Mirjana Ruppel, Christian Klauer* and Thomas Schauer

Enhancing the smoothness of joint motion induced by functional electrical stimulation using co-activation strategies

Abstract: The motor precision of today's neuroprosthetic devices that use artificial generation of limb motion using Functional Electrical Stimulation (FES) is generally low. We investigate the adoption of natural co-activation strategies as present in antagonistic muscle pairs aiming to improve motor precision produced by FES. In a test in which artificial knee-joint movements were generated, we could improve the smoothness of FES-induced motion by 513% when applying co-activation during the phases in which torque production is switched between muscles – compared to no co-activation. We further demonstrated how the co-activation level influences the joint stiffness in a pendulum test.

Keywords: Antagonistic muscle pairs, Co-activation, FES, Neuroprosthetics

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1 Introduction

Functional Electrical Stimulation (FES) is a well-established technology used to artificially activate muscles by recruiting the muscle's motor units using electrical pulses. It is often applied to restore motor functions in paralyzed patients in therapy or to support daily-life activities. However, precise motor functions are often difficult to realize since the stimulation-effect is difficult to model and therefore to predict. Hence, commercial devices mainly focus on reliable, most commonly open-loop strategies often using pre-defined stimulation patters applied to a single muscle.

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In healthy joint-movements, however, motor precision is achieved by a simultaneous contraction of multiple

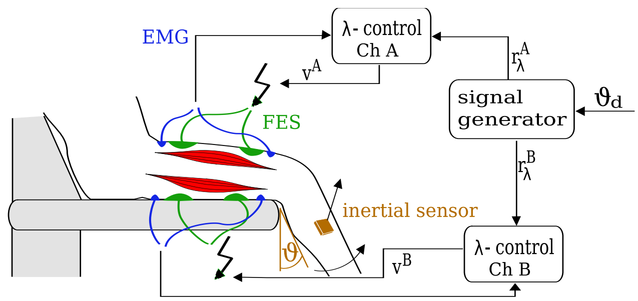

Figure 1: The experimental setup as described in Sec. 2. Recruitment (λ-) controlled FES is applied to the antagonistic muscle pair formed by the quadriceps and the hamstring muscle. Using a feedback of the evoked EMG, the stimulation intensities v^A and v^B are adjusted such that the effects of muscle fatigue are compensated. The knee angle is measured using an inertial sensor attached to the lower leg.

antagonistic muscles. This phenomenon leads to an increase of the mechanical stiffness in the joints and therefore enhanced motor precision and an increased stability of postures [1].

Control of antagonistic muscle pairs with FES is typically performed using a switching strategy that distributes a one-dimensional actuation variable to two activation levels for both muscles (e.g. [2]). Commonly only one muscle is activated at the same time. Co-activations that allow the modulation of the mechanical impedances can also be artificially induced using FES. It is expected that this has a great potential to improve FES-induced motion [3]. Indeed, more than twenty years ago it was demonstrated, that the involvement of co-contractions can significantly increase the performance of control systems that use FES for actuation [2].

In the mid-nineties, Zhou et al. [4] investigated different co-activation strategies under isometric conditions to realize sinusoidal and linear shaped joint torques. Their results show an improved torque tracking performance as the level of co-activation increases.

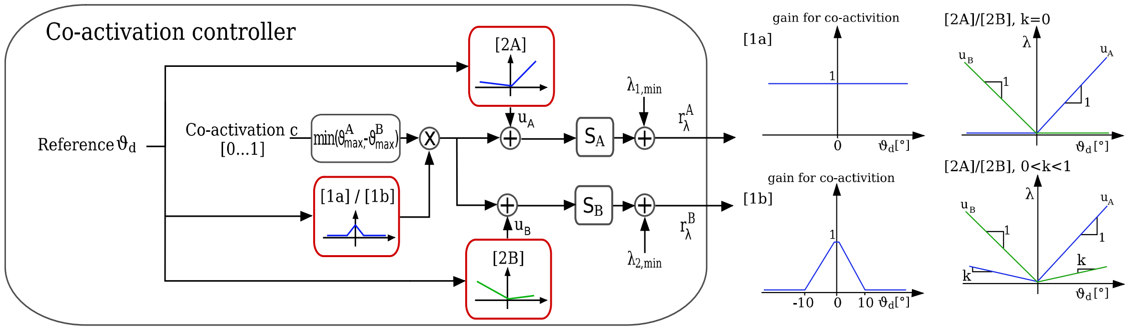

Figure 2: *Left*: The structure of the co-activation controller manipulating the inputs to both $\boldsymbol{\lambda}$-controllers (c.f. Fig. 1). *Middle*: Two options [1a] and [1b] define how the manually adjusted co-activation level is modulated depending on $\boldsymbol{\vartheta_d}$. *Right*: The switching strategy [2A]/[2B] realizes the joint angle tracking, while also introducing muscle co-activation for $\boldsymbol{k} > 0$.

As muscle fatigue increases, the level of co-activation is likely to decrease as time progresses for the same stimulation levels. Most commonly, the individual contribution of each involved muscle cannot be estimated as only measurements of the summation of all individual muscle contraction-effects are typically available. Therefore, a reliable generation of a desired level of co-activation is difficult to realize. Thus, a periodically repeated re-calibration of the control system would be required to adapt the generation of co-activation to the current muscle fatigue state. None of the existing systems consider the difficulty of properly maintaining the level of co-activation in relation to muscle fatigue.

In this publication, we investigate the feasibility of co-activations to improve motor precision in the FES-based control of the knee joint-angle. To overcome issues with fatiguing muscles, we use an underlying feedback of the FES-induced muscle activation obtained by evoked electromyography (eEMG) measurements [5]. The closed loop enforces an almost linear muscle behavior and further compensates the effects of muscle fatigue until the control signal (stimulation intensity) saturates. The results described in [5] apply to the shoulder deltoid muscle and we assume that this similarly applies to the quadriceps- and hamstrings muscle.

In a fundamental investigation, we demonstrate in a pendulum test how the joint stiffness increases as the level of co-activation increases. We further compare the approaches already investigated for torque realization in [4] to control the knee joint angle using different approaches that employ co-activation. A significant improvement in motor precision was observed in the two best-performing approaches compared to the simple switching-strategy that does not use co-activation.

2 Experimental set-up

As shown in Fig. 1, a healthy subject sits on an elevated surface with his/her leg free to swing. FES is applied to the quadriceps (channel A) and the hamstring (channel B) muscle using a current-controlled stimulator (Rehastim, Hasomed GmbH, Germany) and self-adhesive electrodes (ValuTrode® CF4090 (4x9 cm), Axelgaard Manufacturing Co., USA). A stimulation frequency of 25 Hz is used and corresponds to the sampling frequency of the control system (sampling index j). The control signals $v^A[j]$ and $v^B[j] \in [0,1]$ are the normalized pulse charges varying between zero and the maximally tolerable charges Q^A_{max} and Q^B_{max}. Current amplitude and pulse width are computed using the charge control method as described in [5].

The abduction angle $\vartheta[j]$ is measured at 25 Hz by means of an inertial sensor (MPU 9250, InvenSense Inc., San Jose, USA). The orientation is estimated using the approach given in [6]. When FES is active, EMG is measured at 2048 Hz using separate Ag/AgCl electrodes (Ambu Neuroline 720, Ambu A/S, Denmark) and recorded using a signal amplifier embedded into a device combining FES and EMG (RehaMovePro, [7], Hasomed GmbH, Germany). The EMG electrodes are placed outside the stimulation electrodes, as indicated in Fig. 1. By filtering the stimulation evoked EMG, the estimated muscle recruitment levels $\lambda^A[j]$ and $\lambda^B[j]$ of the preceding stimulation period are obtained at stimulation frequency as described in detail in [5]. The recruitment controller adjusts the stimulation intensities $v^A[j]$ and $v^B[j]$ to produce the desired recruitment levels $r^A_\lambda[j]$ and $r^B_\lambda[j]$.

The control algorithms are implemented using OpenRTDynamics (www.openrtdynamics.sf.net) considering all realtime critical parts and Scilab to automate calibration procedures.

3 Methods

An open-loop control strategy to control the knee-joint angle ϑ according to a reference ϑ_d that adjusts the desired recruitment levels $r_\lambda^A[j]$ and $r_\lambda^B[j]$ is used. Different co-activation methods can be applied.

Because of the underlying recruitment controllers, the level of recruitment r_λ can be assumed to relate linearly to the resulting steady-state angle ϑ [5]:

$$\boldsymbol{\vartheta} = \underbrace{\frac{\vartheta_{max}^A}{r_{\lambda max}^A}}_{:=m^A} r_\lambda^A + \underbrace{\frac{\vartheta_{max}^B}{r_{\lambda max}^B}}_{:=m^B} r_\lambda^B. \quad (1)$$

Herein, $\vartheta_{max}^A > 0$ and $\vartheta_{max}^B < 0$ are the angles obtained when applying the maximal possible recruitment levels $r_{\lambda_{max}}^A > 0$, $r_{\lambda_{max}}^B > 0$, respectively. These parameters are automatically determined during the calibration process of the λ-controllers, as described in [5]. We further assume that the contributions T^A and T^B to the joint torque $T = T^A + T^B$ produced by each muscle A and B is also linearly dependent on the recruitment level $r_{\lambda_{max}}^A$, $r_{\lambda_{max}}^B$, respectively, as described by:

$$T^A = \mathrm{g}m^A\, r_\lambda^A \geq 0, T^B = \mathrm{g}m^B\, r_\lambda^B \leq 0, \mathrm{g} > 0. \quad (2)$$

The co-activation torque (amount of counteracting torque) that is expected to modulate mechanical impedances is then defined as follows:

$$T_{co} = \min(T^A, -T^B) = g \min(m^A\, r_\lambda^A, -m^B\, r_\lambda^B). \quad (3)$$

Here, g is a scaling factor. The maximal possible co-activation torque is then given by

$$T_{co,max} = T_{co}\big|_{r_\lambda^A = r_{\lambda_{max}}^A, r_\lambda^B = r_{\lambda_{max}}^B} = g \min(\vartheta_{max}^A, -\vartheta_{max}^B). \quad (4)$$

The used controller is depicted in Fig. 2. Two online-adjustable parameters are provided: A reference angle ϑ_d and the level of co-activation c. The factors m^A and m^B of the linear models (1) and (2) are cancelled by their inverses $S_A := 1/m^A$ and $S_B := 1/m^B$, respectively. A desired joint angle ϑ_d is then realized by the switching strategy [2A]/[2B] as outlined in Fig. 2. The factor $0 \leq k < 1$ allows the respective counteracting muscle to be activated to a desired extent.

A further modulation of the co-activation is possible by choosing $0 < c \leq 1$. For $c = 1$ the maximal possible co-activation torque $T_{co,max}$ (c.f. Eq. (4)) is obtained because c is scaled by $\min(\vartheta_{max}^A, -\vartheta_{max}^B)$. Two options [1a] and [1b] are available: In option [1a] the level of co-activation is constantly applied to the muscles. In option [1b], the co-activation is modulated depending on the reference angle ϑ_d as illustrated in Fig. 2: Within a range $\vartheta_d \in [-10°, 10°]$ the level of co-activation is increased to its maximum for $\vartheta_d = 0°$. Otherwise no co-activation is present. This shall lead to an increased joint stiffness during the phases when muscle activation switches occur.

4 Experimental results

All tests were performed in a healthy subject (age 23, female) asked to always relax. Prior to the respective experiments, the recruitment controllers were calibrated by the automated procedure described in [5].

4.1 Pendulum test

With this experiment, we wanted to investigate the influence of the co-activation level on the knee-joint's stiffness properties. The subject's leg is flexed to approximately 55°. Then it is released and the swinging process is recorded. This is done for multiple levels of co-activation $c \in [0,1], \vartheta_d = 0$ and option [1a]. Fig. 3 exemplarily shows the results for $c = 0$ (no co-activation) and $c = 0.15$. The duration t_{sw} of the swinging is determined by the duration the envelope of ϑ takes to enter 5% of its initial amplitude. As we see in Fig. 3, t_{sw} decreases as the level of co-activation increases.

4.2 Motor precision

Using the controller described in Sec. 3, a piecewise linear movement in a range $\vartheta \in [\vartheta_{low}, \vartheta_{high}]$ shall be realized. Different parameter-combinations are investigated. For this, we generate the reference ϑ_d by means of a two-point controller that alternates between linearly inc.-/decreasing ϑ_d using a given slope $s = 72°/s$. The slope changes its direction when the measured knee angle ϑ crosses the thresholds $\vartheta_{low} = -7°$ and $\vartheta_{high} = 17°$.

We want to achieve a motion that transitions steadily through the switching of the activation between hamstrings and the quadriceps ($\vartheta_d \approx 0°$). Therefore, we tested the following combinations:

(A) No co-activation: $k = 0, c = 0$,
(B) Simultaneous muscle activation: $k = 0.1, c = 0$,
(C) Constant co-activation: option [1a]: $k = 0, c = 0.15$,
(D) ϑ_d-modulated co-activation: option [1b], $k = 0, c = 0.15$,
(E) Combining **(B)** and **(D)**: option [1b], $k = 0.1, c = 0.15$.

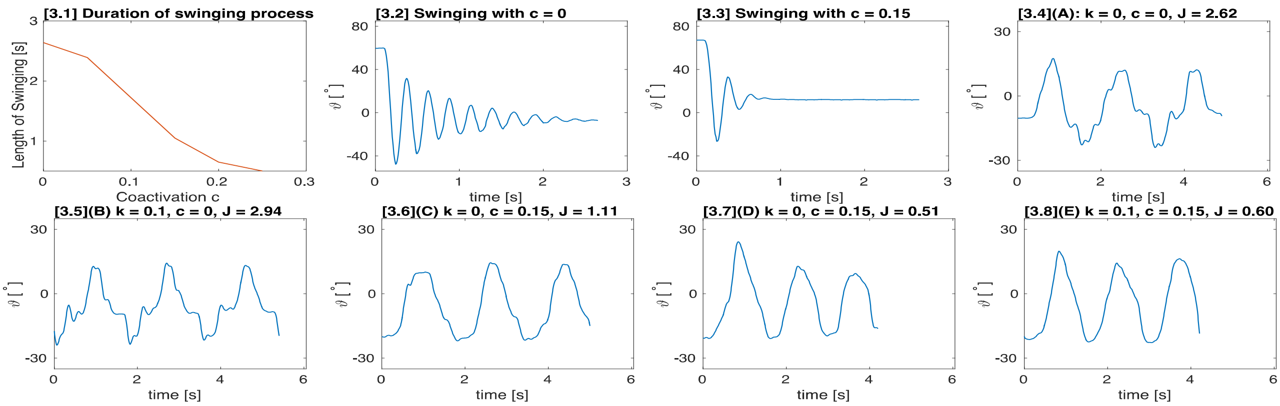

Figure 3: *3.1*: The duration t_{sw} of the swinging process (option [1a] active) decreases with rising co-activation levels as explained in Sec. 4.1. *3.2*: Swinging process for 0% co-activation. *3.3*: Swinging process for 15% co-activation. *3.4-3.8*: **(A)** no co-activation leads to discontinuities of the resulting angle. *3.5*: Variant **(B)** results in pronounced discontinuities. *3.6*: **(C)** a constant co-activation causes fairly smooth transitions during muscle switching. *3.7*: **(D)** leads to very smooth transitions. *3.8*: **(E)** also causes smooth transitions and further yields a nearly triangular movement. The smoothness J improved by 513% when comparing **(A)** to **(D)**.

To evaluate the smoothness during the switching of activation, a linear model was fitted to the measured angles in a range $\vartheta \in [-5°, 5°]$. The RMS-error between this model and the measured data then yields the degree of smoothness J. Further, ideally, the resulting movements are triangularly shaped.

The results are given in Fig. 3. Variants **(A)** and **(B)** result in jerky movements, especially when switching between muscles occurs. Using constant co-activation **(C)**, the smoothness improves strongly. Even better results are obtained in **(D)** when co-activation is induced when switching occurs. Expectations on a triangular movement were mostly fulfilled in **(D)** and **(E)**.

5 Conclusion

Good results were obtained for **(C)**, **(D)** and **(E)**, in which the co-activation level ([1a] or [1b]) was activated $c = 0.15$. Otherwise performance was low. The best results were obtained for variants **(D)** and **(E)**. Using **(E)**, we expect a decreased maximal range of movement and a fast progression of muscle fatigue as the level of muscle activation is higher in average. Therefore, we prefer **(D)**.

Using **(D)**, motion smoothness could be significantly (by 513%) improved compared to no co-activation. We expect the results to also apply to e.g. the control of the elbow-joint angle and the finger and wrist extension/flexion. These investigations aim as a basis for future neuroprostheses to restore paralyzed functions.

Author's Statement

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