# Two-Channel Muscle Recruitment ($\lambda$)-Control using the Evoked-EMG

Christian Klauer, Thomas Schauer

***Abstract*—This contribution is concerned with the use of the Electromyogram (EMG) evoked by Functional Electrical Stimulation (FES) as a feedback variable for controlling muscular recruitment. Feedback controlled FES has shown to be difficult in the past, mainly due to the non-linear recruitment process and fast progression of muscle fatigue. Hence, in previous works we proposed a method for controlling the muscle recruitment that compensates these effects by a fast feedback loop. ($\lambda$-control) In this contribution, the recruitment control method is extended to two-channel operation, which has been difficult in the past because of cross talk effects between closely placed surface EMG-electrodes. Instead of following a decoupling strategy of the electrical signals, we introduce a robust time-multiplexing strategy to separate the responses of the individual muscle parts. In the proposed approach, M-wave overlapping must be avoided limiting the maximal stimulation frequency. Hence, in a study including five healthy subjects and one stroke patient the maximum was determined for each subject. A stimulation frequency of 26 Hz was possible for all subjects. Then, the proposed two-channel $\lambda$-control was successfully tested in one healthy subject showing the feasibility of the proposed concept. To show the applicability of recruitment control to clinical environments, control results for a stroke patient are additionally shown.**

## I. Introduction

FES applied to paralyzed muscles is often used in neuroprosthetic systems to restore motor function to e.g. support or re-train daily live activities. The stimulation intensity is commonly controlled open-loop and typically triggered by events. More elaborated approaches involve a continuous feedback of joint angle- or force measurements with the aim of obtaining a higher motor precision and more natural movements. However, closed-loop control of FES has shown to be difficult in the past because of two main reasons: The effect of muscle fatigue strongly degrades the controller performance as fatigue progresses and, second, the recruitment process due to FES is highly non-linear and hence difficult to model and to adapt to the individual subject [1].

To address these difficulties, we developed a method using a feedback of the FES-evoked EMG, which is the muscle response to an external stimulus, to allow a precise control of the muscle recruitment [2]. The evoked EMG is composed of the stimulation pulse artifact and the M-wave resulting from the superposition of the activation potentials of all recruited motor units. Hence, the evaluation of the M-wave intensity yields an estimate of the muscle-internal recruitment state. The stimulation intensity is then controlled such that a desired recruitment level is realized. The effects of muscle fatigue and the non-linear recruitment behavior are compensated by a high-performance feedback controller without requiring knowledge about these effects.

In the past, the idea of employing the evoked EMG to control FES has already been proposed in [3] but was not further investigated since then. Other approaches involve complex estimation filters to predict muscle force [4] or estimate muscle fatigue using the evoked EMG [5], however there is an increased calibration effort to identify detailed models.

In the control of antagonistic muscle pairs, requiring at least two-channel FES, artificially induced co-activations are expected to improve motor precision [6], [7]. Herein, an effective muscle fatigue compensation is highly beneficial, as it is typically not possible to measure the co-activation strength. The individual feedback of the evoked EMG of both muscles, however, is expected to allow an precise adjustment the co-activation strength.

Up to now, the application of multi-channel recruitment control to muscles close to each other is difficult as strong cross-talk effects, typically leading to a reduced control performance, are present in the EMG measurement and the individual M-waves cannot be distinguished. To reduce cross talk in a two-channel set-up, we propose to apply stimulation pulses not synchronously but alternating for both stimulation channels. In turn, the respective M-waves appear non-overlapped in the measured EMG. We apply the control system to the medial and anterior deltoid as it is clinical relevant to support fully or partially paralyzed arm abduction movements.

## II. Methods

*a) Experimental Set-up:* The experimental set-up is shown in Fig. 1. Herein, FES is applied to the anterior (channel A) and the medial (channel B) part of the deltoid muscle using a current-controlled stimulator (Rehastim, Hasomed GmbH, Germany) and self-adhesive electrodes (ValuTrode® CF4090, Axelgaard Manufacturing Co., USA). The stimulation evoked-EMG caused by both deltoid parts is acquired at 2048 Hz using one pair of Ag/AgCl electrodes (Ambu Neuroline 720, Ambu S/A, Denmark) placed outside of both stimulation electrodes and an amplifier (Porti $32^{TM}$, TMS International, The Netherlands). Further, the shoulder joint angle describing arm elevation $\vartheta$ is measured using an inertial sensor (MTx, Xsens Technologies B.V., The Netherlands). The measured EMG is processed by two digital filters to estimate the normalized muscular recruitments $\hat{\lambda}^A$ and $\hat{\lambda}^B$ respectively. The stimulation intensities $v^A$ and $v^B$ describe the pulse-charges normalized to the respective maximal tolerated values. Based on the pulse-charge, current and pulse-width are calculated using the method described in [2]. Two feedback controllers adjust $v^A$ and $v^B$ respectively, such that the desired recruitment levels $r^A_{\hat{\lambda}}$ and $r^B_{\hat{\lambda}}$ are realized. [2]

The control system was implemented using the real-time framework OpenRTDynamics[1] and runs a computer system running Linux.

C. Klauer (corresponding author) and Thomas Schauer are with Technische Universität Berlin (Control Systems Group), Berlin, Germany (phone: +49 30 314-22276; e-mail: klauer@control.tu-berlin.de).

The experimental trials have been approved by the ethics committee of the Berlin Chamber of Physicians (Ärztekammer Berlin).

[1] http://openrtdynamics.sf.net/

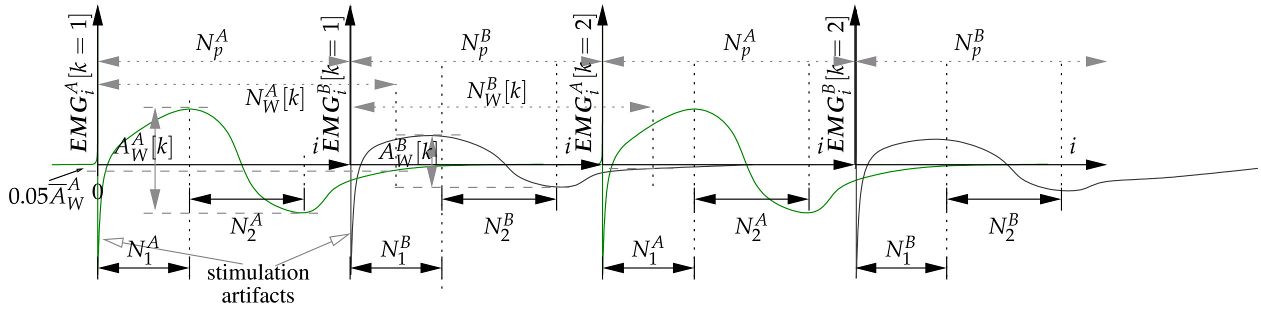


Figure 2. Time multiplexing stimulation and measurement of the FES-evoked EMG.

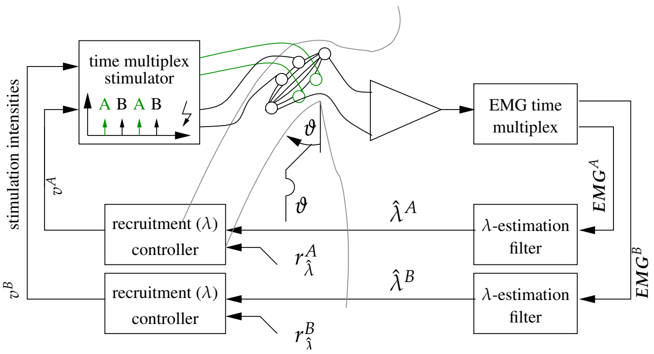


Figure 1. Experimental set-up for two channel recruitment control.

*b) Time-Multiplexed Stimulation and EMG-Measurement:* When applying stimulation pulses to each muscle part simultaneously, both individual responses typically overlap as they are simultaneously generated by the recruited motor units of both muscle parts. A separation of the individual M-waves by means of decoupling strategy is difficult as e.g. cross talk gains must be calibrated in advance and, further, perfect decoupling is difficult to achieve. To reduce calibration effort, we propose a strategy in which both muscle parts are stimulated in an alternating fashion as illustrated in Fig. 2: After one muscle part is stimulated, the decay of the corresponding M-wave is awaited, whereby the durations of the M-waves $N_W^A[k]$ and $N_W^B[k]$ are introduced as shown in Fig. 2, whereby $k$ denotes the sampling instant of the control system. Then, the other part is stimulated. This strategy causes the respective M-waves to appear in different time-windows and hence, ideally no overlapping appears.

The proposed strategy then records the M-waves starting directly after each stimulus during each sampling instant $k$ for both channels $A$ and $B$. Data are stored in the EMG-measurement vectors $\boldsymbol{EMG}^A$ and $\boldsymbol{EMG}^B$ containing $N_p^A$ and $N_p^B$ EMG-samples for channel $A$ and $B$, respectively. Herein, $i$ denotes the sampling instance for the respective EMG-recording.

In this approach, the corresponding inter-pulse time-periods measured in EMG-samples $N_p^A$ and $N_p^B$ must be large enough to avoid overlapping of both M-waves. As described in the next paragraph in detail, the intensity of M-wave is evaluated only in a window whose position and length is described by the parameters $N_1$ and $N_2$ respectively. Hence, it must be ensured that M-waves do not overlap with the window used for the evaluation of the succeeding M-wave appearing in the other channel.

Hence, $N_p^A$ and $N_p^B$ are constrained to a minimum depending on the maximal durations of the M-waves $\overline{N}_W^A$ and $\overline{N}_W^B$ by the conditions

$$\overline{N}_W^A < N_p^A + N_1^B, \quad \overline{N}_W^B < N_p^B + N_1^A. \tag{1}$$

Hence, also the effective stimulation frequency

$$f_{eff} := \frac{2048}{N_p^A + N_p^B}\text{Hz} \tag{2}$$

is introduced, which is limited to a maximal value, in turn.

To experimentally estimate the maximal M-wave duration for one muscle part $\overline{N}_W$[2], the stimulation intensity is slowly increased within 8 s using a stimulation frequency of 25 Hz to the maximal tolerated value, while EMG is recorded and stored into $\boldsymbol{EMG}[k]$ for each sampling instant $k$. In an offline analysis, the amplitude $A_W$ between the maximum and minimum of each M-wave contained in the vector $\boldsymbol{EMG}[k]$ is calculated:

$$A_W[k] = \max_i \boldsymbol{EMG}_i[k] - \min_i \boldsymbol{EMG}_i[k]. \tag{3}$$

Then, the maximally observed amplitude is calculated by

$$\overline{A}_W[k] = \max_k A_W[k]. \tag{4}$$

As the decaying M-wave enters a band between $-0.05\overline{A}_W$ and $+0.05\overline{A}_W$, the length $N_W[k]$ is obtained as illustrated in Fig. 2. The maximal length observed for all sampling instants is then calculated yielding $\overline{N}_W = \max_k N_W[k]$.

*c) M-wave Intensity as a Feedback Variable:* The intensities of all M-waves in terms of the 1-norm are calculated for both channels $A$ and $B$ for a fixed window starting $N_1^A$ and $N_1^B$ samples respectively after the stimulation artifact and lasting $N_2^B$ and $N_2^A$ samples, yielding the indicators $\hat{\lambda}^A$ and $\hat{\lambda}^B$ for the muscle recruitment levels, respectively.

The parameter set ($N_1^A$, $N_1^B$, $N_2^B$, $N_2^A$) is obtained in a calibration procedure performed for each channel $A$ and $B$ separately. The aim is to optimize the filter with regard to improve the degree of linearity between the recruitment level estimated by the filter and the arm elevation joint angle in steady-state (still different horizontal shoulder rotation angles occur for A and B) corresponding to the induced muscle torque. Further, the signal to noise ratio of the estimated recruitment considered in the optimization. The used approach is described in detail in a separate publication to appear [8]. Herein, results obtained from five healthy subjects and one stroke patient showed an improvement of linearity compared to non EMG-controlled FES as well as the compensation of muscle fatigue.

To control the muscle recruitment levels, two discrete-time integral controllers $K^A(q^{-1}) = c_\lambda^A/(1-q^{-1})$ and

[2]The respective superscripts $A$ and $B$ are omitted to simplify notation in this analysis

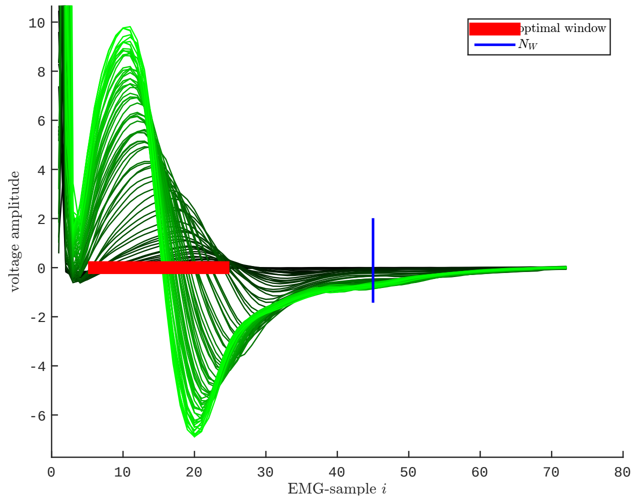


Figure 3. M-waves obtained from the stroke patient P.

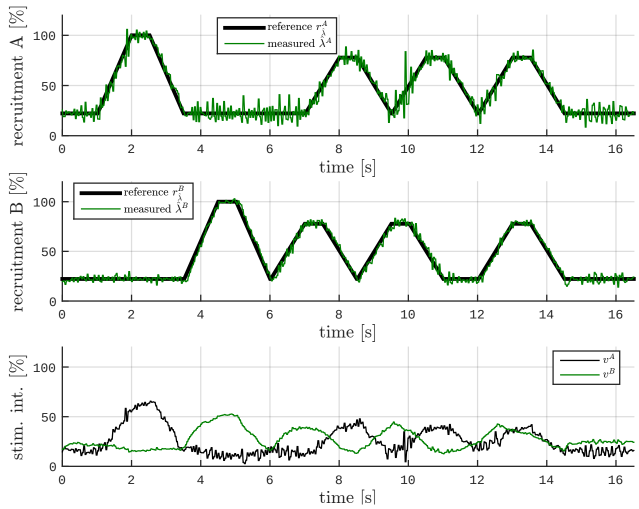


Figure 4. Time series results for two-channel recruitment control in a healthy subject at 25 Hz. Parameters are $N_1^A = 2$, $N_2^A = 16$, $N_1^B = 7$, $N_2^B = 6$, $c_{\hat{\lambda}}^A = 0.01$, $c_{\hat{\lambda}}^B = 0.0049$.

$K^B(q^{-1}) = c_\lambda^B/(1-q^{-1})$ adjust the stimulation intensities $v^A$ and $v^B$ such that the estimated recruitment levels $\hat{\lambda}^A$ and $\hat{\lambda}^B$ follow their given reference trajectories $r_{\hat{\lambda}}^A$ and $r_{\hat{\lambda}}^B$, respectively. Here, $q^{-1}$ is the one-step backwards shift operator $(a(k)q^{-1} = a(k-1))$. The gain factors $c_\lambda^A$ and $c_\lambda^B$ are chosen to maximize closed loop performance, while the standard deviation of the estimation noise – typically amplified in the closed loop – does not exceed 1% of the maximal signal amplitude of $\hat{\lambda}^A$ and $\hat{\lambda}^B$, respectively.

The linearization effect introduced by this control system, applies to the activation threshold of the muscle recruitment function, which is shifted close to the origin. Further, an overall improvement of linearity of the remaining nonlinear behavior of muscle recruitment as well as a compensation of muscle fatigue is achieved. Of course, saturation effects cannot be compensated. FES-based movement control systems can then adjust the desired recruitment levels $r_{\hat{\lambda}}^A$ and $r_{\hat{\lambda}}^B$ (virtual actuation variables) instead of directly using the stimulation intensities, wherein recruitment control serves as an underlying control system.

## III. Results

*d) Maximal Stimulation Frequency in Two-Channel Control:* The typically, maximally achievable stimulation frequency $f_{eff}$ is determined in a study including five healthy subjects (S1 to S5) and one stroke patient (P). Therefor, for each participant, the M-wave duration $A_W^A[k]$ of the medial portion of the deltoid muscle is determined using the method described in Sec. II. The performed analysis is then performed under the assumption that channel B yields similar parameters. Further, the calibration procedure to tune the recruitment controller yielding the parameters $N_1^A$ and $N_2^A$ was performed. Then, the maximal duration $\overline{N}_W^A = \max_k N_W^A[k]$ was obtained. The minimally possible stimulation period duration still fulfilling eq. (1) was then calculated by

$$N_p^A = \overline{N}_W^A - N_1^A \tag{5}$$

and the maximally achievable stimulation frequency was calculated by

$$\overline{f}_{eff} := \frac{2048}{2N_p^A} \text{Hz}. \tag{6}$$

The results for all subjects are summarized in Tab. I. We can state that for the given participants, recruitment control is possible using stimulation frequencies of at least 26.3 Hz (minimum for all subjects). The obtained M-waves in case of the stroke patient are exemplarily shown in Fig. 3 along with the window used to determined the recruitment level (red marking). Further, the maximal duration of the M-wave is displayed (blue line).

TABLE I. Optimal filter parameters, maximal M-wave durations, maximally possible stimulation frequency as well as maximal current amplitude $I_{max}$ and pulsewidth $PW_{max}$.

| Subject | $N_1^A$ | $N_2^A$ | $\overline{N}_W^A$ | $\overline{f}_{eff}$ [Hz] | $I_{max}$ [mA] | $PW_{max}$ [$\mu$s] |
|---|---|---|---|---|---|---|
| P | 6 | 18 | 45 | 26.3 | 62 | 326 |
| S1 | 8 | 23 | 34 | 39.4 | 54 | 270 |
| S2 | 6 | 8 | 36 | 34.1 | 38 | 186 |
| S3 | 7 | 13 | 46 | 26.3 | 38 | 181 |
| S4 | 3 | 17 | 27 | 42.6 | 52 | 272 |
| S5 | 5 | 13 | 44 | 26.3 | 56 | 282 |
| Mean | | | 38.6 | 32.5 | | |
| Stdev | | | 7.6 | 7.3 | | |
| Min | | | | 26.3 | | |

*e) Experimental Validation of Two-Channel Control:* Two-channel recruitment control was applied to one healthy subject and electrodes were placed as shown in Fig. 5. The

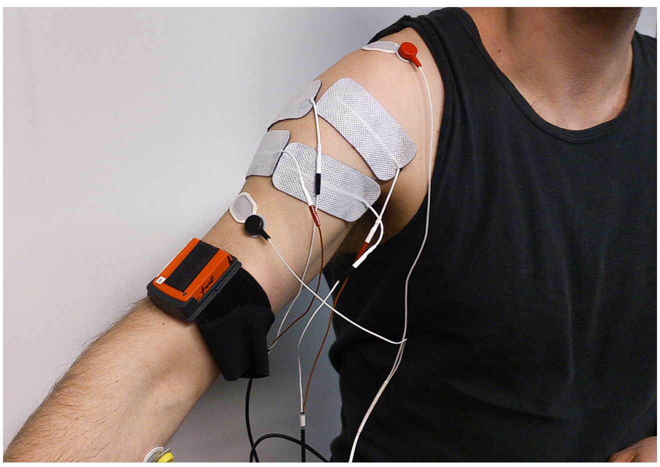

Figure 5. Electrode-placement for two channels.

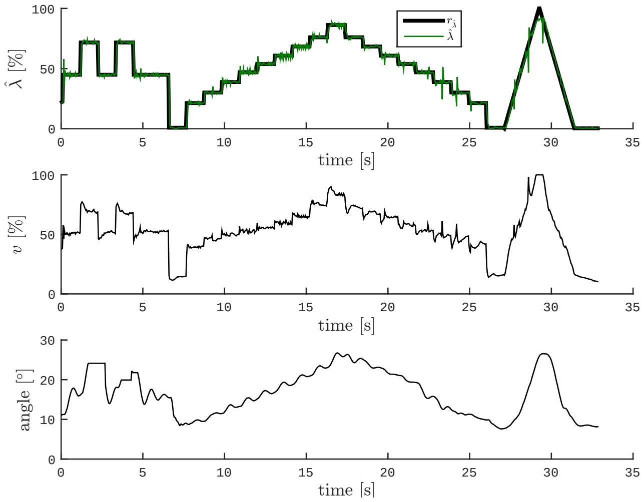


Figure 6. Time series results for single channel recruitment control performed on a stroke patient.

effective stimulation frequency was 25 Hz. The reference trajectories for $r^A_{\hat{\lambda}}$ and $r^B_{\hat{\lambda}}$ were chosen to excite both muscles concurrently. Time series results are given in Fig. 4 showing a precise reference tracking also in case of two-channel recruitment control.

*f) Single Channel Control applied to a Stroke Patient:* To demonstrate the feasibility of the investigated recruitment control strategy in clinical environments, single channel control was applied to the medial deltoid of one stroke patient with almost complete paralysis of the right arm. Results are given Fig. 6 and show a precise tracking of the desired recruitment level along with the resulting abduction angle.

## IV. Conclusion

In this contribution, we investigated an extension of recruitment control for two-channel FES using a time-multiplexing approach to avoid overlapping when obtaining evoked-EMG of two muscle portions. Further, only one pair of EMG-electrodes is required and no additional calibration routine is required, which is an important advantage when donning neuro-prosthetic systems in clinical environments. For single channel recruitment control, we have demonstrated the applicability to clinical environments in a stroke patient. In case of two-channel control, we determined a maximally achievable stimulation frequency of 26.3 Hz for typical subjects by an experimental investigation on six subjects, which is sufficient to the control of most neuro-prosthetic systems. The proposed two-channel recruitment control approach was successfully applied to one healthy subject. An extension to three or more channels is difficult, however, as the maximally achievable stimulation frequency significants degrades with the number of stimulated channels. A hybrid approach of time-multiplexing combined with a classical decoupling strategy may be applied in these cases.

In future research we are going to validate the given approach for a larger group of healthy subjects as well as on stroke patients for different stimulation frequencies. The long-term strategy is to apply two- channel recruitment control to stiffness control in antagonistic muscle pairs.

## Acknowledgement

This work was funded by the German Federal Ministry of Education and Research (BMBF) within the project BeMobil (FKZ 16SV7069K). We further acknowledge Axelgaard Manufacturing Co., USA for donating the used stimulation electrodes.

## References

[1] W. K. Durfee and K. E. MacLean, “Methods for estimating isometric recruitment curves of electrically stimulated muscle,” *Transactions on Biomedical Engineering, IEEE*, vol. 36, no. 7, pp. 654–667, 1989.

[2] C. Klauer, J. Raisch, and T. Schauer, “Linearisation of electrically stimulated muscles by feedback control of the muscular recruitment measured by evoked EMG ,” in *Proc. of MMAR 2012*. Międzyzdroje, Poland: IEEE, 2012, pp. 108–113.

[3] W. K. Durfee and J. T. Dennerlein, “EMG as a feedback signal in surfaces FES applications: issues and preliminary results,” in *Proceedings of the Annual International Conference of the IEEE Engineering in Engineering in Medicine and Biology Society, 1989. Images of the Twenty-First Century*, 1989, pp. 1009–1010 vol.3.

[4] Q. Zhang, M. Hayashibe, and D. Guiraud, “Muscle fatigue tracking based on stimulus evoked EMG and adaptive torque prediction,” in *Proc. of IEEE ICRA 2011*. IEEE, 2011, pp. 1433–1438.

[5] Z. Li, M. Hayashibe, C. Fattal, and D. Guiraud, “Muscle fatigue tracking with evoked emg via recurrent neural network: Toward personalized neuroprosthetics,” *Computational Intelligence Magazine, IEEE*, vol. 9, no. 2, pp. 38–46, 2014.

[6] W. Durfee, “Task-based methods for evaluating electrically stimulated antagonist muscle controllers,” *Biomedical Engineering, IEEE Transactions on*, vol. 36, no. 3, pp. 309–321, march 1989.

[7] E. Burdet, R. Osu, D. W. Franklin, T. E. Milner, and M. Kawato, “The central nervous system stabilizes unstable dynamics by learning optimal impedance,” *Nature*, vol. 414, no. 6862, pp. 446–449, 2001.

[8] C. Klauer, S. Ferrante, E. Ambrosini, U. Shiri, F. Dhne, I. Schmehl, A. Pedrocchi, and T. Schauer, “A patient-controlled functional electrical stimulation system for arm weight relief,” *Accepted presumeably for: Medical Engineering and Physics, Elsevier*, 2016.