Biomechanical and Physiological Evaluation of Multi-Joint Assistance With Soft Exosuits

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***Abstract*—To understand the effects of soft exosuits on human loaded walking, we developed a reconfigurable multi-joint actuation platform that can provide synchronized forces to the ankle and hip joints. Two different assistive strategies were evaluated on eight subjects walking on a treadmill at a speed of 1.25 m/s with a 23.8 kg backpack: 1) hip extension assistance and 2) multi-joint assistance (hip extension, ankle plantarflexion and hip flexion). Results show that the exosuit introduces minimum changes to kinematics and reduces biological joint moments. A reduction trend in muscular activity was observed for both conditions. On average, the exosuit reduced the metabolic cost of walking by 0.21 $\pm$ 0.04 and 0.67 $\pm$ 0.09 W/kg for hip extension assistance and multi-joint assistance respectively, which is equivalent to an average metabolic reduction of 4.6% and 14.6%, demonstrating that soft exosuits can effectively improve human walking efficiency during load carriage without affecting natural walking gait. Moreover, it indicates that actuating multiple joints with soft exosuits provides a significant benefit to muscular activity and metabolic cost compared to actuating single joint.**

***Index Terms*—Assistive robotics, biomechanics, human–robot interaction, soft exosuits.**

I. Introduction

LOWER-LIMB wearable robots have been proposed as a means to augment or assist human locomotion for many applications [1]. Some wearable robots have been designed to augment the locomotion performance of able-bodied individuals through enabling wearers to spend less energy or carry an increased load [2]–[6], while others have been designed to enable individuals to walk if they were not able to previously [7]–[9], or supplement rehabilitation for individuals with chronic stroke-related impairments [10]–[18]. While these are impressive achievements, the rigid nature of existing exoskeletons presents a number of challenges towards the goal of enhancing or restoring locomotion: 1) they may interfere with joint movement due to misalignment between the robot and biological joints and thus cause the wearer to deviate from their natural motion patterns [19], [24], [25]; 2) they may have large inertias or bulky self-aligning mechanisms which increase the metabolic cost associated with wearing the system [20]–[23]; and 3) they may impair the mobility of the wearer.

To alleviate some of these challenges, there has been a recent push to develop next generation wearable robots—exosuits—that use soft materials such as textiles and elastomers to provide a more conformal, unobtrusive and compliant means to interface to the human body [26]–[29]. Exosuits use textiles to interface with the body and apply joint torques via tensile forces in parallel with the muscles in order to reduce the required muscular activation. Previous work by our lab has demonstrated that exosuits can be used to provide controlled torques at the ankle and hip of up to 30% of the biological torques during unloaded walking [26], [29]. One study showed that power could be delivered effectively to the wearer, with a multi-articular ankle/hip exosuit creating ankle plantarflexion and hip flexion torques providing a gross metabolic reduction of $5.1 \pm 3.8\%$ comparing the system on versus off [29]. However, it is still unclear how exosuits affect the kinematics, kinetics and muscular activity during locomotion, or if greater metabolic reductions can be achieved with more optimized systems. Moreover, in the field of both rigid exoskeletons and soft exosuits, previous studies have focused on a single architecture (i.e., full leg system, single joint system etc.). In particular, several groups have performed studies with ankle exoskeletons where they have applied various actuation profiles and evaluated metabolic, kinetic, and kinematic effects on locomotion [30]–[32]. Several groups have found that applying ankle torques can provide a metabolic benefit for tethered [32]–[34] or untethered [29], [35] locomotion. There have also been a few studies with various research platforms that have looked at the effects of applying torques at the hip on various biomechanical and physiological variables [16], [36]–[38]. While a small number of studies has evaluated the effects of wearing a full leg exoskeleton [2], [39], the authors are unaware of any studies that compare the effect of assistance applied to multiple joints on biomechanical and physiological variables to assistance applied at a single joint.

In order to enable rapid experimentation with different exosuit architectures and control strategies, we developed a system

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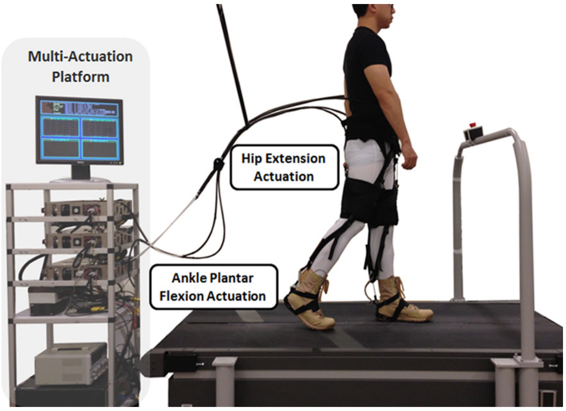


Fig. 1. Multi-joint actuation platform with ankle and hip soft exosuit. Six actuators are packed in the actuator boxes on the top three layers of the multi-actuation platform.

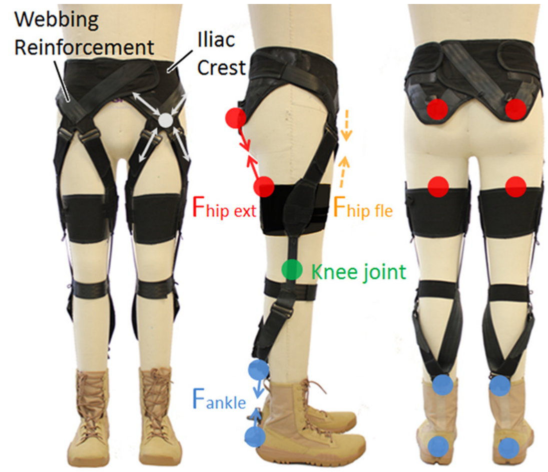


Fig. 2. Ankle-hip soft exosuit, ankle joint anchor points (blue), hip joint anchor points (red). Hip flexion (orange) force is generated by the ankle joint assistance due to multi-articular suit structure.

with a Bowden cable transmission (shown in Fig. 1) that can be mounted next to a treadmill and connected to lower extremity soft exosuits as previously reported in [27]. Using a tethered actuation system is ideal for basic scientific studies to understand how assistance can best be provided to the wearer of a particular exosuit. Experiments can be rapidly performed without having to account for how actuators are placed on the wearer and powerful actuators can generate a wide range of assistance strategies. In addition, the actuation and control strategies developed on this platform can inform and provide specifications for the design of portable body-worn systems.

In this paper, we present an experiment using this tethered actuation system with a multi-joint soft exosuit that examines the effects of applying assistance in two different conditions during loaded walking. The first is applying assistance to only hip extension (hip extension assistance) and the second is applying assistance to hip extension, ankle plantarflexion and hip flexion (multi-joint assistance). The rationale for exploring these joints is that they have been shown to be the largest energetic contributors to forward propulsion during human locomotion [40] and our ultimate goal is to create systems that can reduce the metabolic consumption for the wearer. We collected kinematic, kinetic, muscle activation and metabolic consumption data for these two assistance strategies. This study is the first of its kind that examines the effects of assistance with a soft exosuit on all of these variables. As expected, applying assistance to multiple joints demonstrated a greater reduction in the metabolic cost of the wearer. Furthermore, this experiment also shows the benefits and efficiency of the soft exosuit work input and torque support on human energetics, mechanics and muscle activation.

II. System Design

A. Soft Exosuit

A soft exosuit consists of an integrated garment that includes the attachment points to the body, a structured textile that transmits loads across the body, and anchor points on either side of a biological joint that connect to the actuation units. The goal of an exosuit is to create a secure and comfortable interface that allows the actuation units to transmit forces over the body through beneficial paths such that biologically appropriate moments are created at the joints.

The exosuit, shown in Fig. 2, is an integration of the multi-articular ankle/hip suit and monoarticular hip suit described in [27], combined with improvements to each suit. The suit can transfer forces through either load path independently or in a combined manner. It connects to the body at the pelvis, heel, and thigh, and is constructed of woven textiles oriented in specific directions and reinforced with seatbelt webbing (Seatbelt Planet, Inc.) in this embodiment.

The multi-articular ankle/hip load path, which creates ankle plantarflexion and hip flexion torques on the body with a single actuator as previously reported in [26] and [29], consists of the waist belt and webbing straps that extend from the front of the hip down the sides of the leg to the back of the calf. These straps are held in position down the leg by webbing passing in front of the shin and elastic at the top of the calf. The Bowden cable sheath is attached to the webbing at bottom of the calf, and the inner cable extends further to a shoe attachment bolted to the heel of a military boot (blue circles in Fig. 2).

When the cable is actuated, it shortens the distance between those two points, generating force along the entire length of the load path. Due to the cable passing a distance behind the ankle, force in the suit generates an ankle plantarflexion torque. Because of its multi-articular structure, the exosuit creates a hip flexion torque that occurs concurrently with the ankle torque. The suit is actuated when it can generate beneficial moments on both joints simultaneously, which occurs during 30%–65% of the gait cycle (GC). During this stage of the gait, the calf muscles and tendons push the body up and forward, the hip muscles and ligaments swing the leg forward. Since the suit is intentionally designed to pass through the center rotation of the knee, the kinetics of the knee joint will not be significantly affected [28].

The monoarticular hip load path, assisting hip extension, extends from the waist belt to the thigh braces that surround the leg. Bowden cables are attached between the back of the waist belt and the back of each thigh brace (red circles in Fig. 2). The cable is actuated from 0%–30% GC, during which time the hip

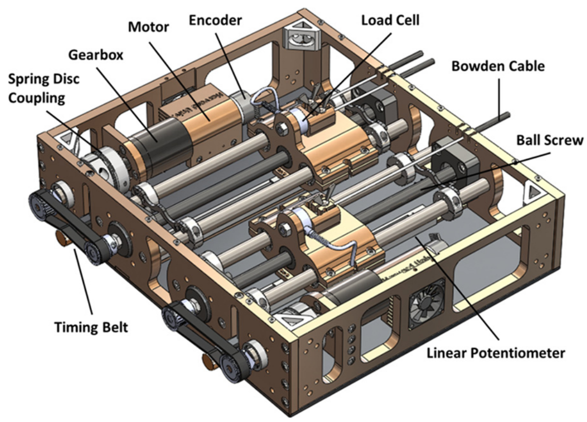


Fig. 3. Modular designed reconfigurable Bowden cable actuation box. Two sets of ball screw-based linear actuators were designed in one actuation modular. The aluminum carriages threaded on the ball nut transmit linear movement to actuate biological joints through the Bowden cable [27].

and thigh muscle groups prevent the body from collapsing under the loading impact and assist forward progression. The cable actuating creates a hip extension torque to assist these muscles.

B. Multi-Joint Actuation Platform

It is well accepted that much basic research remains to be done to understand how to most effectively apply assistance to human walking. To understand the capacity of a soft exosuit to improve human performance in an efficient way, a multi-joint actuation platform was designed and built in a modular form to be able to be quickly attached to a soft exosuit [27]. The multi-joint actuation platform, shown in Fig. 1, consists of three modules each with two linear actuators which are designed to provide biologically realistic joint torques either separately or simultaneously through Bowden cable transmission with a soft exosuit. With this platform, we can rapidly test and optimize the new design of the exosuit and validate the control strategies via human-in-the-loop experiments and a total of four of the linear actuators were used for the purpose of this study.

The Bowden cable sheath is fixed to the actuator frame and the inner cable is connected to the aluminum carriage on the ball screw. Geared motors drive the ball screw to drive the position of the Bowden cable in order to generate tension in the exosuit or go slack when desired. The CAD design of the actuator box is shown in Fig. 3 and a detailed description of its design is provided in [27].

Motor controllers, data acquisition card, power supply and a target PC were installed in a tower and used to measure the sensor signals and control the actuators. Using a real-time control architecture, it was previously demonstrated that desired assistive forces can be delivered to the human joints through soft exosuits with either force or position control [27]. The actuation platform has integrated sensors that monitor voltage, current, force and position at the actuator-side and force sensors in the exosuit to monitor local forces so that consumed electrical energy and delivered mechanical energy can be calculated, which allows us to study the system efficiency.

III. Methods

This section provides details of the design of the assistive controller which was implemented in the human subject test protocol. The human subject experiment protocol and data analysis methods are also described in this section.

A. Assistive Controller

To transmit assistive torques in synchrony with the underlying muscles, it is important that the gait detection is precise and the delivered force is consistent during experiments. A gait detection algorithm was developed based on data from a gyroscope mounted on the back of the boot. The first peak of the angular velocity of the foot from each stride is detected which was shown to correspond to 4% of GC when comparing to lab-based motion capture data. The gait percentage (P_{gait}) is calculated as follows:

$$P_{\text{gait}} = \frac{t}{t_{\text{ave}}} \cdot 100\% + 4\% \tag{1}$$

where t is the past time from the first peak of the angular velocity of the foot from the current stride. t_{ave} is the average stride time calculated from the previous two steps. This estimation method is only valid for lab-based tests on a treadmill, where the variability of stride time is low. More dynamic conditions such as outdoor walking may require a more robust method. A robustness test comparing both foot switches and gyroscopes was performed on an instrumented treadmill at different walking speeds. The ground reaction force was used to segment the collected foot switch and gyroscope signals. The results showed that a heel strike event detected using a foot switch varied from 0% to 5% GC while using a gyroscope, the timing of 4% of the GC could be detected with only a 1% variation. Combined with an iterative force-based position controller described in [28], consistent assistive forces were able to be delivered to both the multi-articular ankle/hip load path and the hip extension load path.

For the multi-articular ankle/hip load path, the actuators are held at a fixed position until 36.5% GC, during which time tension was developed in the suit due to the body's motion [see Fig. 4(a)]. The actuators then execute a position trajectory, pulling on the suit, thus generating a force proportional to the suit-human series stiffness [29]. By implementing this iterative controller, the maximum magnitude and offset of the position profile would increase if the desired peak/pretension force was not achieved or decrease if the desired peak/pretension force was not reached from the last step, so that future steps would have the desired peak force magnitude and specified predefined pretension force at 36.5% GC as described in [28]. For the position trajectory we chose, shown in Fig. 4(a), the resulting force profile had a peak around 50% GC, which is very close to the peak of the biological ankle plantarflexion torque (51% GC) and the peak of hip flexion torque (55% GC). While the force begins at around 18% GC, a time earlier than the start of the biological hip flexion torque (30% GC), the force is low ($<$ 20 N) during this time. The force profile was initiated in this manner to make the force-based position controller more robust, and because removing the slack from the suit permits

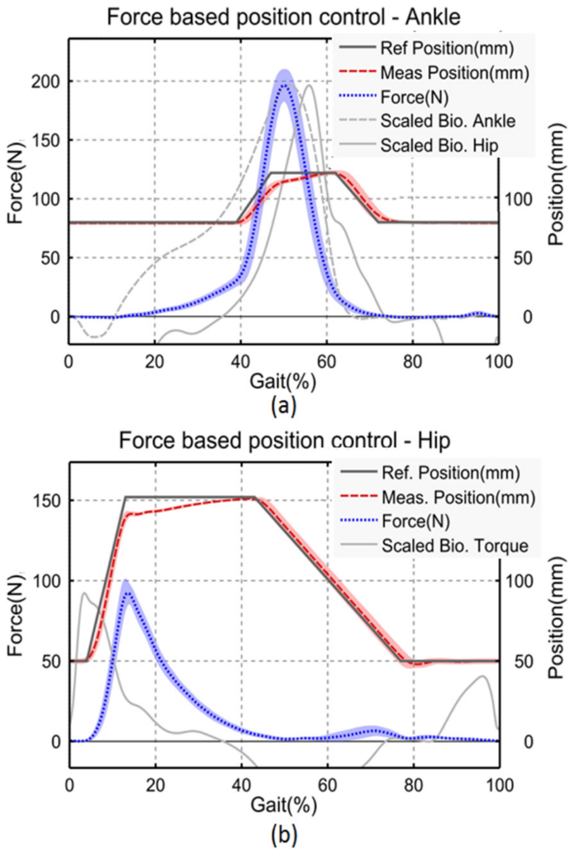


Fig. 4. (a) Ankle joint force and actuator position profile, as well as the biological hip and ankle torques scaled to match the peak force. The hip torque profile is inverted so a flexion torque is positive. The force generated before 36.5% GC is due to the body's motion. (b) Hip joint force profile, actuator position trajectory, and biological hip moment scaled as in (a). Scaled biological joints torque were averaged biological joint data of unloaded baseline walking from our internal speed load study.

the actuator to achieve a higher peak suit force. With this control scheme, the resulting force profile was found to be consistent, with a peak standard $\pm$ deviation of 196 ± 14 N. At 36.5% GC, the force profile had a value of 25 ± 3 N.

For the monoarticular hip extension load path, a similar approach was adopted. The gyroscope peak at 4% GC was used to trigger the hip actuators' position trajectory. While the biological hip extension moment begins prior to heel strike, the gyroscope peak was used to ensure that the actuation timing was very well synchronized to the gait even though it was slightly delayed from the biological moment. The resulting force profile ramped up around 5% GC and reached a peak of 95 ± 10 N around 12% GC as shown in Fig. 4(b). To keep the force consistent between steps, the cable position offset was adjusted so that the suit force was always lower than 5 N during the swing phase in order to not interfere with the body's motion and to minimize the required actuator range of motion during the pull.

B. *Human Subject Experimental Protocol*

To evaluate the effects of our assistive exosuits on locomotion, we conducted an experimental protocol on eight healthy male subjects (age 28.5 ± 6 y.o., weight 78.5 ± 9.9 kg, height 1.76 ± 0.06 m, shank length 0.40 ± 0.03 m, thigh length 0.41 ± 0.03 m, trunk length 0.43 ± 0.02 m) walking on an instrumented treadmill (FIT Bertec, Columbus, OH, USA) at a fixed speed of 1.25 m/s while carrying a 23.8 kg weighted backpack. The study was approved by the Harvard Medical School Institutional Review Board. Consent was obtained from experimental participants after the nature and possible consequences of the exosuit studies were explained.

The protocol was split into a training day and a testing day in which data collection was performed. On the training day, subjects were fitted with the device and familiarized with the system by walking with both assistance conditions for 15 minutes each.

The testing session consisted of four walking bouts: a warm-up, an unloaded baseline walking, followed by two loaded trials for each assistive condition (hip extension assistance and multi-joint actuation assistance). The subject wore the soft suit during the entire test. A 20-minute warm-up bout allowed subjects to acclimate to both assistive strategies for ten minutes each. Subjects then walked for nine minutes with no load and no assistance to provide a baseline measurement ("unloaded mode"). The effect of the suit was evaluated with two 15-minute bouts with load (one for each assistance condition), split into six minutes of walking with the device turned off ("slack mode") and nine minutes with the device turned on ("active mode"). The order in which the two assistance conditions were presented was randomized. Walking bouts were interleaved by resting periods of five minutes. A five-minute quiet standing trial was performed as a baseline at the end of the protocol.

It should be noted that the exosuit in active mode was not compared to not wearing the exosuit in these studies. We are assuming that the slack mode is equivalent to normal walking since the leg-worn components (textile and cables) are lightweight and do not restrict the movement of the joints and the actuation components are offboard to the wearer. This feature of the exosuit really enables efficient experimentation where different conditions can be rapidly compared without the need to take a system off or to readjust markers or EMG electrodes.

Kinematic data were collected through a nine-camera Vicon optical motion analysis system (Oxford Metrics, Oxford, U.K.) using a 75-marker full-body protocol [41], [42]. Muscle activity for eight muscle groups was recorded by means of a Delsys Trigno (Delsys, Natick, MA, USA) wired surface EMG system. The activity of the following muscles was recorded: *Soleus* (SOL), *Gastrocnemius Medialis* (GAS), *Tibialis Anterior* (TA), *Rectus Femoris* (RF), *Vastus Lateralis* (VL), *Vastus Medialis* (VM), *Biceps Femoris* (lateral hamstring, HAM), *Gluteus Maximus* (GM). Electrodes were placed by trained physical therapists following SENIAM guidelines [43]. These muscle groups were chosen in order to highlight potential differences between the different assistance conditions on lower body joints. The metabolic cost of walking was measured through a portable pulmonary gas exchange measurement device ($K4b^2$, COSMED, Rome, Italy).

C. *Data Analysis*

Kinematic data were analyzed using Vicon Nexus (Vicon, Oxford, U.K.) and Visual 3D (C-motion Inc., Rockville, MD, USA). Raw marker and ground reaction force (GRF) data were filtered (fourth-order low-pass Butterworth filter, 12 Hz cutoff frequency) prior to processing. Joint kinematics, moments and powers were calculated using inverse models. All data except joint kinematics were normalized by each participant's body mass. It is worth noting the joint moments computed by the inverse dynamic model represent a combination of the muscular joint torque generated by the subject and of external torque generated by the exosuit.

Joint kinematic, kinetic and power data were segmented and normalized to 0%–100% of the GC, as defined by consecutive heel strikes of the right foot. Gait events (heel strikes and toe-offs) were identified from ground reaction forces and marker data. Kinematic and kinetic data were analyzed for the last minute of walking in each condition, and the average across all measured strides in each condition was used for analysis. In order to compare kinematic profiles across the slack and active modes as well as the two difference assistance conditions, we extracted the minimum and maximum angles for each joint. For kinetics, minimum and maximum moments were extracted for each joint, whereas to compare powers, we extracted positive average power and negative average power.

Electromyographic signals (EMG) were processed to extract the average linear envelope during GC. Raw EMG signals were filtered with a fourth-order band-pass Butterworth filter with cutoff frequencies of 20–400 Hz in order to remove electrical noise and biological artifacts. Signals were then rectified and low-pass filtered (fourth-order low-pass Butterworth, 12 Hz) to extract the linear envelope [44]. Since the amplitudes of EMG signals were different with different subjects, EMG amplitude was normalized by the average of corresponding EMG peak of the baseline walking. Linear envelopes for each muscle group were segmented (heel strike to consecutive heel strike) and normalized to each GC. The root mean square (RMS) was calculated from each normalized curve from the last minute of walking data and the values for the slack and active modes were compared.

The metabolic cost of walking was evaluated by measuring the O_2 and CO_2 gas exchanges between the subject and the environment. The normalized metabolic cost of walking was estimated by applying Brockway's standard equations [45] and by taking the average value of the last two minutes of each walking or standing trial.

It is worth noting that kinematics, kinetics and EMG data that could not be segmented properly, due to the subject stepping in the middle of both force plates on the split-belt treadmill, were excluded from the analysis. On average, 40 strides out of 51 strides were used for data analysis for each trial.

D. Statistical Analysis

Standard errors (SEM) were calculated for EMG RMS, delivered power and metabolic power. Repeated measure analyses of variance (ANOVA) with three modes (unloaded, slack and active) for both assistance conditions were used to verify the effect of load and actuation (slack versus active mode) on each dependent variable. *Post hoc* tests were performed to identify differences when statistically significant effects were identified by the ANOVA.

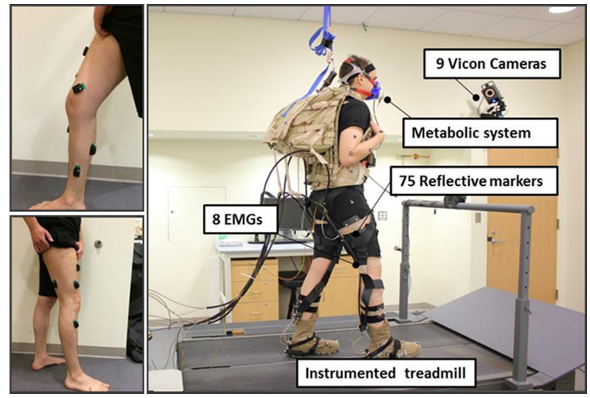


Fig. 5. Lab facilities for human subject experimental protocol and EMG sensor placement.

Paired t-tests were also conducted to test if the changes in EMG and metabolic cost between slack and active modes were different for each assistance condition. The significance level was set at 0.05 for all analysis.

IV. Results

A. Kinematic Analysis

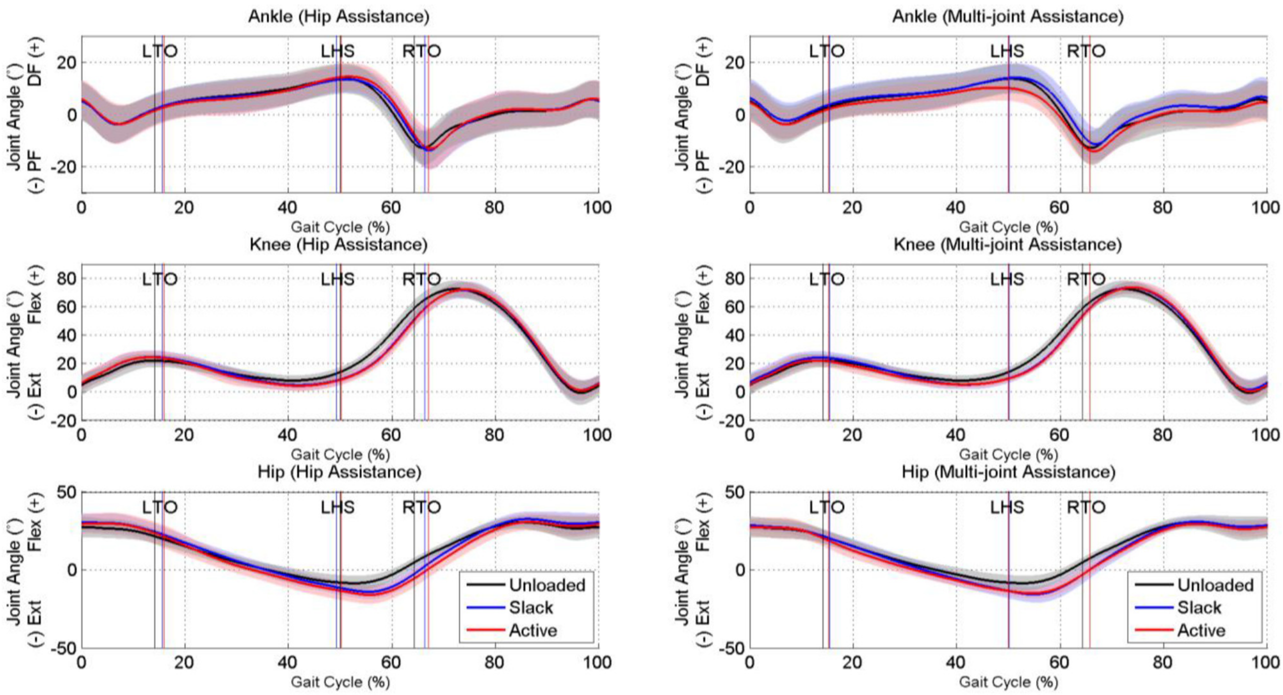
Fig. 6 shows the average kinematic profiles from the eight subjects for the ankle, knee and hip joints. Data is presented in terms of average (solid line) and standard deviation (shaded area) for the three conditions tested: unloaded baseline (black), slack (blue) and active (red). The data is presented for both active modes (hip assistance and multi-joint assistance). The average step frequency among subjects did not change across all walking conditions (average frequency: 130 ± 7 steps/min).

The effect of carrying the load is most apparent at the hip, which showed an increased ROM and peak hip extension in loaded walking (slack mode) compared to unloaded walking. No statistically significant changes were seen at the ankle and knee joint angles between unloaded and slack mode.

A comparison of the active mode to the slack mode, in both hip assistance and multi-joint assistance conditions, shows minimal statistically significant changes in gait kinematics introduced by the exosuit (minimum and maximum angles, and ranges of motion for each joint are all unchanged). However, the peak hip extension angle was found to increase only in hip extension assistance (maximum hip extension: slack $-14.6 \pm 4.7^\circ$, active $-16.5 \pm 5.8^\circ$, $p = 0.015$). The peak ankle plantarflexion angle was increased while peak dorsiflexion angle was decreased only in multi-joint assistance condition (maximum ankle plantarflexion: slack $-15.0 \pm 3.6^\circ$, active $-17.3 \pm 3.7^\circ$, $p = 0.011$; maximum ankle dorsiflexion: active $7.3 \pm 3.0^\circ$, slack $10.4 \pm 3.1^\circ$, $p = 0.014$). In summary, in terms of gait kinematics, the exosuit introduced minimum changes to the wearer's natural movement.

B. Kinetic Analysis

Figs. 7 and 8 present the average joint moments and average joint powers from the eight subjects for the different walking modes and assistance conditions.



(a) (b)

Fig. 6. Ankle, knee and hip kinematics data of the right lower limb for different walking conditions. LTO = toe-off of the left lower limb, LHS = heel strike of the left lower limb, and RTO = toe-off of the right lower limb. PF = Plantar Flexion, DF = Dorsiflexion, Ext = Extension and Flex = Flexion.

As expected, the effect of the load (comparing unloaded to slack) was found to increase peak joint moments for all joints with the exception of hip extension. Similarly, the average power for all of the joints was increased when comparing the unloaded and slack modes.

A comparison of the active mode to the slack mode for both conditions showed no statistically significant changes in the joint moments, agreeing with findings from previous studies by other groups [37], [46].

Minimal statistically significant changes in average power between the slack and active modes were found for the hip extension assistance condition. The only case with statistical significance was for the average positive power at the knee, which was increased compared to slack (slack 0.211 ± 0.077 W/kg, active 0.237 ± 0.077 W/kg, $p = 0.011$). For the multi-joint assistance condition, both the positive and negative knee average power were reduced (positive average power: slack 0.208 ± 0.08 W/kg active 0.19 ± 0.073 W/kg, $p = 0.021$; negative average power: slack -0.401 ± 0.063 W/kg, active -0.364 ± 0.063 W/kg, $p = 0.001$). In addition, the negative ankle average power was significantly reduced (slack -0.215 ± 0.057 W/kg, active -0.149 ± 0.043 W/kg, $p = 0.022$).

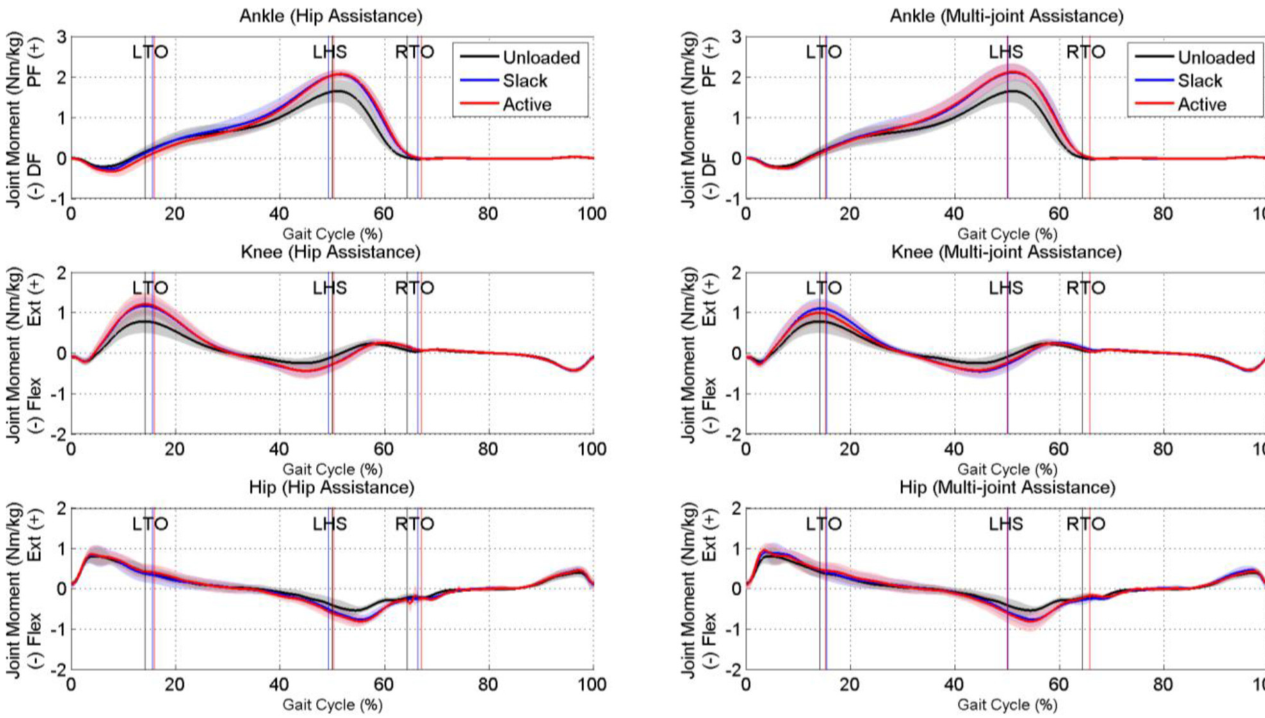
In summary, the sum of the joint moment remained the same and the exosuit introduced minor changes on knee and ankle power.

C. EMG Analysis

Fig. 9 presents the change in the RMS of the average EMG envelopes on all measured muscle groups, in both active modes, compared to slack modes. A reduction trend in the EMG activity was observed for the majority of the muscles except for SOL and HAM in the hip extension assistance condition and TA in the multi-joint assistance condition. Multi-joint assistance reductions were greater on every measured muscle, with the exception of the GM and the TA. However, no significance was found with pairwise comparisons between the slack and active modes from the *post hoc* test.

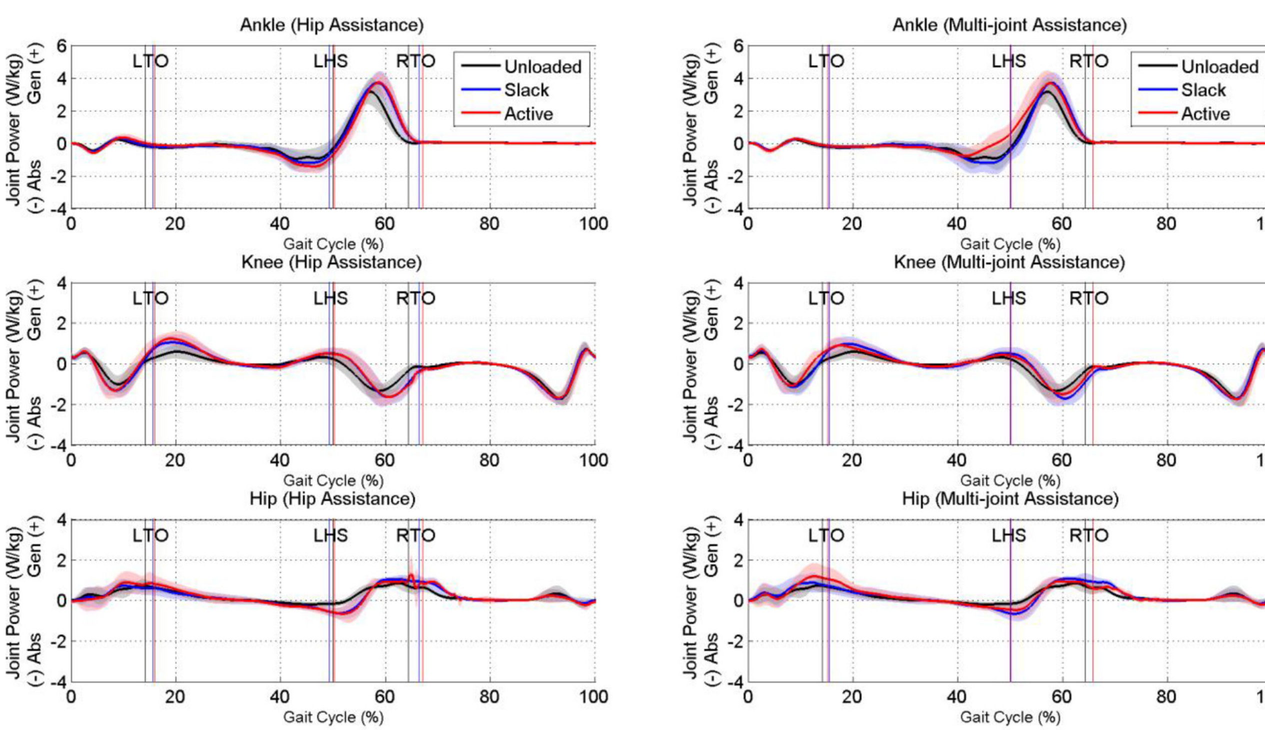
D. Metabolic Analysis

Both assistance conditions resulted in a reduction of the metabolic powered required to walk at 1.25 m/s with a 23.8 kg load when comparing the slack and active modes. Standing required 1.83 ± 0.28 W/kg and walking without assistance required 6.44 ± 0.83 W/kg, agreeing with previous findings [47], [48]. On average, the exosuit reduced the metabolic cost of walking by 16.1 ± 2.9 W and 52.5 ± 6.8 W for hip extension assistance and multi-joint assistance, respectively. Table I shows the average metabolic power normalized by subject's weight in the different tested conditions. Table II shows the average delivered mechanical power for each stride including the estimated hip flexion power, which was generated by the force transferred to the hip through the multi-articular ankle/hip load path when ankle joint was actuated. To characterize the transferred hip flexion moment, the exosuit was instrumented with two additional load cells, one on each leg strap connect to the waist belt. The peak force at the waist belt was compared to that at the ankle joint during walking and found to be approximately 75% across subjects as described in [49]. Therefore, for simplicity, the force assisting the hip during flexion was assumed to be 75% of the force measured by the load cell at the ankle. To estimate the passive hip flexion



(a) (b)

Fig. 7. Ankle, knee and hip moment data of the right lower limb for different walking conditions. LTO = toe-off of the left lower limb, LHS = heel strike of the left lower limb, RTO = toe-off of the right lower limb. PF = Plantar Flexion, DF = Dorsiflexion, Ext = Extension and Flex = Flexion.



(a) (b)

Fig. 8. Ankle, knee and hip power data of the right lower limb for different walking conditions. LTO = toe-off of the left lower limb, LHS = heel strike of the left lower limb, RTO = toe-off of the right lower limb. ABS = Absorption, GEN = Generation.

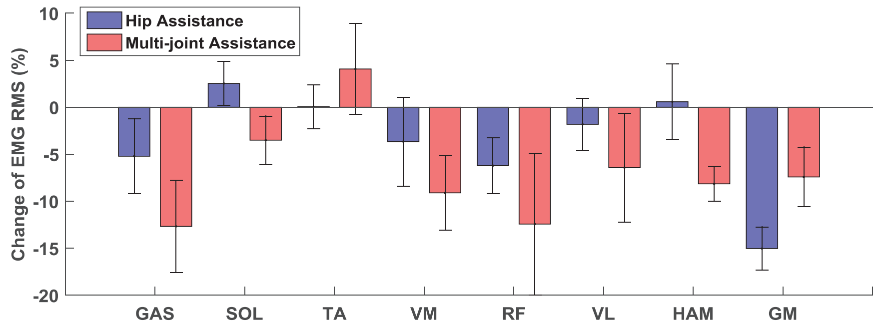


Fig. 9. Improvement (mean ± SEM, N = 8) in the EMG RMS during hip assistance and multi-joint assistance compared to normal loaded walking (slack). GAS = *Gastrocnemius Medialis*, SOL = *Soleus*, TA = *Tibialis Anterior*, VM = *Vastus Medialis*, RF = *Rectus Femoris*, VL = *Vastus Lateralis*, HAM = *Biceps Femoris*, GM = *Gluteus Maximus*. Negative value represents reduction.

TABLE I
METABOLIC POWER AND METABOLIC REDUCTION ON HIP AND MULTI-JOINT ASSISTANCE FOR EIGHT SUBJECTS. VALUES ARE MEAN ± SEM, N = 8. P VALUE IS ACQUIRED FROM REPEATED MEASURES ANOVA, P $<$ 0.05 INDICATES STATISTICAL SIGNIFICANCE

Assistance Strategies	Standing (W/kg)	Slack (W/kg)	Active (W/kg)	Reduction (%)	P Value
Hip Assistance	1.84 ± 0.1	6.44 ± 0.32	6.23 ± 0.29	4.6 ± 0.8	0.006
Multi-joint Assistance		6.45 ± 0.36	5.80 ± 0.36	14.6 ± 1.5	0.001

TABLE II
AVERAGE DELIVERED MECHANICAL POWER AND PEAK FORCE ON HIP AND MULTI-JOINT ASSISTANCE ON EACH LEG FOR EIGHT SUBJECTS. VALUES ARE MEAN ± SEM, N = 8

	Ankle Plantarflexion			Hip Flexion (Estimated)			Hip Extension		
	Pos. (W)	Neg. (W)	Peak Force (N)	Pos. (W)	Neg. (W)	Peak Force (N)	Pos. (W)	Neg. (W)	Peak Force (N)
Hip Assistance	*NA	*NA	200	*NA	*NA	140	7.04 ± 0.70	-0.86 ± 0.12	100
Multi-joint Assistance	7.30 ± 1.55	-1.53 ± 0.19		3.6	-4.5		6.40 ± 0.58	-0.82 ± 0.08	

moment generated by the suit, this force was multiplied by a constant moment arm of 10.7 cm.

The hip assistance condition showed a net metabolic reduction of 4.6%, corresponding to 0.21 W/kg ($p = 0.006$), whereas the multi-joint assistance showed a net metabolic saving of 14.6%, corresponding to 0.67 W/kg ($p = 0.001$).

V. DISCUSSION

This paper presents the first detailed evaluation of the biomechanical and physiological effects on subjects wearing a soft exosuit intended for gait assistance during load carriage. This study was performed in a lab setting using an offboard multi-joint actuation platform [27] that allowed two different assistive conditions, hip extension assistance and multi-joint assistance, to be evaluated so as to gain insight into the effects on assisting multiple joints with a soft exosuit. Significant reductions in the metabolic cost of walking and a reduction trend in muscular activity were shown, demonstrating the potential for future body-worn soft exosuits to provide assistance during loaded walking.

In terms of kinematics, the main change between slack and active mode in the hip extension assistance condition was an increased peak hip extension angle. This may suggest that on average subjects had less forward lean during the pre-swing phase, which has also been reported by test subjects. The increased hip extension may be due to the applied hip torque being somewhat later in time than the biological hip torque [see Fig. 4(b)]. This later torque application may also explain why more positive power was generated at the knee (considering the combined effect of the suit plus the human). The knee is in a serial chain with the hip joint, and so applied hip extension torques should also lead to knee extension.

In contrast, increased peak hip extension during pre-swing was not observed between slack and active mode in the multi-joint actuation condition. A possible explanation for this may be that the multi-joint actuation generates a hip flexion torque that counteracts the hip extension assistance during 20%–45% in the GC when both torques were applied simultaneously. This effect may have also contributed to the decreased knee power generation at around 20% of the GC.

Actuating only hip extension was found not to significantly change the kinematics at the ankle joint. However, for the multi-joint assistance condition, peak ankle plantarflexion was increased and peak ankle dorsiflexion was decreased. This makes sense as the exosuit provides energy absorption during terminal stance, which may lead to less energy absorption at the wearer's ankle joint. The elastic energy storage mechanism on ankle joint is mainly Achilles tendon, thus less energy absorption on the ankle joint may imply less elongation of the Achilles tendon. This may be the reason for a decrease in peak dorsiflexion angle since it seemed that less elongation was required from Achilles tendon to store the energy. The increased peak plantarflexion angle for the ankle may also explain the reduced power absorption at the knee, since the knee may have been required to bend less in order to provide the necessary toe clearance during swing.

The ankle plantarflexion torque created during the multi-joint assistance condition may also explain the decrease in positive and negative power absorption at the knee. When the ankle plantarflexes, the body is propelled upward and forward. This could contribute to decreased power absorption and generation on the contralateral leg's knee joint early in the GC.

In terms of kinetics, besides the changes mentioned above, it is worth noting that the total joint moments of ankle, knee and hip joints were found to be invariant across the slack and active mode for both the hip extension and multi-joint assistance conditions, agreeing with results from previous studies from other groups [37], [46]. Given this finding, it is likely that this resulted in a reduction in the biological joint moment (and corresponding muscle force), if we assume the forces in the exosuit contributed around 12% and 14%, based on the moment arms at the ankle and hip joints. Although no significant changes in muscle activity were found, comparing differences in the activity of the eight muscles in the slack and active mode for the multi-joint assistance condition shown in Fig. 9 shows a trend that is promising. Measurements of muscle activity using surface EMG is challenging with a soft exosuit (due to interference of the textile with the electrodes) and the resulting large standard deviations and limited number of subjects in this study is likely the reason statistical significance was not found. Comparing the EMG reductions in the hip extension and multi-joint assistance condition, similar trends can be seen for the key muscles that would be expected to be reduced in early stance. Further, the effects for both conditions were found to be similar, implying that additionally assisting ankle plantarflexion and hip flexion in the multi-joint assistance condition did not interfere with the hip extension assistance.

For hip extension assistance, the greatest effect was seen in the activity of GM due to hip extension assistance during the loading response phase. This may also reduce co-contractions of muscles in the quadriceps group including RF, VM, and VL. Assistance provided for hip extension accelerated the thigh, which meant more kinematic energy was needed to be absorbed during the terminal stance phase. This may have led to a slightly higher elongation by the Achilles tendon which could cause increased peak EMG activity for the SOL muscle. As was previously mentioned, the total ankle joint moment remained invariant between the slack and active mode. Since the exosuit was not providing any force to the ankle during this condition, this implies that the biological joint developed the same torque during the slack and active modes. The main active contributors to ankle torque in the body are GAS and SOL, and since a slight increase was observed by SOL, GAS would have to be reduced in order to keep the biological moment unchanged; this is in line with the observed results shown in Fig. 9.

During multi-joint assistance, EMG activity of the GM remained unaltered between the slack and active mode, while a reduction trend in SOL muscular activity can be seen in Fig. 9. In contrast with hip extension condition, the exosuit provided assistive force to the ankle joint from terminal stance to pre-swing, which may result in a force reduction on the calf muscle. This may be the reason the EMG reduction trend on both SOL and GAS was observed. The observed increase in EMG activity in the TA during the swing phase (for providing foot clearance through dorsiflexion) may be as a result of the greater peak ankle angle at push off as the exosuit rapidly actuates the cable and accelerates the foot. As was described, the effect of the multi-articular ankle/hip suit architecture is to also generate a hip flexion torque and the EMG data for the RF, VM and VL showed a reduction trend when comparing the slack and active modes.

Although the EMG reduction is not statically significant, the observed trend of the EMG reduction lead likely shows a general reduction in muscular activity that would be consistent with a statistically significant reduction on metabolic cost. The metabolic cost results are also in line with the EMG results in that reductions for the multi-joint assistance condition were greater than for the hip extension assistance. The average reduction for the multi-joint assistance condition was significantly higher (14.6% versus 4.6%) demonstrating the value in assisting multiple joints with a body-worn soft exosuit. While not very large, the average reduction of 4.6% found that for the hip extension assistance condition is promising and implies that a body-worn monoarticular hip exosuit such as [50] could provide benefit by itself. The metabolic reductions observed in hip extension assistance and multi-joint assistance conditions can be estimated to be equivalent to reducing the payload carried by the wearer during walking by approximately 5 and 14 kg, respectively [51], and thus possibly being able to delay the onset of fatigue in loaded walking.

It should be noted that for this study, the mechanical power values being delivered by the exosuit in each condition were not large (as shown in Table II), and yet significant metabolic benefits were still achieved. We delivered an average of 7.0 and 17.1 W positive mechanical power to the wearer and achieved an average of 16.1 and 52.5 W metabolic reduction with energy converting efficiency (metabolic reduction/positive mechanical power) of 2.3 and 3.1, respectively. Comparing these numbers to those from other wearable robotic devices, 2, 2.5, 2.6 for Sawicki [52], Mooney [35], and Malcolm [32], respectively, highlights the benefits of multi-articular suit structure and not having significant inertia or restrictions on the wearer's limbs in order to allow them to largely maintain their normal passive dynamics. The exosuit architecture and control strategy is such that it targets the three key regions that contributed to forward propulsion during walking (hip extension, hip flexion and ankle plantarflexion). As reported in [26], [28], and [50], ongoing work is focused on developing portable body-worn exosuits and the results of this study hold promise that if their total system mass (with the majority of it worn close to the wearer's center of mass) can be sufficiently low then it should be possible to achieve a net metabolic reduction.

VI. Conclusion

In conclusion, we have presented a biomechanical and physiological evaluation of a multi-joint soft exosuit and compared the effects of providing assistance to either a single joint (hip extension) or multiple joints (hip extension, ankle plantarflexion and hip flexion) during loaded walking. This was made possible through a lab-based multi-joint actuation platform that enables rapid reconfiguration of different actuation conditions, in this case at the ankle and hip joints.

The effects of the different assistance conditions on human kinematics, kinetics, muscular activity and metabolic cost were analyzed. Results showed that the exosuit introduced minimal changes to the wearer's natural gait during walking, implying that the wearer's natural movement is not greatly affected when wearing a soft exosuit. However, while peak values did not show large changes, clear differences in the kinematics and kinetics profiles were observed and future work will explore methods for analyzing the time series data.

Reductions in the metabolic cost of walking of 4.6% and 14.6% were obtained for the hip extension and multi-joint assistance conditions, respectively. A reduction trend in the muscular activity was observed for most muscles, with larger effects observed when assisting multiple joints. Results suggest that providing a moderate amount of assistance with a soft exosuit at the right time to both single and multiple joints can effectively improve walking efficiency when carrying a load. Moreover, assisting multiple joints provides a great benefit to human walking and combines the benefits of actuating single joints independently. The results from the study have been used to generate specifications for a body-worn system capable of providing multi-joint assistance. Future work includes comparing different control strategies and increasing the exosuit assistive forces with the goal of achieving greater reductions in muscle activity and metabolic cost.

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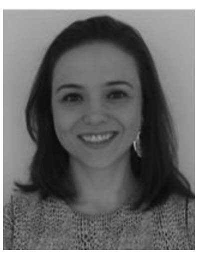

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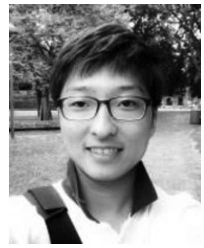

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