Lower Extremity Exoskeletons and Active Orthoses: Challenges and State-of-the-Art

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***Abstract*—In the nearly six decades since researchers began to explore methods of creating them, exoskeletons have progressed from the stuff of science fiction to nearly commercialized products. While there are still many challenges associated with exoskeleton 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 exoskeletons and 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.**

***Index Terms*—Exoskeleton, lower extremity, orthosis, orthotics, rehabilitation, robotics, walking, wearable.**

I. Introduction

BESIDES mention in early patents and science fiction [1], research in powered human exoskeleton devices 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 physically challenged persons. Despite the differences in the intended use, these two fields face many of the same challenges and constraints, particularly those related to portability and interfacing closely to a human operator. For this reason, we address both of these applications.

For the purposes of this review, an exoskeleton is defined as an active mechanical device that is essentially anthropomorphic in nature, is "worn" by an operator and fits closely to his or her body, and works in concert with the operator's movements. In general, the term "exoskeleton" is used to describe a device that augments the performance of an able-bodied wearer. The term "active orthosis" is typically used to describe a device that is used to increase the ambulatory ability of a person suffering from a leg pathology. Occasionally, however, the term "exoskeleton" is also used to describe certain assistive devices, particularly when they encompass the majority of the lower limbs.

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We will focus our review on exoskeleton and orthotic devices for the lower limbs, and only cover devices that operate in parallel with the human legs, as opposed to devices such as the Spring Walker [2] that operate in series with the wearer. For active orthoses, we will limit our scope to devices that provide some means of augmenting 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 gait. Along these lines, we do not include 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 a functional electrical stimulation (FES) control. Finally, exoskeletons used for therapy that are not portable and do not stand-alone mechanically (e.g., treadmill-based devices such as the Lokomat [3]) 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 the areas described before, particularly focusing on the initial development of the different concepts, and less on similar devices built for research purposes. When available, we describe the results of any quantitative evaluation of the effectiveness of the exoskeleton and orthotic devices in performing their intended tasks; however, there are surprisingly 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 as well as the science behind many of the working concepts of the devices that we cover. We then move on to reviewing the literature, beginning with performance-augmenting exoskeletons, and then, exoskeletons and orthoses that act as assistive devices for physically challenged persons. Finally, we present a discussion of this information, summarizing the major accomplishments in the field and identifying research areas that have yet to be addressed.

II. Biomechanics of Walking

Understanding the biomechanics of human walking is crucial in the design of exoskeletons and active orthoses for the lower limbs. Therefore, before getting into our review, we provide a brief background of the most relevant concepts. Fig. 1 (adapted from [4]) shows a simplified diagram of human walking gait, with terms that will be used 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

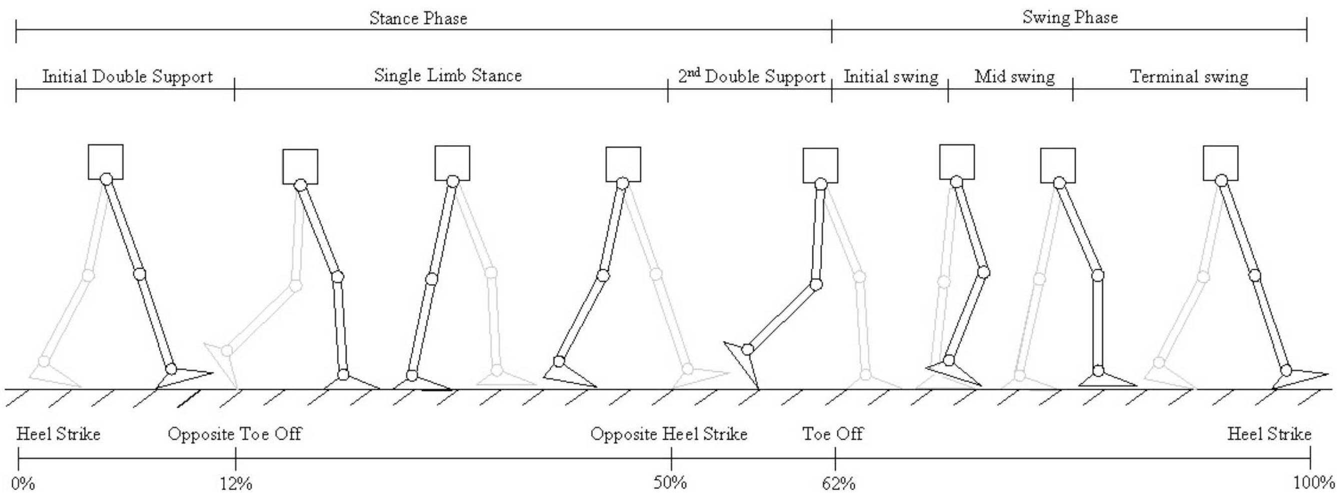

Fig. 1. Human walking gait through one cycle, beginning and ending at heel strike. Percentages showing contact events are given at their approximate location in the cycle. Adapted from [4].

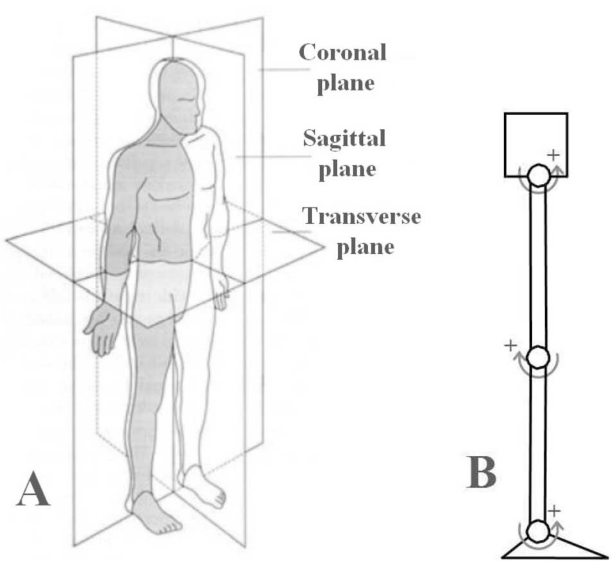

Fig. 2. (A) Description of the anatomical planes. (B) Diagram of the leg shown in the rest position (0 deg at all joints) with the positive direction indicated.

heel strike on the adjacent foot occurring at approximately 62% of gait cycle.

In general, the human leg can be thought of as a 7 DOF structure, with three rotational DOFs at the hip, one at the knee, and three at the ankle. Fig. 2 shows a description of the human anatomical planes [Fig. 2(A)] 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. 2(B)]. In this review, joint motion in this plane is simply referred to as flexion (positive direction) and extension (negative direction). 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 DOFs of the hip and ankle are simply referred to as “rotation.” These various terms are used throughout this paper in describing the kinematic layout of the various exoskeleton and orthosis designs.

Fig. 3 shows biomechanical measures from a normal, healthy individual (82 kg, 0.99 m leg-length, 28-year-old male) walking at 1.27 m/s, showing joint angle, moment, 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 [5]. While walking data may differ somewhat across subject and condition, the qualitative nature of the data remains similar [6]–[9].

It is particularly useful for the understanding exoskeleton and active orthosis design to note the power requirements of each joint. From walking gait data, it can be seen that, particularly at slow 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 evenly split between positive and negative. Note that, during steady-state level ground walking, the net mechanical 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 energetics of ankle, knee, and hip during slow walking, 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 and releasing energy at the ankle using passive elastic structures. However, when the subject walks at moderate to fast speeds, or on a positive incline or ascending stairs, the nature of the power at the individual joints can change dramatically. For this reason, many devices enable the power to also be added at the knee and sometimes the ankle.

A. Metabolic Cost

One key performance measure in demonstrating the effectiveness of a performance-augmenting leg exoskeleton is the metabolic cost required to walk or run. By measuring the rates of oxygen consumption and carbon dioxide production of a subject during a specific task, a measure of how physically taxing the activity is to the subject can be gotten [10], [11]. A number of inexpensive, compact systems for measuring

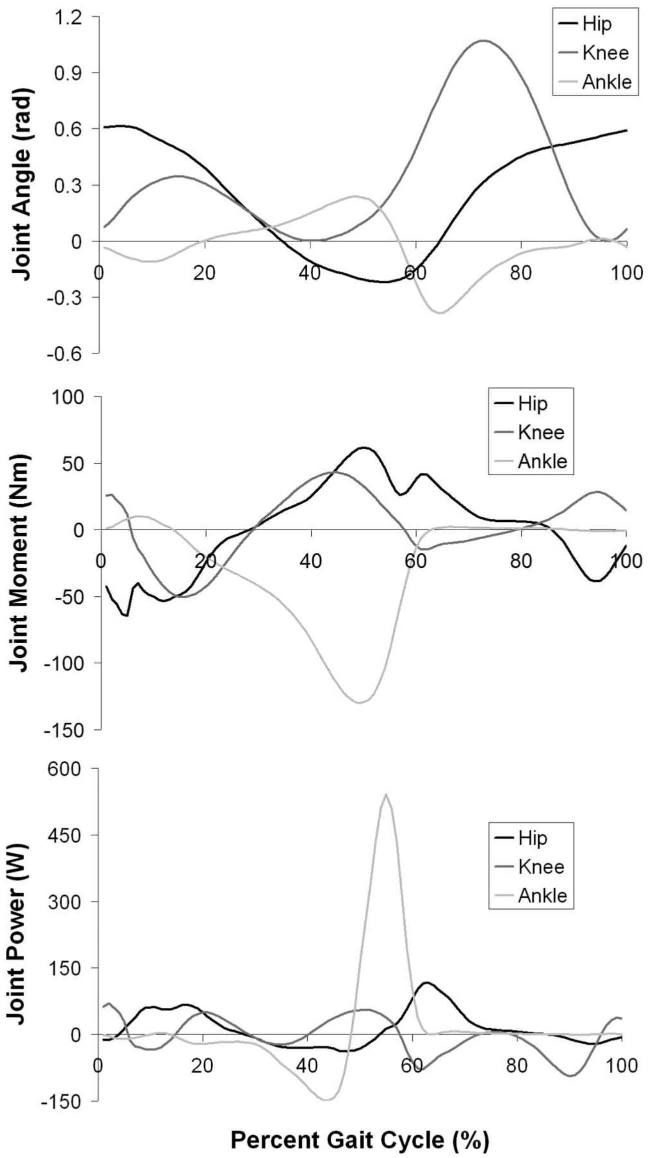

Fig. 3. Representative angles, moments, and power of the leg flexion/extension joints over the gait cycle, beginning and ending at heel strike. Data are average curves from seven walking trials adapted from [5].

these metabolic parameters are commercially available [e.g., the K4 telemetric system (Cosmed srl, Rome, Italy) [12]]. A comparison of metabolic power between performing the task with and without the exoskeleton or active orthosis is a good determinant as to whether there is any energetic advantage of using the device versus other means of performing the task.

III. Performance-Augmenting Exoskeletons

In this section, we describe the research done in developing exoskeletons primarily intended to allow otherwise healthy individuals to perform difficult tasks more easily or enable them to perform tasks that are otherwise impossible using purely human strength or skill (see Fig. 4).

A. Early Exoskeletons

Most of the early work done in developing performance-augmenting exoskeletons were concept studies that never left the drawing board. The few prototypes of these early concepts that were actually built and tested performed poorly.

To our knowledge, the earliest mention of a device resembling an exoskeleton is a set of U.S. Patents granted in 1890 to Yagn [13]. His invention consisted of long bow/leaf springs operating in parallel to the legs and was intended to augment running and jumping. Each leg spring was engaged during the foot contact to effectively transfer the body's weight to the ground and to reduce the forces borne by the stance leg. During the aerial phase, the parallel leg spring was designed to disengage in order to allow the biological leg to freely flex and to enable the foot to clear the ground. Although Yagn's mechanism was proposed to augment running, to our knowledge, the device was never built or successfully demonstrated.

In 1963, Zaroodny of the U.S. Army Exterior Ballistics Laboratory published a technical report detailing his work on a "powered orthopedic supplement" begun in 1951 [14] (note that this work was reportedly started before the publication of [1]). His exoskeleton device was intended to augment the load-carrying abilities of an able-bodied wearer such as a soldier. While mainly a concept paper, Zaroodny identified and began to address many of the fundamentally difficult aspects of implementing such a device, such as a portable power supply, sensing and control, physical interface with the human, and the affectation of the biomechanics of locomotion. The paper ends by describing the results of an informal evaluation of a pneumatically powered prototype device—possibly the first powered performance-augmenting exoskeleton ever created. The 3 DOF device consisted of a large, pneumatic cylinder affixed to a saddle (via a pivot at the hip) and terminating under the toes at the sole of a specially designed shoe. While his proposal did not succeed in securing funding to pursue the project, this report is nonetheless the earliest publication in which the complications of engineering a performance-augmenting exoskeleton device were considered.

In the late 1960s, General Electric Research (Schenectady, NY), in cooperation with researchers at Cornell University and with financial support from the U.S. Office of Naval Research, constructed a full-body powered exoskeleton prototype [15]–[18]. Dubbed "Hardiman" (from the "Human Augmentation Research and Development Investigation"), the exoskeleton, was an enormous hydraulically powered machine (680 kg, 30 DOFs), including components for amplifying the strength of the arms (including hands but without wrists) and legs of the wearer. In comparison to many other augmenting exoskeletons, the intention of the Hardiman project was to drastically increase the strength capabilities of the wearer (approximately 25:1). A patent filed in 1966 describes what was presumably the initial Hardiman concept [19], and was much more sleek and compact than what was eventually constructed.

While satisfactory results were achieved with the arm-amplifying aspects of the prototype, problems with the lower limb components were never resolved and the full-bodied device

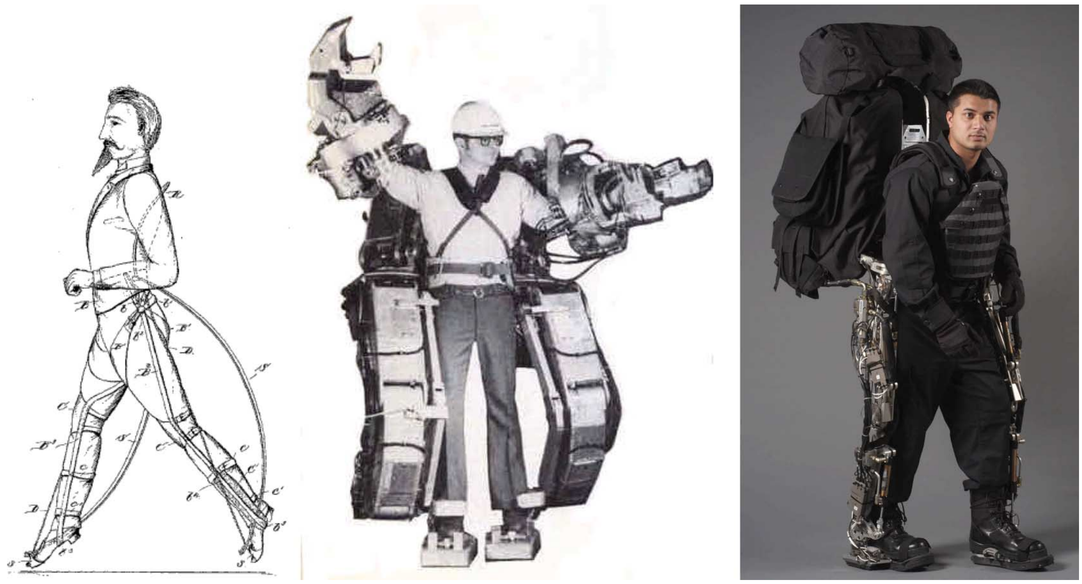

Fig. 4. Yagn's running aid [13], General Electric's Hardiman [18], and the University of California at Berkeley's BLEEX exoskeleton [23] (image credit Prof. Homayoon Kazerooni).

was reportedly never even powered up with a human inside. Perhaps the most important contribution of the Hardiman project was identifying many of the most challenging aspects of the exoskeleton design such as power supply and human/machine interface as well as convincing the research community that the creation of effective exoskeleton devices is extremely difficult.

In the mid 1980s, Jeffrey Moore at the Los Alamos National Laboratory (Los Alamos, NM) wrote a paper on an exoskeleton concept for augmenting the capability of soldiers inspired by Heinlein's concept that he deemed "Pitman" [1], [20]. While the paper did not address how problems such as power supply and implementation were going to be solved, and the concept never left the drawing board due to a failure to secure funding, it undoubtedly planted one of the seeds that grew into the U.S. Defense Advanced Research Projects Agency (DARPA) exoskeleton program a decade later (Section III-B).

An independent researcher named Mark Rosheim expanded on the Hardiman and Pitman concepts in a concept paper by incorporating singularity-free pitch–yaw type joints in order to present a full-body, 26 DOF exoskeleton concept (excluding the hands) [21].

B. DARPA Program Exoskeletons

A major impetus for the recent work in performance-augmenting exoskeletons has come from a program sponsored by the DARPA called Exoskeletons for Human Performance Augmentation (EHPA). The goal of the program is to "increase the capabilities of ground soldiers beyond that of a human" [22]. In particular, the program focuses on augmenting the performance of soldiers during load-carrying, increasing the size of the load that can be carried, and reducing the fatigue on the soldier during the load-carrying task. An in-depth description of the goals of the program as well as the initial role of each of the contractors is provided in [22]. The program began in 2001 and will be transitioned to the Army Program Executive Office Soldier (PEO Soldier) by 2008.

Over the duration of the EHPA program, three institutions demonstrated working exoskeletons, and a number of other institutions made advances in "enabling" technologies such as portable power supplies.

1) Berkeley Exoskeleton (BLEEX): The most visible of the DARPA program exoskeletons has been the Berkeley Lower Extremity Exoskeleton (BLEEX). One of the distinguishing features of this project is that it is energetically autonomous, or carries its own power source. Indeed, its developers claim it as the first "load-bearing and energetically autonomous" exoskeleton [23].

BLEEX features 3 DOFs at the hip, 1 at the knee, and 3 at the ankle. Of these, four are actuated: hip flexion/extension, hip abduction/adduction, knee flexion/extension, and ankle flexion/extension. Of the unactuated joints, the ankle inversion/eversion and hip rotation joints are spring-loaded, and the ankle rotation joint is free-spinning [24]. The kinematics and actuation requirements of the exoskeleton were designed by assuming behavior similar to that of a 75 kg human and utilizing clinical gait analysis data for walking [24], [25].

Interesting features of the kinematic design of the exoskeleton include a hip "rotation" joint that is shared between the two legs of the exoskeleton, and therefore, does not intersect with the wearer's hip joints. Similarly, the inversion/eversion joint at the ankle is not colocated with the human joint, but is set to the lateral side of the foot for simplicity. The other five rotational DOFs of the exoskeleton coincide with the joints of the wearer [24].

The exoskeleton is actuated via bidirectional linear hydraulic cylinders mounted in a triangular configuration with the rotary joints, resulting in an effective moment arm that varies with joint angle. BLEEX consumes an average of 1143 W of hydraulic power during level-ground walking, as well as 200 W of electrical power for the electronics and control. In contrast, a similarly sized, 75 kg human consumes approximately 165 W of metabolic power during level-ground walking [24], [25].

BLEEX was designed with linear hydraulic actuators since they were the "smallest actuation option available" based on their "high specific power (ratio of actuator power to actuator weight)" [24]. However, a further study determined that electric motor actuation significantly decreased power consumption during level walking in comparison to hydraulic actuation [26]. The weight of the implementation of the electrically actuated joint, however, was approximately twice that of their hydraulically actuated joint (4.1 kg vs. 2.1 kg).

The control scheme of the BLEEX seeks to minimize the use of sensory information from the human/exoskeleton interaction, and instead, utilizes mainly sensory information from the exoskeleton. Similarly to a bipedal robot, the exoskeleton can balance on its own, but the human wearer must provide a forward guiding force to direct the system during walking. The control system utilizes the information from 8 encoders and 16 linear accelerometers to determine angle, angular velocity, and angular acceleration of each of the eight actuated joints, a foot switch, and load distribution sensor per foot to determine ground contact and force distribution between the feet during double stance, eight single-axis force sensors for use in force control of each of the actuators, and an inclinometer to determine the orientation of the backpack with respect to gravity [24].

In order to achieve their goal of being energetically autonomous with such an actuator selection, significant effort was invested in developing a hybrid hydraulic–electric portable power supply [27].

In terms of performance, users wearing BLEEX can reportedly support a load of up to 75 kg while walking at 0.9 m/s, and can walk at speeds of up to 1.3 m/s without the load. A second generation of the Berkeley exoskeleton is currently in testing. The new device is approximately half the weight of the original exoskeleton ($\sim$14 kg [28]), in part due to the implementation of electric actuation with a hydraulic transmission system. A laboratory spin-off company called Berkeley Bionics (Berkeley, CA) has been formed in order to market the exoskeleton technology.

2) Sarcos Exoskeleton: The Sarcos Research Corporation (Salt Lake City, UT) has worked toward a full-body "Wearable Energetically Autonomous Robot (WEAR)" under the DARPA EHPA program. As the name suggests, the Sarcos exoskeleton is also energetically autonomous, carrying its own power supply. Similarly to the Berkeley exoskeleton, Sarcos has advanced a hydraulically actuated exoskeleton concept. However, instead of linear hydraulic actuators, the Sarcos exoskeleton employs rotary hydraulic actuators located directly on the powered joints of the device. Although Sarcos has not reported the power requirements of their exoskeleton, they have spent a significant amount of effort developing power supplies and servo–valves for efficient hydraulic actuation of the exoskeleton [28], [29].

The Sarcos exoskeleton utilizes force sensing between the robot and the wearer to implement a "get out of the way" control scheme. The wearer's foot interfaces with the exoskeleton via a stiff metal plate containing force sensing elements, and therefore, the wearer's feet are not allowed to bend.

The Sarcos exoskeleton has reportedly been successful in demonstrating a number of impressive feats: structure supporting entire load of 84 kg, wearer standing on one leg while carrying another person on their back, walking at 1.6 m/s while carrying 68 kg on the back and 23 kg on the arms, walking through 23 cm of mud, as well as twisting, squatting, and kneeling [28], [30].

After the DARPA EHPA program ended, Sarcos secured a large amount of additional funding through the Army PEO Soldier to continue the development of their exoskeleton concept as a personal combat vehicle (PCV), eventually "transitioning" the technology to the Army by fiscal year 2008.

Unfortunately, very little further information regarding the design and performance of the Sarcos exoskeleton has been made public.

3) MIT Exoskeleton: A quasi-passive exoskeleton concept has been advanced in the Biomechatronics Group at the Massachusetts Institute of Technology Media Laboratory under the second phase of the DARPA EHPA program. This concept seeks to exploit the passive dynamics of human walking in order to create lighter and more efficient exoskeleton devices.

The MIT exoskeleton employs a quasi-passive design that does not use any actuators for adding power at the joints. Instead, the design relies completely on the controlled release of energy stored in springs during the (negative power) phases of the walking gait [31]–[34]. The quasi-passive elements in the exoskeleton (springs and variable damper) were chosen based on an analysis of the kinetics and kinematics of human walking.

The 3 DOF hip employs a spring-loaded joint in the flexion/extension direction that stores energy during extension that is released during flexion. This spring mechanism is configured such that the user can freely swing their hip in the flexion direction. The hip abduction/adduction direction is also spring-loaded, but only to counter the moment induced by the backpack load. Additionally, a cam mechanism was incorporated at the hip to compensate for the relative change in length between the thigh of the exoskeleton and the user due to the joint offset during abduction/adduction. Additionally, spring-loaded hip rotation and ankle rotation joints were included to allow nonsagittal plane limb movements.

The knee of the MIT exoskeleton consists of a magnetorheological variable damper (motion in the flexion/extension direction) that is controlled to dissipate energy at appropriate levels throughout the gait cycle. For the ankle, separate springs for dorsi and plantar flexion are implemented in order to capture the different behaviors during these two stages of motion, and store/release the optimum amount of energy. The ankle also features a carbon fiber plate that attaches to the boot and doubles as a subtalar joint inversion/eversion spring. Additionally, there is a carbon fiber spring under the heel that reduces impact losses and aids in lifting the heel at the beginning of the powered plantar flexion. Finally, an artificial elastic spine attaches to the backpack that allows for coronal and sagittal plane human spine movements.

The quasi-passive exoskeleton is controlled simply by using sensory information provided by a set of full-bridge strain gages

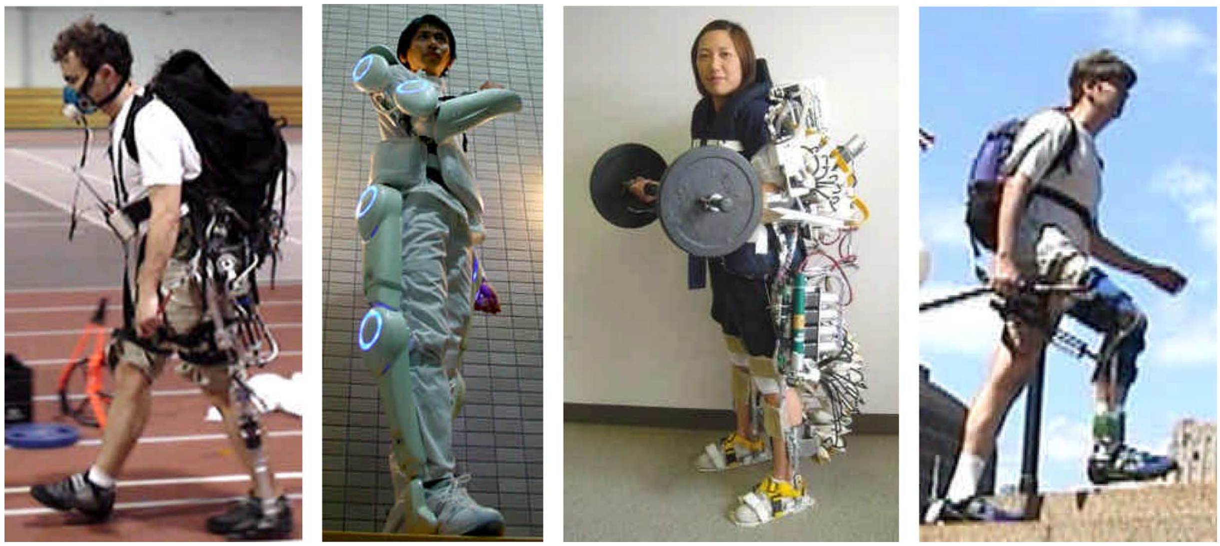

Fig. 5. MIT exoskeleton during metabolic testing [35], HAL-5 exoskeleton [41], nurse-assisting exoskeleton [43], and the RoboKnee [44]. Image credits (from left to right): Prof. Hugh Herr, Biomechatronics Laboratory, MIT Media Lab, Cambridge, MA; Prof. Sankai, University of Tsukuba/CYBERDYNE, Inc., Tsukuba, Japan; Prof. Keijiro Yamamoto, Kanagawa Institute of Technology, Atsugi, Japan; Dr. Jerry Pratt, Institute for Human and Machine Cognition, Pensacola, FL.

on the exoskeleton shin and a potentiometer on the knee joint. The MIT exoskeleton interfaces with the wearer via shoulder straps, a waist belt, thigh cuffs, and specialized shoes. Without a payload, the exoskeleton weighs 11.7 kg and requires only 2 W of electrical power during loaded walking. This power is used mainly to control the variable damper at the knee.

Experimental work with this quasi-passive exoskeleton demonstrated a working device that successfully supported a 36 kg load during walking at 1 m/s. It was also shown that the exoskeleton structure transferred on average 80% of the 36 kg payload weight to the ground during the single-support phase of walking.

However, metabolic studies with the quasi-passive exoskeleton showed a 10% increase in walking metabolic cost of transport, or the metabolic energy required to transport unit weight unit distance, for a subject carrying the 36 kg load via the quasi-passive exoskeleton versus a standard laden backpack [33]–[35]. While this is an undesirable result, it is thought to be the first reported study on the metabolic cost associated with walking under the aid of an exoskeleton. In a separate study, the U.S. Army Natick Soldier Center showed that load-carriage using another quasi-passive exoskeleton design increased metabolic cost of transport on average across three tested loading conditions (20, 40, and 55 kg) by as much as 40% [36]. To our knowledge, no one has yet demonstrated an exoskeleton that reduces the metabolic cost of transport when compared to the load-carriage with a standard backpack.

Further experimental work with the MIT quasi-passive exoskeleton showed a significant reduction in metabolic cost of walking versus the same exoskeleton without the springs at the hip and ankle and the variable damper at the knee, demonstrating the utility of the quasi-passive elements. Additionally, tests were conducted to determine the effect of the added mass and the inertial load of the exoskeleton on the wearer. From these studies, it was concluded that, in addition to the added mass and inertia, a dominant cause for the observed cost of transport increase are the additional kinematic constraints inadvertently imposed on the wearer, upsetting the efficient dynamics normally seen during human walking [34], [35].

4) Enabling Technologies: While this review is limited in scope to fully realized exoskeleton hardware platforms, a number of significant "enabling technologies" were developed by contractors of the DARPA exoskeleton programs. Oak Ridge National Laboratory developed a foot force–torque sensor, control strategies, and power supply technology for exoskeleton applications [37]. Arthur D. Little (now TIAX, Cambridge, MA), Honeywell (Minneapolis, MN), Quoin (Ridgecrest, CA), and Locust USA, Inc. (Miami, FL), worked toward developing specialized power systems to meet the requirements of the exoskeleton project [22], [38]. Boston Dynamics (Cambridge, MA) did predictive modeling, while Will Durfee at the University of Minnesota worked on ways to physically interface the wearer to the exoskeleton while minimizing the discomfort [22], [38]. Additionally, the Vanderbilt Center for Intelligent Mechatronics worked on applying their monopropellant-based power system to the Berkeley exoskeleton [39].

C. Other Lower Limb Exoskeletons

1) Hybrid Assistive Leg: At the University of Tsukuba, Japan, Prof. Yoshikuyi Sankai and his team have been developing an exoskeleton concept that is targeted for both performance-augmenting and rehabilitative purposes [40], [41]. The leg structure of the full-body hybrid assistive leg (HAL)-5 exoskeleton powers the flexion/extension joints at the hip and knee via a dc motor with harmonic drive placed directly on the joints (see Fig. 5). The ankle flexion/extension DOF is passive. The lower limb components interface with the wearer via a number of connections: a special shoe with ground reaction force sensors harnesses on the calf and thigh, and a large waist belt. Note that, in distinction to the load-carrying BLEEX, Sarcos, and MIT exoskeletons, the HAL system does not transfer a load to the ground surface, but simply augments joint torques at the hip, knee, and ankle.

The HAL-5 system utilizes a number of sensing modalities for control: skin-surface electromyographic (EMG) electrodes placed below the hip and above the knee on both the front and the back sides of the wearer's body, potentiometers for joint angle measurement, ground reaction force sensors, and a gyroscope and accelerometer mounted on the backpack for torso posture estimation. These sensing modalities are used in two control systems that together determine user intent and operate the suit: an EMG-based system and a walking-pattern-based system. Reportedly, it takes two months to optimally calibrate the exoskeleton for a specific user [28].

HAL-5 is currently in the process of being readied for commercialization. Modifications from previous versions include upper-body limbs, lighter and more compact power units, longer battery life (approximately 160 min continuous operating time), and a more cosmetic shell. The total weight of the full-body device is 21 kg. Cyberdyne (Tsukuba, Japan, www.cyberdyne.jp), a company spun off from Sankai's laboratory, is responsible for the commercialization of the product.

While there have been many demonstrations of the HAL being worn by an able-bodied operator, results of the performance of the exoskeleton on a physically challenged subject were not able to be found. Along these lines, the ability of the HAL to increase the user's performance in holding large loads in the arms have been shown; however, the effectiveness of the lower limb components of the exoskeleton are unclear. However, on the corporate Web site, the inventors claim that an operator wearing HAL can carry up to 40 kg on the arms and increase the user's "leg press" capability from 100 to 180 kg.

2) Nurse-Assisting Exoskeleton: For more than a decade, researchers at Kanagawa Institute of Technology in Japan have been developing an exoskeleton for the purpose of assisting nurses during patient transfer [42], [43]. The lower limb components of the suit include direct-drive pneumatic rotary actuators for the flexion/extension of the hips and knees. Air pressure is supplied from small air pumps mounted directly to each actuator, allowing the suit to be fully portable. The nature of the unactuated DOFs and methods of attachment to the operator are, however, unclear.

User intent is determined via "muscle hardness sensors" created by attaching force sensing resistors (FSRs) to the surface of the skin above a muscle (the rectus femoris for the knees) via an elastic band. As the knee is flexed and the muscle is contracted, the force on the FSR increases, which, along with the joint angle information from potentiometers, is used to determine the torque required at the joint.

One of the interesting aspects of the mechanical design of the Kanagawa full-bodied suit is that there is no mechanical component on the front of the wearer, allowing the nurse to have direct physical contact with the patient that he or she is carrying. This is an important property for ensuring the comfort and security of the patient.

3) RoboKnee: Yobotics, Inc. (Cincinnati, OH, www.yobotics.com), developed a simple exoskeleton for adding power at the knee to assist in stair climbing and squatting during load-carrying tasks [44]. The device consists of a linear series elastic actuator (SEA) connected to the upper and lower portions of a knee brace, just below the hip and on the calf, respectively. The intention of the device is to apply power to the knee joint while exhibiting a physically low-impedance interface to the wearer, allowing for greater control gains while remaining safe to the operator.

The control of RoboKnee utilizes the ground reaction force (in the vertical direction) and the center of pressure in the sagittal plane (front/back direction). This information, captured via two load cells within each pair of stiff-bottomed shoes worn by the operator, is used in a positive-feedback force amplification control scheme of the torque at the knee.

D. Related Work

There have also been a number of feasibility studies that have not yet led to complete exoskeleton devices [45]–[48]. An interesting paper presented in 1973 presents an idea of a circuit to capture the electric energy generated when the joints of an exoskeleton are passively reversed (when dc motors are used) [49]. This is perhaps the earliest mention of a method to harness the negative power done at an exoskeleton joint.

IV. 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 the energy stored in springs during appropriate phases of the gait (see Fig. 6).

In the United States alone, approximately 4.7 million people would benefit from an active lower limb orthosis due to the effects of stroke, 1 million postpolio, 400 000 due to multiple sclerosis, 200 000 due to spinal cord injury, and 100 000 due to cerebral palsy. In this section, we focus on the development of active orthotic devices to assist this population and others suffering from some leg pathology affecting their locomotion abilities.

A. Early Active Orthoses

As would be expected, early active orthoses 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 U.S. patent from 1935 [50]. 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 a 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 [51]. However, due to the

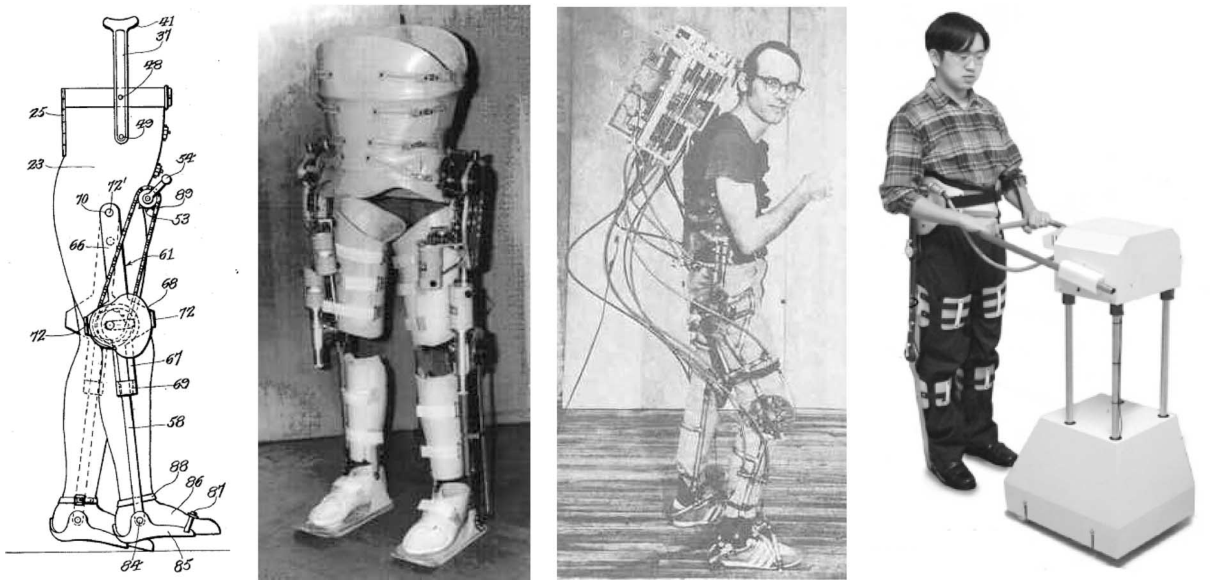

Fig. 6. Cobb's "wind-up" orthosis [50], Pupin Institute "complete" exoskeleton [54], Wisconsin exoskeleton [58], and Sogang orthosis and walker [60]. Image credits (from left to right): U.S. Patent 2 010 482; Prof. Dr. Miomir Vukobratović, Robotics Laboratory, Mihailo Pupin Institute, Belgrade, Serbia; Jack Grundmann, University of Wisconsin, Madison, WI; Kyoungchul Kong and Doyoung Jeon, Sogang University, Seoul, Korea.

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 [52].

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 [53]–[55]. 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 the 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 at 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. 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 held 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 the operation, all of the aforementioned exoskeleton devices 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 the 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 the work with exoskeletons at Pupin Institute is in control methods for robotic bipeds. Indeed, Prof. 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 [56].

A thorough history of the work done with exoskeletons at the Mihailo Pupin Institute is provided in [54]. 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 [57], [58]. Similar to the Pupin Institute exoskeletons, this device was intended to help reambulate paraplegics that have full upper-body capabilities. The kinematic design of the exoskeleton featured universal joints at the hip and ankle (three rotational DOFs 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 DOFs were either completely passive or spring-loaded.

The hydraulic power unit consisted 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 [58].

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 stabilization. The device was programmed to follow joint trajectory data recorded from a similarly sized able-bodied individual in a feedforward, 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 reportedly 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, thus easing the effort required for locomotion [59]. 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 young adults and one 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%. This team of researchers continues to be active in research in this area, and more results are expected in the months following the publication of this review.

An interesting concept meant to alleviate some of the difficulties in creating a portable active orthosis device is presented by researchers at Sogang University, Seoul, Korea [60]. 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 accompanying orthosis. A cable drive transmits mechanical power to the joints of the wearer from the actuators in the walker. Due to this 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 DOFs of the leg, except the rotation of the ankle, which is fixed. User intent is sensed by a combination of joint angle sensors and a pressure sensor that gives a sense of force being applied by the quadriceps muscle.

Another interesting aspect of this design 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. Since most powered orthotic devices still require the use of crutches or another additional support method for the user, this concept is especially promising.

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 [61]. 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, that this strategy may lead to problems with the stability of the wearer.

Researchers at Michigan Tech developed an experimental powered gait orthosis consisting of 1 DOF per leg with actuated hip and knee joints connected by linkages [62]. The device was used to study the power required for a fully actuated device, as well as to determine the amount of force required by the device to support the operator during gait.

Darwin Caldwell, who has been active in upper limb exoskeleton research, has also developed a 10 DOF lower limb exoskeleton device [63]. 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 their own orthosis design that is powered by a custom-designed bilateral hydraulic servo actuator [64]. 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 have not yet progressed past the stage of preliminary investigations [65], [66]. A concept in which the orthosis is controlled via sensed motions of the user's fingers is presented in [67]. 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 [68].

C. Modular Active Orthoses

1) AMOLL Project: The first published work with modular active orthoses is the Active Modular Orthosis for Lower Limbs project (AMOLL, headed by Pierre Rabischong), which incorporated researchers from Montpellier and Toulouse, France, the University of Belgrade, and Stanford Research Institute [69]. The concept advanced the idea of an inflatable interface with the wearer, a concept first introduced by the French company Aerozur as "soft suits" [70]. The modular nature of these devices allowed that 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 DOFs at the hip were to be stiffened by rigidity in the orthosis. Actuation was not

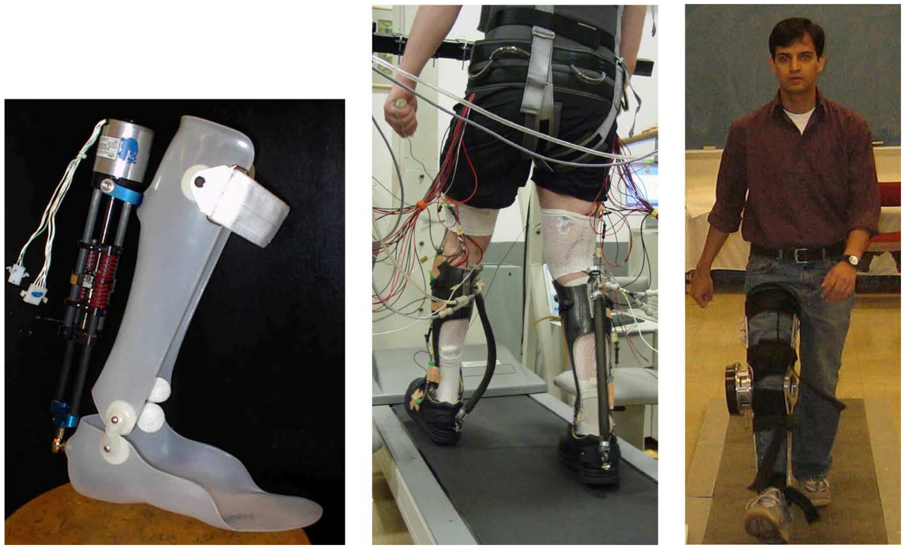

Fig. 7. MIT active AFO [79], Michigan ankle orthoses [82], and the Northeastern University knee orthosis [87]. Image credits (from left to right): Prof. Hugh Herr, Biomechatronics Laboratory, MIT Media Lab, Cambridge, MA; Prof. Dan Ferris, Human Neuromechanics Laboratory, Division of Kinesiology, University of Michigan, Ann Arbor, MI; Prof. Constantinos Mavroidis, Robotics and Mechatronics Laboratory, Northeastern University, Boston, MA.

yet implemented in this initial paper; however, dc servo–motors were proposed. A method of control is proposed in [71].

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 [72]. The work focused 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) University of Belgrade and Zotovic Rehabilitation Institute: 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 of the original AMOLL project paper [70]), 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" [70], [73], devices similar to the inflatable components mentioned under the original AMOLL proposal [69].

Popovic and Schwirtlich's work with these modular devices quickly advanced to developing the first hybrid system combining a powered orthosis with the FES [74]–[76]. This system was intended to extend the use of the 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 last version of their full exoskeleton (Section IV-B1), this "active suit" was actuated via dc motors and was not energetically autonomous at the prototype phase [54], [77].

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 [78]. The device consisted of a dc motor mounted in front of the wearer's shin that assisted in the flexion/extension of the ankle. Also included was a specially designed "junction" that allowed free movement of the ankle. The orthosis was controlled based upon the information from foot switches in the soles.

a) MIT Ankle–Foot Orthosis: The MIT Biomechatronics Group developed a powered AFO to assist dropfoot gait, a deficit affecting many persons who have experienced a stroke, or suffer from multiple sclerosis or cerebral palsy, among others (see Fig. 7) [79]. The device consists of a modified passive AFO with the addition of an SEA to allow for the variation in the impedance of flexion/extension direction of ankle motion, controlled based on ground force and angle position data. Using the SEA, the device varies the impedance of the ankle in plantar flexion during stance, and assists with dorsi flexion during the swing phase of walking.

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," creating better symmetry with the unaffected leg, and providing assistance during powered plantar flexion. Feedback from the subjects was also extremely positive. The device is relatively compact and low-power (10 W average electrical power consumption), and current work is focusing on developing an energetically autonomous, portable version of the device.

b) University of Michigan Orthoses: The Human Neuromechanics Laboratory at the University of Michigan has produced a number of active orthoses, particularly focusing on rehabilitation devices to be used during therapy [80]–[82]. 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) that are mounted to carbon fiber and polypropylene shells, resulting in devices that are extremely lightweight as well as exhibiting high-power outputs. Additionally, the low impedance of the actuators produces safer devices.

The University of Michigan orthoses are primarily designed for the lower leg, with both ankle–foot and knee–ankle–foot devices having been developed. For all devices, carbon fiber and polypropylene shells are custom-built for each subject, eliminating the need for mechanically complex adjustment mechanisms. However, the custom-built nature of these devices has negatives for clinical applications, since a separate device needs to be built for each patient, with a separate fitting visit needed before therapy can commence.

The Human Neuromechanics Laboratory has built AFOs including an agonist/antagonist actuator pair as well as a single plantar flexion actuator [in the positive direction according to Fig. 2(B)]. 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. The results showed that, while providing increased plantar flexion at the end of the stance phase, the AFO 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 AFO has also been developed, and incorporates an additional agonist/antagonist pair of artificial muscles for the flexion/extension of the knee [82].

c) Other Ankle–Foot Orthoses: At Arizona State University, researchers have presented a novel design of an active AFO 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 [83]. These actuators are essentially pneumatic muscles with an internal spring tending to extend the muscle, enabling force to be applied in both plantar and dorsiflexion directions. The tripod configuration allows the ankle to be actuated in flexion/extension (coactivation) and inversion/eversion (single activation). Additionally, the group has also explored using SEAs to power orthosis joints [84].

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 [85].

2) Active Knee Orthoses: Dinos Mavroidis' Laboratory at the Northeastern University has developed a dissipative knee orthosis by combining an electro-rheological fluid-based variable damper with a modified commercial knee brace. 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 ability of the average human during level ground ambulation [86], [87].

Researchers at Berlin University of Technology are developing an orthosis to add power at the knee via a dc motor and ball-screw transmission [88], [89]. However, work up to this point has been focused primarily on developing an EMG-based control system for the device, to be implemented along 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 [90].

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 pendular, straight-legged method of 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 [91]. In this way, the "ground" is effectively raised and lowered in order to compensate for the lack of DOF 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 traditional RGOs were used [92].

An RGO was modified to include actuation at the hip and knees by researchers in Torino, Italy [93]. 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 [94].

A number of researchers have investigated combinations of RGOs and FES [95]–[101]. 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 orthosis incorporating controllable brakes at the hips and knees (flexion/extension) with a method of 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 the FES to be used much more frequently (shorter duty cycle), and it also reduces muscle fatigue [102], [103].

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, as compared to 85% for FES alone.

V. Discussion

In the process of doing this review, a number of themes related to the challenges associated with building functional, autonomous exoskeleton, and 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 exoskeleton 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 [23], [28], [54], [58]. Indeed, this is a problem facing many fields of mobile robotics, particularly those with anthropomorphic architectures.

While these issues continue to be addressed, a number of great advances have been made in the areas related to exoskeletons and active orthoses in the last five decades. The field of biomechanics of human locomotion has matured in recent decades, providing 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, though they still do not meet the demands of many exoskeleton 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.

A. Performance Augmenting Exoskeletons

To this point, the reported advantages of complete, autonomous exoskeleton systems are largely anecdotal. Indeed, there is a marked lack of published quantitative performance results for exoskeleton devices that reportedly improve human locomotion. The few exceptions [34]–[36] give results that do not confirm any benefit of current designs. Considering this, one is left to wonder what the real advantages of these complicated, expensive systems really are. Certainly, there is value in an exoskeleton that enables the wearer to perform a task that he or she could not otherwise perform [23], [28]. However, if exoskeletons intended to facilitate tasks that could otherwise still be performed by the wearer (e.g., load carrying) do not reduce the metabolic cost and/or fatigue of the operator, they have very little value [104]–[109]. Besides locomotory performance as assessed by metabolic cost evaluations, other performance measures that would be appropriate for these types of systems include the reduction of forces borne by the musculoskeletal system, the reduction of muscle fatigue, and the improvement of bipedal stability.

Rather than minimizing the accomplishments that have been made in the field, the lack of quantitative results with exoskeletons instead highlights the numerous challenges associated with creating them. There are, of course, many design challenges that may lead to poor exoskeleton performance: misalignment of joints between operator and hardware, kinematic constraints from attachments such as harnesses and cuffs, design not optimized for load-carrying gait [104]– [109], added forces to the operator that resist motion, and addition of power in a suboptimal manner (e.g., mistiming, too little, too much), among others. All of these problems are very difficult to address, however, and there is much opportunity for fundamental studies addressing these challenges.

B. Active Orthoses

Besides sharing many of the challenges facing performance-enhancing exoskeletons, active orthoses 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 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 [3]; however, there is great value in developing a portable device that can be used outside of the clinic. 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 that limits the application of active orthoses outside of clinical 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.

As with performance-augmenting exoskeletons, there is a lack of published quantitative results on the effectiveness of active orthoses. Comparison with established assistive devices is a logical avenue for these devices. For instance, an active orthosis, meant to assist ambulation in someone who might otherwise be able to ambulate using an RGO, should be tested against results with that device. Appropriate performance measures include the metabolic cost of transport [110], [111], walking speed, smoothness and repeatability of motions, muscle fatigue, and stability, among others.

C. Future Work

Future directions in work related to the creation of exoskeletons and 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 in exoskeleton technology are currently being driven by the exoskeleton research community itself, and not by other, more pervasive applications such as those that drove developments in computing, sensing, and control.

There are a few areas related to the mechanical design of exoskeletons 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 on more effective exoskeleton leg architectures. Gait models based on actual machine elements that capture the major features of human locomotion [112] may enhance the understanding of human leg morphology and control, and lead to analogous improvements in the design of efficient, low-mass exoskeletons.

The investigation of nonanthropomorphic 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 the wearer. Also, there has been little work with "recreational" exoskeletons such as those that augment running or jumping ability, and this area is likely to be a focus in the future.

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 exoskeleton or active orthosis to the human body both mechanically and neurally are important areas for future research.

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