Technologies for Powered Ankle-Foot Orthotic Systems: Possibilities and Challenges

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***Abstract*—Ankle-foot orthoses (AFOs) can be used to ameliorate the impact of impairments to the lower limb neuromuscular motor system that affect gait. Existing AFO technologies include passive devices with fixed and articulated joints, semiactive devices that modulate damping at the joint, and active devices that make use of a variety of technologies to produce power to move the foot. Emerging technologies provide a vision for fully powered, untethered AFOs. However, the stringent design requirements of light weight, small size, high efficiency, and low noise present significant engineering challenges before such devices will be realized. Once such devices appear, they will present new opportunities for clinical treatment of gait abnormalities.**

***Index Terms*—Active assist, ankle-foot orthosis (AFO), fluid power, gait.**

I. Motivation

FOR MOST, walking is a fundamental part of one's daily routine and is a key component in overall quality of life. The efficiency and effectiveness of gait depends on joint mobility and muscle activity that is selective in timing and intensity [1]. The forces and motion generated during gait are related to three main functional tasks: weight acceptance, single limb support, and limb advancement. Weight acceptance and single limb support occur during stance when the foot is in contact with the ground, whereas limb advancement takes place during swing when the foot is off the ground. The ability to walk is impaired by numerous neurological and muscular pathologies or because of injuries. These include trauma, incomplete spinal cord injuries, stroke, multiple sclerosis, muscular dystrophies, and cerebral palsy [1].

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Biomechanical deficits of the lower extremities and their related pathologies affect joint mobility and muscle activity. This paper focuses on the treatment of lower limb deficiencies with ankle-foot orthoses (AFOs). More specifically, the purpose of this paper is to present the background and rationale for developing new technologies for use in fully powered, untethered AFOs.

AFOs can be divided into three groups: passive, semiactive, and active. Passive devices contain no control or electronics, but can have mechanical elements such as springs or dampers to control the motion of the ankle joint during gait. Semiactive devices use computer control to vary the compliance or damping of the joint in real time. Fully active devices have an onboard or tethered source of power, one or more actuators to move the joint, and sensors and a computer to control the application of torque during gait. 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 the functional benefit they are capable of providing. These limitations could be addressed with an active AFO, but despite recent advances in computing, sensing, and other enabling technologies, there are currently no practical portable powered AFO systems in existence.

The development of light, compact, efficient, powered, untethered AFO systems has the potential to yield significant advancements in orthotic control mechanisms and new clinical treatment strategies for rehabilitation and daily assistance. A recent review describes work in lower extremity exoskeletons and active orthoses [2]. The purpose of this paper is to provide a comprehensive review of the state of the art specifically in ankle-foot orthotic technology and to describe the significant technical challenges that remain for AFOs. This paper overviews the biomechanics of normal and pathological gait, reviews existing passive and active AFO devices, and discusses the key enabling technologies required to meet this challenging human scale application.

II. Normal and Pathological Gait

Limb motion during steady-state constant speed locomotion involves intersegment and interlimb interactions during normal and abnormal walking [3]. Each limb segment and joint undergoes a cyclic pattern of flexion, extension, rotation, abduction, and adduction during a stride. An acute injury or pathology that affects a lower limb segment disrupts the cyclic gait pattern and can result in asymmetric deviations during gait [1]. An abnormal gait cycle affects the normal energy conserving characteristics of walking, resulting in increased energy expenditure [4].

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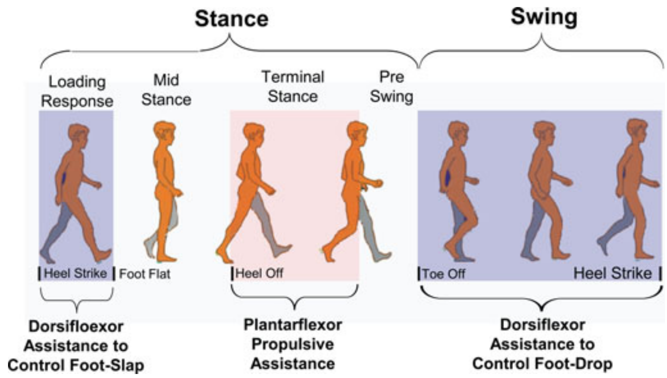

Fig. 1. A gait cycle is defined from heel strike to heel strike and further divided into multiple phases defined by functional task [1]. Ankle foot orthoses assist gait by controlling motion during stance and swing (dorsiflexor assistance) or providing assistive torque during stance (plantarflexor propulsive torque).

During 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. The gait cycle is defined from the initial contact of the heel to the following heel contact (see Fig. 1). At the initiation of the gait cycle, impact forces are dissipated as energy is absorbed by the soft tissues at the heel as the foot comes into contact with the ground [5]. Additionally, the muscles and tendons of the ankle joint complex dissipate energy (i.e., brake) by decelerating the foot before full contact with the ground at foot flat. The ankle joint complex also helps to maintain stability during stance phase. This is particularly important during single limb support in the stance phase when only one limb is supporting the body. In addition to providing stability, energy is stored through tensile loading of the tendons and muscles that traverse the ankle joint complex when the shank pivots. The ankle plantarflexor 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 [6]. Lower limb joint powers for a healthy walker are shown in Fig. 2. The significantly larger peak power at the ankle is shown just before 60% of the gait cycle. As a result, pathology or injury that affects the ankle joint has the potential to significantly impact quality of life by impairing some or all functional aspects of gait.

Weakness in the dorsiflexor and plantarflexor muscle groups is a key cause of impaired gait. Understanding muscle weakness and its effect on gait is essential to the proper design of orthotic devices that compensate for these deficiencies [7]. The dorsiflexors are situated anterior to the ankle joint and include the tibialis anterior, extensor digitorum longus, and extensor hallucius longus [1]. Pathologies that afflict the function of the ankle dorsiflexors affect gait in both swing and initial stance phases. Swing is affected by insufficient toe clearance due to weak or absent dorsiflexor muscles and results in a steppage-type gait pattern that is commonly called foot drop (see Fig. 1). Steppage gait is a compensatory walking pattern characterized by increased knee and hip flexion during the swing phase so that the toe clears the ground during walking. Weak or absent dorsiflexors may also prevent controlled deceleration of the foot shortly after initial contact (see Fig. 1) that often presents as an audible foot slap.

Weakness in the ankle plantarflexor muscle group primarily affects the stance phase of gait. The plantarflexors situated posterior to the ankle joint are comprised of the gastrocnemius, soleus, and the peroneal and posterior tibial muscles [8]. From heel strike to middle stance, the ankle plantarflexors eccentrically contract to stabilize the knee and ankle and restrict forward rotation of the tibia [7]. At the end of stance, the plantarflexors concentrically contract and generate torque that accelerates the leg into swing and contributes to forward progression (see Fig. 1) [9]. Weak ankle plantarflexors affect stability, particularly during single limb support. Individuals with impaired ankle plantarflexors compensate by reducing walking speed and shortening contralateral step length. Reduced walking speed results in a corresponding reduction in torque needed for forward progression. The shortened contralateral step is thought to increase stability by limiting anterior movement of the center of pressure with respect to the ankle [7]. Impaired individuals may maintain a fast walking pace by using their hip flexors to compensate for weak plantarflexor muscles [10].

Muscle weakness can be neurological or muscular in origin and can be due to a multitude of pathologies [1], [7]. Common conditions that may result in muscle weakness include trauma, incomplete spinal cord injury, brain injury, stroke, multiple sclerosis, muscular dystrophy, and cerebral palsy. The ideal orthoses 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 plantarflexion with mild resistance while maintaining ankle and knee stability up to midstance, generate an assistive torque at terminal stance, and block plantarflexion during swing to prevent foot drop. All of these actions need to be accomplished using an orthosis that is compact and light weight to minimize the energetic impact on the wearer. This is particularly true of an orthosis located at the ankle because accelerations at the foot are twice the individual's average walking speed [11]. For example, a 2-kg load placed on each foot of a healthy adult can result in a 30% increase in the rate of oxygen uptake, whereas a 20-kg load placed on the trunk does not result in a measurable increase [4].

Currently, there are no AFOs capable of assisting both dorsiflexor and plantarflexor function. Since maintaining toe clearance during swing has such a dramatic improvement on function, passive orthoses with their relatively simple mechanical resolution and economic viability continue to be the preferred AFO design. The subject-specific nature of these designs in terms of both fit and assistance has made rigorous scientific evaluation of and comparison between novel AFO designs challenging. Instead, the effectiveness of a new AFO design tends to be demonstrated qualitatively using a small number of healthy and/or impaired subjects.

III. Existent Passive and Active AFO Designs

A. Passive AFO Designs

Because compactness is critically important, AFOs used on a daily basis are generally passive. Passive AFOs can be divided into articulated or nonarticulated devices. They can be further

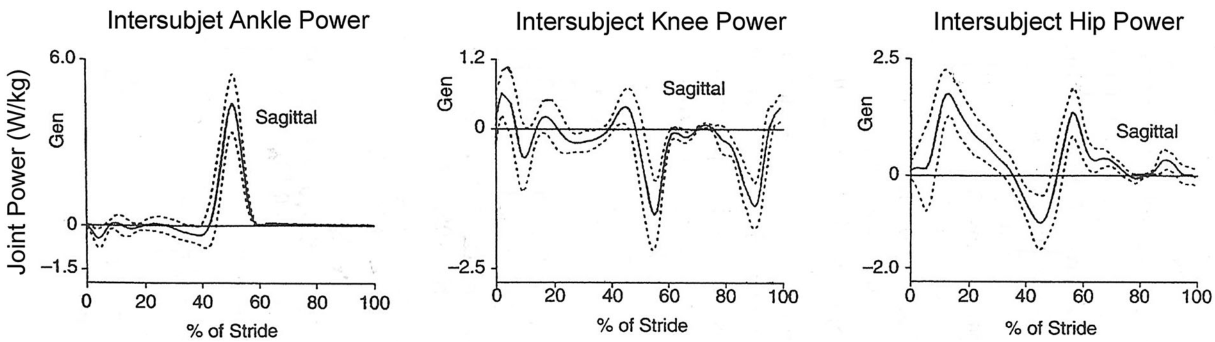

Fig. 2. Healthy sagittal-plane power generation (positive) and absorption (negative) at the ankle, knee, and hip joints normalized to body weight and percentage of the walking cycle. Solid lines are normalized intersubject averages and dotted lines show one standard deviation about the average [51].

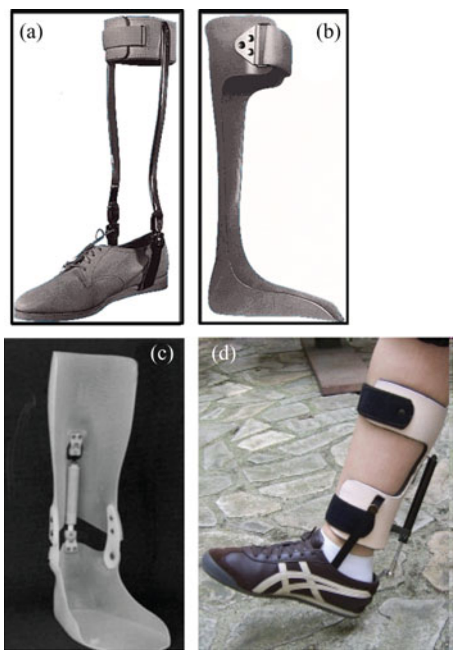

Fig. 3. Passive AFO designs. (a) Metal and leather AFO. (b) Posterior leaf spring AFO. (c) Hybrid AFO with a linear spring and thermoplastic AFO. (d) Pneumatically assisted PneumaFlex AFO. From [14], [18], and [19].

subdivided by material into metal and leather, thermoplastic, composite, and hybrid systems [12], [13]. These AFOs provide assistance by preventing unwanted foot motion with direct physical resistance.

In traditional metal and leather systems, articulated hinge joints with mechanical stops are used to limit motion and optional springs resist or assist movement. Fig. 3(a) shows a metal and leather AFO made by Becker Orthopedic [14]. This AFO uses metal uprights along the length of the shank for stability and an articulating joint that can be configured with various pins and/or springs to control motion in a prescribed manner by blocking, resisting, or assisting movement of the ankle joint. Metal systems can accommodate fluctuations in limb volume due to swelling from inflammation or circulatory problems that would make a fixed dimension total contact fitting plastic or carbon fiber system unfeasible. Metal–leather AFOs are often less expensive than their carbon composite counterparts, but their additional size and bulk make these systems less cosmetically attractive.

A posterior leaf spring AFO [see Fig. 3(b)] is a lighter weight alternative to metal and leather systems and is an example of a nonarticulated AFO [14]. These single-piece AFOs are constructed from both thermoformable plastics (e.g., polypropylene and copolymer) and thermosetting plastics (e.g., carbon graphite composites). The posterior leaf spring is a common strut-type AFO design where the material properties and geometry of the device determine motion control characteristics. The thermoplastic shells of these AFOs encompass the posterior and plantar aspect of the leg and foot, respectively. A one-piece thermoplastic AFO can integrate energy storage and assist elements into its structure. For example, the distal half of the shank acts as a flexible strut or leaf spring providing resistive and assistive forcing as it is deformed during loading. Motion control properties of single-piece thermoplastic AFOs are controlled by the geometry of the material, and the proximity of the medial and lateral trimlines with respect to the anatomical ankle rotation axis. For example, medial and lateral trimlines posterior to the ankle axis will be less stiff compared to trimlines that are anterior [15], [16].

While both metal–leather and posterior leaf spring AFOs are commonly used to treat gait deficiencies, they possess motion control features (e.g., mechanical stops or material deformation) that may inhibit desirable motion. For example, the metal and leather system shown in Fig. 3(a) prevents the foot from dropping during swing with mechanical stops built into the device. However, these stops also prevent plantarflexion that would normally occur during the first half of stance, thereby altering gait. Commercial hinge joints such as the Tamarack Flexure Joint have been used with thermoplastic AFOs to control motion in the sagittal plane, but the assistive force by the joint in one direction (i.e., dorsiflexion) may result in a resistive force in the opposite direction (i.e., plantarflexion) which may improve gait with regard to one direction while inhibiting it in the other [17]. Clinical orthotic intervention strives to provide biomechanical control necessary to improve a functional deficit without perturbing other normal movements and functions.

As a result, several novel hybrid AFOs have been designed to more effectively control the motion of the ankle joint. A hybrid device combines light-weight thermoplastic or carbon

TABLE I
A COMPARISON OF WEIGHT, TORQUE, ADVANTAGES, DISADVANTAGES, PERFORMANCE METRICS, AND EFFECTIVENESS OF THE NOVEL PASSIVE AFOS DESIGNS DESCRIBED IN SECTION III

		Weight	Resistive or Assistive	Active Element	Maximum Applied Force	Advantages	Disadvantages	Performance Metrics	Experimental Evaluation	Results	Control
Passive Hybrid AFOs	DACS AFO (18)	0.3 kg	Resistive	Mechanical Spring	17 Nm (per 10 deg of rotation)	Compact, light weight, untethered. interchangeable springs for patient specific assistance	Constant resistive force impedes motion	Gait speed and qualitative visual inspection of gait	Five Hemiplegic subjects walked with DACS, posterior leaf spring, and metal-leather AFO	DACS AFO users had faster and smoother gait	NA
	PneumaFlex AFO (19)	0.13 kg	Resistive	Air Spring	?	Compact, light weight, untethered. Adjustability of pneumatic springs allows for patient specific assistance	Constant resistive force impedes motion	?	?	?	NA
	University of Illinois AFO (20)	1 kg	Resistive	Locking CAM	?	Variable motion control, untethered and the energy to actuate the locking mechanism harvested during gait	Bulky AFO structure, complicated locking system	Joint angle kinematics, pneumatic line pressure	Single healthy subject	Joint angle data demonstrated proper foot motion and pneumatic pressure data showed correct locking sequence during gait	NA
	Osaka University Hybrid AFO (21)	?	Resistive	Friction Break	4 Nm	Variable motion control, untethered, compact resistive element, energy harvested during gait	The resistance provided by the break in not easily adjustable	Kinematic data from the braced leg and insole pressure sensor data	Single healthy subject	Kinematic data was used to verify the correct restriction of joint range of motion during swing	NA
	Kanagawa Rehabilitation Center AFO (22)	0.4 kg	Resistive	Oil damper	5-14 Nm (at 10 degrees of plantar flexion)	Variable motion control, untethered, light weight and the resistive force is easily adjustable	Resistive force is the same during initial stance and swing	Time, distance and kinematic parameters	Two hemiplegic patients walked with oil damper and conventional AFO	No significant functional difference between the AFOs. Oil damper's largest advantage was ease of adjustability	NA
	Okayama University AFO (23)	0.86 kg	Assistive	Pneumatic Actuator	2 Nm	Variable motion control, untethered, energy to actuate the active element is harvested during gait	Has a bulky AFO structure and only generates small assistive torques	EMG	Single healthy individual	Decrease in EMG signal during trials indicates supplemental assistance	NA

composite shells with articulated joints and passive motion control elements. The hybrid AFOs described in this section are compared in Table I. Researchers at the International University of Health and Welfare, Japan, have developed the dorsiflexion assist controlled by spring (DACS) AFO for the prevention of foot drop in hemiplegic patients [18]. The 300-g DACS AFO has two thermoformable plastic pieces connected with joints on the medial and lateral sides of the ankle. An embedded spring on the dorsal side of the shank provides a peak dorsiflexor toque of 17 N·m (per 10° of rotation) [see Fig. 3(c)]. At heel strike, the spring resists compression and prevents an uncontrolled deceleration of the foot (i.e., foot slap). During stance, the structure of the AFO stabilizes the joint. During swing, the spring resists the foot from dropping below its neutral position, perpendicular to the shank, resisting foot drop and providing toe clearance. To evaluate the design, joint angle data and overall walking speed from five hemiplegic subjects were recorded during trials with the DACS AFO, a plastic posterior leaf spring AFO, and a metal–leather AFO. Subjects walking with the DACS AFO had faster walking speeds and were observed to have smoother gait.

Pneumatic springs have been used in place of mechanical springs to more easily modulate the stiffness of the passive element for patient-specific tuning. The PneumaFlex is an example of this type of design [see Fig. 3(d)] [19]. The central component of the PneumaFlex AFO is a pneumatic spring mounted posterior to the carbon-fiber shank and footplate. The pressure in the spring's cylinder is adjustable and is selected on a patient-specific basis to support the weight of the foot. The resulting light-weight design (130 g) is worn in subject's shoe and is used to control the motion of the foot to prevent foot slap during stance and foot drop during swing. We could not find any published works documenting the performance of this design or comparing the benefits of the PneumaFlex AFO to other designs.

Other innovative hybrid AFO designs have focused on harvesting energy from gait in a different manner from the spring concepts described previously. In these designs, the actuators that control motion or provide assistive torque are separated from the elements used to harvest energy. Researchers at the University of Illinois at Urbana-Champaign, Urbana-Champaign, U.S., have designed an AFO to harvest energy during gait for use in motion control of the ankle [20]. The objective of this AFO is to achieve toe clearance during swing and free ankle motion during stance. The AFO is constructed from a two-part (tibia and foot) carbon composite structure with an articulating ankle joint and weighs 1 kg [see Fig. 4(a)]. Ankle motion control is accomplished with a locking mechanism actuated via a pneumatic circuit connected to a pump embedded in a foam sole under the forefoot. The pump is compressed by subject's body weight during stance. The compressed air is then used to engage the lock during swing to prevent foot drop. At heel strike, a touch valve opens to release the pressurized air and unlock the joint. Joint angle and pressure data were used to demonstrate the effectiveness of the design with a single healthy subject. The joint angle data were used to show that the AFO restricted ankle range of motion during swing, and the pressure data indicated that proper timing of the locking mechanism was achieved.

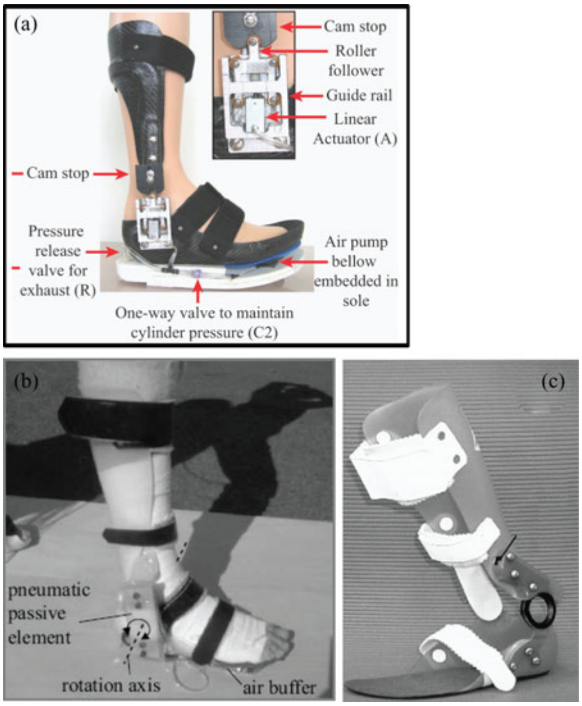

Fig. 4. Energy harvesting passive AFO designs. (a) The Illinois power-harvesting AFO. (b) The Osaka University AFO. (c) The Kanagawa Rehabilitation Center AFO. From [20]–[22].

Researchers at Osaka University, Osaka, Japan, have constructed an AFO that utilizes a passive pneumatic element actuated by subject's bodyweight for motion control of the ankle joint [21]. The motion control element has two subassemblies: an air buffer that functions like a pump and a passive pneumatic element that contains thin laminated sheets enclosed in an airtight plastic chamber located at the axis of rotation of the ankle joint. During stance, the buffer, located under the sole of the AFO, is compressed by subject's weight, forcing air into the passive element. This decreases the vacuum force and pushes the thin sheets of the element apart. The space between layers results in a decrease in rotational friction and allows free motion of the joint. During swing, air returns to the buffer, pressing the laminated sheets together to create a maximum torsional stiffness of around 4 N·m [see Fig. 4(b)]. The motion control provided by this AFO resists both foot drop during swing and foot slap following heel strike. The AFO design was evaluated using both kinematic data from the leg with the AFO and insole pressure data from a single healthy subject. The researchers used qualitative comparisons of the insole pressure and the kinematic marker data between trials collected with and without the AFO to conclude that the design was able to meet its functional objectives.

Oil dampers have been used to control ankle motion by absorbing energy from the system. Researchers from Kanagawa Rehabilitation Center, Kanagawa, Japan, have built an AFO that uses a fixed viscosity damper to control the motion of the foot by creating viscous forces to resist only plantarflexor motion [22]. The 0.4-kg AFO uses this resistive force to prevent toe drop during swing, whereas the lack of dorsiflexor resistance allows free motion during stance [see Fig. 4(c)]. The damper provides a variable resistive force (5–14 N·m) and can be adjusted for subject-specific need by changing the physical parameters of the damper. The gait of two patients with hemipeliga wearing this AFO was compared to subject's gait data collected while wearing a conventional AFO with plantarfexion stops. The resistance provided by the oil damper was adjusted by a physical therapist. Suitable resistance was determined qualitatively based on functional need and the overall patient comfort. Time and distance measures (walking velocity, cadence, step length, stride length, and cycle timing) along with kinematic parameters (peak sagittal plane joint angles) were used to evaluate and compare the performance of the AFO. The researchers conclude that there were no functionally significant differences between the two AFOs, and that ease of adjustability is the oil damper's biggest advantage over the traditional AFO.

Contrasting with the AFO designs that use harvested energy solely for motion control, researchers at Okayama University, Okayama, Japan, built a pneumatic AFO that uses harvested energy from the wearer to help power gait [23]. At heel strike, air is compressed using a bellow pump located under the heel and stored in air balloons located on the lateral side of the ankle. The compressed air is then used to actuate a custom wire type cylinder to produce a dorsiflexor assistive torque during swing. The device weighs 860 g and is capable of producing a peak torque of 2 N·m. The design's effectiveness was examined by qualitatively comparing electromyogram (EMG) signals of a healthy subject walking both with and without the AFO. A decrease in EMG signal was found during the AFO trials indicating that the AFO was successfully supplementing the work done by the muscles.

The passive AFOs presented in this section provide assistance by preventing unwanted foot motion with direct physical resistance. The hybrid AFO designs, with the exception of the Okayama University AFO, have been developed to provide motion control without unnecessary restrictions to walking motion (e.g., unrestricted range of motion during stance) that are created by conventional AFOs (see Table I). The motion control elements used in these designs also offer greater subject-specific tuning options than comparable leaf spring or metal and leather AFOs. The stiffness of these passive AFOs range from a few Newton meter up to ∼20 N·m of resistive torque over a 30° range of motion [24], [25]. However, in current clinical practice, there are no standard methods for the identification of the motion control properties that are most appropriate to improve an individual's gait. It is up to the orthotist to use his/her expertise to select and modify the stiffness of the AFO to effectively assist gait [25].

B. Active and Semiactive AFO Designs

Despite the successful integration of passive elements in hybrid AFOs for both motion control and torque assistance, there are limitations to what can be accomplished with a purely passive device. Passive elements improve gait deficiencies by controlling motion and generally do not provide direct assistance during the propulsive phase of gait. Furthermore, the control

TABLE II
A COMPARISON OF WEIGHT, TORQUE, ADVANTAGES, DISADVANTAGES, PERFORMANCE METRICS, AND EFFECTIVENESS OF THE NOVEL ACTIVE AND SEMIACTIVE AFOs DESIGNS DESCRIBED IN SECTION III

	Type	Weight	Resistive or Assistive	Active Element	Max Applied Force	Advantages	Disadvantages	Performance Metrics	Experimental Evaluation	Results	Controller
Active and Semi-Active AFOs	Osaka University AFO (26)	1.6 kg	Resistive	Magneto Rheological (MR) Damper	24 Nm	Variable motion control, large peak breaking torques	Tethered, only resists motion	Qualitative comparison of AFO sensor data	Single subject with right ankle flaccid paralysis	Qualitatively observed improvement gait with the controlled braking AFO	Finite State Control
	Halstead University AFO (27)	?	Resistive	Magneto Rheological (MR) Damper	?	Variable motion control, untethered	Only resists motion	Qualitative comparison of AFO sensor data	Three healthy subjects	Foot range of motion was properly restricted during gait and stair climbing. Control algorithm successful switched between functional tasks.	Finite State Control
	MIT Active AFO (28-29)	2.6 kg	Assistive	Series Elastic Actuator (SEA)	?	Provides both dorsi- and plantarflexor assistance	Tethered	Kinematic and kinetic data	Two foot-drop subjects and three matched healthy subjects	Foot-slaps per five gait cycles were reduced and foot-drop during swing was prevented	Finite State Control
	Arizona State Robotic Tendon AFO (31-33)	1.75 kg	Assistive	Robotic Tendon (modified SEA)	~ 60 Nm	Provides both dorsi- and plantarflexor assistance	Tethered	Kinematic and kinetic data	Single healthy individual	Control states were triggered correctly during gait. AFO generated power comparable to a healthy individual during level walking	Finite State Control
	BIONic WalkAide (34)	?	Assistive	Function Electric Stimulation (FES)	?	Compact, light weight	Limited patient population	Kinematic data and physiological cost index (PCI)	Single subject with nerve damage	Implantable micro stimulators produced balanced ankle flexion with a low PCI score	Finite State Control
	NESS L300 (35)	?	Assistive	Function Electric Stimulation (FES)	?	Compact, light weight	Limited patient population	Gait speed, heart rate, and temporal gait parameters (stance and swing times)	24 subjects with chronic hemiparsis	Improved walking speed, decreased asymmetry and decreased temporal variability when using the AFO	Finite State Control

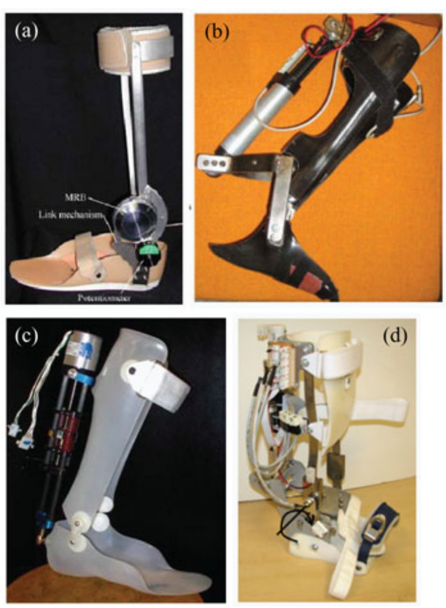

Fig. 5. Semiactive AFO designs. (a) The Oksaka magneto rheology (MR) damper AFO and torque amplifying mechanical linkage. (b) The Halmstad magneto rheology (MR) damper AFO. (c) The MIT active AFO powered by a series elastic actuator. (d) The Arizona State AFO powered by a modified series elastic actuator. From [26]–[28] and [33].

of passive AFO elements relies on the activation of springs, valves, or switches in an open-loop manner as the individual walks. This type of control has limited robustness and does not adapt to changing walking conditions or a changing functional task. Semiactive and fully active AFOs that take advantage of external power supplies and are equipped with powered actuators address both of these limitations. They can be divided into two groups: AFOs that have the potential to be realized as a daily-wear device, and AFOs intended strictly for patient diagnostics and rehabilitation.

1) Daily-Wear Devices: In order for an active AFO to be used as a daily-wear device for improved function, the use of a tether for power or computing is not practical. Current development of external powered orthoses has primarily utilized tethered systems due to the engineering design challenges. Tethered sources of power and computing allow individual components (e.g., actuators) to be designed and tested as other components of the system are developed (e.g., untethered power sources). Active systems rely on sensor feedback to determine both task (e.g., walking) and the functional assistance required by the user. For example, an individual with weak dorsiflexors might require assistance to control foot motion during swing for the prevention of foot drop. In this case, AFO sensors would be used to determine boundaries for swing phase (e.g., the identification of toe off and heel strike) and potentially, in feedback control of the actuator, to track a desirable ankle joint angle to maintain toe clearance (e.g., a 90° angle between the foot and shank). The AFOs described in this section are untethered or the researchers have stated their intentions to continue the development of the device into a future untethered version (see Table II).

The study at both Osaka University and Halmstad University, Halmstad, Sweden, utilizes computer-controlled magnetorheological (MR)-type dampers to modulated viscous damping for semiactive motion control [26], [27]. MR dampers use fluids with viscosities that are modulated using magnetic fields. The Osaka AFO uses an MR damper to create a resistive plantarflexor torque during swing and initial stance. The maximum resistive torque is 11.8 N·m and can be increased to 24 N·m using a mechanical linkage for torque amplification [see Fig. 5(a)]. The amount of resistive force applied by the brake is based on the functional state of the patient during the cycle. The gait cycle is broken up into four states using information from three sensors: a potentiometer to measure the ankle joint angle, a six-axis force–torque sensor mounted in the foot plate of the AFO, and a moment sensor located in the lateral AFO strut. The 1.6-kg AFO was tethered to an external power supply and computer. An experimental evaluation of the AFO was conducted with a single subject with right ankle flaccid paralysis. Data from the AFO sensors were collected while the subject walked with and without controlled braking. The researchers used a qualitative comparison between the conditions to conclude that subject's gait was improved by the controlled braking.

The Halmstad AFO incorporates onboard position sensing, power, and electronics to create an untethered device [27]. An active control algorithm along with an MR damper allows the AFO to assist stair climbing, inclined walking, and level walking [see Fig. 5(b)]. The AFO controller is a finite-state machine with four states: damped, free, locked, and free down (limited damping to allow motion 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. Maximum ankle values during the cycle are used to switch between functional tasks (walking or stair climbing). During walking, the AFO provides moderate damping to control foot slap at heel strike, free range of motion during stance, and large damping to resist foot drop during swing. During stair decent, the AFO operates only in the free down state allowing the toes to point slightly downward as the foot travels from one step to the next. Finally, during stair ascent, the locked state is used when the foot is raised from one step to the next and the free state is used during contact with the step. Three healthy subjects were used to evaluate the performance of the AFO. Data from the AFO sensors were used to demonstrate that foot motion was properly restricted during gait, and that the control algorithm used could successfully switch between functional tasks. Although MR dampers are able to control foot motion, they are not capable of generating torque at the ankle joint for use in push-off.

To address the issue, researchers have utilized a series elastic actuator (SEA) to provide both motion control of the foot and apply plantarflexor torque during gait [28], [29]. The Biomechatronics Group at Massachusetts Institute of Technology, Cambridge, U.S., developed an active SEA AFO to assist foot-drop gait. The AFO weighs 2.6 kg and is tethered to an off-board power supply. The SEA consists of a dc motor powered ball screw mechanism in series with a helical spring [see Fig. 5(c)]. The computer-controlled motor adjusts the rotary compliance of the AFO by driving a lead screw to vary the height of the spring. The use of an elastic element with a motor offers advantages over a direct drive system including greater shock tolerance, lower reflected inertia, more accurate and stable force control, and the capacity for energy storage [30]. While the SEA actuator offers the flexibility to assist patients with both plantarflexor and dorsiflexor weakness, in their paper, the AFO was configured to assist individuals with weak dorsiflexors (e.g., foot-drop gait) but functional plantarflexors. A finite-state controller was used to divide gait into three states each with a separate functional objective. The first control state lasted from heel strike to midstance and prevented foot slap by using the SEA to increase the impedance at the ankle. The second control state lasted from midstance until toe-off and minimized the impedance of the AFO to allow full plantarflexor movement. The third control state occurred during swing and maintained toe clearance by lifting the foot. Ground reaction force and angular position data from onboard sensors were used to transition between states during walking, and an adaptive control algorithm was used to accommodate different walking speeds within the control states. The AFO was evaluated using kinematic and kinetic data from two foot drop and three matched normal subjects. The subjects walked with the AFO set to zero, constant or variable impedance at slow, self-selected and fast gait speeds. The performance of the AFO to assist foot-drop gait was evaluated by counting the number of foot slaps that occurred per five gait cycles and measuring the angular difference between the maximum plantarflexion angle and the maximum dorsiflexion angle during swing. Larger angular differences indicated free motion during plantarflexion and a reduction of foot drop during swing. These metrics were used to show that the SEA AFO with the variable impedance control algorithm outperformed both the zero and constant impedance controllers. This study is focusing on developing an untethered version of this device.

Researchers at Arizona State University, Phoenix, U.S., have also built an AFO around a highly compliant actuator called a robotic tendon [31]–[33]. Like the SEA, the robotic tendon uses a motor/screw/spring arrangement to offer greater compliance than a direct drive system [see Fig. 5(d)]. This AFO also uses the 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. Additionally, the internal compliance of the robotic tendon allows the user to deviate from a prescribed trajectory if the walking environment changes. The researchers suggest that this flexibility provides added safety for the user. The AFO built around the robotic tendon allows motion in the sagittal plane and utilizes an encoder, potentiometer, and one force sensor embedded at the heel for sensor feedback. The control algorithm described in [32] accommodates gait initiation and cessation and allows the device to accommodate different level walking speeds. This earlier design used a 0.95-kg tendon that required 77 W of power to produce a torque comparable to a healthy individual during level walking [32]. An updated AFO design uses a lighter tendon (0.5 kg) and weighs 1.75 kg [33]. This design used 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 are used to switch between states. The first five states occur during stance and alternate between stiffness and velocity control. The sixth and seventh states occur during swing when the foot is raised and then supported at a constant angle until the following heel strike. The functionality of the device was verified using kinematic and kinetic data from a single able-bodied subject. The results showed that the controller switches states correctly and that the actuator generates power comparable to a healthy individual during level walking. While this AFO is still tethered to an external power source and computer, the researchers suggest that the device could be powered for 8 h of continuous operation using a battery worn in a fanny pack.

One form of semiactive AFO that is untethered uses functional electrical stimulation (FES) to create ankle flexion. The BIONic WalkAide [34] and the NESS L300 [35] are commercially available FES devices that 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. The BIONic WalkAide 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). The work conducted by Weber *et al*. used a modified WalkAide stimulator to control implantable microstimulators (BIONS). This study compared the effectiveness of surface and implantable stimulation to correct the foot drop of a single subject with nerve damage resulting from a spinal cord injury. Kinematic data and heart rate were collected during treadmill walking. These data, along with a physiological cost index (PCI), were used for the comparison. The PCI was calculated by dividing the difference between quiet state and walking heart rate by ambulatory velocity. The BION stimulation produced a more balanced ankle dorsiflexion movement and had a slightly lower PCI than the surface stimulation.

The NESS L300 uses a force sensitive resistor placed under the foot to detect swing. A study of 24 patients with chronic hemiparesis was used to show that the NESS L300 enhanced gait and improved dynamic stability. Subjects were asked to walk for 6 min with and without the device at 0, 4, and 8 week assessments. Gait speed, heart rate, and temporal gait parameters (stance and swing times per gait cycle) were collected during the trials. The researchers found improved walking speed, decreased asymmetry, and decreased temporal variability when using the device. Both devices are compact because user's own muscles and skeleton provide the actuation and support that AFOs traditionally provide. Because FES directly stimulates an individual's nerves and muscles, there is the potential to increase the fatigue resistance and strengthen muscles [34]. FES devices, however, require careful positioning of the electrodes on a daily basis, and have a limited suitable patent population.

2) Active AFOs for Patient Diagnosis and Rehabilitation: Fully active AFOs are able to provide net power to the ankle, unlike passive and semiactive designs that can only dissipate or store and release available energy. To date, most active AFOs are tethered because technology capable of meeting the power requirements for full assistance cannot meet the size and weight requirements of a daily-wear device. Tethered devices are suitable for laboratory research and for clinic-based rehabilitation treatments that aid in recovery from the pathology or injury [36]. 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 in locomotion studies to perturb gait. Accordingly, the devices described in this section are not meant to be portable.

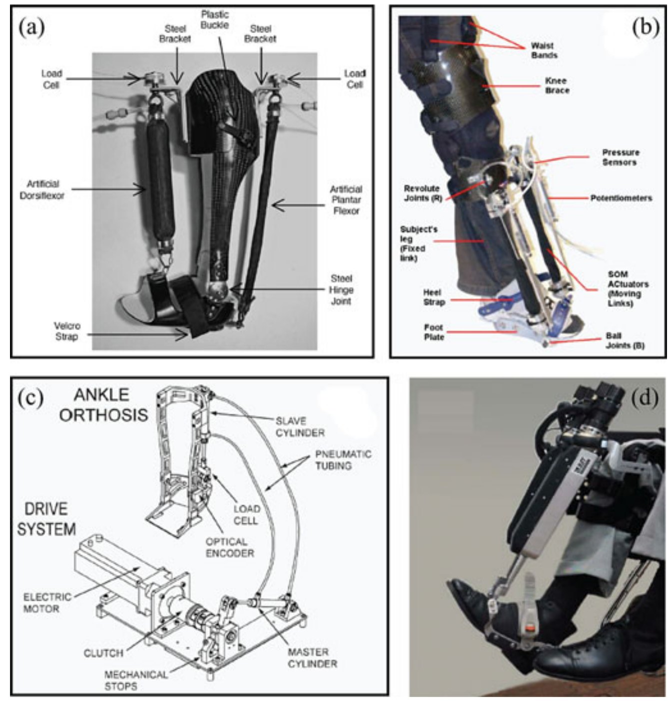

Fig. 6. Tethered active AFO designs. (a) The Michigan AFO powered by pneumatic muscles. (b) The Arizona State Robotic Gait Trainer with spring over muscle pneumatic actuators. (c) A mechanical drawing of the hydraulically powered CIRRIS AFO. (d) The MIT AnkleBot shown with the two front mounted dc-motor-powered-linear-actuators. From [39]–[42].

Researchers at the University of Michigan, Ann Arbor, U.S., Arizona State University, and the Center for Interdisciplinary Research in Rehabilitation and Social Interaction (CIRRIS), QC, Canada, have built fluid-powered AFOs intended for human locomotion study and gait rehabilitation [37]–[41]. The Michigan AFO uses FES to control McKibben-style uniaxial artificial pneumatic muscles in various arrangements to provide dorsiflexor and plantarflexor torque. The AFO has a total weight of 1.6 kg excluding the off-board computer and air compressor [see Fig. 6(a)]. The orthosis provides a peak plantarflexor torque of 70 N·m and a peak dorsiflexor torque of 38 N·m [37].

The rehabilitation AFO, called the robotic gait trainer, built at Arizona State University utilizes pneumatic spring over muscle (SOM) actuators to create bidirectional forcing [41]. The SOM actuators enclose a cylinder plunger containing a compression spring ($K = 1.40$ N/mm) within a McKibben-style pneumatic muscle. The AFO is arranged in a tripod configuration with subject's leg acting as a static link and two SOM actuation links [see Fig. 6(b)]. Both dorsi/plantarflexor and inversion/eversion are allowed by this design. The researchers envision this system being used for home repetitive task therapy and strength building rehabilitation.

The CIRRIS AFO is powered by a water-filled hydraulic master cylinder, which is driven by an electrical motor. The master cylinder is then connected to a slave cylinder mounted posterior to the shank [see Fig. 6(c)] [40]. The AFO produces several torque profiles: constant torque, position-dependent torque, and phase-dependent torque. Additionally, the AFO 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 N·m and a peak torque of 98 N·m to the master cylinder. The weight of the AFO worn by the subject was kept to 1.7 kg by locating the electric motor away from the device. Torque cancellation was performed to minimize the influence of the passive torque induced from the electrohydraulic drive system.

Directly measuring the physical properties of the ankle joint complex is also an important function performed by AFOs used in a rehabilitation setting. The AnkleBot, designed by researchers in 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 [42], [43]. The device is actuated by two dc-motor-powered linear actuators mounted to the front of the shank using a knee brace and footplate. The tripod arrangement of the components allows three degrees of freedom at the foot [see Fig. 6(d)]. A dorsi/plantarflexor torque is produced when both actuators pull/push in the same direction, whereas an inversion/eversion rotational torque is created when the actuators act in opposing directions.

IV. Discussion

Current commercial daily-wear AFO systems are generally limited to passive designs that control undesirable motion of the foot, but do not provide powered assistance during the propulsive phase of gait [12], [13]. These AFOs are successfully used as daily-wear devices because of the simplicity, compactness, and durability of the designs. While motion control can improve the functionality of an individual (e.g., the prevention of foot drop during swing), passive AFOs can impede gait at other points in the cycle (e.g., restrict range of motion during stance) and provide no supplemental torque assistance. Furthermore, the motion control provided by passive systems has limited functional benefit for an impaired individual because they do not adapt to a changing environment and may interfere with nonwalking tasks such as stair assent or decent.

Semiactive AFOs address some of the limitations of passive AFOs by utilizing sensors and controllable braking mechanisms to manage the motion of the foot. The most promising of these designs use MR dampers to modulate the resistance of the AFO based on the phase of gait [26], [27]. Of the MR designs, the untethered Halstead AFO appears to be the most fully developed technologically [27]. This AFO provides variable motion control of the foot and makes use of embedded sensors and predefined states to switch between level walking and stair climbing. While this particular system shows promise as a daily-wear AFO, it lacks the ability to provide supplemental assistive torque.

Fully active AFOs provide net power to the ankle for use in both motion control and propulsive assistance. To date, active AFOs have not been commercialized and exist only in laboratory settings. The size and power requirements of these designs have resulted in systems that are tethered to power supplies, electronics, or both. The active AFOs that make use of SEAs come the closest to the idealized AFO described in this paper, but are currently tethered to computers and power supplies [28], [29], [31]–[33]. The other active systems described in the survey are used in laboratory and clinical settings for rehabilitation and the direct measurement of the physical properties of the ankle joint complex.

The design considerations for the ideal AFO must account for the diverse functionality required to accommodate the many aspects of gait that can be affected by injury or pathology. It also must be compact and light weight to minimize the energetic impact to the wearer. These requirements illustrate the great technological challenges facing the development of nontethered, powered AFOs for daily wear. Currently, state-of-the-art AFOs fall short of the goal of a day scale portable powered orthosis. The core challenges that must be met to realize such a device for both daily wear and rehabilitation are 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; and 3) control schemes that efficiently and effectively apply assistance during any of a number of functional tasks that an individual may be expected to encounter on a daily basis.

Shape memory alloy (SMA), also called "muscle wire" and electroactive polymer actuators (EPAs) both show potential as future actuation methods for a portable AFO [44]. The key feature of SMA is its ability to undergo large seemingly plastic strains and to subsequently recover these strains when a load is removed or the material is heated. SMA has high power to mass ratio, which is consistent with the compactness goal of a portable AFO. However, the relatively slow response rate of this kind of actuator ($\sim$0.25 Hz) and the mechanical inefficiencies that result from poor conversion of thermal energy into mechanical energy (approximately 2–3%) are significant disadvantages that limit the applicability of this kind of actuator at the present time [44]. EPAs have been used in biologically inspired robotic arms, but have not been effectively incorporated into a prosthetic or orthotic device [45]. While EPAs offer potential as an actuator for a portable AFO in the future, the current state of the art lends itself to smaller applications that require soft and flexible actuation schemes (e.g., emulating the hovering ability of an insect) [46]. Electroactive polymers show promise as artificial muscles, but challenges including improved actuator durability and life time at high levels of performance, scaling up of force and stroke requirements to meet the need of orthotics and prosthetics devices, and efficient and compact driving electronics are needed before this can be realized [47].

Hydraulic and pneumatic fluid power systems also show potential as enabling devices for future orthotic devices, with many current research AFOs already using fluid power [19], [21], [23], [26], [37], [40], [48], [49]. Human motion is powered by relatively high torques acting at low velocity. Electric motors, on the other hand, are low-torque, high-velocity actuators. As a result, orthotic systems that make use of electric motors require colocated transmissions with the output at the joint. While the electric motor may be light, current off-the-shelf options for the transmission (planetary gear head or ball screw) are heavy. Fluid power is ideal for high-force, low-velocity applications like gait. The key advantages of fluid power are the high force/weight and force/volume of the actuator, the ability to actuate a joint without a transmission, and the ability to transport pressurized fluid to the actuator through flexible hoses that can be placed where a shaft from a traditional motor would not reach. This allows flexibility in the placement of system components on the assistive device or elsewhere on the body, resulting in compact packaging that does not sacrifice high force and power [50]. The ankle has a brief 200-W peak power with a 13 W average during a single gait cycle [51]. The 200-W peak power during gait is a good match for an accumulator-delivered power burst as accumulators have excellent power density. However, there are significant challenges associated with fluid power. While some aspects of a fluid power system are compact, fluid power generation requires a supply that can be large and noisy [50], e.g., a compressor powered by a combustion engine. Combination systems (electrohydraulic or electropneumatic) may also be a possible solution. Moreover, as the components and transmission lines of a fluid power system are reduced to a size suitable for an orthotic device, losses in efficiency are created by the proportionally larger friction forces that result from the smaller parts. These are among the reasons why most current applications for fluid power are large, heavy equipment.

The creation of a compact, light weight, efficient, portable powered AFO has the potential to dramatically improve both treatment and rehabilitation of injury and pathology at the ankle joint complex. In a daily-wear application, this improvement would result from an ability to accommodate a variety of functional deficits with a single device. Clinicians generally prescribe passive motion control AFOs that only address deficiencies associated with weak dorsiflexors for daily wear. This is due to the fact that a daily-wear assistive device capable of providing a supplemental torque at the ankle joint does not exist. A portable powered AFO would also be of use for in-home rehabilitation because it would allow the clinician to prescribe an at-home physical therapy routine built around the device. This application will broaden the impact of a portable powered device from those with a permanent deficit to any individual recovering from an acute ankle injury. While there are significant technological challenges that must be met to realize the enabling technologies that will result in a portable powered AFO, the successful development of this device will significantly advance the field of orthotics and benefit numerous individuals with lower limb neuromuscular deficiencies.

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References

[1] J. Perry, *Gait Analysis: Normal and Pathological Function*. Thorofare, NJ: SLACK, Inc., 1992.

[2] A. M. Dollar and H. Herr, "Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art," *IEEE Trans. Robot.*, vol. 24, no. 1, pp. 144–158, Feb. 2008.

[3] H. Sadeghi, "Local or global asymmetry in gait of people without impairments," *Gait Posture*, vol. 17, pp. 197–204, 2003.

[4] R. L. Waters and S. Mulroy, "The energy expenditure of normal and pathologic gait," *Gait Posture*, vol. 9, pp. 207–231, Jul. 1999.

[5] L. Ren, D. Howard, L.-Q. Ren, C. Nester, and L.-M. Tian, "A phase-dependent hypothesis for locomotor functions of human foot complex," *J. Bionics Eng.*, vol. 5, pp. 175–180, Sep. 2008.

[6] R. R. Neptune, S. A. Kautz, and F. E. Zajac, "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking," *J. Biomech.*, vol. 34, pp. 1387–1398, Nov. 2001.

[7] M. Meinders, A. Gitter, and J. M. Czerniecki, "The role of ankle plantar flexor muscle work during walking," *Scand. J. Rehabil. Med.*, vol. 30, pp. 39–46, 1998.

[8] S. Simon, R. Mann, J. Hagy, and L. Larsen, "Role of the posterior calf muscles in normal gait," *J. Bone Joint Surg.*, vol. 60, pp. 465–472, 1978.

[9] T. M. Kepple, K. L. Siegel, and S. J. Stanhope, "Relative contributions of the lower extremity joint moments to forward progression and support during gait," *Gait Posture*, vol. 6, pp. 1–8, Aug. 1997.

[10] S. Nadeau, D. Gravel, A. B. Arsenault, and D. Bourbonnais, "Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors," *Clin. Biomech.*, vol. 14, pp. 125–135, Feb. 1999.

[11] S. Barnett, A. Bagley, and H. Skinner, "Ankle weight effect on gait: Orthotic implications," *Orthopedics*, vol. 16, pp. 1127–1131, Oct. 1993.

[12] J. B. Redford, *Orthotics Etcetera.*. Baltimore, MD: Williams & Wilkins, 1986.

[13] G. K. Rose, *Orthotics: Principles and Practice*. London, U.K: Williams Heinemann, 1986.

[14] *Becker Orthopedic Catalog*, Troy, MI, 2009.

[15] H. B. Kitaoka, X. M. Crevoisier, K. Harbst, D. Hansen, B. Kotajarvi, and K. Kaufman, "The effect of custom-made braces for the ankle and hindfoot on ankle and foot kinematics and ground reaction forces," *Archives Phys. Med. Rehabil.*, vol. 87, no. 1, pp. 130–135, Jan. 2006.

[16] S. Yamamoto, M. Ebina, M. Iwasaki, S. Kubo, H. Kawai, and T. Kayashi, "Comparative study of mechanical characteristics of plastic AFOs," *J. Prosthetics Orthotics*, vol. 5, pp. 59–64, 1993.

[17] Tamarack Variable Assist Joint—Ankle Brace—Ankle Orthotic, Tamarack Habilitation Technologies, Inc., Blaine, MN, vol. 2009, 2008.

[18] S. Yamamoto, M. Ebina, S. Kubo, T. Hayashi, Y. Akita, and Y. Hayakawa, "Development of an ankle-foot orthosis with dorsiflexion assist, Part 2: Structure and evaluation," *J. Prosthetics Orthotics*, vol. 11, pp. 24–28, 1999.

[19] H. Teyssedre and G. Lefort, "Dynamic Orthesis," Eur. Patent Office, France, pp. 1–9, 2005.

[20] R. Chin, E. T. Hsiao-Wecksler, E. Loth, G. Kogler, S. Manwaring, K. A. Shorter, S. N. Tyson, and J. N. Gilmer, "A pneumatic power harvesting ankle-foot orthosis to prevent foot-drop," *J. NeuroEng. Rehabil.*, vol. 6, no. 19, Jun. 16, 2009.

[21] H. Hirai, R. Ozawa, S. Goto, H. Fujigaya, S. Yamasaki, Y. Hatanaka, and A. Kawamura, "Development of an ankle-foot orthosis with a pneumatic passive element," in *Proc. 15th IEEE Int. Symp. Robot Human Interactive Commun.*, Hatfield, U.K, 2006, pp. 220–225.

[22] O. Yokoyama, H. Sashika, A. Hagiwara, S. Yamamoto, and T. Yasui, "Kinematic effects on gait of a newly designed ankle-foot orthosis with oil damper resistance: A case series of 2 patients with hemiplegia," *Archives Phys. Med. Rehabil.*, vol. 86, pp. 162–166, 2005.

[23] M. Takaiwa and T. Noritsugu, "Development of pneumatic walking support shoes using potential energy of human," in *Proc. 7th Jpn. Fuild Power System Int. Symp. Fluid Power*, Toyama, Japan, 2008, pp. 781–786.

[24] S. Yamamoto, M. Ebina, M. Iwasaki, S. Kubo, H. Kawai, and T. Hayashi, "Comparative study of mechanical characteristics of plastic AFOs," *J. Prosthetics Orthotics*, vol. 5, pp. 59–64, Apr. 1993.

[25] T. F. Novacheck, C. Beattie, A. Rozumalski, G. Gent, and G. Kroll, "Quantifying the spring-like properties of ankle-foot orthoses (AFOs)," *J. Prosthetics Orthotics*, vol. 19, pp. 98–103, Oct. 2007.

[26] J. Furusho, T. Kikuchi, M. Tokuda, T. Kakehashi, K. Ikeda, S. Morimoto, Y. Hashimoto, H. Tomiyama, A. Nakagawa, and Y. Akazawa, "Development of shear type compact MR brake for the intelligent ankle-foot orthosis and its control; research and development in NEDO for practical application of human support robot," in *Proc. IEEE 10th Int. Conf. Rehabil. Robot.*, Noordwijk, The Netherlands, 2007, pp. 89–94.

[27] W. Svensson and U. Holmberg, "Ankle-foot-orthosis control in inclinations and stairs," in *Proc. IEEE Int. Conf. Robot., Autom. Mechatronics*, Chengdu, China, 2008, pp. 301–306.

[28] J. A. Blaya and H. Herr, "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 12, no. 1, pp. 24–31, Mar. 2004.

[29] S. Hwang, J. Kim, and Y. Kim, "Development of an active ankle-foot orthosis for hemiplegic patients," presented at the 1st Int. Convention Rehabil. Eng. Assistive Technol.: Conjunction 1st Tan Tock Seng Hospital Neurorehabil. Meet., Singapore, 2007.

[30] G. A. Pratt and M. M. Williamson, "Series elastic actuators," in *Proc. IEEE Int. Conf. Intell. Robots Syst.*, Aug. 1995, vol. 1, pp. 399–406.

[31] K. W. Hollander, R. Ilg, T. G. Sugar, and D. Herring, "An efficient robotic tendon for gait assistance," *J. Biomech. Eng.*, vol. 128, pp. 788–792, Oct. 2006.

[32] J. Hitt, A. M. Oymagil, T. Sugar, K. Hollander, A. Boehler, and J. Fleeger, "Dynamically controlled ankle-foot orthosis (DCO) with regenerative kinetics: Incrementally attaining user portability," in *Proc. IEEE Int. Conf. Robot. Autom.*, Apr. 2007, pp. 1541–1546.

[33] A. W. Boehler, K. W. Hollander, T. G. Sugar, and D. Shin, "Design, implementation and test results of a robust control method for a powered ankle foot orthosis (AFO)," in *Proc. IEEE Int. Conf. Robot. Autom.*, 2008, pp. 2025–2030.

[34] D. J. Weber, R. B. Stein, K. M. Chan, G. Loeb, F. Richmond, R. Rolf, K. James, and S. L. Chong, "BIONic WalkAide for correcting foot drop," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 13, no. 2, pp. 242–246, Jun. 2005.

[35] J. M. Hausdorff and H. Ring, "Effects of a new radio frequency-controlled neuroprosthesis on gait symmetry and rhythmicity in patients with chronic hemiparesis," *Amer. J. Phys. Med. Rehabil.*, vol. 87, pp. 4–13, Jan. 2008.

[36] H. Krebs, L. Dipietro, S. Levy-Tzedek, S. Fasoli, A. Rykman-Berland, J. Zipse, J. Fawcett, J. Stein, H. Poizner, A. Lo, B. Volpe, and N. Hogan, "A paradigm shift for rehabilitation robotics," *IEEE Eng. Med. Biol. Mag.*, vol. 27, no. 4, pp. 61–70, Jul./Aug. 2008.

[37] D. P. Ferris, J. M. Czerniecki, and B. Hannaford, "An ankle-foot orthosis powered by artificial pneumatic muscles," *J. Appl. Biomech.*, vol. 21, pp. 189–197, 2005.

[38] D. P. Ferris, G. S. Sawicki, and A. R. Domingo, "Powered lower limb orthoses for gait rehabilitation," *Topics Spinal Cord Injury Rehabil.*, vol. 11, pp. 34–49, 2005.

[39] D. P. Ferris, K. E. Gordon, G. S. Sawicki, and A. Peethambaran, "An improved powered ankle-foot orthosis using proportional myoelectric control," *Gait Posture*, vol. 23, pp. 425–428, 2006.

[40] M. Noel, B. Cantin, S. Lambert, C. M. Gosselin, and L. J. Bouyer, "An electrohydraulic actuated ankle foot orthosis to generate force fields and to test proprioceptive reflexes during human walking," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 16, no. 4, pp. 390–399, Aug. 2008.

[41] K. Bharadwaj, T. G. Sugar, J. B. Koeneman, and E. J. Koeneman, "Design of a robotic gait trainer using spring over muscle actuators for ankle stroke rehabilitation," *J. Biomech. Eng.*, vol. 127, pp. 1009–1014, 2005.

[42] A. Roy, H. I. Krebs, S. L. Patterson, T. N. Judkins, I. Khanna, L. W. Forrester, R. M. Macko, and N. Hogan, "Measurement of human ankle stiffness using the Anklebot," in *Proc. IEEE 10th Int. Conf. Rehabil. Robot.*, Noordwijk, The Netherlands, 2007, pp. 356–363.

[43] H. I. Krebs and N. Hogan, "Therapeutic robotics: A technology push," *Proc. IEEE*, vol. 94, no. 9, pp. 1727–1738, Sep. 2006.

[44] E. T. Esfahani, "Developing an active ankle foot orthosis based on shape memory alloys," M.S. thesis, University of Toledo, Toledo, OH, 2007, p. 161.

[45] Y. Bar-Cohen, "Electro-active polymers: Current capabilities and challenges," in *Proc. SPIE Smart Struct. Mater. Symp.*, San Diego, CA, 2002, pp. 1–7.

[46] S. G. Wax and R. R. Sands, "Electroactive polymer actuators and devices," presented at the SPIE Conf. Electroactive Polymer Actuators Devices, Newport Beach, CA, 1999.

[47] H. M. Herr and R. D. Kornbluh, "New horizons for orthotic and prosthetic technology: Artificial muscle for ambulation," in *Proc. SPIE Smart Struct. Mater. Electroactive Polymer Actuators Devices*, 2004, pp. 1–9.

[48] S. Yamamoto, A. Hagiwara, T. Mizobe, O. Yokoyama, and T. Yasui, "Development of an ankle-foot orthosis with an oil damper," *Prosthetics Orthotics Int.*, vol. 29, pp. 209–219, 2005.

[49] R. Chin, E. Loth, and E. Hsiao-Wecksler, "Fluid-power harvesting by pneumatic bellow during human gait," presented at the 8th Int. Symp. Fluid Power, ASME Fluids Eng. Conf., Jacksonville, FL, 2008.

[50] W. Durfee, Z. Sun, "Fluid power system dynamics," Minneapolis: University of Minnesota, Rep., Apr. 22, 2009.

[51] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, 3rd ed. Hoboken, NJ: Wiley, 2005.

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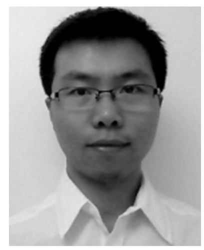

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