34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Reducing Muscle Effort in Walking through Powered Exoskeletons

T. Lenzi, *Student Member, IEEE*, D. Zanotto, P. Stegall, *Student Member, IEEE*
M.C. Carrozza, *Member, IEEE*, and S.K. Agrawal, *Member, IEEE*

Abstract— **This paper presents a novel assistive control for lower limb exoskeletons. The controller provides the user with a scaled version of the Winter's *nominal torque profile*, which is adapted online to the specific gait features of the user. The proposed assistive controller is implemented on the ALEX II exoskeleton and tested on two healthy subjects. Experimental results show that when assisted by the exoskeleton users can reduce the muscle effort compared to free walking.**

I. Introduction

In this paper we explore the use of powered exoskeletons for assisting human walking with the specific goal of reducing the muscle effort.

Several pathologies can decrease the walking ability of affected persons by reducing their muscle strength and endurance, or lowering the maximum sustainable cardiac effort (e.g., heart related diseases). 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. 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 condition.

The production of muscle force is the main contributor to the metabolic cost of walking. In addition, the muscle force is an important source of load on the articulations, which, in the case of e.g. osteoarthritis, produces pain [1]. Assistive exoskeletons may represent a possible way of restoring the normal walking effectiveness in these critical situations. 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 that change along the gait cycle. At the same time, the user adapts his muscle activation patterns in order to exploit such assistive torques in a convenient way (i.e. motor adaptation). If the assistance is successful, the motor adaptation process will result in lower muscle forces and thereby a more efficient walking for the user. A thorough understanding of the human adaptation process is therefore needed to design an effective assistive device for walking [2]. Unfortunately, the adaptation process depends on the specific action of the robot (i.e., the actual assistive torque profile applied on each user's joint), while the effectiveness of the assistive torque (in terms of muscle force reduction) depends on the progression of the adaptation process. These two factors are indelibly interconnected and cannot be studied separately. As a consequence, the effect of any assistive device cannot be evaluated *a priori*, but need to be tested specifically on human subjects. Clearly, the design of an assistive device should take into account fundamental knowledge about biomechanics and neurophysiology [2]. Recent studies exploit proportional EMG control strategies to provide walking assistance. These studies 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 [3],[4]. As a consequence, the muscle torque is decreased and the metabolic effort reduced [5]. On the other hand, the joint position trajectory in the assisted condition seems to be modified by the assistance [3][4]. Similar studies 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 EMG-based control, other approaches have been used, focusing on inertia reduction [6], gait segmentation [7] or position-based compliant force fields [8]. So far these methods did not prove to reduce the user effort compared to free walking.

From a research perspective, the main drawback of existing assistive devices is that they cannot provide the user with any desired assistive torque profile during walking, but are limited to a specific assistive action. This limitation is not only due to the mechanical design of the robot (e.g. actuators selection and placement) but also to the specific control strategy implemented on it (e.g., proportional EMG control). As a result, their potential use for understanding human behavior is restricted. In this paper we present an assistive controller that can overcome this limitation by estimating the wearer's walking cadence on-line (using adaptive frequency oscillators [9]), computing the current percent of stride, and finally providing the user with any desired torque profile along the gait cycle. The proposed assistive controller has been implemented on the ALEX II gait trainer [10] and experimentally tested on two healthy subjects. For the purpose of the experiment, the powered exoskeleton assists the user by providing a scaled version of the nominal torque profile as extracted from Winter's dataset [11] to the hip alone. Results of the experiment along with discussion are reported.

II. Methods

A. The ALEX II gait trainer

ALEX II is a treadmill based lower-limb exoskeleton (Fig. 1) developed at the University of Delaware [10]. The unilateral robotic leg of ALEX II has two active degrees of freedom (driven by geared DC motors, Danaher Corporation, Washington D.C., USA) to power the hip and knee joints of the user on the Sagittal plane. Hip adduction/abduction is allowed through passive degree of freedom. The robotic leg is supported from the rear by a back support, which also attaches to the user. The back support provides configuration-

This work was supported in part by the CYBERLEGs project (FP7-ICT-2011-2.1 #287894) and EVRYON project (FP7-ICT-2007.8.5 #231451).

T. Lenzi and M.C. Carrozza are with the BioRobotics Institute, Scuola Superiore sant'Anna, 56025 Pisa, Italy (corresponding author to provide phone: +39 050 883473; fax: +39 050883497; e-mail: lenzi@ ieee.org).

D. Zanotto, P. Stegall and S. K. Agrawal is with the Mechanical Engineering Department, University of Delaware, Newark, DE 19716, USA.

independent gravity compensation for the device [12]. 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).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.

B. The Assistive Controller

The assistive controller is based on a two-step process: (1) estimate online the user's joint torque and (2) provide the user's joint with a constant fraction of the torque estimate using the powered exoskeleton. This approach allows having assistive torque profiles that are always coherent (in direction and amplitude) with the mechanical action resulting from the muscle activation.

Electromyography can be exploited to estimate the user's joint torque with good results. Nevertheless, it requires an accurate placement of electrodes on the skin, and a complex and time-consuming inter- and intra- subject calibration. An alternative strategy for user's joint torque estimation consists in solving the inverse-dynamic problem. This method requires a good estimate of joint positions and their derivatives, as well as an accurate dynamic model of the user. In addition, any physical interaction with the external environment (including the exoskeleton) should be measured and incorporated in the model. Both these methods are very accurate but also very complex and often impracticable in most real-world scenarios [13]. An alternative solution could be the use of *nominal torque profiles* that can fit the actual user torque during walking inside a certain range of uncertainty. While walking at a constant cadence, the human joint torque follows a periodic temporal pattern that presents good intra- and inter-subject repeatability if normalized by the subject body weight and expressed as function of the relative time of the gait-cycle duration (i.e., time between two consecutive heel strikes of the same foot) [11].

Starting from these considerations, we designed an assistive controller that exploits the *nominal torque profiles*, as defined and computed by Winter [11], to estimate the user's joint torque as a function of the current phase of the gait cycle, the walking cadence and the user's body weight.

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

The first stage of the assistive controller addresses the estimate 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 Oscillator (AFO), a mathematical tool that has been originally developed for other applications [8] and more recently used for estimating the high-level features of periodic human movements for rehabilitation and assistive purposes [8][13].

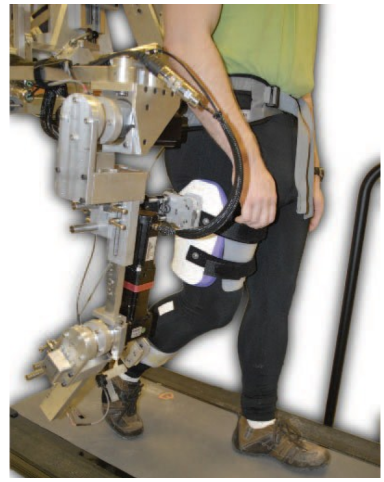

Fig. 1 The ALEX II gait trainer

A thorough presentation of the AFO is out of the scope of this paper. A mathematical description and a detailed experimental validation of the current implementation of the AFO can be found in [8]. Resistive foot-pressure sensors are embedded in the user's shoe insoles (see [10] for implementation) and act as switch 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.

The estimate of the user's joint torque is based on the value reported on the Winter tables [11] 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). These profiles are implemented on a bi-dimensional look-up table (*2D-LUT*) that takes as input 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 2-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 (*Support %*). The obtained value (*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. It is worth noting that by using the same control structure we could have feed any desired torque profile that is defined as a vector of 100 values of the percent of stride.

C. Experimental Protocol

Two 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. Surface EMG activity from Rectus Femoris and Gastrocnemius Medialis of the assisted leg were measured by MA-420 EMG preamplifiers. EMG recordings were 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 hip, knee and ankle flexion-extension by using mechanical resistive potentiometers (PASCO, Roseville, CA, USA). Resistive foot pressure sensors equipped both 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 in walking on a treadmill at a constant velocity of 2.4 mph (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 of 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. 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 trial took place.

Assisted condition: After ten minutes from the beginning of the *zero-torque* condition, the controller automatically started providing the assistive torque. For safety reason, subjects were verbally warned thirty seconds before the onset of the assistance by the experimenter. The desired level of assistance for the trial was controlled by setting the *Support %* command (see Fig. 2) to 50%, which corresponds to providing 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 bi-dimensional LUT. The *assisted condition* lasted 30 minutes. After this period, the *Support %* was set again to zero by the controller, the treadmill was stopped by the experimenter, and the user doffed the exoskeleton.

Free-walking post: After resting, 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.

III. RESULTS

In order to appropriately compare the four different tested conditions, we considered only the last minute of each sequence in the data analysis. We then averaged the variable of interest taking into account the complete strides inside the last minute. Averaged hip, knee and ankle joint trajectories are shown in Fig. 3 for both subjects. This figure shows a significant alteration of the hip trajectories in the *assisted condition*. Specifically, we recorded a shift of the angle trajectory in flexion and a reduction of the movement amplitude. The ankle seems to be minimally affected by the different tested conditions, while the knee profile presents an alteration between 30% and 50% of the stride period, which indicates a more flexed posture during stance phase. Starting from raw EMG signals, linear envelopes have been 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). Fig. 4 shows the averaged EMG envelopes of Gastrocnemius Medialis (GM) and Rectus Femoris (RF) for the last minute of each condition. RF activation is greatly reduced in the *assisted condition* (black line, Fig. 4), between 50% and 80% of the gait cycle, which corresponds to late stance and early swing phases. The RF peak in *assisted condition* is reduced with respect to *free walking condition* by 35.1% and 36.5% for subject 1 and subject 2 respectively. GM activation is

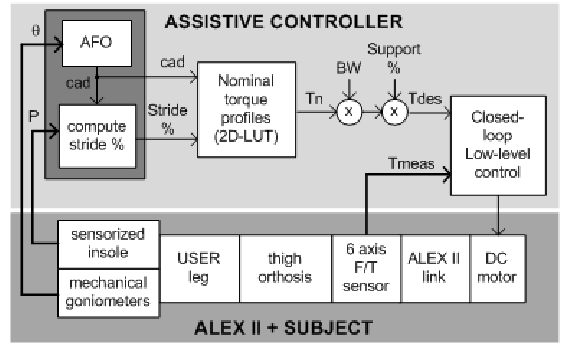

Fig. 2 Block diagram of the assistive controller

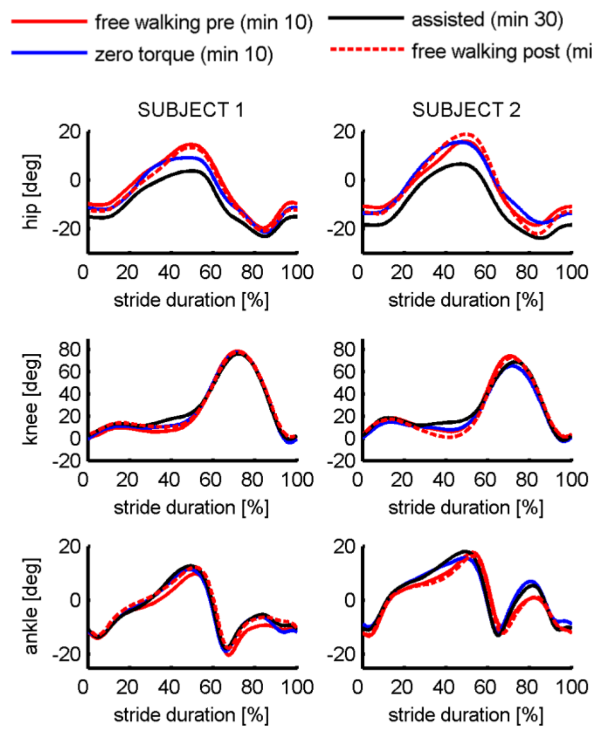

Fig. 3 Averaged joint trajectories for each tested condition

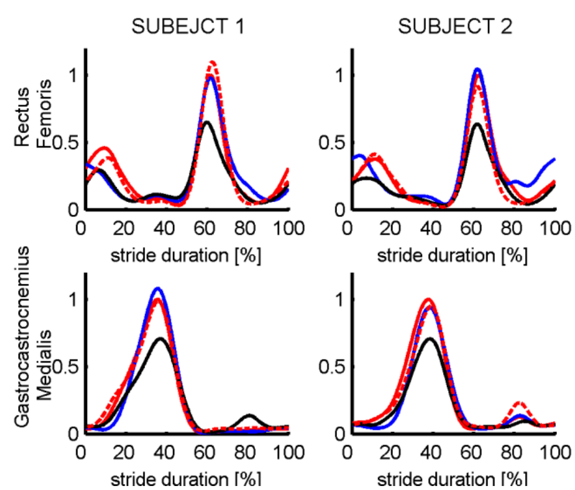

Fig. 4 Averaged muscle envelopes for each tested condition

reduced as well, between 20% and 60% of stride period (i.e., stance phase). In the *assisted condition*, the peak of the GM envelope is reduced with respect to the *free walking* condition by 29.4% for subject 1 and 29.1% for subject 2.

IV. Discussion

The alteration of the joint trajectories resulting from the assistance is not surprising and is coherent with results found in [3][4]. Conversely, we found a completely novel result in terms of muscle activation. Our protocol provides assistance at the hip joint only, nonetheless we found a clear EMG reduction on both the RF (i.e. hip flexor) and the GM (i.e. ankle plantar-flexor). To the best of our knowledge this is the first time that such a relationship has been found. We believe that this result could be attributed to the strategy used by humans at the joint and muscle levels to support the body weight and propel the body mass during walking. Recent studies demonstrate the existence of at least three concomitant strategies that can be used by humans when walking [15][16]: (1) *ankle strategy* i.e. push off of the ankle joint prior to the swing start of the ipsilateral leg; (2) *hip flexor strategy* i.e. pulling the ipsilateral limb into swing; (3) *hip extensor strategy* i.e. contracting the hip extensors to posteriorly rotate the pelvis and help the contralateral limb progression. A tradeoff between these strategies seems to be used by the CNS to produce stable and effective walking [16]. A pathological condition (e.g. diabetes, arthritis) can alter the physiological equilibrium towards one of these concurrent strategies [17][18]. Importantly, this balance can also be altered voluntarily, for example by instructing a healthy subject to exaggerate the ankle push-off [19]. The hip assistance could have altered the normal equilibrium of these strategies by reducing the need for ankle push off. This could explain the lower level of activation of the GM, which is responsible for the ankle plantar flexion that in turn produces the so-called push-off. These findings have a strong relevance for the design of assistive exoskeletons. From our results, we can hypothesize that an external assistance (such as the external torque provided by a powered exoskeleton) could alter the physiological equilibrium by making one of the walking strategies more convenient than the others. If this happened, a different activation pattern should emerge also for the muscles that do not directly power the joint that is assisted by the robot. In our case, assisting the hip flexion resulted in less *ankle strategy,* thus lowering the activation of the shank muscles. Another important outcome of the experiment is about the method we used to generate the assistive torque profile. Our hypothesis was that the lower accuracy of the torque estimate based on Winter's *nominal profile* would have not compromised the effectiveness of the exoskeleton in reducing the user effort. Experimental results seem to confirm this hypothesis. This represents a clear simplification, and then an advantage for the design of assistive exoskeletons.

V. Conclusion

Our assistive method does not require any additional sensor for the robot or the user, it is computationally efficient, and does not need any calibration. For these reasons it could be used outside laboratories, in real-world scenarios, where powered exoskeletons are expected to provide the most benefit for users. As a result of the motor adaptation to the hip assistance, a marked reduction of the hip flexor (i.e. Rectus Femoris) and the ankle plantar-flexor (i.e. Gastrocnemious Medialis) emerge. This result not only proves the effectiveness of the proposed controller in reducing the walking effort but also suggests that the CNS could adapt in order to redistribute the hip assistance on both the hip and ankle joint. Future works will aim to test the controller on a larger number of subjects as well as to verify the effect of the assistance on all the main muscles of the lower limb.

References

[1] Lewis CL, Sahrmann SA, Moran DW. Anterior hip joint force increases with hip extension, decreased gluteal force, or decreased iliopsoas force. Journal of Biomechanics. 2007; 40(16):3725-3731.

[2] Ferris DP, Sawicki GS, Daley MA. A Physiologist's Perspective on Robotic Exoskeletons for Human Locomotion. IJHR. 2007;4:507–28.

[3] Kao PC, Lewis CL, Ferris DP. Invariant ankle moment patterns when walking with and without a robotic ankle exoskeleton. Journal of Biomechanics, 2010;43(2):203-9.

[4] Lewis CL, Ferris DP. Invariant hip moment pattern while walking with a robotic hip exoskeleton. J. of Biomech., 2011; 44(5):789-93.

[5] Sawicki GS and Ferris DP (2009) Powered ankle exoskeletons reveal the metabolic cost of plantar flexor mechanical work during walking with longer steps at constant step frequency. J of Exp. Biol., 212:21-31.

[6] Aguirre-Ollinger, G.; Colgate, J.E.; Peshkin, M.A.; Goswami, A.; , "Inertia Compensation Control of a One-Degree-of-Freedom Exoskeleton for Lower-Limb Assistance: Initial Experiments," IEEE TNSRE, 20(1) ,68-77, 2012.

[7] Kawamoto H., Sankai Y.: Power assist method based on Phase Sequence and muscle force condition for HAL. Advanced Robotics 19(7): 717-734 (2005).

[8] Ronsse R., Lenzi T., Vitiello N., Koopman B., van Asseldonk E., De Rossi S.M.M, van den Kieboom J., van der Kooij H., Carrozza M.C. and Ijspeert A.J., Oscillator-based assistance of cyclical movements: model-based and model-free approaches, MBEC, 49(10):1173-1185.

[9] Righetti L, Buchli J., and Ijspeert A. J., Dynamic Hebbian learning in adaptive frequency oscillators, Physica D, 216, 269–281, 2006.

[10] Winfree, K.N.; Stegall, P.; Agrawal, S.K.; , "Design of a minimally constraining, passively supported gait training exoskeleton: ALEX II," IEEE ICORR 2011.

[11] Winter DA, Yack HJ. EMG profiles during normal human walking: stride-to-stride and inter-subject variability Electroencephalogr Clin Neurophysiol. 1987 Nov;67(5):402-11.

[12] Agrawal A. and Agrawal S.K. Design of gravity balancing leg orthosis using non-zero free length springs. Mechanism and Machine Theory, 40(6):693–709, 2005.

[13] Pandy M.G., Andriacchi T.P., Muscle and Joint Function in Human Locomotion, Annual Review of Biomedical Engineering 12: 401-433.

[14] Ronsse R, Vitiello N, Lenzi T, van den Kieboom J, Carrozza MC, Ijspeert AJ., Human-robot synchrony: flexible assistance using adaptive oscillators, IEEE TBME, 2011 58(4):1001-12.

[15] Kerrigan DC, Todd MK, Della CU, Lipsitz LA, Collins JJ. Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. Arch.Phys.Med.Rehabil. 1998;79:317–322.

[16] Mueller MJ, Sinacore DR, Hoogstrate S, Daly L. Hip and ankle walking strategies: effect on peak plantar pressures and implications for neuropathic ulceration. Arch.Phys.Med.Rehabil. 1994b;75:1196–1200.

[17] DeVita P, Hortobagyi T. Age causes a redistribution of joint torques and powers during gait. J.Appl.Physiol. 2000;88:1804–1811.

[18] Judge JO, Davis RB, III, Ounpuu S. Step length reductions in advanced age: the role of ankle and hip kinetics. J.Gerontol.A Biol.Sci.Med.Sci. 1996;51:M303–M312.

[19] Lewis CL, Ferris DP. Walking with increased push-off decreases hip moment. Journal of Biomechanics. 2008;41:2082-2089.