From muscles to the brain: MU-based EEG filters

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Funded by the European Union
UK Research and Innovation

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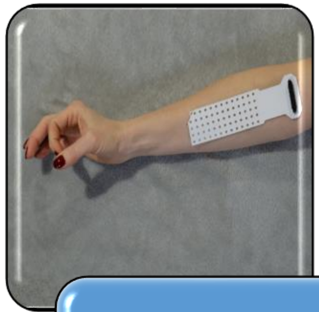
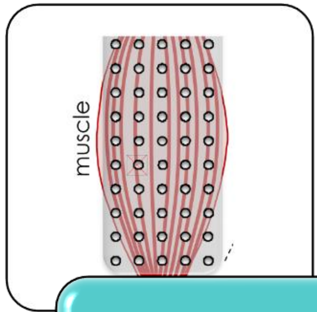
Processing pipeline

Segment A



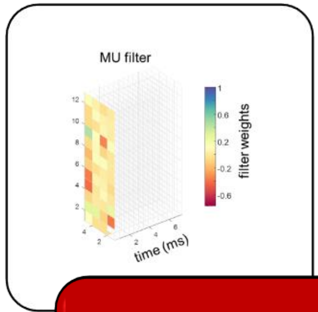
HDEMG acquisition

Segment B



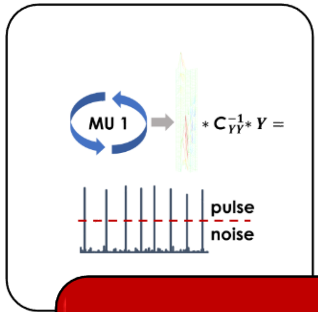
HDEMG channel selection

Segment C



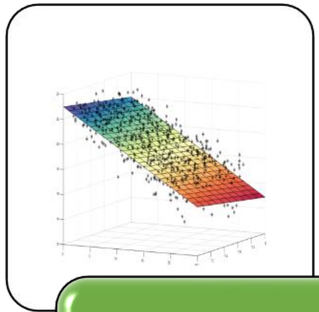
BSS-based MU filter estimation

Segment D



MU filter optimization & firings segmentation

Segment E



Physiological assessment

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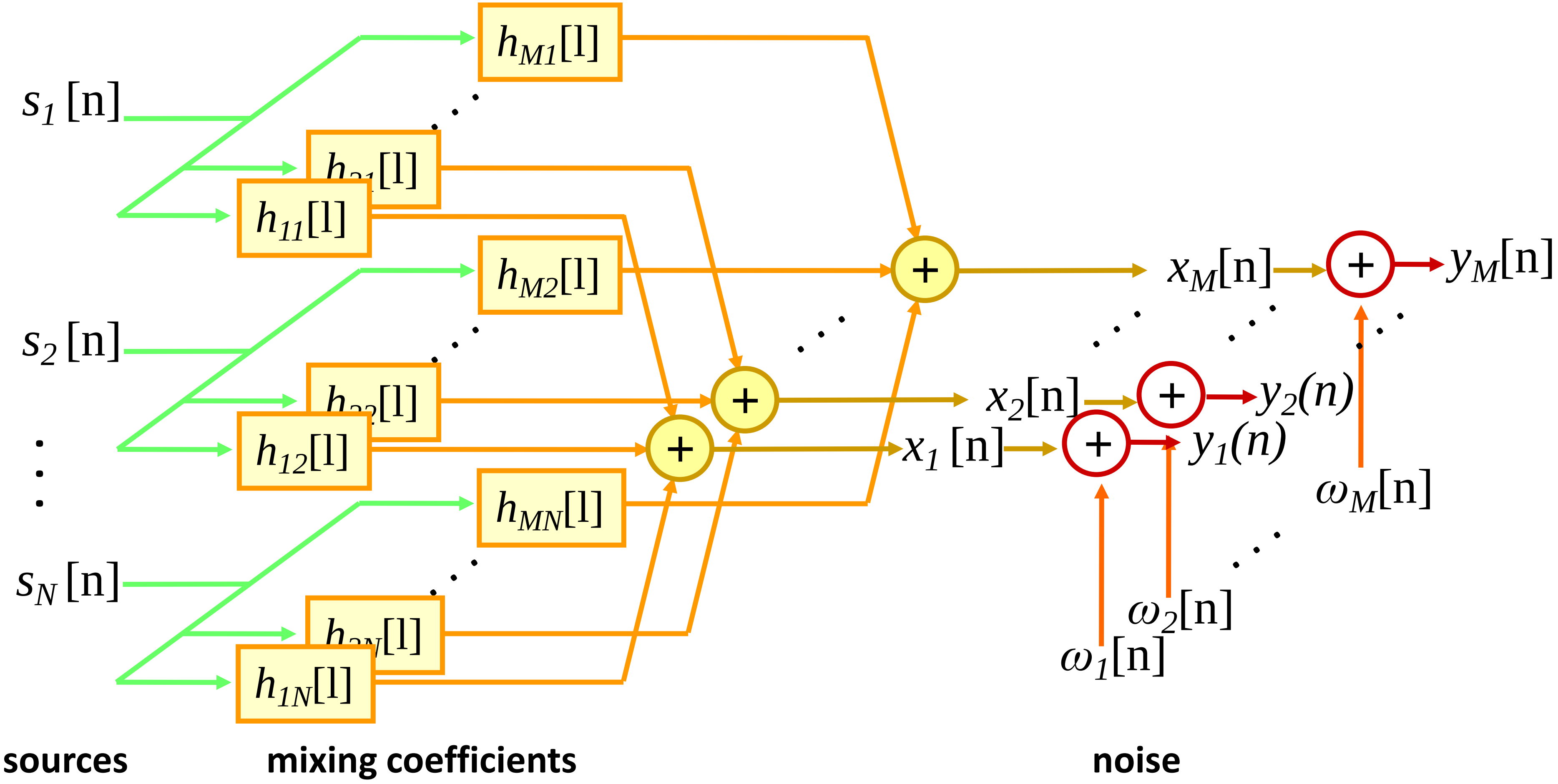
Isometric HDEMG model: Convolutive



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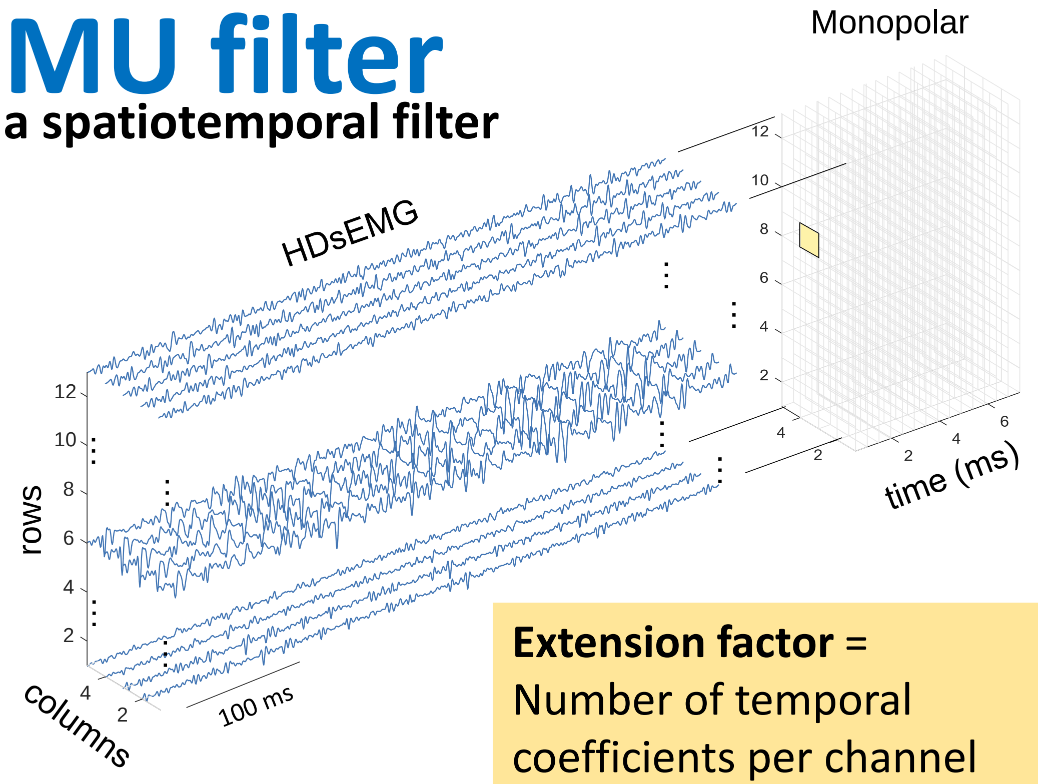
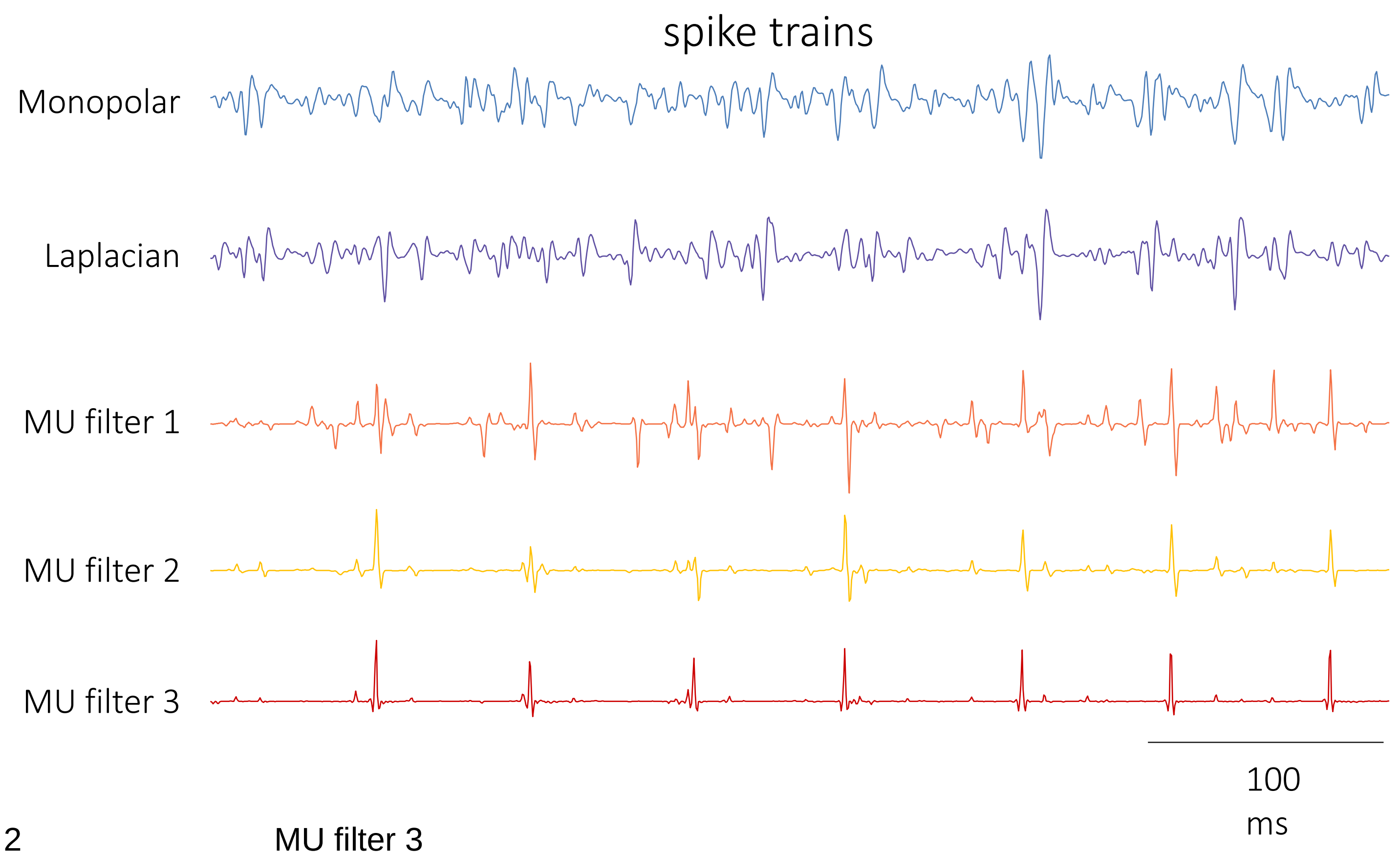
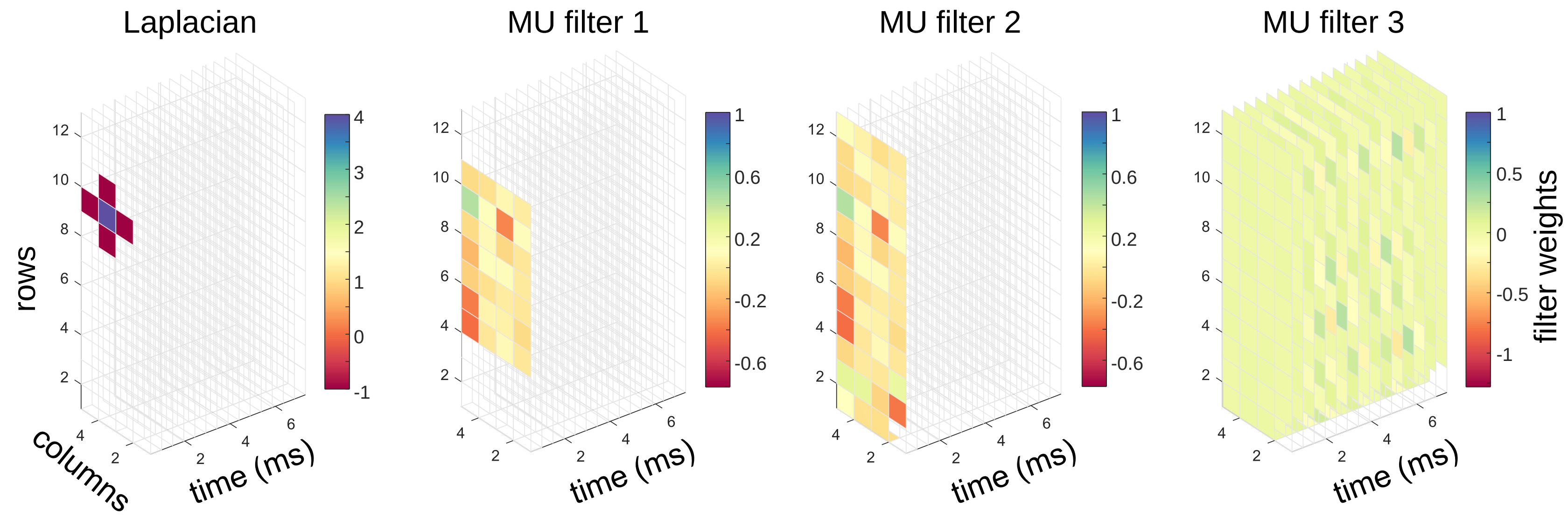
MU filter

a spatiotemporal filter



Extension factor =
Number of temporal coefficients per channel





Škarabot et al. The Journal of Physiology, 2023

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Convolutive EMG model: matrix form

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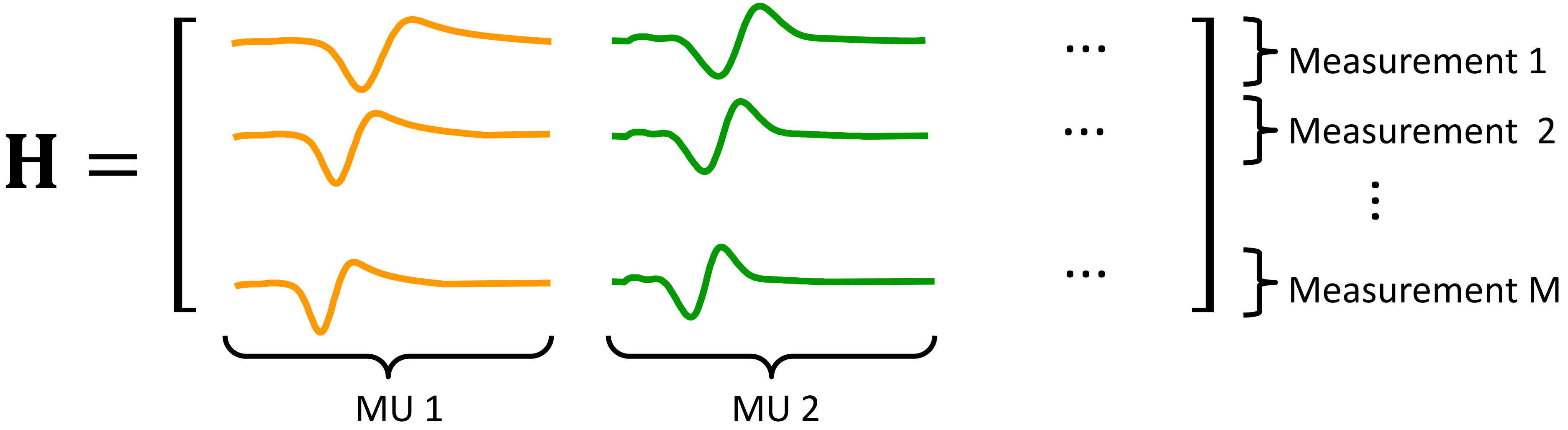
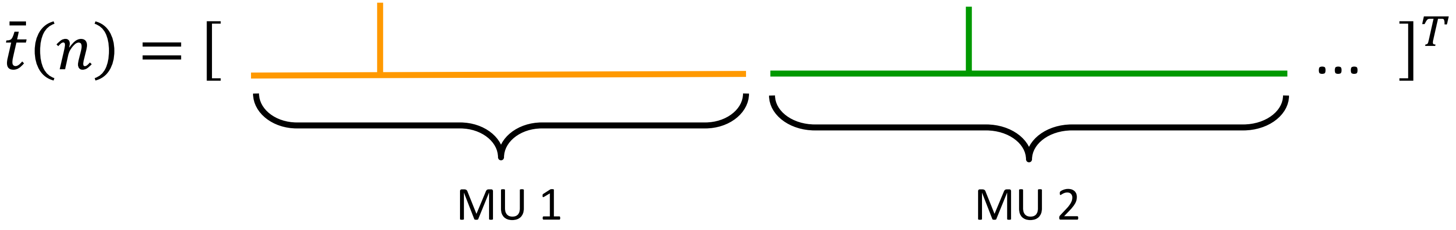
Matrix of convolution kernels (MUAPs):



Extended vector of pulse sources:



A. Holobar, D. Zazula: Convolution Kernel Compensation, *IEEE Trans. Signal Proc.*, 2007

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Data model: MUAPs of Biceps Brachii

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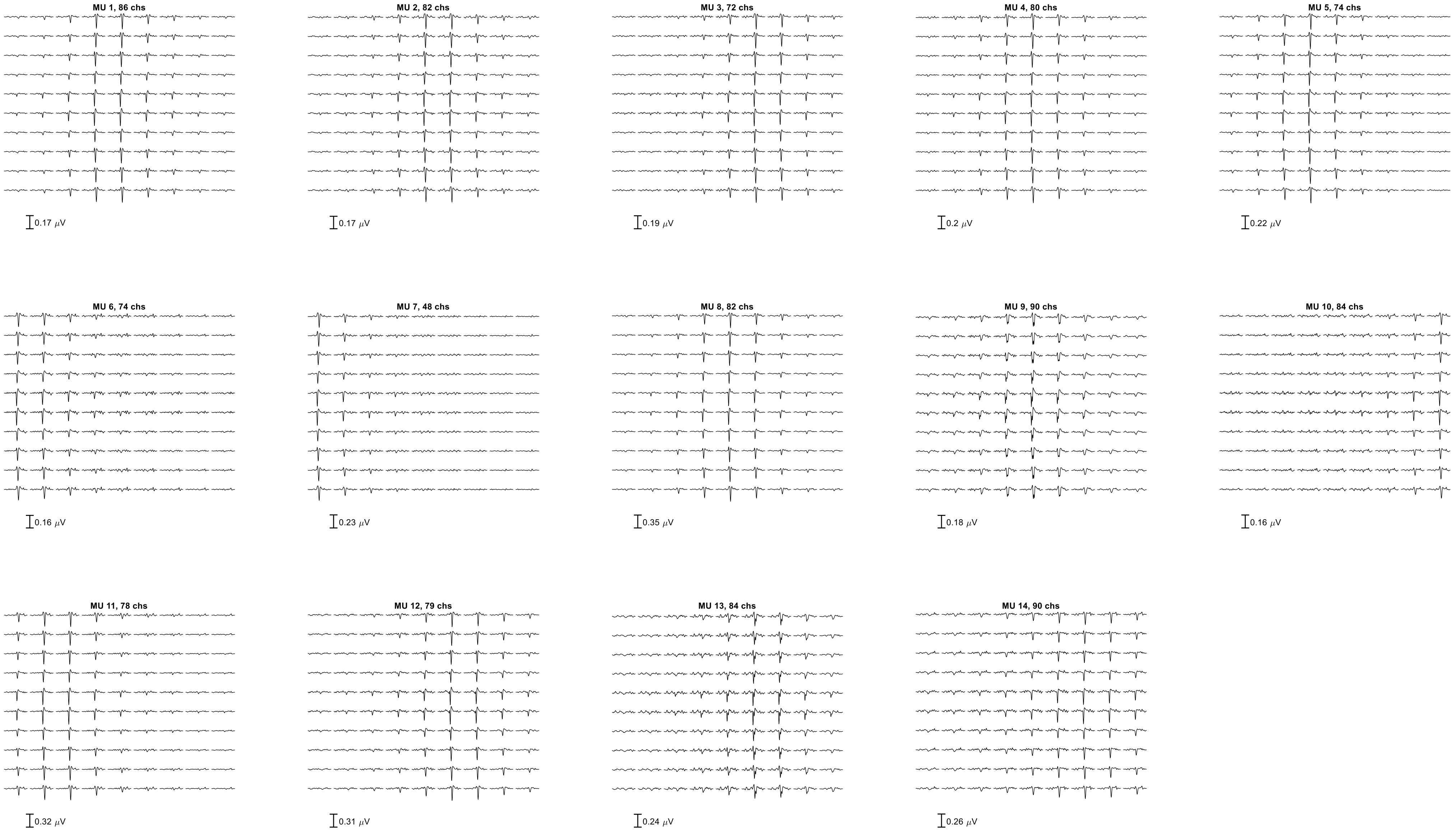

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Data model: MUAPs of Soleus

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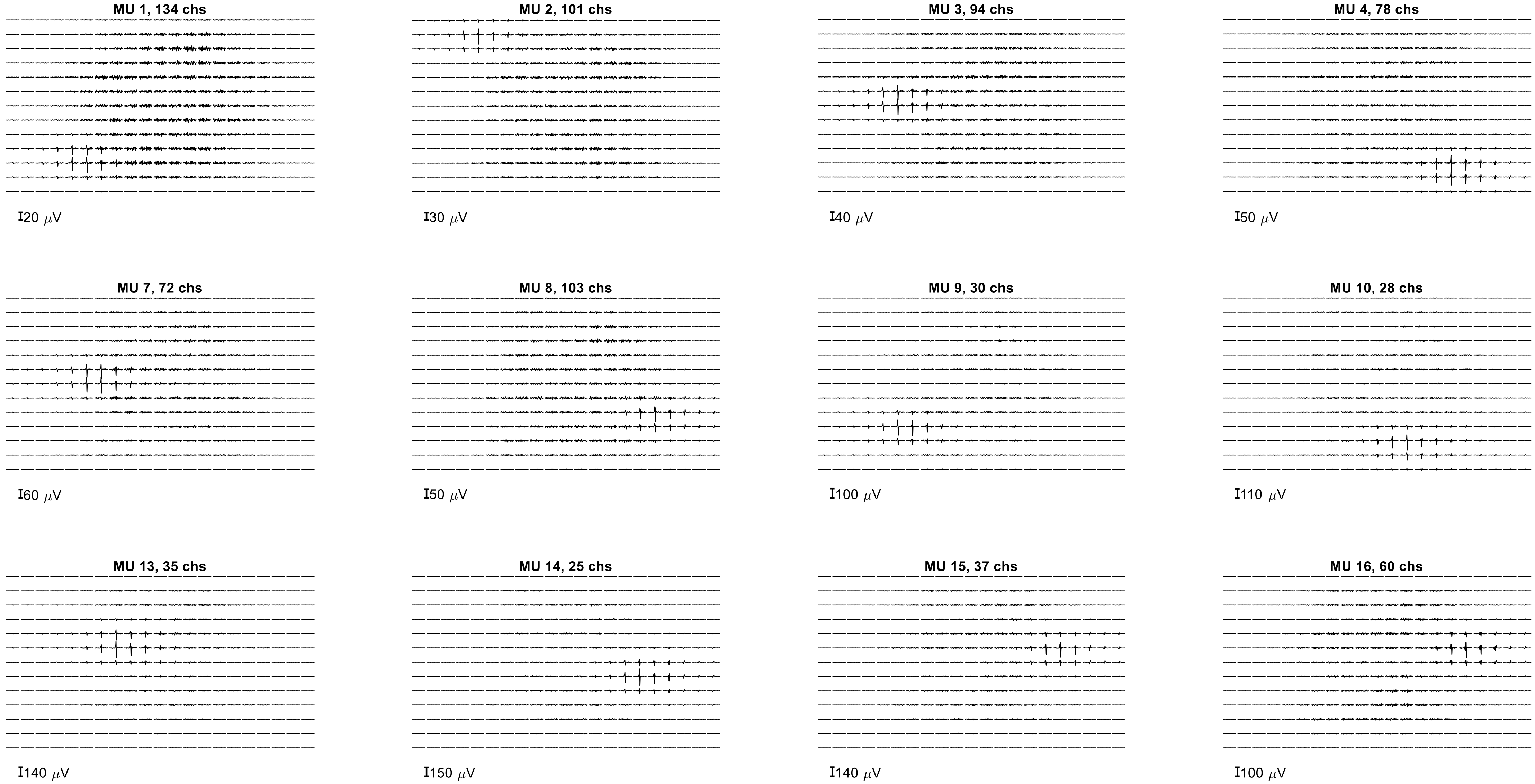

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Convolution Kernel Compensation (CKC)

A. Holobar, D. Zazula: Convolution Kernel Compensation, *IEEE Trans. Signal Proc.*, 2007

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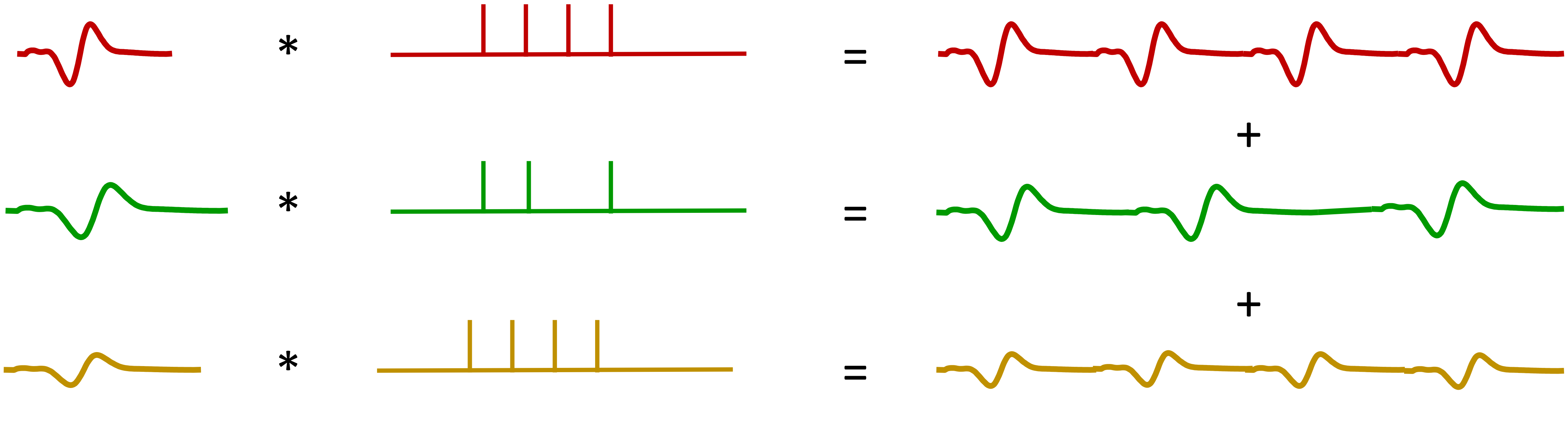
HDEMG model: $\vec{y}(n) = H\vec{t}(n)$



$$= y(n)$$

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Correlation matrixes

Correlation matrix of measurements: $C_y = \sum_n \vec{y}(n)\,\vec{y}(n)$

Correlation matrix of pulse trains: $C_t = \sum_n \vec{t}(n)\,\vec{t}(n)$

$$C_y = \sum_n \vec{y}(n)\,\vec{y}(n) \longleftarrow \vec{y}(n) = H\vec{t}(n)$$

$$C_y = HC_tH^t$$

MU Filter: $y(n_0)^T\, C_y^{-1}$

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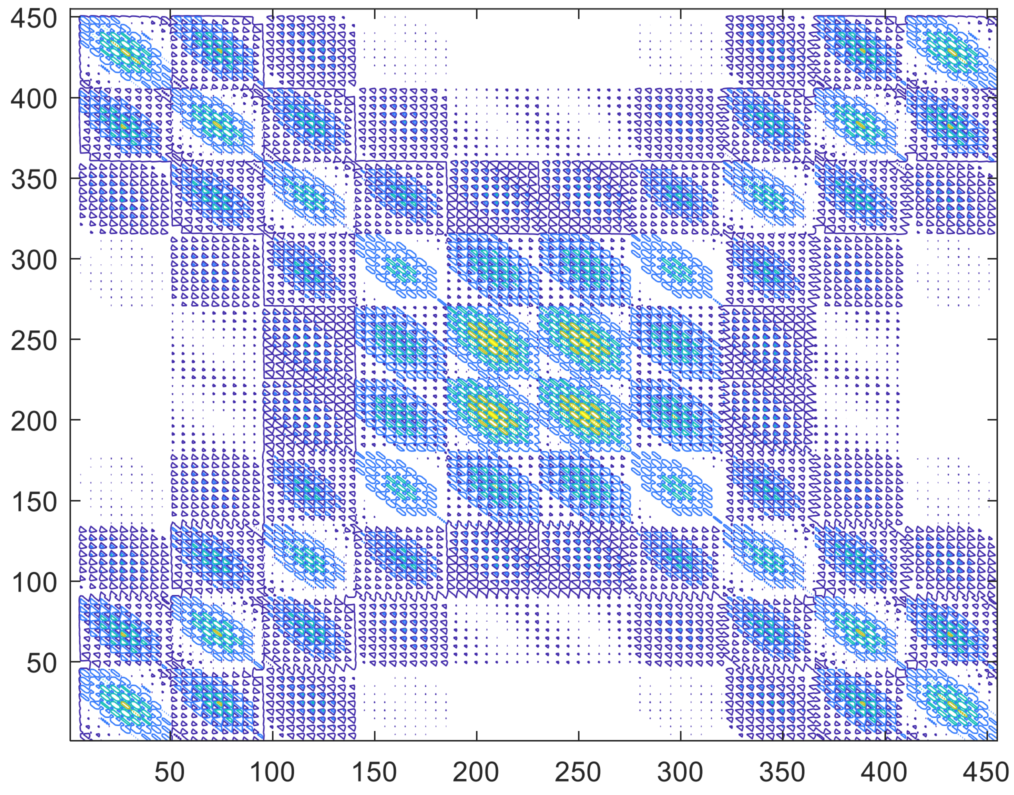
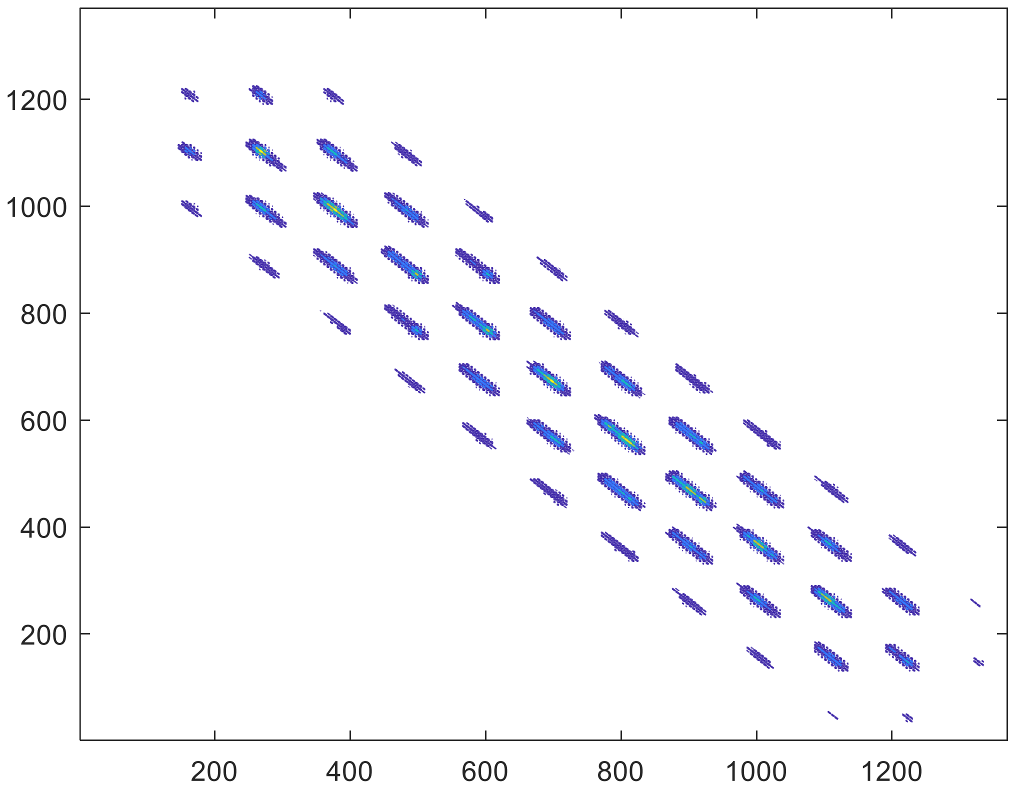
Data model: Correlation matrix of extended measurements

Biceps brachii $\mathbf{C}_{\bar{y}}$



Soleus $\mathbf{C}_{\bar{y}}$



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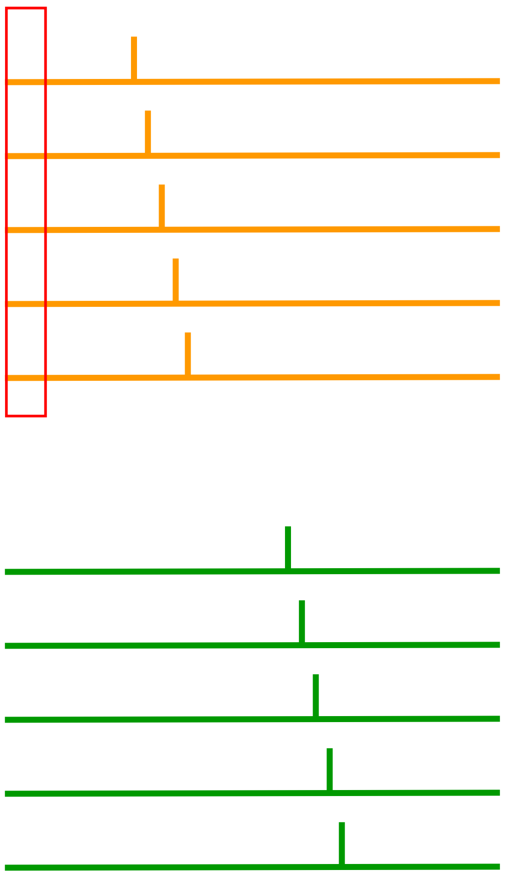
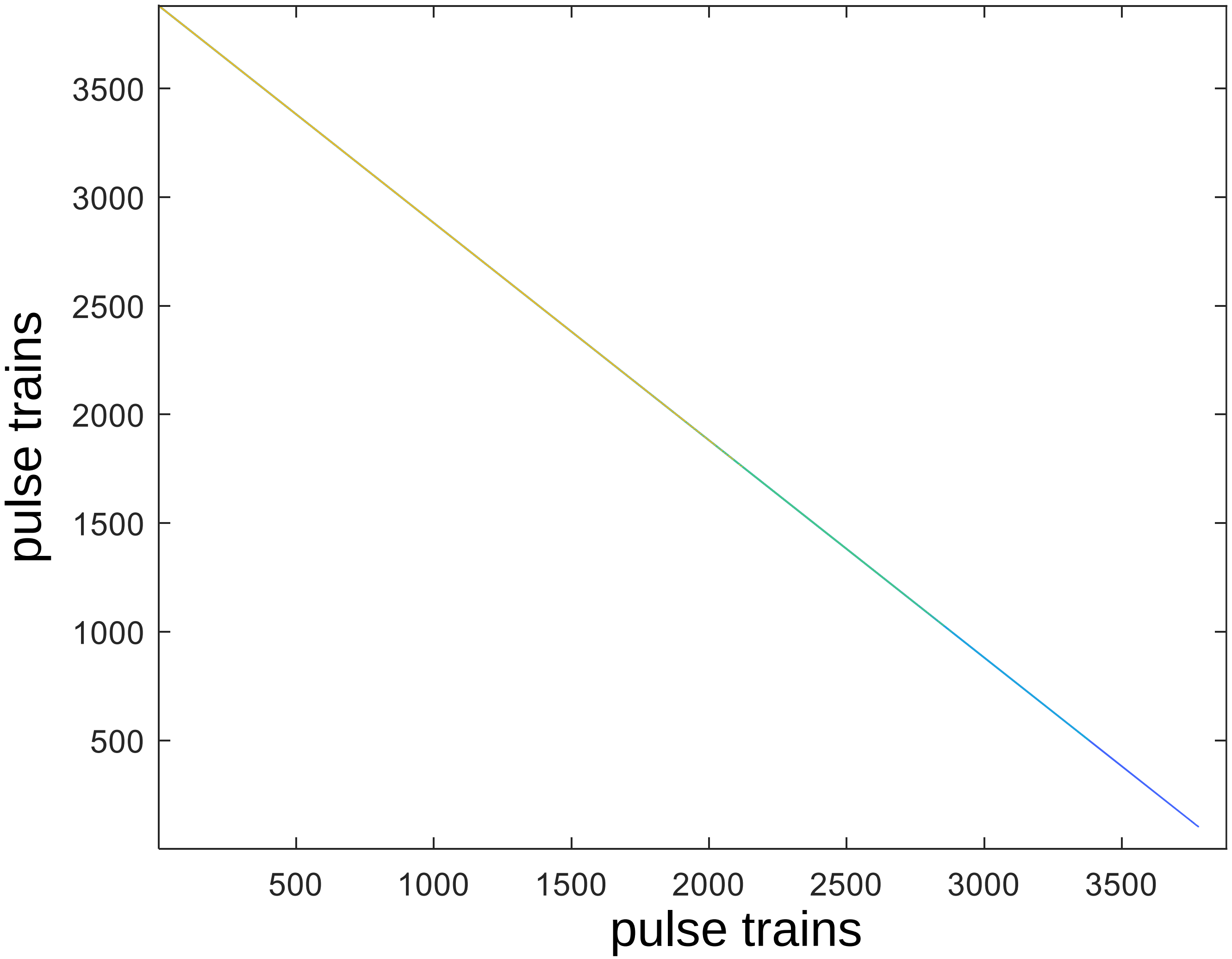
Correlation matrix of pulse trains $C_{\bar{t}}$ is (close to) diagonal





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Convolution Kernel Compensation (CKC)

A. Holobar, D. Zazula: Convolution Kernel Compensation, *IEEE Trans. Signal Proc.*, 2007

HDEMG model: $\vec{y}(n) = H\vec{t}(n)$

HDEMG correlation matrix: $C_y = HC_tH^t$

Example 1: The first approximation of MU filter

MU Filter: $y(n_0)^T C_y^{-1}$

Selector

MU spike train estimation: $y(n_0)^T C_y^{-1} y(n) = \vec{t}(n_0)^T C_t^{-1} \vec{t}(n)$

If MUs are asynchronous: $\vec{t}(n_0)^T \vec{t}(n)$

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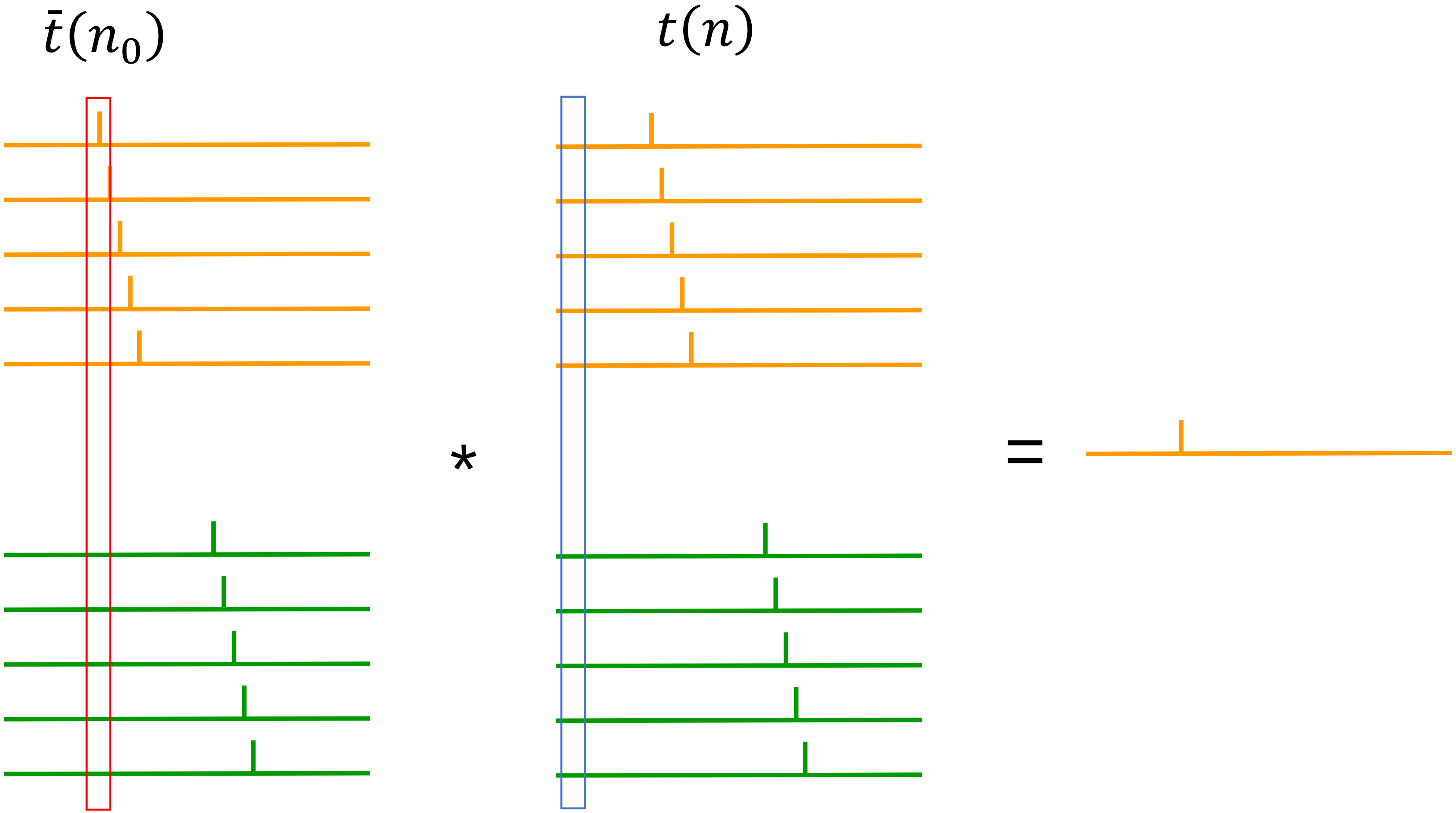
Data model: correlation matrix of pulse trains



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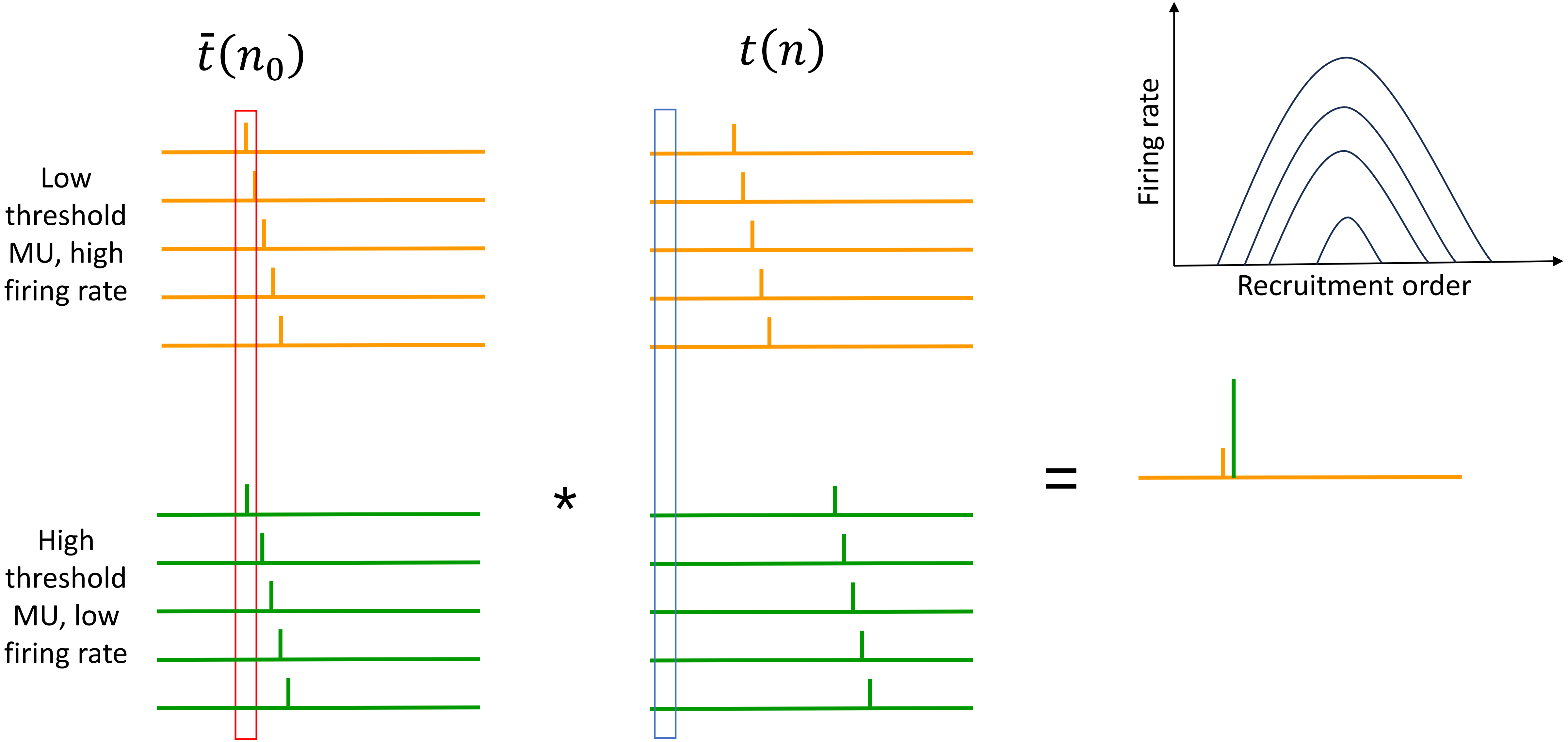
Data model: correlation matrix of pulse trains



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Convolution Kernel Compensation (CKC)

A. Holobar, D. Zazula: Convolution Kernel Compensation, *IEEE Trans. Signal Proc.*, 2007

HDEMG model: $\vec{y}(n) = H\vec{t}(n)$

HDEMG correlation matrix: $C_y = HC_tH^t$

Example 2: The optimal MU filter

MU Filter: $c_{\vec{t_i},y}^{T}\, C_y^{-1}$

MU spike train estimation: $c_{\vec{t_i},y}^{T}\, C_y^{-1} y(n) = c_{\vec{t_i},\vec{t}}^{T}\, C_t^{-1}\, \vec{t}(n)$

If MUs are asynchronic: $c_{\vec{t_i},\vec{t}}^{T}\, \vec{t}(n)$

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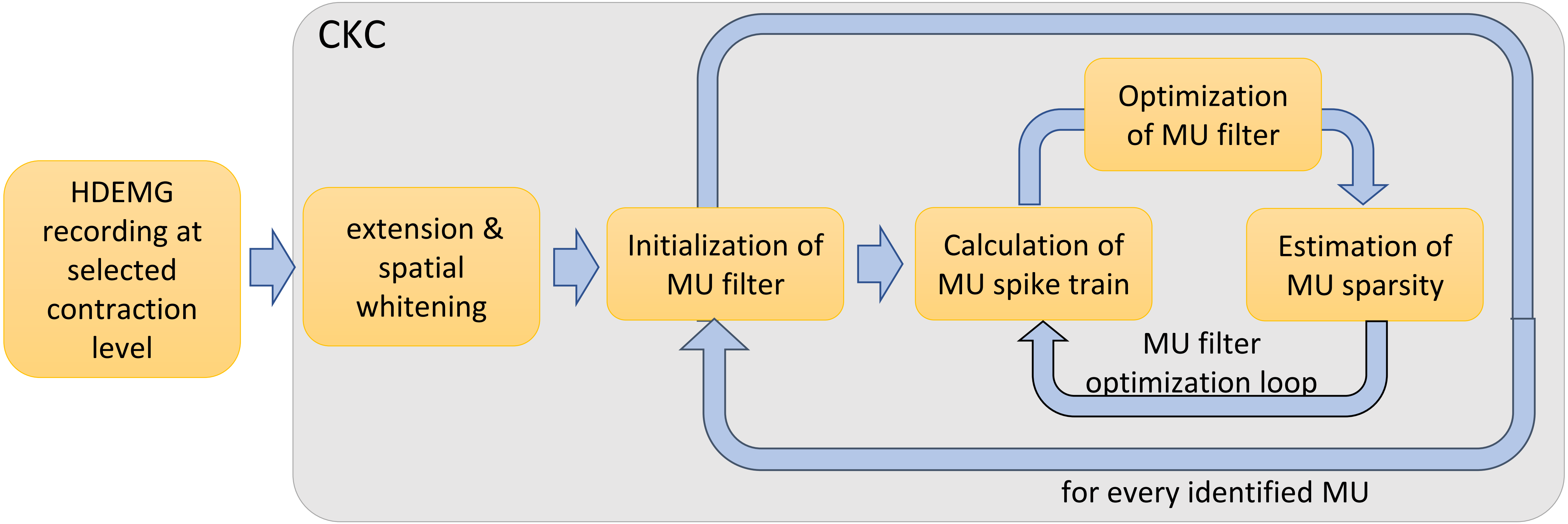
MU filter estimation & optimization



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Processing pipeline

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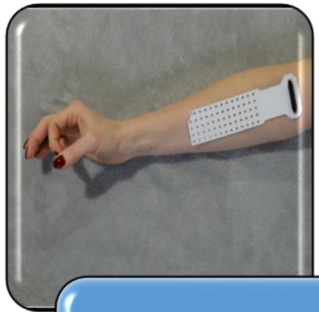
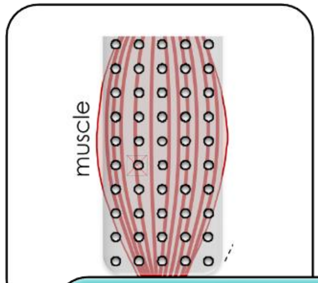
Segment A



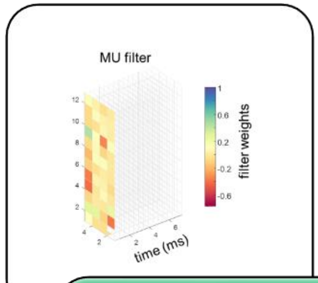
HDEMG acquisition

Segment B



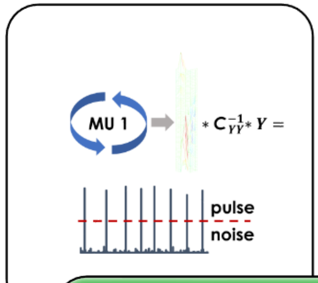
HDEMG channel selection

Segment C



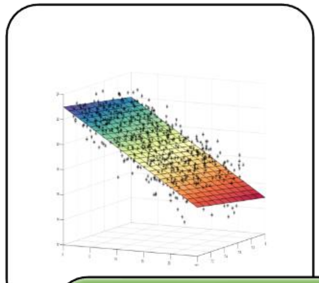
BSS-based MU filter estimation

Segment D



MU filter optimization & firings segmentation

Segment E



Physiological assessment

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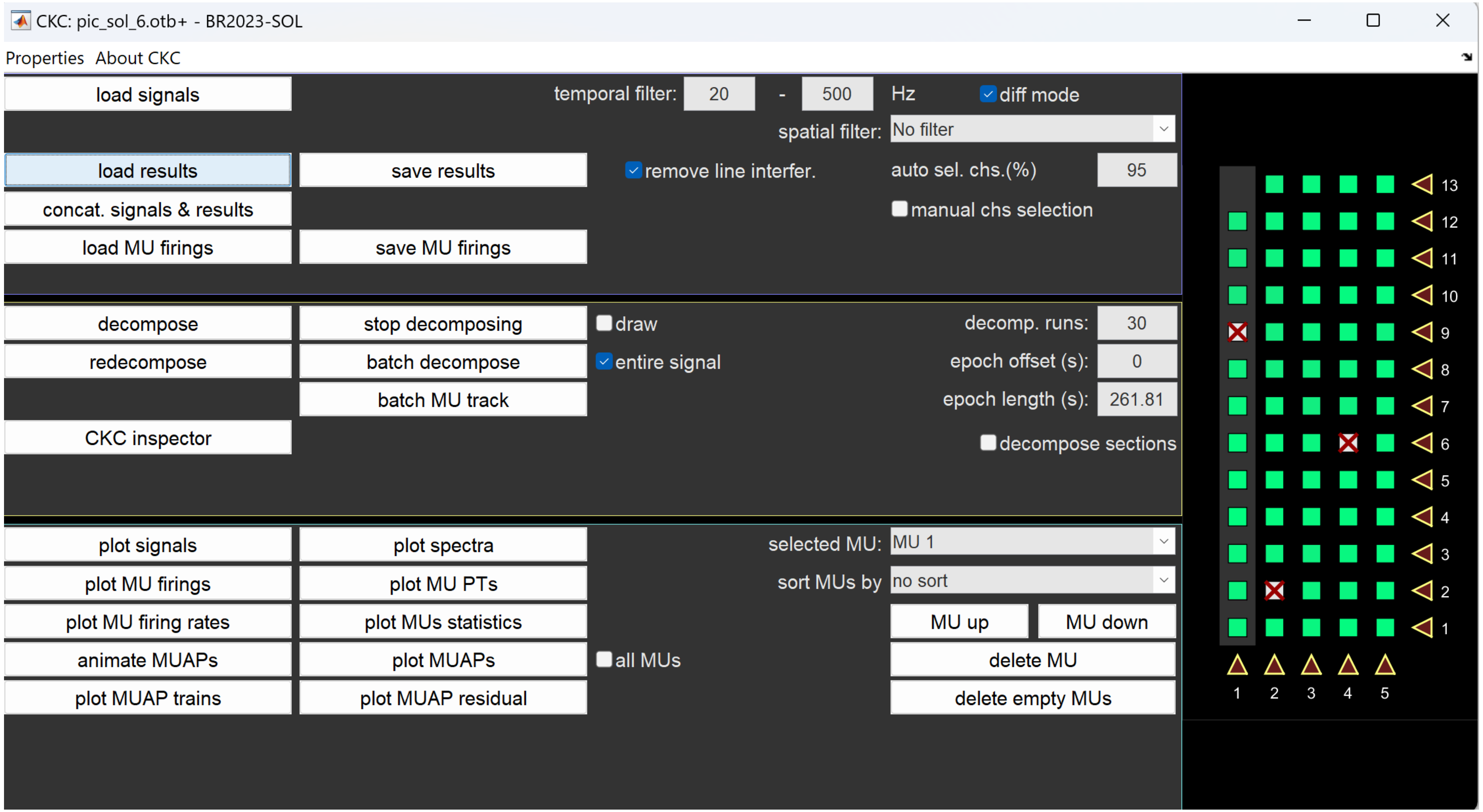
DEMUSE tool



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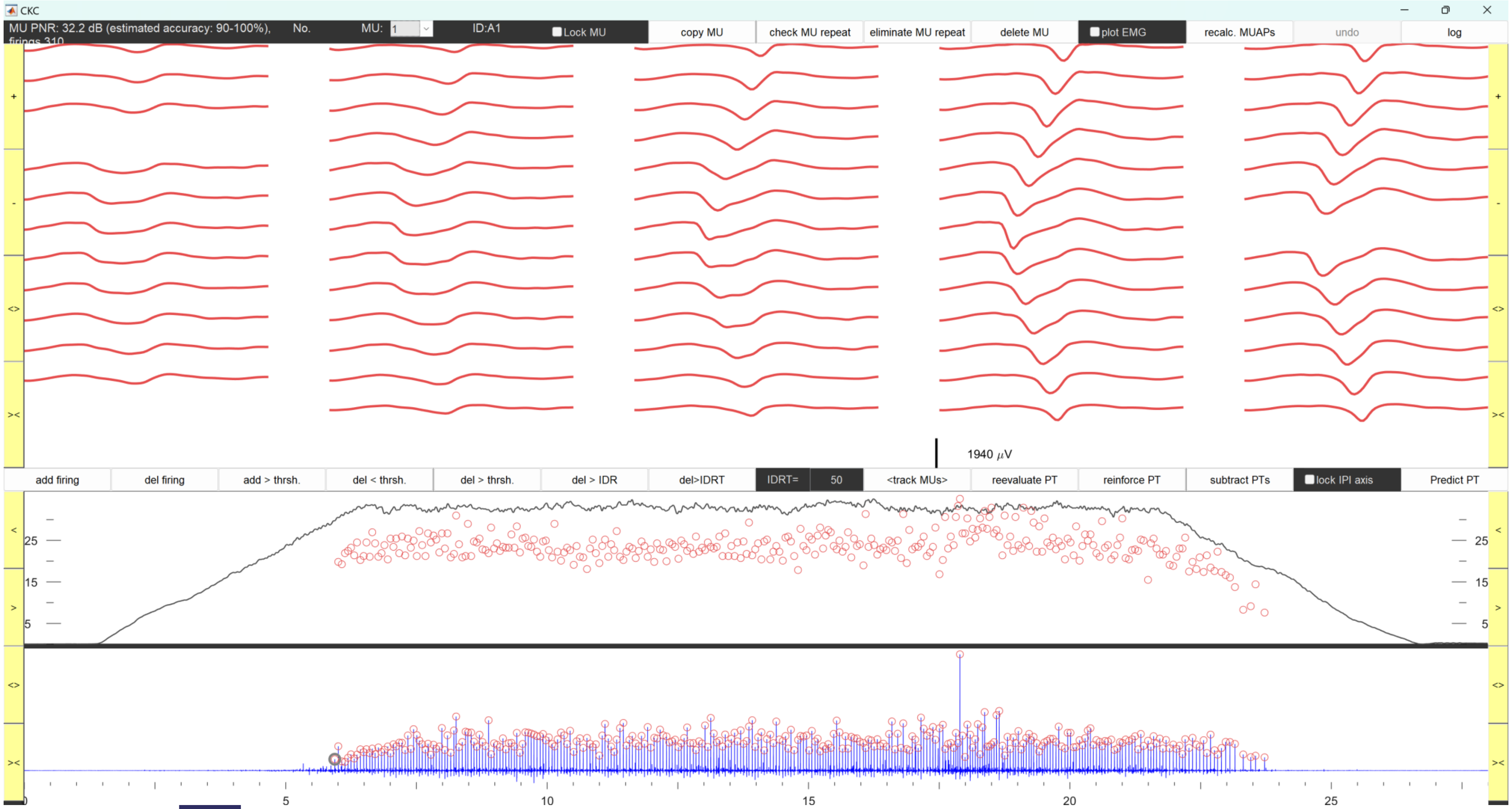
CKC inspector



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MU spike outliers removal

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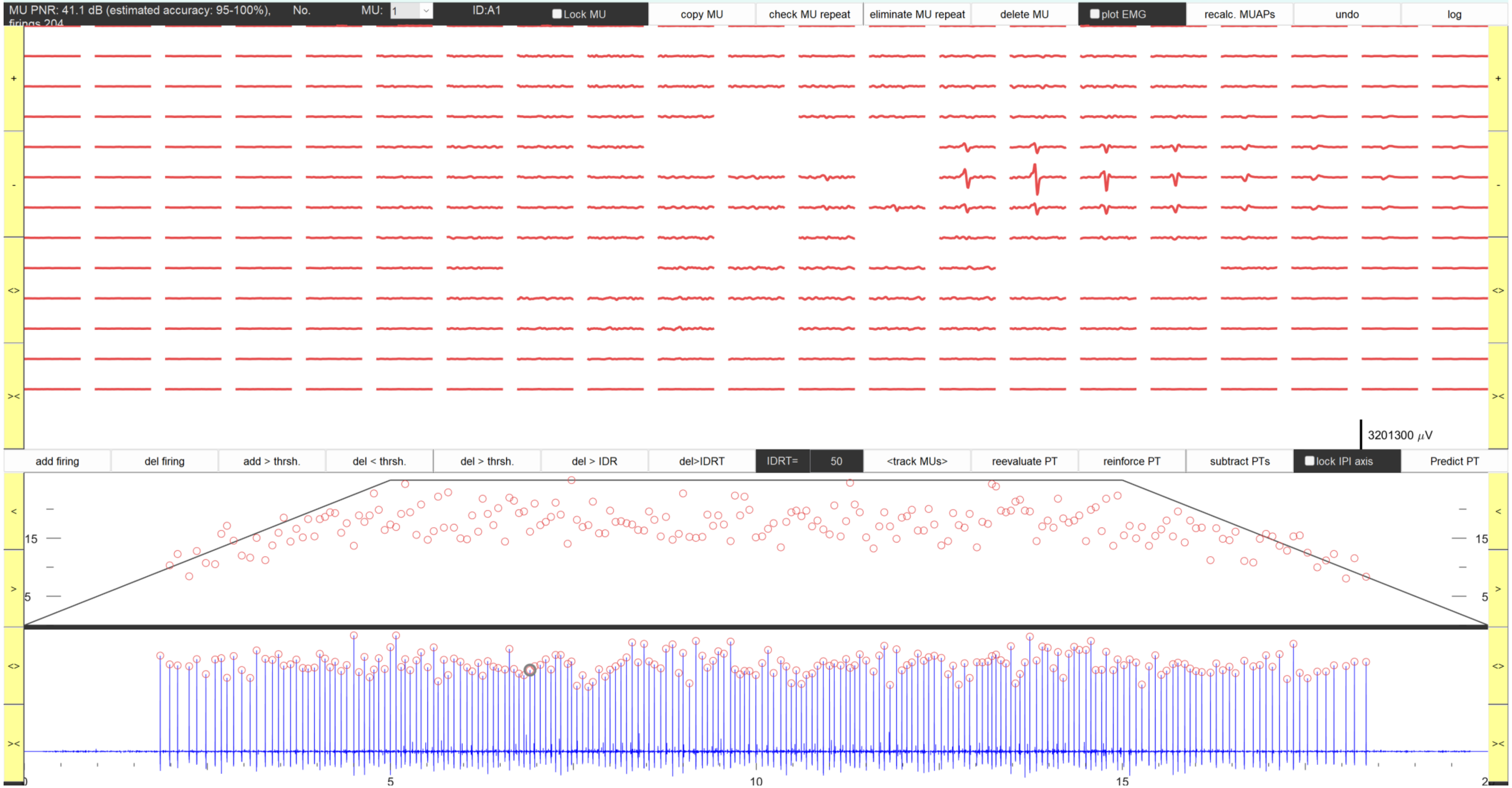
MU spike outliers removal



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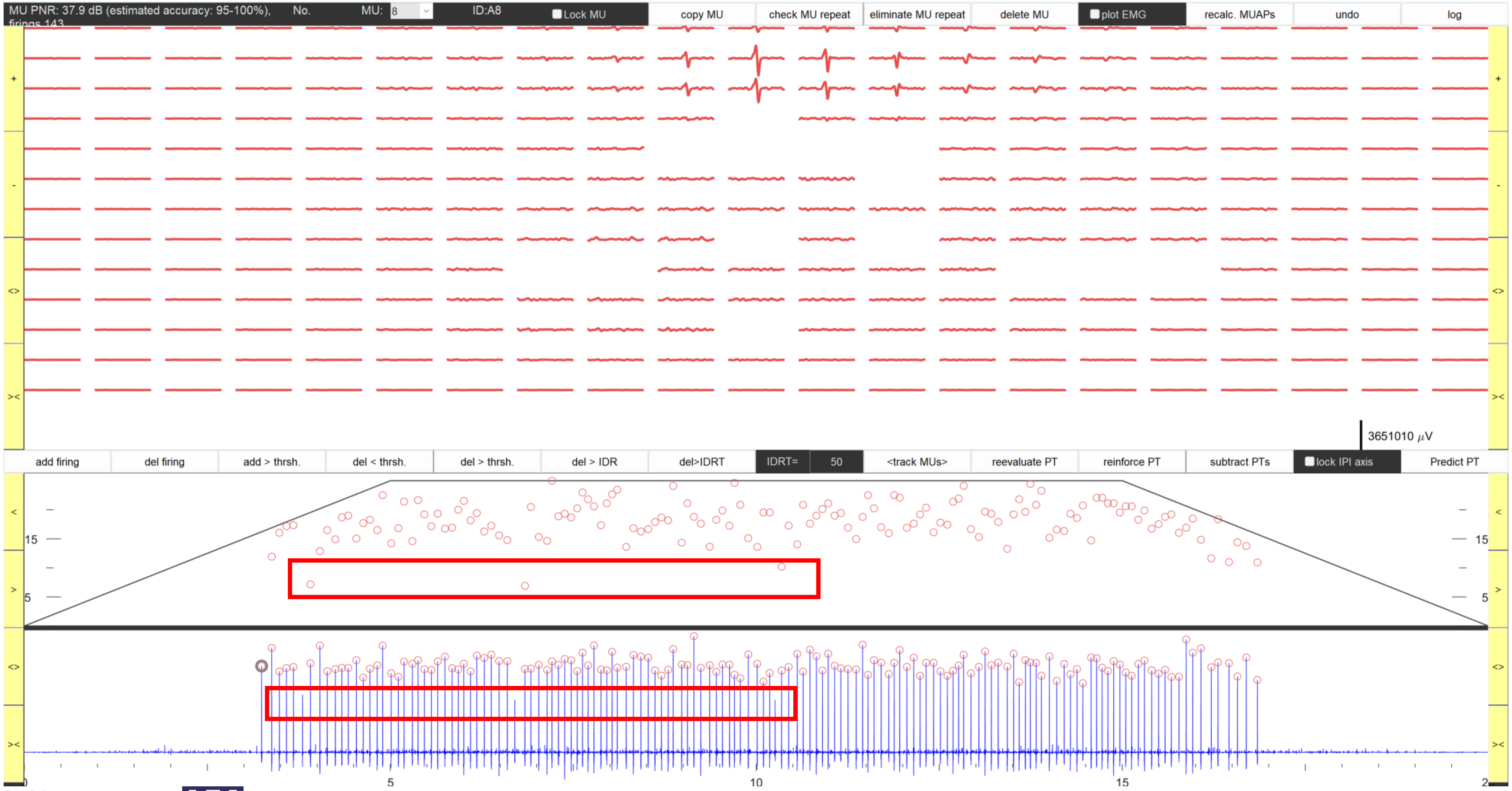
Adding missed firings



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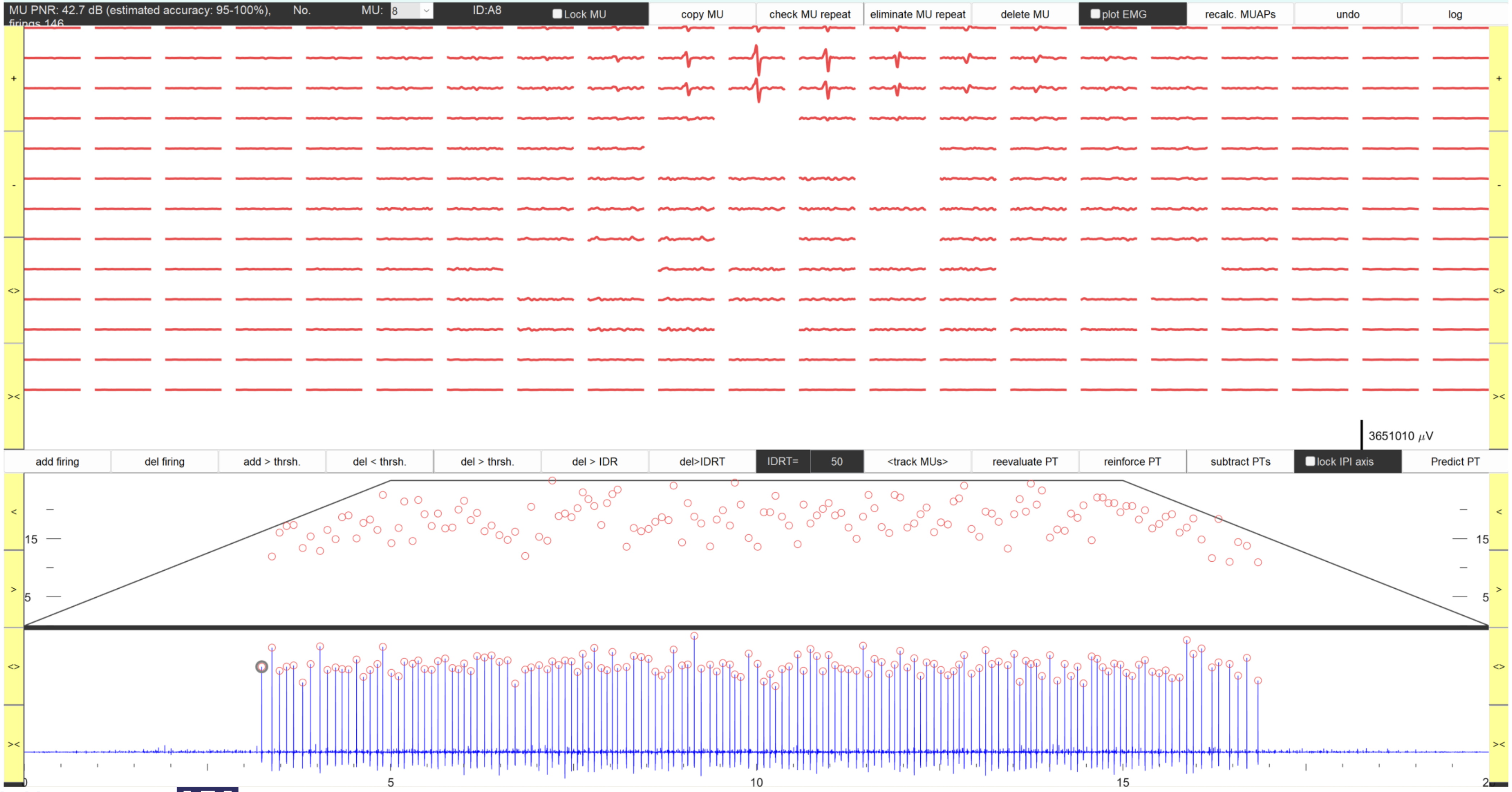
Adding missed firings



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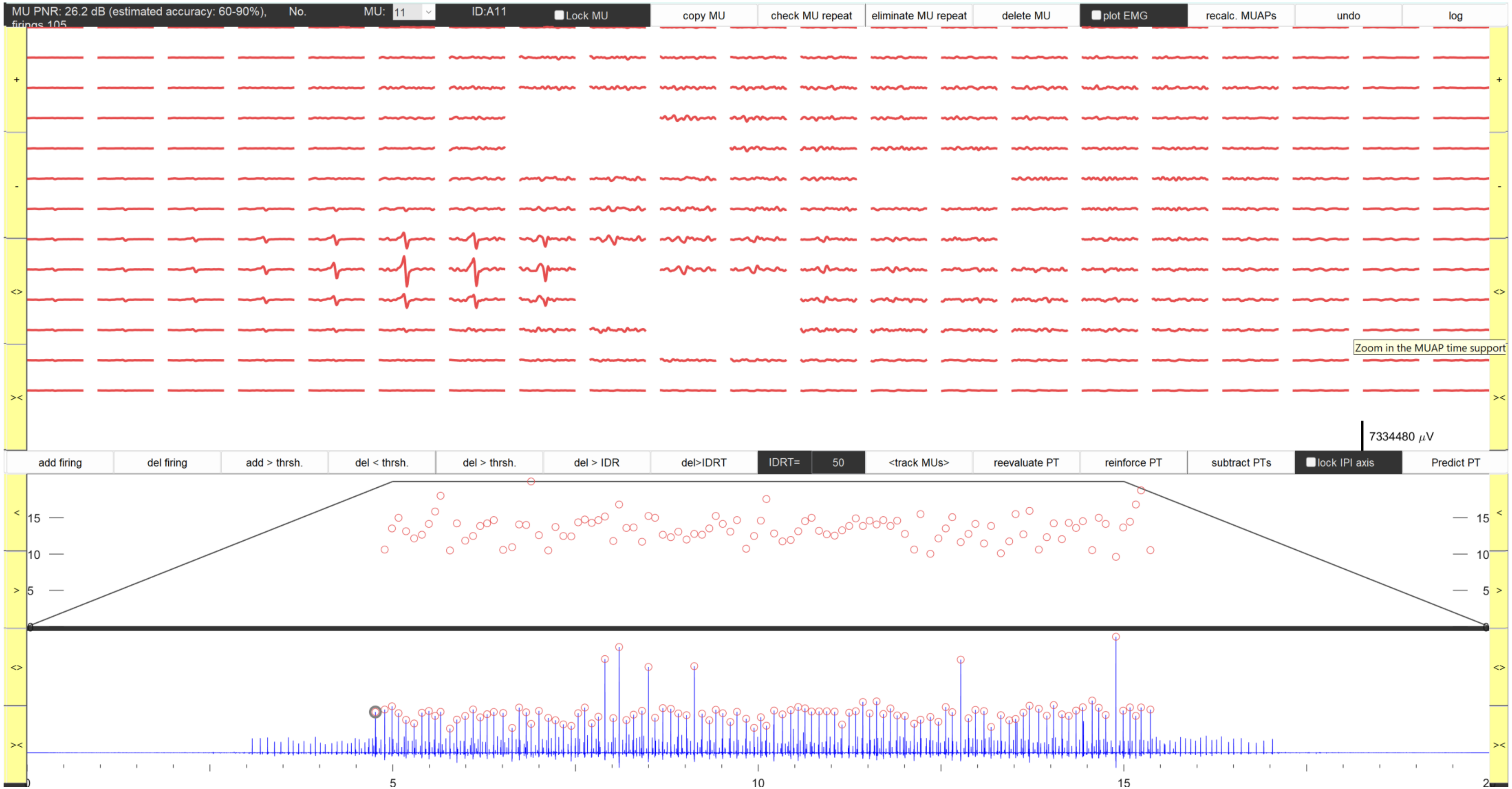
Iterative spike-based MU optimization



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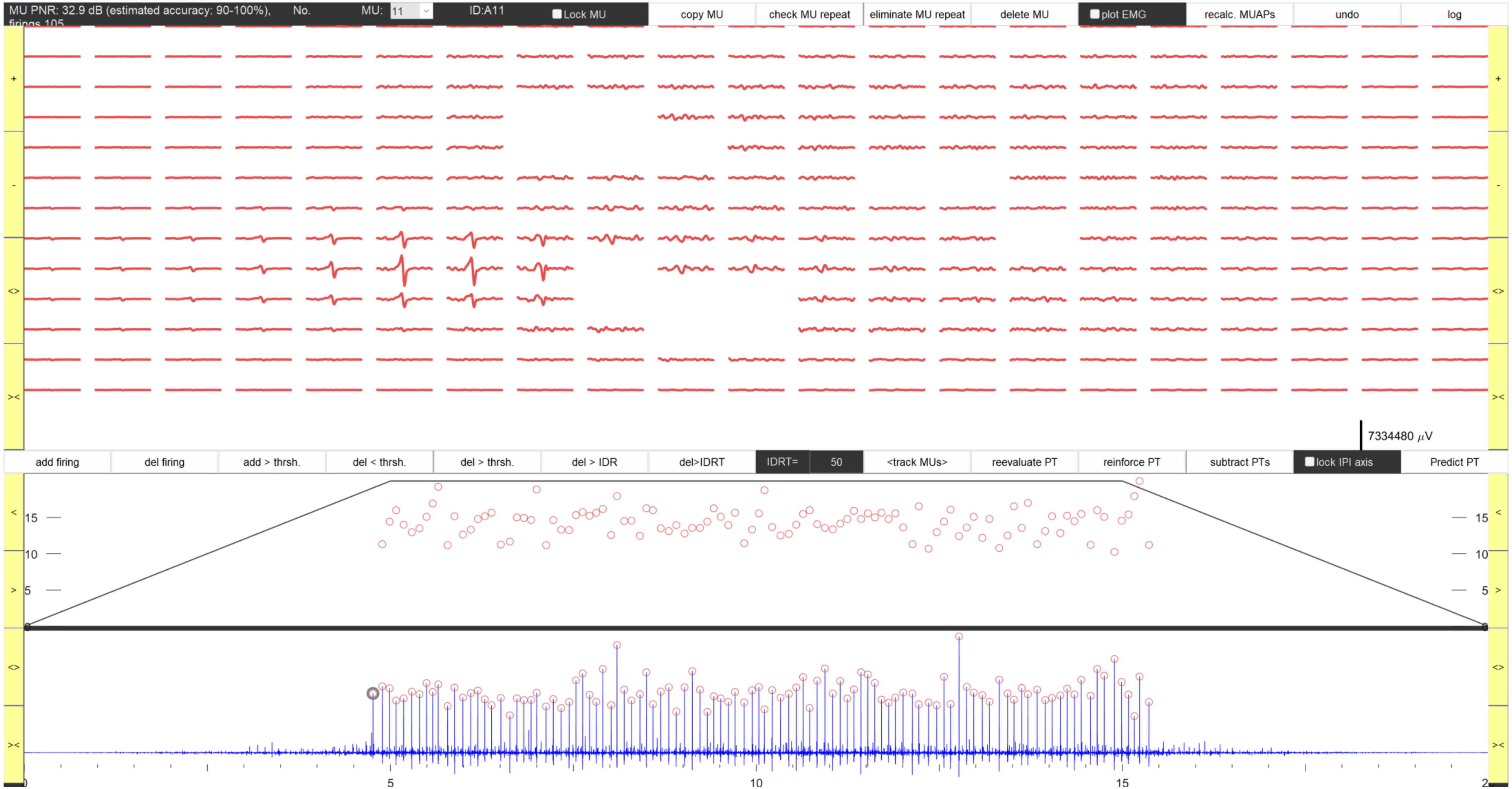
Iterative spike-based MU optimization



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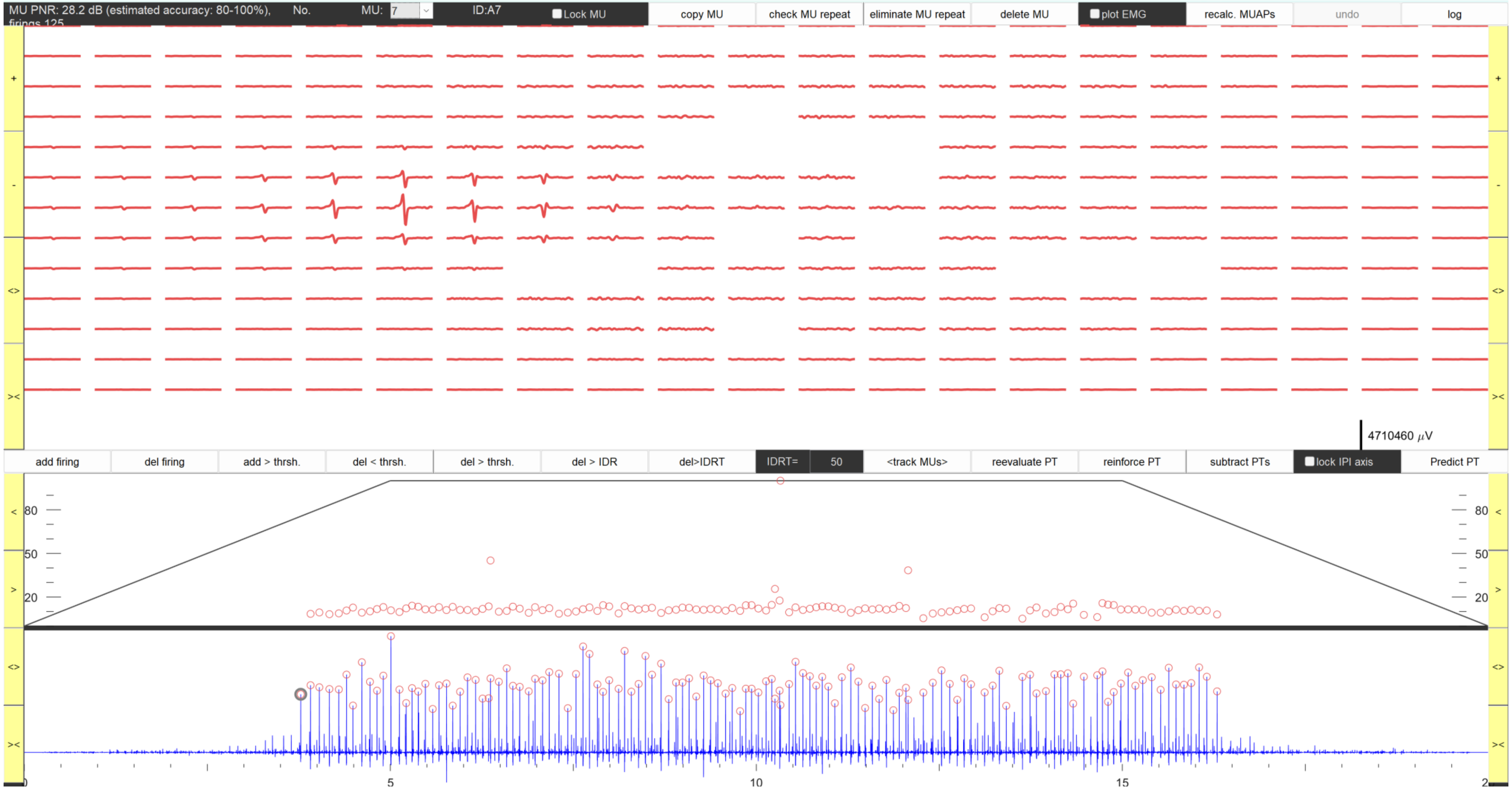
Merged MU



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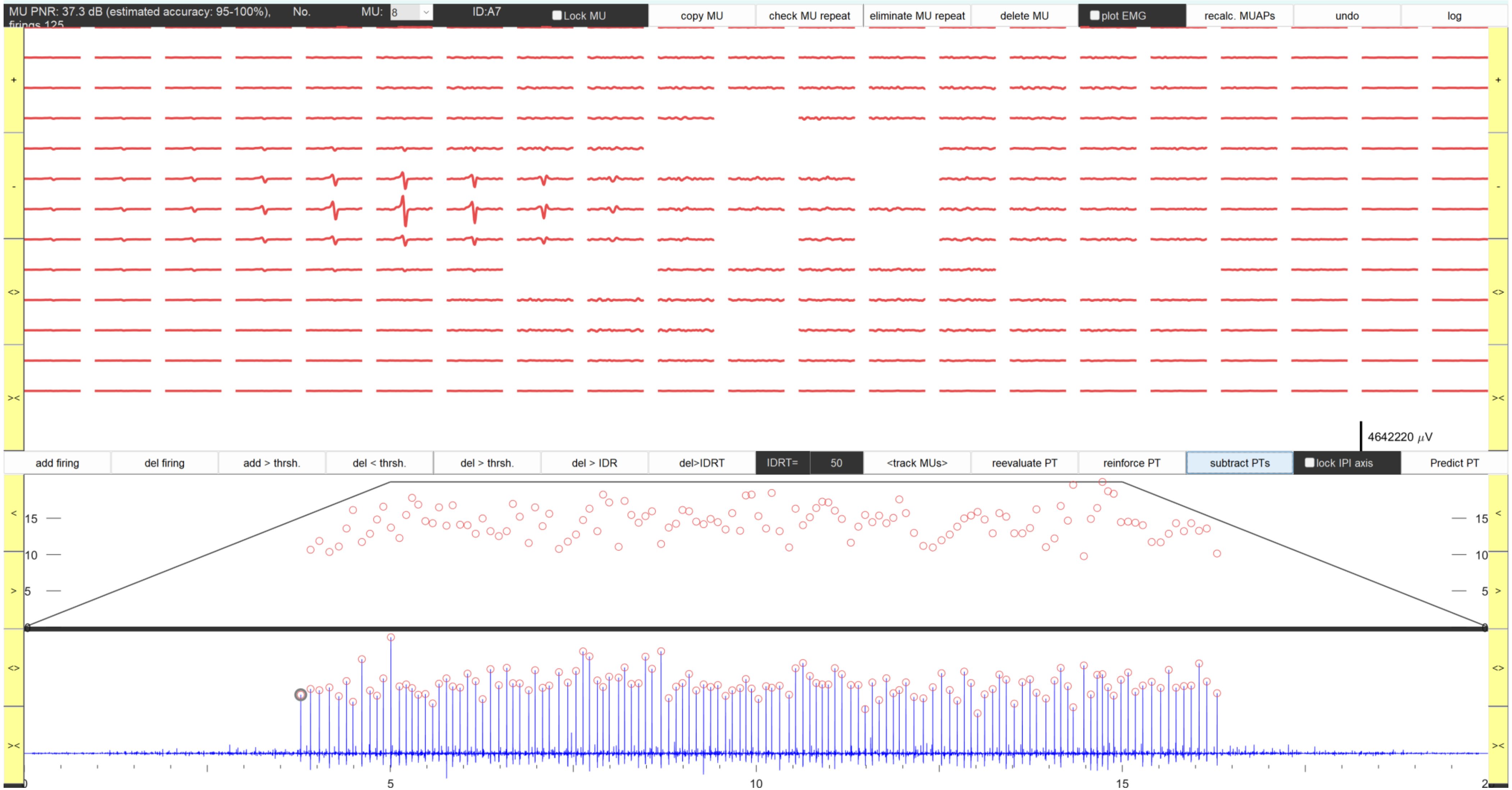
MU 1



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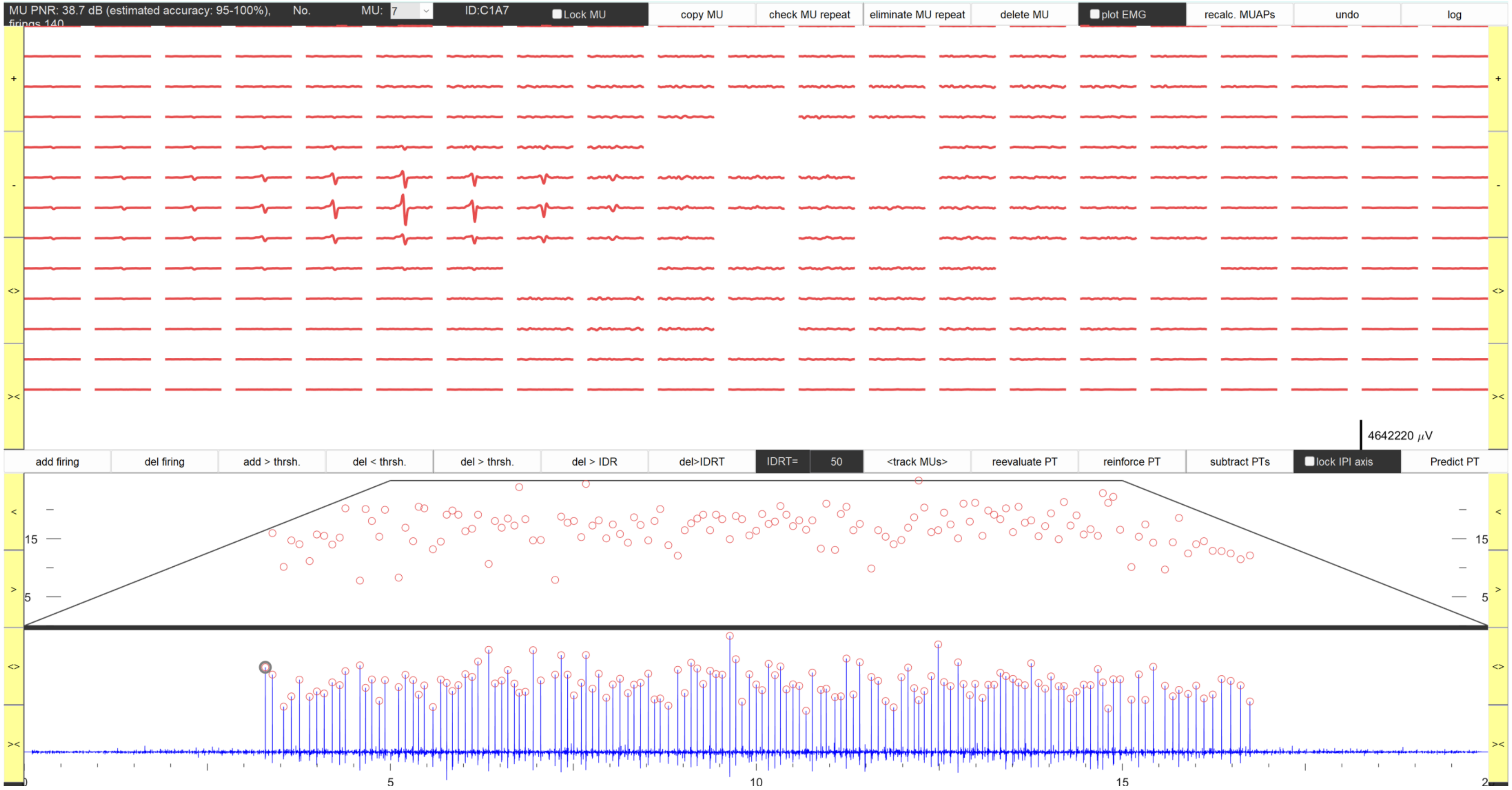
MU 2



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Five steps of consideration for deleting MU

1. PNR < 24 dB.
2. At least 3 different heights in amplitude of PT.
3. Irregular and strange MUAP shapes.
4. The IDR panel doesn’t have a clear firing pattern during different phases of contraction (ramp up, hold, ramp down).
5. MU has < 10 firings.

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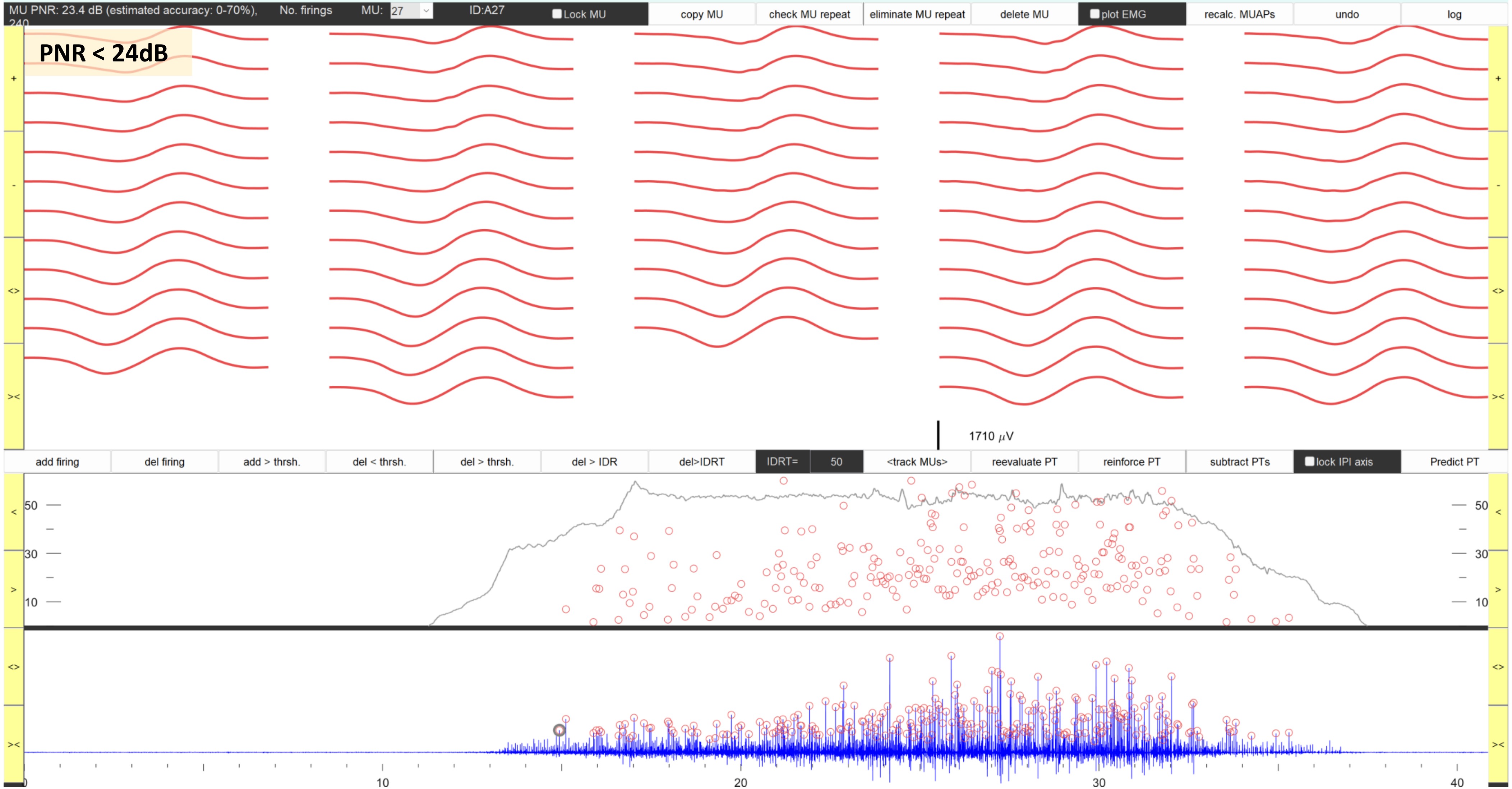
MUs that should be deleted



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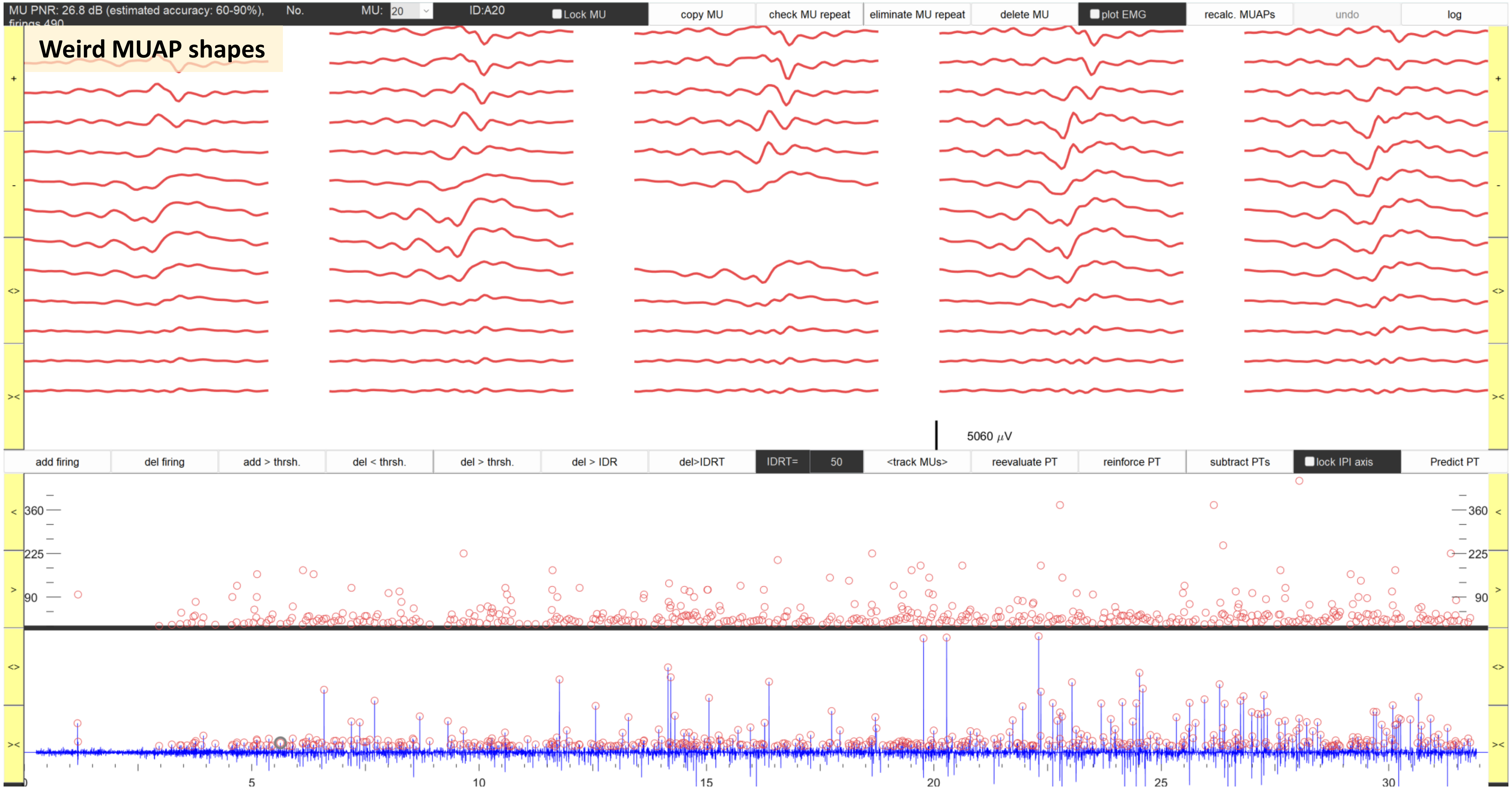
MUs that should be deleted



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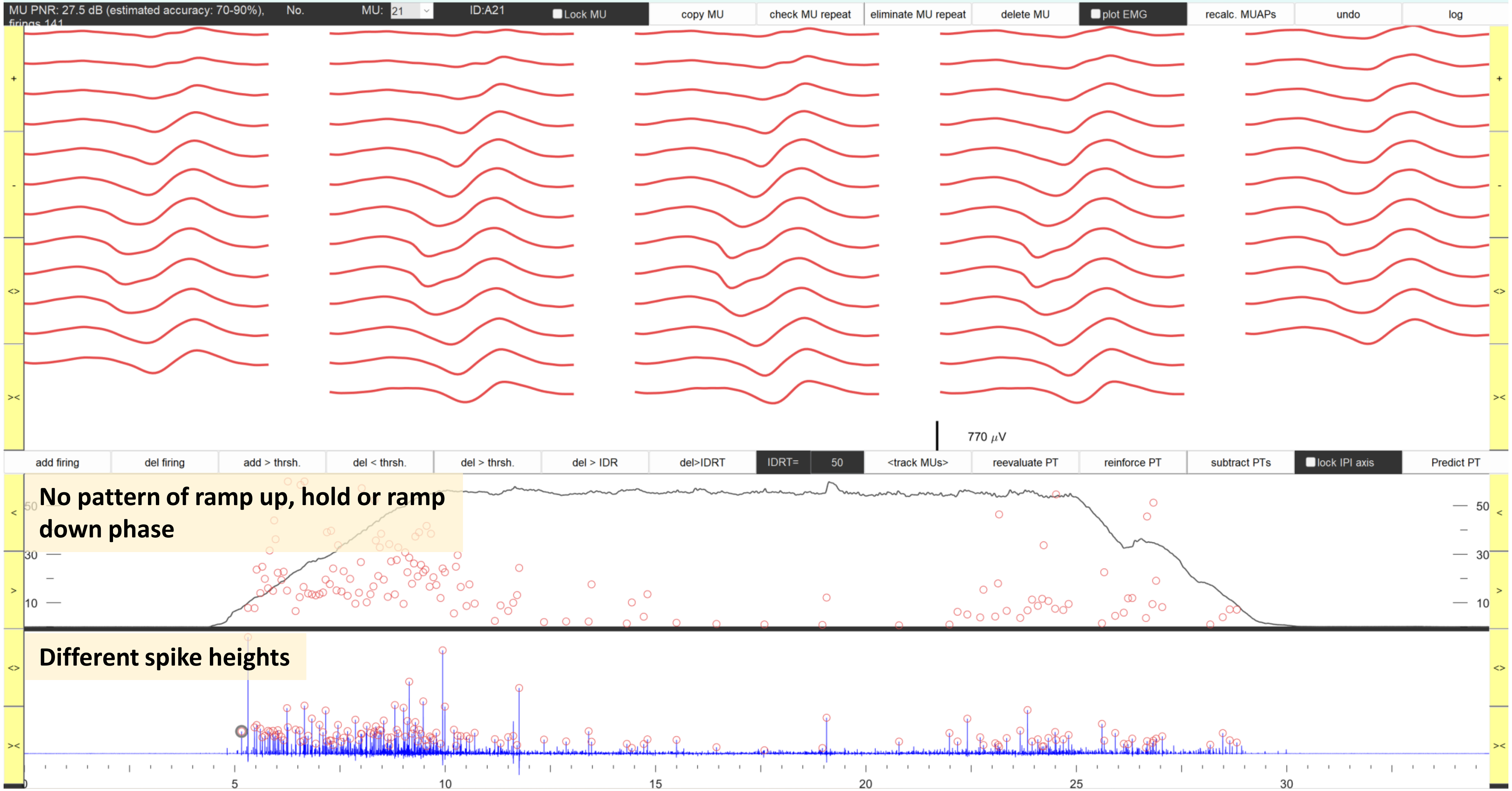
MUs that should be deleted



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Applying MU filter to the EEG (CKC)

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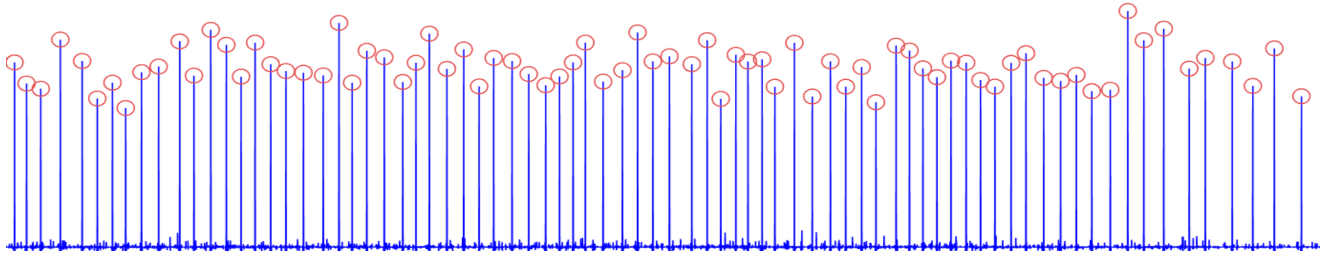
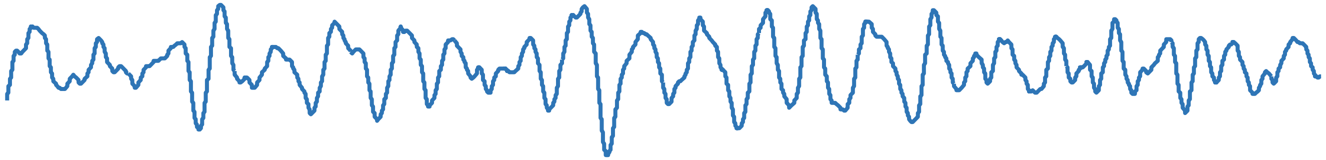
MU spike train estimation: $\hat{t}_i(n) = \mathbf{c}^T_{\bar{t}_i,\, EMG}\, \mathbf{C}^{-1}_{EMG}\, EMG(n)$



EEG component estimation: $\hat{s}_i(n) = \mathbf{c}^T_{cst,\, EEG}\, \mathbf{C}^{-1}_{EEG}\, EEG(n)$



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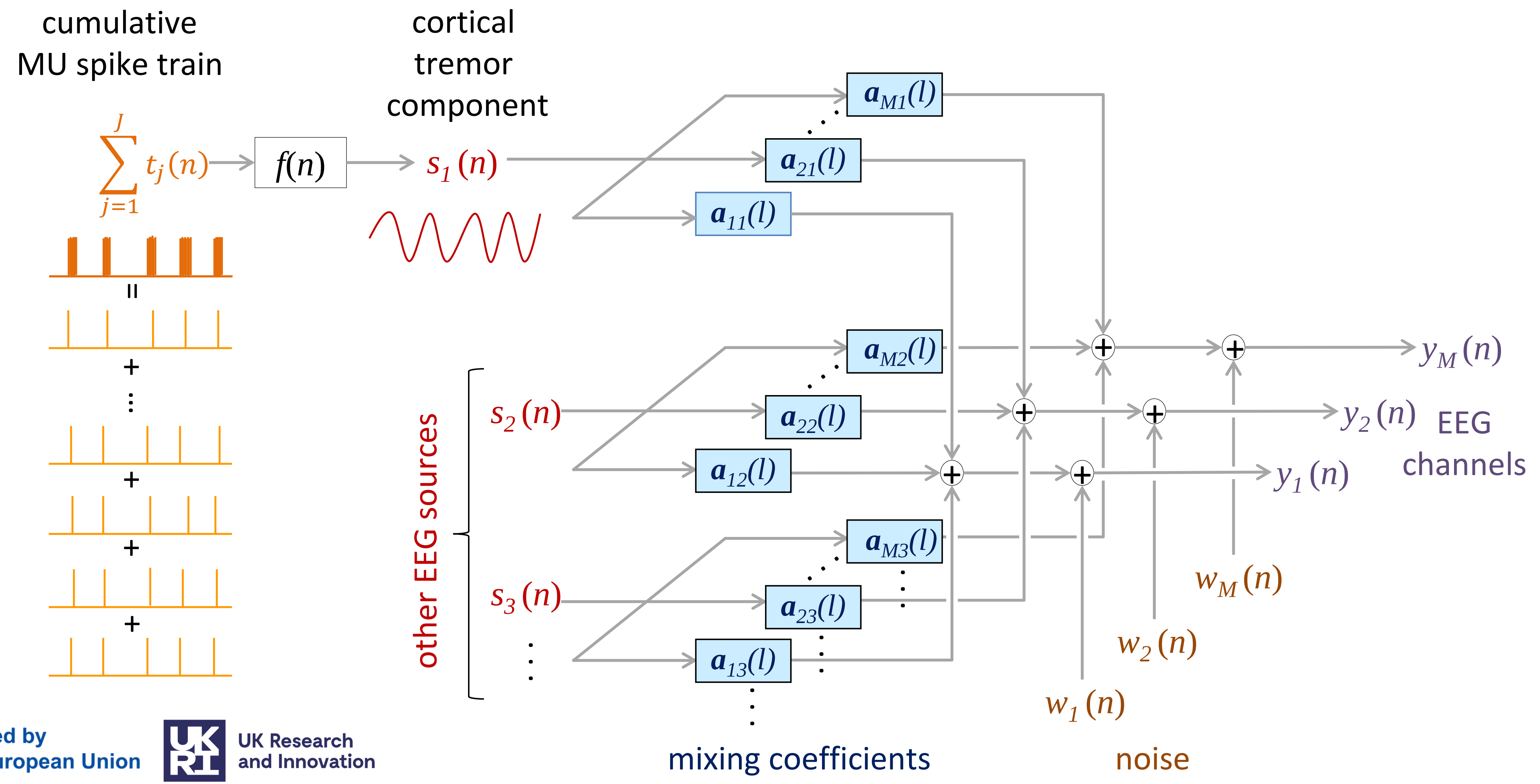
EMG to EEG filter transfer: convolutive EEG model

Holobar et al. Front. Neurol., 2018



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Simulation

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$$t_j(n) = \sum_{k=0}^{K_j} \delta\left(n - k\frac{f_s}{f_j} - d_j - \Delta d_{jk}\right)$$

j=1,...,J,

- d_j is the common firing time lag due to the transmission from cortex,
- Δd_{jk} is MU firing time variability, $\Delta d_{jk} \sim N(0, \sigma_{\Delta d_j})$,
- f_j is the motor unit discharge frequency,
- f_s is the sampling frequency
- K_j is the number of discharges in the observed time interval.

The EEG signals were simulated as comprising ten (J = 10) mutually orthogonal sinusoids $s_j(n)$ and their first higher harmonics as input signals:

$$s_j(n) = \mathrm{a(n)} \cdot \left(B \cdot sin\left(2\pi f_j n - \phi_j\right) + H_1 \cdot sin\left(4\pi f_j n - \phi_j\right)\right),$$

The amplitude B was set equal to 1, whereas the amplitude of the first harmonic, H_1, was varied across simulations and set to 0, 0.2, 0.4, 0.6, 0.8 and 1

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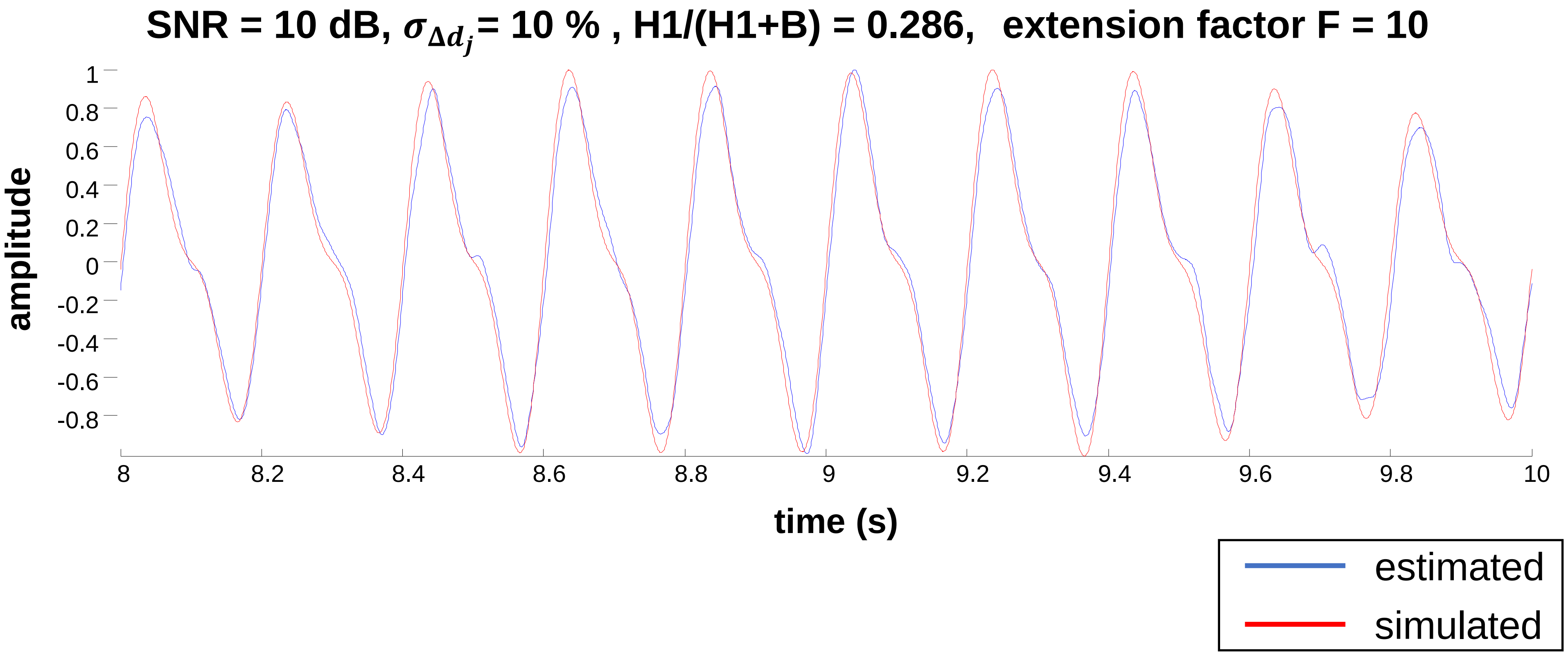
Holobar et al. Front. Neurol., 2018

Simulation



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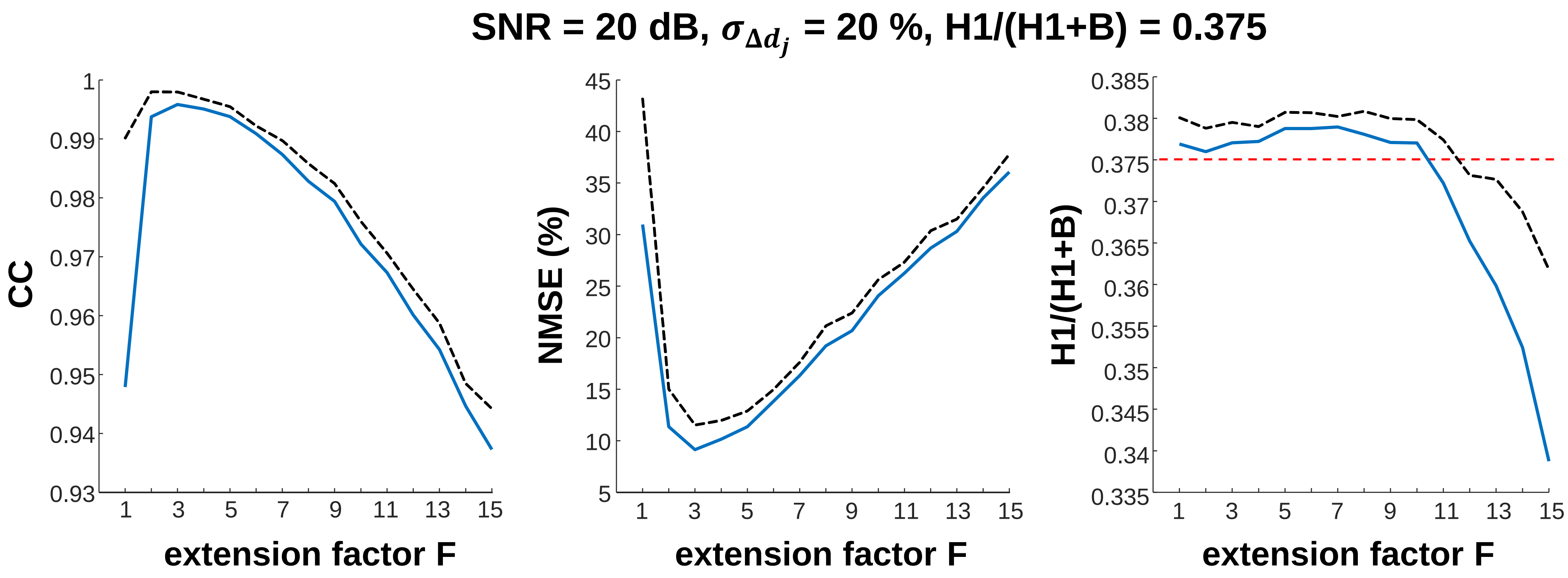
Holobar et al. Front. Neurol., 2018

Simulation



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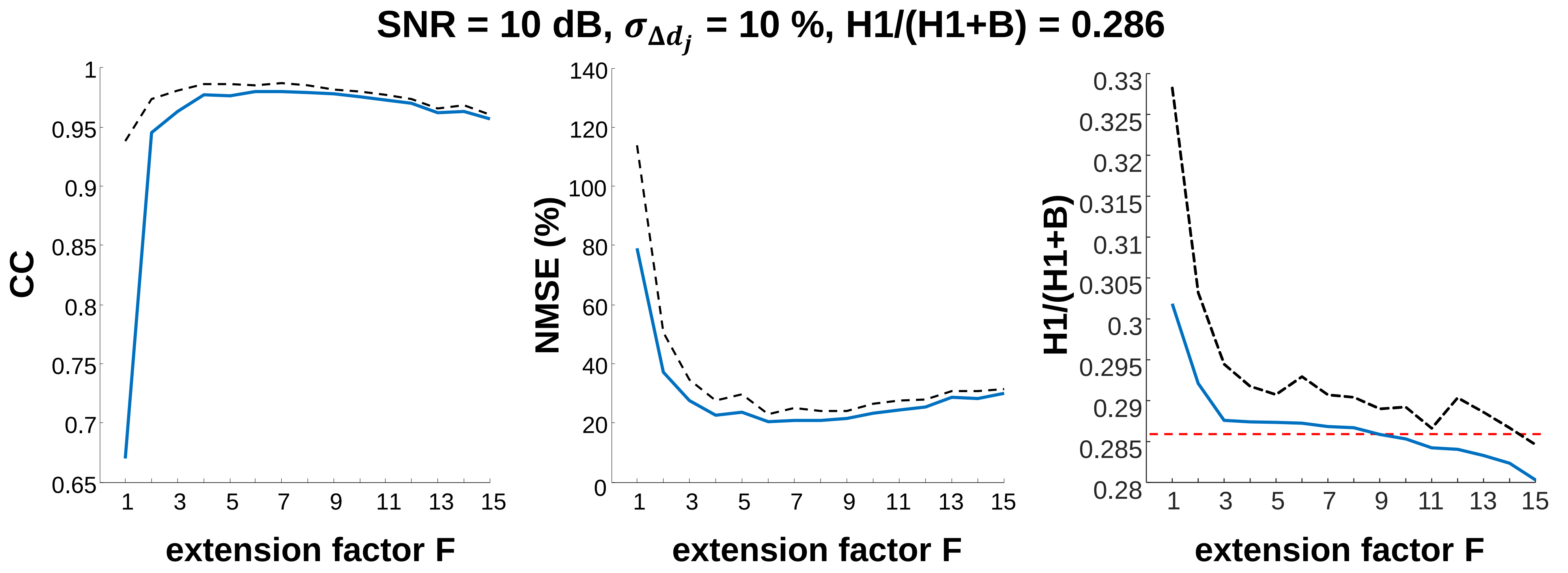
Holobar et al. Front. Neurol., 2018

Simulation



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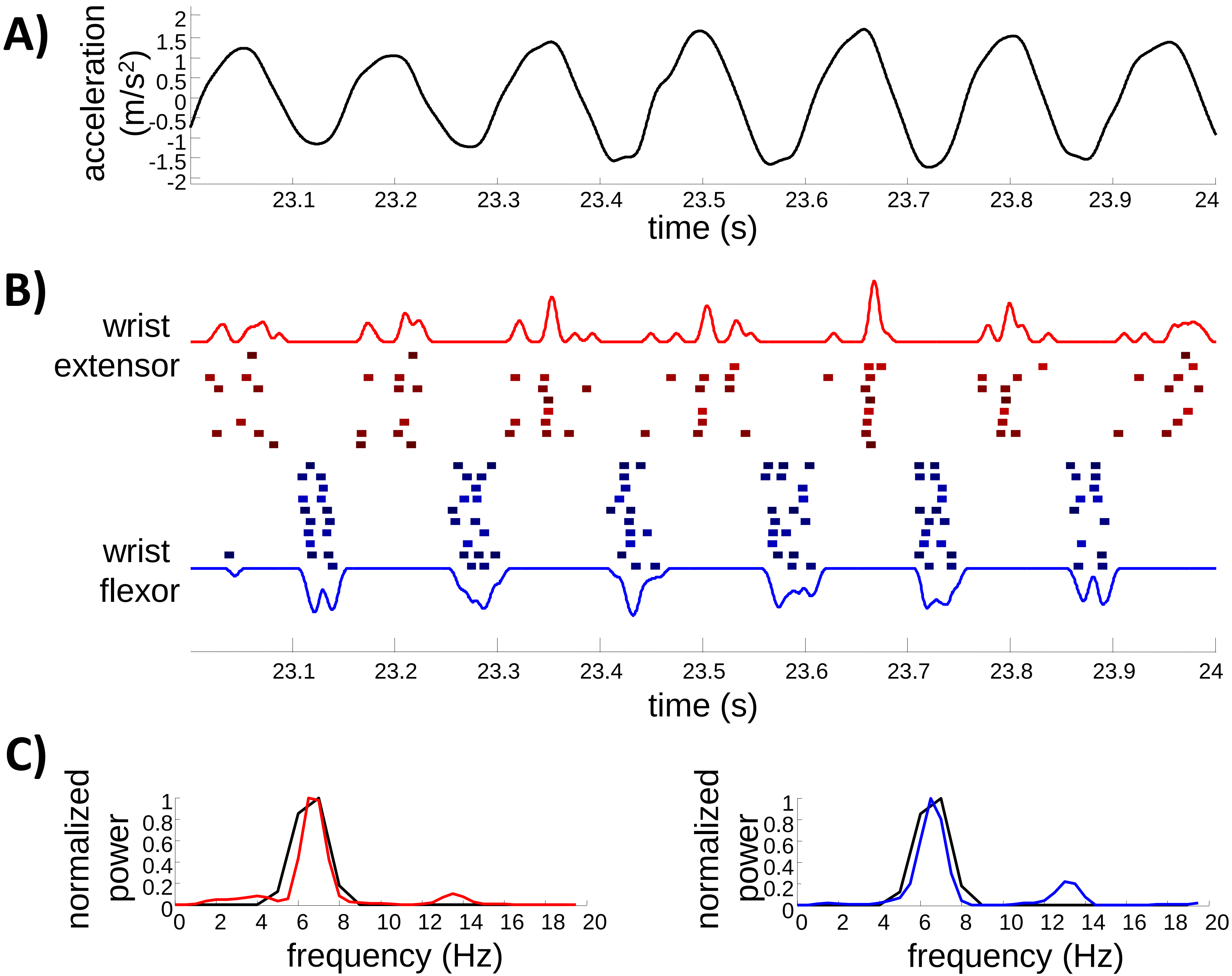
Holobar et al. Front. Neurol., 2018

Experimental conditions

Holobar et al. Front. Neurol., 2018



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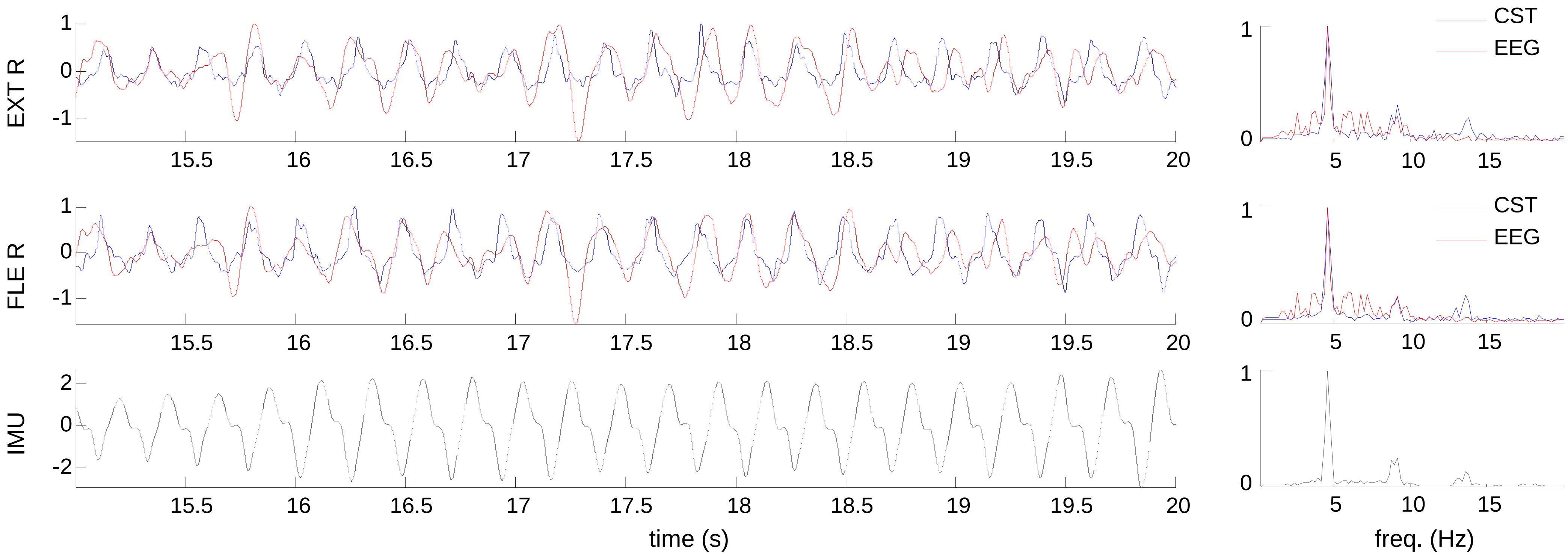
EEG/EMG synchronization



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Holobar et al. Front. Neurol., 2018

EEG-EMG Coherence

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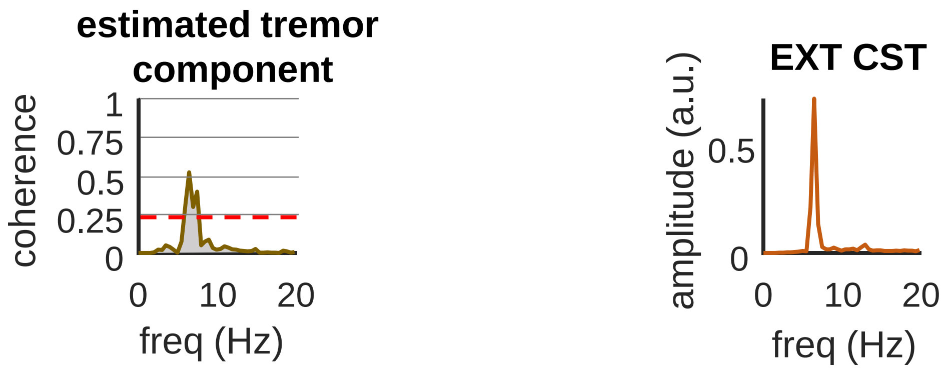

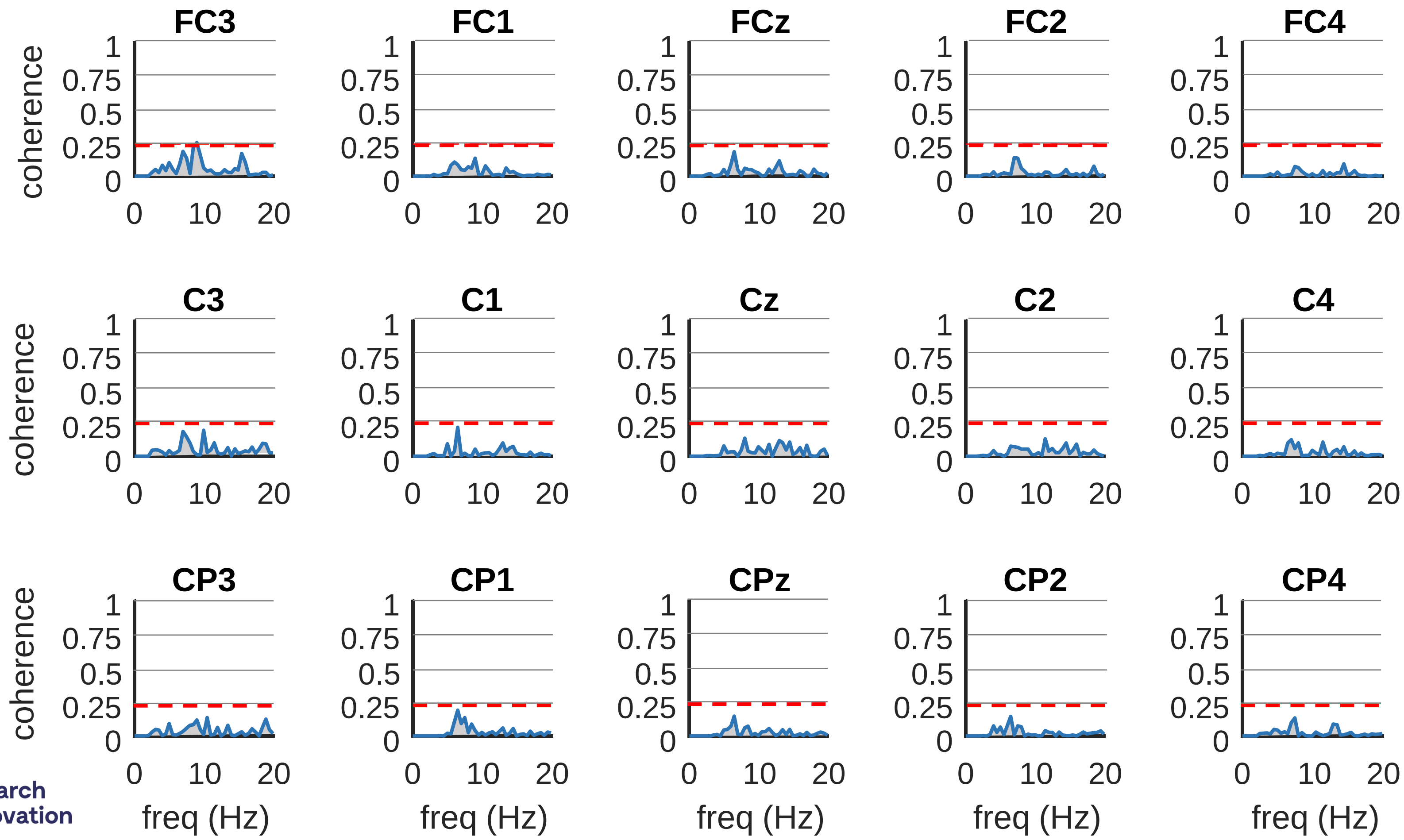

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Inverse tomography and MU related EEG activity

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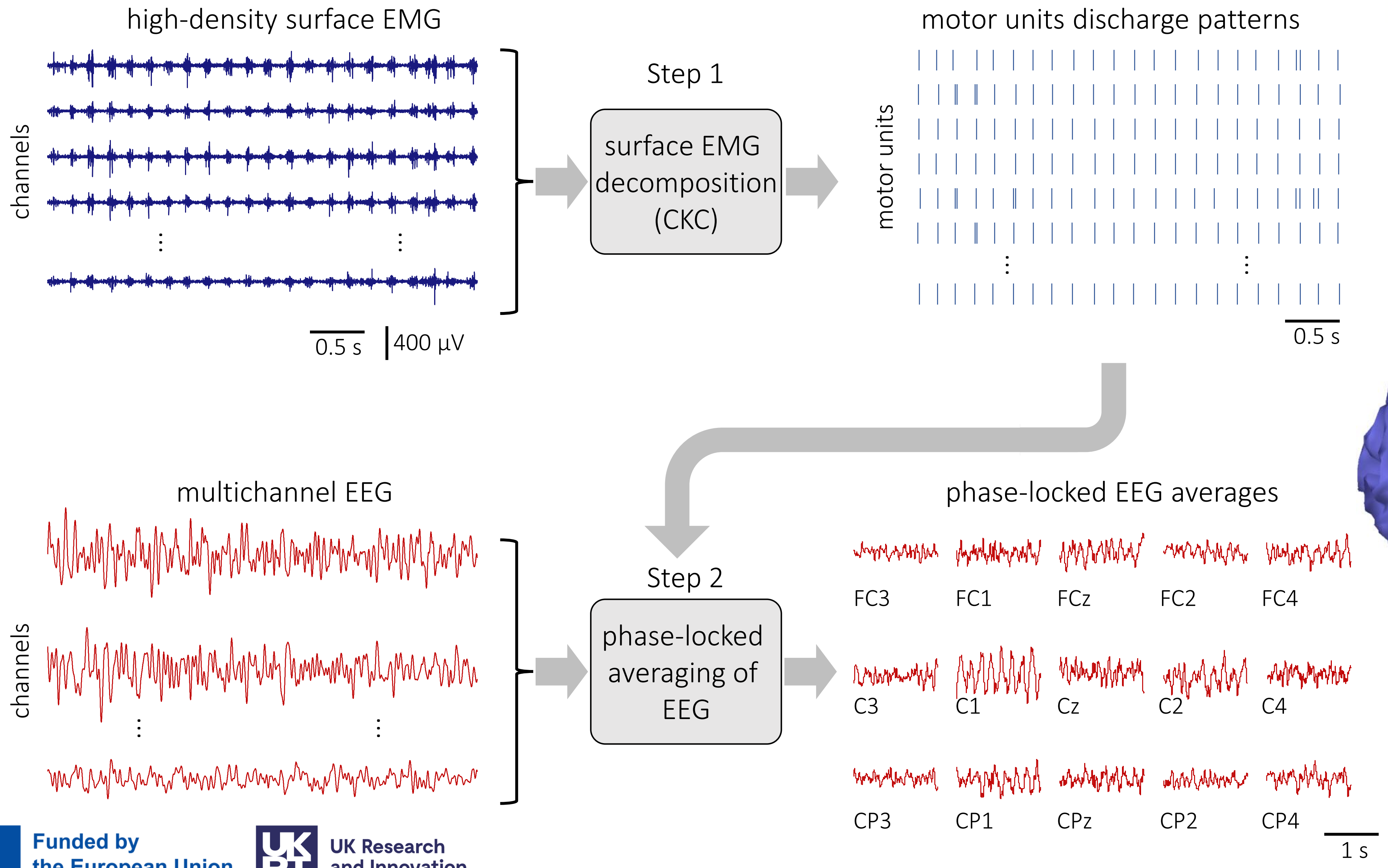

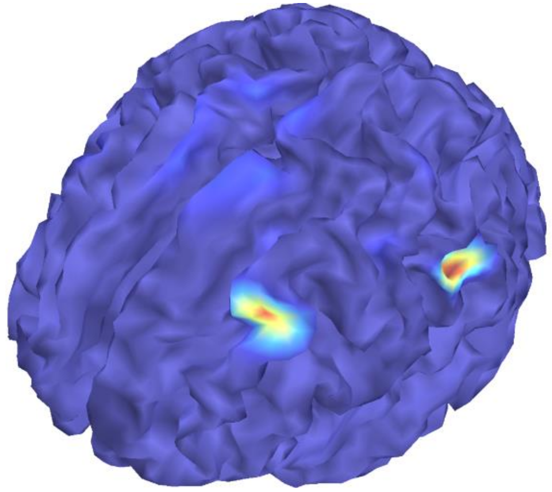

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Holobar et al. Front. Neurol., 2018

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Questions?

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