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Finite Element Models of Transcutaneous Spinal Cord Stimulation

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Definition

Transcutaneous spinal cord stimulation (SCS) is a non-invasive method to electrically stimulate afferent structures of the human lumbar spinal cord. These are the same neural targets as predominantly activated by epidural implants. Biophysical principles derived from computer simulations contributed to the identification of the directly activated neural structures. These simulations combine finite element (FE) models with nerve fiber models and the activating function concept. The transcutaneously generated electric field is inherently non-focal and thus, nerve fiber activation relies on inhomogeneities of the volume conductor and the anatomical paths of the target fibers relative to the electric field. The anatomy, its electrical properties and the neurogeometry cause localized low-threshold sites within the widespread field and allow for the observed selective stimulation. Computer simulations are crucial for understanding the neuromodulative effects of transcutaneous SCS resulting from the trans-synaptic activation of spinal cord neural circuits, and for further improving the stimulation technique.

Detailed description

Epidural SCS involves the activation of neural target structures by leads placed in close distance of a few millimeters to the posterior aspect of the spinal cord. In transcutaneous SCS, the same deeply located neural structures are targeted from a distance of several centimeters, from the body surface. Compared to the neural structures, the stimulating skin electrodes are large. Several anatomical structures between the electrodes and the spinal cord introduce electrical inhomogeneities and anisotropies, which in turn influence the generated electric potential. Yet, electrophysiological studies showed that transcutaneous SCS reliably and selectively stimulates

similar neural structures as with epidural electrodes, i.e. sensory fibers within the posterior roots and rootlets (Minassian et al. 2011).

Computer simulations that calculate the generated electric potential and its effect on neural structures are essential in understanding the immediate effects of SCS (Holsheimer 1998). The problem of calculating the electric potential distribution in a complex volume conductor can be mathematically formulated by partial differential equations. Due to the geometrical complexity and inhomogeneities of the anatomical structures, these equations are usually only tractable by applying numerical approximation methods such as the finite difference, the FE, boundary element or multigrid method (Johnson 1997).

Here, the application of the FE method is elaborated for the approximation of the produced electric potential while assuming quasi-stationary conditions (for details on the FE method see elsewhere in this book, Rattay: Finite Element Model of Volume Conductor). A model is always an abstraction of the reality and can represent only rather simple anatomical properties, can model average dimensions or can be tailored to an individual subject. Current models of transcutaneous SCS (Ladenbauer et al. 2010; Danner et al. 2011) cover a general case with a rather simple geometry.

As opposed to models of epidural SCS where most of the generated currents flow inside the spinal canal, the spine and its geometry are paramount for calculating the electric potential across the neural target structures produced by skin electrodes. The simplified model must include anatomically realistic regions of soft tissues in-between the bony structures with comparably high conductivities, thus allowing for current penetration of the canal. Structures below the electrode (i.e. skin, subcutaneous fat, paraspinal muscles) and the spine, including the ligaments and intervertebral disks, are influencing the produced field near the target structures and thus need to be included in the model of transcutaneous SCS. Furthermore, the detailed anatomical properties of structures inside the spinal canal (e.g. spinal cord gray and white matter, spinal nerves, cerebrospinal fluid, epidural space) need to be modeled. Figure 1a-b illustrate the model geometry, 1c is a sketch of potential target fibers and 1d shows computed current flow lines.

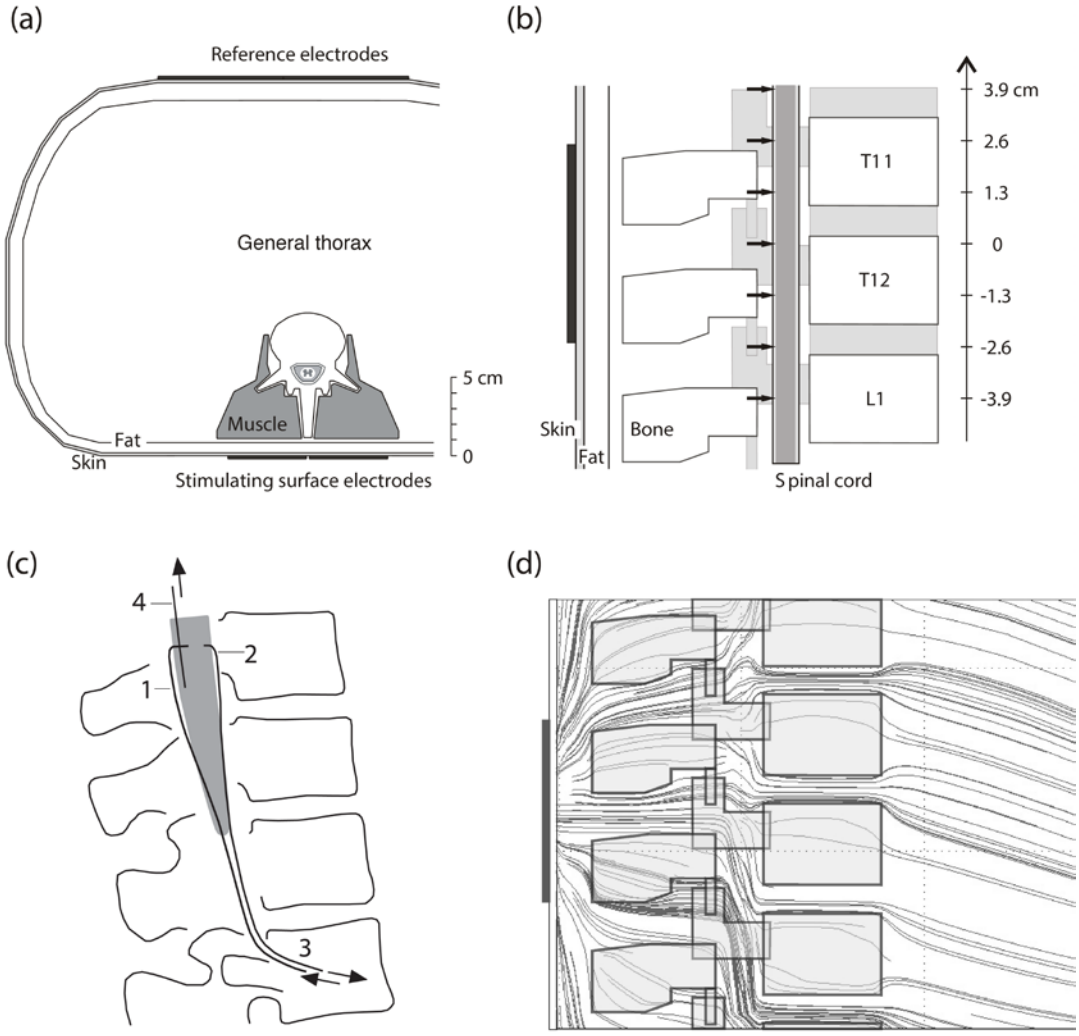

Figure 1 Volume conductor (VC) model geometry and the paths of the simulated axons. (a) Cross section of the VC model. (b) Midsagittal section showing the relation between the spine, spinal cord, intervertebral discs and the transcutaneous paraspinal electrode (black). Model fiber entry and exit levels into the spinal cord are marked with arrows. (c) Sketch of posterior (1) and anterior root fibers (2), joining together at the intervertebral foramina (3), and of the posterior columns (4) in relation to the spinal geometry. (d) Computer simulated current flow within a 2 mm layer at the midsagittal plane (reproduced from Danner et al. 2011 with permission from Wiley Periodicals, Inc.).

In a further step, the electric potential computed by the FE models is evaluated along determined paths, representing the geometry of potentially excitable neural structures. The excitability of nerve structures depends on several factors. The soma is hard to excite due to its relatively high membrane capacitance (Rattay 1999). The activation thresholds of myelinated nerve fibers are lower than of unmyelinated ones, and are lower for nerve fibers of larger diameters (Ranck,

1975). Thus, large diameter myelinated axons can be preferentially selected for the simulation as target structures.

The stimulation effects on the target neural structures can be estimated from the activating function concept (Rattay 1999), which states that the de- and hyperpolarization induced by an externally applied electric field is proportional to the second order spatial differential of the electric potential (or the negative going, first order differential of the electric field) along the nerve fiber. Thus, the spatial trajectory of the fibers relative to the field and the traversal of media of different conductivities greatly influence the excitability. Nerve fiber models can be used to identify the activation thresholds and simulate the influence of the fine structure of the neurons (esp. branchings and collaterals; Struijk et al. 1992). Note that all aforementioned factors can predominantly introduce de- as well as hyperpolarizations along the axons; the direction of the effect on the activation thresholds depends on the fiber's path relative to the electric field. More precisely, convex fiber bends relative to the cathode, traversals of a fiber from a medium with high to one with low conductivity (as seen from the cathode), and fiber collaterals away from the cathode increase the excitability, whereas the respective opposite direction would decrease the excitability.

Computer simulations of transcutaneous SCS showed that the abovementioned factors are paramount for the activation of spinal neural structures through surface electrodes (Ladenbauer et al. 2010; Danner et al. 2011). Hot-spots of maximal depolarization were identified at the entry point of posterior rootlets into the spinal cord, at the entry/exit point of the posterior and anterior roots into/from the spinal canal (Figure 2) and at the branching points of the collaterals from the posterior column fibers. This is somewhat different to epidural SCS, where the points of strongest depolarization are close to the cathode and depend only on anatomically induced hotspots if the leads are located rostral to the fiber’s entry point into the spinal cord (Figure 3). Furthermore, epidural SCS was shown to be more selective. For transcutaneous SCS, the computer simulations suggest that, with increasing intensity, first fibers in the posterior roots, followed by fibers in the anterior roots, and eventually fibers in the posterior columns are stimulated.

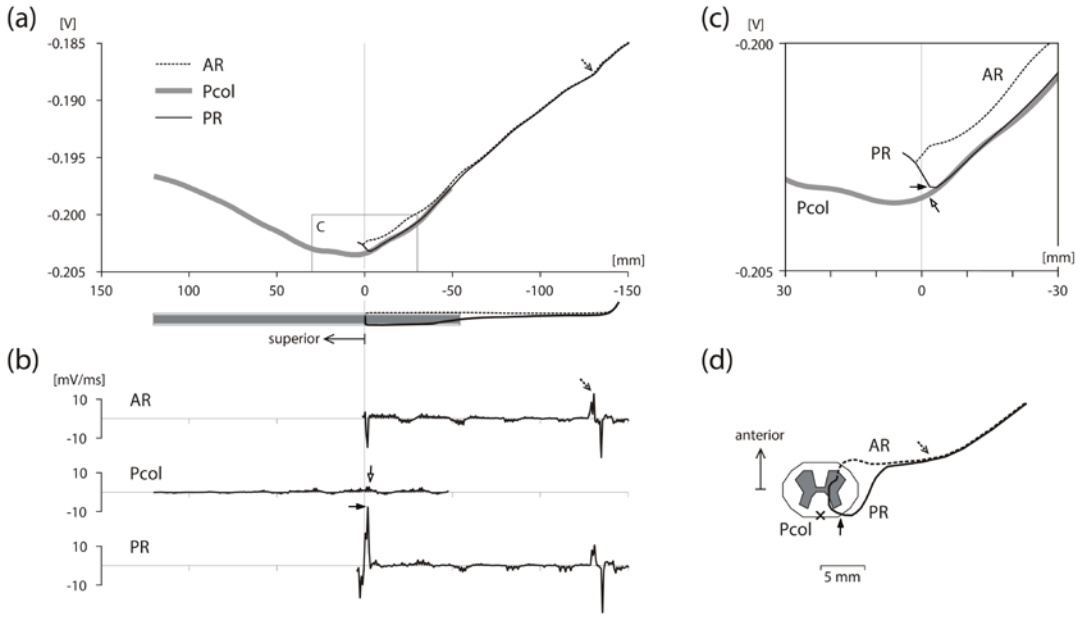

Figure 2 Stimulation effect evaluated along sensory structures and motor fibers of the spinal cord and roots. Low-threshold sites are caused by anatomical influences. The straight posterior column (Pcol) fiber shows only slight deviations from zero in its activating function, whereas the fibers in the posterior (PR) and anterior root (AR) are subject to strong de- (and hyper-) polarization at their entry/exit into/from the spinal cord and canal, respectively. (a) Extracellular potential along selected target fibers generated by transcutaneous stimulation at -1 V. The Pcol fiber is located medially and superficially in the white matter. The abscissa is the distance along the fiber trajectories from the level of the center of the stimulation electrode. Anatomical relations of the PR and AR fiber are illustrated below the abscissa. (b) Activating functions corresponding to cases in (a). (c) Enlarged view of the box in (a). (d) Top view of the trajectories of the AR and PR fibers and of the simulated Pcol fiber location (x). Arrows indicate the lowest threshold sites. Threshold of the fibers with the paraspinal electrode acting as the cathode, calculated using the model of McIntyre et al. (2002): PR, 16.7 V; AR, 51.7 V; Pcol, 67.4 V (reproduced from Danner et al. 2011 with permission from Wiley Periodicals, Inc.).

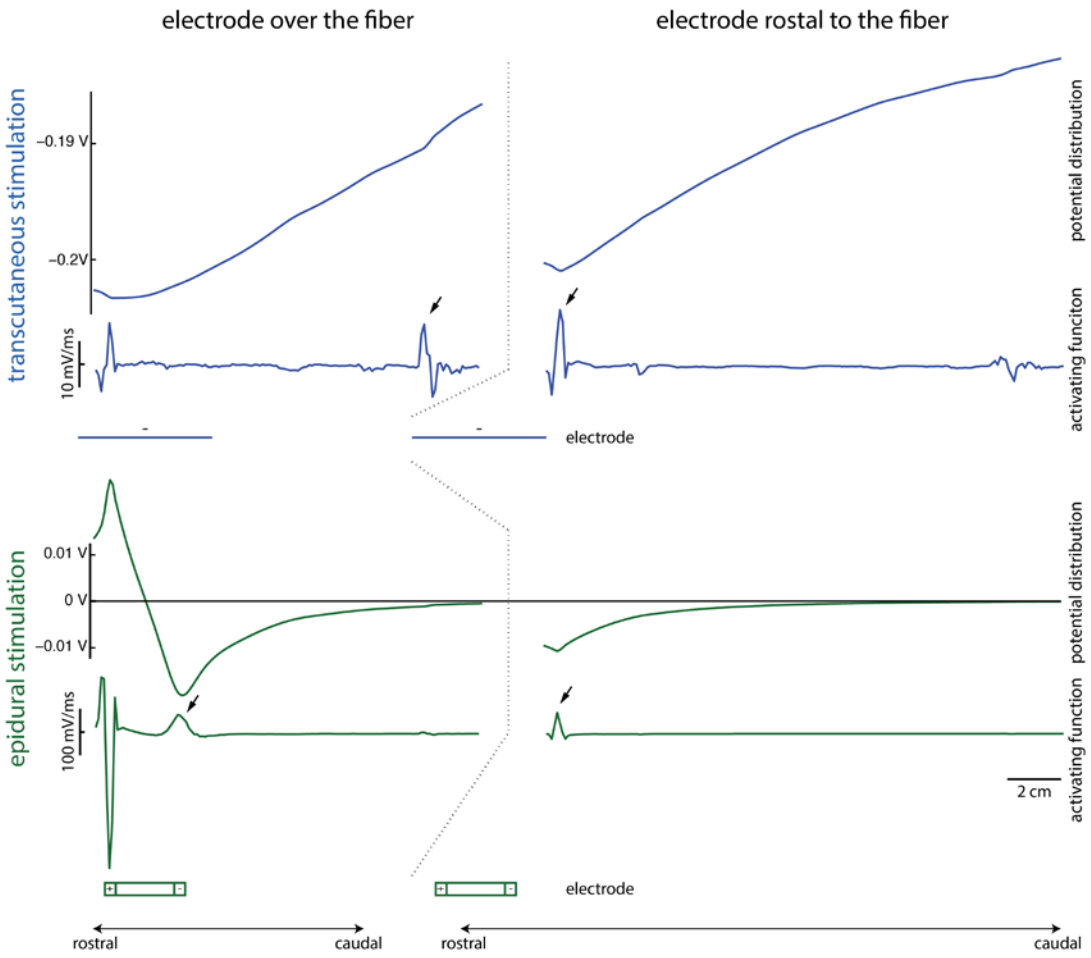

Figure 3 Stimulation effect (1 V) of transcutaneous and epidural electrodes located over and rostral to posterior root fibers. The two low-threshold sites (entry into the spinal cord and the canal, respectively) can be seen in case of transcutaneous stimulation. Proximity of the electrode to these hot-spots increases the amplitude of the corresponding parts of the activating function. In case of epidural stimulation, the electrode placed over the fiber introduces strong, local deflections of the potential distribution and the activating function at the level of the contacts of the electrode. Whereas, the electrode located rostral to the fiber introduces an electric potential and activating function along the fiber similar to those in case of transcutaneous stimulation. The action potential is initiated at the entry of the posterior root into the spinal cord for both stimulation techniques. Arrows indicate the spike initiation sites.

To summarize, modeling transcutaneous stimulation of the lumbar spinal cord helps to identify the directly activated structures, provides explanations for their selective activation by the distant stimulation, and predicts their recruitment order and points of spike initiation. A more general principle was highlighted. The field produced by a distant electrode in a homogeneous medium would be too non-focal to effectively activate neural structures. Rather, the anatomical

properties of neurons and their surrounding tissue allow for strong and localized depolarization. Modeling will further contribute to the development of new, improved and/or tailored stimulation techniques.

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