Advanced Control Strategies for Neuro-Prosthetic Systems

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Abstract – In the field of feedback controlled motor neuro-prosthetic systems, a major problem is the high effort in modelling the dynamic and static properties of muscles, which forms a pre-requisite for the synthesis of high performance control systems. One of the main issues is the static non-linear function between the stimulation intensity and the amount of recruited motor units of the muscle. While being essential for controller synthesis, this function is typically time-variant and hard to estimate. Therefore, this contribution gives an overview of two novel methods, which allow a linearization of this function by cascaded feedback control using additional measurements (acceleration and electromyogramm (EMG)) of the internal neuro-musculo-skeletal state. The EMG-based method enables a precise adjustment of the muscular recruitment level. Both methods are experimentally evaluated and one example application is presented which forms a new approach for the control of drop foot compensation systems. A comparison of both control strategies concludes this contribution.

1 Introduction

Artificial control of human limb movements in neuro-prosthetic systems using Functional Electrical Stimulation (FES) forms a challenging task if natural movements shall be imitated [2]. Related motor control systems generally employ feedforward or feedback control strategies by modulation of the stimulation intensity to achieve a certain goal e.g. tracking a reference trajectory for an joint angle [1].

For controller design, usually intensive effort for identifying muscular models is required [9] whereas the obtained models are still approximative. Furthermore, the modelling of muscular fatigue and hysteresis effects on motor unit recruitment by FES is difficult and a fast model adaptation to individual subjects is hard to achieve.

An important model component is the nonlinear, spatial recruitment curve [3] which describes the number of motor units activated by FES. This static function usually includes a threshold that has to be known exactly for most control strategies. However, the actual threshold value strongly depends on the muscular fatigue, which can significantly decrease control performance. At least an initial estimate can be obtained through several methods. However, the obtained result varies depending on the used method [3].

For improving this situation, two novel methods based on feedback control are investigated. The main idea behind both methods is to obtain additional information about the internal state of the neuro-musculoskeletal system (in contrast to usually exclusively used joint-angle measurements), which is then exploited by cascaded feedback strategies. The fast inner control loops of the cascades enforce a linear behavior for the not exactly known part of the stimulated muscles. The resulting feedback-controlled muscles are then much easier to model and to control. Two internal signals are examined separately to get an estimate of the internal muscle state: the angular acceleration and the FES-evoked EMG (eEMG), i.e. the electrical response of the muscle to an external electrical stimulus.

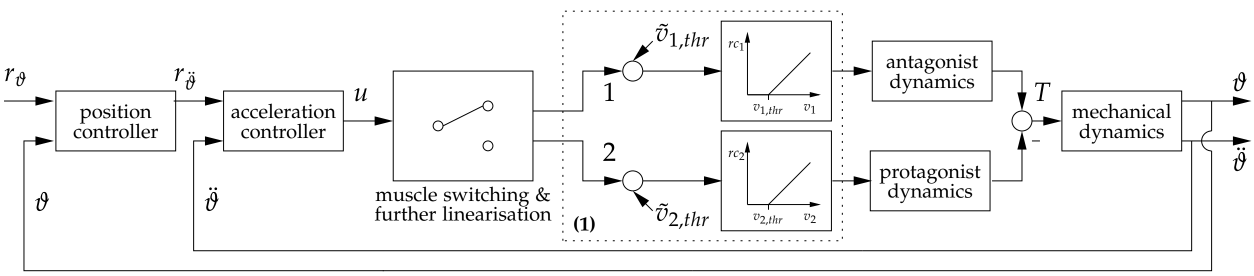

Figure 1: *The acceleration control scheme applied to an antagonistic muscle pair.*

2 Underlying Acceleration Control

Joint movements are induced by muscular torques, that are closely related to accelerations. Therefore, controlling the acceleration can significantly improve the resulting joint movements. For motion control, only the position and optionally the velocity were used for feedback in the past [1].

A closed loop applied to an antagonistic muscle pair involving the joint-angle acceleration is shown in Fig. 1. The angular acceleration and the joint angle can be estimated by mounting an inertial sensor on each

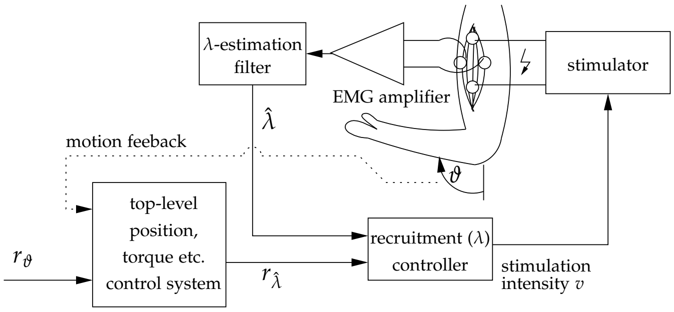

Figure 2: *The λ-Control system exemplarily applied to the biceps*

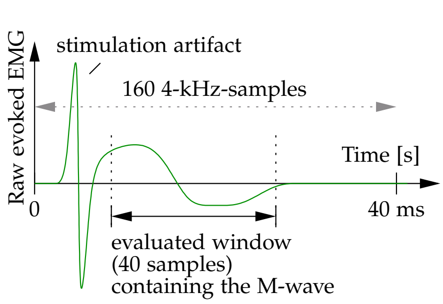

Figure 3: *Stimulation artifact and M-wave occurring after a stimulation pulse.*

shank related to the joint and by using special signal processing algorithms. The muscular thresholds $v_{i,thr}$ ($i = 1, 2$) of the recruitment functions rc_i can be partially compensated if the initial thresholds are estimated. The compensations $\tilde{v}_{i,thr}$ are chosen to be 20% smaller than the estimates of $v_{i,thr}$ in order to avoid an overcompensation. Since two muscles in an antagonistic muscle pair are stimulated, a switching strategy is applied: For a negative control signal u the extensor muscle is activated and for a positive control signal the flexor muscle. For the sake of simplicity the treatment of the muscular dynamic behavior is omitted in this description (for details cf. [7]).

When combining this switching law with the muscular thresholds and their undercompensations, a dead-zone for u around zero occurs in which no motor units are recruited. Generally, the mechanical dynamics are difficult to control in the presence of such a dead-zone, when only a – compared to an acceleration feedback typically slow – position feedback loop is used. Therefore, as an intermediate step between the switching law and the position controller, the high bandwidth acceleration control loop is placed. The unwanted effects of the dead-zone are compensated by this inner closed loop. Moreover, the acceleration feedback also allows for a fast compensation of mechanical input disturbances. A detailed analysis of the acceleration feedback method is given in [7].

Results: The method was exemplary applied to control the elbow-joint angle of a healthy subject via the antagonistic muscle pair formed by the biceps and the triceps. The arm was placed in a way that movements were not affected by gravity. Very small muscular joint moments were required in this case so that an untreated dead-zone would have caused major problems. However, with the used underlying acceleration feedback, a fast positioning with a low rise time of 100 ms was observed. Fig. 4 shows the good results of the position tracking task for a given reference.

3 Recruitment control by measuring the FES evoked EMG (λ-control)

The FES evoked Electromyogramm (eEMG) represents the sum of action potentials that are generated by the synchronous activation of motor units due to a stimulation pulse and is usually called M-wave. The measurement can be performed via additional surface electrodes in between the stimulation electrodes (cf. Fig. 2). Following a stimulation artifact, the M-wave can be observed in the recorded signal (Fig. 3).

The intensity of the M-wave in terms of amplitude and width (area) reflects the number λ of recruited motor units. By using a digital filter algorithm, an estimate $\hat{\lambda}$ of that number can be obtained [6].

In the past, eEMG was mainly used for muscular fatigue estimation [10] or for predicting the joint torque through an additional model of medium complexity [4], [11]. This model prediction was also used to adapt the stimulation intensity. However, this still required the effort of adapting a model to each individual subject.

In contrast to the previous approaches, the below presented λ-control method (cf. Fig. 2) only requires the tuning of one gain parameter – that can be automatically performed. In the λ-control methods, a closed-loop control system adapts the stimulation intensity v such that a given reference $r_{\hat{\lambda}}$ for the motor unit recruitment λ is tracked. This allows for the precise adjustment of the muscular recruitment. As the highly non-linear recruitment function rc is covered by the closed loop, its influence to the actual recruitment is linearized. This is even possible without an estimate of this function. A detailed description of this control method was previously published in [6]. An upper level control system (feedback or feedforward) may then directly adjust the muscular recruitment instead of the stimulation intensity, the impact of which on contraction strength is difficult to predict. Therefore, a more precise control of the joint torque or the angular motion becomes possible.

Results: For demonstrating the effectiveness, the proposed method was applied to control the recruitment of the deltoid muscle of an healthy subject. A fast tracking performance (representing a linearisation) of the desired recruitment was observed as shown in Fig. 5. For the same muscle and subject, the non-linear recruitment curve was estimated by applying a triangular signal (ranging from zero to the tolerable maximum of the stimulation intensity v) while the estimate $\hat{\lambda}$ for the muscular recruitment was recorded. In a scatter

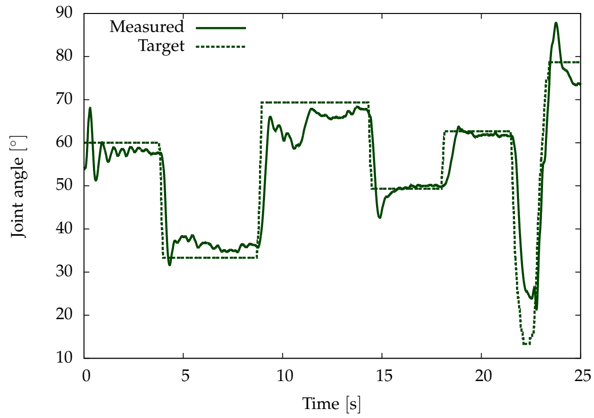

Figure 4: *Elbow joint-angle control experiment.*

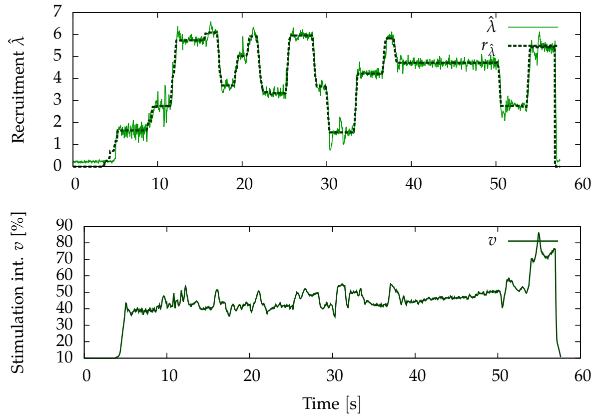

Figure 5: *Tracking of the muscular recruitment reference $r_{\hat{\lambda}}$.*

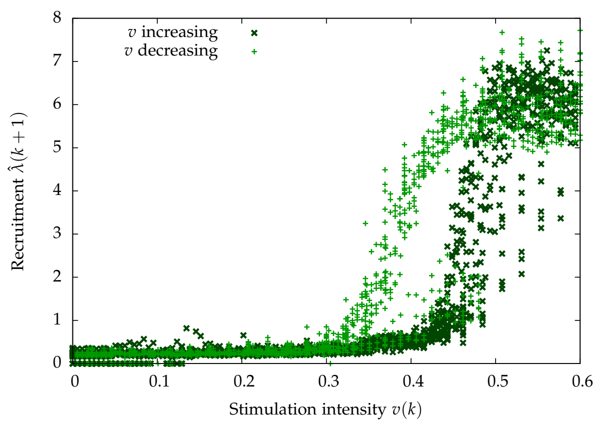

Figure 6: *Observed hysteresis for $\hat{\lambda}$, when applying an triangular stimulation test signal.*

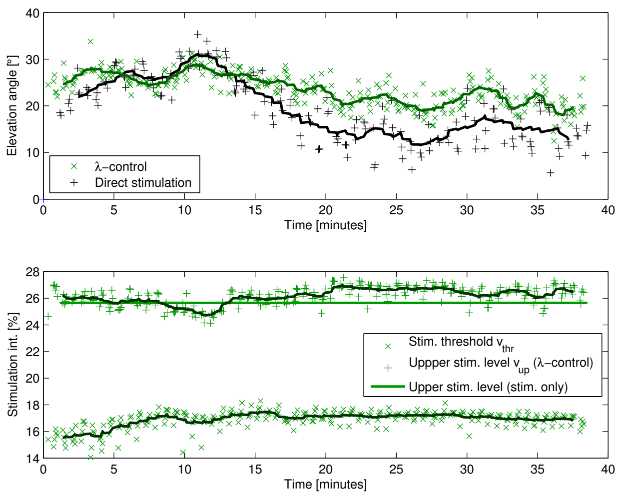

Figure 7: *Results of the λ-control for the drop foot correction.*

plot showing $\hat{\lambda}$ against v (cf. Fig. 6), the typical non-linear behavior – that had to be modeled in the past for improving the controller performance – is visible.

4 Application of λ-control to the drop foot compensation problem

The correction of drop foot forms a typical application of FES requiring long term robustness [8]. Due to a stroke, the voluntary elevation of the foot may be hindered. Commercial stimulation systems for daily use typically apply a pre-defined stimulation profile to the tibialis anterior/peroneal nerve during the swing phase of each step in order to elevate the foot. Muscular fatigue is not taken into account. Hence, in general more stimulation intensity than needed is applied to ensure a sufficient elevation. Research focuses on advanced systems using feedback control of the resulting foot elevation such that only the really essential amount of stimulation is applied. However, an additional angle sensor is needed that requires additional space and mounting effort.

In this example application, λ-control is used instead of a direct stimulation (DS) to apply pre-defined motor unit recruitment profiles. Only for evaluation purposes, an inertial sensor is used for measuring the elevation angle.

In a calibration phase, the recruitment level $\hat{\lambda}_a$ that results in a foot elevation by approx. 25° is determined. During the experiment, the healthy subject is sitting with the foot free to swing. To elevate the foot during a simulated step, the activation level $\hat{\lambda}_a$ is used as reference recruitment $r_{\hat{\lambda}}$ for 2.5 s.

To compare the λ-control strategy with the approach of direct stimulation, five steps using λ-control are carried out followed by three steps of direct stimulation, while the λ-control system is deactivated. This procedure is periodically repeated for approximately 40 minutes. To obtain a comparable stimulation intensity profile for the DS-method, five λ-controlled steps are initially performed while the resulting stimulation intensity profiles (the actuation variable trajectories of the controller) are recorded. The average of these profiles is used as the future intensity profile for DS.

To monitor the threshold of the recruitment curve v_{thr} during the experiment, a very small but constant reference $r_{\hat{\lambda},min}$ (slightly above the noise level of $\hat{\lambda}$) is applied before each λ-controlled swing phase. As a result, the muscle starts to contract. When the closed loop reaches its steady state, the corresponding stimulation intensity v is stored yielding v_{thr}.

Results: The results for an experiment including 290 λ-controlled steps and 171 DS steps are shown in Fig. 7. The foot elevation angle is shown for each step in the upper part for both methods. For all λ-controlled steps, the determined threshold v_{thr} is shown in the lower plot together with the applied upper stimulation intensities v_{up} in the swing phases (λ-controlled and DS).

The quantities v_{thr}, v_{up} as well as the elevation angle were smoothed by using a moving average filter. The result is shown in Fig. 7 as solid lines.

A considerable decrease in the achieved elevation angle is observed for the direct stimulation while the angles resulting for activated λ-control are less decreasing.

For λ-control, the smoothed stimulation intensities v_{up} and v_{thr} increased by up to $12.64\ \%$ and $9.1\ \%$ with respect to the minimum of the corresponding curves, respectively. This indicates an adaptation to muscular fatigue when using λ-control even without angular feedback.

5 Conclusions

The proposed acceleration control scheme counteracts the effects of badly compensated recruitment functions and additionally helps to reject mechanical input disturbances. However, acceleration feedback is not applicable for torque control under isometric conditions (no acceleration present) and for antagonistic muscles with desired co-contraction.

The λ-control method allows the precise adjustment of the muscular recruitment regardless of time-variant muscle behavior. It requires only a minimum of calibration effort. Compared to acceleration control, the achievable performance of the inner feedback loop is even faster.

The feasibility of both methods was experimentally illustrated. In the future, also a combination of both methods is worth to investigate to further improve motion quality.

Concerning the drop foot example, at the current state of development, the λ-control method requires two additional EMG-electrodes. Therefore, the effort is comparable to the feedback solution with an angular sensor, of course. However, the EMG-measurement from stimulation electrodes is likely to be avaible in the near future, yielding a system size comparable to current commercial systems.

Unlike acceleration control, the λ-control method is applied per muscle. For antagonistic muscle pairs, it becomes possible to keep both muscles co-contracted at a desired level even in the long term due to the adaption to fatigue. The exploitation of such co-contractions for modulating mechanical impedances (e.g. stiffness) represents ongoing research [5] and is expected to allow neuro-prosthetic systems to mimic the natural motor control behavior.

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