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Powered Hip Exoskeletons can Reduce the User's Hip and Ankle Muscle Activations during Walking

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***Abstract*— In this paper, we study the human locomotor adaptation to the action of a powered exoskeleton providing assistive torque at the user's hip during walking. To this end, we propose a controller that provides the user's hip with a fraction of the *nominal torque profile*, adapted to the specific gait features of the user from Winter's reference data [34]. The assistive controller has been implemented on the ALEX II exoskeleton and tested on ten healthy subjects. Experimental results show that when assisted by the exoskeleton, users can reduce the muscle effort compared to free walking. Despite providing assistance only to the hip joint, both hip and ankle muscles significantly reduced their activation, indicating a clear tradeoff between *hip and ankle strategy* to propel walking.**

***Index Terms*— powered exoskeletons, biomechanics, gait, human-robot interaction, assistive and rehabilitation robotics.**

I. Introduction

POWERED exoskeletons are intelligent mechatronic systems designed to improve the performance of the wearer. Starting from the 60's, a large number of powered exoskeletons have been developed targeting either human strength and endurance augmentation [1]-[2], neuro-rehabilitation of motor-impaired patients [3]-[6], or movement assistance of subjects affected by permanent movement disorders, such as hemiplegia [7][8], paraplegia [9]-[11] or tremor [12],[13].

Thorough clinical investigations are still ongoing to prove the usability of these assistive robots; besides technical problems, mainly related to autonomy and portability of assistive exoskeletons, a scientific challenge still remains open i.e., how to control the robot in order to promote the human motor-adaptation and provide an effective assistance to the user.

In this paper, we study the human locomotor adaptation to the action of assistive exoskeletons that provide additional torque at the user's hip, with the goal of reducing the muscle activity during gait while still allowing users to control their joint kinematics.

A muscle effort reduction during walking may be desirable for many persons. Several pathologies can decrease the walking ability of affected persons by reducing their muscle strength and endurance (e.g., myopathies) [14], or lowering the maximum sustainable cardiac effort (e.g., peripheral arterial diseases) [15]. A pathological condition of the lower-limb articulations (e.g., hip pain, osteoarthritis) can result in an onset of pain that reduces the patient's walking ability as well [16]. In either case, by using an assistive exoskeleton that can reduce the muscle force required to walk, patients might recover their normal movement ability and improve their physical and social health [17]-[19]. The growing elderly population increases the incidence of these pathologies, and together with longer life expectancy, motivates the study of non-pharmacological solutions. Assistive exoskeletons may present an alternative way of restoring normal walking in these population groups.

Walking assistance requires a strong synergy between the user and the robot. While walking, the assistive exoskeleton provides the user's joints with supplemental torques. At the same time, the user adapts his muscle activation patterns to exploit the available torques in a convenient way. If the assistance is effective, the motor adaptation process will result in lower muscle forces and thereby a less fatiguing walk for the user. A thorough understanding of the human adaptation process is therefore fundamental for designing an effective assistive device for walking [20].

Recent studies exploiting proportional EMG control and gait event detection have showed that when an assistive torque is provided, humans modulate their muscle activation in order to maintain the total torque profile, i.e., the sum of human muscular torque and assistive torque, unaltered along the gait cycle [21],[22]. As a consequence, the muscle torque decreases and the metabolic effort is reduced [23]

On the other hand, the joint angle trajectory in the assisted condition seems to get modified by the assistance [21],[22]. Similar studies have showed that the adaptation time increases with the level of assistance provided, and seems to be equivalent for the hip and the ankle joint.

Besides motor adaptation, the design of an assistive exoskeleton should consider the walking biomechanics with a focus on the strategy used by humans at the joint and muscle level to support the body weight and propel the body mass during walking [24]-[26]. A critical event is the step to step transition, when the body mass changes its motion direction, i.e. from downward to upward and forward to backwards. This transition can be obtained either by using an "ankle strategy" (i.e. push off of the ankle prior to ipsilateral swing [27]), a "hip flexor strategy" (i.e. pulling the ipsilateral limb into swing [28]) or finally by a "hip extensor strategy" (i.e. hip extensors contract to posteriorly rotate the pelvis and help the contralateral limb progression [29][30]). A tradeoff between these strategies seems to be used by the CNS to produce stable and effective walking. A pathological condition (e.g. diabetes, arthritis) can alter the physiological

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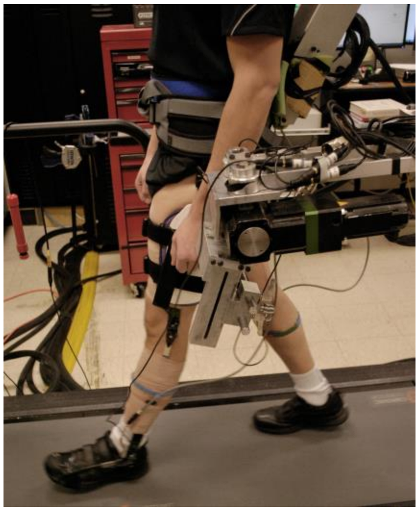

Fig. 1 The ALEX II gait trainer

equilibrium towards one of these concurrent strategies [31]. Interestingly, this balance can also be altered voluntarily, for example, by instructing a healthy subject to exaggerate the ankle push-off [32]. Moreover, wearing a spring-like passive orthosis can alter the hip-ankle equilibrium, reducing the work done by the ankle and increasing the contribution of the hip [33]. These findings have a strong relevance to the design of assistive exoskeletons because they show that the CNS can redistribute the effort needed to produce a stable walking at the joint level.

Our hypothesis is that a similar mechanism of *effort re-distribution* could be used by the CNS when a powered exoskeleton selectively assists one of the user's joints. The external assistance would alter the physiological equilibrium by making one of said *walking strategies* more convenient than the others. As a consequence, a different muscle activation pattern should also emerge for the muscles that do not act directly on the joint for which the robot provides assistance. If this hypothesis is confirmed, by relying on the adaptation of the CNS, a powered exoskeleton providing torque at the user's hip joint could also reduce the effort at the ankle joint by decreasing the *ankle strategy*. This would simplify the structure and control of assistive exoskeletons and provide some more insights into the strategy used by the CNS to exploit robotic assistance for walking.

In this paper, we explore this hypothesis by designing and testing a novel assistive controller that provides additional torque at the hip joint only, while allowing users to control their own kinematics.

Assistive controllers usually define the additional torque provided by the exoskeleton based on an estimate of the user's joint torque [34],[35]. This approach allows having assistive torques that are coherent in direction and amplitude with the mechanical action of muscles. Following this approach, assistive torque profiles have been estimated through electromyography [36]-[38], solving the inverse-dynamic problem [39], or by applying algorithms based on inertia reduction [40], gait segmentation [41] and position-based force fields [42].

In this study we propose an alternative solution based on *nominal torque profiles* that approximate on-line the user's joint torque during walking. While walking at a constant cadence, the human joint torque, normalized by the body weight, follows a characteristic periodical temporal pattern [43], which is quite consistent at the ankle, but presents a high inter-subject variability at the hip and knee. This high variability is however due to stride-by-stride compensations between these two joints, as demonstrated by the high covariance (about 89%) of the hip-knee torque patterns [44]. So the averaged profiles can be considered a basic kinetic pattern of walking anyway.

Based on these observations, we designed an assistive controller that estimates the wearer's walking cadence on-line using adaptive frequency oscillators (AdOs) [45], computes the current percent of stride, and finally provides the user's hip with an assistive torque that is based on a scaled version of the *nominal hip torque profile* extracted from Winter's dataset [43][1]. Such a controller does not require any additional sensors, complex calibration, or estimation algorithms, and can be easily implemented on existing powered exoskeletons.

For the purpose of the experiment, the controller has been implemented on a modified version of the ALEX II gait trainer (Fig. 1) [47] and experimentally tested on ten healthy subjects. Muscle activation and joint kinematics were recorded while subjects walked on a treadmill at a constant speed in three different conditions: free walking (i.e., no exoskeleton worn by the user), zero torque (i.e., exoskeleton working in transparent mode) and assisted (i.e., exoskeleton providing additional torque). Experimental results show that the controller successfully assisted the user as evidenced by a marked decrease of muscle activations, measured through electromyography, compared to free walking. Despite providing assistance only to the hip joint, both hip and ankle muscles significantly reduced their activation, demonstrating a clear tradeoff between these two joints in propelling gait.

Preliminary results have been reported in a conference [48].

II. Methods

This section provides details of the assistive controller and its implementation on ALEX II gait trainer. Then, this section describes the experiment protocol and data analysis.

A. The Assistive Controller

The proposed assistive controller (Fig. 2(a)) is composed of three stages that address the following issues: (1) on-line estimate of the current phase of the gait cycle, (2) planning of the assistive torque, and (3) effective transfer of the desired assistive torque to the user's leg.

The first stage of the assistive controller addresses the estimation of the current phase of the gait cycle. The gait cycle, i.e., stride period, is defined in the controller as the time between two consecutive left heel strike events, while the current phase inside each gait cycle (expressed as a percent of stride period) is obtained as the ratio between the time elapsed from the start of the current cycle, and the expected duration of the cycle. The expected duration of the gait cycle is estimated through an Adaptive Frequency

[1] Note that given the substantial equivalence between over-ground and treadmill walking [46], the Winter's torque profile can be used for experiment performed on treadmill.

Oscillator (AdOs), a mathematical tool that was originally developed for other applications [45] and more recently used for estimating the high-level features of periodic human movements [42],[49]. Resistive foot-pressure sensors are embedded in the user's shoe insoles (see [47] for implementation) and act as switches to detect the heel-strike and toe-off events. By combining the left foot heel strike detection with the estimated cycle-duration, the assistive controller can compute the current stride percent (*Stride %* in Fig. 2(a)).

The estimate of the user's joint torque is based on the value reported on the Winter's tables [43] as a function of the stride percent and the walking cadence. Three different torque profiles are used for slow cadence (86.8 steps/min), normal cadence (105.3 steps/min), and high cadence (123.1 steps/min). Results of pilot studies indicated a harder adaptation of users to extension torques compared to flexion torques. Based on these preliminary observations, we decided to provide less assistance in extension compared to flexion. So we modified the nominal Winter's torque profiles in order to reduce to half the extension torque, while leaving the flexion torque unaltered (final profiles are shown in Fig 2(b)). This modified torque profile resulted in a lower extension assistance, which seemed to improve the feeling of users and facilitate their adaptation. Further studies would be required to understand the reason behind this phenomenon.

These reference torque profiles are implemented on a bi-dimensional look-up table (*2D-LUT*) that takes as inputs the cadence estimate (*cad*) and the current percent of stride (*Stride %*) and gives as output an estimate of the current user's joint torque (*Tn*). The output of the 2D-LUT is then multiplied by the body weight of the subject (*BW*), and subsequently by a factor that allows to regulate the amount of assistance provided by the powered exoskeleton (*Assistance %*). Linear interpolation is exploited to obtain a specific output (*Tdes*) for any possible value of cadence and current percent of stride. Fig 2(c) exemplifies the effect of the linear interpolation showing a mesh having intervals of 0.5 steps/sec and 0.1% for the cadence and stride accomplishment respectively. The output of the linear interpolation on the 2D-LUT (*Tdes*) defines the set point for the closed-loop low-level control, which is in charge of ensuring an effective transmission of the desired torque to the user's leg. This has been achieved by closing the control loop on a direct estimate of the human-robot physical interaction (*Tmeas*) as measured by a six-axis Force/Torque sensor (mini45, ATI Industrial Automation, NC, USA), which is placed at the human-robot interface, between the cuff and the robotic link. In this way, we compensated for the gravity and inertia of the robot, improving the fidelity of the assistive torque. The final interaction-torque controller has a closed-loop bandwidth of 10 Hz, and 0.1 Nm steady-state error.

B. *ALEX II gait trainer*

ALEX II is a treadmill based lower-limb exoskeleton (Fig. 1) developed at the University of Delaware [47]. For the purpose of the experiment, we used a modified version of the ALEX II robot. In this specific version of the exoskeleton, the unilateral robotic leg has only one active degree of freedom driven by geared DC motors, (Danaher Corporation, Washington D.C., USA) to power the left hip joint only in the progression plane. The unilateral robotic leg is composed by

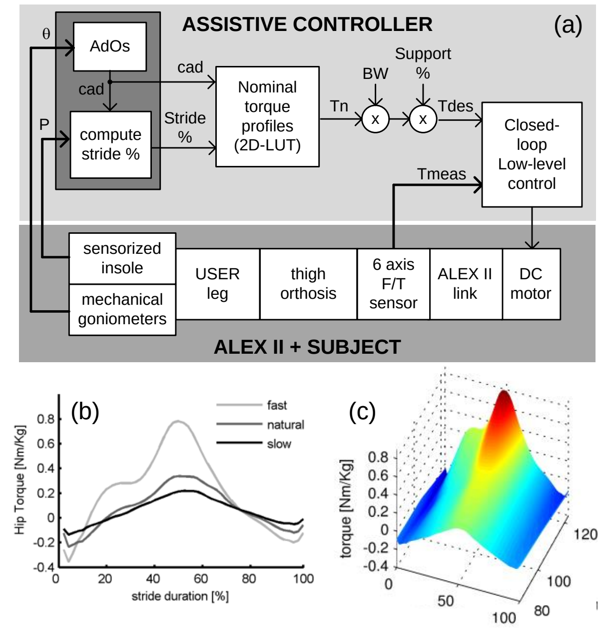

Fig. 2 (a) Block diagram of the assistive controller. The controller takes as inputs the pressure signals, *P*, coming from the sensorized insoles, and the hip angle position, *θ*, measured by the mechanical goniometers. Based on these signals, the controller computes the current gait cadence, *cad*, and the current phase of the gait cycle, *Stride %*. A bi-dimensional look-up table, *2D-LUT*, which implements the Winter's torque profiles, computes the normalized torque, *Tn*, starting from *cad* and *Stride %*. *Tn*, is then multiplied by the subject's body weight, *BW*, and the desired level of assistance, *Assistance %*, in order to define the setting-point of the low level controller, *Tdes*. The desired torque is finally compared to the torque measured, *Tmeas*, by the 6-axis F/T sensor of ALEX II to determine the input for the DC motor;
(b) *nominal hip torque profiles* elaborated from Winter [43] and implemented in the 2D-LUT; (c) Simulation of the linearly interpolated output of the 2D-LUT.

a single link that interfaces the user's left leg at the level of thigh through an orthotic cuff having adjustable size to comply with different users' anthropometry. Hip adduction/abduction is allowed through passive degrees-of-freedom. The robotic leg is supported from the rear, which also attaches to the user. The back support provides configuration-independent gravity compensation for the device [50]. Importantly, the back support is provided with several passive degrees-of-freedom to allow the physiological movement of the pelvis during walking, i.e. antero/posterior, superior/inferior and lateral movement. Thanks to its large workspace, which fully contains the area determined by the treadmill dimensions in the horizontal plane, this mechanism allows the physiological movement of the pelvis during walking without transferring any forces to the user's torso. The real-time control and the data acquisition were managed by a dSPACE 1103 control system (dSPACE GmbH, Paderborn, Germany). For the purpose of the experiment, ALEX II has been modified to interface and assist the movement at the hip only.

C. *Experimental Protocol*

Ten healthy volunteer subjects participated in the experiment. None of them had previously experienced assistive control on the exoskeleton. The participants signed an informed consent before the experiment took place. The protocol was approved by the University of Delaware Institutional Review Board.

EMG activity and joint angular position were measured unilaterally from the left leg only (i.e. the assisted leg). Specifically, surface EMG activity from six muscles of the left leg (Gastrocnemius Medialis, Soleus, Tibialis Anterior, Vastus Lateralis, Rectus Femoris, Semitendonosus) were measured by MA-420 EMG preamplifiers and digitized at 1 kHz using the MA300-28 system (Motion Lab system Inc., Baton Rouge, LA, USA) with an internal band-pass filter (10-500 Hz) and a gain coefficient of 4000. User's joint angular positions were recorded for left hip, knee and ankle flexion-extension by using mechanical goniometers (PASCO, Roseville, CA, USA) independently to the action of the robot (i.e. goniometers were not connected to the robot link in order to avoid any error due to the deformation of the thigh cuff of the robot). Resistive foot pressure sensors were placed both on the left and right insoles and were used as switches to detect heel-strike and toe-off events. Both angle and pressure measures were directly digitized by the ALEX II controller.

The experimental protocol consisted of walking on a treadmill at a constant velocity of 2.4 mph/h (1.07 m/s) under three different conditions.

Free-walking pre: the subject walked for ten minutes without wearing the exoskeleton in order to measure the baseline kinematics and muscle activations.

Zero torque: The subject donned the exoskeleton on the left leg and walked for ten minutes with the robot controlled in transparent mode. In this phase, the desired torque was set to zero. As a consequence, the robot controller minimized the interaction with the user's leg in order to reduce the loading effect of the robot. This session was used to verify the effect of wearing the exoskeleton on the user kinematics and muscle activation. Moreover, it allowed the user to become familiar with the pelvis brace and the leg attachment before the actual assistance experiment.

Assisted condition: After ten minutes from the beginning of the *zero-torque* condition, the controller automatically started providing the assistive torque. For safety reasons, subjects were verbally warned thirty seconds before the onset of the assistance by the experimenter. Starting from zero, the desired level of assistance was gradually increased by regulating the *Assistance %* command (see Fig 2) in order to reach 50% assistance in 50 strides. The gradual increase of the assistance was intended to facilitate the adaptation of the user to the assistive action of the robot. The maximum value of *Assistance %* (i.e., 50) was chosen to provide the user with half the total torque required to walk at the current cadence, as extracted by the Winter's dataset and computed online using the 2D-LUT. This value has been chosen as a compromise between muscle effort reduction and ease of adaptation, based on the results of pilot studies. The *assisted condition* lasted 30 minutes. After this period, the *Assistance %* was set again to zero by the controller. The treadmill was stopped by the experimenter, and the user took off the exoskeleton.

Free-walking post: After resting outside the exoskeleton, a free-walking post-assisted condition lasting five minutes was tested to verify any possible alteration of the baseline values recorded at the beginning of the experimental session.

D. Data Analysis

In order to filter the sensor noise for analyzing the movement kinematics, the actual angular position signals were first offline low-pass filtered (2nd order Butterworth filter, cutoff frequency of 10 Hz), then differentiated to get estimates of the angular velocity. These signals were again smoothed using the same Butterworth filter.

Starting from raw EMG signals, linear envelopes (LE) were computed by full-wave rectification of the band-passed signal (2nd order Butterworth filter, cut-off 10-500Hz) and then low-pass filtering (2nd order Butterworth, cut-off 4 Hz).

For each muscle, EMG data were normalized by the average of the corresponding peak reached in the last minute of the *free walking pre* condition. All data (recorded and derived) were then separated into strides, i.e., the time interval between two consecutive left heel strikes, using the data from the heel-contact sensors located in the insoles. Within each stride, we computed:

1. *cycle duration*, to verify any change in the gait cadence;
2. *gait events*, (left and right heel strike and toe-off) as an indicator of the step symmetry;
3. *flexion/extension peaks of hip, knee and ankle joint position*, to measure the gait kinematics;
4. *EMG Root Mean Square Amplitude (RMSA)* to assess the level of muscle activation and evaluate the effort level over the stride duration;
5. *EMG LE peaks* to evaluate the maximum level of muscle activation during the movement cycle.

The values obtained for each stride were then averaged for each minute of the trial. In this way we reduced the intra-subject step-to-step variability and got a better representation of the trend over the experiment duration. Finally, a mean value was obtained by averaging the variable of interest over all subjects. In order to compare the performance of subjects under the four different conditions experienced during the trial, we analyzed the last minute of each of the four conditions by comparing the average kinematics profiles, interaction torque and EMG envelopes.

Statistical significance was tested using repeated measures ANOVA on each dependent factor separately (as obtained by averaging on the last minute of each tested condition) and the tested condition as the *main factor*. Where appropriate, post-hoc comparisons of the ANOVA levels were tested using the Tukey-Kramer *Honestly Significant Difference* (HSD) method. All data processing was performed using Matlab (The MathWorks, Natick, MA, USA). Statistics tests were computed via SPSS (IBM SPSS, Somer, NY, USA) by setting the significance level at an alpha value of 0.05.

III. Results

A. Kinematic performance

The analysis of the kinematic performance was performed by comparing the mean position and velocity profiles for the hip, knee and ankle joint for the last minute of each tested condition, averaged among subjects. Resulting trajectories are presented in Fig. 3, which reports the averaged joint profiles in different colors for different conditions.

The most evident result was the alteration of the hip trajectories in both *zero-torque* and in *assisted condition*. Specifically, we recorded a reduction of the hip extension peak by 5.44°± 0.53° and 10.44°± 0.47° respectively (see Table I). Given that flexion peak was not significantly altered, the reduction of the extension peak resulted in a lower movement amplitude for the hip.

Knee profile presented visible alterations as well. In particular, a slightly more flexed posture (about 5°) between 30% and 50% of the stride period (i.e., stance phase) was present in the *assisted condition* only. A further modification happened in *zero-torque condition,* which presented a lower extension peak than *free-walking* and *assisted condition* (see knee max in Table I). On the contrary, ankle profiles do not seem to be altered by any tested condition.

Table I reports the quantitative and statistical evaluation of the kinematics performance during the experiment. The repeated measures ANOVA rejected the null-hypothesis for the extension peaks of both hip and knee joint (i.e., hip and knee max). For the hip, post-hoc analysis (Tukey HSD) highlighted a significant difference between *assisted condition* and *free walk pre*, *free walk post*, and between *zero-torque condition* and *free walk pre, free walk post*. For the knee a significant difference has been found only between *zero-torque condition* and *free walk pre, free walk post,* meaning that walking with active assistance from the exoskeleton did not affect this parameter.

By using the switch sensors located in the insoles we detected the heel strike and toe-off events for both feet. These parameters are critical to determine the gait cadence, and the durations of the single and double support phases. As shown in Table I, wearing the robot in transparent mode resulted in a small but statistically significant reduction ($p<10^{-4}$) of the gait cadence (99.2 steps/min), that was not present when the robot provide the assistance (101.4 steps/min). In addition, we found a significant increase of the duration of the left leg swing in the *zero torque mode* ($p<10^{-3}$), which was compensated by decreasing the duration of the first double support phase (DS1 in Table I), at the beginning of the gait cycle ($p<0.001$).

TABLE I

Mean and standard error for variables averaged over all subjects for the last minute of each tested condition along with ANOVA results.

	Independent variable (tested conditions)				
Dependent variable	*Free walk pre*	*Zero torque*	*Assisted*	*Free walk post*	ANOVA (*p*)
Kinematics					
Hip *(deg)*					
max	**14.00 ± 0.48**	**8.56 ± 0.58**	**3.56 ± 0.45**	**15.94 ± 0.65**	**2.5×10^{-5}**
min	-25.45 ± 0.62	-23.02 ± 0.58	-24.32 ± 0.72	-24.65 ± 0.75	0.8507
Knee *(deg)*					
max	**71.96 ± 0.67**	**66.26 ± 0.51**	**69.55 ± 0.47**	**72.73 ± 0.55**	**2.45×10^{-5}**
min	1.092 ± 0.70	-4.09 ± 0.66	-3.77 ± 0.82	-1.53 ± 0.87	0.1034
Ankle *(deg)*					
max	14.66 ± 0.41	15.48 ± 0.85	16.24 ± 0.73	14.58 ± 0.53	0.6392
min	-12.27 ± 0.33	-12.28 ± 0.53	-13.43 ± 0.51	-13.37 ± 0.36	0.4915
Cadence *(steps/min)*					
	102.7 ± 0.48	**99.2 ± 0.52**	**101.4 ± 0.46**	**102.6 ± 0.48**	**7.45×10^{-5}**
Gait phases *(stride %)*					
DS1	**12.46 ± 0.15**	**11.35 ± 0.09**	**11.89 ± 0.10**	**12.18 ± 0.11**	**0.0018**
right swing	37.33 ± 0.14	37.55 ± 0.12	37.70 ± 0.10	37.62 ± 0.11	0.6099
DS2	12.28 ± 0.17	12.09 ± 0.10	12.06 ± 0.11	12.36 ± 0.14	0.5351
left swing	**37.93 ± 0.17**	**39.00 ± 0.11**	**38.34 ± 0.11**	**37.85 ± 0.12**	**4.4×10^{-4}**
Muscle activation					
RMSA *(normalized)*					
GM	**1.00**	**0.913 ± 0.014**	**0.506 ± 0.025**	**0.956 ± 0.010**	**5.05×10^{-8}**
SOL	**1.00**	**0.948 ± 0.008**	**0.784 ± 0.016**	**0.935 ± 0.013**	**6.2×10^{-5}**
TA	1.00	1.021 ± 0.011	0.978 ± 0.038	1.075 ± 0.008	0.7005
RF	**1.00**	**0.845 ± 0.024**	**0.615 ± 0.021**	**0.838 ± 0.017**	**8.65×10^{-5}**
VL	1.00	0.958 ± 0.036	0.792 ± 0.032	0.896 ± 0.033	0.085
SE	**1.00**	**1.011 ± 0.026**	**0.863 ± 0.025**	**1.036 ± 0.025**	**9.73×10^{-4}**
LE Peak *(normalized)*					
GM	**1.00**	**0.971 ± 0.019**	**0.561 ± 0.022**	**0.973 ± 0.012**	**1.00×10^{-6}**
SOL	1.00	0.993 ± 0.010	0.971 ± 0.013	0.960 ± 0.006	0.6630
TA	1.00	0.994 ± 0.013	1.014 ± 0.034	1.037 ± 0.010	0.4841
RF	**1.00**	**0.873 ± 0.026**	**0.607 ± 0.025**	**0.894 ± 0.023**	**7.27×10^{-7}**
VL	1.00	0.864 ± 0.033	0.867 ± 0.036	0.897 ± 0.032	0.1834
SE	1.00	1.198 ± 0.031	1.119 ± 0.041	1.089 ± 0.028	0.3332

B. *Muscle effort reduction*

Starting from the recorded EMG signals, we computed the linear envelope - LE to evaluate the temporal pattern of the muscle activation (Fig. 4), the peak of the linear envelope to show the maximum activation, which is well correlated to the maximum force produced by the muscle (Table I), and finally, the RMSA as an indication of the total effort spent during the stride (Table I, Fig. 5).

Fig. 4 shows the EMG envelopes of all muscles averaged during the last minute of each condition (in different colors) and averaged over all subjects.

The most evident effect of the assistance in terms of muscle adaptation was a marked decrease in the activation of GM (ankle plantar-flexor) and RF (major hip flexor, minor knee extensor). The envelope profiles of these two muscles showed a clear reduction of both the peak of activation and the mean value of the curve over the whole stride. Comparing the *assisted condition* (black solid line) to *free walk pre* (red shaded line) and *free walk post* (green dot-shaded line) respectively, GM peak was reduced by 43.9% ± 2.2% and 41.2% ± 1.7%, RF peak by 39.3% ± 2.5% and 28.7% ± 2.4%, GM RMSA by 49.4% ± 2.5% and 45.0% ± 1.8% and RF RMSA by 38.5% ± 2.1 % and 22.3% ± 1.9%. The statistical significance of these results were confirmed by the

ANOVA (*p* values are reported Table I), which refused the null hypothesis for both GM RMSA ($p<10^{-7}$), GM peak ($p<10^{-5}$), RF RMSA ($p<10^{-4}$) and RF peak ($p<10^{-6}$). Post-hoc analysis (Tukey HSD) showed that for all these dependant variables, the only significant difference was between the *assisted condition* and all the other tested conditions. No significant difference was present between *free walk pre* and *free walk post* or between these two conditions and the *zero torque* one. This analysis indicated that during the assisted condition, GM and RF were less active than in free walking, and that a lower peak of activation is needed to walk the same speed.

A significant reduction of the activation was recorded for SOL (ankle plantarflexor, monoarticular) and SE (hip extensor, knee flexor) too. Nonetheless, the peak of the activation was not altered by the assistance (all *p's* were greater than 0.33). The linear envelopes of SOL showed a reduction of the activation during the assisted condition (Fig. 4, *black line*) between 0 and 30% of the normalized stride duration. The SOL RMSA during the last minute of the assisted condition was reduced by 21.6% ± 1.6% and 15.1% ± 1.4% compared to *free walk pre* and *post* respectively. ANOVA resulted in a rejection of the null-hypothesis ($p<10^{-4}$) while post-hoc analysis found a significant difference between the *assisted condition* and all the others.

Similarly, SE envelope in *assisted condition* showed a reduction between 10% and 25% of the normalized gait stride. ANOVA refused the null-hypothesis for the SE RMSA values ($p<10^{-3}$) but not for the peaks ($p=0.3332$). Post-hoc analysis on RMSA confirmed a significant difference between the *assisted condition* and all the others. Specifically, SE RMSA was reduced by 13.7% ± 2.5% and 17.3% ± 2.5%compare to *free-walk pre* and *post* respectively.

In addition, the peak of activation in *zero-torque* and *assisted condition* seemed to be delayed by about 5% of the stride duration compared to *free walking*.

By analyzing the envelope profiles of VL and TA reported in Fig. 4, small variations are visible between the different tested conditions. Nonetheless, ANOVA accepted the null hypothesis for both RMSA and envelope peak of these two muscles (all *p*'s values were greater than 0.085).

Fig. 4 and Table I compare the results for the last minute of each tested condition. This is useful to understand the effect of the assistance when subjects were more accustomed to it, but cannot give any information about the progress of the adaptation.

Fig. 5 shows the trend of the EMG RMSA for GM, SOL, RF and SE (the muscles that showed a statistical significant difference according to ANOVA) over all the experiment duration, albeit the first 50 steps after the assistance was turned on are not represented, because the assistance level was not fixed but gradually increasing.

As can be seen by comparing the *blue lines* with *red* and *green* ones, wearing the robot in transparent mode (*zero-torque condition*) had no significant effect on muscle activation. This result confirms that the interaction control effectively hides the loading effect of the exoskeleton so that no additional effort was required by subjects.

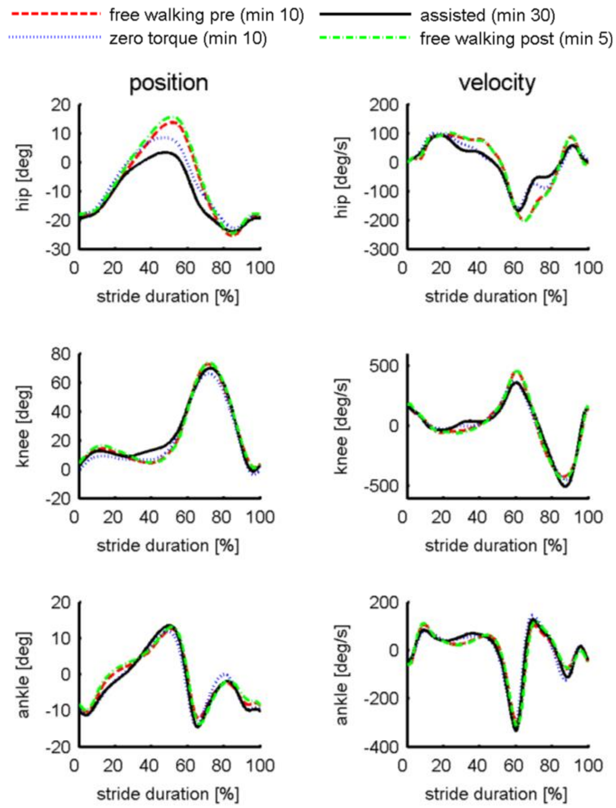

Fig. 3 Joint position and velocity trajectories for the last minute of each tested condition, averaged over all subjects

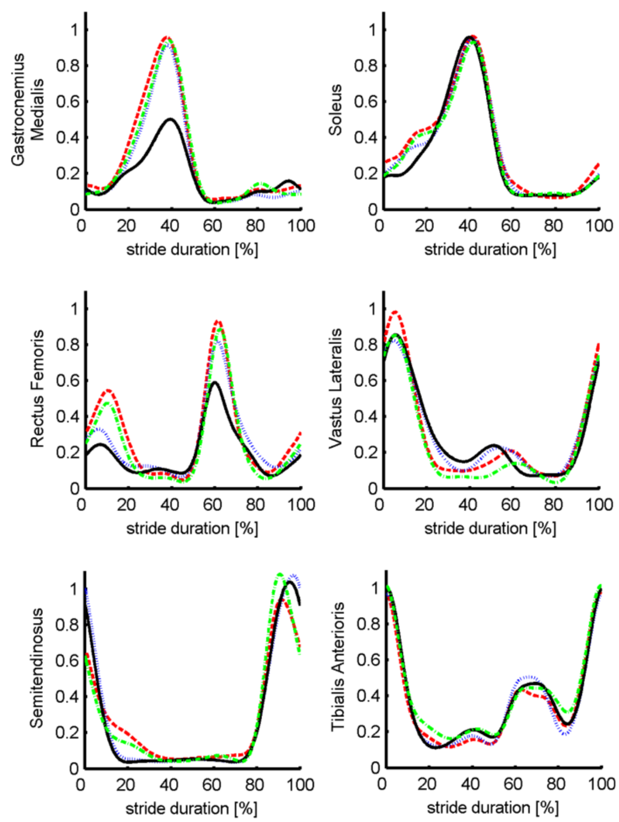

Fig. 4 EMG envelopes for the last minute of each tested condition, averaged over all subjects

By analyzing the trend of muscle RMSA during the assisted condition (*black line*), we can see that soon after the assistance was turned on, the muscle activation started decreasing progressively, without any initial increase of activation. Note however that starting from zero, the assistance was gradually increased until the set-point level was reached (i.e., from 0% to 50% of *Assistance %* in 50 steps), and that this period of gradual increase was not included in Fig. 5.

Despite not being statistically significant, as shown by the post-hoc analysis, RF activation seemed to drop between the *free walking pre* (red line) and the *zero torque* (blue line). Importantly, there was no difference between the *zero torque condition* and the *free walking post* (green line).

Whereas GM, SOL and SE seemed to reach a stationary level in about ten minutes, RF was still decreasing during the last minute of the assisted condition, indicating a longer adaptation time for the latter. Different from other muscles, GM decreased suddenly its activation when the assistance started. This can be seen by the abrupt discontinuity between the *blue* and *black* line around minute 20.

Finally, when subjects doffed the exoskeleton and returned to walk by completely relying on their legs, the muscle activation level increased and returned to the initial value. No statistical difference was recorded between *free walk pre* and *free walk post* condition. This is the final confirmation that the reduction of muscle activation, and then effort was only due to the robotic assistance.

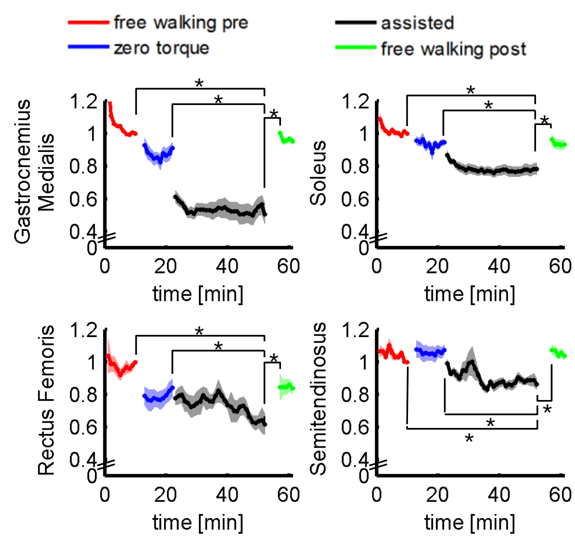

Fig. 5 EMG RMSA of GM, SOL, RF, SE averaged over all subjects for each minute of the experiment. Stars represent the pair wise comparisons reaching significance

IV. Discussion

A. Evidences from the experimental results

The assistance provided by the robot is a modified version of the nominal hip torque profile as extracted from Winter's dataset [43]. By referring to Fig 2, we can see that the hip extensors are initially active (0-20% of stride duration) to control the hip flexion, absorbing energy after the initial contact with the ground, and to stabilize the trunk (regulating its forward rotation). From 20% to 80% of the stride, the hip flexors become active. Initially (20-50% of stride), they control the backward rotation of the thigh, while after the hip flexion movement is arrested, they generate the pull-off of the lower limb (50-80% of stride). In this phase, the hip flexors contract concentrically to revert the movement of the hip and to accelerate the swinging limb upward and forward. Finally, from 80% to 100% of stride, the hip extensors are again active to decelerate the swinging limb, block the hip extension movement, and prepare to support the body weight after heel contact occurs.

Our controller assisted the user over the gait cycle by providing about 50% and 25% of the torque required to walk in flexion and extension respectively. Consequently, we expected a significant reduction in the activation of both hip flexor and extensor muscles. In our experimental set-up, we recorded a hip-extensor muscle (*Semitendinosus*) and a hip-flexor muscle (*Rectus Femoris*). As expected, both of these had a statistically significant reduction of activation in the assisted condition compared to free walking, see mean values and ANOVA results in Table I. This analysis confirms that we were able to assist both hip flexion and extension phase during walking.

Looking at the SE envelope profiles (Fig. 4), we can see that a major reduction of activation happened after heel strike, suggesting that subjects were mainly helped by the robot in controlling the forward rotation of the thigh and reduce the forward acceleration of the body. On the contrary, the peak of activation at the end of the swing does not seem to be very affected, meaning that the assistive controller was less efficient in helping the user to decelerate the swinging lower limb. RF activation was visibly assisted as well. A statistically significant decrement of both RMSA and envelope peak was observed during the assisted condition compared to free walking. This shows that subjects effectively reduce the effort needed to pull-off the swinging limb due to the robotic assistance.

In agreement with what was observed in other studies [21],[22], when assisted, the hip joint trajectory was significantly altered compared to free walking. Specifically, we recorded a clear reduction of the hip extension angle, which supports the previously proposed hypothesis that joint kinematics is less important for motor planning compared to joint kinetics [51]. It is worth noting that a 5.4 degree reduction in hip extension was observed comparing *zero-torque* and *free-walking* condition, without recording any significant effect on muscle activations. Further experiments would be needed to explore this phenomenon.

Besides the reduced effort of hip flexor and extensor muscles, we observed a remarkable decrease in the activation of the ankle plantar-flexors: Both GM (biarticular) and SOL (monoarticular) significantly reduced their EMG RMSA in the assisted condition (-49.4% ± 2.5% and -21.6% ± 1.6% respectively compared to *free walking*). Nonetheless, the analysis of the envelope profiles in the assisted condition (Fig. 4) shows a very different behavior of GM and SOL, which can be explained by considering their different functions in walking in terms e.g. of energy redistribution on the leg segments [52]. The GM had a higher decrease of activation compared to SOL. Moreover, the GM was lower for all the stance period (0%-60% of stride), while SOL was below the *free-walk* value only between 0% and 30% of the gait cycle. Finally, SOL peak of activation was not significantly different in all the tested conditions (Table I). As observed in Fig. 3, the reduction of GM activation resulted in an increased knee-flexion angle in mid and late stance. Again, this can be explained considering that the lower activation of GM in the assisted condition could have reduced the control of the forward rotation of the leg during the stance period [53],[26], affecting as a consequence the knee angle. Ankle plantar-flexion is responsible for the so-called *push-off*, an impulsive push from the trailing limb that happens in the late stance with the goal of redirecting the body center of mass forward and upward [25],[26].

Despite *push-off* being particularly efficient for propelling walking [27], a tradeoff exists between ankle and hip function during gait. In fact, the hip can propel walking as well, by concentrically contracting the flexor muscles in swing [28], or by activating the extensor muscles of the stance leg [29]. The physiological equilibrium between these three strategies depends on age (the elderly use more hip extension and less ankle with respect to the young [54][55]), health condition (persons with diabetes mellitus exaggerate hip flexion [28],[31]), the presence of a passive orthosis [33], and can also be altered voluntarily [32]. In this study, we found that the action of an external assistance, such as the external torque provided by a powered exoskeleton, can modify the physiological equilibrium by altering the normal efficiency of the hip and the ankle walking strategies. As a result, a different activation pattern can emerge also for the muscles that do not directly power the joints assisted by the robot. In our case, assisting the user's hip produced a reduced ankle strategy, thus lowering the activation of the shank muscles, as well as a decrease of the hip strategies (i.e. reduced activation of hip flexor and extensor muscles). These findings are not only important to gain insights into human motor control of walking, but also to guide the design of future lower-limb exoskeletons, which should take into account the human-joint synergies in walking and exploit these in a convenient way.

By focusing on adaptation trends (Fig. 5), further observations can be discussed. Different from most of other studies about assistive exoskeletons [22][42][49], none of the recorded muscles increased their activation levels when the assistance was activated (blue to black transition in Fig. 5).

A possible reason for this result is that in our protocol, we gradually increased the assistance level (i.e., 1% of increment for each step, up to *Assistance %* equals 50) instead of having an abrupt activation of the assistance as in other experimental trials [22][42][49]. The neuroscientific framework supporting this hypothesis can be found in [56]-[58]. These studies analyzed human repetitive movements such as walking and found a mathematical model that can predict the motor adaptation to perturbations. Basically, the progress of adaptation is the result of two concurrent processes that are governed by the trajectory error between expected and actual movement. Considering a single-joint example, if the error exceeds a well-defined threshold, an asymmetric and simultaneous contraction of antagonist muscle pairs happens (note that the activation of the overstretched muscle is greater than the under-stretched one [59]). If the error is under the threshold, both antagonist muscles decrease their activation level. This phenomenon, previously called *slacking* [58], has to be maximized in the case of wearable robot for movement assistance targeting muscle effort reduction.

Our rationale is that if the assistance, i.e., the disturbance to which the user needs to adapt, is increased gradually starting from zero, the trajectory error compared to the previous step remains under the threshold and therefore muscles decrease their activation. Conversely, if the assisting torque is too high, the error would overcome the threshold and then produce muscle co-activation, which in turn increases effort. Our results confirm that a gentle increase of assistance can provide faster adaptation time in terms of muscle activation reduction. In fact, in our study, we found an adaptation time of about 10-15 minutes that is significantly lower than the value observed in other past works [21][22].

Differently from all other muscles, GM had an abrupt decrease of activation when the assistance was activated (see the first minute of assisted condition compared to last minute of zero torque condition in Fig. 5). This sudden drop-off was not likely due to a real process of motor adaptation; rather it may be explained considering the biarticular nature of GM and the assisted-gait kinematics. As can be seen in Fig. 3, the hip assistance had the effect of increasing the knee flexion angle in stance. This could have prevented the GM to correctly lengthen and then contract normally. Again, further investigation would be needed to explore the differences in the adaptation timing of each muscle of the lower limb.

B. Limitations of the study

As in other similar studies [21]-[23], we assisted the gait unilaterally, i.e. robotic assistance was provided to the left leg only. Given the importance of inter-limb coordination mechanisms and reflexes [60]-[62], the unilateral assistance could have increased the difficulty of the task in terms of adaptation time. Although we did not record the kinematics of the subjects' right leg, we assessed the gait symmetry through the foot pressure sensors located in the insoles. Post-hoc analysis on the duration of the gait phases did not show any statistically significant difference between *free walking condition* and *assisted condition*. In addition, we did not record the EMG signals of the users' right-leg muscles, and the ground reaction forces. Thus we could not assess the occurrence of compensation strategies at the level of the right leg [63] during the *assisted* or *zero-torque* condition.

Further insights could be gained in the future by recording kinematics, kinetics, and EMG signals from both legs and by assisting users bilaterally. Finally, heart rate and metabolic consumption measurements would be desirable to verify possible systemic benefits to the cardiopulmonary apparatus.

C. Application fields

The proposed robotic assistance may restore the normal movement ability of persons having muscle weakness caused e.g., by myopathies [14], which preclude production of enough hip force to walk normally. In fact, the robot could provide the amount of assistance that is needed in terms of force, adapting autonomously to the walking cadence.

Reducing hip-muscle forces may also help many persons who suffer from hip pain. For example, osteoarthritis may be mitigated by a reduced load on the articulation [64][65] as a result of the use of the assistive exoskeleton. As already shown by Lewis [32][22], the combination of reduced hip angle extension and decreased hip muscle force may be particularly useful for people with anterior acetabolar labral tear [66]. A further effect of the robotic assistance is the reduced ankle plantar-flexion. This specific gait alteration may be helpful for people having diabetes mellitus or peripheral neuropathies, as already postulated in [22][67]. A decrease of the ankle plantar-flexion could indeed reduce the peak pressure on the forefoot during toe-off, which seems to be a major cause of neuropathic foot ulcers [68]. The use of the assistive exoskeleton could therefore reduce the incidence of this pathology.

A final outcome of the study is the use of a simple model-free assistive method, based on Winter's dataset. This method does not need additional sensor and calibration procedures to compute the assistive torques. This method is not based on predefined walking pattern so leaves users free to use their preferred one. Experimental results confirm that it is a feasible approach for reducing the users' effort while walking.

V. Conclusion

This paper explored the human adaptation to the action of a lower-limb exoskeleton providing hip assistance during walking. Our specific goal was to investigate the interplay among the user's hip and ankle joint during the robot-assisted walking. To this end, we proposed a novel assistive controller that estimates on-line the walking cadence of users and provides them with an assistive torque at the hip through the exoskeleton. Importantly, the assistive torque is based on the *nominal joint torque profiles*, extracted by Winter [40]. Therefore, it is computationally inexpensive and does not require any calibration prior to use. Despite its simplicity, we verified that users can adapt and benefit from this kind of assistance by significantly reducing their muscle activation both at the hip and the ankle level after less than thirty minutes of exercise with the robot.

In the trials, we used the ALEX II exoskeleton [47] to provide assistance at the hip joint only. Nonetheless, experimental data showed a clear reduction of both hip flexors and extensors and ankle plantar-flexors. This result further confirmed the dynamical interaction between several muscles spanning the different joints of the lower-limb. This complex interaction ought to be considered in the design of future assistive exoskeletons in order to be effective.

Experimental results suggest that the use of this kind of walking assistance may be of help for several groups that would benefit from a reduced hip torque (e.g., osteoarthritis, muscle weakness) or ankle plantar-flexion (e.g., person suffering from neuropathies).

Future works will be devoted to test the assistance with actual end-users and to extend the proposed assistive method for bilateral assistance.

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