

A Lightweight Soft Exosuit for Gait Assistance

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Abstract— In this paper we present a soft lower-extremity robotic exosuit intended to augment normal muscle function in healthy individuals. Compared to previous exoskeletons, the device is ultra-lightweight, resulting in low mechanical impedance and inertia. The exosuit has custom McKibben style pneumatic actuators that can assist the hip, knee and ankle. The actuators attach to the exosuit through a network of soft, inextensible webbing triangulated to attachment points utilizing a novel approach we call the virtual anchor technique. This approach is designed to transfer forces to locations on the body that can best accept load. Pneumatic actuation was chosen for this initial prototype because the McKibben actuators are soft and can be easily driven by an off-board compressor. The exosuit itself (human interface and actuators) had a mass of 3500 g and with peripherals (excluding air supply) is 7144 g. In order to examine the exosuit’s performance, a pilot study with one subject was performed which investigated the effect of the ankle plantar-flexion timing on the wearer’s hip, knee and ankle joint kinematics and metabolic power when walking. Wearing the suit in a passive unpowered mode had little effect on hip, knee and ankle joint kinematics as compared to baseline walking when not wearing the suit. Engaging the actuators at the ankles at 30% of the gait cycle for 250 ms altered joint kinematics the least and also minimized metabolic power. The subject’s average metabolic power was 386.7 W, almost identical to the average power when wearing no suit (381.8 W), and substantially less than walking with the unpowered suit (430.6 W). This preliminary work demonstrates that the exosuit can comfortably transmit joint torques to the user while not restricting mobility and that with further optimization, has the potential to reduce the wearer’s metabolic cost during walking.

I. INTRODUCTION

Over the last two decades, a number of exoskeletons have been developed for tasks ranging from heavy lifting [1] to helping the wearer to walk [2-6] and for providing robotic rehabilitation therapy in a hospital setting [7]. Recently, with improvements in actuator and sensor technology we have seen these systems become portable, and begin to transition from academic to commercial applications. Several categories of exoskeletons exist, including those that provide the ability to replace human movements that have been completely lost, e.g. in the case of a patient paralyzed below the waist. To achieve this, the device must provide sufficient control to ensure the user’s full stability, making high speed and agility secondary concerns to balance and safety. In effect, these devices can be thought of as wheelchair

replacements and offer an elegant and potentially life-changing, tool for a specific group of users [8, 9]. Another type of exoskeleton is designed to assist able-bodied users perform tasks more easily or for longer duration. In particular, considerable work has been conducted in the area of active exoskeletons for augmenting load carriage capacity [10-14]. For all these devices, a key challenge is minimizing the weight and power requirements and to this end some groups have proposed quasi-passive architectures in an effort to reduce the exoskeleton’s energy consumption [4].

These previous exoskeletons all rely on rigid frameworks of linkages, coupled to the body at select locations via pads, straps, or other interface techniques. As the wearer flexes or extends their biological joints, these rigid links add considerable inertia to movement which must be overcome by motors or by the user. Though great effort has been made to minimize these effects, they still add considerable impedance to the natural gait dynamics and kinematics. Also, static misalignment of the biological and exoskeleton joints can result in dynamic misalignments of up to 10 cm during normal movement, causing pain and even injury to users [15]. One solution has been to include redundant, passive degrees of freedom to accommodate these misalignments [16]; however, this adds further weight to the system. It is partly for these reasons that these systems do not typically reduce the metabolic power required for locomotion. In order to address these issues there has recently been work on developing active soft orthotics [17, 18] that show great promise in reducing the impedance experienced by the wearer and allowing more natural movement. As an alternative to rigid exoskeletons, we present the design and evaluation of a soft wearable device, we call an “exosuit” (Fig. 1). This device was designed to augment the normal muscle work of healthy individuals by applying assistive torques at the wearer’s joints with the goal of reducing the metabolic cost of transport of the wearer.

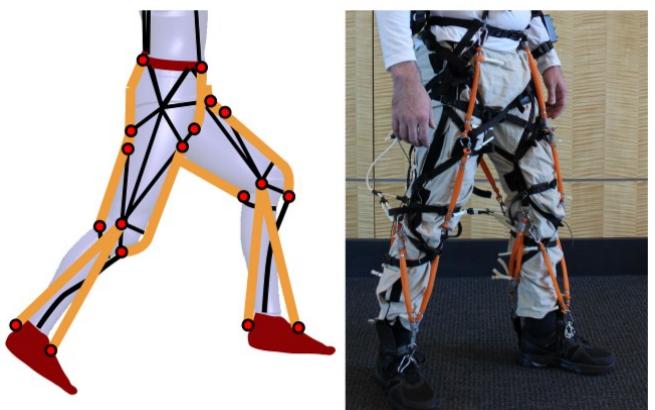


Figure 1. A lightweight, non-restrictive soft lower extremity exosuit that can provide gait assistance through pneumatic actuators (orange) that span the hip, knee and ankle joints.

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This device utilizes flexible materials and actuators to specifically address human factors challenges and does not have a load bearing “skeleton” but rather relies on the biological skeleton to assist with the application of forces and transfer of load. Compared to traditional exoskeletons, the exosuit presented provides minimal additional mechanical impedance and kinematic restrictions.

When considering how a soft exosuit can augment healthy gait, there are several biomechanical studies that provide insight. One strategy is by supplementing normal muscle positive work by adding a small amount of additional energy at the hip and/or ankle during the stance phase and at the onset of the swing phase. With this approach, the system is leveraging the natural dynamics associated with human walking that have been elegantly demonstrated with some walking robots. For example, previously, a passive dynamic walking model was suggested to describe the efficient nature of locomotion [19] and these principles were implemented in a robot that demonstrated a similar cost of transport to human locomotion [20]. Thus when considering actuation for a soft exosuit, it may be possible to have a controller that can add impulses of power at certain joints at the right phase of the walking cycle. In particular, we hypothesize that with such an approach, we can provide assistance to locomotion and reduce the metabolic cost of walking without significantly altering the wearer’s gait.

For the remainder of this paper, we outline the requirements of actuators for the soft exosuit and propose a soft interface concept for providing actuation at the joints to assist with forward propulsion during walking. A prototype was fabricated and its performance evaluated in a pilot study that examined the effect of ankle torque assistance timing on gait kinematics and metabolic power.

II. ACTUATION: REQUIREMENTS & IMPLEMENTATION

With the goal of applying impulses of energy at particular instances during the gait cycle rather than directly controlling limb positions, McKibben actuators provide an attractive solution. These actuators have a high power to weight ratio with the air source being off board, muscle-like force length properties and have intrinsic compliance [21]. Similar actuators have been successfully used in other soft wearable projects [17, 18].

A. Biological Requirements

Requirements for the system were determined from the 50th percentile male, with the specification that it be capable of emulating approximately 50% of the forces and normal ranges of motion of normal walking as defined by Hallenmans et al., 2005 [21, 22].

To translate torques and rotational motion into linear forces for the pneumatic actuators, anthropometric values were found for each joint to estimate moment arm and total required travel [21]. These custom McKibben pneumatic actuators are known to contract 25% during actuation. Thus, knowing the desired range of motion, the required force and displacement can be calculated. Device specifications for the joint actuation for the soft exosuit are given in Table I, showing values derived from anthropometric values and based on McKibben actuator capabilities.

TABLE I. RANGE, MOMENT, AND POWER DURING NORMAL WALKING OF FIFTIETH PERCENTILE MALE, 79KG MASS, 1.75M HEIGHT

Degree of Freedom	Range of motion (deg)	Moment (Nm)	Moment Arm (m)	Max Force (N)
Ankle Plantarflexion	25	100	0.06	1867
Ankle Dorsiflexion	10	5	0.06	67
Knee Flexion	60	25	0.07	457
Knee Extension	-5*	25	0.07	457
Hip Flexion	35	80	0.12	750
Hip Extension	10	50	0.12	367

* Maximum knee extension is less than zero (straight leg) during normal walking.

B. Actuator Design and Characterization

Actuators used for this study were constructed from latex tubing, expandable mesh, pinch clamps, and barbed fittings; all available commercially as previously described in [17]. While, commercial McKibben actuators are available for purchase, custom actuators were made to minimize weight for the size and force requirements based on the biological specifications shown in Table I.

As compressed air is supplied to the actuators, they expand radially, shortening in length and thereby providing actuation. Force can thus be modified by changing input pressure and the stroke length is determined based on the actuator length and properties. An actuator with an active length (excluding end hardware such as fittings and steel loops) of 200 mm was prototyped and force versus displacement data were recorded on an Instron 5544 load frame system for 1 to 5 bar (14.7 to 73.5 psi) input pressures (Fig. 2).

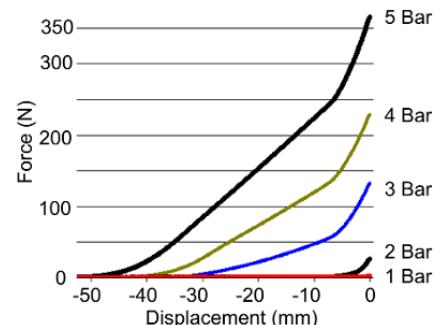


Figure 2. Force versus displacement curves for a 200 mm long custom made McKibben actuator. Negative displacement indicates shortening. As shown, an actuator will generate a peak force when engaged at its rest length, from there, its force producing capabilities decrease with contraction. Increasing supply pressure increases the actuators force output and its amount of maximum contraction.

For this prototype, 4 bar (58.8 psi) was chosen as an operating pressure to provide substantial actuating force, yet provide an additional safety measure, providing forces 38% lower than those at the maximum pressure of 5 bar (230 N vs 370 N). Air flow to the actuators was controlled by inline solenoid valves (Norgren, 2 position, 5/2 valve model E12), and air was supplied from the compressor to the system with 3/8 inch OD (1/4 ID) tubing, and distributed from the valves to the individual actuators with individual 1/4 inch OD (1/8 ID) tubes.

When using these actuators for a dynamic application such as assisting with gait, inflation and deflation times must be considered as the actuators do not actuate instantaneously,

and thus could potentially interfere with joint motion rather than assist if timed incorrectly. To investigate this, force versus time was recorded at 4 bar during inflation and deflation (Fig. 3). For inflation, 90% of max force (235 N) was obtained after 0.316 seconds from when pressure was applied and force dropped from maximum force to 10% of maximum force in 0.098 seconds. Using this actuator as a unit, the force and power requirements listed in Table I could be met by selecting the appropriate number of actuators for each degree of freedom. For example, four actuators could be configured in parallel at each ankle joint to provide 50% of the forces requirement for ankle plantarflexion.

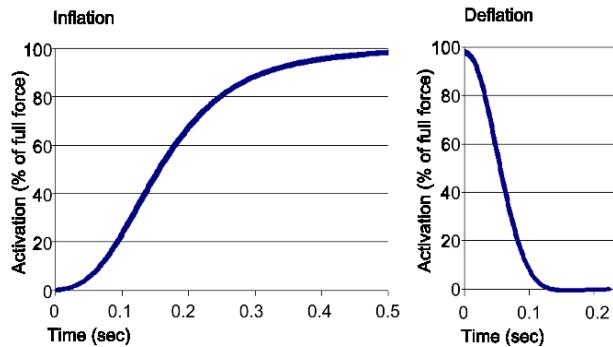


Figure 3. Force development rate for a 200 mm long custom made McKibben actuator in response to a step input of 4.0 bars of air pressure (inflation and deflation). Force is shown as percent of maximum force (235N).

Finally, air consumption was determined for the test actuator. At 4.0 bar, this actuator consumed 0.60 gram (0.021 mol) of air per actuation. These data can be scaled for actuators of other lengths.

III. SOFT HUMAN INTERFACE DESIGN

Design specifications for the soft exosuit human interface were that it be (1) lightweight and add minimal inertia that could potentially disrupt normal gait dynamics; (2) non-restrictive so that it would not disrupt joint kinematics in any body plane and (3) comfortable as reported by the user during operation. In this section, we describe how our interface was constructed in order to meet these requirements.

A. Applying Loads to the Body: Concept of a Key Anchor

In order to develop a suit of primarily soft, non-restrictive components we must develop a method of applying loads to the wearer using tension only while maintaining wearer comfort. Here, the exosuit acts as augmentation for selected muscles, utilizing the wearer's skeletal structure to generate any compressive, bending, or shear loads required in the system.

In order to apply a torque to a biological joint, these actuators require a means to apply a reaction force to another part of the human body. From section II, we see that the values of these forces can be quite high. Previously, techniques such as tight straps or skin-adhesives were used to maintain the position of wearable devices, but experience has shown, that this quickly causes discomfort [16] [23, 24]. Further, such approaches are only effective when required to support small loads such as the weight of the components, rather than larger forces for augmenting motion of the limbs. Forces parallel to the skin cause slippage, chafing, and it has

been reported that non-perpendicular forces are "Intolerable unless very small and intermittent" [23]. In addition to causing pain and discomfort, non-perpendicular forces have a high likelihood of causing device deformation and slippage, which would render any actuation ineffective.

Some areas of the body are known to be better for supporting forces than others and this understanding can provide inspiration for the design of new methods for applying forces to a person using a soft suit. During load-carrying-walking, the payload is primarily supported by the shoulders, back, hips, carried in the hands, or balanced on the top of the head. In other tasks such as sitting, lying, or walking with crutches, other body parts are used to support the ground reaction force. In carrying large or odd-shaped items, and when supporting items while in non-standing postures, other body regions such as the forearms, chest, or lap are used to support loads.

In the context of a soft exosuit, we define these regions known to readily support load, particular the shoulders, iliac crest of hips, plantar aspect of the feet, as "key anchors". These are typically regions that exhibit large bony landmarks near the surface of the skin which can withstand reaction forces applied normal or nearly normal to it. For example, at the hips, we find that loads are borne downward on the top of the iliac crest region, not in shear along the side of the hip [25].

A key question then is how we can leverage the above strategies that humans exploit when carrying heavy loads to apply augmentative torques with a soft exosuit. By observing the human body in motion, we note that during joint motion, some paths on the skin surface change in length substantially relative to one another, while others exhibit quite little relative motion. These low strain paths have been widely studied and quantified as lines of non-extension [26]. Using these regions of high and low strain on the skin surface as inspiration, we propose a new approach defining the concept of a "virtual anchor point" as a way to apply assistive torques at the joints of the biological leg.

B. Virtual Anchor Concept

The goal for the soft exosuit was to apply assistive torques in the sagittal plane at the hip, knee, and ankle joints. Using the concept of a virtual anchor, the reaction force from a desired actuator was redirected to a key anchor. To achieve this we utilized the lines of non-extension concept as inspiration to configure a matrix of connectors from the desired actuation point, triangulating with other connectors to maintain stability during normal range of motion, redirecting forces, and terminating at a key anchor point. This robustly constrains the desired actuation point, minimizing distortion and effect on range of motion, with little force transmitted locally to the wearer. This concept is illustrated in Fig. 4.

Careful selection of triangulation paths along lines of non-extension was performed for hip flexion and extension, knee flexion, and ankle dorsiflexion and plantarflexion. Virtual anchors on the distal end of actuators were connected to the ankles, while those on the proximal end were connected to the hips and shoulders, distributing forces as broadly as possible, and maintaining forces normal to the skin (Fig. 5).

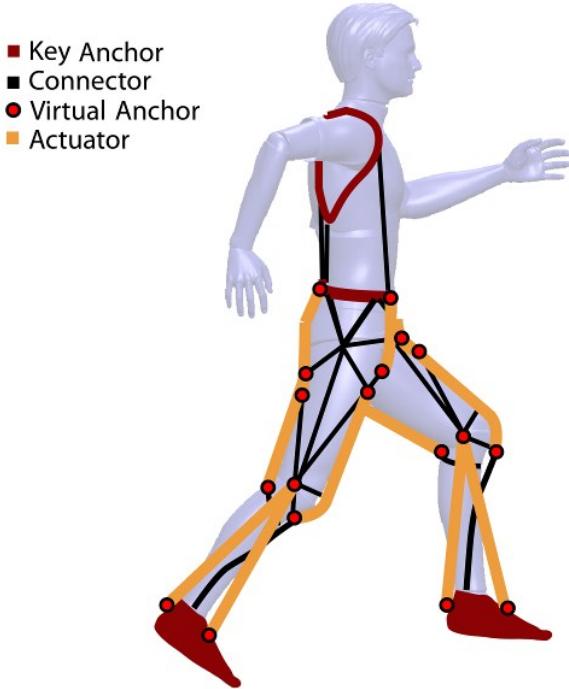


Figure 4. Virtual anchor concept that enabled pneumatic actuators (thick orange lines) to generate joint torques in the sagittal plane at the ankle, knee, and hip. This was accomplished by attaching the actuators to virtual anchor points (red dots) that are constrained from moving significantly by soft inextensible webbing (black thin lines) that distributed the actuator forces to areas known to more readily support loading.

To further illustrate this concept and the forces involved, it is helpful to consider how the knee joint can be actuated (Fig. 5). Virtual anchor points distally and proximally about the knee joint (called VA1 and VA2 in Fig. 5.1) enable tension to be applied between these two points to actuate the knee in extension. As previously discussed, for effective device operation, points VA1 and VA2 cannot move relative to the underlying limb, thus they must be constrained or anchored with sufficient stiffness to resist forces on the order of thousands of Newtons (from Section II). Viewing VA1 from the frontal view (Fig. 5.2), it can be seen that an additional connection is required along a contralateral path to the anchor for stabilization. In order to maintain equilibrium and avoid anchor dislocation, F1 must remain within the angle between F2-1 and F2-2 connectors (red lines). The virtual anchor technique is repeated at the proximal end of the actuator transmitting force to the waist-belt, distributing forces along the iliac crest of the hip and to the shoulders.

To achieve forces required for approximately 50% emulation of human walking range from zero to 933N during ankle plantar-flexion (Table I). These peak forces can be matched with the virtual anchor and McKibben actuator design and this method is inherently safe because the force drops to zero as displacement increases. Further, our assumption is that the virtual anchor locations will be fixed relative to the wearer but in reality there will be some amount of compliance in both the soft suit and where it attaches to the wearer. This effect will reduce the peak force and also result in motion of the suit relative to the wearer.

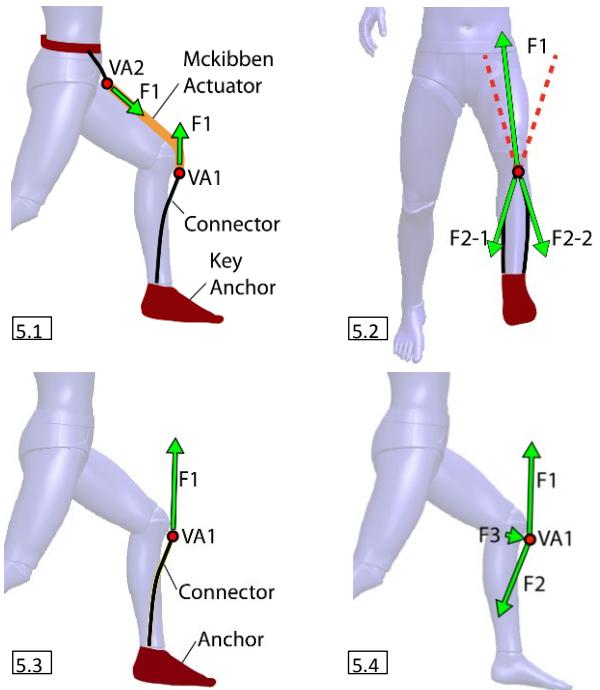


Figure 5. A graphical explanation of the virtual anchor technique. To apply a moment about the knee using soft components (tension only), we would like to apply force F1. Performing free body analysis on VA1 (left), we see that the majority of F1 is counteracted by tension in the connector F2. Only a small net compressive force (almost no shear) is transmitted to the skin to maintain equilibrium.

IV. PROTOTYPE IMPLEMENTATION

Based on the requirements outlined in Sections II and design principles described in Section III, a prototype exosuit was fabricated to demonstrate the concept. Virtual anchors consisted of triangular threaded links (Quik-Links), sewn into a matrix of nylon strapping material attaching virtual anchors to key anchors. Carabiners, squeeze-release buckles, and oval Quick-Links were used as necessary between portions of connector straps to allow donning and doffing of the soft exosuit. The quantity and length of actuators was selected to achieve approximately 50% of the desired force and ranges required for the associated degree of freedom (Tbl. I).

As can be seen from Fig. 6, the human interface for the soft exosuit is inherently flexible and it is primarily made of soft components. The black nylon webbing provided the inextensible elements that redirected forces from the virtual anchors to the key anchors. The orange pneumatic actuators have their ends attached to the key and virtual anchors on either side of a biological joint. The user was able to move their hip, knee and ankle joints through almost their full range of motion in the sagittal plane.

This initial proof of concept prototype was not optimized for quick and easy donning and doffing. The process for putting it on began with loosening the various buckles and straps so that it could be worn on the legs, followed by tightening them so that the device fitted snuggly to the wearer and nylon webbing was in the correct location of the leg and across the joints. However, once fitted to a particular subject, locations of the various components were marked so that future donning could be performed more quickly.



Figure 6. Left, soft exosuit with connector matrix webbing for virtual anchors. Right, with actuators attached to virtual anchor points. A key feature of the exosuit is that it is lightweight, made primarily from soft components and does not restrict the users range of motion.

The pneumatic valves and controller were housed in a back-mounted assembly. The suit was equipped with a tuning box that permitted the onset delays and actuation durations to be adjusted in real time. The box was wrist mounted to allow the wearer to adjust parameters on the fly.

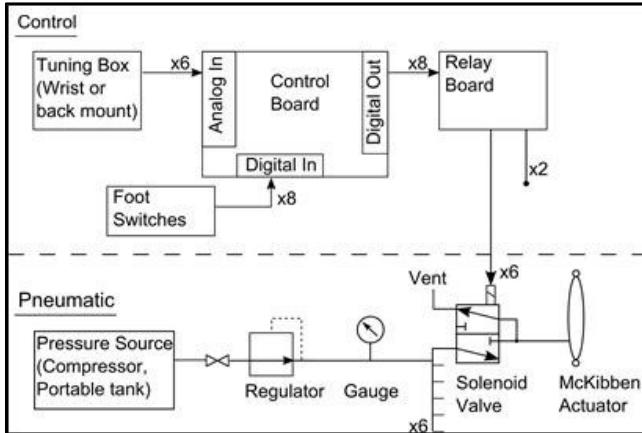


Figure 7. Schematic of the control and pneumatic system illustrating the various components.

Heel strike was sensed via footswitch-instrumented insoles (B&L Engineering), sending a signal to an Arduino Mega 2560 microcontroller (<http://arduino.cc/en/>). Upon sensing heel strike, the controller initiates a timing sequence for actuating the desired degrees of freedom on that limb. Each degree of freedom had a programmable turn on time (actuator turn on time after heel strike) and actuation duration. Heel-strike was sensed for both feet and used to initiate delays and actuations independently. Timing sequences were identical for both legs to maintain symmetry. A schematic of the control/pneumatic layout is shown in Fig. 7, and photo in Fig. 8. Compressed air can be supplied via compressor for stationary/treadmill testing. For untethered testing, compressed air was supplied by a back-mounted 64 cubic inch paintball tank, at 306 bar (4500 psi) as a proof of concept portable implementation.

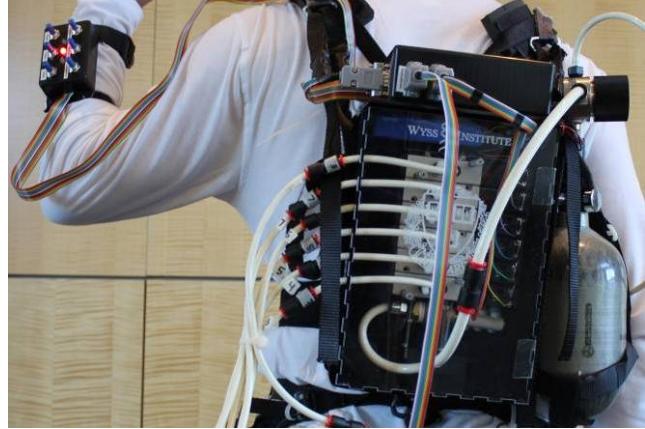


Figure 8. Image of control components

The exosuit has a total mass of 7144 g when tethered to a compressor and 9121 g when including the onboard compressed air tank. The component of the suit that is worn on the lower extremities has a mass of only 3500 g, thus minimizing distal mass which is known to have greater effect on metabolic cost [27]. Masses of major modules are listed in Table 2. The soft exosuit consumed 0.166 mol (4.8 g) of air per gait cycle. Assuming a stride frequency of 1Hz, the suit currently consumes 9.94 mol per minute. The 64 cubic inch, 4500 psi compressed air tank contains 14.3 mol (415 gram) of air, and would last for 1.45 minutes of constant walking.

TABLE II. MASSES OF THE MAJOR MODULES OF THE SOFT EXOSUIT.

Item	Mass (gram)
Suit (pants, shoes, actuators, support straps)	3500
Valve box with batteries	3280
Control module including wrist mount tuning box	364
Total (without air cylinder)	7144
Air cylinder and associated plumbing	1977
Total (untethered)	9121

V. SOFT EXOSUIT EVALUATION

A pilot study, approved by the Harvard Medical School Committee on Human Studies, was performed to examine the performance of the soft exosuit in assisting gait by using the pneumatic actuators to enhance ankle joint torque during powered plantarflexion. All other actuators had their actuation duration adjusted to zero milliseconds so that they generated no force. Kinematic and metabolic data were collected at the Wyss Institute's Motion Capture Laboratory in order to quantify the soft exosuit efficacy. The effect of the ankle plantarflexor actuator's engagement timing on joint kinematics and metabolic power was investigated by varying when the actuator was turned on during the gait cycle occurred. Heel strike of the ipsilateral leg was defined as 0% of the gait cycle. Six actuator turn on times were investigated, ranging from 10 – 60% of the gait cycle in 10% increments, as well as with the suit in a completely passive unpowered mode and with the subject not wearing the suit at all.

A. Kinematics

A Vicon® motion analysis system with 8 infrared cameras (Oxford Metrics, Oxford, UK) was used to obtain the kinematics of one healthy male subject aged 42, 65 kg

and 1.73 m tall. The participant was asked to walk at 1.5 m/s along a 10 meter flat ground (not treadmill) walk-way. Trials with a walking speed greater than $\pm 5\%$ of 1.5 m/s were excluded until three acceptable gait trials were attained. Motion capture data were collected at a sampling rate of 120 Hz. A total of 44 markers were attached to the participant based on a modified Cleveland Clinic marker set [28]. Lower body markers were placed on the following anatomical landmarks: bilateral anterior superior iliac spines, bilateral apex of the iliac crests, dorsal aspect at the L5-sacral interface, lateral and medial femoral condyles, lateral and medial malleoli, calcaneal tuberosities and the superior aspect of the first and fifth metatarsophalangeal joints. Triad marker clusters were placed on the femora and tibiae. Upper body markers were placed at the forehead, left and right temple, seventh cervical vertebra, sternum, tip of the tip of the acromia processes, humeral lateral epicondyles and the midpoint between the radial and ulna styloid processes.

OpenSim 3.0 was used to perform the inverse kinematic analysis [29]. An OpenSim 23 degrees of freedom head, torso and lower limb model was scaled to the subject based on 14 anthropomorphic measurements. After scaling the generic model anatomical joint angles were calculated based on the three dimensional marker trajectories. Means and standard deviations of the ankle, knee and hip joint angles with respect to the gait cycle were computed. It was observed that the values for the joint angles differed (e.g. greater knee flexion during stance) than those generally reported. This was likely due to challenges in marker placement due to the suit. However, in this pilot study, we were only interested in the relative effect of the suit in its power and unpowered configuration, compared to normal walking and so this was not deemed to be of major concern.

The sagittal plane hip and knee joint angles remained similar between the no suit, passive suit, and all the actuated suit conditions (Fig. 9). For all test conditions, the hip joint had typical sagittal plane behavior with initial flexion at heel strike, extension throughout the stance phase and then flexion during the swing phase (Fig. 9). The sagittal plane knee angle for both the passive and actuated test cases also had a typical pattern with the knee initially flexing from heel strike through the loading response, extending from midstance to heel rise, flexing from heel rise to toe off and finally extending during swing (Fig. 9).

Sagittal plane ankle joint kinematics were affected by the actuated exosuit (Fig. 9). Providing additional joint torque at the ankle joint via the McKibben actuators during the stance phase caused the ankle to become more dorsiflexed during the loading response and more plantarflexed from the end of terminal stance to the beginning of pre-swing. As would be expected, the conditions which had the actuator turned on at 10% and 60% of the gait cycle caused the greatest change in ankle joint kinematics with up to approximately 15° difference compared to baseline walking without wearing the exosuit. In contrast, when pressure was supplied to the actuator at 30% of the gait cycle and left on for 250ms, the sagittal plane ankle joint kinematics remained similar to baseline walking without wearing the suit (Fig. 9)

and was similar to the sagittal plane ankle joint kinematics when wearing the passive suit with no actuation (Fig. 9).

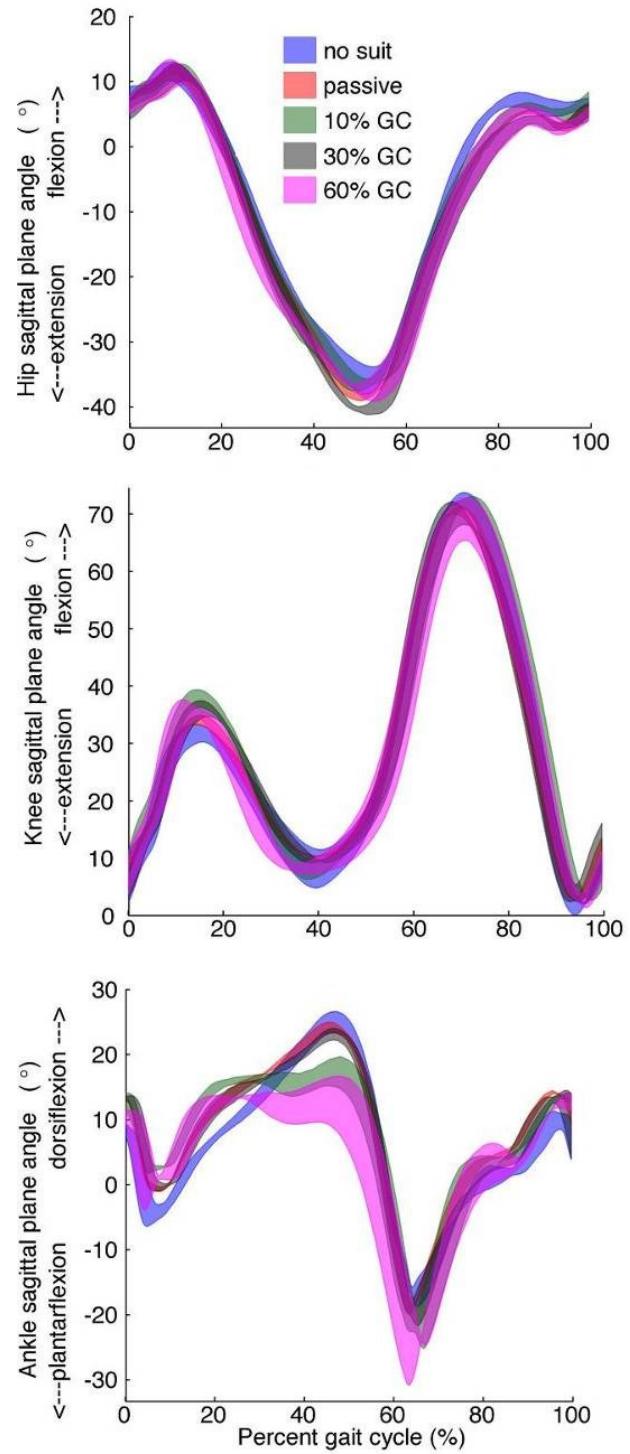


Figure 9. Mean \pm 1SD sagittal plane right hip (top), knee (middle) and ankle joint (bottom) kinematics with respect to the gait cycle. Baseline walking without wearing the suit is shown in blue while actuated conditions with 10%, 30%, and 60% are shown in green, black, and magenta, respectively. As shown, the exosuit had little effect on the hip and knee joint kinematics. For the ankle, actuator turn on times of 10% and 60% of the gait cycle had the greatest effect on kinematics while an actuator that was pressurized at 30% of the gait cycle (green) for 250 ms resulted in ankle joint kinematics which were similar to baseline walking without wearing the suit (blue).

B. Metabolic Power

The subject's metabolic power was measured for the following eight test conditions: 1) standing at rest, 2) walking while not wearing the exosuit, 3) walking with the exosuit unpowered (passive), 4-8) walking while wearing the exosuit with actuator turn on times of 10% through 60% adjusted in 10% increments. For each test case, the same subject previously described (section V, A) walked on a level treadmill at 1.5 m/s for 8 to 10 min after providing written informed consent. A Cosmed K4b2 cardio pulmonary exercise testing device (COSMED USA, Concord, CA) was used to measure the pulmonary gas exchange (VO_2 , VCO_2) during the entire treadmill walking session. The average metabolic power (W) over a 4 min steady state interval was calculated according to the method described by J.M. Brockway [30]. The standard deviation of the metabolic power was calculated according to inter-breath variability.

The average metabolic power for the powered suit conditions were minimized when the ankle plantarflexor actuator turned on at 30% of the gait cycle. With this actuator timing, the average metabolic power while walking was 386.7 ± 4.4 W, almost identical to the average power when wearing no suit at all (381.8 ± 6.0 W), and substantially less than walking with the passive unpowered suit (430.6 ± 8.6 W). There was a 43.9 W or 10.2% reduction in average metabolic power when comparing the powered and optimally tuned suit to the passive unpowered suit. The highest average metabolic power (438.8 ± 3.4 W) occurred when the actuators turn on at 20% of the gait cycle (Fig. 10).

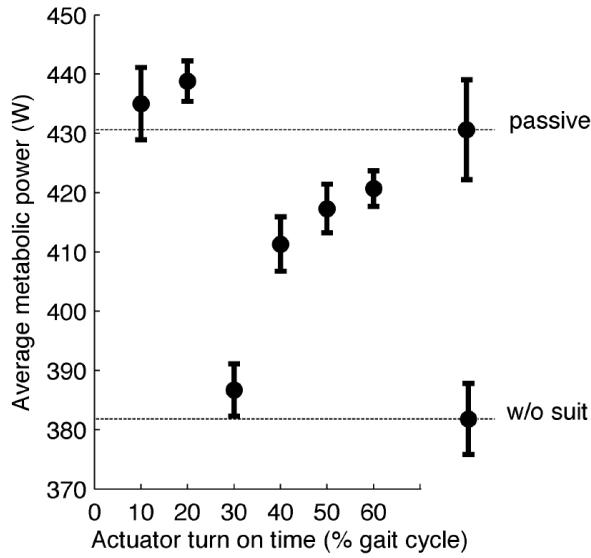


Figure 10. Average metabolic power for six different actuator turn on times. The metabolic power when the suit was not powered (passive) and when the subject did not wear the suit (w/o suit) are also presented for reference. The error bars represent \pm one inter-breath standard deviation.

VI. CONCLUSIONS AND FUTURE WORK

We present here what we believe to be the first engineered soft exosuit, which greatly reduced mechanical impedance and inertia compared to previous exoskeletons and wearable assistive devices. We described a new approach to interface to the wearer via a matrix of soft elements that provides

virtual anchors for actuators to attach to. This was accomplished via a triangulated web of inextensible elements that spanned from desired attachment points to distal and proximal areas of the body known to more readily accept higher forces.

A pilot evaluation demonstrated the effectiveness of this approach. Wearing the suit in a passive unpowered mode had little effect on hip, knee, and ankle joint kinematics as compared to baseline walking when not wearing the suit. We observed that when powered joint torque assistance is improperly provided, it alters gait kinematics and an undesirable increase in metabolic power occurs. However, through proper tuning of actuator timing, the exosuit becomes synergistic with the wearer resulting in kinematics that return to near baseline conditions while also simultaneously minimizing metabolic power. Specifically, we found that engaging the ankle plantarflexor actuators at 30% of the gait cycle for 250 ms resulted in gait kinematics similar to those during normal walking or walking with the unpowered suit. Furthermore, when walking with the powered exosuit with these actuation parameters, the subject's average metabolic power was 386.7 W, almost identical to the average power when wearing no suit (381.8 W), and substantially less than walking with the unpowered suit (430.6 W).

While, we did not demonstrate that the suit can make it easier to walk for the wearer, there are a number of avenues that we plan to explore to further optimize the device design. While, the actuators were designed to provide 50% assistance to the wearer, it is unlikely that this was the case due to some inherent compliance in the interface to the wearer. We plan to add sensors in series with the actuators to measure the actual forces that are transmitted to the soft suit. These data will be used to further optimize the webbing component of the interface. Additionally, for the pilot study presented here, actuation was only applied at the ankle joint and future studies could be performed to examine the effect at other joints as well as the combination of joints. Here we presented a simple timing based control scheme; however, with the addition of sensors to the virtual anchor matrix, additional human-machine interaction control modalities can be examined. Specifically, we can examine those relevant and have been shown to be very important from a physiological perspective [31].

An extensive study of the full range of parameters of this device was beyond the scope of this paper, but we feel that this work opens the door to a broad range of biomechanics studies, not previously possible with other exoskeletons that have a significant effect on natural dynamics and kinematics. Further, we note that while no statistical analyses were performed here, the trends presented for this first case study are promising. Future work will also focus on statistically evaluating the efficacy of the exosuit with an increased number of human subjects.

This new paradigm provides, for the first time, design rules, fabrication methodologies, and control strategies for

soft wearable robots that can assist with various forms of human motion and related activities. Soft robotics is an emerging field that combines classical robotic design and control principles with active soft materials, enabling a new class of applications exemplified by the device presented. In this system, symbiotic human-machine interaction is facilitated by the inherent low weight and compliance of the device. The work in this paper is broadly applicable to a wide variety of wearable assistive devices and in particular will be directly applicable to the next generation Warrior Web suit being developed by DARPA under BAA-11-72. Lastly, while we considered healthy gait during device development, other applications for this design methodology include assisting the elderly, rehabilitation for children and adults with disorders such as Cerebral Palsy. In these applications, rather than augment healthy performance, the system has the potential to provide assistance for limited function, where smaller forces have the potential to achieve greater changes in performance.

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