Hybrid Cooperative Control of Functional Electrical Stimulation and Robot Assistance for Upper Extremity Rehabilitation

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Abstract*—*Objective: **Hybrid systems that integrate Functional Electrical Stimulation (FES) and robotic assistance have been proposed in neurorehabilitation to enhance therapeutic benefits. This study focuses on designing a cooperative controller capable of distributing the required torque for movement between robotic actuation and FES, thereby eliminating the need for time-consuming calibration procedures. *Methods:* The control schema comprises three main blocks: a motion generation block that defines the desired trajectory, a motor control block including both a weight compensation feedforward and a feedback impedance controller, and an FES control block, based on trial-by-trial Iterative Learning Control (ILC), that adjusts the stimulation intensity according to a predefined stimulation waveform. The feedforward motor assistance can be dynamically regulated using an allocation factor. Experiments involving 12 healthy volunteers were conducted using a one-degree-of-freedom elbow testbed. *Results:* The experimental results showcased the successful integration of Functional Electrical Stimulation (FES) with robotic actuation, ensuring precise trajectory tracking with a Root Mean Square Error (RMSE) below 7°. Notably, allocating more torque to FES led to a 51% reduction in motor torque. In conditions where FES operated alone, there was poorer tracking performance with an RMSE of 24° and an early onset of muscle fatigue, as evidenced by a reduced number of achieved repetitions. Furthermore, the hybrid approach enabled 100 fatigue-free elbow flexion repetitions, underscoring the effectiveness of cooperative FES-motor control in extending the benefits of FES-induced exercises. *Significance:* This study proposes a flexible approach which can be extended to a multi-degree-of-freedom hybrid system. Furthermore, it underscores the significance of employing a straightforward and adaptable methodology with a rapid calibration procedure, making it easily transferable to clinical applications.**

***Index Terms*—Functional Electrical Stimulation (FES), hybrid systems, exoskeleton, rehabilitation.**

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I. Introduction

According to the World Health Organization, 250 million people suffered from neurological disorders in 2019, estimated at 50 million years lived with disability (YLDs) [1]. Although recovery can partially occur spontaneously, rehabilitation is required to stimulate a structural remodeling of the central nervous system.

Functional Electrical Stimulation (FES) can enhance motor relearning by enhancing proprioceptive feedback and promoting brain plasticity [2], [3]. Still, FES-induced movements are hard to control due to the non-linear and time-independent nature of FES-induced muscle contractions and the early onset of muscle fatigue [4]. Conversely, robotic devices can promote high-intensive repetitive training with a high trajectory accuracy, but available devices are bulky, heavy, and require high power [5]. Recently, hybrid rehabilitation systems including a combined action of FES and motorized robots have been proposed to take full advantage of both concepts [6]. As compared to FES alone, hybrid systems offer supplementary torque support, ensuring precise movements while preventing early muscle fatigue. Simultaneously, the integration of FES within wearable robotics can reduce power requirements, while preserving FES benefits [7]. The reduction of power requirements may promote the development of lighter, more portable, and less expensive wearable robotic devices, which could be exploited for home-based care of chronic patients. However, to reach this aim, FES and robotic actuators should be applied at the same joint in a *cooperative* manner.

While some attempts of hybrid *cooperative* systems for lower limb restoration have been proposed [6], [8], little effort has been devoted so far to upper limb movements due to the complexity of arm gestures [9]. Indeed, gait rehabilitation permits the exploitation of cyclical movements, involving few degrees of freedom. As an example, Ha et al. [10] incorporated a cycle-to-cycle adaptive FES control, coupled with a position motor controller, to determine the stimulation timing for the quadriceps during walking. This involves maintaining a constant stimulation level while adjusting the timing of stimulation using information from a finite-state machine and hip joint torques from preceding steps. Similarly, Ren et al. [11] introduced a synergistic control framework, merging a model-based feed-forward FES controller with a compliant exoskeleton to support the desired knee trajectory. Alternatively, del-Ama and colleagues [12], [13] proposed an innovative hybrid controller for overground hybrid walking. This controller actively addresses FES-induced muscle fatigue through a combination of an iterative learning controller (ILC) and a Proportional-Integral-Derivative (PID). The control scheme integrates an impedance controller for the exoskeleton, employing an assist-as-needed cooperative strategy.

In the case of hybrid robotic devices for the upper limbs, FES and robotic actuators are usually combined to satisfy separate functions [9]. Passive robotic exoskeletons were combined with FES to concurrently obtain arm angular measurements and support for the arm weight to delay the onset of FES-induced muscle fatigue [14], [15]. Alternatively, robotic devices were used to assist a set of arm sub-movements during reaching tasks. At the same time, FES was only responsible for hand grasping [16] or for the assistance of other degrees of freedom [14], [17]. In other studies, robotic devices were controlled with some variants of position control with FES on top. For example, Wolf et al. proposed a control scheme composed of an empirical feedforward FES controller superimposed on a position feedback motor controller and tested it with the MAHI exoskeleton [18]. In [19], the authors combined RUPERT, a 5 degrees-of-freedom pneumatic exoskeleton, with FES to support grasp/release functions and elbow flexion: elbow movements were supported both by FES and the robot, but the control was not shared, since the pneumatic actuator worked unidirectional and in the opposite direction to FES.

In the latest years, some first examples of *cooperative* control have been proposed for single-joint upper limb hybrid systems. In 2021, Bardi and colleagues [20] proposed a *cooperative* control for elbow flexion that combines a closed-loop torque-based FES control with an impedance-based control strategy to adjust the motor assistance and tested its performance in simulation. In 2022, Burchielli et al. designed and experimentally tested on healthy subjects a hybrid *cooperative* controller involving FES and an arm exosuit to support elbow flexion, showing the possibility of delaying the onset of FES-induced muscle fatigue but they did not evaluate the possibility of reducing motor power consumption [21]. In 2023, Dunkelberger and colleagues presented a hybrid *cooperative* controller based on a model predictive control to combine an exoskeleton (MAHI) and an FES system on a single degree of freedom (either elbow or wrist flexion/extension) [22]. This *cooperative* controller was tested on healthy subjects in the challenging task of tracking time-varying trajectories and showed a reduction of motor torque of about 50% with only a small difference in tracking accuracy compared to the exoskeleton alone. However, this solution required a long-lasting calibration procedure, repeated for every training session, making it not ready for clinical testing. Furthermore, the authors did not investigate the effects of FES-induced muscle fatigue.

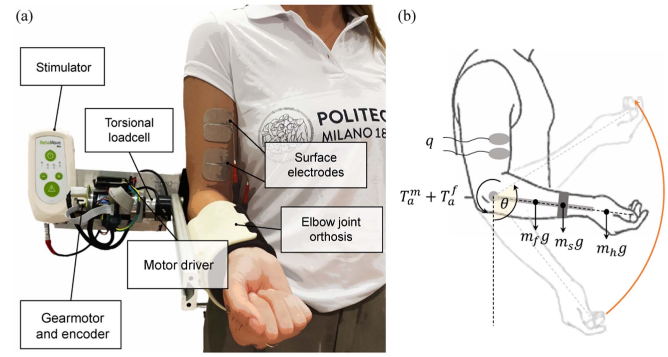

Fig. 1. (a) Hybrid testbed integrating an actuator at the elbow joint and an FES stimulator applied to the *biceps* muscles. (b) Simplified dynamic model of the hybrid system considering planar elbow flexion/extension movements.

In the existing literature, the predominant approach involves using FES to regulate interaction forces between the human limb and the hybrid system. Typically, these methods require the development of a biomechanical FES model for mapping FES-generated torque across the range of motion of the orthosis [11], [22], [23]. However, due to non-linearity, time variance, and uncertainties in FES-torque mapping, this often involves time-consuming calibration procedures. Alternatively, certain approaches necessitate the online measurement of FES-generated interaction torques, posing a challenge in distinguishing them from volitional forces and motor-generated torques [12]. In our work, we proposed a one-degree-of-freedom hybrid *cooperative* controller that relies solely on task-tracking performances, eliminating the need for long-lasting individual mapping procedures and online estimation of FES-induced interaction torques. This controller was experimentally tested on healthy participants to verify the hypothesis that combining FES with a motorized system on one side reduces motor power consumption, preserving high accuracy in trajectory tracking, and on the other side delays the onset of FES-induced muscle fatigue, thus prolonging the benefits of FES.

II. Methods

A. Hybrid System

We developed a single degree of freedom (DOF) elbow hybrid testbed consisting of two main components: a custom-designed mechanical structure to support elbow flexion/extension movements (previously described in [24]), and a programmable neuro-muscular electrical stimulator (Fig. 1(a)), specifically programmed to induce artificial muscle contractions of the *biceps brachii*. We used a 4-channels battery-powered current-controlled electrical stimulator (Rehamove 3, Hasomed GmbH). The stimulator can be programmed in real-time to deliver custom stimulation trains by tuning stimulation frequency, pulse width (PW), and current amplitude (I).

The elbow testbed embodies an aluminum link, which connects to the human arm at the forearm through an ergonomic cuff, as shown in Fig. 1(a). The system is actuated by a brushless electric motor (EC-45 flat, Maxon Motor) coupled with a gearbox (GP-42-C, Maxon Motor with a 156:1 transmission ratio). The

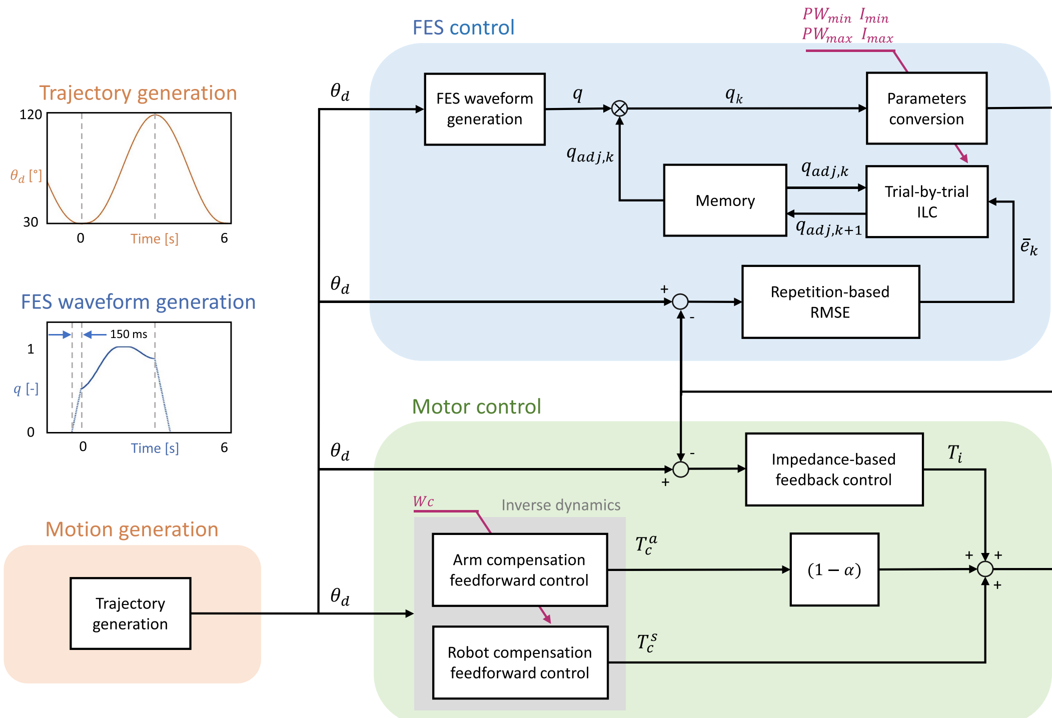

Fig. 2. Block diagram illustrating the cooperative control scheme of the hybrid system. The scheme is structured into three main components: i) Motion generation, ii) FES control, and iii) Motor control.

maximum generated torque is 15 Nm, with a maximum velocity of 3 rd/s. The system also integrates a loadcell-based torque sensor (TRT-200, Transducer Techniques) and an incremental encoder (MILE, Maxon Motor). The torque sensor is linked to the actuator output shaft, enabling the measurement of the output torque acting on the elbow joint. The actuation unit rigidly transmits the generated torque to the orthosis, facilitating elbow flexion/extension movements.

B. Hybrid Cooperative Control

The design of the proposed *cooperative* controller is based on a straightforward approach that depends solely on the average tracking error at the task's conclusion. Our primary goal is not minimizing interaction forces but achieving the overarching objective of completing a reaching task and performing a high number of repetitions without inducing muscular fatigue. The control scheme, depicted in Fig. 2 consists of three blocks: 1) Motion generation, 2) Motor control, and 3) FES control. The hybrid control logic is implemented in a real-time Linux-based machine at a frequency of 1 KHz. Two routines, one for the motor control and one for the FES control, run in parallel and communicate through a shared memory.

1) Motion Generation Block: The motion generation block generates trajectories for elbow cyclical movements in the sagittal plane. The desired elbow angular trajectory θ_d was computed through a minimum-jerk function, which is a good approximation of physiological reaching movements, by employing the $\beta-$density function of (1) and (2), similar to [24].

$$\theta_d(t) = \delta_0 + \delta_1(t-\delta_2)^{\delta_3}(\delta_4 - t)^{\delta_5}, \delta_2 \le t \le \delta_4 \tag{1}$$

$$\delta_1 = \frac{\Delta\theta}{\frac{\delta_4-\delta_2}{2}^{(\delta_3+\delta_5)}} \tag{2}$$

where $t \in [t_0, \Delta T]$ is the task time, ΔT is the task duration, $\Delta\theta$ is the task amplitude, and δ_n are motion parameters. In particular, δ_0 represents the position offset at t_0, δ_1 is related to movement amplitude $\Delta\theta$ by means of (2), δ_2 and δ_4 are the start and the stop time, and δ_3 and δ_5 represent the symmetry orders for the raising and falling slopes. In this work, the elbow flexion/extension range of motion (ROM) was limited between 30° and 120°, since most daily life activities can be performed within this range [25]. The δ_n parameters were computed to obtain a symmetric 5th order flexion/extension movement lasting 6 s with an amplitude of 90°.

2) Motor Control Block: The motor controller, which was previously presented in [24], relies on an impedance-based control strategy. The controller comprises: i) a feedforward controller, derived from the inverse dynamics block, and iii) a feedback controller that corrects for trajectory tracking deviations. The motor feedforward controller action compensates for the dynamics of the arm and the mechanical system, aimed at minimizing the interaction forces with the human arm and maximizing the perceived transparency of the robot, while the feedback controller implements a virtual mechanical impedance that provides a torque field to correct tracking errors.

In detail, the motor control block computes the inverse dynamics model of the elbow human-robot hybrid system to estimate the desired torque of the feedforward controller. Given the low acceleration of the movements, we omitted the compensation for the inertial component of the dynamic model. Moreover, the effect due to the passive dynamics of the musculoskeletal system (passive elements of muscles, tendons, and ligaments), was considered negligible. The simplified feedforward controller is divided into two parts, respectively for the compensation of the weight of the mechanical system T_c^s and of the participant's arm T_c^a. The two terms are computed as in (3) and (4):

$$T_c^s = m_s g l_s \sin\theta_d \tag{3}$$

$$T_c^a = W_c(m_f g l_f + m_h g l_h)\sin\theta_d \tag{4}$$

where θ_d is the desired angular trajectory, g is gravitational acceleration, m_s, m_f, m_h are the masses of the mechanical system, forearm, and hand, respectively, and l_s, l_f, l_h are the respective centers of mass. W_c is a weighting factor that is regulated to adjust the arm weight compensation. This parameter was set experimentally during an initial calibration procedure to minimize the motor torque while assuring a good tracking performance.

To balance the desired torque between the actuator and FES, the arm weight compensation term T_c^a is adjusted through a factor $1-\alpha$, with $\alpha \in [0,1]$. Here, α represents the percentage of torque assigned to FES. A higher α implies a reduced feedforward component required from the motor. Conversely, lower values of α indicate a greater demand for motor assistance.

The feedback controller is implemented through an impedance control strategy that compares the reference position θ_d with the measured position θ_a, correcting for tracking errors via a virtual first-order spring-damper system, and compensating for the friction of the gear motor. The impedance controller is computed as in (5).

$$T_i = K_d(\theta_d - \theta_a) - D_d\dot{\theta}_d + \hat{f}_l\dot{\theta}_d \tag{5}$$

where K_d is the desired virtual stiffness, D_d is the desired virtual damping, and $\hat{f}_l$ is the estimated viscous friction coefficient. K_d and D_d were experimentally tuned and then left unchanged for all participants. To enhance the role of FES, we used a low-stiffness virtual impedance, which allows great deviations from the reference position and promotes compliant behavior of the elbow testbed. Specifically, we used $K_d = 5.0$ Nm/rad and $D_d = 0.5$ Nms/rad.

Therefore, the total desired torque that is commanded to the torque-controlled actuator is computed as in (6):

$$T_d^m = T_c^s + (1-\alpha)T_c^a + T_i \tag{6}$$

Finally, the actuation unit is responsible for delivering the desired torque T_d^m at the joint of the elbow testbed. To do so, we employed a low-level closed-loop torque control strategy. We used a PID control scheme to follow the desired torque based on the actual torque feedback T_a^m measured through the torsional loadcell [24].

3) FES Control Block: The FES control of the hybrid testbed is based on a feedforward controller that computes the stimulation parameters (i.e., current and pulse width) needed to deliver the desired torque assigned to FES. Specifically, the FES controller includes two separate parts: i) a stimulation waveform generator, and ii) a trial-by-trial Iterative Learning Controller (ILC) that regulates the stimulation amplitude according to the kth previous iteration.

First, we shaped the stimulation waveform for a nominal elbow flexion movement. We relied on the simplified assumption of a linear relationship between stimulation charge and torque. We computed the stimulation waveform in terms of normalized charge $q \in [0,1]$ following (7).

$$q(t) = \sin\theta_d(t) \tag{7}$$

where θ_d is the β-function trajectory of (1). Additionally, to compensate for the electromechanical delay of FES-induced muscle contractions, we anticipated the onset of the stimulation waveform of about 150 ms [26], as shown in the top left of Fig. 2. Furthermore, to avoid abrupt changes in the delivered charge, ramps were added before and after the stimulation waveform. Second, the stimulation waveform is multiplied by a scaling factor $q_{adj,k}$, which is dynamically adjusted by the trial-by-trial ILC, as in (8).

$$q_k = q_{adj,k}\,q \tag{8}$$

The scaling factor $q_{adj,k} \in [0,1]$ is iteratively adjusted, following an ILC update law, based on the angular position Root Mean Squared Error (RMSE) $\bar{e}_k$ of the previous repetition to keep the position error between defined boundaries as in (9).

$$q_{adj,k+1} = q_{adj,k} + G\bar{e}_k \tag{9}$$

where $q_{adj,k+1}$ is the updated scaling factor, $\bar{e}_k$ is the tracking error during the kth repetition and $G = 0.1$ is a design parameter, which was found through a trial-and-error procedure. The charge q_k is then turned into stimulation parameters following (10) and (11), for the stimulation pulse width and current amplitude, respectively.

$$PW_k = PW_{\min} + \sqrt{q_k}(PW_{\max} - PW_{\min}) \tag{10}$$

$$I_k = I_{\min} + \sqrt{q_k}(I_{\max} - I_{\min}) \tag{11}$$

The minimal pulse width ($PW_{\min}$) and amplitude ($I_{\min}$) were determined as the ones producing a torque of about 0.5 Nm. Instead, the maximal values of stimulation pulse width and amplitude were set to the limits tolerated by the subject during an initial calibration procedure.

III. Experimental Validation

The aim of the experimental validation was twofold. On the one hand, we aimed to assess the performance of the *cooperative* controller at different allocation levels to verify that increasing FES-induced torque can reduce the motor power consumption while preserving a good tracking of the movement. On the other hand, we aimed to demonstrate that the *cooperative* controller can be employed to prolong training time by delaying the occurrence of FES-induced muscle fatigue. To reach these objectives, two distinct experiments were designed. Both experiments consisted of repetitions of elbow flexion/extension

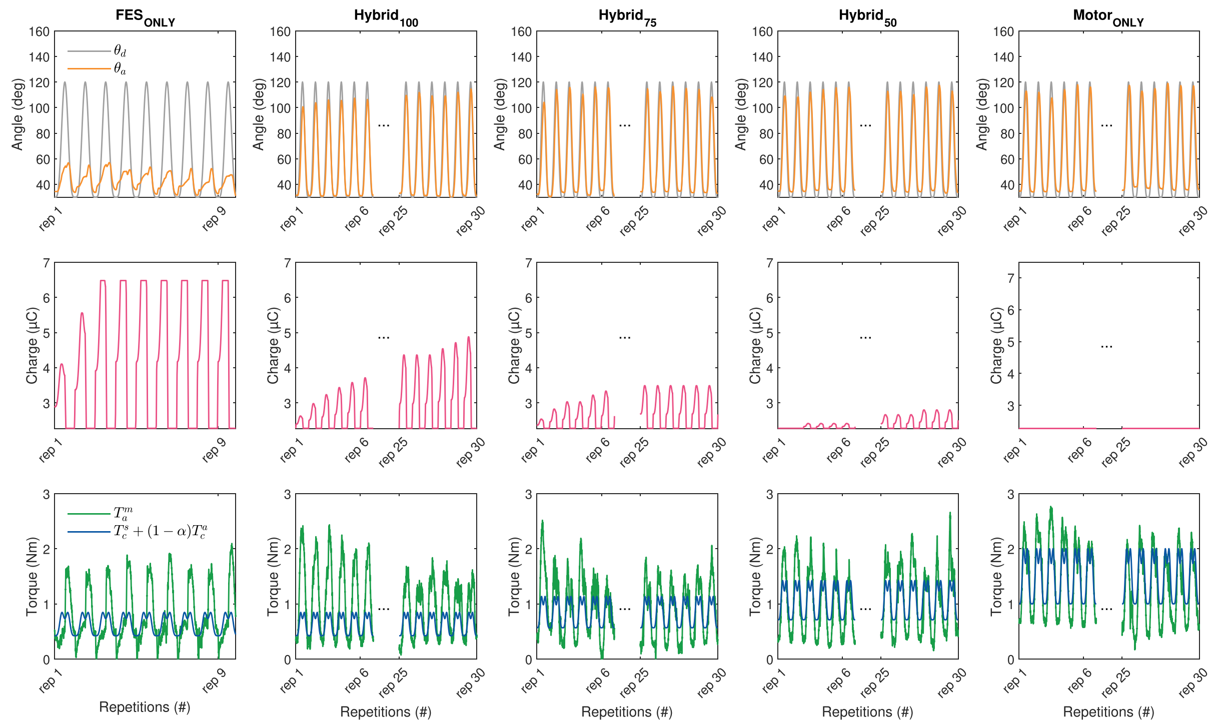

Fig. 3. Results of Experiment A for one representative subject (S05). Columns display the five tested conditions.

movements, but the *cooperative* controller was applied only during the anti-gravitational elbow flexion. Only the *biceps* was stimulated by placing stimulation electrodes (Pals from Axelgaard Manufacturing Co. Ltd.) over the muscle belly, as shown in Fig. 1, and all data were collected from the right arm. Experiments were conducted on healthy volunteers with no prior record of physical, neurological, or cognitive impairments, who were asked to behave passively during the whole experimental procedure to exclude the effect of volitional contractions.

The study was approved by the Ethical Committee of Politecnico di Milano (Opinion n. 13/2021) and participants signed an informed consent before the experiments.

An initial calibration, lasting less than 5 minutes, was performed on each participant. This procedure was aimed to identify the weighting factor W_c and the maximum values of current amplitude and pulse width. The stimulation frequency was kept constant at 40 Hz for all subjects. At first, FES was switched off, α was set to zero as well as the weighting factor W_c. The impedance-based motor control was run and W_c was increased trial-by-trial to obtain a trajectory tracking RMSE below 5° throughout the flexion phase. Indeed, the purpose of the weighting factor was not to perfectly compensate for the arm weight, but to minimize the motor torque while preserving a good tracking performance. Afterwards, the motor control was switched off and an increasing ramp of stimulation charge was delivered to the biceps muscle to identify the maximum tolerated values.

A. Experiment A: Short-Term Evaluation

Elbow flexion movements were performed under five different conditions and the trial was stopped after 30 repetitions or, eventually, 6 repetitions after the saturation of stimulation parameters occurred. Each repetition had a duration of 6 seconds, and the movement involved a 90° elbow flexion/extension. First, we delineated three conditions based on the FES-robot allocation factor: Hybrid_{50}, Hybrid_{75}, and Hybrid_{100}, corresponding to α values of 0.5, 0.75, and 1, respectively. An additional condition, FESONLY, was implemented, where the movement was exclusively controlled by FES. In this mode, the mechanical structure operated in transparent control mode without activating any impedance-based correction. The last condition, $\text{Motor}_{\text{ONLY}}$, involved movement driven solely by the motor, setting α to 0, and providing no FES assistance. To prevent muscular fatigue biases, ten-minute breaks were added between trials, and the order of appearance of the conditions was randomized.

B. Experiment B: Long-Lasting Evaluation

The second experiment aimed to evaluate the capacity of the *cooperative* controller to postpone the onset of muscular fatigue. To achieve this objective, we conducted 100 repetitions of elbow flexion/extension movements under the Hybrid_{100} condition with an additional 0.5 kg payload. While direct measurement of the onset of muscle fatigue was not performed, our hypothesis suggests that the *cooperative* controller enables task completion with more repetitions compared to using FES alone, without the need for escalating stimulation parameters.

IV. Data Analysis and Statistics

All metrics were computed only during the elbow flexion phase when the *cooperative* controller was active. To evaluate the tracking performance, the trajectory tracking Root Mean Squared Error (RMSE) between the desired and the measured

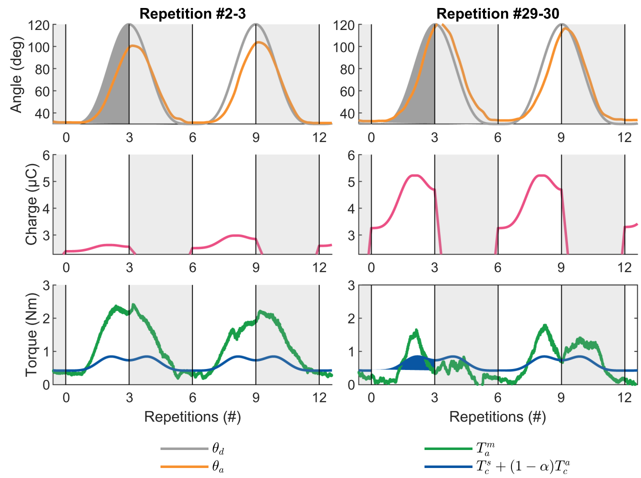

Fig. 4. Results of Experiment A for one representative subject (S05) for first and last repetitions during Hybrid$_{100}$ condition. White and gray areas represent elbow flexion and extension phases, respectively.

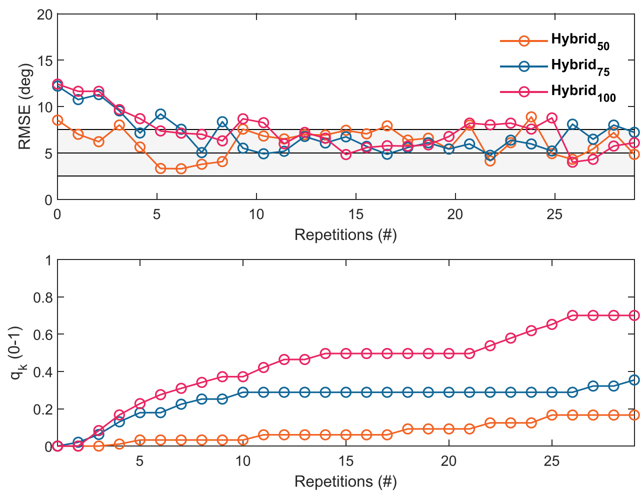

Fig. 5. Results illustrating the trajectory tracking RMSE and the adaptation of the stimulation charge q_k during Experiment A for one representative subject (S05) testing the hybrid conditions.

angular position was calculated. To compare the dynamic behavior of the system across different conditions, the measured interaction torque and the energy consumption were evaluated. The energy consumption was computed as the time integral of the signed mechanical power generated by the actuator, where positive power is obtained when the motor generates assistive torques, and negative power is obtained when the motor resists FES-generated torques. Finally, the time integral of the delivered stimulation charge was computed.

To evaluate whether there were significant differences among conditions, metrics were averaged across repetitions for each subject and condition. Subsequently, a pairwise analysis using a generalized linear mixed model and applying Bonferroni correction for multiple comparisons was performed for each metric.

Data analysis was performed in MATLAB (R2021b version), while statistical analysis was performed in IBM SPSS Software.

V. Results

Twelve healthy subjects participated in the study (8 females, 4 males, age 24 $\pm$ 2 years, height 171 $\pm$ 12 cm, body weight 60 $\pm$ 13 kg).

Overall, the weighting factor (W_c) was 0.15 $\pm$ 0.04. Regarding stimulation parameters, minimum values of current and pulse width were kept constant for all subjects at 9 mA and 250 μs, while maximum values found during the calibration were on average 16 $\pm$ 13 mA and 362 $\pm$ 44 μs, respectively.

A. Experiment A

Figs. 3 and 4 show the results of the short-term experiment for one representative subject (S05). In detail, Fig. 3 compares the results across all conditions. The first row illustrates desired (θd) and actual (θ_a) elbow angular trajectories. The second row depicts the delivered stimulation charge, and the third row shows the actual measured interaction torque (T_a^m) in comparison to the total commanded feedforward torque ($T_c^s(1-\alpha)T_c^a$). Notably, in the hybrid conditions, the subject was able to perform all the required 30 repetitions, and the results of the first and last 5 repetitions are displayed. For the FES$_{\text{ONLY}}$ condition, the test was stopped 6 repetitions after reaching the saturation level, thus only 9 repetitions were performed. In the FES$_{\text{ONLY}}$ condition, there was poor tracking of the desired trajectory compared to the other conditions, despite the high delivered charge. Conversely, in the Motor$_{\text{ONLY}}$ condition, as expected, the tracking performance was optimal. This condition served as a reference to quantify the minimal tracking error and the total motor torque required to complete the intended movement.

In contrast, Fig. 4 illustrates input and output data for the initial and final repetitions within the Hybrid$_{100}$ condition. It is evident from the figure that the FES controller dynamically adjusted the stimulation charge to mitigate tracking errors. Comparing initial repetitions with the final ones reveals an enhancement in tracking performance, correlated with an increase in the delivered charge. Finally, it is noteworthy that in the last repetitions, the measured interaction torque decreased in comparison to the initial repetitions, demonstrating the impact of increasing the delivered charge during FES. Across all hybrid conditions, integrating the motor assistance decreased the necessary stimulation charge to complete the task as compared to the FES$_{\text{ONLY}}$ condition. Notably, in the hybrid conditions, unlike the FESONLY condition, the charge remains below saturation levels.

In Fig. 5, the trial-by-trial ILC convergence performance is presented for the hybrid conditions. The *cooperative* control system achieved convergence, on average, within 5 repetitions. Fig. 6 shows the mean motor power and the mean motor energy consumed during the flexion phase for one representative subject during all trial conditions. The higher power was exerted in the Motor$_{\text{ONLY}}$ condition, as expected, while for higher allocation factors, the mean power exerted decreased.

Fig. 7 shows the results obtained by averaging the metrics for all subjects. In the three hybrid conditions, the trajectory tracking RMSE was significantly higher than the Motor$_{\text{ONLY}}$ condition

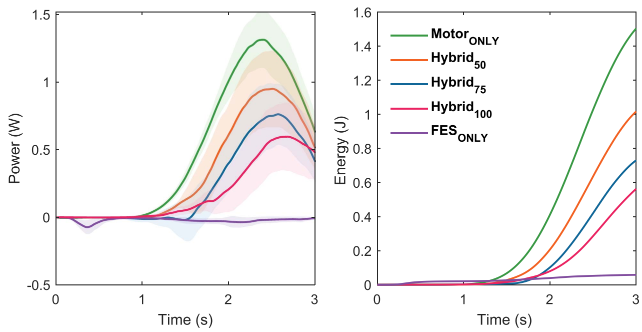

Fig. 6. Results presenting power and energy motor consumption during the elbow flexion phase in Experiment A for one representative subject (S05).

($p < 0.001$), but remained $<7°$. Instead, the FES_{ONLY} conditions showed a strongly worsened trajectory tracking, with a median RMSE of 24°. In terms of motor torque, the hybrid conditions exhibited a significant reduction of 30% in Hybrid_{50}, 41% in Hybrid_{75}, and 51% in Hybrid_{100} compared to the $\text{Motor}_{\text{ONLY}}$ condition. Accordingly, the total energy consumption decreased as the allocation factor increased. On the contrary, the delivered stimulation charge increased significantly ($p < 0.001$) with the increase of the allocation factor, with the highest value achieved in the FES_{ONLY} condition.

Overall, not all subjects weren't able to complete 30 repetitions in FES_{ONLY} condition (mean number of completed repetitions was 21 ± 9), indicating an early onset of muscle fatigue in this condition.

B. Experiment B

Fig. 8 shows the results of Experiment B, which aimed to evaluate the onset of muscle fatigue in the Hybrid_{100} condition. For visualization purposes, the 100 repetitions were divided into 10 blocks of 10 repetitions each. Significant differences ($p < 0.001$) were found between the first 10 repetitions and the other repetitions in terms of trajectory tracking error, measured interaction torque, total charge delivered to the *biceps*, and motor energy consumption. Only the total charge showed a significant difference ($p < 0.001$) also between the second block of repetitions (rep 11–20) compared to all the others. These results showed that 10 repetitions were enough to reach the convergence of the *cooperative* system and that there was not a worsening of the performance over time nor an increase of the required stimulation charge, suggesting that muscle fatigue did not occur over 100 repetitions of elbow flexion with a weight of 0.5 kg.

VI. Discussion

The overarching objective of this work was to validate a *cooperative* control strategy for a hybrid system that combines FES with a powered elbow orthosis. This approach is expected to improve rehabilitation outcomes as compared to FES or robots alone. The proposed control scheme was tested on 12 healthy participants during elbow flexion movements, showing the feasibility of the proposed hybrid approach, which i) adjusted the stimulation current through a trial-by-trial ILC, ii) reduced interaction torque and actuation power consumption proportional to the amount of torque allocated to FES, and iii) achieved better trajectory tracking as compared to FES alone.

In the hybrid condition with the highest allocation factor (α=1), a substantial motor torque reduction of 51% was obtained. However, this reduction was accompanied by a 25% deterioration in tracking performance compared to the condition fully supported by the motor. Despite this, overall satisfactory tracking performance was maintained, with an RMSE below 7°. This performance was strongly better than the condition in which the movement was only supported by FES, in which an RMSE of 24° was achieved. This suggests that, with the proposed *cooperative* control system, the FES control effectively initiates coarse movements, with the motor torque performing small corrections to ensure trajectory tracking. Comparable outcomes were observed in a recent study [22], demonstrating a notable 49% reduction in the sum of squared torque during elbow flexion/extension movements. In their study, the authors developed a hybrid FES-exoskeleton controller capable of accurately tracking time-varying trajectories with minimal error. However, the implementation requires a long-lasting calibration procedure that must be repeated for each training session to fine-tune the predictive model embedded in the control system. For clinical applications, instead, there is a demand for simpler solutions that utilize rapid and automated calibration procedures.

Our findings underscore the feasibility of incorporating FES into wearable robots as an effective approach to mitigate power consumption [7], [9]. This, in turn, contributes to the reduction of weight and bulkiness in wearable robotic devices, all achieved without necessitating extensive and time-consuming calibration procedures. Notably, our *cooperative* control system requires only a quick calibration process, specifically designed to identify the maximally tolerated stimulation parameters.

Regarding our second hypothesis, we indirectly demonstrated that the proposed *cooperative* controller can be employed to postpone the onset of FES-induced muscle fatigue. Specifically, in Experiment A, we illustrated that when relying solely on FES to drive the entire motion (e.g., in the FES_{ONLY} condition), there was a substantial deterioration in tracking performance, coupled with an early onset of muscle fatigue. This was demonstrated by the reduced number of repetitions participants could perform under this condition, highlighting the challenge of controlling FES for inducing precise movements over numerous repetitions. In contrast, in Experiment B, the hybrid solution successfully supported 100 repetitions of elbow flexion (with additional load) without any indication of muscle fatigue. Indeed, the stimulation charge did not increase over time, and the trajectory tracking performance did not deteriorate. This outcome aligns with our initial hypothesis, confirming that our *cooperative* control strategy can be effectively utilized to prolong the benefits of FES-induced exercises and delay the onset of muscle fatigue.

However, the proposed hybrid solution presents some limitations. Firstly, replacing the cycling movement with non-periodic tasks is essential to fully capitalize on the benefits of the presented solution, especially in motor tasks resembling daily life

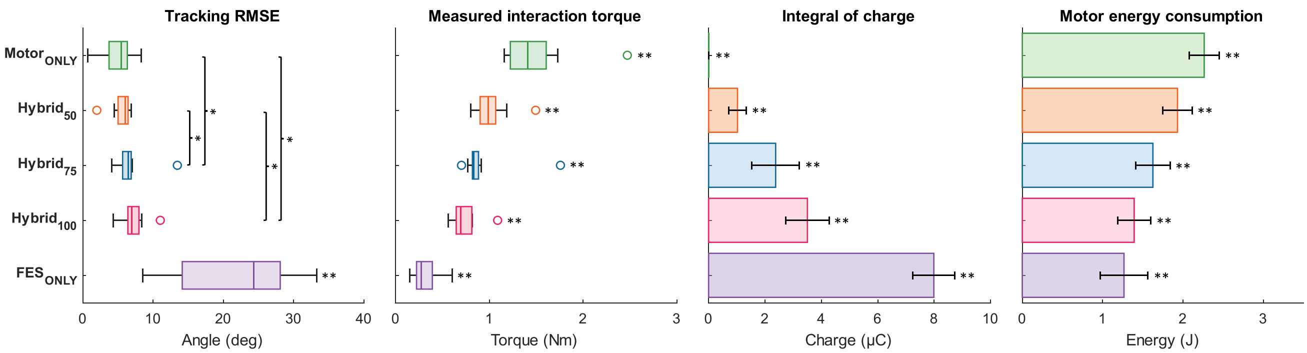

Fig. 7. Averaged results of Experiment A across the 12 subjects for each experimental condition. $*$ indicates a statistically significant difference between conditions; $**$ indicates a statistically significant difference between one condition and all the others.

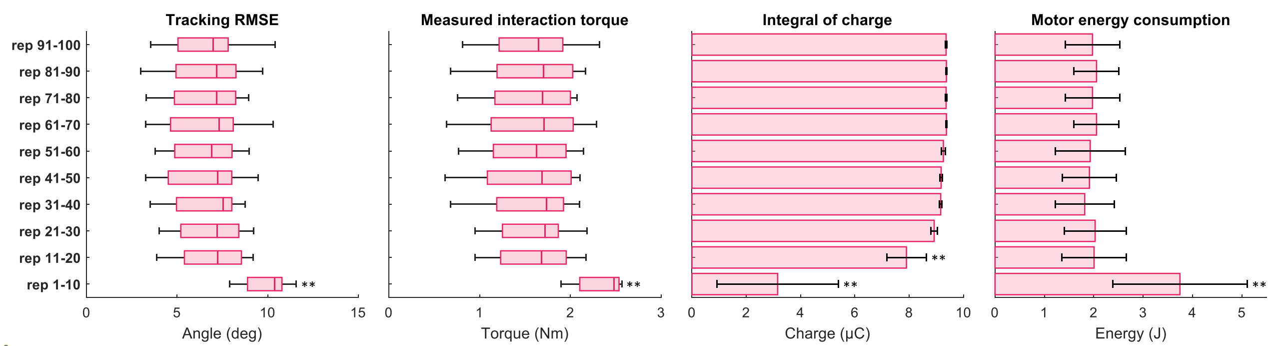

Fig. 8. Averaged results of Experiment B across the 12 subjects during 100 repetitions. The results are shown in blocks of 10 repetitions each. $**$ indicates a statistically significant difference between one repetition block and all the others.

activities. In such scenarios, the trial-by-trial ILC, responsible for amplitude tuning of the stimulation profile, should also consider the timing of stimulation by performing a point-by-point shaping of the charge waveform to accommodate the dynamics of the movement.

Secondly, we tested the proposed solution with fixed allocation factors, whereas future research should focus on developing methods to dynamically adjust the allocation factor α, which is the tunable gain between the two power sources (FES and robot assistance). This adjustment should be based on monitoring fatigue levels, allowing for an increase in motor assistance when the effectiveness of FES diminishes, following an *assisted-as-needed* approach.

Although we speculated that the proposed hybrid solution could delay the onset of muscle fatigue, evidenced by a high number of repetitions (n=100) without the need to increase the stimulation charge and without a decrease in tracking performance, this study lacks a direct measure of muscle fatigue. Further investigations are necessary to understand the complex relationship between stimulation, fatigue, and task performance.

While our study focuses on one degree of freedom (e.g. elbow joint), future work should extend this concept to multi-degree-of-freedom exoskeletons. This extension will introduce several challenges, such as addressing coordination issues among multiple joints, optimizing control strategies for diverse movement patterns, and ensuring seamless integration with the user's natural motion. Overcoming these challenges will require innovative approaches to adapt the hybrid control framework to the increased complexity of multi-joint exoskeletons.

Furthermore, the proposed control system was tested in a simplified condition where healthy volunteers were asked to not voluntarily participate in the task. However, the literature suggests that FES-induced cortical plasticity is enhanced when FES is synchronized with voluntary drive [3], [27]. Therefore, it is crucial to design control systems that distribute the required assistance among the subject's residual voluntary capability, FES, and the robotic system. We envision that the proposed *cooperative* control should be applicable in scenarios where participants actively engage in the task voluntarily. Indeed, we expect that the proposed controller will gradually decrease the delivered stimulation charge with increasing active participation. FES activation would occur only when participants face challenges in completing the task. This approach prioritizes voluntary effort, which is faster, with the robotic assistance employed to refine movements solely when the actual trajectory deviates from the desired one. However, further experiments are essential to validate this hypothesis. Finally, future tests should involve neurological patients, such as stroke survivors and individuals with spinal cord injuries, to demonstrate the practicality of the proposed approach with target users.

VII. Conclusion

In this paper, a *cooperative* control approach for a hybrid FES-motor system was presented. We intended to combine the therapeutical benefits associated with FES, with the reliability of the haptic feedback generated by the robotic device. Experimental results on healthy participants demonstrated that the proposed hybrid approach reduced the motor power consumption while preserving tracking accuracy as compared to full motor assistance, and delayed the occurrence of muscular fatigue as compared to FES alone, without the need for long-lasting calibration procedures. Therefore, we believe that the presented hybrid approach could be used as an effective solution for more integrated FES-hybrid rehabilitation solutions.

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