# SIMULATION ENVIRONMENT FOR THE DEVELOPMENT OF NEURO-MUSCULAR STIMULATION SYSTEMS

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## Introduction

Stroke is one of the major causes of disability worldwide, which often leads to unilateral sensorimotor dysfunction. One possible form of therapy is functional electrical stimulation (FES). It has, however some disadvantages, such as insufficient selective muscle activation for different grasps, early fatigue, and low force generation [1]. The high time consumption and poor availability as well as the rapid muscular fatigue of the patients limit the experimental studies. The low repeatability, for example due to misplacement of the electrodes and daily changing condition of the patients, further complicates the evaluation of the results. Computational simulations can be used as a fast and standardised alternative. Therefore, we developed a four-step simulation environment that allows us to evaluate selective forearm muscle activation and force estimation as a function of the different stimulation parameters stimulation current, pulse length, pulse shape, and frequency as well as electrode types, shapes, and orientation.

## Methods

In the first step we build a finite element (FE) model based on personal MRI data. The model includes simplified geometric and anatomical features: electrodes, skin, fat, muscles, and bones. Stimulation near the terminal nerve branches of the muscles leads to muscle activation. We map this region of interest (ROI) by a defined volume in every muscle [2]. In the second step, we calculate the potential distribution in the tissue electrically transient. As the FE model is a linear time-invariant system, we can derive any stimulus from a short pulse (1 µs 1 mA) via linear combination of the system response using impulse response calculation [3]. If electrode arrays are considered, each element of the array can be assumed as an independent current source and the resulting potentials can then be superposed using Helmholtz superposition principle. For further calculations, only the potentials of the ROIs are used. In the third step, the nerve fibers within the ROIs are defined geometrically to determine points at which the extracellular potential is applied to the nodes of Ranvier of the nerves. Then, nerve activation is determined using either the activation function (AF) [4] or an adapted non-linear nerve model (CRRSS) [5]. In the final step, muscle forces are determined using an adapted twitch model [6]. Since each activated nerve of the model innervates a motor unit, the forces resulting from the twitches can be superimposed to a total force depending on the stimulation frequency. Finally, we adjusted and tested the model parameters via force, movement, and EMG measurements on a person-specific basis.

## Results

The lowest electrical stimulus strength (rheobase) for contraction of the extensor digitorum muscle (ED) is 2 mA (CRRSS) or 3 mA (AF) higher than the experimental measurement. The minimal time to double rheobase (chronaxy) differs from the experimental measurement by 44 µs (CRRSS) or 70 µs (AF) (Figure 1A). The normalised force of the ED shows the characteristic recruitment curve and a good overlap with a RMSE of 0.76% for the CRRSS nerve model (Figure 1B). The AF cannot reproduce that curve.

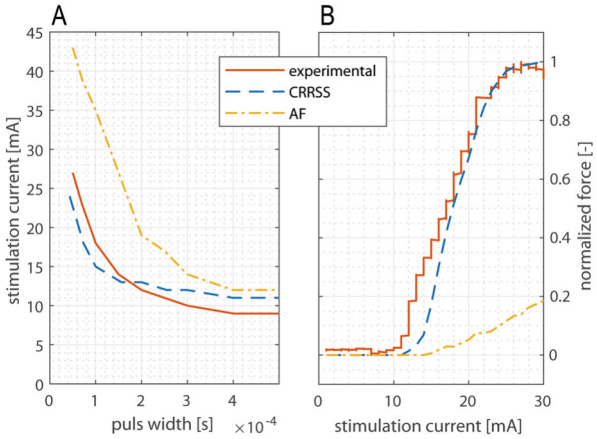


*Figure 1: Experimental and simulated strength-duration curve (A) and normalised force (B) for the M. extensor digitorum (ED) at 30 Hz stimulation frequency.*

## Discussion

With this simulation environment, we can map the selective activation behavior of the superficial musculature in accordance with experimental data using CRRSS. Thus, specific investigation of the influence of various stimulation and electrode parameters can be done. The environment can also be used to identify suitable stimulation sites by using optimisation algorithms or reinforcement learning without negative effects such as muscular fatigue.

## References

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