# Energy Minimization during Transcutaneous Electrical Stimulation by Charge Efficient Stimulation Pulses

## *Benefits of using Short Duration and High Amplitude Stimulation Pulses*

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Abstract: In transcutaneous electrical stimulation motor axons are activated by externally applied electrical pulses. More efficient stimulation pulses could lead to less stress for the patient and to prolonged battery lifetime of the stimulation device. In this study a minimization problem was solved to find charge efficient stimulation pulses, which could potentially reduce the necessary energy provided by the stimulator. The findings of the minimization problem as well as simulations with an active axon model lead to the conclusion that short duration, high amplitude pulses are favorable and that the choice of the correct stimulation amplitude and pulse duration is more important than using the most efficient pulse shape.

# 1 INTRODUCTION

Transcutaneous electrical stimulation (TES) of muscles and motor neurons as a rehabilitation technique can be used to treat patients suffering from stroke or spinal cord injury (Knutson et al., 2007; Mangold et al., 2005). Stimulation parameters that can be controlled by the physician or patient are stimulation amplitude (mA), pulse duration (µs) and stimulation frequency (Hz) (Hunter Peckham, 1999; Gregory et al., 2007). In most cases biphasic rectangular pulses with a short interphase are used. The question whether pulses different to rectangular ones could lead to better stimulation outcomes has been discussed several times (Jezernik and Morari, 2005; Jezernik et al., 2010; Wongsarnpigoon and Grill, 2010; Meza-Cuevas et al., 2012; Krouchev et al., 2014). Especially the question whether energy could be saved by using more efficient pulse shapes has been of great interest.

It is necessary to differentiate between the energy applied to the patient, which depends on the voltage drop over the attached electrodes, or the total energy the stimulation device consumes. Most stimulation devices are supplied from with a constant high voltage source and therefore the energy consumed depends solely on pulse current amplitude and duration. The product of both these parameters is equivalent to the electric charge of the stimulation pulse.

In previous studies different pulse shapes were often compared for fixed pulse durations here we would like to use a different approach and regard the pulse duration as a part of the pulse shape. Consequently the pulses will be compared for fixed stimulation amplitudes.

The goal of this study is to elaborate possible advantages by using pulse shapes other than rectangular ones in regards of delivered electrical charge and to give some advice on performing efficient stimulations which can benefit the patient as well as the battery lifetime of the used device.

# 2 METHODS

## 2.1 3D Finite Element Simulation

A 3D finite element model described in (Loitz et al., 2015) was used to calculate the response of a motor neuron to TES (figure 1). The forearm model consisted of several layers including skin, fat, muscle bone and electrodes. All of these had a specific conductivity and permittivity associated to them. To reduce computation time and to perform an optimization the response to a 1 mA, 1 µs stimulation pulse $I_{pulse}$ along a line was simulated. The electric potential along this line was used as the external electric po-



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tential around a motor axon lying parallel to the skin surface at a specific depth of 8 mm.

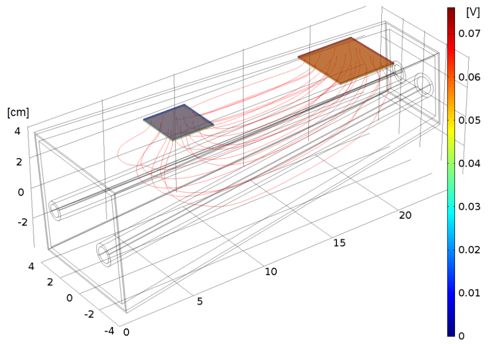


Figure 1: Simple 3D model of the human forearm after $I_{pulse}$. The red lines represent the current flow between the two multicolored electrodes.

## 2.2 Minimization Problem

The first step was to investigate what kind of waveform was suited best to increase the membrane potential, using a passive axon model. Therefore the external potential $V_e$ caused by $I_{pulse}$ was used to calculate the membrane potential $V_m$ according to (McNeal, 1976):

$$\begin{aligned}\frac{dV_{m,n}}{dt} = \frac{1}{C_m}[G_a(V_{m,n-1} - 2V_{m,n} + V_{m,n+1} + V_{e,n-1} \\ -2V_{e,n} + V_{e,n+1}) - G_m V_{m,n}]\end{aligned} \quad (1)$$

Thereby $C_m$ represents the membrane capacitance, $G_a$ the conductance along the axon and $G_m$ the membrane conductance.

The membrane potential $V_{m,n}$ constitutes the unit impulse response of the arm and axon system. $V_m$ at a single node of Ranvier $n$ was then used as the basis for the minimization problem. The goal of the minimization problem was to find the pulse shape which needs the least amount of charge to exceed a threshold potential. The final membrane potential elicited by an arbitrary stimulation pulse was computed as a discrete convolution of a vector $\vec{x}_{stim}$ and a matrix $K$ containing the response of $V_m$ as follows:

$$\vec{V}_{m,stim} = K\,\vec{x}_{stim} \quad (2)$$

with $K$ =

$$\begin{pmatrix} V_m(t_0) & 0 & \dots & 0 \\ V_m(t_1) & V_m(0) & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ V_m(t_{end}) & V_m(t_{end-1}) & \dots & V_m(t_0) \end{pmatrix}$$

The vector $\vec{x}_{stim}$ defines the stimulation pulse shape as its amplitude in mA in 1 µs steps.

During the optimization of the stimulation pulse, some boundary conditions were set: The pulse was not allowed to be longer than 500 µs, the pulse was not allowed to exceed a preset maximum amplitude ($a_{max}$), the minimum amplitude of the pulse had to be larger than zero and the evoked change in membrane potential had to be smaller or equal to the threshold potential until $t = t_{end} = 500\,\mu s$ was reached. For the optimization a starting solution $\vec{x}_{start}$ with a duration of 500 µs and an amplitude of $a_{start}$ was defined that satisfied the boundary conditions. The difference between the electrical charge of the starting solution and the optimized pulse was used as the optimization objective. This task constitutes a linear programming problem, and therefore the Matlab function 'linprog' (equation 3) was used to find the solution.

$$\min_{\vec{x}} \vec{f}^{\,T}\vec{x} \text{ such that } \begin{cases} A\,\vec{x} \leq \vec{b}, \\ lb(t) \leq x(t) \leq ub(t). \end{cases} \quad (3)$$

with $\vec{x} = \vec{x}_{start} - \vec{x}_{stim}$, $lb(t) = x_{start}(t) - a_{max}$, $ub(t) = x_{start}(t)$, $\vec{f} = [-1,-1,...,-1]^T$, $\vec{b} = [0,0,...,0]^T$ and $A$ =

$$\begin{pmatrix} -V_m(t_0) & 0 & \dots & 0 \\ -V_m(t_1) & -V_m(0) & \dots & 0 \\ \vdots & \vdots & \ddots & \vdots \\ +V_m(t_{end}) & +V_m(t_{end-1}) & \dots & +V_m(t_0) \end{pmatrix}$$

## 2.3 Active Axon Model Simulation

To check whether an active model containing ion channels supports the results of the minimization problem the charge necessary to elicit a propagating action potential by a rectangular pulse with fixed maximal amplitudes was compared to a linear increasing and a sinusoidal pulse (figure 2). Previous studies have claimed that linear increasing and/or sinusoidal pulses are more energy efficient compared to rectangular ones (Wongsarnpigoon and Grill, 2010; Meza-Cuevas et al., 2012). The active model used the same equivalent circuit as described in equation 1 but with ionic currents described by (Hodgkin and Huxley, 1952) like equations from (McIntyre et al., 2002).

Divergent from other studies, in this study comparisons of energy efficiency were performed for fixed pulse amplitudes with the pulse duration allowed to vary, since the pulse duration was regarded as a part of the pulse shape.

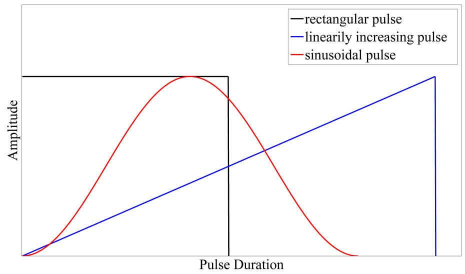


Figure 2: The three pulses used for the active axon model simulation. The amplitude was always the same for all pulses, but the pulse duration was adjusted to elicit an action potential.

## 3 RESULTS

The result of the minimization problem described in section 2.2 can be seen in figure 3. Three different maximum amplitudes were used in this example: 20, 40 and 60 mA. The optimized solution always used the maximum available amplitude to reach the predefined threshold potential and had a rectangular shape. No limitation of the maximum amplitude would result in a Dirac impulse as the optimal solution. In order to achieve a charge optimal stimulation apparently the necessary electrical charge should be applied as fast as possible which results in a rectangular shape. However, increasing the maximum amplitude after a certain value did not reduce the required charge by a lot.

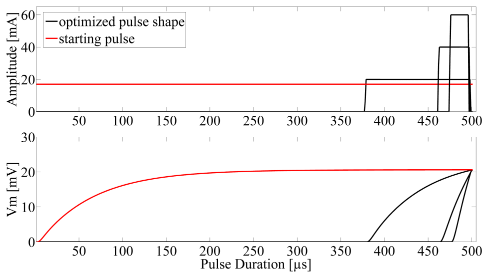


Figure 3: The upper figure shows the starting pulse in red and three optimized solutions for maximal amplitudes of 20, 40 and 60 mA. The figure below shows the elicited change in $V_m$ . Thereby all pulses reach the same potential at the end of the stimulation.

Since the minimization problem just used a simple passive axon model, a comparison to other pulse shapes was performed with an active axon model to see whether the observed results are valid. Figure 4 shows the necessary charge for different stimulation amplitudes. As predicted by the minimization problem, the necessary charge reduces when the stimulation amplitude is increased. This behavior can be observed for all three pulse shapes. However, both of the other pulses require more charge for all amplitude levels. For higher amplitudes saturation appears and the charge requirement of the different pulses comes closer to each other. It can be assumed that at some point the amplitude will be so high that the pulse shape has no impact.

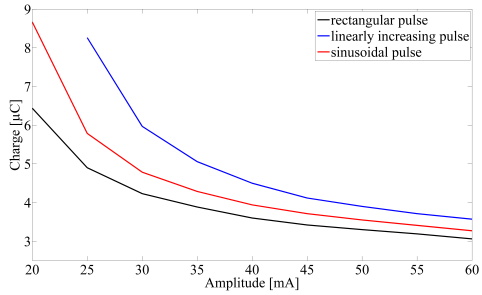


Figure 4: Required charge plotted over the stimulation amplitude.

Since using high stimulation amplitudes will result in a high voltage drop over the patient, an additional computation of the energy dissipation between the electrodes was performed (figure 5). For each pulse shape a minimum in the energy dissipation can be observed at one specific amplitude level. Opposed to the charge, the energy dissipation does increase for higher amplitudes, although the rate is not very high. The linearly increasing pulse as well as the sinusoidal pulse dissipates less energy compared to the rectangular pulse at higher amplitudes.

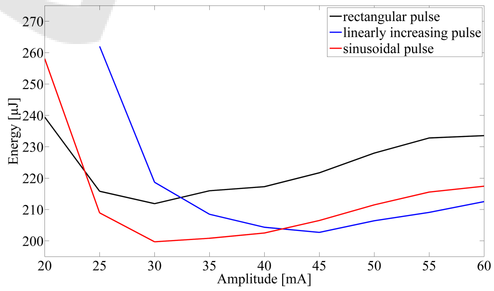


Figure 5: Dissipated energy plotted over the stimulation amplitude.

## 4 CONCLUSION

The minimization problem showed that using stimulation pulses with high amplitudes does reduce the required charge to cause a desired change in $V_m$. This finding was supported by the active model simulation. Nevertheless increasing stimulation amplitudes over a certain point will not decrease the required charge in an efficient way, especially if the increased energy dissipation over the patient is taken into account. The linearly increasing pulse and the sinusoidal pulse were both showing a higher charge requirement than the rectangular pulse for fixed amplitudes. However, the sinusoidal pulse dissipated the least amount of energy over the patient whereas the rectangular one dissipated the most.

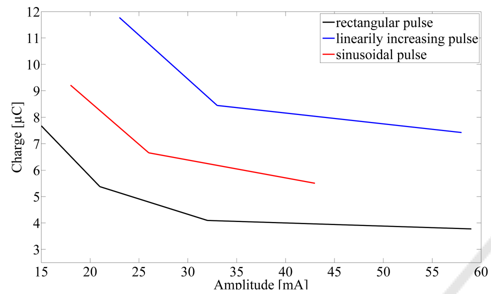


Figure 6: Required charge plotted over the stimulation amplitude. Experimental data from (Meza-Cuevas et al., 2012).

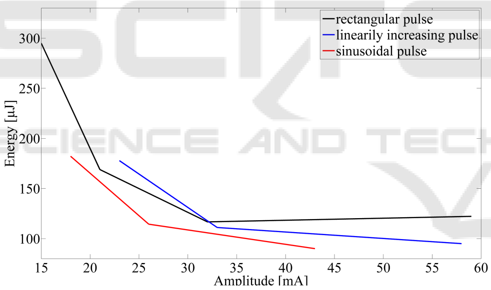


Figure 7: Dissipated energy plotted over the stimulation amplitude. Experimental data from (Meza-Cuevas et al., 2012).

Comparing the simulation results from this study (figure 4 and figure 5) to older experimental data of our institute (figure 6 and figure 7) from (Meza-Cuevas et al., 2012) a very similar behavior could be observed. The rectangular pulse showed the least amount of required charge whereby the sinusoidal pulse dissipated the least amount of energy. Only the slow increase in dissipated energy for higher stimulation amplitudes of figure 5 was not visible in figure 7. One possible reason is the low number of data points. Moreover in (Meza-Cuevas et al., 2012) symmetric biphasic pulses without an interphase were used, whereas this study focused on monophasic pulses.

The low energy dissipation of sinusoidal pulses described in this present study and by (Meza-Cuevas et al., 2012) are comparable to the findings of (Wongsarnpigoon and Grill, 2010) who identified a Gaussian function as the most efficient stimulation pulse in their study.

It was stated by (Jezernik et al., 2010) that a Dirac impulse will provide the most charge efficient stimulation, which is supported by our study.

Even though differences in efficiency across the pulse shapes could be observed, it should be stated that choosing an efficient stimulation amplitude and duration has a much larger effect than just changing one pulse shape to another. Before investing in finding a different pulse shape it is important to know what should be achieved with the new pulse shape and also to investigate whether the available pulse shapes have already been used accordingly.

One suggestion that could be extracted from our findings is that there is no need to worry about short duration and high amplitude stimulation pulses. This does not mean that in every case the stimulation amplitude should be set to the available maximum value. We would rather advice to start with a relatively high amplitude and short pulse duration and from there on start to increase the duration step wise. Pulse durations between 50 and 150 µs for rectangular pulses showed good results in this simulation study.

At this point it has to be noted that the presented graphs and the containing numbers are very specific to the simulation environment used. Usage of different axon properties will change the results. Nevertheless, we are convinced that the shown trends are valid and will be supported by experimental evidence and further simulations in the future. In a future study the impact of a charge balance pulse and its time delay after the actual stimulation pulse should be investigated. A new device able to deliver programmable stimulation pulse shapes could be used to support our results and would help us to design efficient charge balanced stimulation pulses to improve the treatment of patients with TES.

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