



## Array electrode design for transcutaneous electrical stimulation: A simulation study

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### ABSTRACT

Array electrodes are a promising technology that is likely to bring transcutaneous electrical stimulation (TES) a step forward. The dynamic adaptation of electrode size and position helps to simplify the use of electrical stimulation systems and to increase their clinical efficacy. However, up to now array electrodes were built by trial and error and it is unclear how, for example, the gaps between the array elements or the resistivity of the electrode–skin interface material influence the current distribution. A TES model that comprises a finite element model and a nerve model has been used to analyze the influence of array electrode gaps and gel resistivities on nerve activation. Simulation results indicate that the resistivity of the electrode–skin interface layer should be adapted depending on the size of the gaps between the array elements. Furthermore, the gap sizes should be smaller than 3 mm in order to keep losses small.

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### 1. Introduction

In transcutaneous electrical stimulation (TES) pairs of surface electrodes are placed on the skin in order to stimulate motor nerves. Clinically, TES is often applied in the rehabilitation of stroke subjects or spinal cord injured subjects [1], or for supporting tasks of daily living [2] using so called neuroprostheses. In the past such neuroprostheses used single stimulation electrodes [3,4]. Recently array electrodes were proposed to improve the efficacy of such TES systems [5,6]. Array electrodes consist of multiple elements which can be individually activated to form a virtual electrode of arbitrary size and location. The position and size of the activated region (virtual electrode) can be dynamically changed [7,8]. For good wearability textile array electrodes are produced by embroidering silver coated fibers in the form of array electrodes into garments [9]. One large layer of self-adhesive hydrogel is used as skin interface between the skin and the array electrode [10]. The gaps between array elements and the resistivity of the interface layer (gel) influence the current distribution flowing into the human limb.

In previous works the sizes of the gaps between array elements and the resistivity of the gel layer were chosen intuitively [5,6,10]. In this paper a TES model comprising a finite element (FE) and a

nerve model is used in order to analyze and better understand the influence of the gap sizes and the gel resistivity on nerve activation. Using the TES model it was investigated how large the gaps are allowed to be and how this decision is related to the choice of the gel resistivity.

Previously, the indifferent electrode (anode) was placed separately from the array [5,6]. However, in [9] it was shown that the active (cathode) and the indifferent electrode (anode) can be placed on the same array. This simplifies the application of TES because only one electrode has to be applied to the human body instead of one array plus a separate indifferent electrode. Obviously, a part of the applied current is lost because it will directly flow from the cathode to the anode through the hydrogel. It is unclear how much of the applied current is lost in the gel layer depending on different parameters (e.g., gel resistivity, skin resistivity, . . .). To address these issues different parameter combinations were applied to the TES model in order to quantify and to reduce these losses.

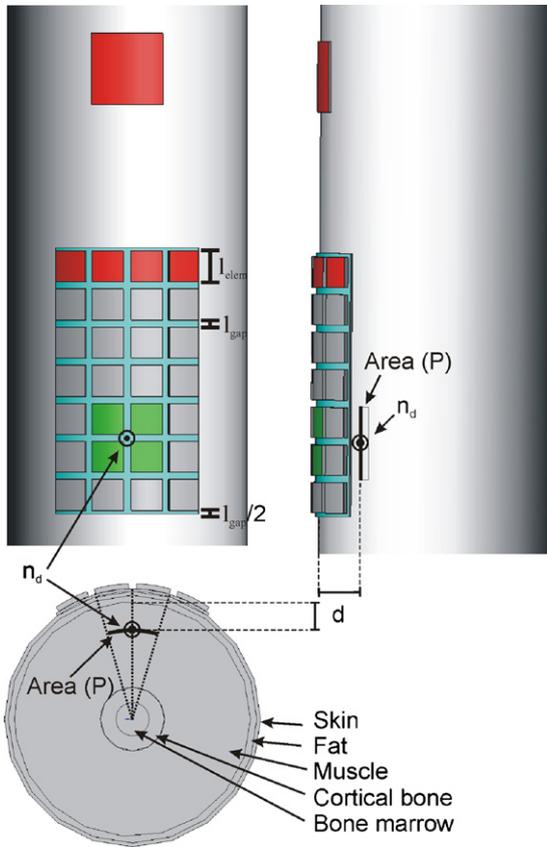
### 2. Methods

#### 2.1. Combined FE and nerve model

The experimentally verified TES model comprised a FE model and a nerve model [11]. The FE model calculates the 3D electric potential distribution in a geometry representing the forearm and the nerve model relates the potential distribution to nerve activation [12].

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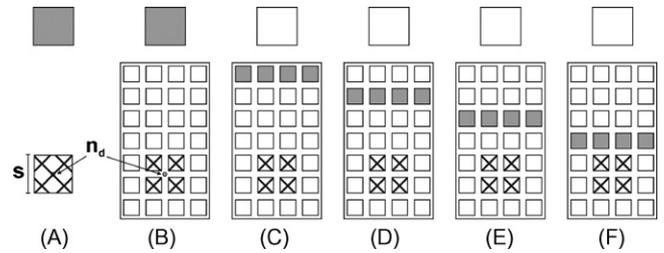


**Fig. 1.** Cylindrical arm model (length: 60 cm, diameter: 10 cm) with array electrode and a separate single electrode. Underneath the array one large gel layer provides the interface between array elements and skin. A region of  $2 \times 2$  array elements was used as cathode. The anode was either on the separate single electrode or on the array.

The electric scalar potential distribution ( $V_{FE}$ ) in the arm model generated during TES by array electrodes was calculated by using Laplace's Eq. (1) with the FEM software Ansys (EMAG, Ansys Inc., Canonsburg, PA). It takes only the resistive properties ( $\rho$ ) into account. Capacitive and inductive effects were neglected in the model [13,12]. The geometry was a multiple layer cylindrical representation of the forearm consisting of skin, fat, muscle and bone layers (Fig. 1). The resistivity of the skin ( $\rho_{skin}$ ) depends non-linearly on the applied current density [14]. The current densities that are typically applied are between  $1 \text{ mA/cm}^2$  and  $15 \text{ mA/cm}^2$  [9] leading to skin resistivities between  $670 \Omega \text{ m}$  and  $2100 \Omega \text{ m}$  [14]. In order to cover a wide range of these experimentally found skin resistivities, low ( $700 \Omega \text{ m}$ ) and high skin resistivities ( $2000 \Omega \text{ m}$ ) were used in the TES model. The thickness of the skin in the model was 1.5 mm. The thicknesses and electrical properties of the other tissues were 2.5 mm and  $33 \Omega \text{ m}$  for fat, 33.5 mm and  $3 \Omega \text{ m}$  for muscle, 6 mm and  $50 \Omega \text{ m}$  for cortical bone, and 6.5 mm and  $12.5 \Omega \text{ m}$  for bone marrow [15,16]. The total cylinder diameter is therefore 10 cm. The anisotropy of the muscles' resistivity was considered by using a factor of three between the axial and radial direction ( $3\rho_{axial} = \rho_{radial}$ ) [16]. The cylinder was chosen long enough (60 cm) that the boundary of the cylinder did not have an influence on the potential distribution underneath the electrodes.

$$-\nabla \cdot ([\sigma] \nabla V_{FE}) = 0 \quad (1)$$

The TES model was used to compare array electrodes with standard single electrodes. The influence of gaps between individual array elements and the influence of the electrical properties of the electrode gel on nerve activation were investigated. The goal is to



**Fig. 2.** (A) Two single electrodes for comparison with array electrodes. Grey electrodes indicate anodes and electrode with crossed lines cathodes. (B) Array electrode with separate single anode. (C–F) Array with different distances between anode and cathode (C: 51 mm, D: 35 mm, E: 19 mm, F: 3 mm).

identify parameters where activation of array electrodes is similar to the activation of single electrodes. Therefore, two different electrode layouts were considered:

- An array electrode and a single electrode were placed on the cylinder (sketches B to F in Fig. 2).
- Two single electrodes were used (sketch A in Fig. 2).

Between cathode and anode biphasic current regulated pulses with a pulse amplitude  $I$  and a pulse duration of  $300 \mu\text{s}$  were used.

The single elements of the array electrode were modeled as good conducting substrate (silver) and had a size of 13 mm ( $l_{elem}$  in Fig. 1), the gap sizes ( $l_{gap}$ ) were between 1 mm and 5 mm ( $l_{gap}$  in Fig. 1), and the resistivity was  $1.6 \times 10^{-8} \Omega \text{ m}$  (silver). These parameters were chosen according to the specifications of array electrodes used in another study [9]. Underneath the array elements a single hydrogel layer provided an interface to the skin. This gel layer had thicknesses ( $l_{gel}$ ) between 0.25 mm and 2 mm, and resistivities ( $\rho_{gel}$ ) between  $1 \Omega \text{ m}$  and  $10,000 \Omega \text{ m}$ . This is the range of hydrogels that can be produced by manufactures such as Sekisui Plastics Co., Ltd., Tokyo.

In order to compare single with array electrodes their edge lengths have to be of comparable size. The size  $s$  of the single cathode (sketch A in Fig. 2) was always adapted to the size of the corresponding 2 by 2 electrode (array, sketches B to F in Fig. 2) it was compared with. The edge length of the single cathode was  $s = 2l_{elem} + l_{gap}$ . The centers of the single electrode and the  $2 \times 2$  virtual electrode were kept congruent.

Passive nerve bundle models consisting of 100 axons with diameters distributed according to the bimodal distribution in human nerve bundles with peaks at  $6 \mu\text{m}$  and  $13 \mu\text{m}$  [17,18] were placed within the volume representing muscle and were oriented in axial direction of the cylinder. Hundred axons are enough to accurately predict recruitment of one nerve bundle [19]. The nerve bundles contained passive axon models that were derived from the MRG-model [20]. It was previously shown that such passive axon models can be used to predict activation of active axon models [21,22]. Computer simulations of the MRG model were performed in NEURON (Yale University). An implementation of a single axon can be downloaded from the model database (ModelDB) on the NEURON web page (<http://www.neuron.yale.edu/neuron>). Multiple of these single axon models with the mentioned bimodal diameter distribution were used in order to model a nerve bundle containing axons of different diameters. The axons intracellular voltage is influenced by the extracellular potential  $V_{e,n}$  at the nodes of Ranvier ( $n$ ) and at non-nodal compartments ( $n-n$ ) between the nodes of Ranvier  $V_{e,n-n}(t)$  [20].

The link between the FE model and the nerve models was established by assigning the spatially interpolated potentials from the FE model  $V_{FE,n}$  and  $V_{FE,n-n}$  to the corresponding extracellular potentials of the axon models  $V_{e,n} = V_{FE,n}$  and  $V_{e,n-n} = V_{FE,n-n}$ .

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