33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

Proportional EMG control for upper-limb powered exoskeletons

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***Abstract*—Electromyography (EMG) has been frequently proposed as the driving signal for controlling powered exoskeletons. Lot of effort has been spent to design accurate algorithms for muscular torque estimation, while very few studies attempted to understand to what extent an accurate torque estimate is indeed necessary to provide effective movement assistance through powered exoskeletons. In this study, we focus on the latter aspect by using a simple and "low-accuracy" torque estimate, an EMG-proportional control, to provide assistance through an elbow exoskeleton. Preliminary results show that subjects adapt almost instantaneously to the assistance provided by the exoskeleton and can reduce their effort while keeping full control of the movement.**

I. Introduction

Powered exoskeletons are wearable robots designed to assist humans in performing movements. The assistance provided by these devices can be exploited to several ends: either to augment the performance of healthy humans, enhancing their endurance [1] or strength [2], or to restore normal abilities in patients affected by movement disorders, such as tremor [3], hemiplegia [4] or paraplegia [5], or finally for neuro-muscular rehabilitation [6].

The most common way of realizing this kind of assistance consists in estimating the joint torques needed to perform the intended movement, and then in providing the wearer with a constant fraction of said torque through the robot, depending on the needs of the user and on the specific goal of the treatment [7][8]. As a result of the assistance, the wearer is supposed to adapt his/her motor behavior, in terms of muscle activations, to exploit the extra torque and to reduce the metabolic effort required to perform the movement [9].

Despite encouraging results, this method presents two main drawbacks. First, an accurate estimation of the torque needed by the user to perform the movement is very difficult to obtain in real-time. In addition, the user's reaction to the extra-torque provided by the robot cannot be modeled, but should be taken into account in order to design an effective assistive control. A possible strategy for estimating the torque needed to perform the intended movement consists in solving the inverse dynamic problem [10]. This method requires a good estimate of joint positions, velocities and accelerations, as well as an accurate dynamic model of the user's limb to be assisted. Moreover, any physical interaction with the external environment (including the exoskeleton) should be measured by means of force sensors and included in the model. These requirements are hard to satisfy, and make this approach impractical in most real-world scenarios.

Another torque estimation method consists in exploiting the electro-myographic (EMG) signals that are generated by the neuro-musculo-skeletal system to produce the intended movement. EMG signals, resulting from the motor neuron impulses that activate the muscle fibres, can be correlated with the force produced by muscles and the torque exerted at the joint level [11]. EMG-based torque estimation presents some intrinsic advantages. First of all, the incipit of EMG signal starts about 20-80ms before the muscle contraction takes place [12]. This delay can be exploited for real-time computation of assistive torques, as well as to compensate the limited bandwidth of the robot actuation system. In addition, EMG-based control does not require to take into account if the user is interacting with the external environment or is moving freely. EMG-based methods do not need a dynamic model of the user's limbs and, most importantly, allow to estimate the torque before the movement takes place, so that the user can be assisted even if not able to initiate the movement autonomously.

Different approaches have been proposed in the past to estimate the muscular torques starting from EMG activation, ranging from black-box neural networks [13], neuro-fuzzy classifiers [14], and Hill models [15][16]. In these studies, a lot of effort was spent in designing algorithms aiming at the best possible torque estimate accuracy. Particular attention was given to develop estimation methods that reduce the need for complex subject- and session-dependent calibration procedures, lowering, at the same time, the required computational power. Despite the huge effort spent to achieve high estimation accuracy, very little attention has been paid to understand if an accurate torque estimate is indeed necessary to provide effective movement assistance through powered exoskeletons.

In this paper we focus on the latter aspect by exploring to what extent the EMG-torque estimation algorithm can be simplified, while still providing effective movement assistance. Specifically, we are not interested on raw open-loop prediction of EMG but rather on closed-loop usability of the system. We postulate that the adaptation capacity of the user (i.e. the motor learning ability) can compensate for torque estimate imprecision, without adding further cognitive effort to that the user normally spends in

This work was partly supported by the EU within the EVRYON Collaborative Project STREP (Evolving Morphologies for Human-Robot Symbiotic Interaction, Project FP7-ICT-2007-3-231451).

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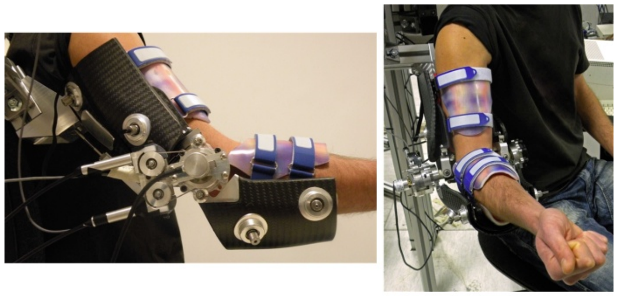

(a) (b)

Fig. 1 Overview of NEUROExos: (a) lateral view, (b) front view.

controlling his own arm movement. With this aim, we test the user reaction to a simple and "low-accuracy" assistive torque provided through an elbow exoskeleton. This simplified torque estimate is obtained by applying a proportional gain to the envelope of EMGs recorded from the muscles involved in the movement. As already shown by Hogan in [17], this approximation holds only for isometric contractions, and therefore introduces a systematic error in the estimate when applied in dynamic conditions. Surprisingly enough, we observed that subjects are able to adapt almost instantaneously to the assistance provided by the exoskeleton and can reduce their effort while keeping full control of the movement.

This paper presents the description of the EMG-based proportional control system along with its implementation on the NEUROExos platform, a powered exoskeleton for the elbow assistance. The proposed assistive control was tested on two subjects performing an elbow flexion/extension movement against gravity (i.e. in non-isometric, non-isotonic muscle contraction conditions). Results of the experiment along with discussions are reported.

II. Methods

A. The NEUROExos platform

The NEUROExos [18][19], shown in Fig. 1, is an elbow powered exoskeleton. User's upper- and lower-arm are rigidly connected to the robotic links, which have a double shell structure: an outer carbon-fiber shell allows to transmit the assistive force, and an inner flexible shell improving the comfort of the user [20]. NEUROExos is provided with an adaptive, passive-compliant actuator, implemented by means of an antagonistic non-linear elastic actuation system located remotely from the user [21]. Torque is transmitted to the NEUROExos joint by means of steel wire ropes routed trough Bowden cables. The driving block is composed by a driving pulley (radius of 19 mm), which the antagonistic steel ropes wrap around and a planetary gear amplifying the input torque of a factor four. The platform is equipped with two cable force sensors, and with a 4096 ppr rotary encoder.

B. EMG processing and proportional controller

Surface EMG activity from the biceps brachii and triceps brachii muscle were picked up by pre-gelled Ag/AgCl 8 mm diameter bipolar surface electrodes (Pirronse&Co., Italy) attached about 2 cm apart along the longitudinal axis of the muscle belly. EMG recordings were digitized at 1 kHz using the Telemyo 2400R G2 analog output receiver (Noraxon USA Inc., AZ, USA) with an internal band-pass filter (10-500 Hz) and a gain coefficient of 2000. Raw EMG signals were processed to obtain the linear envelope (LE) profiles which resemble the muscle tension waveforms during dynamic changes of isometric forces [22]. LEs were obtained on-line through full-wave rectification of band-passed EMG signals and post-filtering by means of a second-order low-pass Butterworth filter with a cut-off frequency of 3 Hz [23]. As showed in the NEUROExos control scheme (Fig. 2), LEs gathered from biceps and triceps muscle (i.e. LE_{bic} and LE_{tric}) were multiplied by two different constant factors K_{bic} and K_{tric}, to obtain the force set-points for the NEUROExos flexor and extensor cable respectively ($\hat{F}_{flx}$ and $\hat{F}_{ext}$). Cable forces (F_{flx} and F_{ext}) were regulated by the NEUROExos closed-loop low-level controller to produce the final assistive torque on the user joint. EMG recordings and sensor outputs were synchronized and saved by means of a Labview® routine running at 1 kHz on a real-time controller NI PXI-8196 (National Instrument, TX, USA).

C. Experimental procedure

After the set-up of EMG recording apparatus, subjects sat on a chair and wore the NEUROExos on their right arm. The weight of NEUROExos was supported by an external frame, which also constrained the upper arm to an inclination of about 30 deg to the gravity vector, as shown in Fig. 1. The experimental procedure was divided in two phases. In the first part, the subject chose the two gains of the proportional controllers (K_{bic} and K_{tric}), one after the other, starting from the biceps. Both gain values were initially set to 0. The subject was instructed to increase the gain gradually using a knob while moving his/her elbow freely. The experimenter exhorted subjects to increase the gain as long as they felt comfortable with the level of assistance. After K_{bic} was properly set, the same procedure was repeated for the K_{tric}. No time constraint was given to subjects for this procedure, but no more than 2 minutes were needed.

Subjects took rest for 10 minutes outside the exoskeleton before the second part of the experiment started. In this phase, participants were asked to make cyclical flexion/extension movement with a target amplitude of 50 deg, and pace of 1Hz. Augmented visual feedback was

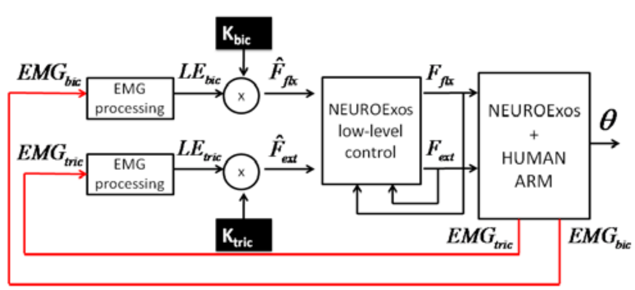

Fig. 2 Block diagram of the proportional EMG controller.

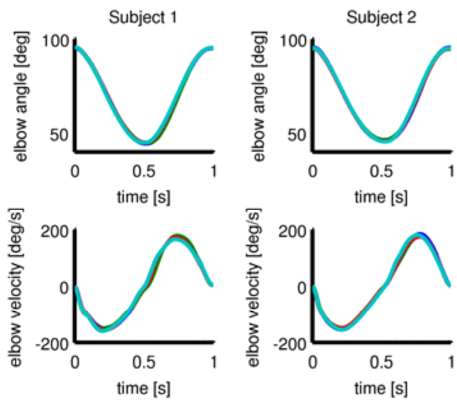

Fig. 3 Mean position and velocity profiles over different assistance levels for the two subjects. Different color lines are used for each assistance level.

provided to subjects using a computer screen, which displayed current elbow angle and target movement range by means of a vertical cursor, an upper and a lower bound. In addition, a metronome supplied the desired movement pace to which the user was asked to synchronize. While performing this cyclical movement, subjects experienced three increasing level of assistance obtained by setting the actual force gain values (K_{bic} and K_{tric}) to 50%, 100% and 150% of the preferred values previously chosen. Each level of assistance lasted one minute, and was interleaved by 1 minute of no-assistance condition (K_{bic} and K_{tric} equal to 0).

III. Results

Data acquired during the flexion/extension trials (e.g. joint angle, cable force sensors, EMG envelopes) were divided into sequences of 60 seconds, with homogeneous levels of assistance provided to the subjects. Each sequence was separated in flexion/extension cycles using a peak detection algorithm. The cycles during which a transition occurred were not included in the analysis.

Within each cycle, we computed cycle amplitude (difference between maximum and minimum angular position) and cycle duration, in order to assess the fulfillment of the kinematic task across the different assistance conditions. Table I reports the mean cycle amplitude and duration along with standard deviations for each assistance condition experienced by the two subjects.

Mean position and velocity profiles were also calculated for each assistance level and are reported in Fig. 3 with different colors, in order to verify if movement kinematics was altered by the provided assistance.

To explore the effect of the proposed assistive control on the subject's muscular activity, mean and maximum value of biceps and triceps EMG envelopes were computed. Fig. 4 shows the mean biceps envelope profiles of the two subjects, for each level of assistance, using different color lines. Finally, Fig. 5 reports the peak of biceps envelope, for each movement cycle overall the trial in order to investigate if any kind of trend was present.

Table I
Average cycle amplitude and duration for the each assistance level.

		0	50%	100%	150%
Subject 1	Cycle amplitude	53.43°± 4.33°	53.62°± 2.98°	53.48°± 3.56°	54.46°± 4.62°
	Cycle duration	0.997± 0.042 s	1.00± 0.038 s	0.983± 0.064 s	0.989± 0.066 s
Subject 2	Cycle amplitude	48.77°± 2.40°	48.91°± 2.61°	49.71°± 2.59°	50.47°± 3.47°
	Cycle duration	0.991± 0.041 s	0.982± 0.037 s	1.002± 0.054 s	1.001± 0.052 s

IV. Discussion

Mean cycle amplitude and duration (Table I) did not change significantly across different assistance levels. Moreover, standard deviation values, gathered from different assistive levels, are very similar, meaning that elbow movement variability was not altered by the assistive control. Besides mean amplitude and duration, the kinematic profiles, as performed by subjects under different assistance level, were also very similar, to such an extent that lines of different colors are hardly visible on Fig. 3. These experimental results clearly demonstrate that movement kinematics was not significantly altered in any tested condition. Subjects could keep the full control of their arm movement despite the "simplified" assistive controller.

On the contrary, biceps EMG envelopes, as shown in Fig. 4, were significantly decreased by means of the extra torque provided by the exoskeleton, indicating an effective reduction of the effort spent by subjects for movement generation. This result is particularly encouraging if we consider the well-known difficulties of EMG-based control in assisting movements that require very low muscular effort, and consequently produce low EMG signals, such as the unconstrained elbow movement that we tested [14]. Similar conclusions can be drawn by analyzing the peaks of the biceps EMG envelope reported in Fig. 5. Despite the observed variability, a clear reduction can be seen if the mean of the peak over each assistance level is considered (horizontal black segments of Fig. 5).

Quite surprisingly, the analysis of the peaks of biceps EMG envelope, does not reveal any clear trend that could underline a motor learning process such e.g. the one observed by Ferris [9] who used a similar EMG proportional

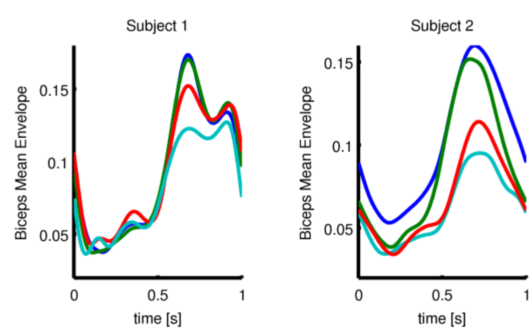

Fig. 4 Mean biceps envelope profiles of the two subjects. Different color lines are used for each assistance level (blue: K =0%; green: K = 50%; red: K = 100%, cyan: k = 150%).

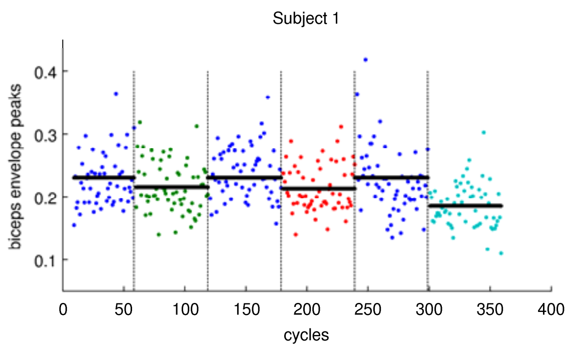

Fig. 5 Peaks of biceps EMG envelope recorded on each cycle overall the experiment. Different colors are used for each assistance level. The vertical dashed lines indicate the time when the assistance level changed. The black line corresponds to the mean of the envelope peaks over a specific assistance level (blue: K =0%; green: K = 50%; red: K = 100%, cyan: k = 150%).

controller for an ankle exoskeleton. EMG envelop peaks were apparently constant during the 60 seconds performed at a constant assistance level, while one should have foresee that as subjects adapted and learned how to exploit the assistance provided, the effort spent, and then the EMG peaks, should have been reduced. Further analyses and studies should be performed to answer these questions.

V. Conclusions and Future Works

This work investigates about the possibility of using a proportional EMG controller for providing assistance through powered exoskeletons. We postulate that humans can rely on their motor adaptation ability to compensate for the inaccuracy of the controller, which provides the user with an additional torque that does not correspond to a constant fraction of that actually needed to perform the intended movement. Preliminary results reported in the paper show that subjects can effectively adapt to this kind of assistance and take advantage of the assistive torque.

Acknowledgment

Authors would like to thank Dr. Stefano Roccella and Dr. Fabrizio Vecchi, for their valuable support.

References

[1] C.J. Walsh, K. Endo, and H. Herr, "A Quasi-Passive Leg Exoskeleton For Load-Carrying Augmentation," *Int. J. of Humanoid Robotics*, vol. 4, 2007, p. 487.

[2] A. Zoss, H. Kazerooni, and A. Chu, "Biomechanical design of the Berkeley lower extremity exoskeleton (bleex)," *IEEE/ASME Trans. on Mechatronics*, vol.11, no.2, pp. 128 –138, April 2006.

[3] E. Rocon, J.M. Belda-Lois, F. Ruiz, M. Manto, J.C. Moreno, and J.L. Pons, "Design and validation of a rehabilitation robotic exoskeleton for tremor assessment and suppression," *IEEE Trans. on neural systems and rehabilitation engineering*, vol. 15, 2007, pp. 367-78.

[4] H. Kawamoto, T. Hayashi; T. Sakurai; K. Eguchi and Y. Sankai "Development of single leg version of HAL for hemiplegia," *IEEE EMBC*, pp.5038-5043, 3-6 Sept. 2009

[5] http://berkeleybionics.com/exoskeletons-rehab-mobility/

[6] Mihelj, M.; Nef, T.; Riener, R.; , "ARMin II - 7 DoF rehabilitation robot: mechanics and kinematics," IEEE ICRA, 2007 pp.4120-4125.

[7] Kyoungchul Kong; Tomizuka, M.; , "Control of Exoskeletons Inspired by Fictitious Gain in Human Model", IEEE/ASME Transactions on Mechatronics, , vol.14, no.6, pp.689-698, Dec. 2009.

[8] Ronsse, R.; Vitiello, N.; Lenzi , T.; Kieboom , J. v. d.; Carrozza, M. C.; Ijspeert , A. J.; , "Human–Robot Synchrony: Flexible Assistance Using Adaptive Oscillators" IEEE Transactions on Biomedical Engineering, vol.58, no.4, pp.1001-1012, April 2011.

[9] K.E. Gordon and D.P. Ferris, "Learning to walk with a robotic ankle exoskeleton," Journal of Biomechanics, 2007, vol.40 pp. 2636-44.

[10] Kazerooni, H, R. Steger, L. Huang "Hybrid Control of the Berkeley Lower Extremity Exoskeleton, "The International Journal of Robotics Research", Vol 25, No 5-6, pp. 561-573, May-June 2006.

[11] T.S. Buchanan, D.G. Lloyd, K. Manal, and T.F. Besier, "Neuromusculoskeletal modeling: estimation of muscle forces and joint moments and movements from measurements of neural command.," Journal of applied biomechanics, v 20, 2004, pp. 367-95.

[12] R.W. Norman and P.V. Komi, "Electromechanical delay in skeletal muscle under normal movement conditions.," Acta physiologica Scandinavica, vol. 106, Jul. 1979, pp. 241-8.

[13] R. Song and K.Y. Tong, "Using recurrent artificial neural network model to estimate voluntary elbow torque in dynamic situations.," Medical & biological engineering & computing,, 2005, pp. 473-80.

[14] Kiguchi, K.; Kariya, S.; Watanabe, K.; Izumi, K.; Fukuda, T.; "An exoskeletal robot for human elbow motion support-sensor fusion, adaptation, and control," IEEE Transactions on Systems, Man, and Cybernetics, Part B: Cybernetics, vol.31, no.3, pp.353-361, Jun 2001.

[15] Cavallaro, E.E.; Rosen, J.; Perry, J.C.; Burns, S.; , "Real-Time Myoprocessors for a Neural Controlled Powered Exoskeleton Arm" IEEE Transactions on Biomedical Engineering, vol.53, no.11, pp.2387-2396, Nov. 2006.

[16] Fleischer, C.; Hommel, G.; , "A Human--Exoskeleton Interface Utilizing Electromyography" IEEE Transactions on Robotics, vol.24, no.4, pp.872-882, Aug. 2008.

[17] Clancy, E.A.; Hogan, N.; "Relating agonist-antagonist electromyograms to joint torque during isometric, quasi-isotonic, nonfatiguing contractions" IEEE Transactions on Biomedical Engineering, vol.44, no.10, pp.1024-1028, Oct. 1997.

[18] T. Lenzi, N. Vitiello, S.M.M. De Rossi, S. Roccella, F. Vecchi, M.C. Carrozza, "NEUROExos: a variable impedance powered elbow exoskeleton", in Proc. of the 2011 IEEE International Conference on Robotics and Automation (ICRA), pp. 1419-1426, 2011.

[19] T. Lenzi, S. De Rossi, N. Vitiello, A. Chiri, S. Roccella, F. Giovacchini, F. Vecchi, M.C. Carrozza, "The neuro-robotics paradigm: NEURARM, NEUROExos, HANDEXOS", IEEE EMBC 2009, Minneapolis, USA, pp. 2430-2433.

[20] T. Lenzi, N. Vitiello, S.M.M. De Rossi, A. Persichetti, F. Giovacchini, S. Roccella, F. Vecchi, M.C. Carrozza, "Measuring human-robot interaction on wearable robots: A distributed approach", *Mechatronics* (2011), doi:10.1016/j.mechatronics.2011.04.003.

[21] N. Vitiello, T. Lenzi, S.M.M. De Rossi, S. Roccella, M.C. Carrozza, "A sensorless torque control for Antagonistic Driven Compliant Joints", *Mechatronics*, vol. 20(3), pp. 355-367, 2010.

[22] H.S. Milner-Brown, R.B. Stein, and R. Yemm, "The contractile properties of human motor units during voluntary isometric contractions.," The Journal of physiology, 1973, pp. 285-306.

[23] T.K.K. Koo and A.F.T. Mak, "Feasibility of using EMG driven neuromusculoskeletal model for prediction of dynamic movement of the elbow.," Journal of electromyography and kinesiology, vol. 15, Feb. 2005, pp. 12-26.