Multi-joint Soft Exosuit for Gait Assistance

Alan T. Asbeck[1,2], Kai Schmidt[1,2], Ignacio Galiana[1,2], Diana Wagner[1,2], and Conor J. Walsh[1,2]

***Abstract*— Exosuits represent a new approach for applying assistive forces to an individual, using soft textiles to interface to the wearer and transmit forces through specified load paths. In this paper we present a body-worn, multi-joint soft exosuit that assists both ankle plantar flexion and hip flexion through a multiarticular load path, and hip extension through a separate load path, at walking speeds up to 1.79m/s (4.0mph). The exosuit applies forces of 300N in the multiarticular load path and 150N in hip extension, which correspond to torques of 21% and 19% of the nominal biological moments at the ankle and hip during unloaded walking. The multi-joint soft exosuit uses a new actuation approach that exploits joint synergies, with one motor actuating the multiarticular load paths on both legs and one motor actuating the hip extension load paths on both legs, in order to reduce the total system weight. Control is accomplished by an algorithm that uses only a gyroscope at the heel and a load cell monitoring the suit tension, and is shown to adapt within a single step to changes in cadence. Additionally, the control algorithm can create slack in the suit during non-level-ground walking motions such as stepping over obstacles so that the system can be transparent to the wearer when required. The resulting system consumes 137W, and has a mass of 6.5kg including batteries.**

I. INTRODUCTION

Many lower-limb assistive devices have been developed to assist or enable human walking in various forms. Some exoskeletons have been designed to make load carriage easier by providing a parallel structure to ground [1]–[3], while others apply torques to the wearer's body directly in order to assist impaired [4] or able-bodied [5]–[10] individuals. For most of these devices, the engineering goal is to make it easier for the wearer to move, comparing wearing the system to not wearing the system. Traditionally, rigid exoskeleton frames have been used in these assistive devices. With a rigid frame, the systems can apply very high torques to the body [8] or support super-human payloads with an external structure [11].

However, for devices that aim to augment the wearer's joint torques directly, rigid frames may restrict natural movement or may apply parasitic torques to the body if they are imperfectly aligned with the wearer's biological joints [12]. Alternately, self-aligning mechanisms may be bulky and heavy [12], [13]. Rigid exoskeletons may also have large inertias which can further hinder motion if the wearer is imperfectly tracked by the control system. Adding mass to a person is counterproductive since it increases the metabolic effort required to walk, particularly if the mass is located distally. For example, mass carried on the waist increases a wearer's net metabolic rate during walking by 1-2%/kg, while mass at the feet raises it 8-9%/kg [14].

Recently, several exosuits composed of soft textiles have been developed in an attempt to mitigate against some of these effects. These both interface to the body and transmit tensile forces through specified load paths over the body with textiles, using the bone structure of the body to support compressive forces across the joints. Exosuits have been made to support lifting tasks [15], [16], aid with grasping via an actuated glove [17], [18], and assist with walking [19]–[24]. Exosuits are particularly suited to assisting locomotion since they have extremely low inertias and intrinsically create torques centered at the biological joints. In addition, they can be worn under an outer layer of clothing since they are very low-profile. Several potential applications of exosuits exist, including providing gait rehabilitation, permitting first responders or hikers to carry loads with less energy, and enabling impaired individuals to regain mobility. However, several technical challenges must be solved in order to achieve a practical device: the exosuit must create sufficient joint torques while having a low total system mass; the applied forces must be consistently synchronized with the wearer's gait; the exosuit must not be restrictive during other motions; and it must be comfortable if worn for long periods of time.

In this paper, we present the design and evaluation of an exosuit and portable actuation system, shown in Figure 1(a), that achieves several advances toward this goal of a practical assistive device. The exosuit is an integration of the multiarticular ankle/hip suit and monoarticular hip suit described in [22]–[24], combined with improvements to each part. The exosuit assists walking by creating forces in two load paths, one that aids ankle plantarflexion and hip flexion (multiarticular load path), and one that aids hip extension (monoarticular load path). These load paths were chosen since the ankle and hip are the major power contributors to level-ground walking [25]. By applying torques to both the ankle and hip, we expect that the biomechanics of walking will be better maintained as compared to applying torques to either joint in isolation. The suit's architecture and construction enable high forces to be applied to the multiarticular ankle/hip flexion load path (300N) and the hip extension load path (150N), which correspond to 21% and 19% of the nominal biological torques at the ankle and at the hip during level-ground walking, respectively. To reduce the carried mass, we developed a new actuation approach in which one motor actuates the multiarticular load path on both legs, and a second motor actuates the hip extension

[1]Harvard University, School of Engineering and Applied Sciences, Cambridge, MA 02138, USA

[2]The Wyss Institute for Biologically Inspired Engineering, Cambridge, MA 02138, USA. Email: aasbeck@gmail.com, walsh@seas.harvard.edu

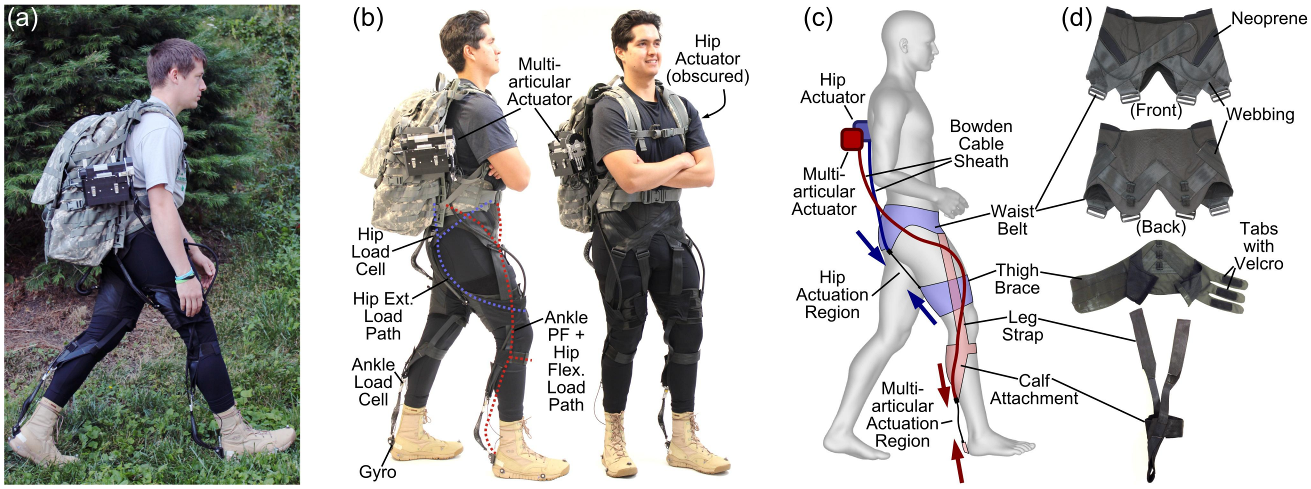

Fig. 1. (a), Photograph of the system in use during walking outside. Actuators are mounted on the sides of an empty backpack, and the exosuit is worn from the waist down. (b)-(d), Load paths and components of the suit. The suit includes load paths to actuate hip extension (monoarticular) and the combination of ankle plantarflexion and hip flexion (multiarticular). Load cells are mounted where the Bowden cable sheaths connect to the suit, and a gyroscope is mounted on each heel.

load path on both legs. Finally, we demonstrate a robust control strategy that utilizes minimal sensory information, yet actuates the suit consistently and in synchrony with the wearer during level-ground walking and makes the exosuit slack (non-restrictive) during other motions. In the following sections we describe the design of the multi-joint soft exosuit and present some experimental results with the system worn during outdoor walking.

II. SUIT DESCRIPTION

Several views of the exosuit presented in this paper are shown in Figure 1. The exosuit builds on the designs of the multiarticular exosuit described in [22]–[24] assisting ankle plantarflexion and hip flexion, as well as the hip extension exosuit described in [22].

The exosuit consists of a spandex base layer, a waist belt wrapping around the pelvis, a thigh brace and calf attachment on each leg, and two vertical straps per leg connecting the calf attachment to the waist belt. These leg straps attach to buckles in the waist belt at the top of the thigh, and pass through channels in the thigh braces so they are held close to the body. The calf attachment secures around the front of the wearer's shin just below their knee, and includes a triangular piece of fabric covering their calf. The system attaches to the wearer's shoe with a metal bracket that bolts onto the back of the heel.

Connecting to the textile exosuit are Bowden cables which provide a flexible transmission from the proximally-mounted actuators. Discussed further in Section III, one actuator powers the multiarticular load path on both legs and one actuator powers hip extension on both legs. To create hip extension torques, one Bowden cable sheath per leg connects to the back of the waist belt, while the inner cable extends further to the back of the thigh brace (see Figure 1(c)). A second Bowden cable per leg powers the multiarticular load path. The sheath of this cable connects to the suit at the back of the calf, and the inner cable extends further to the shoe attachment. The textiles in the suit are patterned to route the force in this load path over the front of the hip, through the center of the knee, and behind the ankle, thereby creating hip flexion and ankle plantarflexion torques simultaneously when tension is created in this part of the suit [22]–[24].

In general, the load paths of the exosuit create forces on the body in parallel with the wearer's muscles, so that if forces are created in the suit at appropriate times, the wearer's muscles should adapt to the assistance and decrease their activation, letting the suit do some of the work instead. Previous work has shown that when assistive devices are worn, typically the body adapts and decreases the muscle forces so that the total joint moment (suit plus human) is the same as the original biological joint moment [26], [27]. If this occurs, wearing the exosuit may actually decrease the axial loading on the bone structure, since the exosuit is offset at larger radii from the joint centers of rotation than are the underlying muscles. Because a joint moment is the product of the force and radius, to achieve a given joint moment the suit may require a smaller axial force than that created solely by the muscles.

The suit design presented here includes a number of improvements over previous work. The waist belt is constructed to be a single piece, securing in the front with Velcro. While this means that the belt only fits waist sizes within a 10cm range, it is easier to don and stretches less due to Velcro than previous versions. The belt also includes neoprene inserts over the iliac crest of the pelvis to provide cushioning and force distribution there.

The front of each thigh brace has three independently-adjustable Velcro-covered tabs along its height, enabling it to be secured to the wearer's leg in a conformal manner. The back of each thigh brace is reinforced with seatbelt webbing angled in an inverted "V"-shape, with the apex at the connection to the inner cable. This transfers force to the front of the leg with minimal displacement of the textile.

The multiarticular load path is also improved in several ways as compared to previous versions. At the front of the waist belt, the vertical leg straps are connected closer to the sides of the leg than they were previously to reduce the displacement of the tissue there. The overall load path now includes a strap in front of the shin, instead of going directly to the back of the calf. This is beneficial because the front of the shin is bony and hard, reducing suit displacement, and the shin strap additionally holds the webbing in the correct location with respect to the knee. Finally, the architecture of the calf attachment helps resist downward motion of the suit by gripping the calf muscle in conjunction with the straps extending up to the waist belt.

The suit uses a minimal number of sensors, with load cells (Futek #LSB200) mounted at the ends of the Bowden cable sheaths where they connect to the suit, and gyroscopes (ST #LPY503AL) mounted at the heels.

III. ACTUATION APPROACH

The ankle is actuated in plantarflexion (plus hip flexion through the multiarticular load path) and the hip actuated in extension because these are the portions of the gait cycle during which the largest amount of positive power is created by the body. During level-ground walking, the ankle has a peak power of 3.7 W/kg at around 53% in the gait cycle (which extends from one heel strike to the next), while the hip has a peak power of 0.8 W/kg at around 12% in the gait cycle [25]. These periods of peak power correspond to the times when the body is transitioning from one leg to the other: the ankle on the trailing leg propels the body upward and forward, while the leading leg accepts the weight of the body and redirects its momentum. The remainder of this section discusses how our system powers ankle plantarflexion, hip flexion and hip extension with a single actuator per leg, and presents the hardware used to accomplish this.

A. Joint Synergies

In order to reduce the mass carried by the wearer, it is desirable to minimize the number of motors used to actuate the four load paths (multiarticular and hip extension load paths for each leg). Previously, Bowden-cable-driven lower-body exosuits have used one motor per load path. In this case, each motor retracts the cable to create joint torques and feeds it out again to create slack in the suit. This permits perfect control over the cable during all stages of the gait cycle, but can require a large number of motors as the number of controlled load paths increases.

However, synergies in timing between the joints of each leg can be used to reduce the number of motors actuating an exosuit. In particular, during walking the legs are 180 degrees out of phase. While there is some overlap in the stances period of the legs, usually one leg is in stance while the other is in swing; the leg uses different muscles–and hence has different joint torques–during each part of the gait cycle. This can permit one motor to be used for the same joint in both directions (e.g. hip flexion and hip extension), one motor to be used for two different joints on the same leg (e.g. hip extension and ankle plantarflexion), or one motor to be used for the same joint on both legs (e.g. right ankle plantarflexion and left ankle plantarflexion). In our case, due to the timing of when the joints are actuated, it is necessary to use one motor for the same load path on both legs. A key feature of exosuits that enables this behavior is that the suit becomes slack and non-restrictive if the actuated segment is lengthened.

The scheme of using one motor to operate two load paths is illustrated in Figure 2(a). If a spool is used with cables wound around it in each direction, as the spool rotates one way, one of the cables will become shorter and create force in the suit while the other will extend and create slack. The cable that is fed out will protrude away from the body a small amount, as can be seen in Figure 1. In our exosuit, this leads to the cable routing shown in Figure 2(b), where each actuator has one Bowden cable that goes largely straight down the leg and a second cable that crosses the body to the opposite leg.

To understand how the motor moves to create forces in both legs, first consider the actuator trajectory and resulting force if there were one actuator per leg, shown in Figure 2(c). This example uses the force pattern for the multiarticular load path, as explained previously in [22]–[24]. This single-leg trajectory remains at a certain value, the "Tension Position," for most of the gait cycle, only changing position between 46-70% in the gait cycle in this example. The force present on the leg from 25-46% in the gait cycle is passively induced due to body changing its pose while the actuator remains at a fixed position, so the suit becomes stretched over the body. From 46-70%, the force increases dramatically due to the actuator's shortening the suit length.

Figure 2(d) explains how one motor can be used to actuate both legs. In the middle panel, if the motor rotates to have a positive position, the cable pulls on the right leg, creating the force shown in the top panel. Conversely, if the motor rotates to a negative position, force is created in the left leg.

Several additional features on the graph are important to note. First, the forces on the right and left legs are non-overlapping and have a small transition period in between when no force is applied to either leg. This space is necessary because the motor needs to have time to switch between legs.

Second, there is a substantial amount of cable that must be reeled in between the legs, shown by the position difference between the "Right Tension Position" and "Left Tension Position." In order to make the exosuit non-restrictive during standing and other non-walking motions, both legs must have a small amount of slack in them, which is the offset of each Tension Position from the zero centered position. The switching time between the force profiles is necessary so that the motor can reel in this slack in the suit.

While the specifics are different, the hip actuation scheme is very similar to that shown here for the multiarticular load path, except that the force profile extends from approximately 0-30% in the gait cycle.

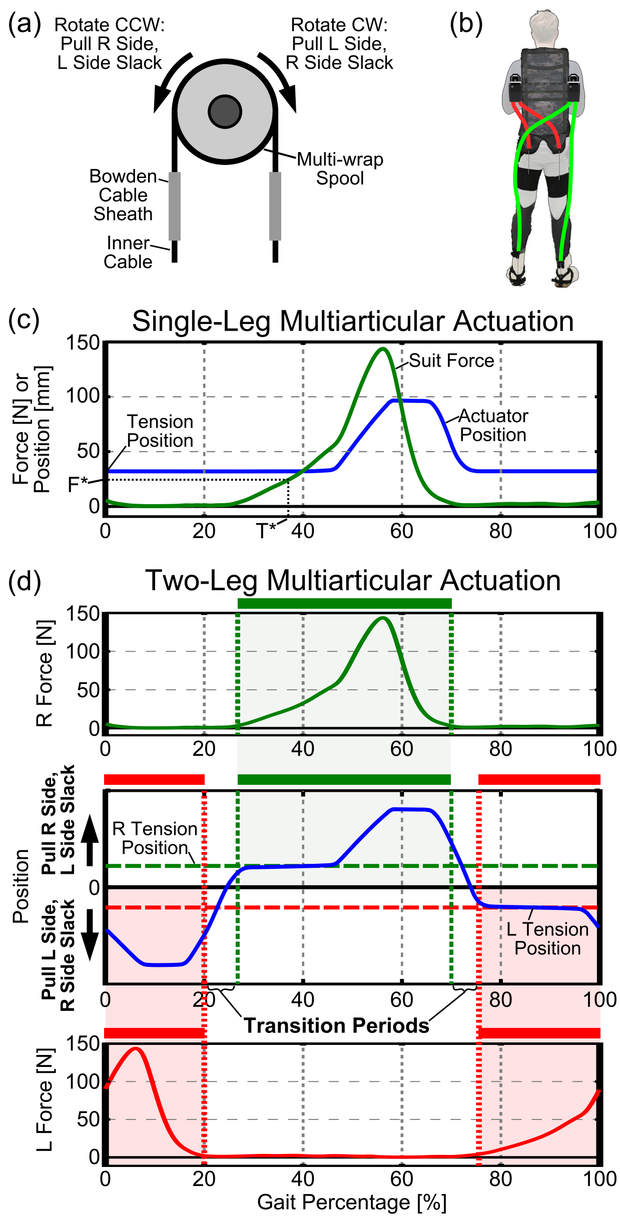

Fig. 2. Actuation scheme for the system. (a) Each actuator consists of a multi-wrap spool with two Bowden inner cables wrapped around it to form a pull-pull actuator. (b) Routing of cables over the body. (c) Actuation scheme for one actuator powering a single multiarticular load path. The gait cycle (GC) extends from one heel strike to the next. The actuator is held at an initial "tension position" in order to generate force in the suit due to the body's motion (25-46% GC in this example). When the suit tension exceeds F*, the wearer is assumed to be at T* in the gait cycle. The actuators then move starting at 46% GC (in this example) to generate additional force in the suit. (d) Switching actuation scheme in which one actuator powers the multiarticular load path for both legs; here the graphs are in terms of the right leg's gait cycle. The actuator is able to power both legs because force is only present in the suit from 25-70% GC, which is less than half of the gait cycle, so the actuator has a small amount of time to switch between legs.

B. System Calculations

Previous studies reported measurements of how the suit interacted with wearers when actuation was applied [22]–[24]. Tension Positions of approximately 1-3cm were found to be required at the ankle and hip to achieve a full range of motion, and varied slightly depending on the subject's anatomy. Also, initial experiments indicated that pull amplitudes of 4-6cm were necessary to create 300N in the multiarticular load path and 150N in the hip extension load path. Thus, switching between the peak of the pull and the Tension Position of the opposite leg requires a travel of $\Delta_{position}$ = (Pull Amplitude) + 2(Tension Position) which is 8-12cm for both load paths.

The minimum gait period for subjects walking at 1.79m/s (4.0 mph) was measured to be 0.9 seconds. With the multiarticular force profile in Figure 2 extending from 25% to 70% in the gait cycle, this allows only 5% of the gait cycle (or 45 msec) for each switching phase. However, it was noted that the actuator could reverse direction earlier than is shown in this figure without causing significant changes in the force profile. One contributing factor to this was that the motor takes time (30ms) to switch direction during which time there is relatively little actuator motion. Having the actuator reverse direction earlier permitted the multiarticular load path to have up to approximately 20% of the gait cycle (180ms) for each transition between the legs. The nominal hip force profile only extends for 30% of the gait cycle, which gives the actuator 20% of the gait cycle to switch legs.

Transmitting forces of 300N and 150N to the multiarticular load path and hip extension load path respectively during the pull phase requires linear cable velocities of 0.5-0.6m/s and 1.0m/s, respectively. For comparison, the actuators require a speed of 1.0m/s to switch between legs in the worst case, traveling a distance of 12cm in 120ms (180ms total, minus 30ms for acceleration and 30ms for deceleration). It is convenient that the two required velocities are the same for the hip load path and relatively close for the multiarticular load path, so a gearbox that can produce the necessary cable speeds for switching is also fairly optimized for the pull.

C. Actuation Units: Mechanical Design

The mechanical design of the actuation units is shown in Figure 3. Both the multiarticular and hip actuator units had identical mechanical designs except for the gearbox. In each unit, a Maxon EC 4-pole 30, 200W motor (part #305013) was connected to a gearbox (79:1 for the multiarticular, 33:1 for the hip) which drives a 3cm radius multi-wrap pulley. The pulley module is removable for ease of replacing cables, and so the sensor and actuator cables could remain with the exosuit if a wearer wanted to disconnect from the actuator units. Each motor was controlled by a Copley Accelnet Module motor controller which plugged into a custom circuit board. A removable battery unit clipped onto the bottom of each actuator unit, containing two 4-cell Li-Po batteries (5200mAh for the multiarticular actuator, 3900mAh for the hip) wired in series. In total, the system weighed 6.5kg

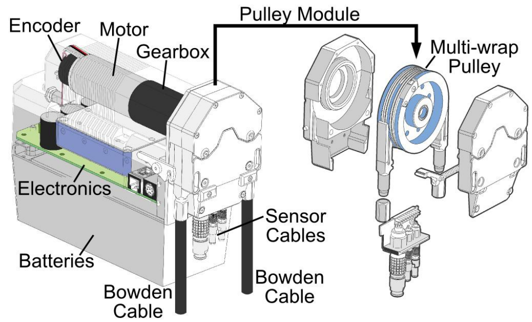

Fig. 3. Diagram of the actuator units.

including the textile exosuit, and consumed 65W in the multiarticular load path and 72W at the hip during walking at 1.79m/s (4.0mph).

The motor chosen had a no-load speed of 16,700rpm. A 79:1 gearbox was chosen for the multiarticular unit in order to maximize the torque; the motor was overdriven to 20,000rpm during switching in order to cover the switching distance quickly. Combined with the spool radius, this produced a maximum linear cable speed of 0.795m/s which was slightly less than the desired 1.0m/s; as such, the multiarticular units required a slightly smaller Tension Position to switch in the allotted time at high walking speeds. The hip unit used a 33:1 gearbox, which produced a linear cable velocity of 1.59m/s at the no-load speed. The higher speed was chosen to permit increased slack in the suit, which sometimes occurred if suit components migrated on the body over time.

The multi-wrap pulley needed more than one complete wrap of cable because in the worst-case situations, the cables required 9cm of travel on each side, and additional wraps were used to create friction with the spool and minimize force on the cable termination. Particular attention was paid to the design of the multi-wrap pulley so that the cable did not jam between the pulley and case. With a cable diameter of 1.3mm, the tolerance between the pulley and case was set to 0.3mm, which is less than a quarter of a cable diameter. Even with this close tolerance, it was found that the cable had a tendency to jam between the pulley and case at times. To eliminate this phenomenon, spring tensioners were added to each end of the Bowden cables. Compression springs were placed around each inner cable after it exited the sheath, pulling the inner cable out with respect to the sheath with a force of 2-6N. These were covered by a black mesh material in order to protect the wearer from the springs, as can be seen in Figure 1.

IV. CONTROL

The goal of the controller was to create the simplest possible system that still effectively created forces in synchrony with the human body. The force profiles used were chosen to mirror the biological joint torques, with the assumption that this would enable the muscle activation to be reduced proportionally.

A. Sensors

To sense the body's motion, only two different sensors were utilized: a gyroscope at each heel, and a suit tension sensor (a load cell) at each location where a Bowden cable sheath attaches to the suit (at the hip and at the calf). The gyroscope was used to measure when the foot made contact with the ground. More specifically, after a person lands on their heel during walking, their foot rotates forward until it lies flat on the ground. This motion was detected by the gyroscope, which showed a consistent peak in the signal at 4% in the gait cycle that was easily identified. The load cells were used to monitor the tension in the suit across each joint, which was proportional (in conjunction with the radius from the joint) to the torque being applied to the joint.

B. Zero Step Delay and Hip Control

A force-based position control system was used to drive the actuators. A motor position trajectory induces forces in the suit because the human tissue over which the suit rests is compliant, and the suit material itself can stretch. The combination of these, which we refer to as the "suit-human series stiffness," is a repeatable force-displacement function that appears to the motor as a series stiffness [24]. An example position trajectory is shown in Figure 2(d): the motor position starts increasing at a specified gait percentage, reaches its maximum at a later specified gait percentage, holds this position until a third gait percentage, and then decreases at the maximum velocity until it reaches the Tension Position of the opposite leg. While this scheme generates consistent force profiles for a given walking speed, a force-based feedback loop must be used since human walking varies over time. Furthermore, the forces must be correctly aligned with the biological torques within each step.

To accomplish this, we use an algorithm we call "zero step delay" control since it can detect the wearer's gait period accurately within a single step. This is accomplished by using two sensor measurements within a single gait cycle. The first of these is the gyroscope peak, which is known to be at 4% in the gait cycle. The second measurement is a suit tension measurement in the multiarticular load path due to the wearer moving into a pose that corresponds to just before push-off during walking. In particular, this pose is with their hip extended (knee toward the rear) and their ankle dorsiflexed (toe pointed up). When the wearer moves into this pose, with the actuators held at a constant position, force is passively induced in the suit, as shown in Figure 2(c). This passively-induced force is monitored, and when it exceeds a certain threshold (25N), that time is assumed to be 36.5% in the gait cycle. In Figure 2(c), this force threshold is labeled "F*" and the corresponding assumed time in the gait cycle is labeled "T*". With these two points in the gait cycle known, the gait period can be calculated and used to time-scale the position trajectory of the motor.

To maintain the appropriate forces over time, the Tension Position and pull amplitude are adjusted after the step is complete, using the subsequent heel strike gyroscope reading to generate the true step period. If the force threshold F* was

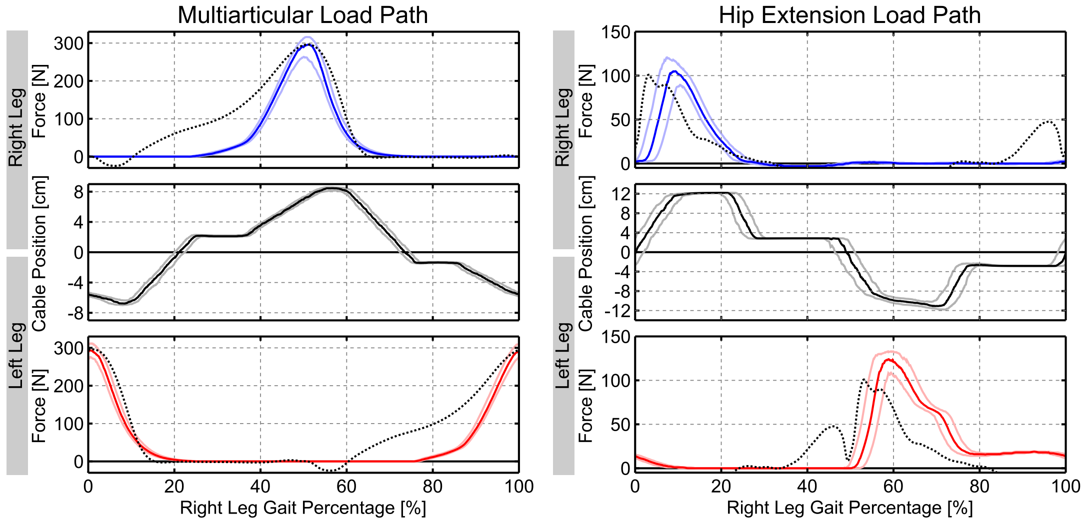

Fig. 4. Forces and actuator positions for the hip and multiarticular load paths during overground walking at 1.6 m/s (3.6mph). In each plot of the forces, the three solid lines are the 25th, 50th, and 75th percentiles of the force, computed per instant in the walking cycle. The dotted lines are the biological torque for the ankle and hip joints, respectively, with the amplitude scaled so that the peak matches the peak of the right leg's force. This is shown so the general shape and timing of the applied forces can be compared to the biological moment. The forces for the right and left leg are both plotted in terms of the gait percentage of the right leg, so the left leg forces are shifted 50% from the usual convention. In the plot of the force for the left hip, the non-zero force shown from 80-100% in the gait cycle is an artefact of lateral force on the load cell, and this was not applied to the wearer.

reached at a time before 36.5% in the gait cycle, the Tension Position is reduced by a fixed amount (1mm), while if it is reached after 36.5% the Tension Position is increased by that amount. Similarly, the pull amplitude is adjusted over time to maintain a desired peak force level by recording the peak force and examining it after the step is complete, then increasing or decreasing the pull amplitude by 1mm.

Actuating the hip in extension also relies on the gait percentage estimate from the zero step delay algorithm, but from the opposite leg. The leading leg's hip is actuated just after the trailing leg's multiarticular load path is actuated, since the leading leg contacts the ground shortly after the trailing leg begins the push-off motion. From the perspective of the trailing leg, the hip actuation profile begins at 46.5% in the gait cycle. This time is just after the passive pretension is detected in the trailing leg (36.5%), so the gait period information derived from it is fairly accurate even if the wearer is accelerating or decelerating. As such, the hip of the leading leg is actuated from 46.5-71.5% in the gait cycle of the trailing leg. The pull amplitude adjustment algorithm is also applied to the hip so that the force magnitudes remain consistent.

C. System Architecture

The electronics of the system are based on the Arduino Due circuit, with additional circuitry added including the load cell amplifiers and CAN transceivers. The microcontroller (MCU) communicates to the motor controller using the CANopen protocol, sending positions that the motor should move to at a specified velocity; the motor controller is responsible for the low-level motor control. In addition to the load cells for the multiarticular load path, the gyroscopes are connected to the multiarticular actuation unit since it implements the zero step delay algorithm. The hip unit is connected to the load cells for the hip extension load path so that the hip force level can be monitored. Since the hip unit's actuation timing is based on information from the zero step delay algorithm, the multiarticular unit broadcasts the gait information to the hip unit over a CAN bus.

V. RESULTS

A. Overground Walking

Human subjects testing of the system was approved by the Harvard Medical School Institutional Review Board. Results from the system in operation during outside walking at 1.6m/s (3.6mph) for three minutes on one subject are shown in Figure 4. The path followed by the wearer was primarily level paved segments but also included grassy areas. To explore how the system could assist with load-carriage applications, the wearer carried a load of 25.3kg plus the system mass (6.5kg). In each plot in Figure 4, the darker solid line indicates the median force or position, while the lighter solid lines are the 25th and 75th percentiles. As can be seen, the system applied consistent forces that are very well-aligned in time. The rising edge of the hip assistive forces can be seen to have a larger variation than the multiarticular forces. This is due to the fact that the multiarticular pull is initiated immediately after the suit tension measurement (36.5%) while the hip pull is initiated after a delay of 10% in the gait cycle, during which time the cadence may have varied slightly.

Also shown in the plots of the force are dotted lines, which are the nominal biological torque profiles for the ankle and hip during level-ground walking with a 25.3kg load. These are scaled so that the maximum amplitude matches the right leg's peak force amplitude, and are shown so the timing of the nominal biological torques can be compared with the timing of the assistive force profiles. Notably, the peak of the multiarticular force profile coincides with the biological ankle force peak. The peak of the hip profile is later than the

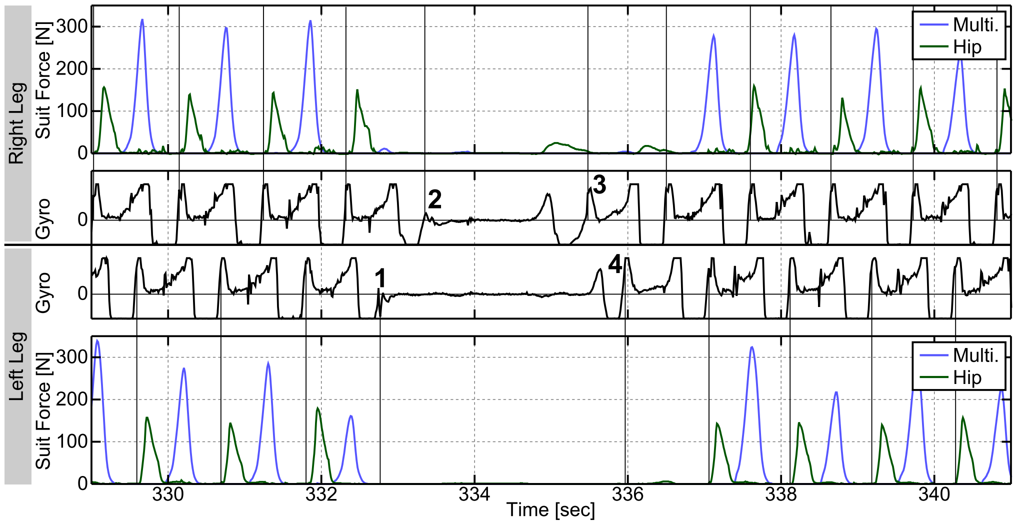

Fig. 5. Forces and gyroscope readings for both legs during a walking sequence in which the operator was walking at 1.25 m/s (2.8mph) on a treadmill, stepped off the treadmill, and then stepped back onto it and continued walking. "Multi." is the force in the multiarticular load path, "Hip" is the force in the hip load path. Solid vertical lines indicate heel strike times. (1) Left foot steps off treadmill, (2) Right foot steps off treadmill, (3) Right foot steps back onto treadmill, (4) Left foot steps back onto treadmill.

biological peak, but this was found to be more comfortable empirically. However, further testing is likely required to determine if a different force profile at the hip may be more beneficial than the one used here. Over the course of this test, the system correctly detected all of the steps taken by the wearer. In other tests for longer periods of time (>30 min) over similar terrain, with more than 8 different wearers, the system typically failed to detect 1-2 steps out of every 1000.

B. Zero Step Delay Controller Behavior During Stop-Start Motion

A key feature of the zero step delay controller is that it initiates or terminates forces within a single step. This is demonstrated in Figure 5, which shows the system in operation during steady-state walking at 1.25 m/s (2.8mph) on a treadmill followed by the user stopping and then starting to walk again. The plot shows the force measurements on both legs as well as the gyroscope readings. In this plot, heel strike events are indicated by thin solid vertical lines for the corresponding foot. At point 1, the wearer steps off of the treadmill with their left foot. During the preceding step of the left leg, the system detected the passive force in the suit and applied actuation to the multiarticular load path, but the kinematics of the wearer had changed such that the peak force was much lower than usual. On the right leg, the hip is actuated one final time with a normal amplitude (hip pulse directly above point 1). A very small passive tension force on the right multiarticular load path can also be seen immediately after this, but the force does not cross the 25N threshold and so the system does not actuate. At point 2, the wearer steps off the treadmill with their right leg. At point 3, the wearer steps back onto the treadmill with their right leg. During this step, again a very small pretension force can be seen, but due to the wearer's unusual kinematics of stepping back onto the treadmill, this force is not large enough to initiate actuation. On either side of point 3, small (30N) forces can be seen on the right hip. This is due to the hip actuator stopping in a position that created small forces passively on the right leg. At point 4, the left leg steps onto the treadmill again. After this, the right leg has normal kinematics, and so the passive force crosses the threshold and actuation resumes as normal.

VI. DISCUSSION AND CONCLUSIONS

In conclusion, we have presented a body-worn multi-joint soft exosuit capable of robust and long-term operation in natural outdoor environments. The actuation scheme described here reduced the carried actuator weight substantially compared to prior designs, and can apply consistent forces to the wearer. The switching operation of the actuators functions well if the suit is positioned properly on the wearer, although it may present difficulties if the suit changes position, thereby increasing the cable-travel distance required to transition between legs. In extreme cases, the peak forces may be reduced or the shape of the force profile may be altered if the motor cannot reach the opposite leg fast enough. The control scheme presented here was found to be able to consistently detect when it is beneficial to apply forces, and can rapidly accommodates changes in the wearer's speed. The system was tested with more than 20 individuals on a treadmill, and found to provide consistent forces to each wearing at a variety of walking speeds. In the future, we intend to increase the assistive force magnitudes, and perform human subject studies with biomechanical and physiological analysis to determine the optimal force profiles for each joint. Since the system is relatively light and creates substantial joint moments without disrupting the wearers natural biomechanics, we expect that it will be able to create a net metabolic reduction (system worn vs. no system). Finally, while subjects have worn the suit for an hour of continuous walking and found it to be comfortable, the long-term effects of wearing the suit still need to be determined.

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