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Active Orthoses for the Lower-Limbs: Challenges and State of the Art

Aaron M. Dollar, *Member, IEEE,* and Hugh Herr, *Member, IEEE*

***Abstract*—In the many decades since researchers began to explore methods of creating them, active orthoses have progressed from complex research devices to nearly commercialized products. And while there are still many challenges associated with their development that have yet to be perfected, the advances in the field have been enormous. In this paper we review the history and discuss the state of the art of lower-limb active orthoses. We provide a design overview of hardware, actuation, sensory, and control systems for most of the devices that have been described in the literature, and end with a discussion of the major advances that have been made and hurdles yet to be overcome.**

I. Introduction

In the United States alone, approximately 6.7 million people could benefit from an advanced leg orthosis due to the effects of stroke, post-polio, multiple sclerosis, spinal cord injury, and cerebral palsy. In this review we focus on research in the development of active orthotic devices that might allow members of this population to ambulate in a more natural, efficient manner than they might with traditional passive orthoses.

Besides mention in early patents and science fiction [1], research in active orthoses and exoskeletons began in the late 1960s, almost in parallel between a number of research groups in the United States and in the former Yugoslavia. However, the former was primarily focused on developing technologies to augment the abilities of able-bodied humans, often for military purposes, while the latter was intent on developing assistive technologies for handicapped persons. Despite the differences in intended use, these two fields face many of the same challenges and constraints, particularly related to portability and interfacing closely to a human operator.

In this paper, we will present a review of the work done on active orthoses for the lower limbs in the past decades. The term 'active orthosis' is typically used to describe a device intended to increase the ambulatory ability of a person suffering from a leg pathology by providing some means of augmenting the power at one or more joints of the lower extremities. This includes both adding and dissipating power, as well as the controlled release of energy stored in springs during various phases of the gait. Occasionally, the term 'exoskeleton' is used to describe certain rehabilitation devices of this type, particularly when they encompass the majority of the joints of the lower limbs.

Unlike passive orthoses, active orthotic devices have the potential of actively controlling the joints of the devices, rather than just simple mechanical coupling that exists with the most common commercial assistive devices. Architectures in which power or torque is added at appropriate phases of the gait cycle might be able to enable users to walk who otherwise could not with passive devices, or allow them to walk more naturally and/or efficiently. Additionally, portable devices such as these have the potential of providing both assistance and therapy at the same time, an extremely desirable property in rehabilitation.

In addition to limiting our scope to active orthotic devices for the lower limbs, we do not cover devices whose active components simply lock and unlock joints of an orthosis, nor systems that are purely a hybrid of a passive orthotic brace and a method of functional electrical stimulation (FES) control. Finally, exoskeletons for therapy that are not portable and stand-alone mechanically (e.g. treadmill-based devices such as the Lokomat [2]) are not discussed, as these are not subject to the vast number of constraints associated with portable devices.

We attempt to cover all of the major developments in these areas, focusing particularly on the initial development of the different concepts, and less on similar devices built for research purposes. When available, studies on clinical efficacy are described, however there are surprising few instances of such studies being reported.

We begin with a brief background on the biomechanics of human walking in order to describe some of the terminology used in this review. We then move on to reviewing the literature on active orthoses, covering full exoskeletal, modular, and single joint devices. Finally, we present a discussion summarizing the major accomplishments in the field and identifying areas that still need to be addressed.

II. Biomechanics of Walking

Before getting into our review, it will be useful to provide a brief background on the biomechanics of human walking, as this information plays a crucial role in the design of such systems. Fig. 1 (adapted from [3]) shows a simplified diagram of human walking gait, with terms that will be used

A. M. Dollar is with the Harvard/MIT Division of Health Sciences and Technology and the MIT Media Lab, Massachusetts Institute of Technology, Cambridge, MA 02139 USA (e-mail: adollar@mit.edu).

H. Herr is with the MIT Media Lab and the Harvard/MIT Division of Health Sciences and Technology, Massachusetts Institute of Technology, Cambridge, MA 02139 USA (corresponding author phone: 617-314-3661; fax: 617-253-8542; e-mail: hherr@media.mit.edu).

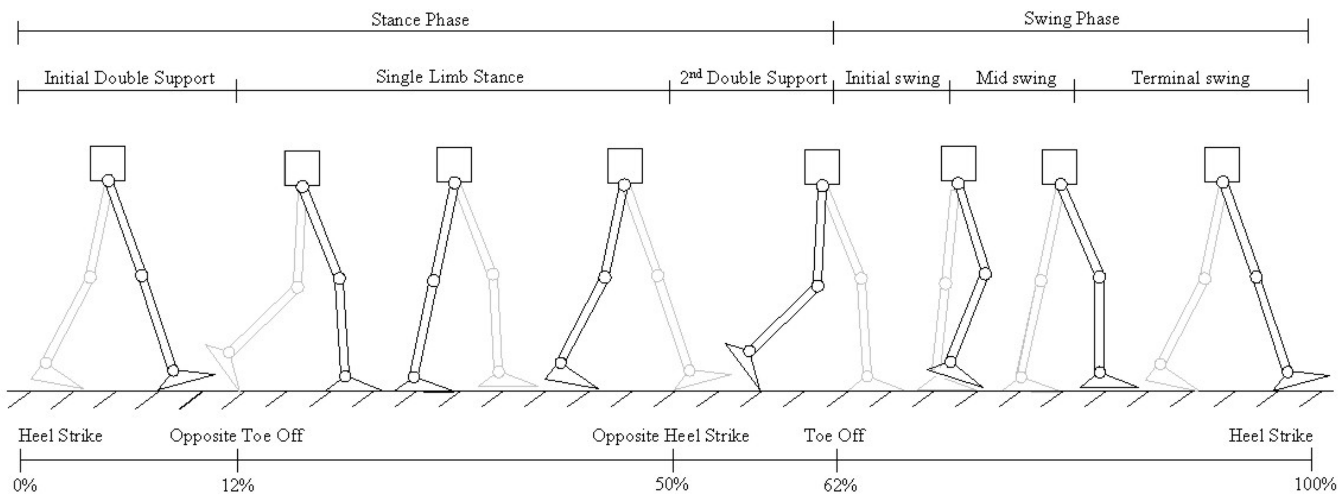

Fig. 1. Human walking gait through one cycle, beginning and ending at heel strike. Percentages showing contact events are given at their approximate location in the cycle. Adapted from [3].

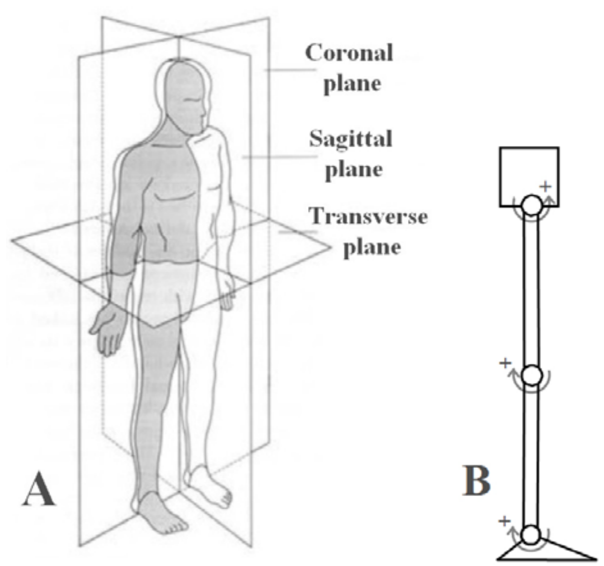

Fig. 2. Description of the anatomical planes (A) and a diagram of the leg shown in the rest position (0 deg at all joints) with the positive direction indicated (B).

throughout this paper. Note that the timing of the labeled events during the gait cycle is approximate, and varies across individuals and conditions. The human walking gait cycle is typically represented as starting (0%) and ending (100%) at the point of heel strike on the same foot, with heel strike on the adjacent foot occurring at approximately 62% of gait cycle.

In general, the human leg can be thought of as a seven degree of freedom structure, with three rotational degrees of freedom at the hip, one at the knee, and three at the ankle. Fig. 2 shows a description of the human anatomical planes (Fig. 2A) as well as a kinematic model of the human leg in the sagittal plane, which is the dominant plane of motion during human locomotion (Fig. 2B). In this review paper, joint motion in this plane is referred to as flexion (positive direction) and extension (negative direction). Additionally, motion of the hip in the coronal plane is referred to as abduction (away from the center of the body) and adduction. Further, motion of the ankle in the coronal plane is referred to as eversion (away from the center of the body) and inversion. The remaining degrees of freedom of the hip and ankle are referred to as simply 'rotation'. These various terms are used throughout this paper in describing the kinematic layout of the various orthosis designs.

Figs. 3 and 4 show the biomechanics of a normal, healthy individual (82 kg, 0.99m leg-length, 28 year-old male walking at 1.27 m/s), showing joint angle and power for hip, knee, and ankle flexion/extension motions during level-ground walking. Details of the experimental methods used to capture these data can be found in [4]. While walking data can differ somewhat across subject and condition, the qualitative nature of the curves remains similar (e.g. [5-8]).

It is particularly useful to note the power requirements of each joint. From gait data it can be seen that, particularly at slow walking speeds, power at the hip is positive or near zero, power at the knee is predominantly negative (dissipates power), and power at the ankle is somewhat evenly split between positive and negative. Note that during level ground walking, the net power of the individual as a whole *should* be close to zero, since no net work is being done and resistance to motion is small.

Considering the results of Fig. 4, powered exoskeletons and orthoses often incorporate means of adding power at the hip, dissipating power at the knee (e.g. brake or damper), and storing energy at the ankle using passive elastic structures. However, these results change dramatically when the subject walks at moderate to fast speeds, or on a positive incline or upstairs. For this reason many devices enable power to also be added at the knee and sometimes the ankle.

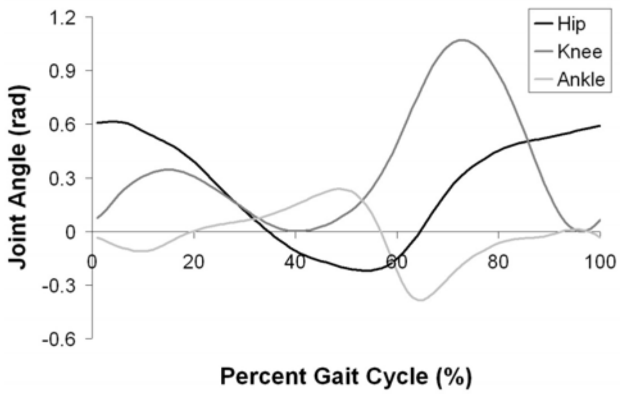

Fig. 3. Representative angles of the leg flexion/extension joints over the gait cycle, beginning and ending at heel strike. Data is from motion in the sagittal plane (flexion/extension).

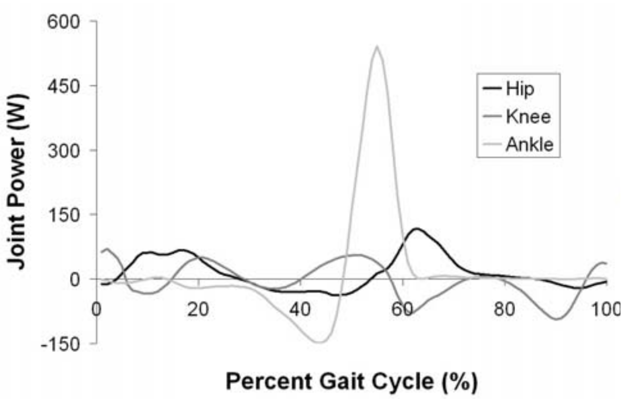

Fig. 4. Representative power at the leg flexion/extension joints over the gait cycle, beginning and ending at heel strike. Data is from motion in the sagittal plane (flexion/extension).

A. *Metabolic Cost of Transport*

One key performance measure in demonstrating the effectiveness of an active orthotic device is metabolic cost of transport (COT). This measure of how physically taxing the activity is to the subject is attained by measuring the oxygen consumption and carbon dioxide production of human breathing during a task [9,10]. A number of inexpensive, compact systems for measuring these parameters exist (e.g. the K4 telemetric system (Cosmed srl, Rome, Italy) [11]). Comparison of metabolic COT between performing the task with the active orthosis and another enabling device is a good determinant as to whether there is any energetic advantage to using the active orthosis.

III. Review of Active Orthoses

In this section, we describe work done in developing orthotic devices that improve upon traditional passive braces by some combination of adding or dissipating power at the joints of the device and/or the controlled release of energy stored in springs during appropriate phases of the gait.

A. *Early Active Orthoses*

As would be expected, early active orthotic devices were essentially standard braces that were modified to provide some sort of active assistance. The first mention of such a device that could be found is a US patent from 1935 (Fig. 5) [12]. The device was essentially a leg brace with reciprocating motion at the knee. A crank located at the hip was used to wind up a torsional spring located on the knee joint, which drove the joint through a preset motion determined by a cam and follower. The brace interfaced with the wearer via a foot connection, straps around the thighs, and a torso strap.

The first controllable active orthosis that could be found is a patent for a hydraulically-actuated device from 1942 for adding power at the hip and knee joints [13]. However, due to the state of the art in controls technology at the time, the device was "controlled" by the physical opening and closing of the hydraulic valves by a cable and linkage system that activates at certain joint angles in the gait cycle. Another early patent from 1951 describes a similar passive device that uses spring-loaded pins for locking and unlocking the joints of the brace at various stages of the wearer's gait [14].

B. *Full Lower-limb Exoskeletons*

1) *Mihailo Pupin Institute Exoskeletons*

The pioneering work done with exoskeletons by Miomir Vukobratovic and his associates at the Mihailo Pupin Institute in Belgrade in the late 1960s and 1970s is some of the most extensive to date [15-17]. The work started with a passive device for measuring the kinematics of walking and then quickly progressed to the development of powered exoskeletons. The earliest of these, the 'kinematic walker', featured a single hydraulic actuator for driving the hip and knee, which were kinematically coupled. In 1970, the so-called 'partial active exoskeleton' was developed, which incorporated pneumatic actuators for flexion/extension of hip, knee, and ankle, as well as an actuated abduction/adduction joint in the hip for greater stability in the frontal plane. This concept was later slightly modified into the 'complete exoskeleton' by extending the attachment at the torso to enclose the entire chest of the patient, providing greater trunk support (Fig. 5). More than 100 clinical trials were performed with this device, and a number of patients with varying degrees of paralysis mastered walking using the complete exoskeleton with support from crutches.

These devices interfaced with the wearer via shoe bindings, cuffs around the calves and thighs, and a 'corset' on the torso. This corset also holds the 14 solenoid valves for the control of the pneumatic pistons. The total weight of the 'complete' exoskeleton, after incorporation of lighter valves, was 12 kg. This value does not include the power source and control computer, which are not located on the device.

During operation, all of the above exoskeleton devices

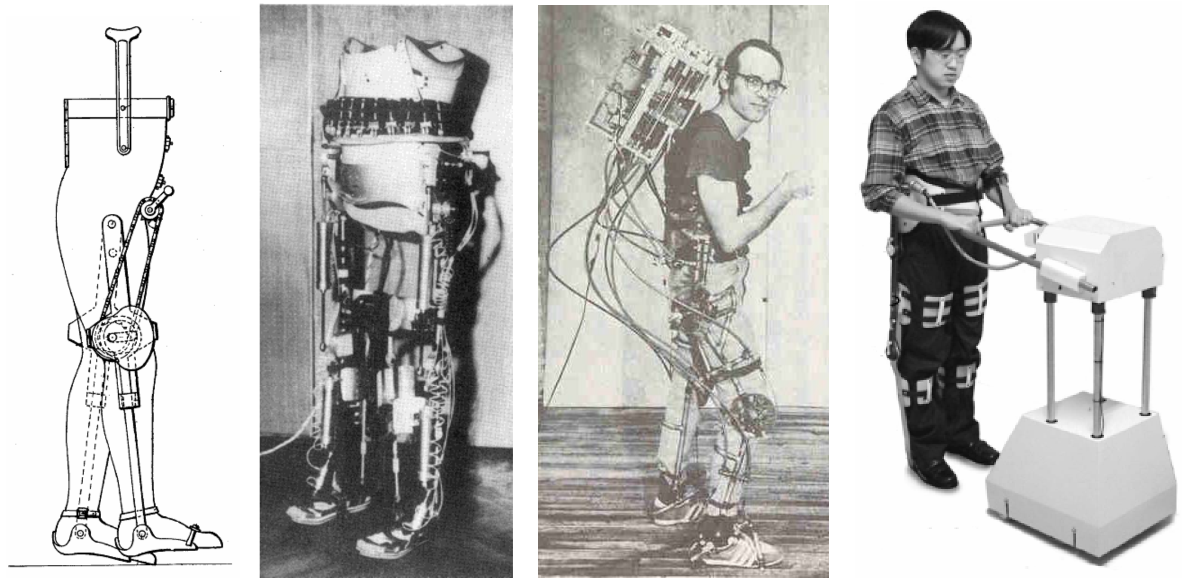

Fig. 5. Cobb's "wind-up" orthosis [12], Pupin Institute 'complete' exoskeleton [15], Wisconsin exoskeleton [19], and Sogang orthosis and walker [22]. Image credits (from left to right): U.S. Patent 2,010,482; Prof. Dr Miomir Vukobratović, Robotics Laboratory, Mihailo Pupin Institute, Volgina 15, 11060 Belgrade, Serbia; Jack Grundmann, University of Wisconsin; Kyoungchul Kong and Doyoung Jeon, Sogang University, Seoul, Korea.

were driven through a predetermined reciprocating motion via an 'electronic diode' function generator. However, a set of three piezo-ceramic force sensors were soon incorporated into the sole of the 'complete' exoskeleton foot for use in determining the location and magnitude of the ground reaction force, which in turn was used in the control of the device.

In order to begin to address the problem of being energetically autonomous, a version of the exoskeleton actuated by DC motors was developed. Although the state of motor, battery, and computer technology limited the true portability of the device, this new actuation scheme offered further improvements such as smoother motion and better tracking ability.

One of the most lasting contributions of their work with exoskeletons is in control methods for robotic bipeds. Indeed, Professor Vukobratovic along with Devor Juricic are credited with developing the concept of the 'zero moment point' and its role in the control of bipedal locomotion [18].

A thorough history of the work done with exoskeletons at the Mihailo Pupin Institute is provided in [16]. The same text also briefly describes exoskeletons developed at the University of Tokushima in Japan in 1973 and the Central Institute for Traumatology and Orthopaedy in Moscow in 1976. However, no references are given in the text concerning these devices and none could be found during this review.

2) University of Wisconsin Exoskeleton

Another full, lower-limb exoskeleton was developed at the University of Wisconsin beginning in 1968 (Fig. 5) [19,20]. Similar to the Pupin Institute exoskeletons, this device was intended to help re-ambulate paraplegics with full upper-body capabilities. The kinematic design of the exoskeleton featured universal joints at the hip and ankle (three rotational degrees of freedom each) as well as a single rotational joint at the knee. The flexion/extension joints at the hip and knee were powered by rotary hydraulic actuators, and the remaining degrees of freedom were either completely passive or spring-loaded.

The hydraulic power unit consists of a battery-powered DC motor driving a hydraulic pump. These systems, including the servo-valves for each of the four actuators, are located on the fiberglass corset around the waist of the operator. The entire exoskeleton device was physically autonomous except for its control, which was done on an off-board computer. A thorough discussion of the design and control of the device can be found in [20].

The Wisconsin exoskeleton was intended to provide the wearer with the ability to sit down and stand up in addition to walking at half normal speed. The operator needed to use a pair of canes for better stabilization. The device was programmed to follow joint trajectory data recorded from a similarly sized able-bodied individual in a feed-forward, open loop manner.

It is unknown whether tests with a paraplegic operator were ever conducted. However, experiments with an able-bodied wearer using two canes for support showed stable, 'natural-seeming' operation. Additionally, the operator was able to wear the device for several hours at a time without discomfort.

3) Other full Lower-limb exoskeletons

Researchers in the Departments of Mechanical Engineering and Physical Therapy at the University of Delaware have developed a passive leg orthosis that is designed to reduce the forces of gravity on the patient during walking, easing the effort required for locomotion [21]. This device utilizes an interesting combination of springs and linkages in order to geometrically locate the center of mass of the leg-orthosis system and then balance out the effect of gravity.

The authors present thorough experimental work with their device on five able-bodied and individual with paralysis in the right leg due to stroke. Among other things, the results showed that the current implementation of the device, while not affecting required torques at the knee, reduced the average torque required from the patient's hip by 61%.

An interesting concept to alleviate some of the difficulties in creating a portable active orthosis device is presented by researchers at Sogang University in Seoul, Korea [22]. The device consists of a full lower limb orthosis paired with a specially designed walker that houses the battery, DC motors, and control computer, greatly reducing the weight of the orthosis (Fig. 5). A cable drive transmits power to the joints of the wearer from the actuators in the walker. Do to the transmission, the wearer is held to a fixed distance from the walker. The orthosis adds power in the flexion/extension directions of the hips and knees, and allows motion in the other degrees of freedom of the leg, except rotation of the ankle, which is fixed. User intent is sensed by a combination of joint angle sensors and a pressure sensor giving a sense of force being applied by the quadriceps muscle.

Another interesting aspect of the design of this device is that the handlebars of the walker move up and down with the operator by sensing joint angles of the brace, facilitating sitting and standing. The walker moves actively with the operator, mounted on powered casters.

Another novel idea proposed in the literature is a combination of powered orthosis, powered telescoping crutches, and roller skate-like mobile platforms under the user's feet [23]. The orthosis and crutches are designed to assist in standing and sitting as well as ascending and descending stairs. The mobile platforms are only intended to be used to assist motion over level ground, during which the joints of the orthosis lock the user in an upright posture. One can imagine, however, this strategy leading to problems with the stability of the wearer.

Researchers at Michigan Tech developed an experimental powered gait orthosis consisting of one degree of freedom per leg with actuated hip and knee joints connected by linkages [24]. The device was used to study the power required for a fully-actuated device, finding that it requires a peak electrical power of 304W, nearly identical to the approximately 300W peak power required during human walking [3]. Their results are particular to their electric motor actuation scheme.

Darwin Caldwell, who has been active in research in upper-limb exoskeletons, also developed a ten degree of freedom lower-limb exoskeleton device [25]. Actuation is provided to the flexion/extension directions of the hip, knee, and ankle, and abduction/adduction of the hip via pneumatic muscle actuators.

Researchers at Tokyo Denkai University have proposed proposed their own orthosis design that is powered by a custom-designed bilateral hydraulic servo actuator [26]. This device is intended for use in therapy for gait training, and requires the use of a custom frame that houses the power supply and also aids walking.

A number of groups have published work on active orthotic devices that did not progress past the stage of preliminary investigations [27,28]. [29] presents a concept in which the orthosis is controlled via sensed motions of the users fingers. Another concept uses contact sensors at the base of crutches to determine whether the user is in a stable stance and then allows the joints of the orthosis to be appropriately activated [30].

C. Modular Active Orthoses

1) AMOLL Project

The first published work with modular active orthoses is the AMOLL project (Active Modular Orthosis for Lower Limbs, headed by Pierre Rabischong), which incorporated researchers from Montpellier and Toulouse, France, University of Belgrade, and Stanford Reasearch Institute [31]. The concept advanced the idea of an inflatable interface with the wearer, a concept first introduced by the French company Aerozur as "soft suits" (see [32]). Being modular, only components necessary for the ambulation of the specific patient needed to be utilized. Actuation was to be available for both the hip and knee components in flexion/extension, while the unactuated degrees of freedom at the hip were to be stiffened by rigidity in the orthosis. Actuation was not yet implemented in this initial paper, however DC servo motors were proposed. A method of control is proposed in [33].

Later, J.W. Hill of the Stanford Research Institute described work on the design of a hydraulically powered orthosis that he performed under the AMOLL project [34]. The work focuses on methods of increasing the efficiency of a hydraulic power source and a control algorithm based on joint angles for walking with the device. The author also mentions the potential benefit of an unpowered hydraulic device, as it can still be used to lock the joints of the orthosis during appropriate phases of the gait.

2) Univ. of Belgrade and Zotovic Rehab. Inst.

It is unknown whether their work continued to fall under the umbrella of the AMOLL project, but researchers in Belgrade continued work in modular orthoses until the 1990s. Rajko Tomovic (one of the authors on the original AMOLL project paper [31]) and more significantly Dejan Popovic (both from the University of Belgrade) and Laslo Schwirtlich (from the Dr. Miroslav Zotovic Rehabilitation Institute) continued with what they called "self-fitting modular orthoses" [32,35], devices similar to the inflatable components mentioned under the original AMOLL proposal [31].

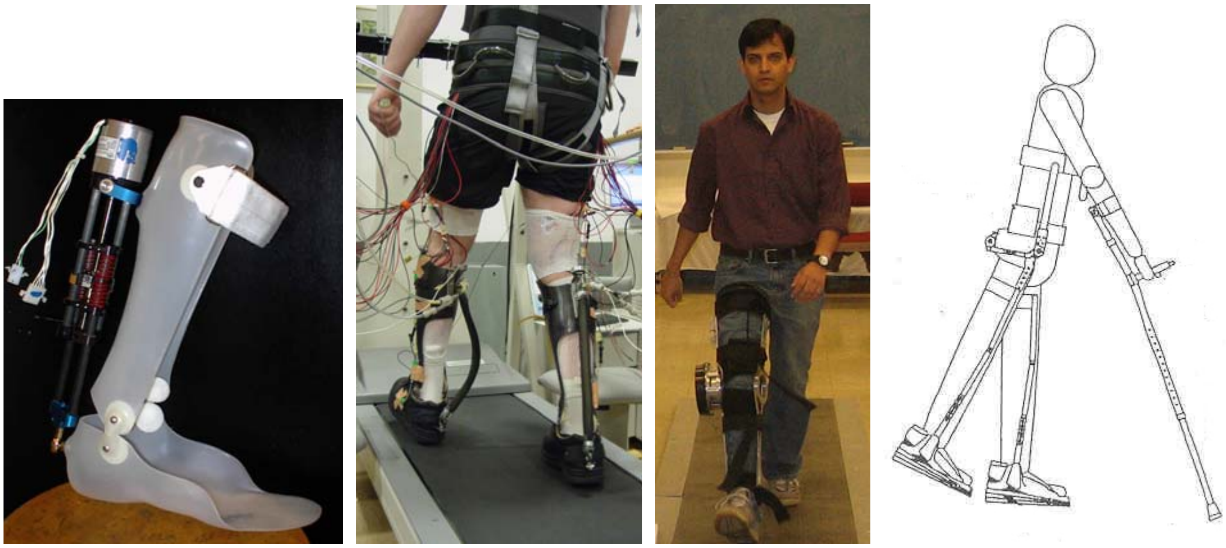

Fig. 6. MIT active ankle-foot orthosis [41], Michigan ankle orthoses [42], Northeastern University knee orthosis [48], and the weight-bearing control orthosis [53]. Image credits (from left to right): Prof. Hugh Herr, Biomechatronics Laboratory, MIT Media Lab, Cambridge, Massachusetts; Prof. Dan Ferris, Human Neuromechanics Laboratory, Division of Kinesiology, University of Michigan, Ann Arbor, Michigan; Prof.Constantinos Mavroidis, Robotics and Mechatronics Laboratory, Northeastern University, Boston, MA, http://www.robots.neu.edu; Dr. Noritaka Kawashima, National Rehabilitation Center for the Disabled, 4-1 Namiki, Tokorozawa City, Saitama Pref. 359-8555, Japan.

Popovic's and Schwirtlich's work with these modular devices quickly advanced to developing the first hybrid system combining a powered orthosis with Functional Electrical Stimulation (FES) [36-38]. This system is intended to extend the use of FES to patients lacking the control or muscle strength needed for the established combination of passive brace and electrical stimulation. These systems were shown to allow a patient to walk faster than either a self-fitting modular orthosis or FES individually.

3) Mihailo Pupin Institute

Vukobratovic and his associates at the Mihailo Pupin Institute also investigated modular active orthoses, allowing for hip and/or knee sections to be added depending on the ability of the individual patient. An interesting aspect of their device is the microprocessor control system mounted on the torso support, allowing the wearer to select level ground, stair ascension, and stair descension gaits, as well as gait pace, stride length, and turn direction. Like the latest version of their full exoskeleton, this "active suit" was actuated via DC motors and was not energetically autonomous [39].

D. Single Joint Active Orthoses

1) Active Ankle-Foot Orthoses (AFOs)

An early active ankle orthosis was presented in 1981 by Jaukovic at the University of Titograd in the former Yugoslavia [40]. The device consisted of a DC motor mounted in front of the wearer's shin which assisted in the flexion/extension of the ankle via some means of transmission that is unclear from the paper. Also included was a specially-designed "junction" that allowed free movement of the ankle. The orthosis was controlled based upon information from foot switches in the soles.

a) MIT Ankle-Foot Orthosis

The MIT Biomechatronics Lab developed a powered ankle-foot orthosis to assist drop-foot gait, a deficit affecting many persons who have experienced a stroke, or suffer from multiple sclerosis or cerebral palsy, among others (Fig. 6) [41]. The device consists of a modified passive ankle-foot orthosis with the addition of a series elastic actuator (SEA) to allow for variation in the impedance of flexion/extension direction of ankle motion, controlled based on ground force and angle position data.

In clinical trials, the MIT active AFO was shown to improve the gait of dropfoot patients by increasing walking speed, reducing the instances of "foot slap", better symmetry with the unaffected leg, and assistance during powered plantar flexion. Feedback from the subjects was also extremely positive. The device, however, was built mostly as an experimental tool and requires further work to make it portable.

b) University of Michigan Orthoses

Dan Ferris's Lab at the University of Michigan has produced a number of active orthoses, particularly focusing on rehabilitation devices to be used during therapy [42-44]. Accordingly, these devices are not meant to be fully portable, and are mostly pneumatically actuated, with a tether to a stationary compressor. The pneumatic actuators used are artificial pneumatic muscles (McKibbon muscles), and these are mounted to carbon fiber and polypropylene shells, resulting in devices are extremely lightweight as well as have high power outputs. Additional benefits lie in the low impedance of the actuators, resulting in safer devices.

The University of Michigan orthoses are primarily for the lower leg, having created both ankle-foot and knee-ankle-foot devices. For all devices, carbon fiber and polypropylene shells are custom built for each subject, eliminating the need for mechanically complex adjustment mechanisms.

For ankle-foot orthoses, versions including an agonist/antagonist pair as well as a single plantar flexion actuator (in the positive direction according to Fig. 2B). The latter device was tested on six subjects with chronic incomplete spinal cord injury walking at slow speeds (0.54 m/s) under partial body weight support (30% or 50% depending on the abilities of the individual) provided via a harness (Fig. 6). The results showed that while providing increased plantar flexion at the end of the stance phase, it did not decrease muscular recruitment as measured by surface EMG on the soleus and gastrocnemius muscles.

A knee-ankle-foot orthosis that is an extended version of the ankle-foot orthosis was also developed, and incorporates an additional agonist/antagonist pair of artificial muscles for the flexion/extension of the knee [44].

c) Other Ankle-Foot Orthoses

At Arizona State University, researchers have presented a novel design of an active ankle-foot orthosis with two 'spring over muscle' actuators attached to the left and right sides of the foot under the toes, forming a tripod with the heel [45]. These actuators are essentially pneumatic muscles with an internal spring tending to extend the muscle, enabling force to be applied in both directions. The tripod configuration allows the angle to be actuated in flexion/extension (co-activation) and inversion/eversion (single activation). Additionally, the group has also explored using series elastic actuators to power orthosis joints [46].

Researchers in the Departments of Mechanical Engineering and Physical Therapy at the University of Delaware have also proposed a design of an active ankle orthosis that adds power to the wearer in both the flexion/extension and inversion/eversion directions [47].

2) Active Knee Orthoses

Dinos Mavroidis' Lab at Northeastern University has developed a dissipative knee orthosis by combining an electro-rheological fluid-based variable damper with a modified commercial knee brace (Fig. 6) [48,49]. This device is intended to provide resistive torques to the user for rehabilitation purposes, and was designed to provide approximately 30 Nm of torque to the wearer, approximately 25% of the maximum knee torque of the average human ability.

Researchers at the Berlin University of Technology are developing an orthosis to add power at the knee via a DC motor and ballscrew transmission [50,51]. However, work up to this point has been focused primarily on developing an EMG-based control system for the device, with this system to be implemented with the hardware in future work.

Finally, a knee orthosis powered by pneumatic muscles supporting the wearer during deep knee bends is briefly reported in [52].

E. Other Orthotic Devices

Although they are not within the scope of this review, reciprocating gait orthoses (RGOs) are worth briefly mentioning. These devices lock the wearer's knees and couple the two hip joints in such a way that the flexion of one hip occurs by the extension of the opposite hip. By this method, the wearer is able to support their body weight and perform a straight-legged method ambulation, although with the support of canes or a walker.

An interesting concept proposed by researchers in Saitama, Japan is essentially a standard RGO with a modified shoe in which the thickness of the sole is actively controlled in order to compensate for the pendular motion enforced by the locking of the knees in an RGO (Fig. 6) [53,54]. In this way, the ground is effectively raised and lowered in order to compensate for the lack of degree of freedom at the knee. Experimental results with this device show a significant increase in walking speed and decrease in energy cost as compared to the results of other studies in which a traditional RGO were used [54].

An RGO was modified to include actuation at the hip and knees by researchers in Torino, Italy [55]. The orthosis uses double-acting pneumatic cylinders for actuation, with an off-board compressor. Another modified RGO, with power added at the hip via a brushless DC motor is presented in [56].

A number of researchers have investigated combinations of reciprocating gait orthoses and functional electrical stimulation (FES) [57-63]. Will Durfee at the University of Minnesota has been actively involved in research with orthotic devices for many years. One device is a full lower-limb othosis incorporating controllable brakes at the hips and knees (flexion/extension) with a method of functional electrical stimulation (FES). By activating the brakes to stiffen the orthosis during standing, the device only requires the patient's muscles to be used during motion. This enables FES to be used much more frequently (shorter duty cycle) as well as reduces muscle fatigue [64,65].

Results of testing on a T6 complete paraplegic utilizing the hybrid controlled brake-FES system showed a much more repeatable gait than with FES alone. Additionally, with the hybrid system the patient's muscles only needed to be stimulated during 10% of the gait cycle, in comparison with 85% for FES alone.

IV. DISCUSSION

In the process of doing this review, a number of themes related to the challenges associated with building functional, autonomous active orthotic devices kept reappearing. Power supply, lightweight actuators, and efficient transmissions are among the many issues that all researchers in this area have had to face. It has become obvious, particularly to those in the more advanced stages of development that, for many of the power, actuation, and other subsystems, off the shelf components do not meet the low weight, high efficiency, and other criteria needed to accomplish their design objectives. Indeed, this is a problem facing also facing many fields of mobile robotics, particularly with anthropomorphic architectures.

And while these issues are still being addressed, a number of great advances have been made in the areas related to active orthoses in the last five decades. The field of biomechanics of human locomotion has fully matured in recent decades, providing the necessary background science for the design of devices that closely mimic the dynamics of the operator's motion. Battery and DC motor technology has greatly advanced in recent years, although they still are not at a point where they meet the demands of many exoskeleton and orthosis applications. The state of the art in computing, sensing, and control has, of course, advanced so dramatically that these areas are no longer major obstacles to the implementation of robotic hardware.

There is a marked lack of published quantitative results on the performance of the active orthotic devices that have been developed. Considering this, one is left to wonder what the real advantages of these complicated, expensive systems really are. For these devices, comparison with established assistive devices is a logical avenue. For instance, a device meant to assist ambulation in someone who might otherwise be able to ambulate with the assistance of a reciprocating gait orthosis, should be tested against results with that device. Performance measures that can be used include metabolic cost of transport [66,67], walking speed, smoothness and repeatability of motions, muscle fatigue, and stability, among others.

Rather than minimizing the accomplishments that have been made in the field, the lack of quantitative results instead highlights the numerous challenges associated with creating active orthoses. There are of course many design issues that may lead to poor performance: misalignment of joints between operator and hardware, kinematic constraints from attachments such as harnesses and cuffs, design not optimized for the specific gait, added forces to the operator that resist motion, and addition of power in a sub-optimal manner (e.g. mis-timing, too little, too much), among others.

Active orthoses also face the daunting issue that the specific nature of a disability varies widely from one patient to the next. This makes the development of a generally applicable device difficult. This is in fact a challenge for many assistive devices. To our knowledge, there are currently no commercially available autonomous orthoses that provide active assistance to the wearer. Exoskeletons that are purely meant for clinical therapy purposes are currently effective as stand-alone, treadmill-based devices such as the Lokomat [2], however there is great value in developing a portable device that can be used outside of the laboratory. Ideally, one would like a compact, energetically autonomous orthosis that can provide both assistance and therapy during the wearer's every day life.

The issue of portability is one of the major factors in limiting the application of active orthoses outside of laboratory-based therapy. The vast majority of the orthotic devices covered in this review were not energetically autonomous, typically being tethered to some external power supply – air compressors, hydraulic pumps, or electrical power.

Future directions in work related to the creation of portable active orthotic devices will likely center around the 'enabling' technologies such as power supplies, actuators, and transmissions that are lightweight and efficient. Interestingly, a large portion of these developments necessary for further advances are currently being driven by the exoskeleton and active orthosis research community itself, and not by other, more pervasive issues like those that drove developments in computing, sensing, and control.

There are a few areas related to the mechanical design of active orthoses that show promise and have been largely overlooked. An improved understanding of muscle and tendon function in walking and other movement tasks may shed light into more effective exoskeleton leg architectures. Models based on actual machine elements that capture the major features of human locomotion (e.g. [68,69]) may enhance understanding of human leg morphology and control and lead to analogous improvements in the design of efficient, low-mass orthotic devices.

Investigation of non-anthropomorphic architectures may provide solutions to some of the problems associated with closely matching the structure of the exoskeleton to the wearer such as the need for close alignment between the joints of the robot and wearer.

Besides enabling technologies and mechanical design, there are a few issues related to the implementation of exoskeletons and active orthoses that have been largely ignored. Studies on the safety of the human operator, who is strapped inside the powerful exoskeleton device, have yet to be performed. Additionally, effective strategies for interfacing an active orthosis to the human body both mechanically and neurally are important areas of future research.

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