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Stronger, Smarter, *Softer*: Next Generation Wearable Robots

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I. Introduction

Humans have long dreamed of and created ways of improving our strength, speed, and endurance through wearable assistive devices. Science fiction authors have inspired our imaginations with remarkable exoskeleton devices like those worn in Robert Heinlein's Starship Troopers and that worn by Marvel Comics' Iron Man, and many researchers have spent countless hours and resources toward making these machines a reality.

One particular area of interest is that of devices to assist the lower body for tasks such as walking, running, and supporting heavy loads. The vast majority of these are rigid exoskeletons, with links in parallel with the body that can impart torques to the joints, support compressive forces, and in many cases transmit loads to the ground.

Some exoskeletons have enabled individuals to walk if they were not able to previously, supporting their entire body weight or a large percentage of it [1]–[3], while others are designed to help able-bodied individuals walk while expending less energy [4]–[9], assist impaired individuals [10], or characterize the impedance of a wearer's joints [11]. Other approaches have been to assist with load carriage by providing a parallel path to ground, thereby offloading the wearer's musculature [12]–[14], and some systems also provide gait rehabilitation in conjunction with a treadmill [6], [15], [16]. Each of these systems' operation has been possible through a large number of clever and innovative design features and control schemes.

Nevertheless, exoskeletons still present a number of ongoing challenges, including: 1) rigid links with pin joints resist the movement of the biological joints if they are not perfectly aligned [17], and 2) exoskeletons may require bulky self-aligning mechanisms [17]–[19]. Rigid systems also have the problem of having large inertia; in particular, adding mass to the legs distally increases the metabolic cost of accelerating and decelerating them (8%/kg for mass at the feet vs. 1-2%/kg for mass at the waist) [20]. Due to these effects, wearing such devices often disrupts the natural biomechanics of walking, leading to discomfort or increased metabolic expenditure.

For scenarios in which an assistive device would be worn for extended periods of time, such as endurance augmentation, load-carriage, or potentially medical applications, avoiding increased metabolic expenditure is especially important. A few devices have been able to reduce the metabolic cost of certain activities, including tethered walking [7], [21], untethered walking with load [22], or stationary activities such as squatting [23] and hopping [24].

Fig. 1. Photos of two soft exosuits developed by our lab. The left suit is an early pneumatically-powered design that controls each of the joints in the leg in both directions in the sagittal plane. The right suit is our latest multi-articular design aiding ankle plantarflexion and hip flexion, and is actuated by geared motors driving Bowden cables.

Our long-term goal is to create a portable wearable robot that assists the wearer during walking and can reduce their metabolic expenditure compared to regular walking. To work toward these goals, we have proposed a new paradigm in assistive device design which we call soft clothing-like "exosuits" [25], [26]. These are devices that use textiles to interface to the body, and apply joint torques via tensile forces over the outside of the body in parallel with the muscles, utilizing the bone structure to support compressive loads. Previous research at Harvard focused on the exciting approach of designing

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soft wearable robots that could use actuators and sensors that were sufficiently compliant so as to not restrict movement [27]–[29]. In addition, work at Chuo University proposed a pneumatically-powered orthosis that used low forces to assist hip flexion and encourage longer steps during walking [30]. Compared to these prior approaches, we are focusing on systems intended to assist with forward propulsion during walking. A significant challenge with this approach is ensuring that the exosuits we describe have sufficient bandwidth and force generating capability to apply biologically relevant torques to the joints of the wearer during walking.

In comparison with rigid exoskeleton devices, exosuits have a number of advantages: they can be very light and have extremely low inertias, which reduces the metabolic cost of wearing them; they intrinsically transmit moments through the biological joints, since they can only apply tensile forces; and they are low-profile and can be worn underneath regular clothing, so the wearer can either blend in with normal society or can take advantage of protective outerwear. Since they are composed of textiles, they are easy to put on and take off, and can adapt easily to anatomical variations. A key feature of exosuits is that if the actuated segments are extended, the suit length can increase so that the entire suit is slack, at which point wearing an exosuit feels like wearing a pair of pants and does not restrict the wearer whatsoever. An effective exosuit for gait augmentation meets three requirements: (1) it leaves the user in full control over his/her own gait; (2) it introduces minor to no kinematic changes to natural gait; and (3) it assists the lower body during walking. Figure 1 shows two examples of exosuits designed by our lab, including an early pneumatically-powered exosuit and a more recent electromechanically-driven exosuit. Exosuits do have a few drawbacks however, including being able to transmit lower maximum forces than rigid-frame devices, not supporting compressive loads, and presenting challenging requirements for sensing and actuation. A summary of the differences between rigid exoskeletons and exosuits is in Table I.

In this paper, we describe the anatomy of a soft exosuit, showing examples of exosuits our lab has built over the last couple of years. Each component of an exosuit has presented unique challenges and design opportunities we have had to overcome in order to achieve a practical and useful device. We conclude by presenting metrics for exosuit evaluation and some initial results from the effect our exosuits have on the body.

TABLE I
COMPARISON OF THE KEY FEATURES OF RIGID EXOSKELETONS VS. EXOSUITS

FEATURE	RIGID EXOSKELETONS	EXOSUITS
Construction of leg components	Metal, plastic, etc.	Textiles
Mode of operation	Torques– tension and compression forces	Tensile forces only
Joint alignment, system adjustability	Alignment and adjustability are difficult, or require complex mechanisms	Alignment and adjustability are easily achieved
Bulkiness, inertia	Can be bulky and high inertia, requiring energy to move	Very low profile and low inertia
Bandwidth	Very high due to rigid frame	Low-medium due to compliant suit and human interface
Maximum torques	1-10x the nominal biological torques	0.1-1x the nominal biological torques
Effect on gait	Usually alter normal walking kinematics	Little to no effect on kinematics

II. STRUCTURED FUNCTIONAL TEXTILES

An exosuit consists of an integrated garment that includes attachment points to the body, a structured textile that transmits loads across the body, and actuated segments that can reduce their relative length to provide controlled tensile forces in the suit. The suit creates moments around the joints as these forces are offset from the joint centers of rotation due to the tissue and bone structure surrounding the joints.

A. Exosuit Architecture

In order to obtain high-performance soft exosuits, some considerations should be taken into account in the design process. Exosuits should attach to the body securely and comfortably, and transmit forces over the body through beneficial paths such that biologically-appropriate moments are created at the joints. Our lab initially developed the concept of virtual anchor points as a method of describing how an exosuit could be designed. In [25], we define "key anchors" as those parts of the body which are good at supporting loads and have high stiffness, such as the foot and pelvis. The suit must connect the ends of the actuators to the key anchors in order to support the high forces from the actuators. We do this by creating a matrix of connectors along lines of non-extension [31] to minimize motion of the suit. We denote the points at which the actuated segments attach to the connector matrix as "virtual anchors."

This principle is illustrated in Figure 2 at the top left. We presented the first embodiment of this technology in [25], which is shown on the left side of Figure 1. This exosuit used pneumatic actuators to actuate the joints in the sagittal plane. We call this a "mono-articular" architecture, since each actuator assists only a single joint in one direction.

When designing the webbing, care must be taken to route the intermediate connector matrix so it does not apply detrimental moments to other joints. For example, in a mono-articular suit the connectors going from the top of the ankle plantarflexion actuator to the waist must pass through the centers of the knee and hip joints, and be carefully routed in between, to avoid applying moments to those joints when the ankle is actuated.

A second possibility is to create a multi-articular suit that intentionally routes the forces between the actuators and key anchors so they create beneficial moments on the intervening joints [26]. Similar to bi- or multi-articular muscles such as the hamstrings (which cross the hip and knee) or gastrocnemius (which crosses the ankle and knee), a multi-articular suit can efficiently aid specific motions or transfer energy between joints [32]. In this case, the suit must be routed over the appro-

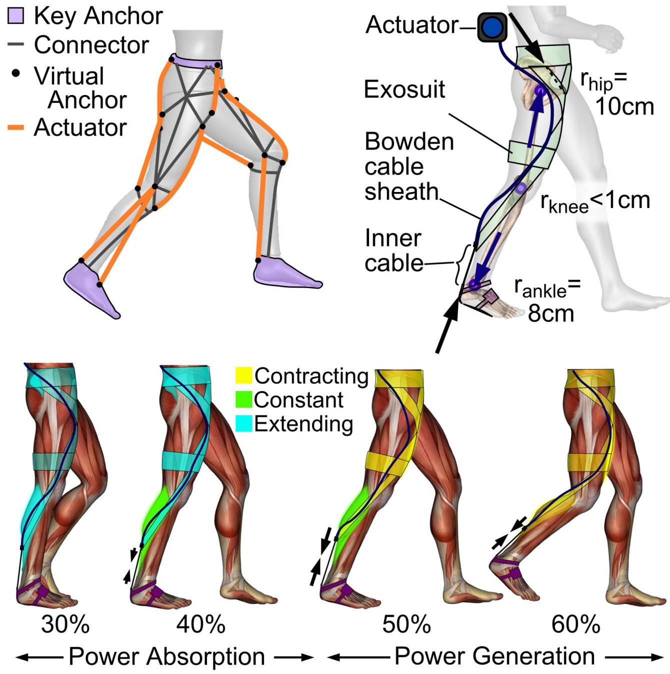

Fig. 2. The top left diagram illustrates the concept of virtual anchors for connecting actuators to a mono-articular exosuit. The top right figure shows an overview of a multiarticular suit, showing the moment arms at each joint and the reaction forces at the hip and ankle. The bottom row sequence shows the body position, active muscles, and multi-articular suit behavior during several points in the gait cycle. Colors indicate if the suit and muscles are contracting, constant in length, or extending.

priate side of each joint, so the desired moments are generated when tensile forces are induced in the suit. The timing of the induced moments at each joint is necessarily simultaneous, although the moment magnitudes and power transmitted may be different due to varying moment arms and differing joint velocities, respectively. Based on this biological inspiration, we created a multi-articular exosuit (Figure 1 (right)) that passes over the front of the hip, creating flexion moments, and behind the ankle, creating plantarflexion moments.

The suit intentionally passes close to the center of rotation of the knee to generate a negligible moment there. This suit is illustrated in the top right of Figure 2, which shows how the suit is actuated by a Bowden cable. The suit connects primarily to the body at the heel and iliac crest of the pelvis, and distributes tensile forces through various paths between the two locations. The top right illustration also shows the forces in the sagittal plane: the black arrows are the forces on the body due to the suit behavior, and the blue arrows are the reaction forces at the centers of the joints which are supported by the bone structure.

With both mono- and multi-articular exosuits, the moments on the body must be applied in a manner synergistic with the underlying muscles. With a mono-articular design, this is easy because the joints are independent. With a multi-articular design, the exosuit will be beneficial only for motions in which the moments at each joint are simultaneous, and it should be made slack in other situations.

The bottom half of Figure 2 shows how our multi-articular exosuit applies moments at the hip and ankle simultaneously with the underlying muscles during 30-60% in the gait cycle, which extends from one heel strike to the next for a given leg. During this stage of the gait, the calf muscles and tendons push the body up and forward, and the hip muscles and ligaments swing the leg forward. Initially, the calf and hip absorb power by stretching as the body's center of mass falls downward and forward over the planted foot. After around 50% in the gait cycle, this absorbed power is returned to the body as the tendons and ligaments elastically recoil. The muscles in the calf and hip actively contract to supplement this returned power with additional energy. Our exosuit absorbs and transmits power in this manner as well: with the actuators held at a fixed length initially, the exosuit material itself stretches and the tissue under the suit compresses as the body falls forward. This induces a tension in the suit and absorbs power from the body. Thus the multi-articular exosuit architecture has the unique property in that the exosuit only becomes tense when the body is in the correct position for forces to be applied. After the period of power absorption, the suit retracts elastically, returning the energy to the body. This is supplemented by the actuators contracting starting at 40% in the gait cycle to propel the body upwards and forwards.

B. *Structured Textiles for Load Distribution*

In addition to the architecture of the exosuit transferring forces over the body effectively, the suit itself must be comfortable and have high axial stiffness. We accomplish this with structured textiles made from specially designed patterns and materials. As a concrete example, we consider the design of the waist attachment in the multi-articular exosuit in Figure 2, which is also shown in Figure 3(a) and (b).

Figure 3(a) shows the front view of the waist attachment, with lines showing forces within the garment. The exosuit is designed to distribute forces from a node on the crease of the hip (shown with a circle) up to both sides of the waist. On the opposite side of the body, the forces are delivered to the top of the iliac crest of the pelvis; on the same side, forces follow two paths both above and below the iliac crest for improved load distribution. For the suit to be comfortable, the forces must be distributed as evenly as possible over the body to avoid points of high pressure which may cause discomfort or restrict blood flow [33], [34].

To achieve this pattern of load distribution, we use the suit layout shown in Figure 3(b). The waist attachment is composed of three different textiles, layered and oriented in different directions. The majority of the fabric is a plain weave nylon, chosen due to its high dimensional stability (it holds its shape) and its higher stiffness in extension as compared to other fabrics. This fabric, like all woven fabrics, has threads in two perpendicular directions, the warp and the weft. The fabric is strongest and stiffest in these directions (principal fabric axes) since along them the threads are pulled lengthwise. In a direction 45° from either of these axes, the fabric is less stiff since the weave structure of the fabric must support forces instead of just the thread. The relative strains of the fabric in different directions are shown in Figure 3(c), which is the result of evaluating the mechanical properties of swaths of fabric 5cm wide in an Instron mechanical testing machine.

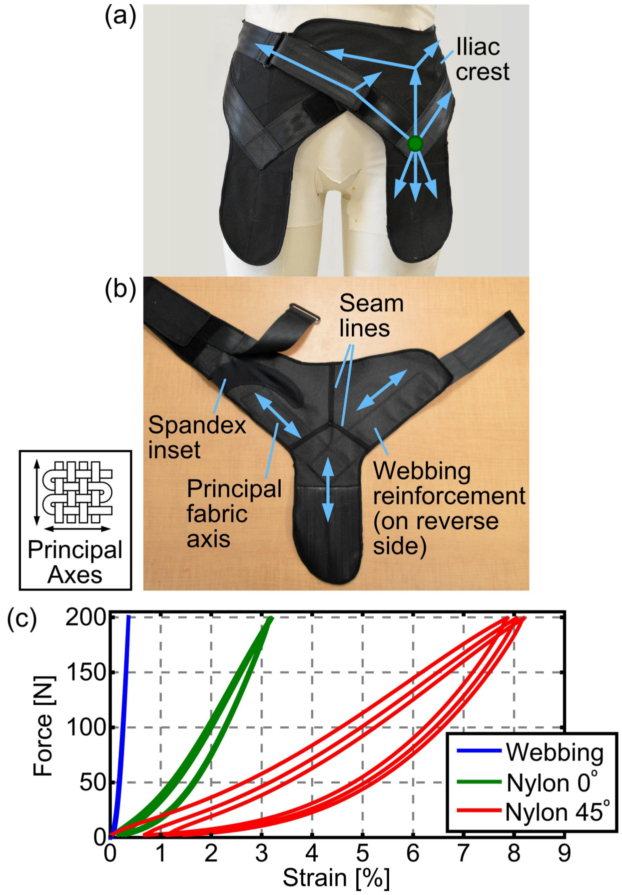

Fig. 3. (a) Front view of waist belt with arrows indicating force paths throughout the garment. (b) View of the reverse side of the left leg of the waist belt in (a), showing key features for load distribution. (c) Results from testing three textiles showing different strains under load and different hysteresis.

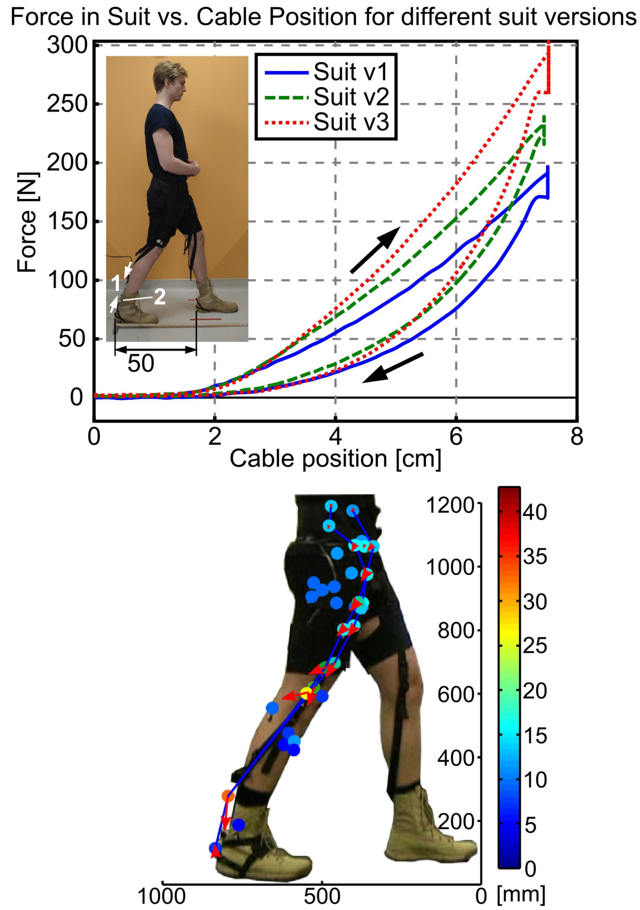

Fig. 4. Top, suit-human stiffness testing results for several different suit versions. The test setup is shown in the inset, with cable displacements applied between the heel and the back of the calf (1), and forces measured at the heel attachment point (2). Arrows indicate the direction around the hysteresis loop. Bottom, results of tracking points on a suit during stiffness testing. The color of the markers shows the total net displacement during a 7cm pull, while the arrows show the direction of motion.

To best utilize the fabric to convey forces in the desired pattern, we use three panels oriented so that the principal fabric axes are parallel to the desired force paths. The strain of the fabric under load matters greatly, since large displacements will reduce the stiffness of the exosuit and require increased power when actuated [35].

The nylon base layer is further stiffened by seatbelt webbing (Seatbelt Planet, Inc.) in the main load distribution paths across the body and around the side of the leg. As shown in Figure 3(c), this webbing has a much lower strain than the nylon fabric (0.3% vs. 3.2% under a 200 N load) due to its dense weave structure and increased thickness (1.2 mm), but at the cost of decreased conformability.

Finally, we use spandex fabric directly over the iliac crest of the pelvis to virtually eliminate vertical shear forces, since the spandex stretches 45% under forces of only 5 N. This causes the forces to be routed through the front of the exosuit, which is the desired path for this embodiment.

C. Textile Evaluation

Finally, we need to evaluate how well the textile portion of the exosuit works, both for modeling from a systems-level perspective and for understanding how the exosuit moves relative to the human when force is applied. Since our compliant exosuits are interacting with a compliant human body, we have needed to devise new tools that properly characterize and define the performance of our systems.

We characterize the suit-human effective stiffness and overall force-displacement properties, as shown in the top of Figure 4. The testing procedure is as follows: a subject stands stationary in a pose close to that at 50% in the gait cycle, as shown in the inset. The actuators then command a position profile at (1), reducing the suit's length between the back of the heel and the back of the calf, and we measure the force in the suit at the back of the heel (2). By plotting the resulting force-displacement curve, we can determine the effective stiffness of the suit-human system resulting from motor displacements. The figure shows how we have made improvements to the suits over time, with successive versions having higher stiffness. Each subsequent version incorporated lower-stretch materials, utilized load paths that followed more direct routes from the waist to the calf along the leg, and included increased fabric coverage around the waist, thigh, and calf. As shown in Figure 4, the human-suit system does have hysteresis, typically losing 35% of the input energy; so it is not a perfect energy harvesting mechanism. The resulting force-displacement curves can be fitted with equations and used in calculations and simulations of the suit behavior. In [36] we utilize these equations to calculate the energy flow process between the human and the suit.

A second way of characterizing the suits is to observe their motion over the body as shown in the bottom of Figure 4. We placed reflective markers on both the suit and body, and repeat the stiffness testing procedure by capturing the positions of the markers through a VICON motion capture system. By observing the deflection of these points, we can determine how the suit is moving relative to the body, and detect regions where the suit is stretching large amounts to optimize our designs and fabric selection in these regions.

III. Low-power actuation with flexible transmissions

The next key component of an exosuit is the power source and transmission. These need to be able to convey power to distal body segments while conforming to the body and not restricting its motion. Furthermore, the actuation scheme needs to be fast enough to move with the limb and displace the series compliance of the human body, the suit, and the interface between the human and the suit. During human walking, positive power is generated by the muscles at the joints in short bursts. Thus, the actuators of an exosuit must be able to function with this timing and utilization as well.

To determine the actuator specifications, the starting point is the biological moments and kinematics of the joints. From there, the series compliance of the suit-human system (from measurements in Figure 4) must also be considered. With the human tissue and suit displacing under applied forces, to achieve a given joint moment the actuators must move further than would be required if there was a rigid connection to the body. In our exosuits, accommodating this additional displacement means that the actuators must move roughly twice as far (and thus twice as fast) as if they were connected to a rigid system.

There are several ways of achieving this flexible transmission with a high power source. One is to mount a motor directly on the suit which pulls a cable, at the cost of increased distal mass. Another option is to use a proximally-mounted geared motor driving a Bowden cable, similar to that which was done in the LOPES exoskeleton and others [15]. Bowden cables are able to transmit force between the motor and the region where the inner cable exits the sheath without any restrictions on the intermediate path. Their main drawback is their efficiency, which can vary from 50-85% depending on the sheath and cable construction, and bends in the cable can reduce this further. However, these effects can be minimized by reducing the cable length and by routing the cable along the leg such that it is mostly straight when actuated.

In parallel with the development of portable systems, we have developed a lab-based actuation platform that can drive Bowden cables with high-power motors [36]. This is shown at the top of Figure 5, and is useful for rapidly optimizing design and control strategies to actuate several joints. Such an

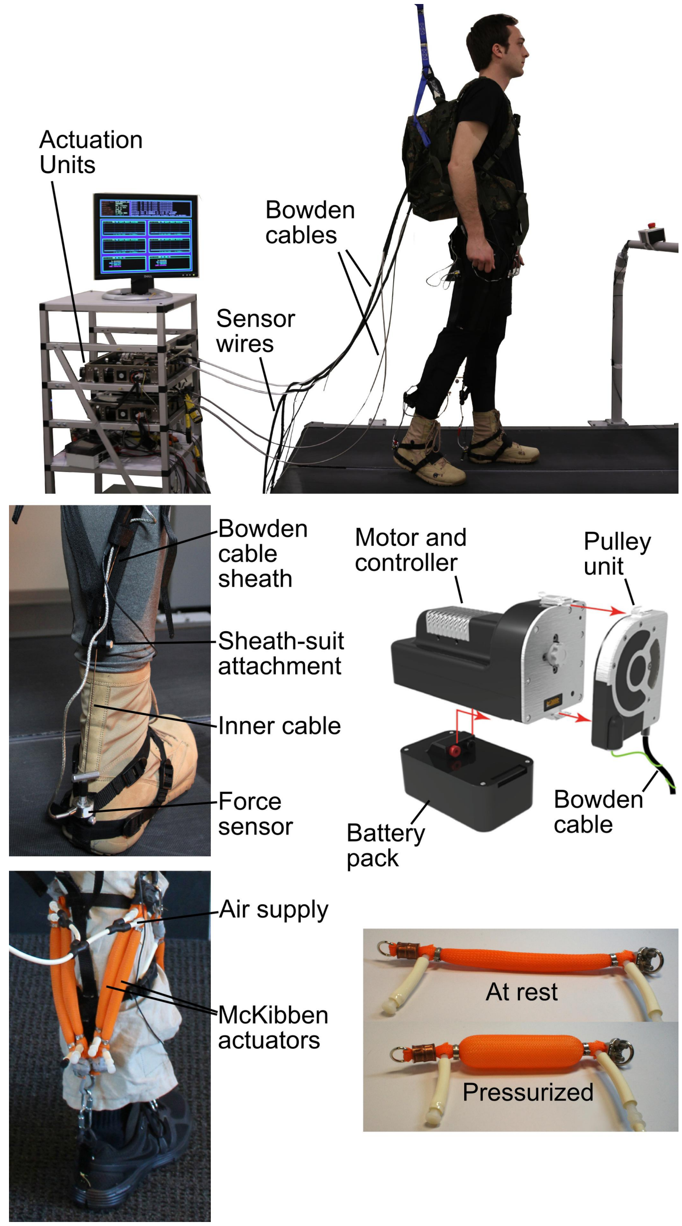

Fig. 5. Top, off-board actuation system capable of powering multiple joints with Bowden cables. Middle, detailed view of how a Bowden cable attaches across the ankle joint, and a mobile actuator unit used to retract Bowden cables. Bottom, detailed view of a pneumatically-powered exosuit actuating the ankle.

approach allows us to rapidly explore the basic science around human-machine interaction with such systems that can then be used to guide the design of our portable systems.

In addition, we have developed portable actuators so our exosuits can be used outside of the laboratory. Our recent implementations weigh only 5.5 kg including batteries for up to 4 hours of continuous walking, and consume approximately 50 W on average. These are shown in the middle row of Figure 5, as well as a detailed view of how the Bowden cable attaches to the exosuit around the ankle. The portable actuators drive the Bowden cable by winding the inner cable around a pulley driven by a geared motor. Since the Bowden cables and sensor wires are integrated into the exosuit, we have designed the pulley and cable to be removable from the motors and batteries to easily disconnect the suit from the actuators and controller.

As was mentioned earlier, an alternative approach to minimizing the mass on the wearer is to use pneumatic actuation, shown in the bottom row of Figure 5. While McKibben actuators are lightweight and intrinsically compliant, pneumatic systems able to deliver high forces (>150 N) typically require a powerful (>1 kW) air compressor to provide sufficient air flow and pressure for walking applications, and can also be more difficult to control than electromechanical actuators.

IV. Sensor Systems

New sensor systems that are easy to integrate with textiles and soft components are required in order to properly control and evaluate soft exosuits. Rigid exoskeletons usually include sensors such as encoders or potentiometers in robotic joints that accurately track joint angles, but these technologies are not compatible with soft structures. We are designing new sensors to measure human kinematics and suit-human interaction forces that are robust, compliant, cost effective, and offer easy integration into wearable garments. Some examples of the sensors we use are shown in Figure 6 including a soft kinematic sensing suit, suit-human interaction force sensors, a foot-mounted accelerometer, and footswitches.

A. Integrated Kinematic Sensing

Kinematic sensors are useful for monitoring joint angles in real-time, so control systems can have an estimate of the body's motion. This approach is especially important for using these systems outside of the laboratory in challenging environments and when performing activities of daily living.

Previous work on wearable sensors to measure human kinematics include compliant sensors such as nanotubes or silicon encapsulated in soft polymers, which require complex fabrication techniques, or inertial measurement units (IMUs). While extensive work has been done to properly measure human kinematics with IMUs, these systems require additional sensors or aggressive filtering techniques to avoid problems related to integration drift [38], [39].

To address the limitations of previous soft sensors, the Microrobotics Lab at Harvard has designed a suite of soft sensors that can measure strain, pressure, curvature and shear [40]–[44]. These sensors are all based on the concept of embedded a liquid metal (eutectic gallium indium alloy) in channels in a hyper-elastic silicone material that acts as a variable resistor. Deformation of the material due to external disturbances changes the geometry of the channels and thus the resistance, which can easily be measured. The compliant nature of these sensors means they can be integrated into wearable garments and robots [28], [37], [41], [45], [46].

To demonstrate the potential of these hyper-elastic strain sensors to measure joint kinematics, a soft sensing suit was developed in our lab and is shown in Figure 6. The sensors spanned the hip, knee and ankle joints and strain as a function of the joint angle and therefore can be used to measure joint kinematics in the sagittal plane. In initial walking experiments on a treadmill with three subjects, the sensor data was compared to that from a motion capture system (Vicon), and we found that the resulting root-mean-square errors for estimating joint angles were less than 5° [37] which may be sufficient for understanding the motion of the wearer. The bottom of Figure 6 shows the joint angles from Vicon in thin gray lines, as compared to the soft sensor data in thicker lines.

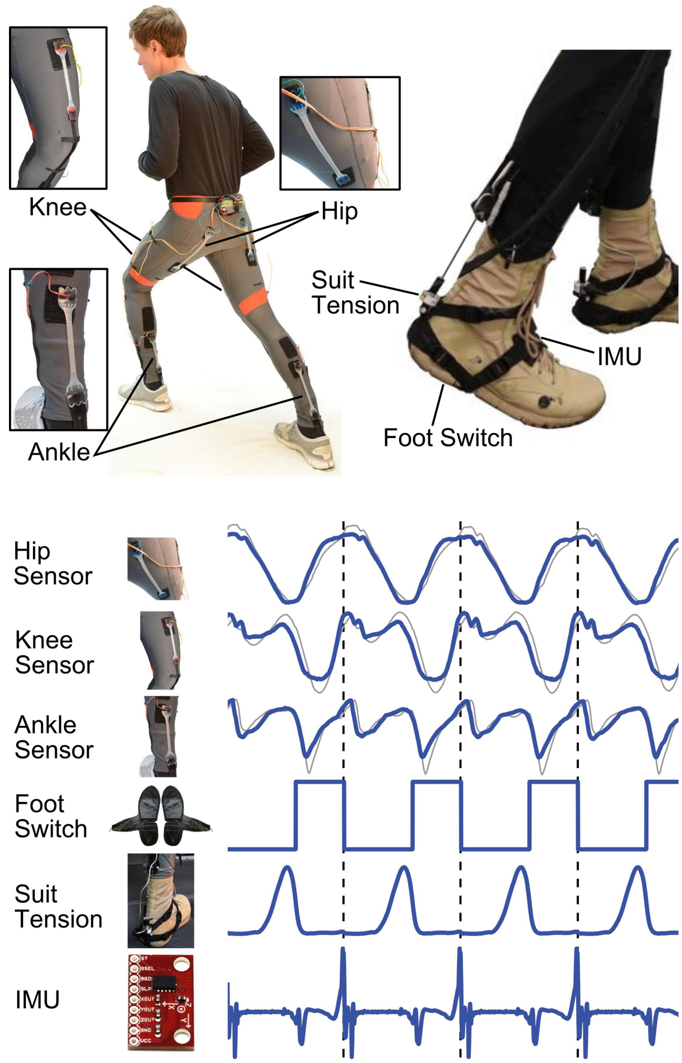

Fig. 6. Sensor systems. Top left, sensor suit to measure gait kinematics [37]. Top right, sensors integrated in the boot and human-suit interface to measure gait events and interaction forces. Bottom, different sensors integrated in our systems and example signals over time, with vertical dashed lines indicating heel-strike events. Thin gray lines for the hip, knee, and ankle sensors are ground truth joint angle data from a Vicon motion capture system.

B. Detecting Gait Events and Interaction Forces

In addition to joint kinematic measurements, we use several additional sensors in conjunction with our soft exosuits. These include footswitch insoles and accelerometers as shown in Figure 6. Both accelerometers and footswitches have been extensively used by different research groups to detect gait events [47]. A final sensor, used in our exosuits, is a load cell located at the connection between the foot attachment and the actuation cable. Since this sensor is in series with the suit and actuators, it can be used to monitor the suit tension or perform closed-loop control with the actuators.

V. Human-suit Interaction Methods

Previous work on control methods to assist with locomotion has been influenced by the mechanical characteristics of traditional rigid exoskeletons. These systems add significant inertia to the human leg that places bounds on the types of interactions between the wearer and the device [48], [49]. As was discussed earlier, the exosuits we are designing are lightweight, soft, and do not restrict the human natural kinematics or range of motion. Moreover, they can easily become fully transparent to the wearer by extending the actuators so the suit is not under tension. Control methods can take advantage of this feature to economize battery, or become fully transparent when the user is doing challenging actions or in the event of a low-battery condition. On the other hand, while rigid exoskeletons can apply higher forces, they typically require a lot of power and control considerations to become fully transparent to the wearer. These inherent differences result in different control strategies and opportunities for new research on human-machine interaction.

A. Assistive Force Generation

Our exosuits are intended to apply torques at the joints in synchrony with the underlying muscles, as illustrated in Figure 2. When applying forces with actuator, it is important that the forces are applied gradually to mimic the onset of forces in muscles as too rapid an increase in force may cause the muscles to react adversely. In our mono-articular exosuit described in [25], we used pneumatic actuators that were controlled in a straightforward open-loop manner by pressurizing and depressurizing them at a desired time in the gait cycle. The inherent fluidic and mechanical compliance of these actuators resulted in a smooth first-order time response where the actuators increased to 90% of the desired maximum value over approximately 200 ms. In a pilot human walking study, we varied the actuator turn-on time as a function of the gait cycle to determine when it would be most beneficial. We found that an actuator turn-on time of 30% in the gait cycle was metabolically optimal, and corresponded to the force profile extending from 35-62% in the gait cycle.

B. Force Control

While pneumatic actuation was sufficient for our proof-of-concept work, achieving accurate position or force control with this type of actuation is challenging. Thus, to enable better control over the applied force profile, we switched to using electromechanical actuation and Bowden cable transmissions for our subsequent systems as were described previously.

In order to transmit biologically-realistic torques to the human joints, one option is to use a real-time force controller. We have implemented this with our non-portable actuation system in Figure 5 using the suit tension load cell for feedback [36]. Implementing a force controller requires an actuator with a relatively high force bandwidth due to the compliance of soft exosuits and Bowden cable transmission. We have characterized the force bandwidth of this system to be 20 Hz when delivering a 200 N peak-to-peak force with the distal ends of both the Bowden inner cable and sheath clamped to a rigid plate. Through human subject experiments, we have demonstrated that our real-time controlled system can accurately deliver high forces to the user (up to 250 N) through soft exosuits when walking at 1.25 m/sec.

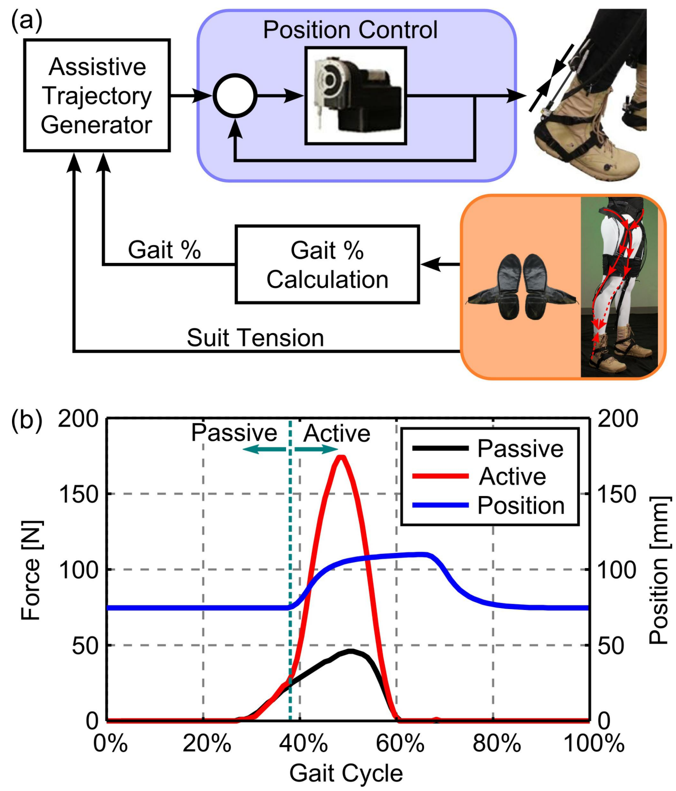

Fig. 7. Exosuit-Human Interface based on Integrated Sensor Measurements. (a) Force-based position control architecture. (b) Generated force profile with the suit in passive mode (black line), commanded position profile to assist ankle plantarflexion (blue line), and resulting force profile (red line).

C. Force-based Position Control

An alternative to force control that requires a much lower actuator bandwidth is to use position control of the Bowden cable. By driving a Bowden cable through a specified position trajectory, consistent forces are created in the suit assuming repeatable force-displacement characteristics of the suit and human (Figure 4). With the suit-human force-displacement model described in section II, we can generate the correct cable position profile, and play it back as a function of the percentage through the gait cycle. When no force should be present in the suit, the actuators are commanded to stay at a fixed initial position so that the suit is slack. We used this scheme in earlier work, which resulted in consistent force profiles delivered to the wearer [50].

As discussed in section II, the multi-articular exosuit architecture becomes stretched when the body is in the correct pose for forces to be applied, and absorbs energy and returns it to the body even when the actuators are in the initial offset position. The resulting passively-induced force as a function of the percentage through the gait cycle is shown in Figure 7(b) as the black line.

From a control standpoint, this passively-induced force is extremely useful. If the actuators are held at a fixed initial position, force in the exosuit means that the wearer is beginning to transition between legs, and so additional assistance would be beneficial. As such, we monitor this passive force with the suit tension load cell, and use the measurements for control.

Figure 7(b) also shows the result of actuating the suit, with the actuator position in blue and the resulting force in the suit in red. When the actuator shortens the effective suit length, the force in the suit increases substantially, imparting an extra boost of power to the user at the correct time.

Due to their highly compliant nature, soft exosuits can deform over time or move relative to the body if worn for extended periods of time while walking. Moreover, changes in gait may modify the resulting force profile and amplitude with this position-control scheme. This presents a challenge when commanding the suits in position control, since the assistive profiles resulting from the position controller will vary over time for different human motions and suit alignments. To improve upon this, we developed a controller that monitors key force profile features including the peak force and the passively generated force before actuation, and automatically adjusts the assistive position profile to keep the desired force consistent over time or between users. If during a gait cycle, the resulting peak force or the passively generated force is different than desired, the initial offset and the maximum amplitude of the position profile are increased or decreased so that the forces are corrected for future steps. The maximum correction per step is limited to a low value so that there is no significant difference in the applied force between two consecutive steps. Moreover, when the wearer turns on the system the position profile will ramp-up slowly until the desired force profile is achieved. This controller corrects the position profile so that the desired forces are achieved independently of the way that the suit is initially positioned for a particular wearer or of the relative motions between the suit and the human.

VI. Exosuit Performance Metrics

Performance metrics for the evaluation of wearable robots are strongly dependent on the application of the device, as robots with different purposes have different requirements and should be evaluated differently.

We propose a set of metrics that are appropriate to evaluate lower-body soft exosuits (and wearable robots in general) for performance enhancement (i.e. augmentation). These were specified in the Introduction, namely that an exosuit leaves the user in full control over his/her own gait, it introduces minor to no changes to their natural gait, and it assists the lower body during walking.

In line with other groups past work on powered exoskeletons (e.g. [7], [51]), our approach to evaluating exosuits is to define a specific task (e.g. 10 minutes of treadmill walking at 1.25 m/s with a 25 kg backpack) and measure gait kinematics, dynamics and energetics comparing three different conditions: wearing the exosuit in active mode ('active') vs. wearing the exosuit in transparent mode ('slack') vs. not wearing the exosuit at all ('no suit').

A. Gait Kinematics

We analyze the effect of the exosuits on gait kinematics by calculating the average hip, knee and ankle angles in the sagittal plane, as well as in the frontal and transverse planes. By comparing the average profile and range of motion of each joint in the three conditions, we can identify how the soft exosuit itself impacts gait (slack vs. no suit) and how the assistance applied by the exosuit changes kinematics (active vs. slack). It is desirable that such changes are minimal and in any case not disruptive to natural gait. The analysis of ground reaction forces (GRF) also allows us to determine whether the active suit promotes changes to the natural gait frequency compared to normal walking, or if it changes the relative duration of stance and swing.

B. Gait Dynamics and Energetics

We study to what extent the active exosuit is assisting the human by analyzing gait dynamics and kinetics (joint moments, power, force delivered by the exosuit). Inverse dynamics is an effective way to determine to what degree the exosuit is augmenting the body function at a joint level. The comparison of joint moments and suit assistive forces allows us to monitor the degree of synchronicity between the user and the robot.

Our motion capture lab utilizes a Vicon T-series 9-camera system for motion capture, together with a Bertec fully instrumented split-belt treadmill to measure GRFs. The Vicon Nexus software is used in combination with C-Motion Visual 3D and custom Matlab processing scripts to calculate inverse kinematic and dynamic variables.

Surface electromyography (sEMG) can be used to selectively monitor muscular activity focusing on the muscle groups that are most relevant for the task under consideration (for walking, the calf muscles and hip flexors and extensors). Comparing the ensemble average profiles of sEMG activity between the slack, active and no suit conditions allows us to determine effects on the maximum force being delivered by each muscle (peak sEMG activation) and on the energy cost of each muscle activation (integral sEMG). sEMG is measured with a Delsys Trigno or Bagnoli dry-electrode system supporting up to 16 electrodes. A typical electrode configuration during walking would include electrodes to measure ankle plantarflexors (*Soleus* and *Gastrocnemius Medialis*), ankle dorsiflexors (*Tibialis Anterior*), knee flexors and extensors (e.g. *Sartorius* and *Biceps Femoris*) and hip flexors and extensors (e.g. *Gluteus* and *Quadriceps Femoris*).

Metabolic Cost of Walking (MCW) is a global physiological measurement to determine to what extent the suit is assisting the wearer (reduction in MCW between active and slack) and if assistance offsets the weight of the device (reduction in MCW between active and no suit). Metabolic cost is assessed using a COSMED K4b2 portable system (COSMED Srl, Rome, Italy) for pulmonary gas exchange measurement.

C. Challenges in Evaluating Exosuit Performance

The evaluation of exosuit performance suffers from several confounding factors related to the complexity and duration involved in the experimental sessions. A typical experiment would involve multiple sessions interleaved by rest periods leading potentially to multiple hours of continuous experimentation.

With this time frame, effects such as fatigue, motor learning and gait adaptation can play a relevant role in changing gait kinematics and energetics. For example, the onset of fatigue is known to change the frequency spectrum of muscle activations [52], as well as to increase the metabolic cost of walking. In addition, motor learning effects leading to changes in gait kinematics and muscle activation have been demonstrated in other lower-body wearable robots [51].

Other confounding factors, such as perspiration, air humidity, digestion and mental fatigue can create challenges in drawing conclusions from experimental data collected on human subjects. Thus, accurate control over the experiment timing and consistency are paramount in achieving a reliable evaluation.

D. Results

For our early tethered prototype shown in Figure 1 (left) and presented in [25], we demonstrated that a wearable, pneumatically-powered soft exosuit can assist normal, unloaded walking with minimal changes to gait kinematics. The pneumatic exosuit was programmed to generate boosts of assistive force with different delays from heel strike, ranging from 0% delay (at heel strike) to 60% delay (at toe off). An activation delay of 30% of gait duration in conjunction with the gradual actuator response (peaking after approximately 200 ms from the control signal being sent to the valve) generated a smooth force profile synchronized with ankle plantarflexion during the push-off phase, the most-energy relevant phase of walking. We found that this resulted in minimal changes to the kinematics of the hip, knee and ankle. Early energetics results also showed that in this best case (one subject) the MCW of wearing the 7.1 kg suit and control box was substantially identical to that of wearing no suit at all (386.7 $\pm$ 4.4 W Active vs. 381.8 $\pm$ 6.0 W No suit), showing that the exosuit could effectively offset the added metabolic cost of wearing the device. A best-case reduction of 10.2% was demonstrated when comparing the active suit vs. slack suit. In addition, we found that the MCW was quite sensitive to changes in the actuation timing. A variation of 10% in the activation delay had a detrimental effect on the MCW by more than 13% (438.8 $\pm$ 3.4 W when actuating at 20% of gait).

In evaluating our latest prototype shown in Figure 1 (right) we focused our attention on the analysis of loaded walking (1.25 m/s with a 24.5 kg backpack + weight of the device). Figure 8 shows an example result of wearing an exosuit prototype on gait kinematics and kinetics. The bottom plot shows the assistance delivered by the exosuit at the heel.

The top three plots show how the exosuit does not significantly affect hip and knee kinematics. The ankle shows reduced dorsiflexion and increased plantarflexion at push-off, in accordance with the assistance the exosuit is providing. Such a change in gait is minimal and ensures a very natural gait pattern. It can be also seen how the suit causes a reduction in the first peak of vertical GRF during early stance. This may be a consequence of a reduced acceleration towards the ground during the load acceptance phase caused by the exosuit action on the contralateral leg during late stance. These results suggest how the suit is capable of assisting gait while not causing any disruptive change compared to natural walking.

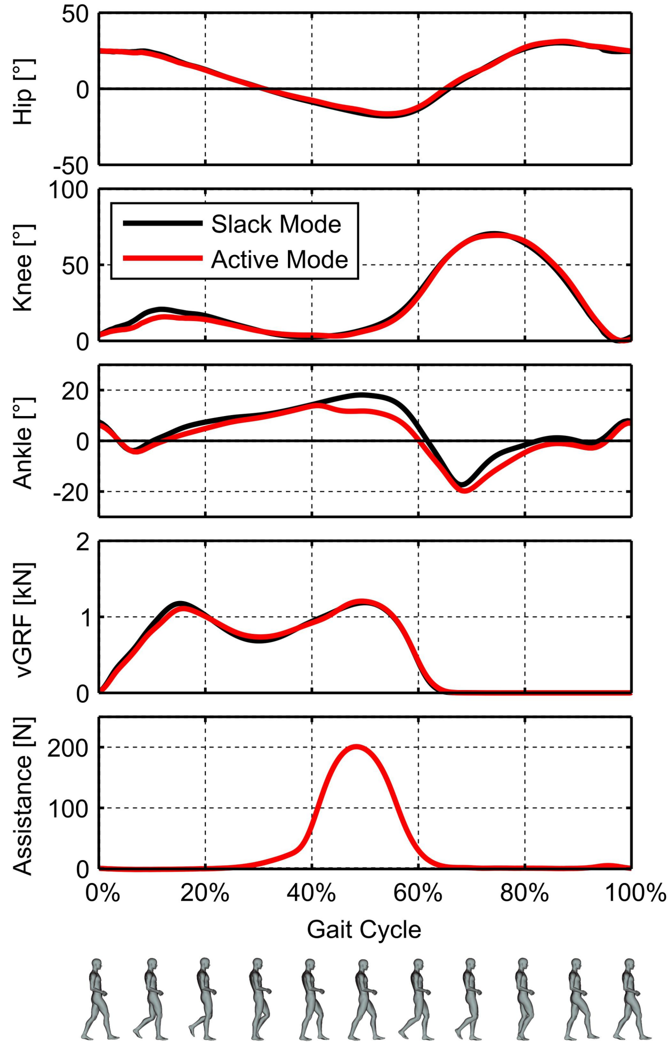

Fig. 8. Effect of a soft exosuit on gait. Top to bottom, hip, knee and ankle angles in the sagittal plane, showing no substantial changes to hip and knee kinematic and reduced dorsiflexion / increased plantarflexion at push off; vertical ground reaction force showing a reduced peak at early stance; assistive force generated by suit at the heel. The bottom row of figures shows the position of the body during each stage of the walking cycle.

For an earlier revision of this device, having a weight of 10.1 kg, energetic results published in [50] show an average reduction of approximately 6.4% in the best-case MCW (active vs. slack) on a pool of 5 healthy subjects, showing that the suit is capable of effectively assisting gait. The metabolic cost of carrying the system mass was experimentally measured to raise the metabolism by approximately 16.7%, which is 1.55%/kg of system mass. This value is commensurate with previous

TABLE II
METABOLIC RESULTS

System	N	Weight carried	MCW *Active vs. Slack*	MCW *Slack vs. No suit*	MCW *Active vs. No suit*
Pneumatic, ***tethered*** [25]	1	7.1 kg system	-10.2%	+12.8%	+1.3%
Electro-mechanical, ***mobile*** [50]	4	10.1 kg system + 24.5 kg payload	-6.4%	+16.7%*	+9.3%**

This table shows the effect of soft exosuits on the metabolic cost of walking (MCW) at 1.25 m/s. Our early, pneumatic prototype [25] was tested during walking without any additional load beyond the weight of the system, which included the exosuit and control box. The air compressor was not carried by the subject. Our mobile electromechanical system presented in [50] was tested during loaded walking (24.5 kg for backpack and load, and 10.1 kg for the exosuit and actuation units) on $N = 4$ subjects. In this case, the metabolic savings produced by the device was not sufficient to offset the added cost of carrying the actuator mass. *$N = 2$ subjects. **Calculated from values in previous two columns.

studies, which estimate the cost of carrying load to be between 1-2%/kg for mass carried on the torso, and 8%/kg for mass at the foot [20].

Table II summarizes the energetic effects of the two different systems for walking at 1.25 m/s in unloaded and loaded conditions. Reducing the weight of these systems will be a key element of future work to bring exosuits to achieve a net metabolic benefit (active vs. no suit). Our most recent exosuit embodiment shown in Figure 1 has approximately half the weight of the system in [50], and we are currently in the process of evaluating its effect on the MCW.

VII. Conclusion

Exosuits show much promise as a method for augmenting the body with lightweight, portable and compliant wearable systems. We envision such systems can be further refined so that they can be sufficiently low-profile to fit under a wearer's existing clothing. Our focus is on creating an assistive device that provides a fraction of the nominal biological torques and does not provide external load transfer. In early work, we have shown that the system can substantially maintain normal biomechanics and positively affect a wearers metabolic rate.

Many basic fundamental research and development challenges remain in actuator development, textile innovation, soft sensor development, human-machine interface (control), biomechanics and physiology that provide fertile ground for academics in many disciplines. While we have focused on gait assistance thus far, numerous other applications are possible, including rehabilitation, upper-body support, and assistance for other motions. We look forward to a future where wearable robots provide benefits for people across many areas of our society.

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