ARCHIVIO ISTITUZIONALE
DELLA RICERCA

Alma Mater Studiorum Università di Bologna
Archivio istituzionale della ricerca

Robotic Muscular Assistance-As-Needed for Physical and Training/Rehabilitation Tasks: Design and Experimental Validation of a Closed-Loop Myoelectric Control in Grounded and Wearable Applications

This is the final peer-reviewed author’s accepted manuscript (postprint) of the following publication:

Published Version:

Availability:
This version is available at: https://hdl.handle.net/11585/874198 since: 2022-02-28

Published:
DOI: http://doi.org/10.1007/978-3-030-71356-0_2

Terms of use:

Some rights reserved. The terms and conditions for the reuse of this version of the manuscript are specified in the publishing policy. For all terms of use and more information see the publisher's website.

This item was downloaded from IRIS Università di Bologna (https://cris.unibo.it/).
When citing, please refer to the published version.

(Article begins on next page)

27 April 2024

This is the final peer-reviewed accepted manuscript of:

Meattini, R., Chiaravalli, D., Hosseini, M., Palli, G., Paik, J., Melchiorri, C. (2021). Robotic Muscular Assistance-As-Needed for Physical and Training/Rehabilitation Tasks: Design and Experimental Validation of a Closed-Loop Myoelectric Control in Grounded and Wearable Applications. In: Saveriano, M., Renaudo, E., Rodríguez-Sánchez, A., Piater, J. (eds) Human-Friendly Robotics 2020. HFR 2020. Springer Proceedings in Advanced Robotics, vol 18. Springer.

The final published version is available online at: https://doi.org/10.1007/978-3-030-71356-0_2

Rights / License:

The terms and conditions for the reuse of this version of the manuscript are specified in the publishing policy. For all terms of use and more information see the publisher's website.

This item was downloaded from IRIS Università di Bologna (https://cris.unibo.it/)

When citing, please refer to the published version.

Robotic Muscular Assistance-As-Needed for Physical and Training/Rehabilitation Tasks: Design and Experimental Validation of a Closed-Loop Myoelectric Control in Grounded and Wearable Applications

Roberto Meattini[1], Davide Chiaravalli[1], Mohssen Hosseini[2], Gianluca Palli[1], Jamie Paik[2] and Claudio Melchiorri[1]

[1]DEI - Department of Electrical, Electronic and system Engineering "Guglielmo Marconi", University of Bologna, Bologna, Italy
{roberto.meattini2, davide.chiaravalli2, gianluca.palli, claudio.melchiorri}@unibo.it
[2]Reconfigurable Robotics Lab (RRL), École polytechnique fédérale de Lausanne (EPFL), Lausanne, Switzerland
{mohssen.hosseini, jaime.paik}@epfl.ch

Abstract. In this work a solution for the design of an assistive system for both *muscular effort compensations* and *muscular effort generations* for physical and rehabilitation tasks is presented. The proposed human-in-the-loop (HITL) control directly exploits the subject muscle sEMG signals measures to produce a specified and repeatable muscular response, without the need for human joint torque estimations. A set of experimental tests addressing different assistive tasks are proposed to validate the control design. Moreover different robotic devices, both grounded and wearable, are considered to assess the control under different working scenarios. The experimental results, involving four healthy subjects, show the efficacy of the proposed approach and the successful compensation/generation of the subject effort in the different assistive tasks considered.

Keywords: Robotic Assistance-As-Needed · Human-Robot Physical Interaction · Human-In-The-Loop · sEMG · Myoelectric Control.

1 Introduction

In the last decades the advancements in control capabilities and precision of robotic platform have stimulated the development of many applications in the

This work was supported by the European Commission's Horizon 2020 Framework Programme with the project REMODEL - Robotic technologies for the manipulation of complex deformable linear objects - under Grant 870133

assistive robotic field, focused on the enhancement of human performances in tasks concerning both neuromuscular rehabilitation and/or physical strength augmentation [1]. The robotic system, characterized as a kinematic chain physically connected with the user [2], is set to react to the estimated human effort through the generation of an assistive torque, that can reduce the workload for hard labour workers or improve the effectiveness of many physical muscle therapies. In neuro-muscular rehabilitation, the resistive torque provided by the assistive robot induces the patient to generate a required specific effort profile [3]. Similarly in *prehabilitation*, the preparation for future limb inactivity or hospitalization [4], the patient muscles are trained and strengthened with cycles of specific muscle contractions, generated in response to resistive torques [5]. This kind of training proves determinant to mitigate the effect of muscle weakness that can be caused by long period of reduced mobility, as it might happen after surgery or severe diseases [6], [7].

The control of such robotic systems proves very challenging because of the many different aspects involved. Indeed, the kinematics of the robotic platform can vary greatly from *grounded assistive robots*, where the assistive effort is trasmitted to the human arm through a physical connection with the robot end-effector only, to *wearable assistive robots*, where the human-like structure allows for a more distributed application of forces. Moreover the human intent and action must be considered and handled in a so called human-in-the-loop control framework [8]. Eventually the techniques that can be used to estimate the human intentions and effort can lead to many different control strategies and results, that are still in most cases of limited applicability in a real case scenario. Therefore robotic assistive applications are still considered an open problem of wide interest in the research community.

Among the techniques for the human effort estimation, the surface skin electromyography (sEMG) has proved to be an effective way to acquire accurate information for the detection of the human intentions [9]. Many studies have exploited them in task- and user-dependent algorithm calibrations, for example using models of the muscular system [2], neural network and fuzzy classifiers [10], or dynamic-model-based estimations of the human torque in order to provide needed robotic assistance [11, 12]. These approaches require complex training procedures and/or identification/optimization procedures, resulting to be arduously usable outside laboratories and therefore with a limited applicability. Other methods avoiding human joint torque estimations have been proposed, e.g. see [13], however limited to an *a posteriori* analysis of the assistance exploited by the user, without the possibility of imposing quantifiable assisting targets.

Differently, in this work a sEMG-based, assistance-as-needed, HITL control system is proposed, directly exploiting the filtered sEMG signals to achieve a target user muscle activity level. This way it is possible to produce a specific measurable and repeatable response from the human operator by relying on its adaptability to external inputs without the need of imprecise and complex torque evaluations. In particular this paper focuses on some specific assistance goals for muscle training/rehabilitation tasks related to the elbow joint. Other

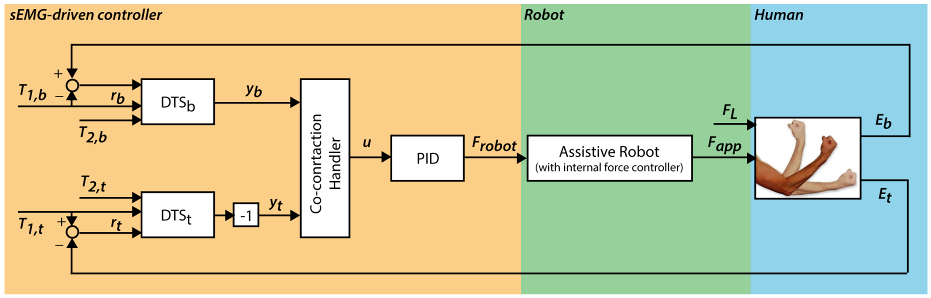


Fig. 1. Block diagram of the proposed sEMG-based control.

recent works have focused on muscle minimization activity, but have been limited by time demanding algorithms [14] or numerical simulations [15]. Conversely in this work an experimental test with four healthy subjects involved in physical and training/rehabilitation tasks is proposed, both with grounded and wearable devices. The final aim of the work is to prove the feasibility of the proposed approach in obtaining a target effort compensation/generation related to the elbow joint and directly driven by myoelectric signals.

2 Methods and Tools

2.1 Description of the sEMG-based Control

The sEMG-based control proposed in this work exploits the sEMG signal measurements from the human arm biceps and triceps muscles to bring the muscular activity within a predetermined level set, chosen according to the specific assistive task. The system, described in Fig. 1, is characterized by a robotic device physically attached to the human limb for the generation of the assistive torque F_{app} and two sensors placed on the user arm that read the sEMG signals E_b and E_t generated by the biceps and triceps muscles respectively. The sEMG driven controller exploits the data received by the sensors to define the torque reference F_{robot} for the assistive robot in order to reduce/increase the operator effort. The required muscular activity is obtained by taking into consideration both the robotic assistive force and the external force F_L applied to the forearm by the user. The controller enforces a control loop where the assistive force value is adapted to the user motion until the expected level of muscular activity $T_{1,i}$ is reached. The generation of the assistive force reference is composed by four steps. At first the signal error r_i is evaluated:

$$r_i = T_{1,i} - E_i, \quad (1)$$

with $i = \{b, t\}$ representing the biceps and triceps muscles respectively. Then a double threshold strategy (DTS blocks in Fig. 1) is applied through the definition of an hysteresis band to prevent unwanted oscillations of the reference torque.

Threshold Values		Assistance Modality
$T_{1,b} > T_{2,b}$		*Effort Generation* (Biceps, see Eq. (6))
$T_{1,b} < T_{2,b}$	$T_{1,t} < T_{2,t}$	*Effort Compensation (both biceps and triceps)*
	$T_{1,t} > T_{2,t}$	*Effort Generation* (Triceps, see Eq. (6))
	$T_{1,t} = T_{2,t}$	No Assistance/System Inactive
$T_{1,b} = T_{2,b}$		No Assistance/System Inactive

Tab. 2. Combinations of thresholds for the assistive modalities.

To this end a second threshold value $T_{2,i}$ is defined to properly characterize the hysteresis behaviour: the closed loop adaptation of the assistive force is activated only when the muscular signal exceedes the activation threshold $T_{2,i}$ but is stopped only after the reference threshold $T_{1,i}$ has been reached. The dimension of the hysteresis band B is there described by

$$B = |T_{1,i} - T_{2,i}|. \tag{2}$$

The two threshold values are defined according to the assistive task and the target muscle and are summarized in Tab. 2. The double threshold strategy can

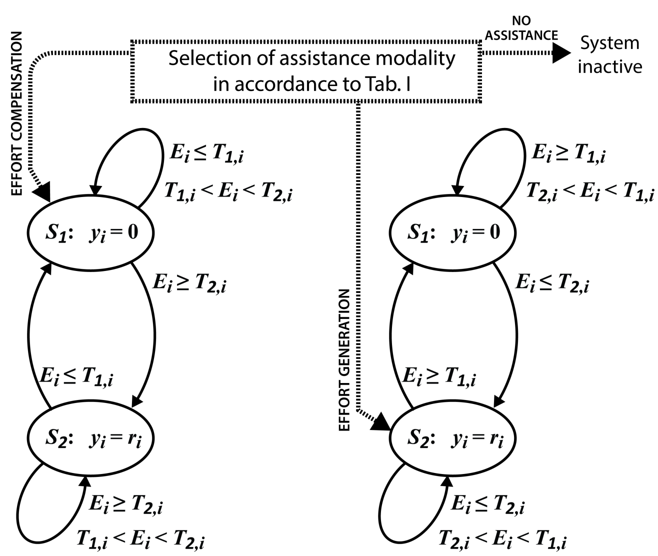


Fig. 3. Finite State Machine logic of the DTS_i block of Fig. 1.

be addressed more in details according to the state machine defined in Fig. 3. Let's define two states S_1 and S_2 of the system S corresponding to the activation and deactivation of the force adaptation. When the system is in state S_1 no adaptation is in place and the assistive reference y_i is kept constant to the last learned value. Adversely when in state S_2 the adaptation is active and the assistive reference y_i is given in accordance to the sEMG readings as follows:

$$y_i = \begin{cases} 0, & \text{if S} = S_1 \\ r_i, & \text{if S} = S_2 \end{cases}. \tag{3}$$

The system enters in state S_2 when the sEMG reading E_i surpasses the activation threshold $T_{2,i}$. Conversely when the reading reaches the reference threshold $T_{1,i}$ the state S_1 is restored. The threshold activation condition is dependend on the assistive task as shown in Fig. 3. Once the assistive reference y_i is generated for both the biceps and triceps muscles the two signals are exploited to define the reference input u for the PID controller according to a co-contraction strategy

$$u = f(y_y, y_b). \tag{4}$$

that is dependent on the specific assistive task (Co-contraction handler block in Fig. 1). This way any co-contraction is automatically filtered out and the system is left unaffected. More details on the specific co-contraction functions for each task will be presented in each task description. Eventually the tuning of the PID controller defines the dynamics for the generation of the assistive force reference F_{app} sent to the robot.

2.2 Effort Compensation for Physical Tasks

In an effort compensation assistive task the user forearm is requested to generate a force $F_L \neq 0$ using a maximum predefined effort value characterized by the activation threshold $T_{2,i}$. An increase in effort E_i above the threshold would represent and excessive workload on the human arm and would cause a change of state of the control system from the idle state S_1 to the assistive state S_2. The activation of the assistance would cause an increasing robotic support F_{app} that would produce a consequent effort reduction until the reference threshold $T_{1,i}$ is reached. This would cause an new change to the idle state S_1 where the adaptation of the assistance is stopped and the robot generates the required assistive torque value. The state machine cycle of the system is repeated any time the activation threshold is reached, until the system stabilizes on the final assistive torque and the user effort is kept below the maximum value for the whole task. In this scenario the Co-contraction Handler function acts in case both the biceps and triceps muscle activates at the same time (in case of (un)voluntary contractions), according to:

$$u = \begin{cases} y_t, & \text{if } \ y_b = 0 \\ y_b, & \text{otherwise} \end{cases}. \tag{5}$$

This way the triceps muscle is considered only when the biceps is at rest.

2.3 Effort Generation for Training/Rehabilitation Tasks

In an effort generation assistive task the user forearm is requested to generate a specific muscle activity during the motion. Therefore no external force should be present ($F_L = 0$). In this scenario the activation threshold defines the minimum muscular activity $T_{2,i}$ required for the human forearm during the motion of the elbow. The assistive control is activated when the human effort E_i goes below the minimum value and the system switches to state S_2. Conversely when the human reaches the required effort the state S_1 is restored. The closed loop adaptation follows a cyclical change of the system state in a similar fashion to the effort compensation task. The Co-contraction Handler function described by

$$u = \begin{cases} y_b, & \text{if } T_{1,b} > T_{2,b} \\ y_t, & \text{if } T_{1,t} > T_{2,t} \wedge T_{1,b} < T_{2,b} \end{cases}, \tag{6}$$

ensures, in this case, that only the selected target muscle is actually considered for the generation of the assistive force reference.

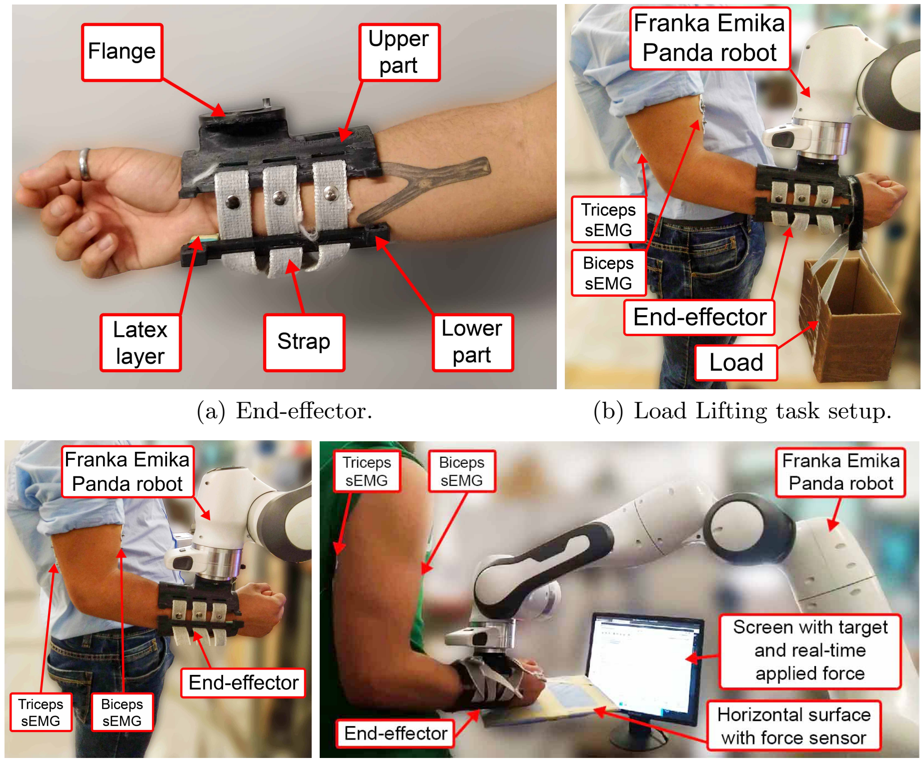
3 Experiment

In the following an experimental test characterized by two experimental cases related to the use of a grounded and a wearable assistive device respectively is reported. For each experimental case results concerning different assistive tasks are reported. The test involved four subjects, from now on denominated as U_1, U_2, U_3, U_4. The aim of the test is to prove the feasibility of the proposed control scheme and to show that is possible to achieve a specific measurable and repeatable response by directly exploiting sEMG signals in the control loop.

3.1 Experimental Case #1: Grounded Assistive Application

The first experimental case focused on grounded assistive applications. The control system, showed in Fig. 3, consists of a robotic manipulator connected with the user forearm through a purposedly designed end-effector. The design of the tool has considered both reusability for different subjects and ergonomicity. Two 3D printed enclosures are solidly joined with the arm by means of tight straps. At the same time the flange on the upper part of the end-effector allows for a fixed connection with the assistive robot. A layer of anti-allergic latex acts as contact medium and protects the arm. On the robotic side the Franka Emika Panda collaborative lightweight robot has been chosen as assistive platform [16]. The robot, characterized by a redundant kinematic chain, can provide both position and torque control capabilities and a real time estimation of its dynamic

Four healthy participants have been considered (males, right-handed, age: 30.5 ± 4). The experiments have been carried out in accordance with the Declaration of Helsinki. All test subjects received a detailed explanation of the experimental protocol and signed an informed consent form.



(a) End-effector.

(b) Load Lifting task setup.

(c) Muscle training/rehab task setup.

(d) Pressure on Surface task setup.

Fig. 4. Experimental setup of the grounded assistive application.

parameters, proving therefore a valid choice for the kind of application considered. The grounded test case considered three different assistive tasks: "Load Lifting", "Pressure on Surface" and "Muscle Training/Rehabilitation".

Experimental Task Description and Protocol The *load lifting task* deals with the cyclical lifting and lowering of a load of predefined weight. For this test a load of $2kg$ applied to the wrist is considered. Initially the subject is asked to wait with the elbow flexed at 90^o. Then, after 5–10s, the assistive control is enabled, and the robot starts to progressively assist the subject. When the control system completes the adaptation phase and enters the idle state S_1, the user is requested to perform a sequence of 8 slow and steady extension-flexion motions, moving the elbow in the range 90^o to 10^o. The sEMG and force assistive signals are recorded for the duration of the whole test.

The *pressure on surface task* considers the application of a predetermined force on an horizontal surface by means of the elbow joint alone. The objective of the assistive task was to reduce the user effort below a specific threshold level.

In this case, each subject is asked to keep a 90^o elbow angle and apply a vertical force of $35N$ on a surface located $10cm$ below her wrist. An ATI force sensor (ATI Multi-axis Force/Torque Sensor System ISA F/T-16), placed beneath the surface provided the measure of the applied force through a screen placed in front of the subject. This way the user receives a precise feedback on its motion and the expected level of pressure was successfully applied in a continuous manner (the setup for this test can be seen in Fig. 4(a),4(b),4(c),4(d)). As in the previous test, the robotic assistance is activated only after an initial period of about $15s$. During the whole test each subject is asked to mantain constant the overall level of pressure of $35N$ on the surface.

In the *muscle training/rehabilitation task* the objective was to obtain a specific minimum muscular activity in the subject during elbow flexion/extention motions. Initially each user is asked to place the elbow in a 90^o position and wait for $10s$, similarly to the load lifting case. Then, she is asked to perform 8 extension-flexion repetitions, moving the elbow from 90^o to 10^o in a slow and smooth way. This test was repeated for each subject for both the biceps and the triceps muscles.

Single-Subject Results In Fig. 5 the result for the load lifting task performed by the subject U_2 are shown. Ten time windows $C_1, C_2, ..., C_{10}$ have been defined in the graph to properly highlight the 8 extension-flexion elbow motions and the starting phase (top graph). During the motion a direct comparison of the sEMG signal E_b (middle graph) and of the assistive force F_{robot} (bottom graph) is reported. Initially the assistive system is disabled (time window C_1) and the sEMG signal stands above the activation threshold $T_{2,b}$. Then as soon as the assistive system is enabled (time window C_2), an increase in the assistive force can be seen and a consequent reduction of the muscular activity. Once the reference threshold $T_{1,b}$ is reached the system enters the state S_1 and the force assistance remain constant (C_3 time window). Each time during the flexion motion the activation threshold is passed, again the system enters the state S_2 and an additional increase in force can be noticed (time window C_3, C_4). From C_5 the sEMG signal remain below the activation threshold, proving that the expected reduction in muscular activity has been successfully obtained.

Fig. 6 reports the result for the Pressure on Surface task for subject U_3. In the second time window C_2, the user applies the $35N$ required force on the horizontal surface and a consequent increase of muscular activity can be noticed. Then in C_3 the assistive system is enabled and the assistive force produce a decrease of muscular activity until in C_4 the sEMG muscular signal is kept below the maximum target value $T_{2,b}$. The system is therefore stabilized to the state S_1. Also in this case it is possible to see that an effective reduction in muscular effort has been achieved.

The results for the muscle training/rehabilitation task can be observed in Fig. 7. In particular Fig. 7(a) shows the results of the biceps muscle effort generation for U_3 and Fig. 7(b) the results of the triceps muscle effort generation for U_4. As with the load lifting task, in both cases after an initial resting period the assitive

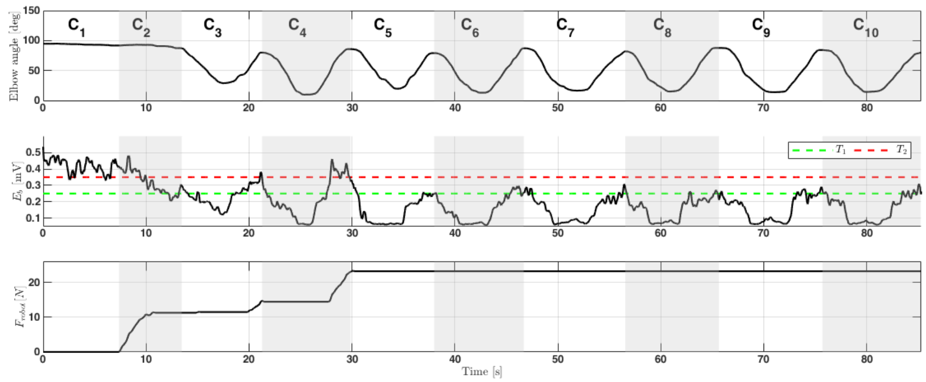


Fig. 5. Single-subject results for the Load Lifting task (subject U_2; $2kg$ load.)

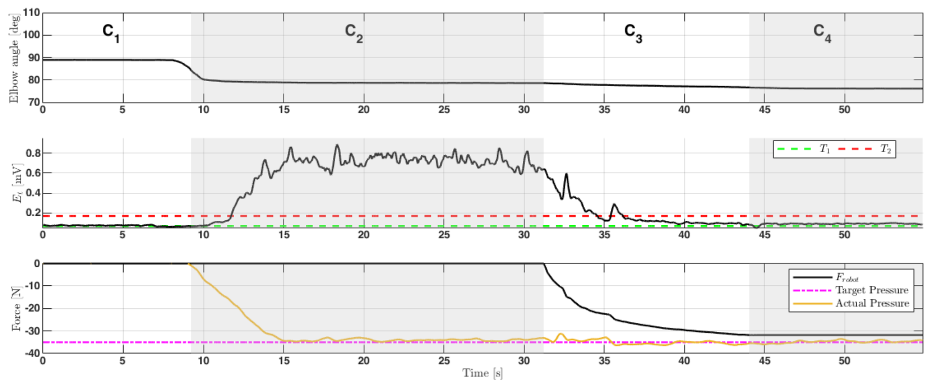
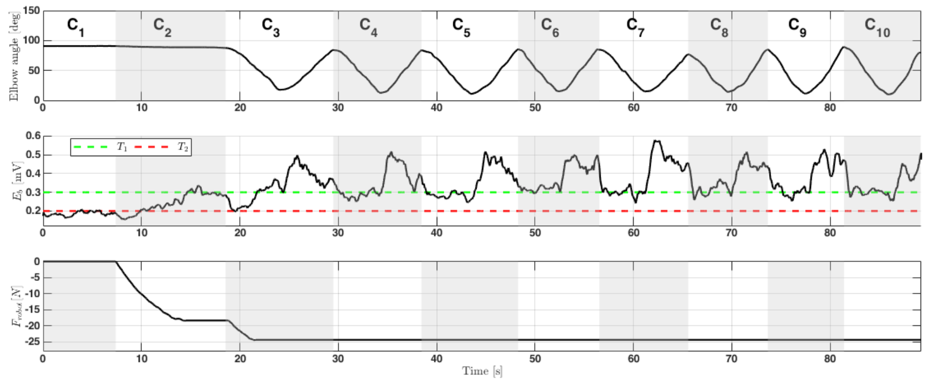


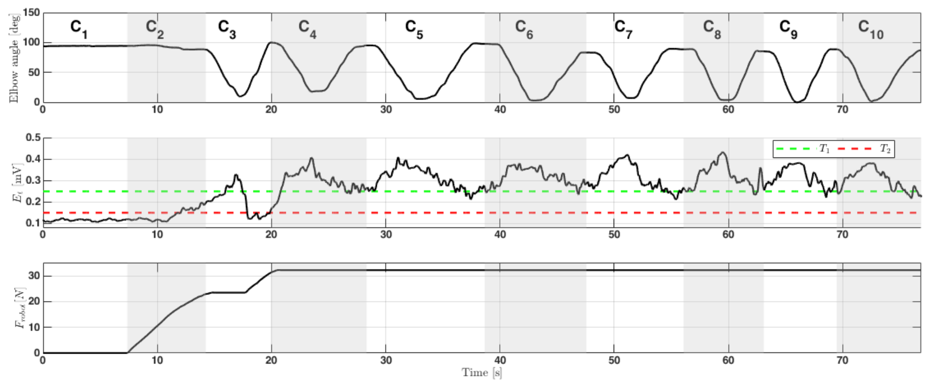
Fig. 6. Single-subject results for the Pressure on Surface task (subject U_3; $35N$ pressure force.)

control is activated (time window C_2) and an increase in muscular activity can be seen. In this test case, the objective was to induce a minimum level of muscular activity on the subject, therefore the activation threshold $T_{2,i}$ is placed below the each reference threshold $T_{1,i}$ and an increase in the sEMG signal is expected. After a single adaptation cycle C_3 the expected level of muscular activity is successfully achieved.

Global Results In Fig. 8 the global results of the assistive test over the four test subjects are reported. Both the sEMG biceps and triceps muscular signals and the assistive force results are proposed in each time window. A normalization with respect to the reference threshold $T_{1,i}$ has been applied to all sEMG data to allow comparation between the different test subjects. In both the Load lifting task (Fig. 8(a)) and the muscle training task (Fig. 8(c)) it is possible to see that



(a) Subject U_3 results for the biceps effort generation.



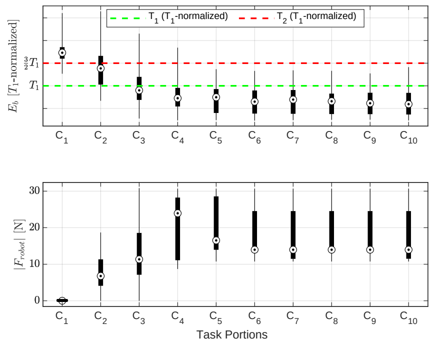
(b) Subject U_4 results for the triceps effort generation.

Fig. 7. Single-subject results for the Muscle Strength Training task.

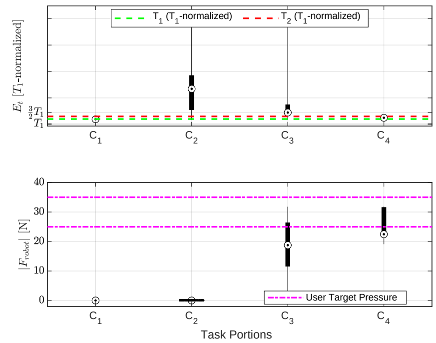
for all subjects after the first elbow flexion motion (time window C_4) the target level of muscular activity has been successfully reached. Therefore the system proved to be able to quickly adapt to the user in order to obtain the expected muscular performances. Also the results of the Pressure on surface task (Fig. 8(b)) show that for all subjects after the activation of the assistive control the triceps sEMG signal is successfully kept steady below the reference threshold. The result proposed confirm the feasibility of the proposed approach in setting an upper/lower limit to the muscular activity required in a specific assistance task by means of a grounded assistive device.

3.2 Experimental Case #2: Wearable Assistive Application

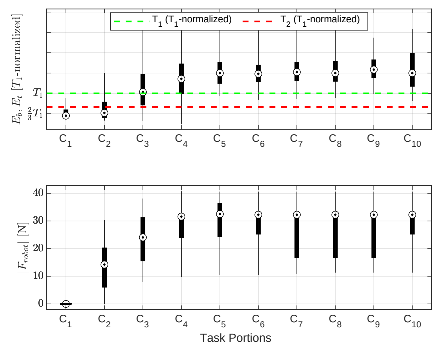
The second experimental case focused on wearable assistive applications. The control system, as showed in Fig. 9 is characterized by a soft elbow ExoSuit weared by the subject and able to apply the assitive force through the forearm support connected to the user arm. The suit exploits Twisted String Actuation



(a) Aggregated results for the Load Lifting task.



(b) Aggregated results for the Pressure on Surface task.



(c) Aggregated results for the Muscle Training/Rehabilitation task.

Fig. 8. Global results of the experimental tasks over the four subjects.

(TSA) technology to produce a linear motion by means of the twist of a string operated by an electric motor. Two TSA modules encase the motors and are mounted on the back of the user to provide the actuation for the system. A force sensor inside each of the module allow a proper evaluation of the tension on the string. A tendon connects each string to the user arm support by means of a sliding element on the shoulder. A set of straps allows an easy fastening of the suit to the body. Since with TSA no additional transmission element is required, the overall weight of the ExoSuit is only of 1650 g. For all the details about TSA technology and related development of the ExoSuit please refer to [17–20], being this knowledge outside of the scope of the present work.

Experimental Task Description and Protocol For this experimental case a single assistive test was performed under a load lifting scenario. Similarly to the grounded case, the subjects was asked to perform a slow and smooth cyclical extension-flexion motion of the elbow between 90° and 30°. The user was required to put particular attention in avoiding shoulder motions in order

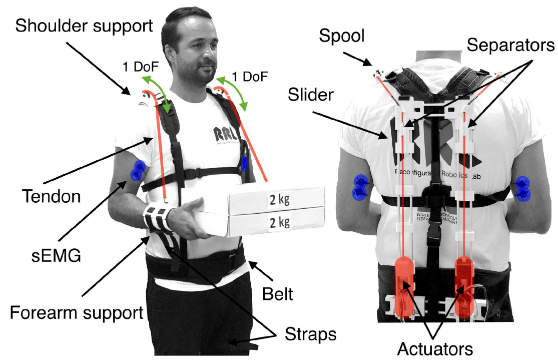


Fig. 9. Overview of the TSA-based ExoSuite presented in [18].

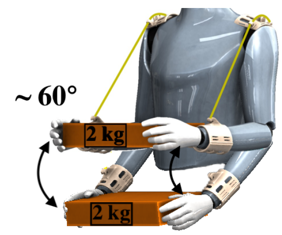


Fig. 10. The 2 kg box lowering-lifting task.

to keep the shoulder joint passive. In this test, a load of 2 kg directly held by the subjects was chosen (see Fig. 10).

Single-subject Results In Fig. 11 the result of the load lifting task for U_1 are reported. Both the data related to the assistive system for the right arm (graphs on the left) and the left arm (graphs on the right) are reported. At the beginning the system in not enabled, and the sEMG signals (top graphs) show a muscular activity beyond the required threshold. Then, as soon as the assistive system is enabled (red zone), an immediate reduction of the biceps effort, alongside an increase in the tendon tension (bottom graphs) and a consequent change in string length (p_{ref} middle graphs) can be noticed. Once the system has stabilized the user is requested to perform the cyclical lifting task (blue and yellow zones), causing the consequent adaptation of the system according to biceps and triceps signals. During the task the signal remain within the required threshold band almost at all times. Only a quick surpassing of the threshold was necessarily present for (and bounded to) the onset of the lowering/lifting motions.

Global Results The global results for the load lifting task proposed in Fig. 12, show that, overall, the mean muscular effort of the subject is successfully reduced within the expected band. Also for this test case the sEMG signals have been normalized to allow an effective comparison between subjects.

3.3 Selection of the Threshold Values

For the Load Lifting and Pressure on Surface task each subject was asked to keep the forearm elbow at a 90° orientation while a $1kg$ load was applied to it. During a $10s$ time window, a calibration set E_C was therefore recorded and

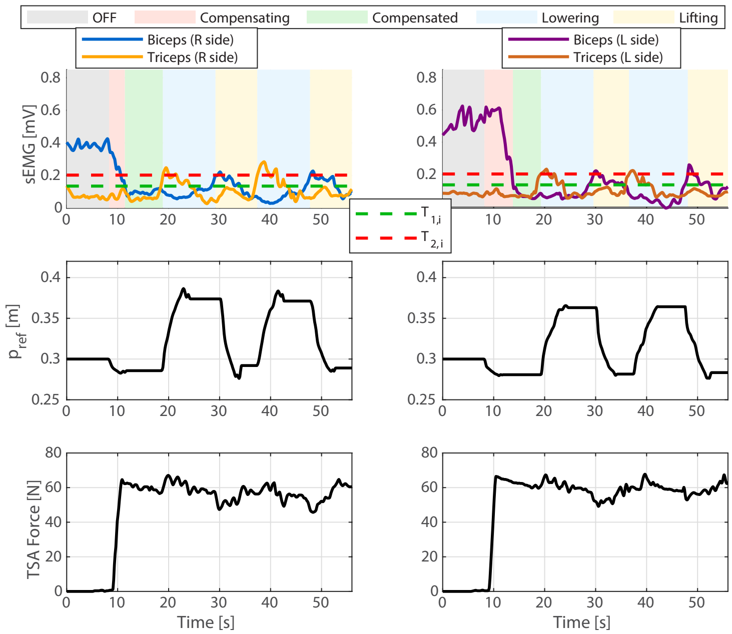


Fig. 11. Lowering-lifting experiment of the subject U1.

exploited to evaluate the required thresholds:

$$\begin{aligned} T_{2,b} &= \mu_{C,b} - \sigma_{C,b}, \quad T_{1,b} = 2T_{2,b}/3, \\ T_{1,t} &= \mu_{C,t} + \sigma_{C,t}, \quad T_{2,t} = 3T_{1,t}/2. \end{aligned} \tag{7}$$

where $\mu_{C,b}$, $\mu_{C,t}$ are the mean values of the biceps' and triceps' sEMG signals in the calibration set, and $\sigma_{C,b}$, $\sigma_{C,t}$ are their standard deviations. A similar procedure was applied for the Muscle Training/Rehabilitation task without any load applied. In this case the thresholds were evaluated as

$$\begin{aligned} T_{2,b} &= \mu_{R,b} + \sigma_{R,b}, \quad T_{1,b} = 3T_{2,b}/2, \\ T_{2,t} &= \mu_{R,t} + \sigma_{R,t}, \quad T_{2,t} = 3T_{2,t}/2. \end{aligned} \tag{8}$$

where $\mu_{R,b}$, $\mu_{R,t}$ and $\sigma_{R,b}$, $\sigma_{R,t}$ are the mean values and the standard deviations over the calibration set.

3.4 Conclusions

In this work a novel assistance-as-needed HITL control for assistive tasks relying on sEMG signals has been proposed. The control system directly exploits sEMG signals of the biceps and triceps muscles of the subject to impose a specific muscular activity. Both effort generation and effort compensation assistive scenarios have been considered in the experimental evaluation, that has been

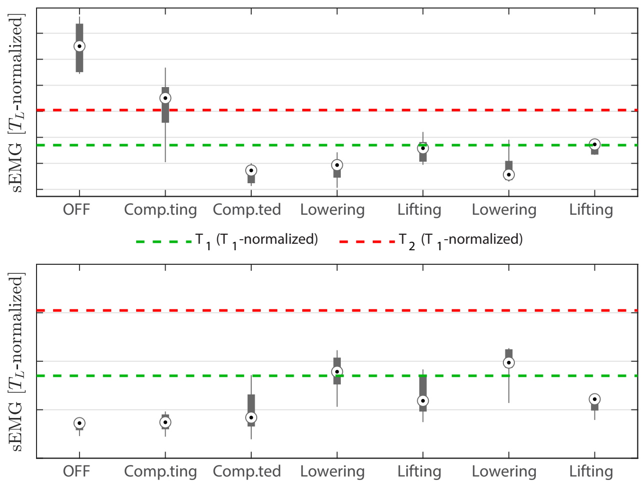


Fig. 12. Lowering-lifting experiments: mean sEMG over the 4 subjects (boxplot) for the task zones of Fig. 11.

carried out both with wearable and grounded assistive devices. The experimental results prove the feasibility of the proposed approach. Future studies will involve a complete statistical analysis on the performance of the presented method and extensionded to the participation of impaired subjects. Moreover the control strategy will be expanded to take into consideration also muscles co-contractions more effectively.

References

1. Pons, J.L.: Wearable robots: biomechatronic exoskeletons. John Wiley & Sons (2008)
2. Lo, H.S., Xie, S.Q.: Exoskeleton robots for upper-limb rehabilitation: State of the art and future prospects. Medical engineering & physics 34(3), 261–268 (2012)
3. Ada, L., Dorsch, S., Canning, C.G.: Strengthening interventions increase strength and improve activity after stroke: a systematic review. Australian Journal of Physiotherapy 52(4), 241–248 (2006)
4. Ditmyer, M.M., Topp, R., Pifer, M.: Prehabilitation in preparation for orthopaedic surgery. Orthopaedic Nursing 21(5), 43–54 (2002)
5. Iruthayarajah, J., Mirkowski, M., Reg, M.M.O., Foley, N., Iliescu, A., Caughlin, S., Harris, J., Dukelow, S., Chae, J., Knutson, J., et al.: Upper extremity motor rehabilitation interventions (2019)
6. Donatelli, R.A., Wooden, M.J.: Orthopaedic Physical Therapy-E-Book. Elsevier health sciences (2009)
7. Bowker, L., Price, J., Smith, S.: Oxford handbook of geriatric medicine. OUP Oxford (2012)

8. Yang, C., Zhang, J., Chen, Y., Dong, Y., Zhang, Y.: A review of exoskeleton-type systems and their key technologies. Proceedings of the Institution of Mechanical Engineers, Part C: Journal of Mechanical Engineering Science 222(8), 1599–1612 (2008)
9. Farina, D., Jiang, N., Rehbaum, H., Holobar, A., Graimann, B., Dietl, H., Aszmann, O.C.: The extraction of neural information from the surface emg for the control of upper-limb prostheses: emerging avenues and challenges. IEEE Transactions on Neural Systems and Rehabilitation Engineering 22(4), 797–809 (2014)
10. Song, R., Tong, K.Y.: Using recurrent artificial neural network model to estimate voluntary elbow torque in dynamic situations. Medical and Biological Engineering and Computing 43(4), 473–480 (2005)
11. Wang, S., Wang, L., Meijneke, C., Van Asseldonk, E., Hoellinger, T., Cheron, G., Ivanenko, Y., La Scaleia, V., Sylos-Labini, F., Molinari, M., et al.: Design and control of the mindwalker exoskeleton. IEEE transactions on neural systems and rehabilitation engineering 23(2), 277–286 (2014)
12. Pehlivan, A.U., Losey, D.P., O'Malley, M.K.: Minimal assist-as-needed controller for upper limb robotic rehabilitation. IEEE Transactions on Robotics 32(1), 113–124 (2015)
13. Lenzi, T., De Rossi, S.M., Vitiello, N., Carrozza, M.C.: Proportional EMG control for upper-limb powered exoskeletons. Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS pp. 628–631 (2011)
14. Peternel, L., Noda, T., Petrič, T., Ude, A., Morimoto, J., Babič, J.: Adaptive control of exoskeleton robots for periodic assistive behaviours based on emg feedback minimisation. PloS one 11(2), e0148942 (2016)
15. Mghames, S., Santina, C.D., Garabini, M., Bicchi, A.: A neuromuscular-model based control strategy to minimize muscle effort in assistive exoskeletons. IEEE International Conference on Rehabilitation Robotics 2019-June, 963–970 (2019)
16. Emika, F.: Panda. Tech. rep., Accessed 2017-10-06.[Online]. Available: https://www. franka. de (2018)
17. Palli, G., Natale, C., May, C., Melchiorri, C., Wurtz, T.: Modeling and control of the twisted string actuation system. IEEE/ASME Transactions on Mechatronics 18(2), 664–673 (2013)
18. Hosseini, M., Meattini, R., San-Millan, A., Palli, G., Melchiorri, C., Paik, J.: A semg-driven soft exosuit based on twisted string actuators for elbow assistive applications. IEEE Robotics and Automation Letters 5(3), 4094–4101 (2020)
19. Hosseini, M., Meattini, R., Palli, G., Melchiorri, C.: A wearable robotic device based on twisted string actuation for rehabilitation and assistive applications. Journal of Robotics 2017 (2017)
20. Hosseini, M., Meattini, R., Palli, G., Melchiorri, C.: Development of semg-driven assistive devices based on twisted string actuation. In: 2017 3rd International Conference on Control, Automation and Robotics (ICCAR). pp. 115–120. IEEE (2017)