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Adaptive Cooperative Control for Hybrid FES-Robotic Upper Limb Devices: a Simulation Study*

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Abstract— **Robotic systems and Functional Electrical Stimulation (FES) are common technologies exploited in motor rehabilitation. However, they present some limits. To overcome the weaknesses of both approaches, hybrid cooperative devices have been developed, which combine the action of the robot and that of the electrically stimulated muscles on the same joint. In this work, we present a novel adaptive cooperative controller for the rehabilitation of the upper limb. The controller comprises an *allocator* - which breaks down the reference torque between the motor and the FES a-priori contributions based on muscle fatigue estimation - an FES closed-loop controller, and an impedance control loop on the motor to correct trajectory tracking errors. The controller was tested in simulation environment reproducing elbow flexion/extension movements. Results showed that the controller could reduce motor torque requirements with respect to the motor-only case, at the expense of trajectory tracking performance. Moreover, it could improve fatigue management with respect to the FES-only case. In conclusion, the proposed control strategy provides a good trade-off between motor torque consumption and trajectory tracking performance, while the *allocator* manages fatigue-related phenomena.**

Clinical relevance—**The use of allocation proves to be effective in both reducing motor torque and FES-induced muscle fatigue, and might be an effective solution for hybrid FES-robotic systems.**

I. INTRODUCTION

In the context of motor rehabilitation of subjects suffering from neuromuscular and neurodegenerative diseases, two important technology-aided approaches are rehabilitation robotics and Functional Electrical Stimulation (FES). Rehabilitation robotics brings advantages in terms of duration of the session, training intensity, repeatability, patient involvement and objective measurements [1], [2]. However, robots are bulky, heavy and expensive, thus preventing their adoption outside the clinical context. FES, instead, brings peripheral advantages in terms of muscular tone, vascular health, metabolic consumption, and prevention of osteoporosis. Moreover, it promotes functional reorganization and changes in the excitability of the cortex, favoring motor recovery [4]. The main drawbacks of using FES rely upon the difficulty of generating a precise movement and in the premature onset of muscular fatigue [5], which makes sessions short.

Recently, in order to enhance rehabilitation robotics and FES respective advantages, active hybrid cooperative systems have been developed both for the lower [6] and the upper limbs [7]. Hybrid cooperative devices combine the action of electrically stimulated muscles with that of robotic systems on the same joint. Preliminary results proved that this approach may bring significant advantages [6], [8]. On one side, the contribution of the FES-induced muscle contraction can lower the motor torque requirements, which results in the possibility to use smaller motors and to decrease energy consumption, improving the portability of the robot. On the other side, using a robotic device during an FES session allows both to share the effort and to correct deviations from the ideal trajectory, thus delaying muscular fatigue, performing more functional-oriented tasks, and prolonging the therapy session.

Several works focused on the development of active hybrid cooperative systems for the lower limb. To cite a few, Del Ama et al. in 2014 [10] proposed a way to estimate muscular fatigue from the interaction torque between the limb and the exoskeleton. The motors were included in a feedback impedance control to correct trajectory errors. Zhang et al. in 2017 [11] proposed an online parameter regulator which dynamically allocated the reference torque between motors and FES. Kirsch et al. in 2018 [12] proposed a Non-linear Model Predictive Control (NMPC) to optimally dynamically allocate FES and motor contributions in order to minimize the total control input. Regarding upper limb active hybrid cooperative control, few studies have been conducted so far. Tu et al. in 2017 [13] implemented a 5-DOF cooperative Iterative Leaning Controller (ILC) for repetitive reach-to-grasp tasks. Wolf et al. in 2017 [14] developed an empirical feedforward FES controller and a position feedback motor controller.

In this context, there are two connected aspects that require further research: fatigue management and torque allocation. While the first one has been deeply investigated in literature, the latter has been proposed through feedforward FES control [11], [12]. However, given the model dependency of this approach, it requires precise parameter estimation, which is time-consuming.

This work aims to design, develop and test in simulation a hybrid cooperative controller for the rehabilitation and assistance of the elbow flexion/extension movements. The architecture integrates the action of motor and FES, and an adaptive *allocator* divides the required torque between them.

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II. MATERIALS AND METHODS

A. Models

In order to test the control architecture, two models were implemented: the dynamic model of the coupled arm-robot system and the FES-torque model. The motor dynamics was neglected for sake of simplicity and thus, the motor was considered an ideal torque source.

The upper limb dynamic model was developed in OpenSim (SimTK) [15], starting from ”Arm26”, which includes 2-DOF (shoulder elevation and elbow flexion/extension). Then, the model was modified adding a simplified exoskeleton CAD model drawn in Solidworks (Dassault Systèmes SolidWorks Corporation) and an external actuator at the elbow joint. The external actuator was used to provide at the elbow joint the total torque computed as the sum of the motor torque and the FES-induced torque, as shown in Fig. 1. This model was used to compute both the inverse and the forward dynamics and, thanks to the OpenSim API, it was included in Matlab/Simulink (MathWorks, Inc. USA).

The second model simulated the biceps response to FES. In particular, the model takes as inputs a normalized pulse charge ($0 < q < 1$), and its integral throughout the exercise, to take into account fatigue effects, and returns the torque produced at the elbow joint. The FES-torque model was obtained by training a NARX (Non-linear AutoRegressive with Exogenous Input) neural network in Matlab. The dataset was obtained by stimulating the right biceps of a healthy subject in isometric conditions in sessions of 3-4 minutes in order to be able to include muscle fatigue. Experiments involving human subjects were approved by the ethical committee of Politecnico di Milano. Pulse Width (PW) and Pulse Amplitude (I) were computed and provided to the stimulator as follows:

$$I = I_{min} + \sqrt{q}(I_{max} - I_{min}) \tag{1}$$

$$PW = PW_{min} + \sqrt{q}(PW_{max} - PW_{min}) \tag{2}$$

Minimal and maximal values were defined as the values inducing a visible muscle contraction and maximally tolerated by the subject, respectively.

B. Control architecture

The cooperative control architecture, shown in Fig. 1, is composed of i) an inverse dynamics module, which computes the torque necessary to execute a movement, ii) an *allocator* that subdivides the reference torque between the motor feedforward contribution and the FES contribution, iii) an PI FES torque closed-loop controller, which determines the charge to be delivered to the muscle, and iv) an impedance loop that computes an additional motor torque to correct for trajectory errors. The FES and the motor torque sum up at the joint level and determine the dynamic behavior of the system.

Regarding the motor contribution, in addition to the feedforward torque term, τ_{rmotor}, a feedback term is added as follows:

$$\tau_{motor} = \tau_{imp} + \tau_{rmotor} \tag{3}$$

where τ_{imp} is determined by an explicit impedance control law, which consists of an external position loop that corrects trajectory errors, and an internal torque loop which guarantees compliance. The equation describing the impedance control law is:

$$\tau_{imp} = K_d(\theta_r - \theta) + D_d(\dot{\theta}_r - \dot{\theta}) \tag{4}$$

where K_d is the stiffness, D_d is the damping, θ_r and $\dot{\theta}_r$ describe the desired trajectory, while θ and $\dot{\theta}$ are the measured angle and angular velocity, respectively. The control parameters were empirically tuned so as to display an overall stable behavior and to display a trade-off between trajectory tracking and rendered compliance of the system.

At the inner level, a PI control on the torque signal was implemented to modulate FES intensity. The proportional and integral parameters were tuned by trial-and-error. In this case, the torque produced by the stimulated muscles was computed thanks to the NARX model. However, in real life applications, this type of control would require an estimation of the FES-induced torque. Indeed, in non-isometric conditions it is not possible to directly measure the torque produced by the muscles, but the impedance torque can be used as an estimate of the error of the FES controller, assuming a precise inverse dynamics model and a stiff impedance control, as proposed in [8].

C. Allocator

Once the inverse dynamics is computed, the *allocator* breaks down the reference torque between the motor feedforward contribution and the FES contribution as follows:

$$\tau_{rFES,i} = (1 - \alpha_i)\tau_{r,i} \tag{5}$$

$$\tau_{rmotor,i} = \alpha_i \tau_{r,i} \tag{6}$$

where $\tau_{r,i}$ is the reference torque, α_i is the allocation factor, $\tau_{rFES,i}$ is the torque allocated to FES and $\tau_{rmotor,i}$ is the torque allocated to the motor at the iteration i.

The torque percentage to be allocated to FES is updated by estimating muscular fatigue. When fatigue arises, the trajectory tracking performances will become poorer due to a reduction of FES-induced torque. Since the impedance controller is meant to correct trajectory by providing a torque proportional to the position and velocity errors, the impedance torque can be used as an indicator of muscular fatigue. The allocation factor was computed as follows:

$$\alpha_i = \frac{\overline{\tau_{imp}}}{\overline{\tau_{rFES}}} \tag{7}$$

where $\overline{\tau_{imp}}$ is the average of the torque determined by the impedance controller over the chosen window length at the iteration i, and $\overline{\tau_{rFES}}$ is the average of the torque allocated to FES over the chosen window length at the iteration i. The initial allocation factor was set $\alpha_0 = 0$.

Two different types of allocation were proposed and tested:

- Discrete allocation: the allocation factor was updated at the end of each movement repetition in order to adjust the required contribution according to the performance of the previous task.

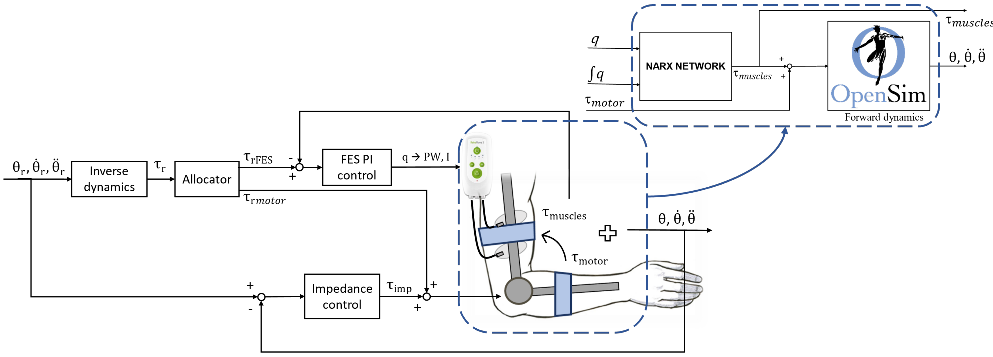

Fig. 1: Cooperative control architecture composed by: i) an inverse dynamics block, which given the desired trajectory $\theta_r, \dot{\theta}_r, \ddot{\theta}_r$ computes the reference torque τ_r to be applied at the elbow joint, ii) a dynamic allocator which subdivides the reference torque between the FES contribution τ_{rFES} and the motor feedforward contribution τ_{rmotor}, iii) a closed-loop FES torque controller, which determines the pulse charge q to stimulate the muscles with current I and pulse width PW in order to produce a torque $\tau_{muscles}$ which is equal to τ_{rFES}, iv) an impedance controller which takes the error between the desired trajectory and the actual trajectory as the feedback term and determines the impedance torque τ_{imp}. The total motor contribution, τ_{motor}, is given by the sum of τ_{rmotor} and τ_{imp}. τ_{motor} and $\tau_{muscles}$ sum at the elbow level and determine the kinematics of the arm $\theta, \dot{\theta}, \ddot{\theta}$.

- Continuous allocation: the allocation factor was computed with a window length of 0.5s and updated at each iteration in order to dynamically allocate the contribution according to the differences in FES performance among repetitions.

D. Simulations

Simulations were performed in Simulink. Four different scenarios were simulated: i) the torque was completely allocated to the motor, ii) the torque was completely allocated to FES and the motor contribution was limited to the impedance control, iii) the reference torque was allocated with the discrete allocation, iv) the reference torque was allocated with the continuous allocation. Simulations of 300 seconds were performed where the reference trajectory for each repetition was a complete elbow flexion/extension $(0-135°)$. Each movement lasted 5 seconds, and between two consecutive repetitions a pause of 2.5 seconds was inserted.

Performance was assessed according to the following metrics: i) root mean square (RMS) of the reference torque allocated to FES, as an indicator of the allocation effects, ii) RMS of the total motor torque generated during the movement, as an indicator of power consumption, iii) angle root mean square error (RMSE), as an indicator of trajectory tracking performance, iv) FES-induced muscle torque RMSE normalized with respect to the reference torque allocated to FES, as an indicator of the PI controller performance and v) charge integral.

III. RESULTS

Results are shown in Table I for each simulation scenario at the 2^{nd}, 20^{th} and 40^{th} movement repetitions. As it can be observed, the reference torque allocated to FES is the highest for the FES+impedance case, followed by the continuous strategy and the discrete strategy. The lower motor torque RMS is achieved by the FES+impedance case followed by the continuous strategy and the discrete strategy. The lowest angle RMSE is achieved by the motor only case, followed by the continuous strategy, the discrete strategy and the FES+impedance case. The normalized FES-induced torque RMSE is in general the highest for the FES+impedance case, while it is the lowest for the continuous allocation strategy. Finally, the integral of the normalized charge does not conspicuously change among different strategies and it slightly increases with the number of repetitions.

As an example, we report in Fig. 2 results related to the continuous allocation strategy during the 2^{nd}, 20^{th} and 40^{th} repetitions. As it can be observed, the normalized charge saturates in each repetition, determining the error on the muscle torque and the intervention of the motor.

IV. DISCUSSION

As expected, the lowest trajectory tracking error is achieved when relying solely on the motor at the expense of the highest motor torque. The use of FES in combination with the impedance control allows to reduce the motor generated torque, but it significantly worsens the trajectory tracking performance with respect to the motor-only case. The use of allocation strategies also permits to improve the trajectory tracking performance while maintaining low torque requirements. The deterioration of the trajectory tracking performance can be explained considering the compliant nature of the controller, which should allow FES to actively contribute to the movement.

The good performance of the continuous allocation strategy might be explained by the fact that the reference torque is allocated according to the current muscle performance, which might change inside the same movement repetition.

Regarding fatigue effects, results suggest that fatigue was better managed when an allocation strategy was introduced, in particular the continuous one. This was proved by the overall lower normalized muscle torque RMSE. Thanks to the allocation based on fatigue detection, the muscles were asked to contribute only with their residual capability.

An advantage of the developed control architecture is that it does not require time-consuming subject-specific FES parameter estimation and it is still able to manage fatigue. This makes it a good candidate for the clinical context.

TABLE I: Simulation results are shown in terms of: i) Reference torque allocated to FES, ii) total motor torque generated during the movement, iii) trajectory tracking error, iv) normalized FES-induced muscle torque tracking error, and v) charge integral. Results are presented for Motor-only, FES+impedance, Discrete and Continuous allocation cases, for the 2^{nd}, 20^{th} and 40^{th} repetitions.

	Ref. FES torque RMS [Nm]			Motor torque RMS [Nm]			Angle RMSE [°]			Norm. FES torque RMSE			Charge integral		
	2^{nd}	20^{th}	40^{th}	2^{nd}	20^{th}	40^{th}	2^{nd}	20^{th}	40^{th}	2^{nd}	20^{th}	40^{th}	2^{nd}	20^{th}	40^{th}
Motor only	0.00	0.00	0.00	2.22	2.22	2.22	0.00	0.00	0.00	NA	NA	NA	NA	NA	NA
FES+imp.	2.22	2.22	2.22	0.74	0.81	0.90	4.14	4.57	5.04	0.28	0.31	0.36	3.17	3.41	3.52
Discrete	1.40	1.24	1.35	0.94	0.97	1.04	2.16	2.66	2.64	0.23	0.33	0.31	2.95	3.26	3.31
Continuous	1.83	1.76	1.65	0.80	0.92	1.01	1.90	2.08	2.26	0.16	0.19	0.23	3.20	3.28	3.31

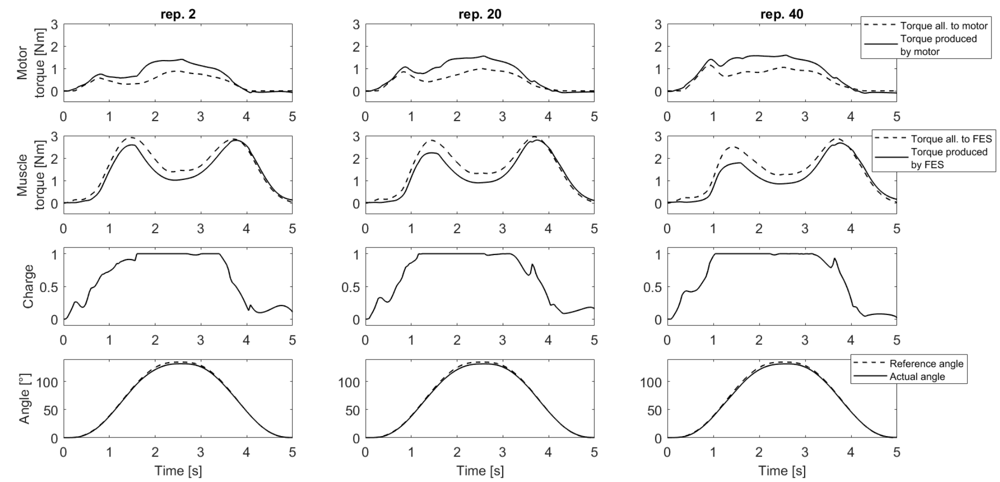

Fig. 2: Simulation results are shown in terms of allocated and produced motor torque, allocated and produced muscle torque, normalized charge and angle tracking for the continuous allocation strategy a the 2^{nd}, 20^{th} and 40^{th} repetitions.

This work is a preliminary study of a novel cooperative controller for elbow flexion/extension movements. Although results seem promising, limits must be acknowledged. First of all, the upper limb model was simplified. In particular, it did not include the passive muscle behavior and the actuator dynamics. Regarding the FES-muscle model, the NARX training dataset was acquired in isometric conditions, thus not including the dynamic model of the muscle.

In the future, the actuator dynamics and the muscle passive dynamics should be included in the model. The FES controller could be improved by considering strategies for the adaptation of the parameters. The work could be expanded to multiple joints and, finally, the controller should be implemented and tested with an experimental hybrid system.

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