Virtual Weight-Compensating Exoskeleton using λ-Controlled FES

Christian Klauer[1] Arne Passon[1], Jörg Raisch[1,2] Thomas Schauer[1]

[1]Control Systems Group, Technische Universität Berlin, Germany
[2]Max-Planck-Institut für Dynamik komplexer technischer Systeme, Magdeburg, Germany
Contact: klauer@control.tu-berlin.de

Introduction

Stroke patients often suffer from a reduced volitional muscle activity that may hinder functional movements. To support such small voluntary forces, functional electrical stimulation (FES) can be applied to the corresponding muscle for gaining additional force. In neuro-prosthetic systems supporting the patients movement, the stimulation is generally triggered by an estimate γ of the volitional muscular activity which is obtained by electromyography (EMG).

In practice, simple control strategies are usually used that apply a pre-defined stimulation level if a pre-defined threshold for the volitional EMG is exceeded. These system are robust to muscular fatigue, but generally apply more stimulation than needed. This leads to a faster proceeding of muscular fatigue. An arbitrary control of a joint angle profile is also not possible.

The more elaborated control strategies adjust the stimulation intensity proportionally to the estimated volitional activity [1] or use an integration based scheme. For the latter, stimulation is either ramped up or down or stays constant depending on the volitional EMG crossing thresholds of a hysteresis. In the past, both methods did not compensate for muscular fatigue.

A general problem of EMG-controlled methods is the very low signal to noise ratio of the volitional EMG-signal especially for patients with less volitional muscular activity. If the signal to noise ratio is too low, it becomes difficult to reliably differentiate between multiple thresholds. Also for the EMG-proportional control the ability of the patient to control e.g. a joint angle is reduced. As the signal to noise ratio gets better, a more precise control of the stimulation intensity becomes possible.

In this work a different approach is investigated: The support by FES is controlled by a static, linear feedback of the joint angle. Hereby, the joint-torque component induced by the artificial muscle activation shall be equal to a fraction of the required torque for holding the current angle in steady state (under compensation). A similar strategy was applied to support standing up in [2], however the control system did not involve an under compensation. Therefore, the stability of the employed closed loop presented in [2] is questionable.

As shown in the here presented results, a higher sensitivity of the control system to the volitional muscle activations could be achieved, allowing a precise positioning even with low voluntary activity. Hereby, the system was exemplary applied to support the shoulder abduction by stimulating the medial deltoid muscle.

To allow an adaption to muscular fatigue, instead of a direct stimulation of the muscle, the λ-control method [3] is applied for precisely controlling the muscular recruitment regardless of muscle fatigue.

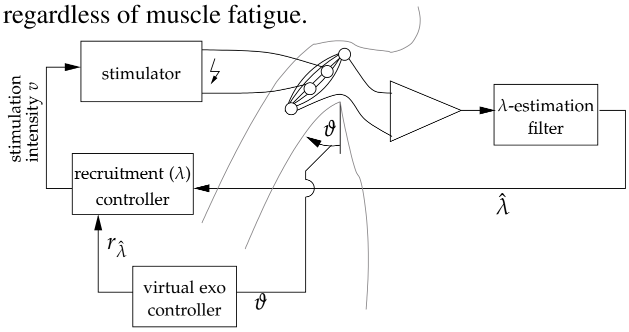

Figure 1: Experimental set-up applied to the shoulder.

Experimental Set-up & Controllers

The set-up is shown in Fig. 1. The λ-control method involves a recording of the stimulation evoked EMG (eEMG), obtained from AgCl electrodes located around the pair of hydro-gel surface stimulation electrodes. After each stimulation pulse, causing a recruitment of motor units, an electrical wave (the so called M-wave) appears in the EMG, which is the sum of action potentials of all stimulation-activated motor units. The intensity $\hat{\lambda}$ of this wave is taken as an estimate for the muscular recruitment level. A fast feedback of $\hat{\lambda}$ allows to adjust the stimulation intensity v such that a reference level $r_{\hat{\lambda}}$ is tracked. The ability to compensate for non-linear behaviour in the recruitment function like hysteresis effects or activation thresholds as well as muscular fatigue has been shown in [4]. The muscle is then much easier to model and control and the resulting feedback-enforced behaviour from $r_{\hat{\lambda}}$ to $\hat{\lambda}$ can be considered as a linear system. The calibration procedure involves the tuning of only one parameter [3].

For measuring the abduction angle ϑ, an inertial sensor is attached to the upper arm.

On top of the underlying λ-control system, the actuation variable of the virtual exoskeleton controller is applied to r_{λ}. For the proposed control scheme it is assumed, that the recruitment level λ is proportional to the joint torque induced due to FES.

The virtual exoskeleton controller linearly maps the obtained joint angles within a range $\Omega_{\vartheta} := [\vartheta_{min}, \vartheta_{max}]$ to a virtual actuation variable $r_{\hat{\lambda}} \in \Omega_{\hat{\lambda}}$ in the range $\Omega_{\hat{\lambda}} := [\hat{\lambda}_{min}, \hat{\lambda}_{max}]$:

$$r_{\hat{\lambda}} = \frac{\hat{\lambda}_{max} - \hat{\lambda}_{min}}{\vartheta_{max} - \vartheta_{min}} (\vartheta - \vartheta_{min}) + \hat{\lambda}_{min}. \quad (1)$$

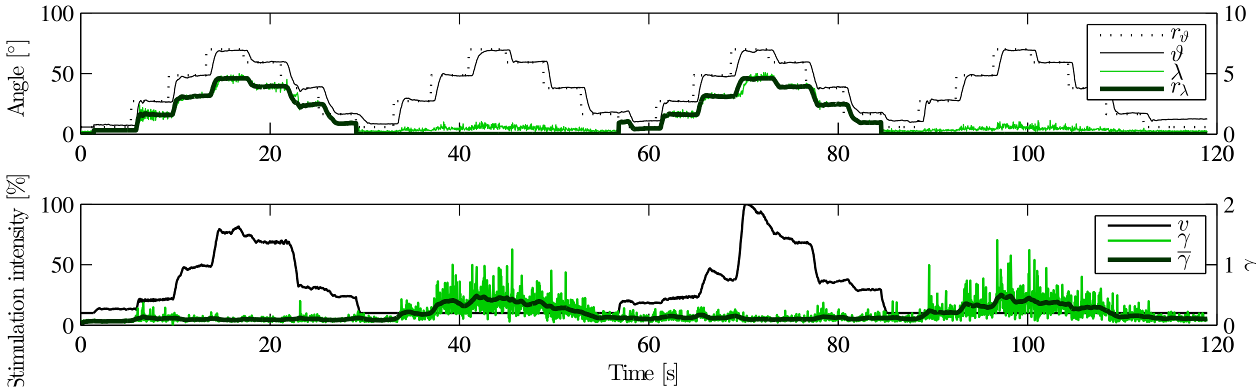

Figure 2: Results for a comparison of the FES-induced weight compensation to a pure voluntary muscle activation. The symbols are: r_ϑ - target joint angle as displayed to the subject, ϑ - joint angle, λ - muscular recruitment index, r_λ - desired recruitment level given to λ-control, v - stimulation intensity resulting from λ-control, $\gamma, \overline{\gamma}$ - voluntary muscle activity.

For the case of the shoulder abduction, the angular range is set to $\Omega_\vartheta = [0°, 90°]$. The base recruitment level $\hat{\lambda}_{min}$ is slightly below the noise level of $\hat{\lambda}$. Therefore, for the minimum joint angle $\vartheta = \vartheta_{min}$, the muscle is at the onset of a contraction due to FES. The upper bound for the reference recruitment $\hat{\lambda}_{max}$ is adjusted such that this level leads, when applied to $r_{\hat{\lambda}}$, to the joint angle $\vartheta_{s,max}$ that is about 30% less than the upper bound of the operable range ϑ_{max}. This ensures an under compensation of arm weight and a static gain of the closed loop less than one. Under the assumption of a stable, second order, linear system without transmission zeros for the mechanical dynamics, it can be shown that the control system is stable. Measured joint angle values outside the operable range Ω_ϑ are saturated to the respective bounds before being applied to the controller (1).

The control system is implemented using OPENRTDYNAMICS[1] on a PC running Linux with RT-Preemption-Patch. All devices are connected via USB-interfaces.

Results

The control scheme was applied to a healthy subject and the results are shown in Fig. 2. For evaluation purposes, the subject was asked to voluntary follow a given reference trajectory r_ϑ for the abduction angle ϑ. The trajectory was displayed along with a real-time visualisation of the measured angle ϑ. This task had to be repeated for four times, whereby during the first and third run, the weight-compensation controller was activated. For comparative purposes, the controller was deactivated during the second and the fourth trial.

For determining the subject's activity, the volitional EMG γ was additionally evaluated. As observed in Fig. 2, the volitional activity is significantly lower during active FES. For each trial, the mean of the volitional EMG γ was calculated for a time window of 2 s during which the maximal joint angles occurs. In case of the activated controller, the voluntary activity could be reduced to 6.8% of the activity without support.

Conclusions

The high sensitivity of the control system to changes in the joint angle allows the control of the weight compensating neuro-prosthesis even for levels of voluntary muscle activity that cannot be sufficiently estimated by EMG measurements. Therefore, even in such cases a precise joint angle positioning can be achieved by the subject. The use of the underlying λ-control scheme compensates for muscular fatigue and additionally allows a for a low-effort calibration procedure as the main non-linearities do not have to be identified during calibration.

Although the proposed control scheme is linear, the real neuro-musculo-skeletal system is non-linear (e.g. due to the mechanical dynamics and the dependency of the joint torque on the joint angle ϑ and velocity). Therefore, more elaborated control approaches could further improve the control system and comfort for the user. Additionally, it could be beneficial to implement a haptic control scheme that allows for a virtual therapist to enable training sessions for relearning functional movements at home.

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