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NOVEL TECHNOLOGY AND CONTROL FOR ANKLE-FOOT ORTHOSIS IN FOOT DROP: A REVIEW

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Abstract

Foot drop is one of the most common causes of movement disorder. Various designs of ankle foot orthosis (AFO) have been developed in the last few decades. These devices aim to reduce the impact of impairments in the lower limb neuromuscular motor system on gait. Existing AFO devices include passive devices with fixed and articulated joints, semi-active devices that modulate damping at the joint, and active devices that make use of a variety of technologies to generate power to move the foot. This paper reviews a variety of designs with their relevant actuating mechanisms, control strategies, characteristics, advantages, and drawbacks with a futuristic intelligent ideal AFO.

Index Terms: Foot Drop; Ankle-Foot Orthosis; Actuator; Gait Cycle; Artificial Neural Network(ANN); Brain Computer Interface (BCI); Electromyography (EMG)

1. INTRODUCTION

Individuals with neuromuscular disorders often have residual deficits and require interventions to improve gait and mobility[3].Foot drop often results from damage to the peroneal nerve supplying to the muscles of the foot and ankle[4].This reduces the patient's walking capacity, which greatly impacts their daily activities [5].Spasticity, contracture, and weakness can also result in a reduction in walking speed, an elevation in energy cost, and an increased risk of falling. Plantar flexor muscle weakness results in a reduction of push-off power and the incapability of the dorsal muscle to lift the foot adequately in mid-swing, resulting in toe drag and foot slap just after heel strike[4].Muscle weakness/paralysis results from medical conditions such as stroke, brain injury, spinal cord injury polio, muscular dystrophy, and diabetes[6, 7].

The gait cycle begins from the initial contact of the heel and ends with the following heel contact. While walking, the stance phase is defined as the stage during which the foot touches and then leaves the ground, which includes five steps: (a) initial contact; (b) loading response; (c)mid-stance; (d) terminal stance;(e) pre-swing. The swing phase is the stage when the same foot has no contact with the ground. This phase, which constitutes 40% of the walking cycle, is composed of three sub-stages: (a) initial swing; (b) mid-swing;(c) terminal swing[4].At the initiation of the gait cycle, impact forces are dissipated as the energy is absorbed by the soft tissues at the heel as the foot comes into contact with the ground[8]. The phases during stance are controlled plantarflexion (CPF: initial contact till maximum plantarflexion), controlled dorsiflexion (CDF: maximum

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plantarflexion till maximum dorsiflexion), and powered plantarflexion (PPF: maximum dorsiflexion till toe off) [9]. The ankle plantar flexor torque generated at push-off results in the highest power output for any joint during walking and is the primary source of power for forward propulsion[10]. In normal gait, the ankle joint, shank, and foot play important roles in all aspects of locomotion, including motion control, shock absorption, stance stability, energy conservation, and propulsion. As a result, pathology or injury affecting the ankle joint has the potential to significantly impact the quality of life by impairing some or all functional aspects of gait[2].Biomechanical deficits of the lower extremities and their related pathologies affect joint mobility and muscle activity. The efficiency and effectiveness of gait depend on joint mobility and muscle activity, which is selective in timing and intensity [6]. Orthotics are included in rehabilitation plans to improve function by adjusting joint deformities, optimizing structural alignment, providing support, and assisting motion at a joint. Ankle-foot orthotics (AFO) are often recommended to minimize gait dysfunction caused by abnormal muscle tone or weakness [11].It is a mechanical device used to support and align the ankle and foot to assist the weak and paralyzed muscles and help to improve their functionality[12].

The ideal orthosis for the compensation of muscle weakness should be flexible enough to accommodate both plantar and dorsiflexor weakness. For optimal function, the orthosis would control the deceleration of the foot at the start of stance, permit free ankle plantar flexion with mild resistance while maintaining ankle and knee stability up to mid-stance, generate an assistive torque at terminal stance, and block plantarflexion during the swing to prevent foot drop. All of these actions need to be accomplished using an orthosis that is compact and light to minimize the energetic impact on the wearer [2].The development of light, compact, efficient, powered, and untethered AFO systems has the potential to yield significant advancement in orthotic control mechanisms and new clinical treatment strategies for rehabilitation and daily assistance. However, despite the recent advances in computing, sensing, and other enabling technologies, there are currently no practical portable powered AFO systems in existence[2].

In general, there are two types of AFO devices: passive devices and active devices. Passive devices contain no control or electronics, but they can have mechanical elements such as springs or dampers to control the motion of the ankle joint during gait. Fully active devices have an on board or tethered source of power, one or more actuators to move the joint, and sensors with a computer to control the application of torque during gait[2].When worn, an AFO can make an immediate, short-time improvement in the energy cost of walking [13].

2. PASSIVE DEVICES

Passive AFOs make up the bulk of the devices prescribed by clinicians to treat weakness at the ankle joint complex; however, the passive nature of these AFOs limits their functional benefit. A passive orthosis acts like braces and prevents foot drop in the swing phase by maintaining the foot position at an angle of 90° and prevents toe drag and foot

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slap. The long-term use of this type of AFO leads to the weakening and atrophy of muscles. This causes increased dependency on orthosis daily[14].

Passive AFOs can be divided into articulated or non-articulated devices. Depending on the material, they can be further subdivided into metal and leather, thermoplastic, composite, and hybrid systems [15, 16](Fig. 1a).

Articulated mechanical hinges are incorporated to limit the motion while using spring it resist and assist motion. Metal and leather systems accommodate the volume fluctuation in the limb, although they are not cosmetically appealing. Plastic and carbon fibre composites are light, expensive, and cannot accommodate volume fluctuation.

A posterior leaf spring AFO is a single piece made out of polypropylene, co-polymer thermoplastic, or thermosetting resin with carbon composite(Fig.1b).Investigations are going on in the fields related to 3D scanning, 3D printing, and computer-aided designing for the manufacturing of a customized AFO with modular elements with localized composite reinforcement[17]. The first step is the acquisition of geometrical data from the patient's foot, followed by data conversion and import into a CAD modeler; in this study, Solid Works™ is chosen for this purpose.

The AFO was modelled parametrically around the foot mesh and optimized in order to resist the predicted mechanical stresses. Finally, the device was 3D-printed on an FDM printer[18].The model fit was then tested for mechanical properties, vibration analysis, and flexibility of the AFO[19].

The consensus drawn is that the best approach would be represented by a combination of advanced computational models and experimental techniques, capable of being used to optimally mimic real-life conditions[20].

Motion control properties of this AFO control using the trim line posterior to the ankle axis[21, 22].Hybrid AFOs have been designed with thermoplastic shells with articulated joint and passive motion control elements such as DACS and pneumatic springs[22] (Fig 1c**).**Another hybrid AFO was designed with the principle of harvested energy for motion control[23, 24].

It uses a passive pneumatic element actuated by the subject's body weight for motion control of the ankle joint [25] (Fig 1d).An oil damper was used to control ankle motion by absorbing energy from the system [26].

The damper provides a variable resistive force (5–14 Nm) and can be adjusted for subject-specific needs by changing the physical parameters of the damper. The stiffness of these passive AFOs range from a few Nm up to 20 Nm of resistive torque over a 30°-range of motion[22, 27].

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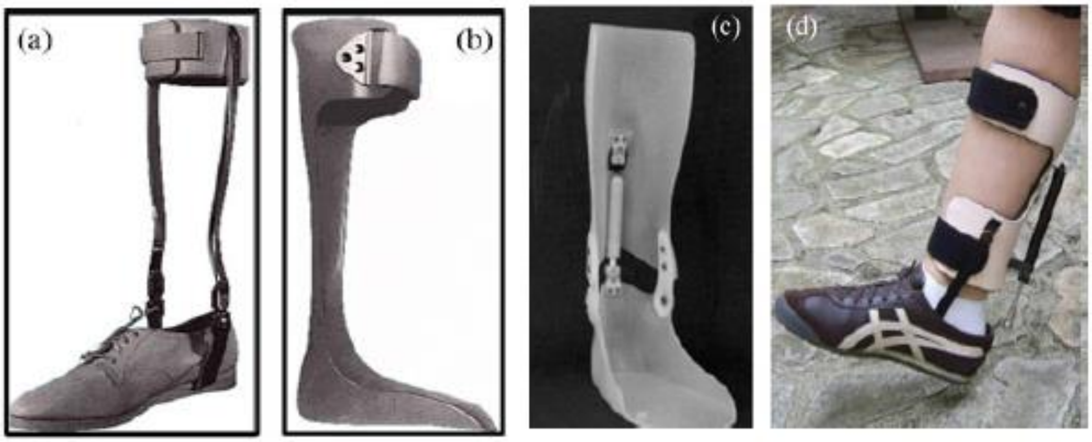

Fig 1: Passive AFO designs. (a) Metal and leather AFO. (b) Posterior leaf spring AFO. (c) Hybrid AFO with a linear spring and thermoplastic AFO. (d) Pneumatically assisted AFO[16, 22, 25, 28]. This figure is reproduced from Shorter, Xia [2] with permission

Though the hybrid AFO is found to be better than a pure passive AFO in terms of controlling motion, the latter does not only provide power assistance during the propulsive phase but also acts in an open loop system while walking. A gait control strategy for AFOs involves the control of the mechanical properties, such as the control of the bending stiffness, damping stiffness, assistive torque, and motion path[29].Active power AFO takes the advantage of this situation and addresses this limitation.

3. ACTIVE DEVICE

It is incorporated with a control system that not only provides power assistance but controls the impedance in the ankle joint during walking. Additionally, the active performance of the assisting orthosis helps re-educating the neuro-motor system and expedites the rehabilitation process. The current active system relies on sensor feedback to determine both the task and functional assistance of the actuators. In the recent past, various types of actuators have been developed.

3.1 Magneto-Rheological Fluid

Magneto-rheological (MR) fluid is a solution carrying magnetic metal particles in a carrier fluid, usually oil. The viscosity of this fluid changes rapidly when a magnetic field is applied[30] (Fig. 2a**).**The controlling system of this device is such that after dorsiflexion movement, a current is transmitted through a coil, and upon generating a magnetic field and viscosity change for MR fluid, the brake gets activated[30].Swenson and Holmberg[31]fabricated an AFO using an angle measuring sensor to detect different walking conditions like stairs ascending and descending using an MR damper. Naito et al.[4]developed an AFO prototype in which a servomotor is used to change the distance between the pipe with MR fluid and a magnetic field. The Halmstad AFO incorporates on board position sensing, power, and electronics to create an untethered device [30]

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(Fig.2b). An active control algorithm along with an MR damper allows the AFO to assist with stair climbing, inclined walking, and level walking. The AFO controller is a finite-state machine with four states: damped, free, locked, and free down (limited damping to allow movement during stance and swing). The transitions between the states are determined by the position of the ankle angle and the direction of the angular motion [31].Though MR dampers can control the motion but cannot generate torque at the ankle for push-off.

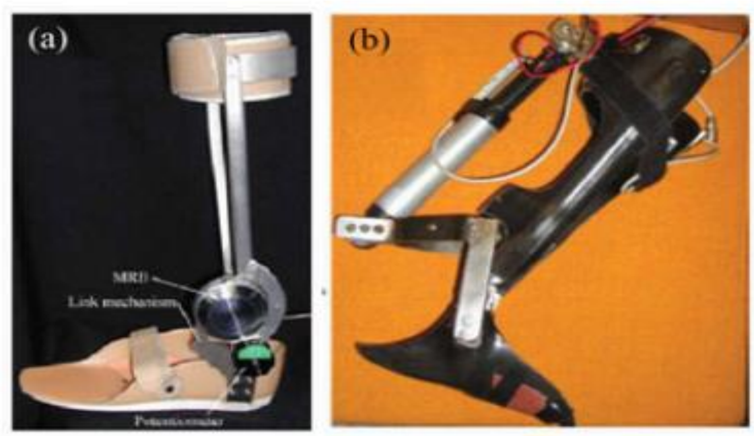

Fig 2: Semi active AFO designs. (a) The Oksaka magneto theology (MR)damper AFO and torque amplifying mechanical linkage. (b) The Halmstad magneto rheology (MR) damper AFO. This figure is reproduced from Shorter, Xia [2] with permission

A multifunctional AFO was developed. The key component of this orthosis is a novel MR actuator and a magnetic circuit. The actuator can transmit both bi-directional changeable assistant and resistant torque, whereas the magnetic circuit of the actuator can be simulated using ANSYS[32].A 1-degree-of-freedom linear-motion system is capable of regulating its viscosity via a magneto-rheological fluid and an electromagnetic coil. The link is also integrated with a compression spring that allows it to store and release energy based on its coil current[33]. A passive controllable AFO (PICAFO) system, such as the gait detection method, with a small-scale actuator design and controller, was developed to achieve personalized gait treatment goals; this utilizes MR brake to control the walking and gait based on ankle velocity reference, for which the body mass index and walking speed are taken into calculation [34].

3.2 Series Elastic Actuator

It provides both motion control and plantarflexion torque during gait. The SEA consists of a DC-motor-powered ball screw mechanism in series with a helical spring. The computer-controlled motor adjusts the rotary compliance of the AFO by driving a lead screw to vary the height of the spring [35] (Fig. 3).The impedance of the built device can also be changed considering the walking phase by dividing the walking stages into three different phases. Using an adaptive controlling system, theground reaction force and angular position data from on board sensors were used to transition between states during

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walking. Despite acceptable experimental results, this device cannot be used for rehabilitation of persons afflicted with foot drop due to its weight (2.6 kg) and large battery since it makes it impossible for the patient to sit[36].

Fig 3: The MIT active AFO powered by a series elastic actuator. From [27, 30, 31, 35]. This figure is reproduced from Shorter, Xia [2] with permission

3.3 Robotic Tendon

It uses a motor/screw/spring arrangement to offer greater compliance than a direct drive system [37]. This AFO also uses increased elasticity to harvest energy from the gait cycle, reducing both average and peak motor power requirements, which in turn results in a reduction in motor size and weight. Researchers at Arizona State University, Phoenix, U.S., have also built this AFO; the robotic tendon allows motion in the sagittal plane and utilizes an encoder, potentiometer, and a force sensor embedded at the heel for sensor feedback. The control algorithm described in[38] accommodates gait initiation and cessation and allows the device to accommodate different levels of walking speed(Fig. 4**).**This earlier design used a 0.95-kg tendon that required 77Wof power to produce a torque comparable to a healthy individual during level walking[39]. An updated AFO design uses a lighter tendon of 0.5 kg and weighs 1.75 kg [40]. This design uses a seven-state finite-state machine to control the stiffness or the velocity of the AFO. In this design, a digital incremental encoder is used to control the position of the motor. An absolute angle encoder and foot switches in the heel and toe of the AFO foot plate help to switch between states. The first five states occur during stance and alternate between stiffness and velocity control. While this AFO is still tethered to an external power source and computer, the researchers suggest that the device could be powered for 8 hr of continuous operation using a battery worn in a fanny pack.

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Fig 4: Robotic tendon [38, 40]. This figure is reproduced from Shorter, Xia [2] with permission

Fig 5: FES Wail Aid [41]

3.4 Functional Electrical Stimulation

FES devices use small surface electrical stimulation signals to stimulate the peroneal nerve to activate the ankle dorsiflexors to provide functional toe clearance during swing. These devices are customized to the individual, using trial-and-error methods during the initial fitting. Bionic Walk Aide[42] and the NESS L300[43]are commercially available FES devices (Fig 5).The Bionic Walk Aide uses a tilt sensor to monitor the orientation of the shank and initiates surface FES stimulation when the tilt sensor passes through a set threshold (indicating the onset of swing)[41]. The NESS L300 uses a force-sensitive resistor placed under the foot to detect swing.

3.5 Shape Memory Alloy (SMA)

The most highlighted characteristic of SMA is its ability to bear large plastic strains and subsequently recover the strains upon removal of the load or under heat. It is a lightweight and flexible actuator, also called “muscle wire”. This and electro-active polymer actuators (EPAs) both show potential as future actuation methods for a portable AFO [44]. Van Kuren[45] used shape memory alloy actuators to design a wearable AFO for rehabilitating cerebral palsy patients with spastic paralysis. A conceptual design was also developed by Esfahan in 2007 with a control algorithm to prevent foot drop [44].

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A shape memory-activated device (SHADE) utilizing the property of pseudo-elastic behaviour was developed by Pittaccio et al. [46-48].The control was based on the EMG signal. Due to the non-linear behaviour of shape memory alloys (SMA), it had a long response time and complex control. Therefore, implementation was challenging. In 2010, using Ni–Ti wire, two rotary actuators were developed and used in an AFO. In this system, software was used to analyse the electromyographic signals from TA muscle and control the orthosis in line with providing the trigger for its activation [49].

In 2012, SMA wire Ni–Ti was used as an actuator with variable stiffness. It is in accordance with the impedance change of an ankle while walking. To establish the required movement in this design, 14pulleys were used to pass 90 inches of SMA wire with a diameter of 0.25 mm through. The response time of this prototype was about 8 sec, which is quite less compared to the sample manufactured by Bhadane and Deshpande[50] (Fig.6), which responds in 30 seconds. Zhang et al.[51-53]used SMA-based artificial muscle, which combines two articulated plastic shells with two artificial muscles. To reduce the response time, airflow was supplied using a vessel and a mini pump.

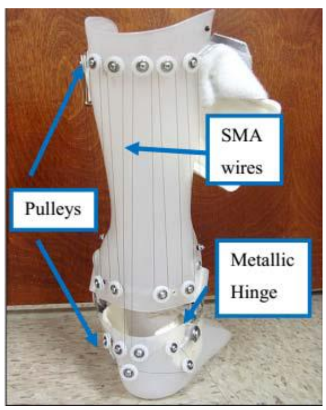

Fig 6: Shape memory alloy[50]

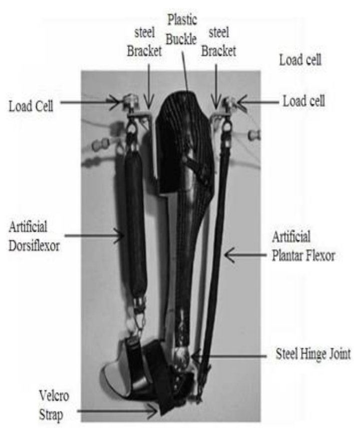

Fig 7: Pneumatic artificial muscle [12, 54, 55]. This figure is reproduced from Shorter, Xia [2] with permission

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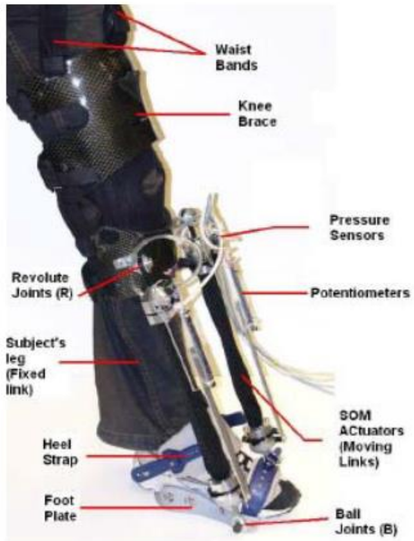

Fig 8: Robotic gait trainer This figure is reproduced from Shorter, Xia [2] with permission

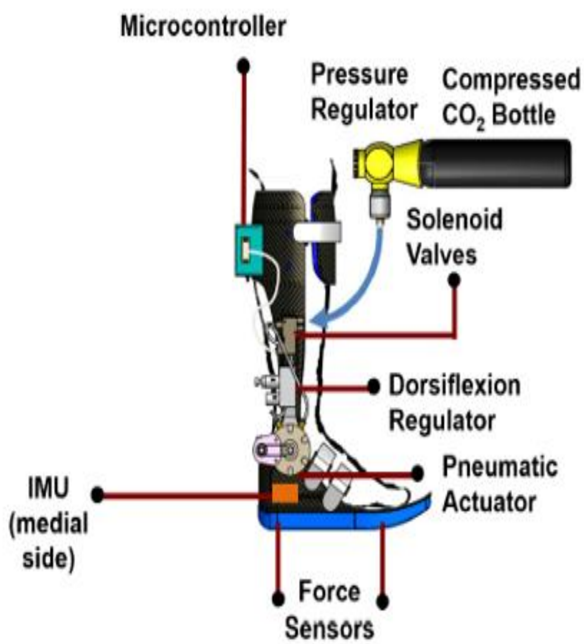

Fig 9: Pneumatic actuator [1]

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3.6 Pneumatic Actuator

Ferris et al.[56]developed pneumatic artificial muscle (McKibben artificial muscle), which was commercialized. The AFO had two artificial muscles to have both dorsiflexion and plantar flexion. Other designs were developed based on the pneumatic muscle, from 2006 to 2010 [54, 55] (Fig.7).Park et al. [57, 58] used four pneumatic muscles to develop a bio-inspired active soft orthotic device, which provides all four major movements at the ankle joint. It is made of plastic and composite material and various types of sensors, such as strain sensor inertial measurement unit (IMU) and pressure sensors. Moreover, they provided the possibility for designing and fabricating AFOs with higher degrees of freedom and lighter weights by introducing elastomeric artificial muscles composed of several miniature McKibben actuators and positioned inside an elastomeric monolith sheathe made of hyper-elastic composites [57].Hirai et al. presented a novel design for an AFO equipped with a passive pneumatic actuator system [25].Researchers at the University of Michigan used FES to control McKibben-style uniaxial artificial pneumatic muscles in various arrangements to provide dorsiflexor and plantar flexor torque. The AFO has a total weight of 1.6 kg excluding the off-board computer and air compressor. The orthosis provides a peak plantar flexor torque of 70 Nm and a peak dorsiflexor torque of 38 Nm [56]. The rehabilitation AFO, called the robotic gait trainer, built at Arizona State University, utilizes pneumatic spring-over-muscle (SOM) actuators to create bidirectional force[59] (Fig 8). The SOM actuators enclose a cylinder plunger containing a compression spring (K = 1.40 N/mm) within a McKibben-style pneumatic muscle. Li and Hsiao [1] developed a portable powered ankle-foot orthosis(PPAFO) that uses a pneumatic bi-directional rotary actuator powered by compressed CO_2 to create an untethered device (Fig.9).

3.7 Hydraulic Actuator

This uses comparatively heavy and transfers higher forces to produce plantar flexion and dorsiflexion. Here, an electric motor drives a water-filled hydraulic master cylinder that is connected to a slave cylinder mounted posterior to the shank. A classic PID controller was used by Noel et al.[60].It produces several torque profiles: constant torque, position-dependent torque, and phase-dependent torque. Additionally, it can be used to produce a high-velocity displacement disturbance at the ankle joint for the study of proprioceptive reflexes during human locomotion. The motor delivers a continuous torque of 70 Nm and a peak torque of 98 Nm to the master cylinder. The weight of the AFO worn by the subject was kept at 1.7 kg by locating the electric motor away from the device. It was further enhanced by Houle, who designed a power transmission system and made it a prototype [61](Fig 10).

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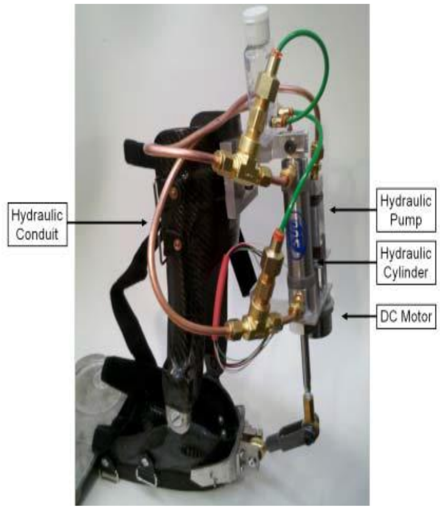

Fig 10: Hydraulic actuator from [12, 61]

4. ANKLEBOT AND REHABILITATION ROBOTICS

The AnkleBot, designed by researchers at the Newman Laboratory, Massachusetts Institute of Technology, has been used for both rehabilitation and direct measurement of the passive stiffness of the ankle joint complex [62, 63]. The device is actuated by two DC-motor-powered linear actuators mounted to the front of the shank using a knee brace and a footplate. Tethered devices are suitable for laboratory research and clinic-based rehabilitation treatments that aid in recovery from pathology or injury[64].Rehabilitation and diagnostic AFOs have been used as training devices to help restore normal walking function, instruments for the measurement of motion and force at the ankle joint, and locomotion studies. Mostly, the pneumatic and hydraulic actuators, which are heavy and are tethered with power and computation, are used for this purpose. As they are heavy and bulky and have a short battery life, AFOs powered by dielectric elastomeric actuators (DEAs) were developed [65].

The prototypes are mathematically modelled as a two-degree-of-freedom mass–spring system. The unknown disturbances are modelled as a finite sum of sinusoidal signals with unknown amplitudes, frequencies and phases, and an unknown constant. The back-stepping control algorithm is designed for the force input supplied to the system, and the stability of the equilibrium point is proved[66]. With the 1-DOF(degree of freedom) PAFO(powered ankle-foot orthosis), the stability deteriorated compared to that with the 2-DOF PAFO[67]. The study is the first attempt to show that, together with the assistive torque profile, also the stiffness level of a compliant ankle actuator can influence the assistive performance of a powered AFO[68].

In recent years, various types of robots have been developed for lower limb rehabilitation. Generally, these robots can be divided into two categories: exoskeleton and end-effectors robots[69]. For example, Lokomat [70], BLEEX [71], and LOPES [72, 73]are typical exoskeleton robots, while Rutgers Ankle [74] and Haptic Walker [75] are end-effectors

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robots. According to their mechanisms and rehabilitation principles, exoskeleton robots can be grouped as treadmill-based devices and orthosis-based robots, while the end-effectors robots have footplate-based and platform-based types. Table 1 describes various types of actuators with their features.

Table 1: Most leading designs of AFO

AFO	Actuating element	Moment (Nm)Max	Structure	DOF	Motion Control	Mechanism	Ref
AFO with MR Damper	MR Fkuide	24	rigid	1	sensor	Resistive	[31]
AFO with SEA	SEA	60	rigid	2	sensor	Assistive	[36]
AFO with Robotic tendon	McKibben pneumatic	PF 110 DF 53 Eversion 21 inversion 20	soft	4	sensor	Assistive	[38]
AFO with SMA	SMA	18	rigid	1	EMG	Assistive	[46, 47]
AFO with Passive pneumatic	Passive Pneumatic	4	rigid	1	Mechanical	Resistive	[54, 55]
AFO with hydraulic	Hydraulic	20	rigid	1	sensor	Assistive	[61]

5. CONTROL STRATEGIES

Control strategies reviewed in this section can generally be divided into four categories: position tracking control, force and impedance control, bio-signal-based control, and adaptive control.

5.1 Position Tracking Control

Gait is essentially a periodic motion. Pre-programmed patterns that may be adjusted as a function of the stride time and information about the current kinematic/kinetic state have been proposed to mimic the ankle behaviour. Oymagil et al.[38, 76] used a trajectory tracking controller for an active orthosis actuated by a linear motor with a series elasticity element. This low-level controller has a proportional-derivative (PD) structure and uses an adjustable gait pattern as a reference for the motor position. The reference pattern is generated by a polynomial fit of a normal-gait pattern and is a function of the stride time[77]. The system determines the duration between two heel strikes and adjusts the polynomial function accordingly. Ward et al.[78]reported some performance improvements in terms of gait speed, maximum ankle moment, and peak power.

5.2 Force and Impedance Control

A finite-state machine is proposed to regulate the mechanical impedance of the orthosis. The first state covers the period between heel strike and midstance. In this state, the associated controller simulates a linear-rotary spring at the ankle with variable stiffness. The control algorithm, based on the analysis of the GRF at the moment of forefoot impact, modifies the stiffness of the simulated rotary spring to prevent foot slap. The second state

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goes from mid-stance to toe-off. As foot drop does not affect the stance period, the actuator is set to zero force to not disturb the gait of the user during this phase. The last state corresponds to the swing phase of walking. For this state, the ankle is modelled as a linear second-order under-damped mechanical system with a PD position controller [36, 79].

5.3 Bio-Signal Base Control

Bio-signals contain useful information about human limb movements. They enable the robot to be controlled more naturally by using EMG signals recorded from the participant's muscles. It has been found that a considerable correlation exists between EMG signals, limb movement, and muscle activities [80].EMG-triggered and continuous control are two typical EMG-based strategies[81, 82]. EMG signals are generated before limb muscle contraction, so it can be used to predict the movement intention in advance[83]. For example, Krebs et al.[84]proposed a performance-based progressive robot control mode, which allows the patient to move the limb without assistance first, and when the EMG value reaches certain threshold, the robot assistance gets triggered. EMG-based continuous control methods have been developed recently to improve the patient-active performance. Song et al. [85]developed a myoelectric control robotic system to provide continuous stretching assistance torque whenever the EMG signals are generated. The provided assistance driven by myoelectric signals are controlled to be proportional to the amplitude of EMG signals[86].

5.4 Adaptive Control

Adaptive impedance control is more likely to achieve better rehabilitation effects for its ability to make the robot's behaviour more flexible and adjustable to the patient's capabilities, progress, and participation. In the movement ability-based type, the patient's movement ability can be estimated from contact force/torque[87, 88], quantitative efforts [89], or trajectory tracking errors[90]. Ankle reference trajectory is updated online based on the main gait cycle events and is adapted with respect to the self-selected speed of the wearer.

By using an adaptive controller, the robot assistance force can be adjusted according to the patient's physical movement ability. In assisting with the needed control, the system adaptively takes into account the patient's ability and provides assistance only when needed, rather than imposing an inflexible control strategy. A pilot study with eight healthy participants was conducted, and the experimental results demonstrate that, with the proposed control algorithm, the robotic AFO has the potential for ankle rehabilitation by providing adaptive assistance[91]. Controllers based on this principle are also referred to as 'patient-cooperative', 'human-centred', or 'progressive' controllers [92].In EMG-based adaptive control, the patient's muscle activity and recovery conditions can be reflected by EMG signals. To sense the patient's muscle ability, many recent studies[93, 94] have focused on the estimation and evaluation of muscle activity levels based on EMG signals. The isometric muscle model is traditionally used to establish the non-linear relationship between EMG and muscle forces[95].

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6. ARTIFICIAL INTELLIGENCE-BASEDORTHOSIS AND PROSTHESIS

Powered orthotic or prosthetic devices opened new challenges to detect gait modes (level ground, ascent, and descent during walking on stairs or ramps). A novel algorithm based on an artificial neural network (ANN) was proposed that continuously analysed the input signals for automatically detecting the gait mode using an inertial measurement unit(IMU). This algorithm recognized new gait modes faster and with higher accuracy than a previous method [96]. Data on gait patterns was collected with the help of analogue gyro sensors. From this data, a fuzzy rule base was created to complete the neuro-fuzzy system which was used to control the gait pattern. The control design for an active AFO to provide the required plantar flexion and dorsiflexion movement is developed using fuzzy logic. Depending on the input rule invoked by the fuzzy controller, the control signal is generated for controlling the position of the servo motor. An embedded control system using fuzzy logic for an active AFO can be designed by getting real-time gait data[97]. Proxy-based sliding mode control (PSMC) offers great performances in both position tracking and safety guarantee. Adaptation to the basic PSMC can be done with an online adaptation of the proportional, integral, and derivative parameters[98].

The angular velocity and angle of feet served as inputs for the controller and the output was actuation[99]. The ankle angular velocity is the better choice for use as a control reference parameter, which can be estimated based on walking speed and BMI as revealed by the results[100].A wireless signal acquisition system (WAS) was designed specifically, forming a platform to demonstrate and record individual sEMG and acceleration data simultaneously[101] (Fig.11). AI(artificial intelligence)-assisted marker-less motion capture is also suitable for analysing gait disturbance in an ankle-foot orthosis. Using the open Pose software prevented any influence on the body part recognition accuracy [102]. Syamantak etal. developed bionic leg orthosis (SBLO) that detects the wearer's walking motion and intelligently bends the brace at the proper time during the walking cycle[103].

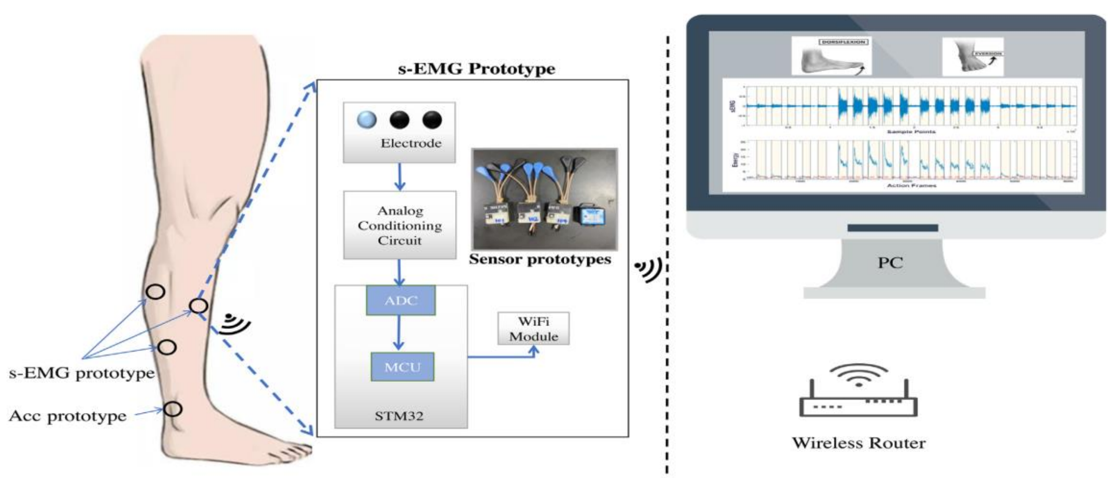

Fig 11: The block diagram and prototypes of proposed Wireless data acquisition platform for sEMG [101]

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A bio mechatronics system has four units: biosensors, mechanical sensors, controllers, and actuators[104]. A technique that allows the integration of a smart sensor system into a thermoplastic material (polypropylene, PP) is vacuum lamination. This laminated stack can then be thermoformed from a flat sheet into the desired 3D shape[105]. Sensor number optimization is conducted using neural network methods by a non-linear autoregressive exogenous (NARX) neural network developed for predicting the damping stiffness based on the ankle position [106].

The areas of use of bio mechatronics are orthotics, prosthesis, exoskeleton and rehabilitation robots, and neuroprosthesis. A bio-inspired soft wearable robotic device was designed and prototyped using soft actuators and sensors. The prototype showed the capability of assisting sagittal ankle motions with actuation, providing a range of 27°for the test subject. The prototype also demonstrated the capability for feedback control of the ankle joint angle using an identified LTI model of the system [57](Fig.12).A soft robotic AFO was developed containing a wearable gait-sensing module for measuring the leg trajectory and foot pressure in real-time for feedback control**.** A bi-directional tendon-driven actuator was used for assisting both dorsiflexion and plantarflexion with the help of a gait assistance algorithm (Fig13). The sock-like AFO is composed of soft actuators made from fabric-based, thermally bonded nylon and designed to be worn over the users' shoes. The dorsiflexion actuator can achieve a linear tensile force of 197*N*at 200kPa. The variable stiffness actuator generates up to 1.2*Nm* of torque at the same pressure[107].

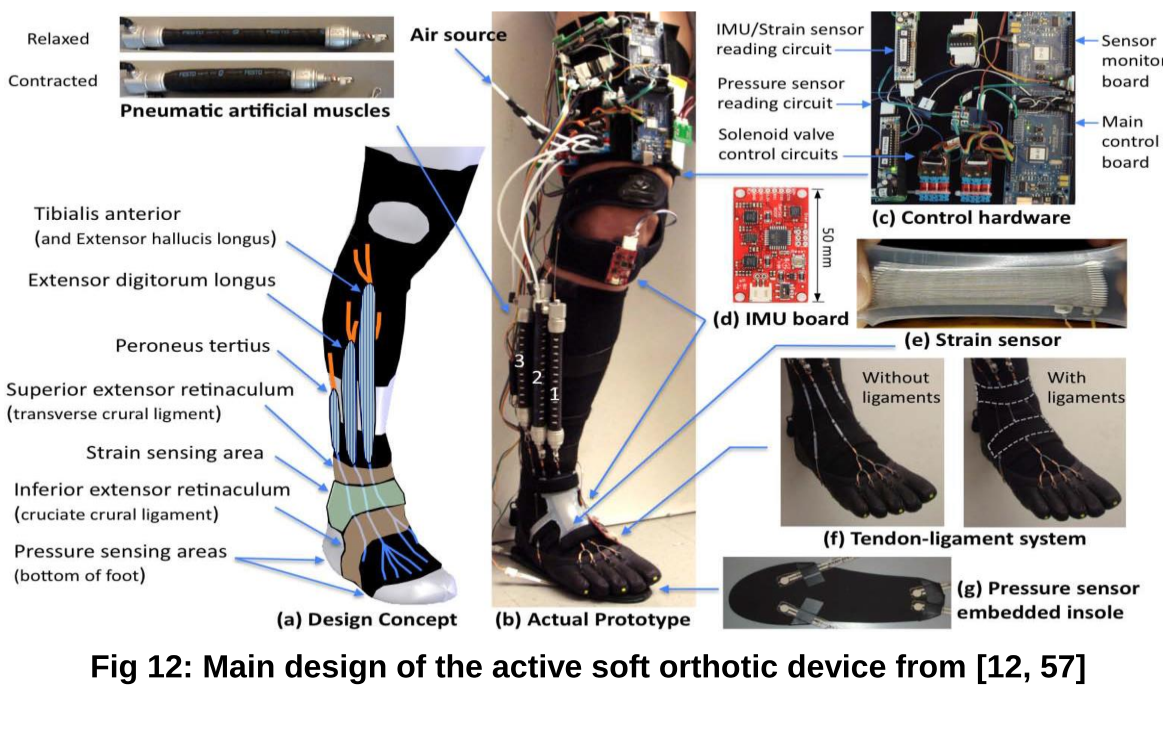

Fig 12: Main design of the active soft orthotic device from [12, 57]

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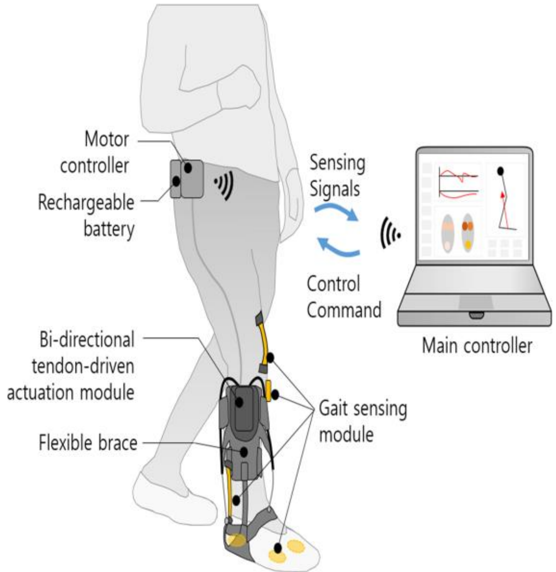

Fig 13: Illustration of a prototype of the soft wearable robotic ankle-foot orthosis. This figure has been reproduced with permission fromKwon, Park [108]

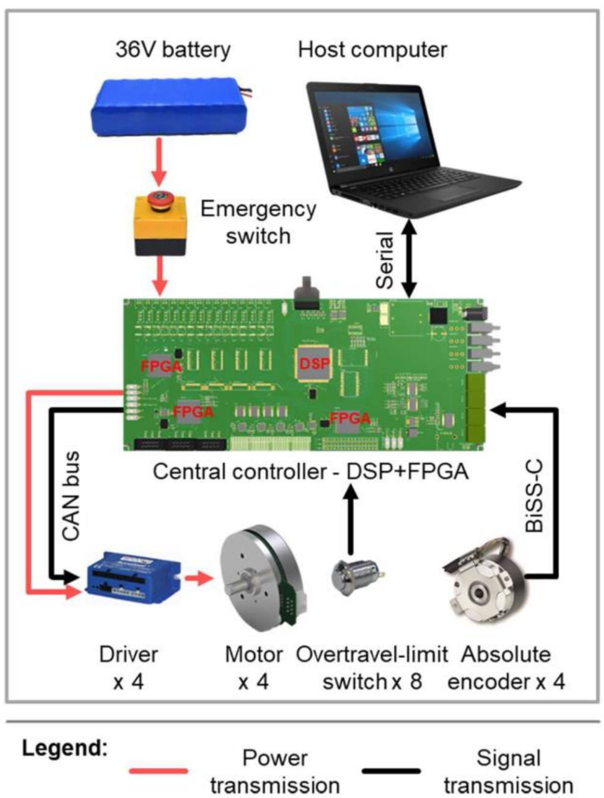

Fig 14: Electronic design of Exoskeleton[109]

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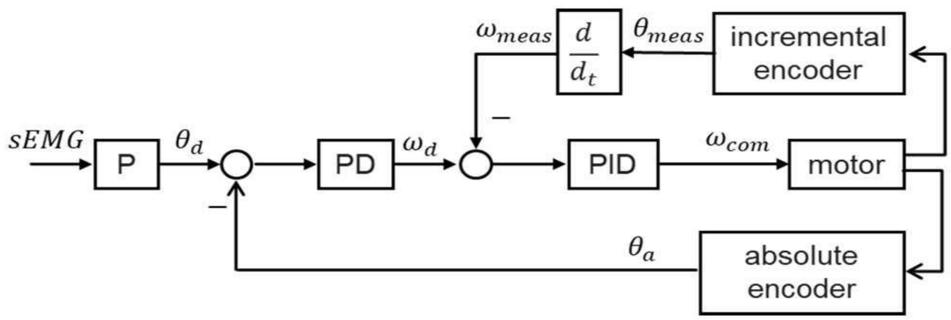

Fig 15: Block diagram of the control algorithm. The joint positions of the exoskeleton are proportionally controlled by the filtered sEMG signal of the subject[109]

At present, development in prosthetics has gone higher than that in the orthotics field. Prosthetic limbs that would be fully controlled by sensors implanted in the brain and even restore the sense of touch by sending electrical impulses from the limb back to the sensory cortex are in development[110]. Subjects could use their EMG signals to control the exoskeleton to assist them in playing the game for rehabilitation (Fig 14, 15). The advancement in EMG controls myoelectric prosthesis using an EMG pattern recognition-based control strategy [111]. The most advanced and developed neural machine interface technology is TMR or targeted muscle reinnervation [112]. It uses a multi-electrode EMG recording (up to 16 channels), classifier training, testing in offline mode, and real-time virtual and physical prosthesis control to regulate performance [113]. The diagram of EEG-based control and EMG pattern recognition based-control is utilized in upper extremity prosthesis, which is a mind- or thought-controlled prosthesis. It uses EEG signals and ANN [114]. Exoskeletons use BCI or EMG controllers to control orthotic devices [115] (Fig 16).

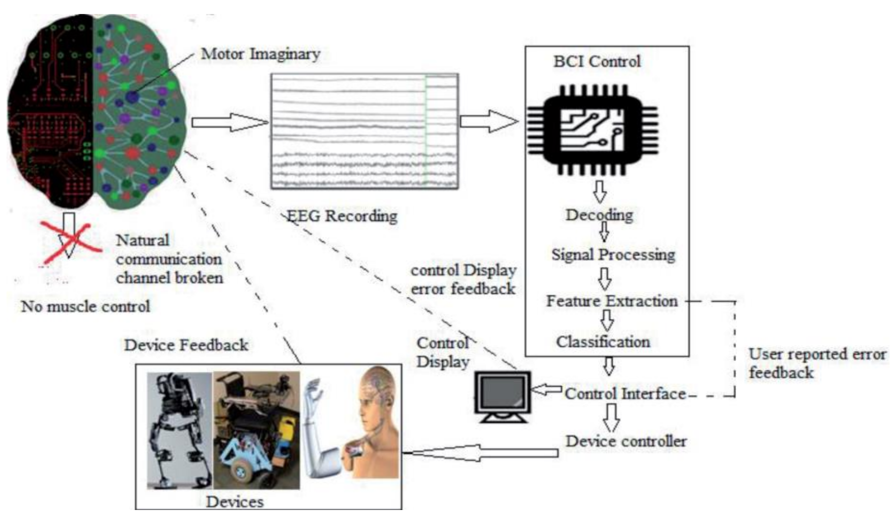

Fig 16: Brain computer Interface (BCI), controlling prosthetic and orthotics devices[115]

7. CONCLUSION

Current commercial AFO systems are generally limited to passive designs that control the undesirable motion of the foot but do not provide powered assistance during the propulsive phase of gait. Semi-active AFOs address some of the limitations of the passive ones by utilizing sensors and controllable braking mechanisms to manage the motion of the foot. However, they cannot provide supplemental assistive torque. Fully active designs provide net power to the ankle for both motion control and propulsive assistance. To date, active AFOs have not been commercialized and exist only in laboratory settings. The core challenges that must be met to realize such a device for both daily wear and rehabilitation are as follows: 1) a compact power source capable of day-scale operation; 2) compact and efficient actuators and means of power transmission capable of providing force comparable to healthy individuals; 3) control schemes that efficiently and effectively assist with any of a number of functional tasks that an individual may be expected to accomplish. In addition, the application of artificial intelligence (AI) and robotics to AFOs are in the initial stage of development and are yet to be commercialized widely.

The design of the AFO should be patient-specific and must depend on the muscle power availability around the ankle joint. It should be light, user-friendly, capable of meeting the biomechanical deficit, durable, and affordable. Developing intelligent AFO systems can help to improve the quality of life of individuals with foot drop.

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Conflict of Interest Statement

There is no Conflict of interest

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