Biologically-inspired Soft Exosuit

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Abstract*— In this paper, we present the design and evaluation of a novel soft cable-driven exosuit that can apply forces to the body to assist walking. Unlike traditional exoskeletons which contain rigid framing elements, the soft exosuit is worn like clothing, yet can generate moments at the ankle and hip with magnitudes of 18% and 30% of those naturally generated by the body during walking, respectively. Our design uses geared motors to pull on Bowden cables connected to the suit near the ankle. The suit has the advantages over a traditional exoskeleton in that the wearer's joints are unconstrained by external rigid structures, and the worn part of the suit is extremely light, which minimizes the suit's unintentional interference with the body's natural biomechanics. However, a soft suit presents challenges related to actuation force transfer and control, since the body is compliant and cannot support large pressures comfortably. We discuss the design of the suit and actuation system, including principles by which soft suits can transfer force to the body effectively and the biological inspiration for the design. For a soft exosuit, an important design parameter is the combined effective stiffness of the suit and its interface to the wearer. We characterize the exosuit's effective stiffness, and present preliminary results from it generating assistive torques to a subject during walking. We envision such an exosuit having broad applicability for assisting healthy individuals as well as those with muscle weakness.

Keywords—exosuit; walking; wearable robot, soft robot; exoskeleton

I. INTRODUCTION

Robotic exoskeletons have been developed for a large number of applications, with the goal of assisting or enhancing human activities. For the lower extremity, devices have been developed to apply assistive torques to the biological joints to augment healthy individuals or assist those with disabilities to walk [1-10]; assist with load carriage by providing a parallel path to transfer load to the ground [11-14], thus off-loading the wearer and bypassing their musculature; or finally, provide gait retraining or rehabilitation for those with disabilities [15-18]. These systems all are based on the principle of having a robotic mechanism (with rigid links, joints, and actuators) that runs in parallel with the biological limb. Typically, the links of these exoskeletons are connected to the wearer at a few locations with a waist belt, leg strapping, or foot attachment. In our recent work we have been exploring the use of soft flexible materials as an alternative means to not only interface to the

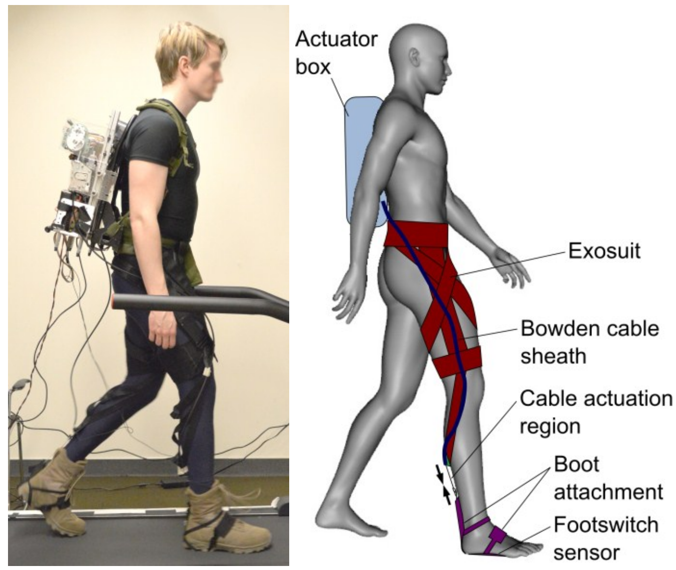

Fig. 1. Left, prototype exosuit described in this paper. Right, illustration showing the main components.

wearer, but also provide a flexible structure so that assistive torques can be applied to the biological joints [19]. Fig. 1 shows the design of such an "exosuit" that is the focus of this paper.

The optimum form and function of an exoskeleton will be closely tied to its intended application; however, there are two main challenges that apply to the design of these systems. The first major challenge relates to the exoskeleton imposing kinematic restrictions on the wearer's natural degrees of freedom. By having a parallel robotic mechanism, care has to be taken to ensure alignment between the robotic joints and those of the biological limb. This is challenging because the biological joints do not have a fixed center of rotation [20]. Misalignment with the wearer's kinematics will cause forces or torques that resist the wearer's motion, increasing the effort required to move, altering natural gait patterns and decreasing the exoskeleton's efficacy [21]. To address this, self-aligning mechanisms have been proposed [22, 23]; however, this results in an overall increased system mass. An approach often taken to reduce the challenge associated with matching the kinematics of the exoskeleton and wearer is to have a device that provides assistance at a single joint only. Most of these systems have focused on the ankle [6, 7, 24] or the knee [4] as

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they can be approximated as pin joints in the sagittal plane; however, alignment with the joint remains important.

The second main challenge associated with traditional exoskeletons is that they are relatively heavy. While actuation can help accelerate and decelerate this inertia, often the wearer must assist with this. This effect is often an increased effort for the wearer and this is most significant when the mass is located distally near the foot and ankle [25]. For instance, a mass of 4kg near the wearer's center of mass has been shown to be associated with a 7.6% increase in metabolic effort, but this increases to 34% for the same amount attached at both of the feet. The mass of a heavy exoskeleton leads to large power requirements to power it which in turns leads to needing large payloads of batteries or other energy storage means in order to achieve a mobile system. To address the power requirement issue, a number of groups have looked at using passive and quasi-passive systems that can exploit the natural generation and absorption of energy during the gait [8, 9].

Despite the drawbacks associated with restricting kinematics and added inertia, a number of exoskeletons have demonstrated the ability to augment or assist the wearer. In two human subjects studies with a leg exoskeleton for load carrying, significant off-loading was demonstrated [12, 26]; however, at the cost of an increased metabolic effort to walk. If such systems can be improved to the point where they are neutral in terms of metabolic effect, the benefits of off-loading the skeletal and musculature could reduce the risk of injury. A number of exoskeletons have also recently demonstrated that the metabolic cost of the wearer can be reduced under certain conditions. A single degree of freedom tethered pneumatic exoskeleton for the ankle demonstrated that at fast walking (1.75 m/s), metabolic effort could be reduced by 13.8% [7]. Recently, a squat-assistance exoskeleton was shown to reduce the wearer's metabolic effort [27] and an exoskeleton in parallel with the legs was shown to reduce the metabolic effort required to hop in place [28]. However, it remains an open research question if a net metabolic reduction can be achieved for walking or running with a non-tethered portable system.

In contrast to traditional exoskeleton systems to augment human capability, a new paradigm is to use soft clothing-like "exosuits" to achieve many of the same objectives [19]. In this case, the suit does not contain any rigid elements supporting compressive loads, so the wearer's bone structure must sustain all the compressive forces normally encountered by the body plus the forces generated by the exosuit.

Exosuits offer a number of benefits as compared to traditional exoskeletons. The suits themselves, composed primarily of fabrics, can be significantly lighter than an exoskeleton frame or linkage system, leading to very low inertias and lower cost of transporting the suit mass. Since the suit does not contain a rigid frame, it also provides minimal restrictions to the wearer's natural kinematics, avoiding problems relating to joint misalignment. Further, control for the system can be less precise as the inherent compliance in the system can make it more forgiving. Finally, the suit can also provide torques on multiple joints simultaneously but having it span multiple joints, which may assist with reducing the total number of actuators.

This paper focuses on the initial development of a biologically-inspired soft exosuit designed to create joint torques on the wearer during walking. A battery-powered portable prototype system was demonstrated using a motor-actuated cable-drive system. Preliminary human studies indicate that it is possible to apply significant forces and torques with a soft suit, without causing discomfort or injury.

II. Biologically-inspired Soft Exosuit

To achieve a lightweight and efficient exosuit, there is much insight to be gained through understanding how humans walk. By designing the architecture of the exosuit to mimic some of that of the biological limb, it is hypothesized that a more transparent, safe, and effective design can be achieved.

A. Motivation

The goal for the exosuit is to assist with forward propulsion during walking on level ground at 1.25 m/s by applying appropriate assistance at the ankle, knee, and hip joints. Specifically, the goal is to duplicate the forces generated by muscles and tendons in the biological leg. In principle, if the suit generates forces on the body that mirror what the muscles do, the muscles may be required to do less work, thus reducing the wearer's metabolic cost of transport. Human walking is very efficient due to the tuned limb lengths, stiffnesses, and muscle sizes in the leg. The system can be well-modeled as a passive dynamic walking system requiring minimal energy input [29], and these principles have been implemented in robots that demonstrated a similar cost of transport to human locomotion [30].

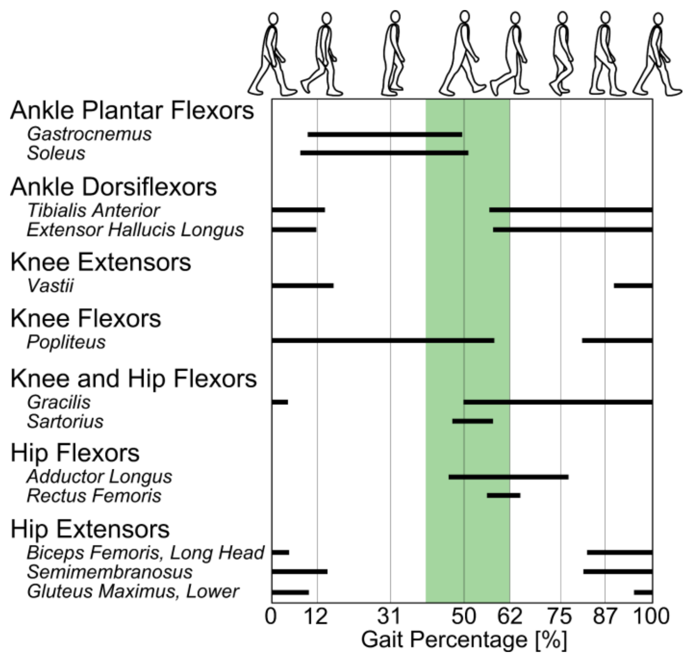

Fig. 2. Diagram of when the muscles in the leg are active. Our suit is active for the duration specified by the shaded region, to assist ankle plantar flexion, knee flexion, and hip flexion. The gait percentages correspond to the following periods of the gait: 0-12%: Loading Response; 12-31%: Mid-stance; 31-50%: Terminal stance; 50-62%: Pre-swing; 62-75%: Initial swing; 75-87%: Mid-swing; 87-100%: Terminal swing.

During normal walking, power is expended by the body primarily at the transitions of support from one leg to the other and this power is provided largely by the hip and ankle. During stance, the calf muscles (gastrocnemius, soleus, etc.) contract isometrically, which stretches the Achilles tendon due to the natural motion of the body falling forward. Subsequently, these muscles contract concentrically and the Achilles tendon recoils, giving a large positive power burst from 40-60% of the gait cycle, and providing 0.39 J/kg of energy to help redirect the body's momentum and prepare for the opposite leg's touchdown [31]. This creates a moment about the foot which propels the body upward and forwards. The gastrocnemius muscle also flexes the knee during this time, permitting the foot to clear the ground for swing. The activity of the muscles in the leg as a function of the gait cycle is shown in Fig. 2.

The hip provides a smaller power burst and a period of power absorption during the transition between legs. The power absorption occurs from 35-50% of the gait cycle, absorbing 0.12 J/kg to help stop the body's falling forward, and occurs through the extension of the tendons at the front of the hip. The power burst occurs from 50-75% of the gait cycle, providing 0.14 J/kg of energy to help swing the leg [31]. This is accomplished by the contraction of the muscles at the front of the thigh (rectus femoris, adductor longus, adductor magnus, etc.). The exosuit is designed to assist the power absorption through passively extending, as well as assisting the muscle contractions by actuating. To aid both the hip and the ankle, the suit is primarily actuated from 40-60% of the gait cycle, shown in Fig. 2 by the shaded region, then is made slack otherwise.

B. Design Overview

Based on this understanding of the biomechanics of walking, an exosuit consisting of various fabrics sewn together in a conformal form factor was designed. Fig. 3 shows a diagram of the forces in the suit when tension is applied to it with a cable close to the ankle. The numbers referred to in the text are labels in the figure. The exosuit attaches around the waist (1) and above the knee (6), and transfers force between the back of the calf and the waist through a series of webbing straps (2-5), (7). The webbing attaches to the sheath of a Bowden cable at the bottom of strap (7), which is located at the back of the calf. The cable itself extends downward from this sheath termination point to a small webbing strap coming up from the heel, where it is attached to the wearer's shoe. When the Bowden cable is actuated, it pulls these two points together. Thus, the back of the ankle is pulled upward and the bottom of the exosuit is pulled downward during actuation. The exosuit then transfers the force up to the wearer's waist, so the pelvis bone is pulled downward. The skeletal structure of the wearer then transfers this downward force back to the ankle joint and to the ground through the foot.

The actuation applied by the cable creates tension in the suit that creates moments about the ankle, knee, and hip. A moment is generated about the ankle because the webbing on the back of the heel, pulling upward, is several centimeters away from the ankle pivot point, which experiences the downward force through the bone structure. Moments are generated about the hip joint since the majority of the straps (2-4) are positioned in front of the hip. The lower straps (7) create a small moment flexing the knee due to their attaching above the knee joint on strap (6), and the knee being bent slightly during the time in the gait when the cable is actuated. On a typical subject, the moment arms about the hip, knee, and ankle are 8cm, 1cm, and 8cm, respectively.

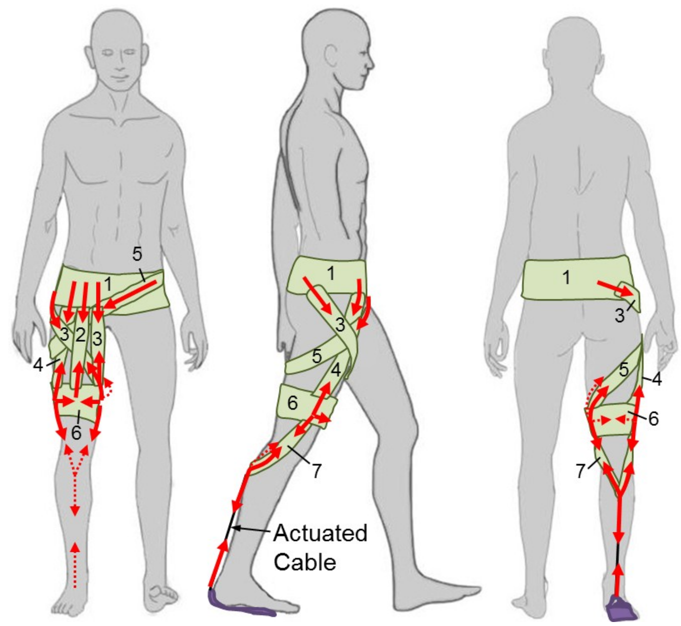

Fig. 3. Left, several views of the forces in the suit as it is actuated. Dotted lines indicate forces in straps on the obscured side of the leg. The cable sheath is attached to and pulls down at the bottom of straps (7), and pulls up at the back of the boot.

A key feature of the exosuit is that forces are generated in the webbing both passively due to the natural kinematics of walking and actively from the Bowden cable contracting; thus mimicking the action of the biological limb. Referring toFig. 2, it can be seen that the hip flexor muscles are active when the hip is maximally extended. Due to the fact that the suit is located primarily at the front of the hip, it will become taut and thus generate a resistive moment as the hip extends maximally, absorbing power which can later be returned. The suit tension is enhanced when the cable is actuated in a manner similar to how biological muscle stretches pre-stressed tendons. When the hip flexes, the suit becomes slack and is transparent to the wearer. Similarly, the suit tenses when the ankle is dorsiflexed, and actuating the cable provides additional force. The suit becomes slack when the ankle is plantarflexed. Due to this construction, the suit can be made to apply beneficial moments purely passively during walking if it is adjusted properly.

C. Suit Design Principles

The suit was designed to create a path to transfer loads between the ankle and the pelvis. As such, high effective exosuit stiffness (resultant of fabric and the series compliance due to the interface to the wearer) is required in order to transfer power to the body efficiently from the actuator instead of putting energy primarily into stretching the suit. Moreover, high suit stiffness is beneficial because it corresponds to lower suit displacements during actuation, and thus less risk of chafing. If the stiffness is high enough, the suit designer can place other elastic elements in series with the suit, thereby having control over the effective exosuit stiffness. The suit construction follows several principles to provide a comfortable, functional device.

1) Terminate on bony parts of the body

First, the top of the suit is terminated on the pelvis because it is a bony part of the body with a relatively thin layer of skin coverage. When the skin is compressed against the bone during the suit's operation, the displacement is low as compared to other parts of the body covered by thicker muscle or fat, leading to high stiffness. The shins and shoulders also have thin layers of muscle over the bone, but these were not chosen because pulling on the shoulders during preliminary human subjects experiments was uncomfortable and forced the wearer to hunch over, and the shin orientation makes it difficult to create the upward forces needed at the heel.

On the suit, strap (5) transfers loads to the suit from the opposite iliac crest of the pelvis bone. This geometry is important to prevent the suit from sagging. The center area of the abdomen does not have any rigid elements to support vertical loads, so if strap (5) connected to the center of the waist belt, it would pull down several centimeters during loading.

2) Load the body normally as much as possible

Second, the bulk of the load is transferred to the pelvis to load the body normal to the skin and bones as much as possible. Shear forces against the skin will cause slipping of the suit and chafing if the friction force with the skin is exceeded [32]. It is impossible to avoid shear forces entirely throughout the suit, as vertical loads at the bottom of the suit will cause some displacement along the length of the leg. Some displacement is possible, though, without ill effect if there is a large amount of tissue over the bone, and the skin is not stretched taut. In this scenario, the skin and underlying muscle and fat can move back and forth over the bone under relatively low forces where the thicker the layer of tissue, the more displacement is possible. For example, when a person is standing vertically, the skin can deflect up or down several centimeters at the thigh without discomfort. The exosuit displaces less than 3 cm at the thigh under loads of up to 200N at the ankle, and so does not slip or cause chafing there. The lack of slipping is aided by the normal force against the skin from the knee strap (6), which increases the friction force needed there before slipping occurs. Lower on the suit, some motion of the webbing does occur, around the knee and over the calf. In these regions, the normal force is small and the motion is not problematic. Further protection from chafing can be accomplished by wearing a close-fitting knee-length garment under the suit that the suit can slide over.

3) Preload the suit against the body

The lack of slipping at the thigh is also facilitated by a third principle; preloading the suit against the body. When donning the suit, first the knee straps (6) are tightened and the belt (1) put around the waist. Then, the vertical straps between the knee straps and waist belt (2-5) are tightened. This pulls up on the knee straps and down on the waist belt when standing vertically, causing the thigh skin to displace 1-2cm upward. This preloading is possible because the thigh expands in diameter going upward from the knee, while the iliac crest of the pelvis becomes larger in the downward direction; this wedge effect limit suit motion. With the thigh muscle and fat initially pulled slightly (1cm) vertically, the skin can displace down a larger distance downward before it reaches its travel limit and leads to suit slippage or chafing. On the waist, this preloading compresses the skin and fat against the pelvis somewhat and prevents the suit from riding upward on the pelvis, thereby increasing the resulting stiffness.

4) Minimize normal pressures on the body

Fourth, the suit makes use of wide straps throughout to minimize normal pressures on the body and minimize displacement. The skin can accommodate a certain amount of normal pressure before discomfort or lack of blood flow occurs. Estimates of the maximum comfortable normal pressure are typically around 0.5 N/cm^2 [32, 33]. Furthermore, the body will compress if forces are applied normally, such as occurs over the thigh with our suit as straps (2)-(5) push inward against it (see Fig. 3 (center)). To maximize suit stiffness, this displacement must be minimized, which can be achieved by having wide contact areas. The suit stiffness also increases if it is actuated while the underlying muscles are tensed, which causes them to become stiffer and compress less. Ideally, the suit will assist muscles that are firing, which will result in this behavior. At the back of the calf, the straps (7) extend low on the leg to bypass the bulk of the calf muscle. Having the straps higher results in the calf compressing when the suit is actuated, which is uncomfortable and results in a lower suit stiffness.

5) Other factors

In addition to minimizing normal pressures from wide straps, pressures can be distributed through equalizing the tension in various parts of the suit. During the suit donning process, straps (2)-(5) are adjusted iteratively until they have around equal tension, which we have found maximizes both comfort and performance.

Finally, the suit should not restrict muscles from expanding during operation. In our suit, the knee strap is made from a stretch cotton material to permit the thigh muscles to bulge when they contract. Our experiments have shown that a non-extensible strap is uncomfortable there.

D. Suit Implementation

Three pictures of the front, side, and back of the exosuit are shown in Fig. 4. As shown in the front view, the suit begins with a waist belt (1) which sits on the iliac crests of the pelvis. The waist belt is constructed of rip-stop nylon and tensioned by two polyester 1" webbing straps that are above and below the iliac crest. The top strap supports loads while the bottom strap provides snugness and prevents lateral motion. This top strap supports almost the entire downward force of the suit, which is then spread out across the leg through the rip-stop nylon and straps attached to it.

Attached to the waist belt are four 2" polyester straps per leg, (2)-(5), extending down to a strap above the knee (6) that is positioned with the bottom edge 3-4 cm above the top of the patella. Straps (2)-(5) are sewn onto the waist belt, loop through slides (buckles) on the knee strap, and then attach back on themselves (2,4,5) or to the waist belt (3) with 2"-wide industrial strength hook Velcro. This hook Velcro mated with standard 2" loop Velcro. Although the industrial strength hook Velcro performs better than a standard hook, it still displaces several millimeters in shear under loads due to the length of the loops. Using the Velcro on the vertical webbing straps, the suit can be adjusted to fit individuals from 1.75 to 2.00 meters tall.

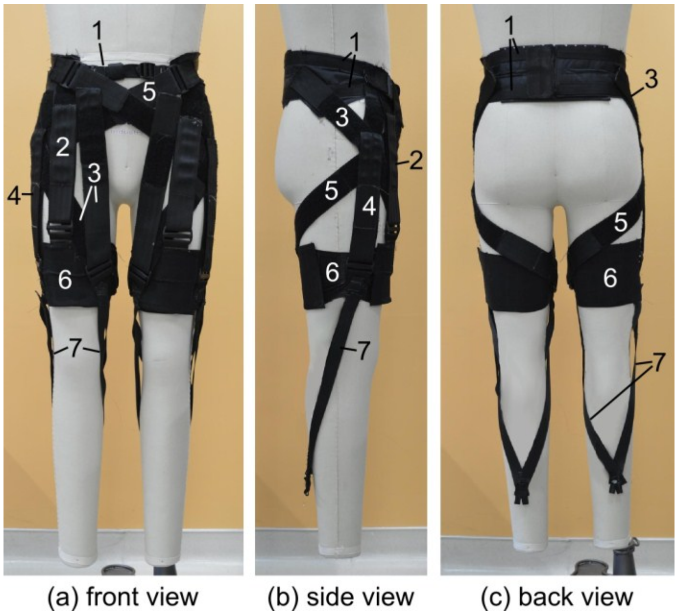

Fig. 4. Several views of the exosuit. Labels correspond to those in Fig. 3.

Strap (2) holds up the center of the knee strap (strap 5). Strap (4) provides a direct path from the calf straps (7) to the side of the waist belt, and holds up the outside edge of the knee strap. Straps (3) and (5) support the inside edge of the knee strap, with strap (3) crossing the front of the thigh and strap (5) going behind the leg to support it evenly. Adjusting the relative tension of these straps allows the wearer to control the position of the knee strap rotationally around the leg. Straps (3,4,5) are sewn together where they intersect on the side of the hip, to distribute the forces from each of them. Strap (3) comes down from the side of the waist, loops through the buckle on the knee strap, then continues vertically to terminate on strap (5).

III. Portable Actuation Unit

The Bowden cable extends up the wearer's leg and to an actuator module carried on the wearer's back. This module contains Maxon motors, Copley motor controllers, a PC/104 computer, and batteries. Just below the motors is another module which contains a sensor to measure the cable tension. At the ankle and foot, the webbing behind the heel connects to webbing straps attached to the back of the boot heel. Footswitches are placed inside the shoes to measure heel strike and toe off. The mass of the cloth exosuit is 1.07 kg, the footswitches have a combined mass of 0.44 kg, and the actuator/control system and backpack frame have a mass of 10.64 kg, for a total system mass of 12.15 kg.

A. System overview

A diagram of the initial portable actuator and control hardware is shown in Fig. 5, with a detailed list of components and their respective masses in Table I. The system was not optimized for mass or volume, but rather designed to provide flexibility to explore a range of gait assistance strategies.

B. Cable-actuation unit

The Bowden cable ends are pulled by actuator units, and the cable then passes through a force sensor module and continues down the leg. Each actuator unit consists of a motor and 111:1 gearbox connected to a secondary shaft via 1:1 gearing. The secondary shaft contains a pulley (radius=2.45cm) capable of winding up the cable one rotation, which corresponds to 15cm of cable travel. This range allows for some pre-tensioning of the cable to ensure a snug fit before it pulls to provide assistance during walking (corresponding to a typical cable travel of 5-6cm).

TABLE I. Mobile Actuation System Components

Component	Mass
Batteries to power motors *5000 mAh of LiPo batteries to power system for 4 hours*	1.97 kg
Batteries to power computer *4000 mAh 14.8V LiPo battery to power system for 5 hours*	0.42 kg
Motor Controllers (2) *Copley Accelnet ADP-090-36*	0.86 kg
Direct Drive Actuator Units (2) *Maxon EC 4-pole 30 200W, 24V motor with 111:1 gearbox, connected spool with radius 2.45 cm*	2.27 kg
PC/104 Computer *Diamond Systems Aurora PC/104*	0.53 kg
Frame *ALICE pack frame, aluminum and acrylic structure*	2.72 kg
Force sensor modules (2) *Phidgets 3135 50kg Micro Load Cell, Futek CSG110 amp.*	0.87 kg
Other *Connectors, switches, etc.*	1.00 kg
Total	**10.64 kg**

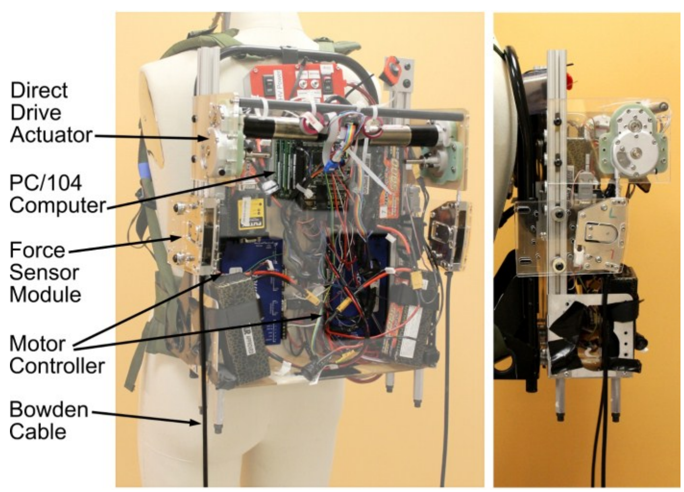

Fig. 5. Left, mobile actuation system used with the exosuit with PC/104 and Copley Accelnet motor controllers highlighted. Right, side view of system showing actuator and force sensor module.

C. Force-sensing unit

Below the actuator unit, along the cable, is a force sensing unit. This contains an idler pulley connected to one end of a cantilever-style force sensor. The cable glances off the pulley, changing angle by θ=8 degrees on each side and causing a force on the pulley. The force sensor has strain gages, and measures the pulley force as it causes shear in the force sensor. The force sensor is located next to the motor to reduce wiring down the body. This location means that the force at the ankle is masked by the efficiency of the Bowden cables, but with high efficiency cables (Nokon Part# KON05020), a good estimate of the force at the ankle is still achieved.

IV. EXOSUIT EVALUATION

The system was designed to be intrinsically safe in several ways. The maximum force that the suit can apply to the person, under normal operation, is <30% of the maximum ankle moment present during normal walking. The travel range of the cables was limited mechanically by the spool and Bowden cable ends, and the velocity was limited by the no-load speed of the motor followed by a large gear reduction. Finally, in case of undesired operation, the stoppers (ferrules) at the ends of the cable act as mechanical fuses, coming off consistently at 600-650N and preventing any further force transfer to the wearer.

A. *Suit stiffness characterization*

The suit stiffness was characterized by positioning a subject in two poses, tensioning and releasing the suit, and simultaneously recording the forces and displacement. Forces were recorded by a load cell at the ankle (Omega LC201). Loading and unloading were performed with a ramp profile, over a period of 3 seconds each, and the displacement of the motor was recorded. The results are shown in Fig. 6.

As seen in Fig. 6, the suit exhibits a nonlinear stiffness and significant hysteresis. The suit stiffness increases to 7350 N/m or 5400 N/m depending on the subject's pose (stepping forward or standing upright with legs parallel). As seen in the lower portion of the figure, with the cable retracted from 1 cm to 9 cm, the suit itself displaces 4.7cm downward while the boot attachment displaces 2.0cm upward and the cable system stretches 1.0 cm.

In these trials, the energy contributed to the suit during loading was 7.9J and 12.0J for the legs parallel and stepping forward cases, respectively. The hysteresis in the suit caused 59.7% and 53.8% of the energy in the suit to be lost for each test actuation cycle, respectively.

B. *Human walking trials*

For human subjects experiments, a preliminary control system was implemented that could function during steady state walking. A position-controlled trajectory for the cable was programmed, which was transformed into a force at the ankle through the compliance of the suit. The position trajectory was pre-computed as a function of gait percentage based on the joint angles of the normal walking cycle and an estimated suit stiffness, to generate an estimated force at the ankle. The controller was then tuned in realtime while a person was wearing the system to correct for modeling errors. The controller is triggered from a signal from the footswitch indicating that heelstrike has occurred, and then the position trajectory is played back as a function of time.

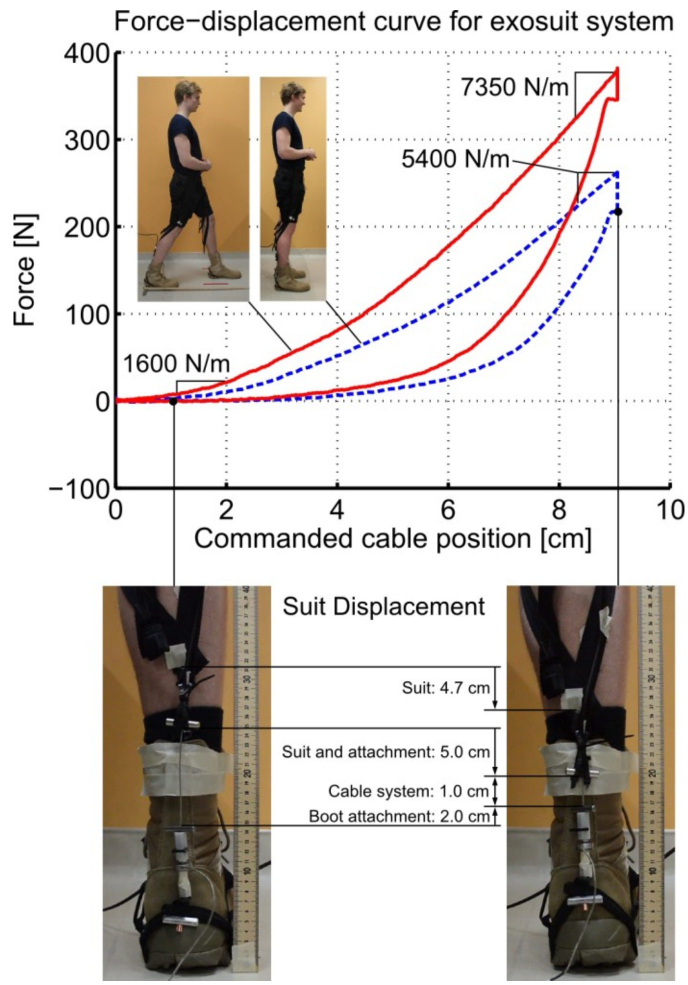

Fig. 6. Top, stiffness curves for the exosuit, when the wearer is positioned with both legs together (blue curve) and one leg 60cm in front of the other (red curve). Bottom, a depiction of how much the suit and boot attachments displace during actuation. An additional load cell is used at the heel to measure the forces.

During walking, forces and displacements of the cable were recorded. A graph of the cable displacement and force as a function of gait percentage are shown in Fig. 7. In this figure, the force is observed to begin increasing at around 22% in the gait cycle, although the cable is still at zero displacement. This occurs because the suit naturally tensions due to the motion of the ankle and hip. Once the cable is actuated, here at 30% of the gait cycle, the force increases rapidly with cable displacement. During walking trials, the stiffness of the suit was observed to be close to 225N/5.5cm = 4090N/m, which corresponds closely to the stiffness measured in the angled-leg standing test with a displacement of 5.5cm and small preload.

The power input and output of the system were measured, as well. For each leg, the motor used 17.6 Watts on average over a gait cycle of 1.16 seconds, for a total energy of 20.4 Joules. The direct drive actuator unit output 12.3 Joules, and the ankle received 7.0 Joules for an overall efficiency of 39.7%. As seen in Fig. 8, the power at the ankle is 56.4% of the power at the output of the actuator unit. The losses come from the Bowden cables, which were inefficient due to the cables being worn and bent.

The moment arms of the suit at the ankle, knee, and hip were 8cm in front of the hip, 1cm behind the knee joint, and 8cm behind the ankle, respectively. With a force of 200N, for example, the suit will generate moments of 16 N-m at the hip and ankle, and a moment of less than 2 N-m at the knee. Compared to normal walking data [31], this corresponds to 12% of the ankle moment and 20% of the hip moment. In other trials, the suit has been able to produce peak forces of over 300N—corresponding to 18% of the ankle moment and 30% of the hip moment.

I. Conclusions and Future Work

In conclusion, we have demonstrated a soft exosuit capable of applying forces to the body during walking. The suit can apply moments to the hip and ankle of at least 18% of the normal human walking moments, enough to be noticeable to the wearer. Our experiences have shown that if the timing of actuation is adjusted poorly, the wearer can feel the suit resisting them, and one or more of their muscles tire quickly. If the timing and suit tension are adjusted well, the suit can feel like it is helping the wearer, lifting the heel and helping the leg swing. More studies need to be performed to determine the optimal timing for suit actuation and the accompanying best suit architecture.

The suit is extremely light, minimizing distal mass, and adds negligible inertia to the legs. The suit feels comfortable and does not cause chafing, even though it moves with the skin a small amount during actuation. The suit also does not constrain any of the degrees of freedom in the legs, and permits the wearer to move through their full range of motion.

The current suit stiffness is relatively low, which sets bounds on the forces that can be applied for a given motor power. It also is nonlinear, increasing substantially after several centimeters of displacement. As such, control schemes can benefit from fast motor motions to remove slack in the system, followed by power transfer at higher stiffnesses. The maximum forces that the human body can tolerate via the suit are unknown; measurement of the pressures applied by the suit is needed. One risk of high forces is that they create increased suit displacements, which will lead to chafing. Based on our experience with the suit, we estimate that chafing will become a problem at forces above 250-300N, unless suit stiffness can be increased further.

There are many areas for future improvement of the system. Measurements must be made of the suit's benefit, including metabolic and electromyography measurements. The actuator unit must be substantially lightened to create a unit that does not increase the wearer's metabolic rate due to the additional load, and must be packaged into a smaller form factor. We estimate that the entire system can be constructed with a mass of only 5kg, including batteries. Finally, further tuning of the control system is needed to understand how it can benefit the body most. Additional work is needed to determine the most effective strategies for aiding human locomotion, including schemes that assist the load-acceptance phase of the gait, instead of push-off as currently.

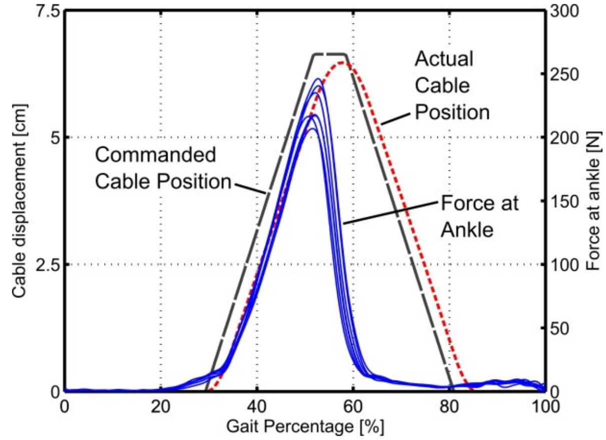

Fig. 7. Cable position and force at the ankle while walking seven steps. The actual cable position is computed by the motor controller, taking into account acceleration limits of the motor.

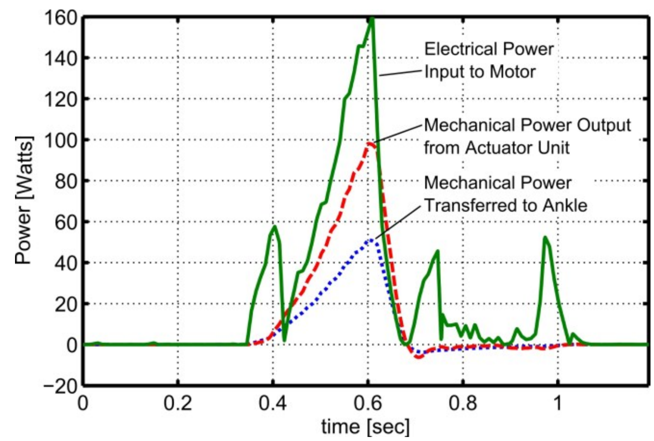

Fig. 8. Powers input to motor, output from direct drive actuator unit, and applied to the ankle during a walking trial.

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