors at the level of the extensor muscles and the foot, the adaptation time should be decreased to a minimum. A short adaptation time suggests that the actuator-control combination more closely approaches the physiological process and simplifies determining the relationship between muscle activation and orthosis assistance, and thus facilitates motor learning. Following the initial adaptation period, healthy subjects achieve sagittal plane kinematics more closely approaching normal overground walking in all actuator-control combinations (51). These changes in joint kinematics are accompanied by muscle activation amplitudes returning to values comparable to normal overground walking, except for the muscle for which the EMG signal is used as control signal. The activity of these muscles decreases below their normal overground gait value (41, 43, 51, 53–55). If similar mechanisms can be assumed in patient populations, this mechanism has the potential to contribute to a more appropriate afferent input from the mechanoreceptors at the level of the AFC (51).

Despite the fact that AAFOs contribute to afferent peripheral input similar to normal overground walking, it does not always produce functional meaningful afferent feedback. During dorsiflexion actuation, for example, the AAFO provides for pressure at the plantar surface of the foot, and thus contributes to inappropriate afferent input during the swing phase. This inappropriate afferent input is inherent to the system used. An additional factor contributing to inappropriate afferent input is the restriction to 1 degree of freedom, differing from normal ankle joint kinematics during normal overground walking.

Although both methods improve joint kinematics, PMC in combination with PAM induces joint kinematics more closely resembling a normal kinematic pattern compared with an "on/off" control system (41). A possible explanation for these differences could be the fact that the PMC is a more physiologically inspired control system that more closely resembles the normal physiological control used by the nervous system to generate motion. Consequently, ankle-foot actuation through PMC might be experienced as a relatively minor change compared with normal control, whereas a footswitch provides for a more non-natural substitution for neuromotor control (41). A second possible explanation is that the "on/off" controller evokes too much ankle torque, due to the fact that the level and timing of the control system is inadequately tuned (41). When applied for dorsiflexion actuation, PMC through swing control is preferred over continuous control: with swing control inducing joint kinematics and muscle activation patterns best corresponding to normal gait (55).

The second critical factor in enhancing normal neuromotor control of walking is the presence of a critical level of step-to-step variation (60, 61). In the AAFO designed step-to-step variation can currently be introduced in two ways: (*i*) at the level of the control system and (*ii*) at the level of the actuator. Compared with an "on/off" and a position control system, PMC lends itself excellently to achieving step-to-step variability. Proportional myoelectric control uses the subject's own EMG signal to control the assistance provided by the AAFO. The rationale for the implementation of a patient's own weak and disordered signal to guide an AAFO is based on the principals of plasticity of the CPGs. Proportional myoelectric control of PAM on a robotic orthosis provides a means of amplifying the consequences of the electrical signal sent to the muscles, making it easier for the nervous system to detect performance error and alter the subsequent electrical commands to the muscles (46, 62–66). At the level of the actuator, PAMs as well as SEAs are characterized by an intrinsic compliance. The beneficial effect of performance errors on the functioning of CPGs is proven (60, 61), although, to date, no clear consensus exists on the relative compliance of the different systems available, nor on the critical amount of step-to-step variability necessary to achieve maximal known adaptation. Therefore, it is difficult to judge which combination will provide the largest amount of adaptation.

Despite the large number of AAFO described in the literature, studies evaluating their ability to satisfy the attributes of walking in patient populations are limited (47, 67). Preliminary test results in incomplete SCI patients show that plantarflexion assistance through PAM, in combination with an "on/off" control system, contributes to joint kinematics more closely resembling joint kinematics during normal overground walking. This might contribute to instantaneous improvement in afferent input from mechanoreceptors at the level of the AFC during push-off. On the other hand, plantarflexion actuation results in a decrease in hip range of motion at low walking speed, amounting to a less favourable situation for CPG activation. The additional load of the AAFO is possibly attributable to this. Some concerns can be formulated in relation to the application of AAFO for the restoration of normal human gait in patients with SCI. First, the isolated AAFO is only suitable for a limited patient population characterized by minor muscle weakness in the lower extremities or with an isolated drop foot. For more severe SCI patients, the AAFO might serve clinical goals if implemented in a DGO, mutually actuating the knee and the hip joint. The implementation of ankle-foot actuation in a DGO entails the creation of a closed chain at the level of the lower limb, leading to additional conflicts. When the lower limb is actuated at the level of the hip and knee joint, but no ankle-foot actuation is implemented, minor irregularities or minor deviations from the hip and knee joint trajectory can be compensated for at the level of the foot. When the foot is restricted to a single-dimensional specific joint pattern, this might cause a conflict at the level of the hip and knee joint. Therefore, the inclusion of an AAFO in a DGO needs to be considered carefully. Although, in healthy subjects an AAFO does not affect hip or knee kinematics this should be evaluated explicitly in patient populations with motor impairments.

There are several questions that require further investigation. Instead of solely evaluating the instantaneous effect of ankle-foot actuation, future research should focus on the evaluation of both the short- and long-term influence of gait rehabilitation training. Research could focus on the application of different AAFO in specific groups of patients with SCI, in order to determine whether improvement is actuator- and/or control-specific and to determine from which AAFO a particular patient benefits most. The implementation of an additional functional, clinically relevant parameter, in the form of plantar pressure measurements, might allow conclusions to be drawn about changes in foot enrolment/foot-to-ground contact and consequent sensory input. With the exception of two research groups (67, 68), most AAFO currently restrict the ankle-foot actuation to 1 degree of freedom, simplifying the ankle joint to a hinge joint and considering the foot as a single rigid body. This is in strong contrast with the functional anatomical perspective of the AFC, which consists of the talocrural, talocalcaneonavicular and subtalar joints, which each allow 3-dimensional movements. Furthermore, the coupling of the individual bones occurring at the level of the synovial and syndesmotic junctions of the foot affects foot enrolment and may change foot-to-ground contact. Therefore, in order to enhance correct afferent input, the implementation of extra degrees of freedom might be advisable.

In conclusion, the mutual interaction of the actuator and control type applied in a robotic exoskeleton has a substantial influence on human-machine interaction and the modification of normal locomotor function. For that reason, both the actuator type and the control system should be taken into account when drawing definite conclusions on the most appropriate design to achieve and stimulate optimal recovery. Despite the shorter adaptation time and similar muscle activation encountered in

AAFO driven by PAMs and controlled by an "on/off" system, PAM in combination with PMC meets most of the attributes for normal human walking. In future it might be interesting to determine the instantaneous influence of AAFO, as well as the short- and long-term influence of RAGT on gait performance in different SCI patient populations.

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