

Soft Exosuit for Hip Assistance

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Abstract

Exoskeletons comprised of rigid load-bearing structures have been developed for many years, but a new paradigm is to create “exosuits” that apply tensile forces to the body using textiles and utilize the body’s skeletal structure to support compressive forces. Exosuits are intended to augment the musculature by providing small to moderate levels of assistance at appropriate times in the walking cycle. They have a number of substantial benefits: with their fabric construction, exosuits eliminate problems of needing to align a rigid frame precisely with the biological joints and their inertia can be extremely low. In this paper, we present a fully portable hip-assistance exosuit that uses a backpack frame to attach to the torso, onto which is mounted a spooled-webbing actuator that connects to the back of the user’s thigh. The actuators, powered by a geared brushless motor connected to a spool via a timing belt, wind up seat-belt webbing onto the spool so that a large travel is possible with a simple, compact mechanism. Designed to be worn over the clothing, the webbing creates a large moment arm around the hip that provides torques in the sagittal plane of up to 30% of the nominal biological torques for level-ground walking. Due to its soft design, the system does not restrict the motion of the hip in the ab- and adduction directions or rotation about the leg axis. Here we present the design of the system along with some initial measurements of the system in use during walking on level ground at 1.25 m/s, where it creates a force of up to 150 N on the thigh, equivalent to a torque of 20.5 Nm to assist hip extension.

Keywords: exosuit, wearable robotics, walking assistance

1. Introduction

A large number of lower limb exoskeletons have been developed over the years, usually with the purpose of assisting or augmenting human walking. For individuals needing to carry heavy loads such as soldiers or recreational backpackers, the promise of a mechanical assist is welcoming, potentially reducing muscle fatigue, metabolic expenditure, or injury rates. For individuals needing assistance with walking such as the elderly or paraplegic, or patients requiring gait rehabilitation, such devices could potentially restore walking function.

Many previous exoskeletons have comprised of rigid load-bearing structures, designed to transmit forces to the ground while tracking or apply-

ing torques to the wearer’s joints at points along the structure. Some of these use the exoskeleton to support the weight of a hiker’s backpack [1, 2, 3, 4], while others support a paraplegic’s body-weight [5, 6, 7, 8, 9, 10].

An alternative approach that has been applied to assist walking is to apply torques to the user’s joints in parallel with the musculature, but without transmitting a load to ground. In particular, devices have been constructed to augment the strength of the hip and knee in healthy individuals [11, 12]. Many other such devices are powered orthoses [13, 14, 15] that are designed to support disabled individuals by providing assistance at both the hip and knee to compensate for reduced muscle strength. These systems are also rigid, and may require an ankle-foot orthosis to be used with them to provide stability for the ankle joint and prevent migration of the rigid components that can often have significant weight. Two previous devices for

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assisting the hip in flexion used pneumatic actuators [16, 17, 18] along with rigid plastic orthoses to secure to the leg and pelvis such that a pin joint was approximately colocated with the hip. Several other devices have been made to provide assistance to the ankle alone, both for plantarflexion and dorsiflexion [19, 20]. These units offer the potential to increase power generation capability at the ankle to support forward propulsion in addition to ensuring sufficient toe clearance from the ground during swing.

A more recent rigid exoskeleton is the Honda Walking Assist Device with Stride Management System [21, 22], which actuates the hip joint in both flexion and extension. This device consists of a hybrid soft-rigid waist belt and struts that extend partially down the thigh, where they attach with narrow straps. It is powered by a motor approximately colocated with the wearers hip joint in the sagittal plane and includes a passive joint just above the motor to permit small amount of hip ab-/adduction. It functions differently from the previously-discussed examples, in that it only applies small forces and serves primarily to regulate the cadence of the wearer. Humans are known to walk most efficiently at a certain cadence (steps per minute) for a given forward velocity, and the Honda system applies small torques to the hips to entice the user into walking at that cadence.

As an alternative to rigid exoskeletons, a new paradigm is to create “exosuits” that are comprised of fabrics and utilize the body’s skeletal structure to support compressive forces. Exosuits are used to augment the musculature by providing small amounts of assistance at crucial times in the walking cycle, as opposed to being able to apply large torques or support significant weight. These systems have a number of substantial benefits: with their fabric construction, exosuits eliminate problems of needing to align a rigid frame precisely with the biological joints and their inertia can be extremely low. These two features virtually eliminate resistance to motion, thus permitting close to natural kinematics. Furthermore, exosuits can be light and sleek, permitting them to be worn constantly; thus enabling rehabilitation or strength augmentation to occur for longer periods of time throughout the day. However, while these systems do offer potential benefits over rigid exoskeletons, there are limitations related to controllability and maximum applied force and thus much research remains to characterize them and understand their capabilities.

ties.

While the concept of an exosuit is relatively new, a number of designs have already been developed. Wehner et al. [23] created a pneumatically-powered system that assists ankle plantarflexion, and uses a garment of cloth and webbing to secure to the legs, waist, and shoulders. Asbeck et al. [24] developed a Bowden-cable-driven biologically inspired exosuit that assists both ankle plantarflexion and hip flexion with a single actuator for each leg. This is achieved with a multi-articular exosuit architecture that transmits forces from the waist and thigh through the knee and to the back of the calf, where it connects to the heel via the Bowden cable. Kawamura et al. [25] made a pneumatically-powered exosuit to assist hip flexion, with the device attaching to the wearer with a wide waist belt and a brace that crosses the knee. In addition to assisting walking, several other exosuits have been developed that support the back and torso while the wearer is lifting heavy loads [26, 27, 28].

The system described in this paper is a new example of a soft exosuit that uses the body’s bone structure to support compressive forces and does not provide kinematic restrictions to the joints. The design assists hip extension, and is shown in figure 1.

In this paper, we focus on the design of the exosuit. In the following sections, we present an overview of the system, model how it applies torques to the body, and compute the requirements for the actuators. We describe and perform analysis for a new spooled-webbing actuator. We also present preliminary measurements of the system in use during walking on level ground, demonstrating its functionality.

2. System overview

Our design (figure 1) utilizes a backpack frame to attach to the torso, and uses geared motors to retract webbing ribbon connected to a thigh brace on each leg. The actuators wind up the webbing ribbon onto a spool, which permits large (>25 cm) travel with a simple mechanism. The bottom of figure 1 shows how the device functions during the walking cycle. The actuators retract the webbing just before heelstrike, which in conjunction with the leg’s motion creates forces pulling the thigh back. The force peaks just after heelstrike, concurrent with the wearer extending their hip to support

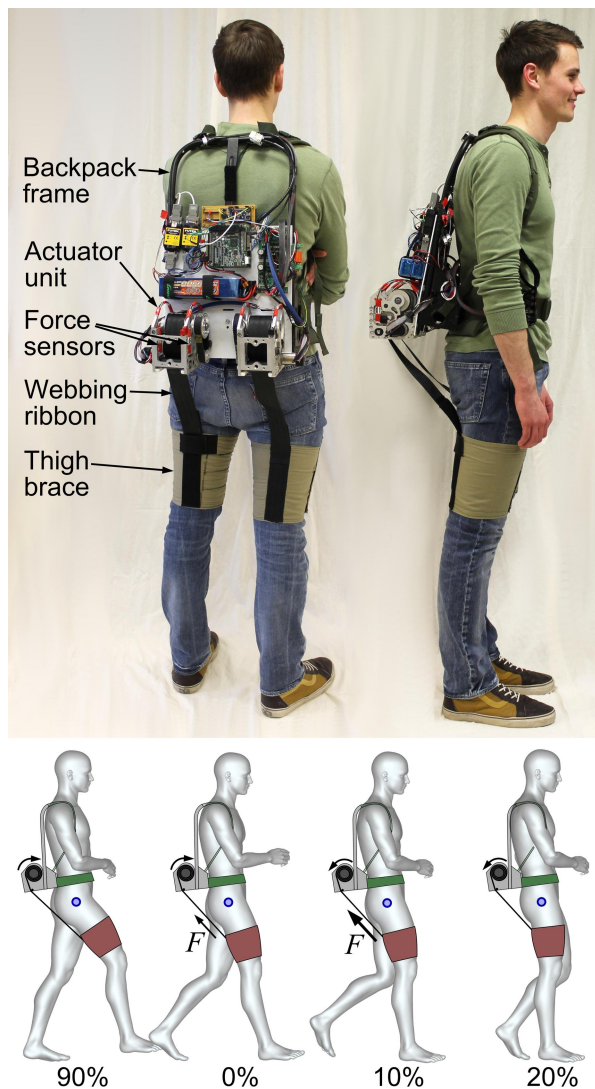


Figure 1: Top, exosuit to assist hip extension described in this paper. Two actuator units are mounted on a backpack frame, and connect to cloth thigh braces with webbing. Bottom, overview of device operation. Starting at 90% in the gait cycle, which extends from one heelstrike to the next, the actuator units retract the webbing. This produces a force on the thigh that increases until 6% in the gait cycle. The actuator units then spool out the webbing so the force decreases, reaching zero at 20% in the gait cycle. At other times in the gait cycle, the webbing is slack so that no force is applied to the leg.

themselves on the newly-planted leg. The actuators then extend the webbing again, so that there is no force in the webbing for the majority of the gait cycle.

This actuation scheme is unique in that the ribbon is wound on top of itself, increasing the spool diameter as the ribbon retracts. Mechanisms that wind or unwind a thin ribbon from a spool with an actuator have been used in several applications previously, including retracting seat belts, reading audio and video magnetic tapes, mopping floors, and deploying drip system tubing [29, 30, 31, 32]. For comparison, many devices wind up a thin cable around a spool, but the cable typically travels down the length of the spool as it winds, instead of solely winding on top of itself as it does in our actuators. We use webbing instead of a thin cable so the device is comfortable if the wearer is seated or if their leg is at large ($> 90^\circ$) angles in flexion. In these scenarios, a narrow cable would dig into the body, while the 2" wide webbing will press in only a small amount due to its much larger contact area. The webbing further forms a wide attachment to the thigh brace, helping distribute forces around the thigh.

The webbing path between the backpack and thigh brace creates a large moment arm around the hip, providing torques in the sagittal plane of up to 30% of the nominal biological hip torques for level-ground walking. A key feature of this soft design is that it does not restrict the motion of the hip in the ab- and adduction directions or in rotation about the leg axis. The webbing also permits the force to be measured inside the actuator unit, since it extends directly from it to the thigh attachment. Measuring forces internally permits the unit to be compact and robust, with no wiring or sensors outside of the actuation units. The system also includes two footswitches that are used to estimate the wearer's cadence and phase through gait.

The prototype device is capable of applying forces to the leg of up to 150 N at walking speeds up to 1.79 m/s (4 mph). The device utilizes the backpack frame to transmit torques to the torso, pushing forward and downward on the lower back with forces up to 188 N and pulling back on the shoulders with forces up to 50 N. The entire device weighs 7.57 kg including batteries to last for 2.8 hours of use. The bulk of the weight is on the torso, with each thigh brace providing minimal inertia to the leg with a weight of 0.17 kg.

3. Calculations

In this section we present a simple model of the system and perform calculations to determine the necessary velocity of the ribbon at the motor in order for the actuators to achieve sufficient forces necessary to apply a scaled version of the nominal hip moment during walking on level ground at 1.43 m/s (3.2 mph). It should be noted that these values depend on the compliance of the body and suit in addition to the leg's motion.

3.1. System Modeling

A diagram of the system is shown in figure 2. This shows the backpack frame worn on the user's torso and the webbing ribbon extending downward to the thigh brace. Within the leg, a number of springs are shown, representing the compliance of the tissue between the surface of the leg and the underlying bone. The muscle, fat, and other tissue in the leg will compress as they transfer forces to the femur bone, which is the only rigid component in the leg. Additional springs are shown at the waist and the shoulder, where the body also has some compliance. We model the total effective stiffness of all of these locations as they interact with the system together as the stiffness k_{body} . The compliance of the ribbon itself can be ignored since it is comparatively inextensible, with a stiffness of 600,000 N/m for a 25 cm length (Young's modulus = 2.41 GPa).

The top right of the diagram in figure 2 shows a geometric model of the wearer's leg and how the exosuit attaches to it. The pivot point of the hip in the sagittal plane is shown by a large black dot. The leg extends downward from this point at an angle θ_{hip} with respect to vertical (shown in the figure as positive), and the leg bone forms an angle ψ with respect to the webbing ribbon. The pivot point of the hip is assumed to be horizontal from the bottom of the actuators where the webbing exits, and the distance between these two points is denoted a . The distance between the hip pivot point and the point at which the ribbon would intersect the femur bone, if it continued through the leg tissue, is denoted l . As the user walks, the perpendicular moment arm r between the ribbon and the hip joint will vary due to θ_{hip} changing. The force on the webbing, F , and the moment arm r , combine to form the moment exerted on the hip, M_{hip} .

The bottom right of the diagram shows the distances related to the ribbon travel. In each case,

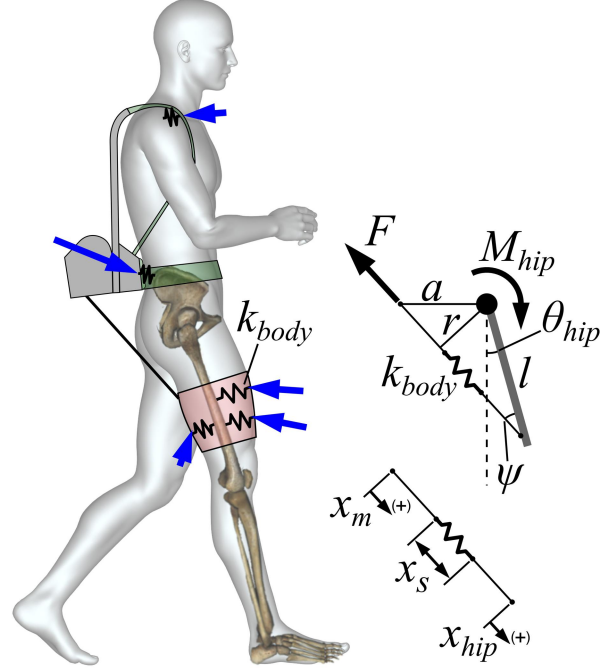


Figure 2: Left, diagram of person wearing the system, showing the forces exerted on the wearer by the device. The wearers body compresses at the thigh, waist, and shoulder, the behavior of which is represented by the lumped stiffness k_{body} . Right, model used to calculate the required motor displacement in order to apply a specified moment M_{hip} at the hip as the hip moves.

these distances are along the ribbon, which will change orientation as the wearer walks. x_m is the position of the ribbon at the actuator, with positive out of the actuator (downward and to the right in the figure). x_{hip} is the position of the ribbon at the virtual point where it connects to the femur, also with positive downward and to the right, with $x_{hip} = 0$ when the hip is furthest in extension. In between, x_s is the length of the simulated spring representing the compliance of the leg tissue k_{body} . While in reality the leg tissue is compressing, this is equivalent to an extension spring extending backward from the femur, and it is drawn as such in the model for simplicity. For each of these positions, we ignore constant offsets since we are only concerned with the net travel of the tissue and actuator.

The length of ribbon between the actuator and femur is defined as:

$$l_{ribbon} = \sqrt{a^2 + l^2 - 2al \cos(\pi/2 + \theta_{hip})} \quad (1)$$

The travel of the ribbon needed to follow the

hip motion (x_{hip}) is a function of this distance. Subtracting $\sqrt{a^2 + l^2}$ so that $x_{hip}(\theta_{hip} = 0) = 0$, we get:

$$x_{hip} = l_{ribbon} - \sqrt{a^2 + l^2} \quad (2)$$

To calculate the external moment M_{hip} that the hip exosuit is capable of applying to the person, it is necessary to determine the angle ψ between the thigh and the webbing.

$$M_{hip} = rF \quad (3)$$

$$r = l \sin(\psi) \quad (4)$$

$$\psi = \arccos\left(\frac{l^2 + l_{ribbon}^2 - a^2}{2l l_{ribbon}}\right) \quad (5)$$

$$\Rightarrow F = \frac{M_{hip}}{l \sin(\psi)} \quad (6)$$

Beyond modeling the geometric changes as the hip moves during walking, the stiffness of the body must be taken into account. Intuitively, in addition to tracking the hip's motion, to apply forces the motor must retract the ribbon additionally to compress the thigh tissue.

Experiments discussed in section 3.2 found that the body stiffness can be approximated by a second-order equation:

$$F(x_s) = p_1 x_s^2 + p_2 x_s \quad (7)$$

For a force to be applied to the webbing, it must displace the suit-human spring k_{body} :

$$F = p_1 x_s^2 + p_2 x_s \equiv \frac{M_{hip}}{l \sin(\psi)} \quad (8)$$

$$\Rightarrow x_s = \frac{-p_2}{2p_1} + \sqrt{\left(\frac{p_2}{2p_1}\right)^2 + \left(\frac{M_{hip}}{p_1 l \sin(\psi)}\right)} \quad (9)$$

Finally, the motor position x_m can be calculated from the sum of the hip motion and the spring motion. According to the defined coordinate frame, the motor position x_m that is required to track and apply forces to the hip can be calculated by:

$$x_m = x_{hip} - x_s \quad (10)$$

We use this modeled motor position to determine requirements for the actuator units, and as a basis for generating motor trajectories in practice.

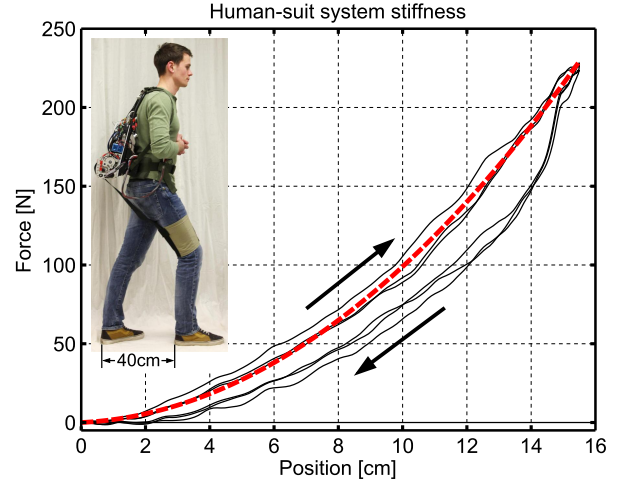


Figure 3: Measured stiffness of the suit-human system, determined by retracting the webbing while holding the hip in a fixed vertical position and measuring the force at the actuator unit. Three cycles of a sine wave actuation profile are shown with thin black lines. Also shown is a quadratic line of best fit to the rising edge (thicker dashed line).

3.2. Measurements of Suit-Human Stiffness

To determine the actual value of the lumped stiffness k_{body} , a set of experiments were performed using the assembled device. This and other human-subjects experiments in this paper were approved by the Harvard Institutional Review Board. For the protocol, the wearer stood with the measured leg forward with their feet 40 cm apart as shown in the inset in figure 3, and the actuator was commanded to retract the ribbon 16.5 cm and then release it following a sinusoidal profile at 0.5 Hz. The ribbon displacement was determined experimentally so that a force of approximately 200 N would be applied to the wearer. This was repeated three times, and the results plotted in figure 3 with an offset removed where no force was applied to the body.

The rising portion of the curve (corresponding to when the actuator was pulling) was fitted to a quadratic equation, also plotted in figure 3, resulting in the line of best fit:

$$F(x_s) = p_1 x_s^2 + p_2 x_s \quad (11)$$

$$p_1 = 8899.7, \quad p_2 = 99.546 \quad (12)$$

where x_s is in meters and F in Newtons.

3.3. Simulations

Using the model of the system and the measured body stiffness, we next compute the required actuator travel in order to apply a scaled version of the nominal hip moment to the body.

We begin with the nominal biological sagittal hip moment for level-ground walking, duplicated from [33]. This is plotted in figure 4(top) with respect to the percentage through the gait cycle, which extends from one heel strike to the next for a given foot. The biological moment begins slightly before heel strike to decelerate the leg, and peaks just after heel strike to prevent the leg from collapsing as it accepts the weight of the body. We next form an approximated hip moment which is smoothed relative to the nominal moment and is limited to positive (extensor) torques, also shown in figure 4(top). This curve was constructed by selecting seven points along the nominal moment profile to match the starting point, peak, end point, and several relevant points in between, then interpolating and smoothing with a low-pass filter. It differs slightly from the nominal moment between 90-100% in the gait cycle to avoid rapid changes in force which an actuator would have difficulty following, and because a smooth moment profile will feel more comfortable to the wearer. Since the exosuit is designed to apply a partial assistance torque to the wearer, we then scale the smoothed approximation by 31% for use in subsequent calculations. This scaling corresponds to a peak force of 145 N at the leg for a 76 kg individual. When the smoothed approximation is scaled by 31%, it is close in magnitude to the unscaled biological moment between 90% and 100% in the gait cycle. We use this scaled approximation as M_{hip} in our model. Using this along with the nominal biological hip angle (not shown), also from [33], and plugging them into equations 2-8, we generate positions x_{hip} , x_s , and x_m , with offsets removed so that x_{hip} reaches 0 when the hip is furthest back and x_s is zero when no force is applied. These are shown in figure 4(middle). In these calculations, we use $a = 0.32$ m, $l = 0.266$ m, and we assume a gait period of 1.15 seconds which was empirically measured with a subject walking on a treadmill at 1.25 m/s (2.8 mph).

In figure 4(middle), the hip position x_{hip} starts at a large positive value at heel strike (the leg is forward, in flexion) and decreases until around halfway through the gait cycle when the leg is furthest back, in extension, and then increases in value again. If the motor were to perfectly track the hip's motion,

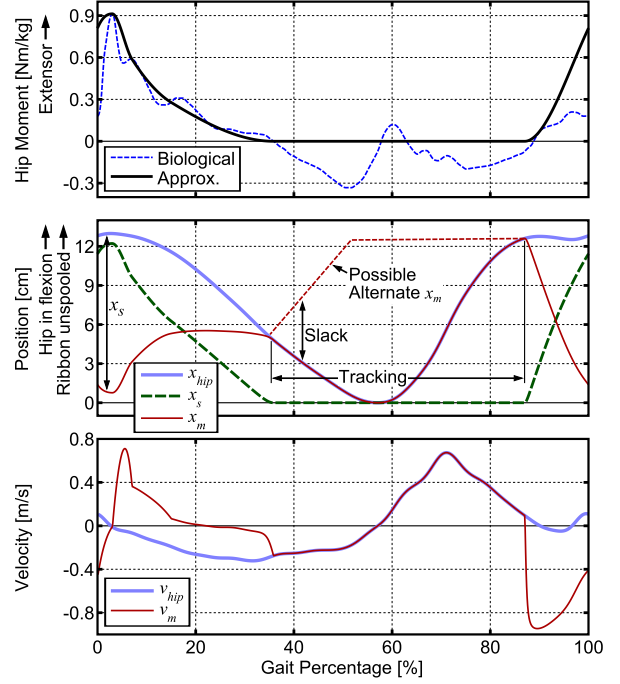


Figure 4: Top, biological hip moment profile vs. approximated profile, which is used for the simulation after scaling by 31%. Hip extension moments are positive, which corresponds to the muscles pulling the leg toward the posterior. Middle, resulting position profiles of the hip (x_{hip}), spring (x_s), and motor (x_m) during ground level walking. Here flexion positions are positive, which corresponds to the leg angled towards the front of the body or the motor spooling out the ribbon. The spring position is plotted with the spring extending being positive. Bottom, hip speed (v_{hip}) and motor speed (v_m) that is necessary both to track and to actuate the hip. All graphs are shown as a function of percentage through the gait cycle, which extends from one heel strike to the next.

applying no force to the ribbon, then the motor trajectory would follow this profile as well. Instead, the motor must retract the ribbon additionally (more negative values of x_m in the plot) to generate force across the body's compliance. As indicated on the far left of the plot, the difference in height between the hip position x_{hip} and the motor position x_m is the amount the spring is extended (x_s). In the center of the gait cycle, when the applied moment is zero, the simulation shows the motor tracking the hip ($x_m = x_{hip}$), as indicated in the figure.

In addition to the solid lines in the figure, a possible alternate path for the motor position is shown in a red dashed line. This line is above the hip position, which corresponds to slack in the ribbon. Slack in the ribbon guarantees that the force is

zero during 35-85% in the gait cycle; if the actuators were tracking the leg imperfectly, antagonistic forces could be applied to the leg during that time. When operating the device in practice, we use a motor trajectory similar to this that creates slack in the ribbon.

In figure 4(bottom), the resulting velocities of the ribbon at the hip (v_{hip}) and motor (v_m) are plotted. The motor velocity at around 90% in the gait cycle shows that a speed of 0.95 m/s is required to compress the body's compliance and apply the specified moment to the hip. For comparison, if the body was extremely stiff, the actuators would have to follow a path very close to that of the hip itself, and thus have a velocity of v_{hip} . To track the hip's motion, this would require a speed of 0.67 m/s, which occurs at around 70% in the gait cycle when no force is on the ribbon. During the period when force is applied to the ribbon, the maximum speed of the leg is only 0.32 m/s, a value much slower than 0.95 m/s. In summary, a major effect of the body's compliance is to increase the required motor speed for applying forces by more than a factor of two.

One interesting point is how to optimize the positioning of the ribbon with respect to the hip joint. The simulation assumed relatively large distances between the actuator and the hip joint ($a = 0.32$ m) and the distance down the leg ($l = 0.266$ m). It is interesting to consider what would happen if these distances were varied. This analysis is illustrated in figure 5, which shows the peak force F and peak motor velocity v_m required to achieve the approximated moment profile in figure 4 as distances a and l are varied.

According to the measured stiffness data, the body's compliance requires 12 cm of ribbon travel to achieve 145 N of force. This displacement would likely be similar if the distances a and l were changed, since the combined stiffness of the leg, waist, and shoulder is largely independent of the strap's positioning on the leg. For small values of a and l , in the bottom left corner of the plots, larger ribbon forces are required because the moment arm about the hip has been reduced. The motor velocities are also increased, because the larger forces require additional displacement of the body's compliance, and it must occur in the same amount of time as before. The motor would then be much faster than needed in order to track the hip's motion, which would be slower by a factor of two due to the smaller moment arm. Thus, smaller values of a and l result in a higher peak motor power and

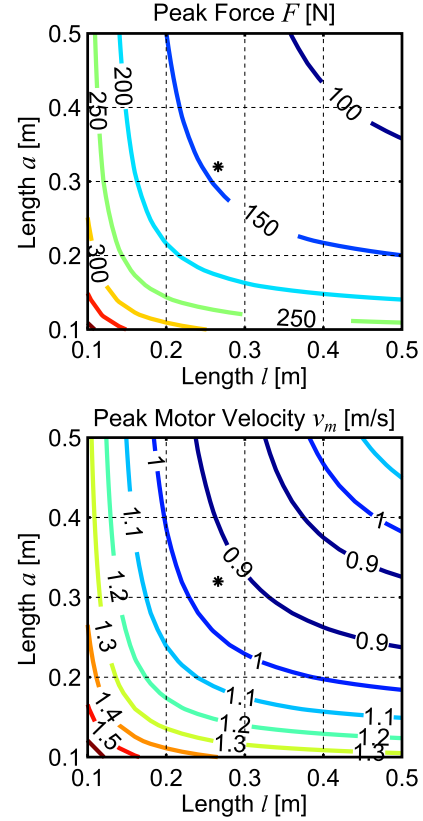


Figure 5: Peak force F (top) and peak motor velocity v_m (bottom) required to produce the approximated hip moment shown in figure 4 after it is scaled as distances a and l are varied. As the distance between the ribbon and the hip joint increases (a and l increase), the required peak force goes down. The required motor velocity decreases up to a certain point due to the motor being more able to compress the body's compliance. For large values of a and l , the peak motor velocity increases due to the necessity of tracking the leg at 70% in the gait cycle. The values of a and l used in the device are shown by an asterisk in each plot.

a mismatch in required motor speeds at different points in the gait cycle.

For larger values of a and l , the force decreases monotonically as the radius between the ribbon and hip joint increases. The peak motor velocity also decreases as a and l increase, but only up to a point. In the top right corner of the plot, with large values of a and l , higher motor velocities are again needed, but now in order to track the hip's motion at around 70% in the gait cycle. In this region, the motor power during the pull phase is reduced, but the motor may be required to move faster than desired. The minimum required motor velocity corresponds to a ribbon position which results in near-

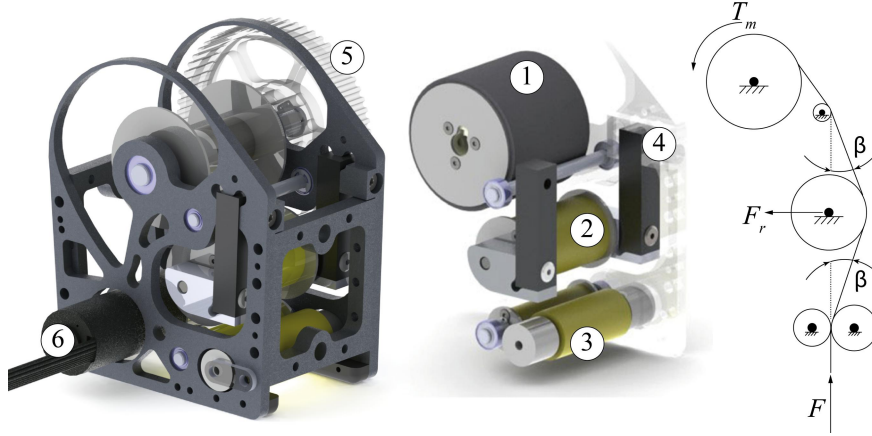


Figure 6: Left, assembled actuation unit, and center, view of the inside. Right, diagram of the pulley system, which measures the ribbon force F indirectly by measuring F_r . 1 - spool; 2 - idler; 3 - feeder system; 4 - cantilever load cell (Phidgets 20 kg Micro Load Cell); 5 - timing pulley; 6 - Maxon EC 4-pole 30 motor plus 32 mm gearbox

equal speeds for compressing the body's compliance and tracking the joint's motion.

The values of a and l used in our device are close to this point, and were prevented from being any larger due to practical concerns. The thigh brace was made tall (20 cm) to distribute the force broadly over the wearer's thigh, which limited the maximum possible value of l . Similarly, the value of a was restricted to prevent the device from protruding too far from the wearer.

For the device we constructed, we chose a maximum speed of 0.75 m/s to design towards, slightly slower than the 0.95 m/s predicted by the simulation, in order that the peak force delivered to the body might be slightly higher than would be possible with a 0.95 m/s peak speed. This slower speed means that the position and force pulse will be slightly more spread out as compared to the simulated one. This speed is still capable of tracking the hip's motion, however, which only requires a speed of 0.67 m/s.

4. Mechanical Design

Following our determination of the required ribbon speed at the motor, we can design the actuator accordingly. We first present the overall design of the actuator unit, and then discuss the ribbon spool and the electrical design.

4.1. Overall Design

A rendering of the actuator unit is shown in figure 6. The unit consists of a geared motor (6) connected

to the spool (1) that winds up the webbing ribbon via a timing belt (5). After exiting the spool, the ribbon passes over an idler pulley connected to two cantilever-style load cells (4) before exiting the unit through a feeder system (3) comprised of two additional idlers. In addition to guiding the ribbon, this system of idlers and load cells is used to detect the force in the ribbon, as shown in the far right of figure 6. The ribbon passes over the instrumented idler at an angle β on each side, so that when the ribbon pulls with force F , a horizontal force F_r is induced on the instrumented idler. By measuring this force with a load cell on each side of the idler, the ribbon force can be computed via

$$F = \frac{F_r}{2 \sin(\beta)} \quad (13)$$

where F_r is the sum of the readings from the two load cells. The load cells achieve an accurate reading of the ribbon force because the friction in the feeder system is low, and the ribbon does not contact anything between the feeder and where it attaches to the leg.

4.2. Ribbon Spool

The actuator functions by winding up webbing onto a spool, depicted in figure 7(top). The spool has a minimum usable radius $r_{s,min}$, a maximum radius $r_{s,max}$, and the webbing has thickness $t_{webbing}$. The webbing exiting the spool is at position x_m and pulls with force F , as previously discussed. The shape of the webbing in the spool is

an Archimedes' spiral [34], with radius r a function of the angle the webbing forms around the spool ϕ . In general, an Archimedes spiral has the equation

$$r(\phi) = \alpha\phi \quad (14)$$

In our situation, we have $\alpha = t_{webbing}/(2\pi)$, the spool has a minimum radius $r_{s,min}$, and we denote the radius at which the webbing exits the spool r_{spool} , resulting in

$$r_{spool}(\phi) = \frac{t_{webbing}}{2\pi}\phi + r_{s,min} \quad (15)$$

If the spool rotates at angular velocity $\omega = \frac{d\phi}{dt}$, the linear velocity of the webbing exiting the spool is

$$v = r_{spool}\omega = \frac{t_{webbing}}{2\pi}\phi\frac{d\phi}{dt} + r_{spool,min}\frac{d\phi}{dt} \quad (16)$$

and the arc length between $r_{s,min}$ and r_{spool} is

$$\begin{aligned} s(\phi_{spool}) = & \frac{t_{webbing}}{4\pi} \left((\phi_{spool} + c)\sqrt{1 + (\phi_{spool} + c)^2} \right. \\ & + \ln \left((\phi_{spool} + c) + \sqrt{1 + (\phi_{spool} + c)^2} \right) \\ & \left. - c\sqrt{1 + c^2} + \ln c + \sqrt{1 + c^2} \right) \end{aligned} \quad (17)$$

where

$$c = \frac{2\pi}{t_{webbing}}r_{s,min} \quad (18)$$

These equations are derived by taking the arc length of an Archimedes' spiral from [34], substituting in the appropriate variables for our application, and subtracting the equivalent arc length of the center of the spool ($r < r_{s,min}$).

Since the spool will change radius as it winds up or feeds out the ribbon, the velocity of the ribbon and torque produced by the actuator will change slightly over time. Figure 7(middle) examines how the radius changes for a generic spool as a function of several parameters. The x -axis of this plot shows the rotations of a sample spool, with a fully-wound spool at radius $r_{s,max}$ having N turns, and each tick mark to the left of that corresponding to one fewer winding of ribbon around the spool. On the vertical axis, the radius of the spool is plotted

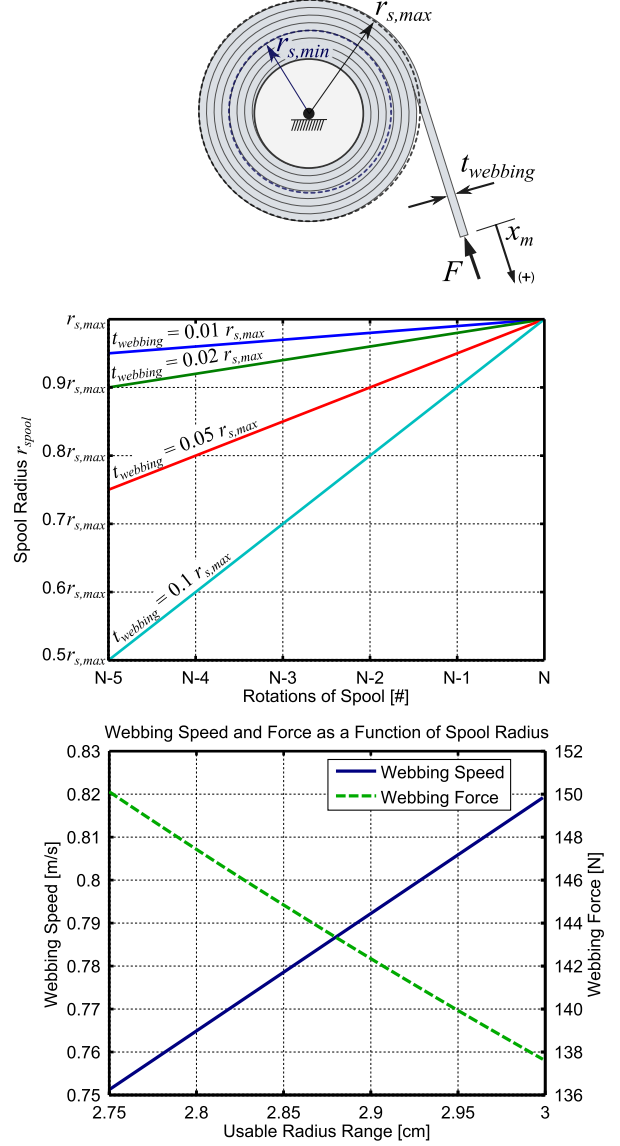


Figure 7: Top, diagram of the spool depicting the relevant variables. Middle, plot of how the spool radius varies as a function of the number of rotations of the spool for various ribbon thicknesses. Data are normalized to be fractions of the maximum spool radius r_{max} . Bottom, webbing speed and force as a function of spool radius in our actuators.

as a fraction of the maximum spool radius. The plot shows results for several different thicknesses of ribbon: the top line shows a very thin ribbon of thickness $0.01r_{s,max}$ while the bottom line shows a thick ribbon (relative to the spool maximum radius) of $0.1r_{s,max}$. For the thinnest ribbon shown, the radius varies by only 5% over five rotations of the spool, while for the thickest ribbon shown it varies by 50%. Since the velocity of the ribbon exiting the spool is proportional to the spool radius for a constant angular velocity, this graph can also be used directly to understand the variation in velocity with a spooled ribbon. For most systems, a thin ribbon is desired so the velocity and torque will not vary significantly ($< 10\%$) during operation.

4.2.1. Parameters for our system

Our device uses seatbelt webbing for the ribbon, which has a thickness of $t_{webbing} = 1.26$ mm. We chose a spool radius with $r_{s,min} = 2.75$ cm and $r_{s,max} = 3$ cm. The minimum radius is large enough to provide space inside the spool to secure the webbing to the spool, and so that the ribbon velocity is relatively constant. The size also minimizes the inertia of the spool. With these numbers, $t_{webbing} = 0.042r_{s,max}$, which would be located on figure 7(middle) just above the $t_{webbing} = 0.05r_{s,max}$ line. The spool radius is large enough that one rotation of the spool corresponds to approximately 18 cm of ribbon travel, which is more than enough to actuate the hip during level-ground walking. This means that the change in radius during walking (and hence velocity) is less than 5% throughout the gait cycle. The minimum and maximum values of our spool radius correspond to 1.98 wraps around the spool and 35.8 cm of ribbon travel. This travel gives extra room for adjusting the device to different sizes of individual as well as permitting the actuators to feed out enough ribbon to permit the wearer to bring their knee to their chest, which requires 25 cm of ribbon.

The gear reduction can be computed after the spool radius is determined. The encoder on the motor is limited to 12000 rpm = 1256.6 rad/sec. With the output spool at its minimum radius of 2.75 cm, and a required ribbon velocity of 0.75 m/s, the output spool must have an angular velocity of $\omega = v/r_{spool} = 27.27$ rad/sec. This gives a required gear reduction of 46.07:1 between the motor and output spool. We choose a gearbox of 23:1 and use a timing belt to give another 2:1 reduction to achieve 46:1 for the combination.

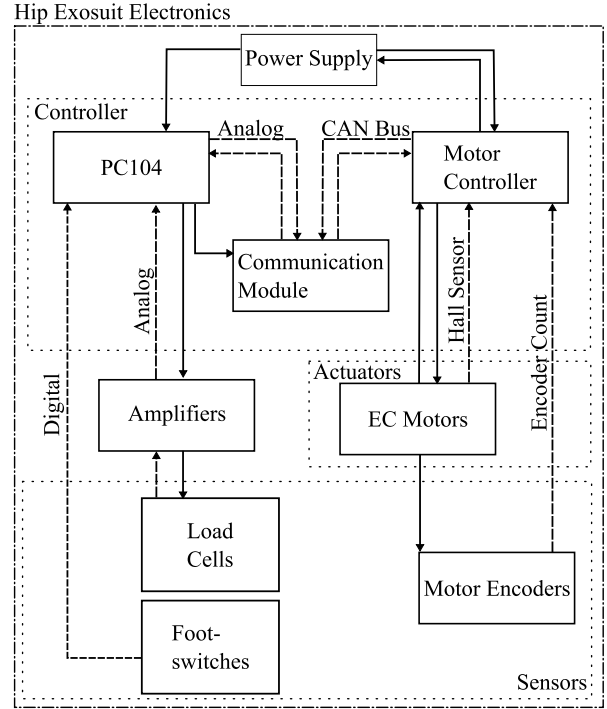


Figure 8: Electronics block diagram.

To compute the expected ribbon force at the output of the actuator, we estimate the efficiency of the drive train components. The Maxon 32 mm 23:1 gearbox has a quoted efficiency of 75%, the timing belt is assumed to have an efficiency of 95%, and the other idler pulleys and friction with the feeder have an assumed efficiency of 90-95%. This combines to make a total mechanical efficiency of 64.1-67.7%. With a motor torque of 0.14 Nm, this leads to a torque at the minimum spool radius of 4.128 Nm and a force on the ribbon of 150 N. The final ribbon speed and force are plotted as a function of spool radius in figure 7(bottom).

4.3. Electronics Overview

A block diagram of the electronics is shown in figure 8. An Aurora PC104 realtime computer running Matlab xPC Target is the primary controller. The PC104 reads in the amplified load cell signals and the footswitches, and sends an analog position control signal to an mbed LCP1768 microcontroller which acts as a communications module. This in turn runs a position controller, and uses a CAN bus to transmit desired motor currents to the Roboteq SBL1360 motor controllers. To complete the feedback loop, the motor controllers transmit the motor

encoder readings back to the mbed communications module, which then converts them to an analog signal and sends them back to the PC104 for logging.

5. Results

For preliminary walking experiments, we used footswitches to segment the gait, and a position controller to move the ribbon spool through a trajectory at specified times in the gait cycle. Following our calculations of x_m in section 3.3, we form a position trajectory profile by using a triangular pulse with a similar amplitude as x_m (12.3 cm actual vs. 11.8 cm x_m) and similar timing, but with the motor acceleration- and velocity-limited to 3500 rotations/sec² and 12000 rpm, respectively. The resulting profile extends from 80-25% in the gait cycle. The controller maintains a buffer of the last 5 footstrike times and computes the average gait period. It assumes the wearer is walking at a constant velocity, and uses the average gait period to estimate when they are at 80% in the gait cycle. At that point, the motor executes the position-controlled motor trajectory, which creates a force in the ribbon through the compliance of the body. This scheme depends on the wearer walking at a constant cadence in order to have the motor's pull line up well with the biological hip moment. Even so, this simple control scheme is useful for determining the behavior of the device.

Results from an individual wearing the system while walking at 1.25 m/s (2.8 mph) are shown in figure 9. In this figure, we compare simulated data with measured data to validate the model. For the simulated data, we use the reference hip angle θ_{hip} from [33], lengths a and l measured on the subject, and the actual motor trajectory x_m . These are used with the model developed in section 3 to calculate the resulting webbing force and hip position. In the figure, the top graph shows the actual motor trajectory x_m along with the simulated hip position x_{hip} and spring length x_s . The bottom plot shows the simulated force as compared to the measured force and its standard deviation.

As shown in the bottom graph, in practice the device achieves 9% less force (133 N vs. 146 N) despite 0.5 mm additional cable travel as compared to the simulation. This difference is likely due to additional torso lean which occurs during walking, especially when the force is applied at the shoulder. The measured force also drops off faster than the simulated force; this is likely due to the fact that the

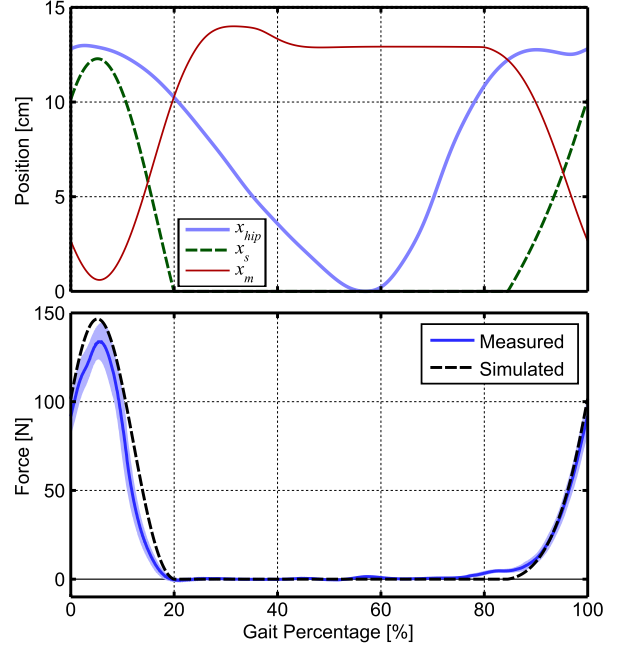


Figure 9: Results for one leg for a subject walking at 1.25 m/s (2.8 mph). Top, graph of motor position x_m used in the experiment, and simulated x_{hip} and x_s . Bottom, graph of measured force in the ribbon with standard deviation as compared to simulated force using the model.

model did not include the hysteresis in the body-suit stiffness curve. In general, though, the model fits the measured data well.

The resulting assistive pulses of force feel natural due to their being aligned well with the nominal biological torques. During the period of force application, the leg is approximately at a constant angle of $\Psi = 35^\circ$, which also can be seen from the approximately constant position of x_{hip} in figure 9 from 90-15% in the gait cycle. At this angle, the thigh brace pulls up along the leg as well as backward. This does result in some translation of the thigh brace vertically along the leg, but this motion is minimal due to the conical shape of the thigh just above the knee. If the thigh brace is initially wrapped snugly around the thigh, there is no motion between the skin and thigh brace, making it comfortable to wear during actuation for the level of force we propose to apply.

6. Conclusion

In conclusion, we have modeled and designed an exosuit that effectively applies moments to the hip

joint, contributing up to 30% of the nominal biological moment for walking. The design does not restrict the hip's motion with rigid linkages, but relies on the bone structure to support compressive forces across the joint. Substantial forces can be applied to the leg through the soft interface, and the limiting factor is not the comfort of the system but the available motor power. The soft interface does have substantial compliance, which requires higher motor speeds compared to a rigid system. Compared to classical exoskeletons, the system is lighter and less restrictive to the wearers motion. In addition, it can be put on and taken off very rapidly or simply disengaged from the wearer while assistance is not required. While having these advantages, the system is limited in the amount of assistance that can be applied to the wearer and uses a single motor that only assists unidirectionally (in that it assists hip extension but not flexion). However, we believe this will be sufficient for improving walking economy.

For the current control strategy, we assumed that an appropriate torque profile to apply to the hip was a scaled version of the biological joint moment, but further work is required to validate this. For assistive applications such as hiking or load carriage, it may be beneficial to provide the wearer with as much power as possible. For rehabilitation, individuals may require moment profiles specific to their particular injury. If a scaled biological joint moment is indeed best, increasing the speed and bandwidth of the actuation system would be useful to duplicate this moment more precisely, since the device currently provides a smoothed version of the biological moment. At force levels higher than those used currently, matching the biological moment may be especially important. Increasing the bandwidth of the actuators could also permit using force-control instead of a position-control scheme. Improving the control scheme will be a major thrust of future work, so the system can adapt on a per-step basis to changes in gait. Adding additional sensors such as gyroscopes to estimate the hip angle over time may be useful. Further testing of the system is also necessary to determine its effect on gait, both kinematically and metabolically.

In future work, the system will be made significantly lighter and smaller, so that it can be packaged compactly to fit onto a hiking backpack or orthosis supporting the trunk. Further optimization will also be performed to select the vertical height of the actuators with respect to the hip joint to max-

imize the overall efficiency. The current design allows a person to sit down while wearing the device, but positioning the actuators lower may improve the stiffness and increase the moments transmitted to the joints. Alternately, for some applications a lower-profile unit that could be worn under clothing may be beneficial. In all of these situations, the ability to provide moments at the hip through a compliant, non-restrictive mechanism should be very useful.

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