Influence of electrode impedance on threshold voltage for transcranial electrical stimulation in motor evoked potential monitoring

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Abstract—*Motor potentials evoked by transcranial electrical stimulation (TES) are used for monitoring the motor pathways, with emphasis on the spinal cord and brainstem. The stimulus voltage threshold is the voltage below which no motor response can be elicited. It has frequently been used as a monitoring parameter. However, its value can be limited, because it is affected by the impedance of the stimulus electrode. For example, the voltage threshold can change owing to formation of oedema of the scalp. The relationship between the TES voltage threshold and the electrode impedance of different electrode types was studied and discussed in the context of neuromonitoring: 323 impedance and voltage threshold pairs were studied, and TES was performed with disc cup EEG electrodes (six), corkscrew electrodes (type I: seven, type II: eight), multiple EEG needle electrodes (16) and a large needle electrode Cz' (anode) together with a ground strip over the forehead (cathode) (286). The study found the voltage threshold to be strongly dependent on electrode impedance when the impedance was higher than 460Ω (correlation: $R^2 = 0.87$; $p < 0.001$). Below 460Ω, which included 91% of the category with the largest electrode surfaces, 25% of the multiple EEG electrodes and 75% of type II corkscrew electrodes, no significant correlation ($R^2 = 0.0064$; $p = 0.15$) was found. It was concluded that the correlation between the TES voltage threshold and electrode impedance can be markedly reduced by using TES electrodes with large contact surfaces, resulting in limit values for these parameters. This also may improve the reliability of TES motor evoked potential monitoring.*

Keywords—*Transcranial electrical stimulation, Safety, Impedance, Motor evoked potentials, Brain, Threshold monitoring*

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1 Introduction

THE VOLTAGE threshold V_{th} of motor potentials evoked by transcranial electrical stimulation (TES) using a voltage stimulator has been used as a monitoring parameter in many patients (CALANCIE *et al.*, 1998, 2001). This voltage threshold is, for a large part, dependent on the impedance of the electrodes used for stimulation. This electrode impedance is the sum of the impedance in the immediate vicinity of the electrodes, the so-called local electrode impedance, and the impedance of the tissues of the head between the electrodes, the so-called head impedance. Large electrode surfaces imply low local impedances.

When the local electrode impedance is large and becomes a dominant factor in the impedance, we can expect the measured V_{th} to be largely determined by it. Consequently, the stimulation voltage threshold depends on the electrode dimensions. In this paper, the relationship between the TES voltage threshold and electrode impedance of different electrode types is studied and discussed in the context of neuromonitoring.

2 Material and methods

Data were analysed retrospectively from routine TES motor evoked potential (MEP) monitoring in patients during corrective spine surgery, between 1 March 1997 and 1 September 2003, in our institute. Eighteen cases where the impedance values or V_{th} values were missing, or where MEPs were poor or absent, were excluded. A total of 323 V_{th}–impedance pairs were available for analysis. The following anaesthetic protocol was used: induction with propofol, $3\,mg\,kg^{-1}$, and remifentanil, $2\,g\,kg^{-1}$; anaesthesia maintenance: propofol $1{,}5–2\,mg\,kg^{-1}\,h^{-1}$, remifentanyl, $15\,g\,kg^{-1}\,h^{-1}$, with titration to pain responses and ventilation with a mixture of 35% O_2/65% N_2O. No muscle relaxants were used.

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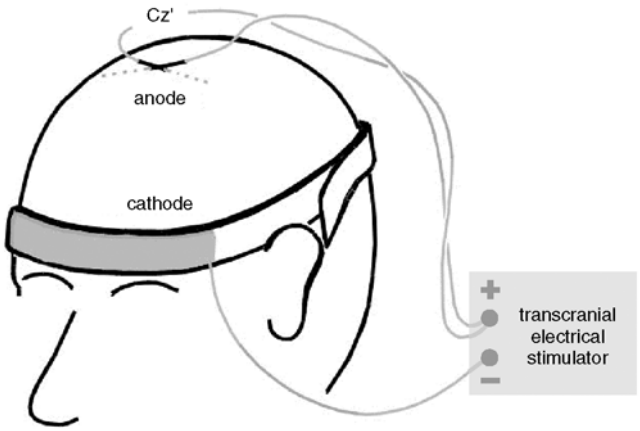

Fig. 1 *Set-up of large surface electrodes used for transcranial stimulation. For impedance measurements, impedance-measuring device replaces transcranial stimulator*

Impedance values were routinely measured at 1 kHz to check the correct placement and electrical connection of the TES electrodes. TES-MEP response curves were then measured in 25 V steps between 0 and 450 V at $n = 4$ pulses per stimulus, interpulse interval (IPI) = 2 ms. The curves were used to determine a supramaximum TES voltage for monitoring. TES threshold voltages V_{th} for the anterior tibial (TA) muscles were determined. Standard EEG disc silver cup electrodes with a diameter of 10 mm were applied in the first six cases. Two of these electrodes placed at Cz′ were used for anodal stimulation, and four, forming a 10 cm long strip over the forehead, were used for cathodal stimulation.

Two types of corkscrew electrode were used. Type I (seven patients) were corkscrew electrodes[+] that were obtained in 1997, and type II (eight patients) were CS corkscrew electrodes[#] that we obtained from Dr Deletis (Department of Neurophysiology, Beth Israel Medical Center, New York) in 2003. When compared with type I, the surface area of the electrodes of type II was greater owing to the larger diameter (0.4 mm) of the helix needles (helix diameter: 6 mm).

In 16 patients, two needle EEG electrodes, 1.25 cm long and 0.3 mm diameter, were used for anodal stimulation at Cz′, and three were used for cathodal stimulation over the forehead. Multiple electrodes at one location were intended to decrease the electrode impedance. In six cases, the impedance between two single needle electrode pairs was measured before multiple insertions were performed. There are no V_{th} values available for single needle electrode pairs, as the TES-MEP response curves were measured after multiple insertions.

A further decrement in impedance was obtained in the next 286 cases, by the use of electrodes with a larger contact surface. Two interconnected stainless-steel needle electrodes (3 cm and 0.4 mm diameter) were inserted in a lateral direction at Cz′, defining a 5–6 cm long anodal contact strip in the coronal plane and a Velcro ground strip, with a contact surface area of $15 \times 1\,cm^2$, immersed in saline, was placed supra-orbitally. The impedance was always measured before TES stimulation. The set-up is illustrated in Fig. 1. Cases where stimulation was performed in a coronal plane across C3-C4 or C1-C2, which is used less frequently in our centre, were excluded from this study.

The monitoring equipment* comprised eight EMG channels and a multipulse TES voltage stimulator with an output impedance of 18 Ω (developed by one of the authors). The impedance values were corrected according to a table that was obtained from calibration tests of the impedance meter using a series of 1% load resistors between 100 and 2000 Ω. The same resistors were used to calibrate the output voltage of the TES stimulator using an oscilloscope. The calibration table obtained was used to correct the V_{th} values. The EMG signals were measured with surface ECG electrodes.[†] The filter settings were high-pass: 50 Hz; low-pass: 2.5 kHz.

Statistical computations were performed by the SPSS version 11.0 software package using ANOVA computation for linear regression analysis.

3 Results

The mean and standard deviation of the impedance and stimulation voltage thresholds V_{th}, are listed in Table 1. Six impedance values of single EEG needle electrodes are also listed. No individual V_{th} values are given, as these were only measured with the multi-electrode montages. The mean and standard deviation of the impedances of the corkscrew (type I) and single EEG needle electrodes are about equal to each other, so that these two can be considered together within the same electrode category.

Fig. 2 shows the TES voltage threshold values V_{th} as a function of the TES impedance. Two impedance regions can be distinguished: one below 460 Ω, where V_{th} appears to be independent of the impedance, and one above 460 Ω, with a distinct relationship between V_{th} and impedance. This is confirmed by regression analysis.

For impedance values $I \leq 460\,\Omega$ (258 of 286 = 90% of the large contact-surface type electrodes and 4 of 16 = 25% of the multiple EEG electrode category), no statistically significant ($p = 0.15$) correlation ($R^2 = 0.0064$) was found between V_{th} and the TES impedance. The regression line $V_{th} = 37 + 0.039I$ is almost horizontal. For $I > 460\,\Omega$, the Spearman correlation coefficient of 0.93 ($R^2 = 0.87$) is significant for $p < 0.001$; regression line $V_{th} = -186 + 0.516I$. This regression line intersects the mean level of the uncorrelated points in the region below 460 Ω at 461 Ω. Both regression lines, which intersect at 466 Ω, are shown in Fig. 2.

Fig. 3 gives a survey of which impedance regions the four electrode categories belong to. The categories of the disc cup, corkscrew (type I) and single EEG electrodes are all found in the region above 460 Ω, with a high correlation between electrode impedance and V_{th}, together with 75% of the multi-EEG needle, 25% of the type II corkscrew electrodes and 10% of the large contact-surface electrodes.

4 Discussion

There is no correlation between V_{th} and the TES impedance for stimulus electrode impedances below 460 Ω. Twenty-five percent of the multi-EEG needle electrodes, 75% of the type II corkscrew electrodes and 90% of the large contact-surface electrodes were found to be within this impedance region. This implies that, for impedances below 460 Ω, the threshold TES voltage V_{th} has reached its lower limit whereby the influence of the local electrode impedance is minimised. The wide scatter of V_{th} and impedance values in this low impedance region is ascribed to inter-individual differences in geometrical parameters, such as the thickness and shape of the scalp, skull and CSF layer and specific conductance of the skull, and to physico-chemical effects from electrochemical polarisation at the contact surfaces. All measurements were performed before any stimula-

[+] Neuromedical Supplies, Inc., Sterling, VA

[#] Nicolet Biomedical, Madison, WI

*NeuroGuard

[†] 3M Red-Dot

Table 1 Survey of TES electrode impedance values and V_{th} for different types of electrode. (*) No data as not used for stimulation. Impedance was measured at 1 kHz; linear range, small signal

		TES impedance, Ω		V_{th}, V	
Electrode type	Number of cases	mean	SD	mean	SD
silver disc cup EEG	6	1056	186	347	52
corkscrew:					
type I	7	801	126	251	81
type II	8	514	171	90	42
EEG needle electrodes:					
multiple	16	542	107	135	49
single	6	773	121	*	*
large needle electrodes and ground strip	286	382	67	56	23

tion, so that impedance changes due to TES, which in our experience showed decrements of between 20 and 90 Ω, could be excluded.

The impedance region above 460 Ω shows a significant linear correlation between V_{th} and the TES electrode impedance. This applies to all disc cup EEG electrodes and most type I corkscrew electrodes and EEG needle electrodes (Table 1). EEG disc cup electrodes and corkscrew electrodes (BURKE *et al.*, 1992; DELETIS and CAMARGO, 2001; HICKS *et al.*, 1992; KOTHBAUER *et al.*, 1997; MEYLAERTS *et al.*, 1999; PECHSTEIN *et al.*, 1996; VAN-DONGEN *et al.*, 1999) are used for TES in almost all centres. Table 1 shows that the mean V_{th} of 56 V with large contact-surface electrodes is increased to 251 (+250%) for the corkscrew (type I) and single EEG needle electrodes and to 347 (+346%) for the EEG disc cup surface electrodes. This means that the local electrode impedance dominates the overall impedance of the TES electrodes. This has the following implications in clinical practice:

First, extra electrical energy corresponding to the V_{th} increment will be dissipated in the direct vicinity of the TES electrode. Relatively high local electrode impedance will result in a high local voltage drop in the direct vicinity of the electrodes. The resulting thermal effects will be most pronounced with small electrodes for two reasons: the high local electrode impedance of small electrodes requires a higher TES voltage, and the dissipation occurs into the small tissue volumes of small electrode contact surfaces, whereas large surface area electrodes dissipate less energy in a larger tissue volume, which has a larger thermal capacity, resulting in a smaller temperature increment. For any given stimulation current, the dissipated energy at the electrode is linearly related to the local electrode impedance.

It must be emphasised that the impedance values in this study were determined at 1 kHz and at low voltages, where tissue impedances are linear. These values will differ significantly from the actual quotient between the TES voltage and current during a pulse that defines the true TES impedance. The short pulse widths of rectangular TES pulses of 50 and 100 μs (GEDDES and BAKER, 1967) have most of their spectral power in a frequency band in a range of 4–30 kHz. This is one magnitude higher than the frequency of 1 kHz that is used for impedance measurements. In that higher frequency range, most biological tissues have lower impedances (FOSTER *et al.*, 1979; GEDDES, 1987; SAHA and WILLIAMS, 1995; STOY *et al.*, 1982). Moreover, the impedance will be highly non-linear at high stimulus voltages, resulting in a further impedance reduction (MCADAMS and JOSSINET, 1994; MERTON *et al.*, 1982; POLETTO and VAN

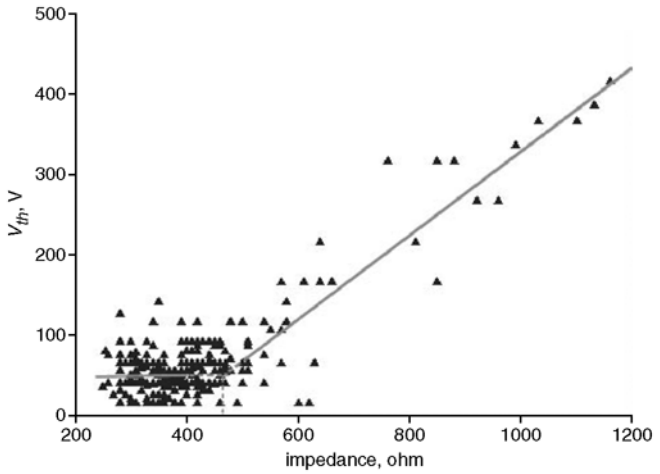

Fig. 2 *Scatter diagram of anterior tibial V_{th} as function of TES electrode impedance. Impedance is measured at 1 kHz, linear range, small signal. Diagram comprises 315 data pairs, many of which overlap. Regression lines and their intersection for impedances ⩽ 400 Ω and ⩾ 500 Ω are shown*

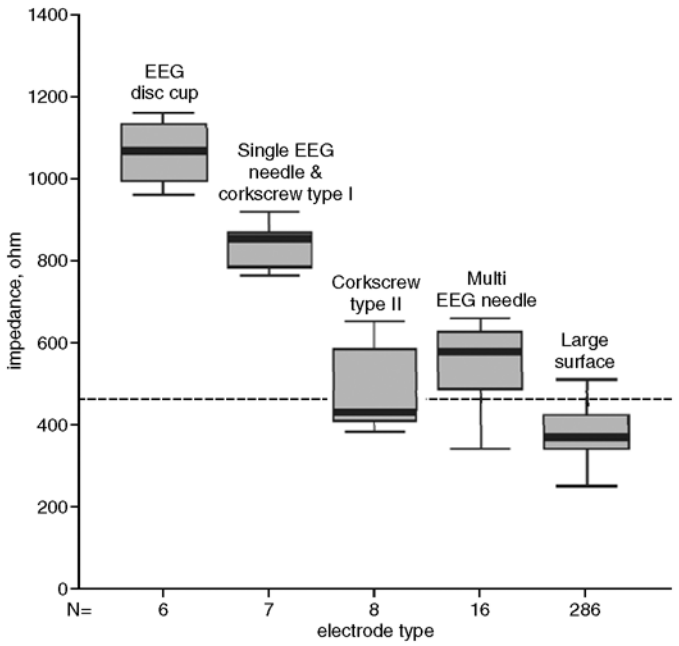

Fig. 3 *Box plot of anterior tibial electrode impedance of five types of electrode used in this study. Impedance was measured at 1 kHz, linear range, small signal. Single EEG electrodes, which have approximately same impedance distribution as type I corkscrew electrodes, are considered with them. Boxes represent interquartile range that contains 50% of values. Whiskers are lines that extend from box to highest and lowest values, excluding outliers. Line across box indicates median. Six outliers out of 286 impedance values of large surface electrodes are not shown*

DOREN, 1999). We cannot therefore estimate absolute values of tissue heating during high-voltage TES using the 1 kHz linear impedance values, which leads to overestimates. Nevertheless, the impedance data at 1 kHz are useful for a qualitative description of heating effects at the electrodes.

Secondly, the local electrode impedance adds to the output impedance of the TES stimulator. A high local electrode impedance can seriously affect the characteristics of a voltage stimulator. When the local electrode impedance is unknown, absolute V_{th} values cannot be determined. The TES voltage threshold monitoring method of CALANCIE *et al.* (1998, 2001) uses corkscrew electrodes. Their impedance values of type I corkscrew electrodes are within the range of about 550–1050 Ω (Table 1), which is the region where V_{th} is linearly correlated with the impedance at 1 kHz, as shown in Fig. 2. This means that the voltage threshold method may be sensitive to clinically irrelevant impedance changes.

This can be illustrated by an example that is an oversimplification of the real situation. Assume an electrode impedance of $Z_{tot} = 900\,\Omega$, equal to the sum of the head impedance $Z_{head} = 270\,\Omega$ and local electrode impedance of Z_{local} 630 Ω (case A). This implies that $V_{th} = 3.33$. $V_{th,abs}$, in which V_{th} is measured and $V_{th,abs}$ is the absolute threshold voltage. For another case, B, with the same Z_{head} and $Z_{tot} = 300\,\Omega$, this results in $V_{th} = 1.09 \cdot V_{th,abs}$. When Z_{head} is decreased to 200 Ω by scalp oedema, which shunts the net electrode impedance (encountered in our material), V_{th} increases from 3.33 $V_{th,abs}$ to 4.15 $V_{th,abs}$ (25% increase) in case A, whereas, in case B, V_{th} increases from 1.11 $V_{th,abs}$ to 1.15 $V_{th,abs}$ (4% increase). This means that the sensitivity of V_{th} to scalp oedema is 25% in case A, which used corkscrew electrodes, and 4% in case B, which used large contact surface electrodes. The higher local impedance of case A makes the threshold method over six times as sensitive to the effects of scalp oedema as in case B. It is concluded that the sensitivity of the voltage threshold in TES monitoring to changes in the local electrode impedance can be reduced by large contact surfaces.

Thirdly, voltage stimulation basically requires a circuit in which the sum of the output impedance of the stimulator and the local electrode impedances is smaller than the load impedance. When excluding the local electrode impedance, the net load impedance at 1 kHz is between 200 and 450 Ω. When measured during a TES pulse in a patient, the net load impedance is between 100 and 150 Ω (calculated from the curves in Fig. 2 in JOURNÉE *et al.*, 2003). The local impedances of large contact surface electrodes and interconnected type II corkscrew electrodes (data not shown) are small compared with the net load impedance. However, there are, as yet, no commercially available transcranial electrical stimulators that fulfil the basic criterion for TES voltage stimulation (output voltage independent of load).

The only FDA approved TES voltage stimulator is the Digitimer D185. The manufacturer specifies an output resistance of 120 Ω, and the output voltage is specified for an output load of 1000 Ω. The design incorporated surface electrodes originally used by MERTON and MORTON (1980) in clinical applications (KALKMAN *et al.*, 1995) where relatively high local electrode impedances were accepted (UBAGS *et al.*, 1996, 1997). Moreover, for load impedances below 470 Ω, the pulse shape and the duration of the stimuli become distorted, and the voltage available for neural stimulation can drop by over 50% (JOURNEE, *et al.*, 2003).

As the D185 stimulator has a current limit of 1.5 A, the low electrode impedances of the stimulation electrodes sometimes result in an additional decrement in the output voltage. The available stimulation voltage can become inadequate and affect the reliability of neuromonitoring in some patients. The influence of local electrode impedances on threshold currents can also be eliminated by replacing a voltage by a current stimulator, although the threshold current will then be very sensitive to variations in the inter-electrode tissue impedance, such as those resulting from scalp oedema. The choice of electrode type is not important if the voltage reserve of the stimulator is sufficiently high.

5 Conclusions

It is concluded that the correlation between TES voltage threshold and electrode impedance can be reduced markedly by the use of TES electrodes with large electrode contact surfaces, resulting in limit values for these parameters. This may also improve the reliability of TES-MEP monitoring.

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References

BURKE, D., HICKS, R., and STEPHEN, J. (1992): 'Anodal and cathodal stimulation of the upper-limb area of the human motor cortex', *Brain*, **115**, pp. 1497–1508

CALANCIE, B., HARRIS, W., BROTON, J. G., ALEXEEVA, N., and GREEN, B. A. (1998): ''Threshold-level' multipulse transcranial electrical stimulation of motor cortex for intraoperative monitoring of spinal motor tracts: description of method and comparison to somatosensory evoked potential monitoring', *J. Neurosurg.*, **88**, pp. 457–470

CALANCIE, B., HARRIS, W., BRINDLE, G. F., GREEN, B. A., and LANDY, H. J. (2001): 'Threshold-level repetitive transcranial electrical stimulation for intraoperative monitoring of central motor conduction', *J. Neurosurg.*, **95**, pp. 161–168

DELETIS, V., and CAMARGO, A. B. (2001): 'Transcranial electrical motor evoked potential monitoring for brain tumor resection', *Neurosurgery*, **49**, pp. 1488–1489

FOSTER, K. R., SCHEPPS, J. L., STOY, R. D., and SCHWAN, H. P. (1979): 'Dielectric properties of brain tissue between 0.01 and 10 GHz', *Phys. Med. Biol.*, **24**, pp. 1177–1187

GEDDES, L. A., and BAKER, L. E. (1967): 'The specific resistance of biological material—a compendium of data for the biomedical engineer and physiologist', *Med. Biol. Eng.*, **5**, pp. 271–293

GEDDES, L. A. (1987): 'Optimal stimulus duration for extracranial cortical stimulation', *Neurosurgery*, **20**, pp. 94–99

HICKS, R. G., WOODFORTH, I. J., CRAWFORD, M. R., STEPHEN, J. P., and BURKE, D. J. (1992): 'Some effects of isoflurane on I waves of the motor evoked potential', *Br. J. Anaesth.*, **69**, pp. 130–136

JOURNEE, H. L., SHILS, J., BUENO DE CAMARGO, A., NOVAK, K., and DELETIS, V. (2003): 'Failure of Digitimer's D-185 transcranial stimulator to deliver declared stimulus parameters', *Clin. Neurophysiol.*, **114**, pp. 2497–2498

KALKMAN, C. J., UBAGS, L. H., BEEN, H. D., SWAAN, A., and DRUMMOND, J. C. (1995): 'Improved amplitude of myogenic motor evoked responses after paired transcranial electrical stimulation during sufentanil/nitrous oxide anesthesia', *Anesthesiology*, **83**, pp. 270–276

KOTHBAUER, K., DELETIS, V., and EPSTEIN, F. J. (1997): 'Intraoperative spinal cord monitoring for intramedullary surgery: an essential adjunct', *Pediatr. Neurosurg.*, **26**, pp. 247–254

MCADAMS, E. T., and JOSSINET, J. (1994): 'Physical interpretation of Schwan's limit voltage of linearity', *Med. Biol. Eng. Comput.*, **32**, pp. 126–130

MERTON, P. A., and MORTON, H. B. (1980): 'Stimulation of the cerebral cortex in the intact human subject', *Nature*, **285**, p. 227

MERTON, P. A., HILL, D. K., MORTON, H. B., and MARSDEN, C. D. (1982): 'Scope of a technique for electrical stimulation of human brain, spinal cord, and muscle', *Lancet*, **2**, pp. 597–600

MEYLAERTS, S. A., DE-HAAN, P., KALKMAN, C. J., LIPS, J., DE-MOL, B. A., and JACOBS, M. J. (1999): 'The influence of regional spinal cord hypothermia on transcranial myogenic motor-evoked potential monitoring and the efficacy of spinal cord ischemia detection', *J. Thorac. Cardiovasc. Surg.*, **118**, pp. 1038–1045

PECHSTEIN, U., CEDZICH, C., NADSTAWEK, J., and SCHRAMM, J. (1996): 'Transcranial high-frequency repetitive electrical stimulation for recording myogenic motor evoked potentials with the patient under general anesthesia', *Neurosurgery*, **39**, pp. 335–343

POLETTO, C. J., and VAN DOREN, C. L. (1999): 'A high voltage, constant current stimulator for electrocutaneous stimulation through small electrodes', *IEEE Trans. Biomed. Eng.*, **46**, pp. 929–936

SAHA, S., and WILLIAMS, P. A. (1995): 'Comparison of the electrical and dielectric behavior of wet human cortical and cancellous bone tissue from the distal tibia', *J. Orthop. Res.*, **13**, pp. 524–532

STOY, R. D., FOSTER, K. R., and SCHWAN, H. P. (1982): 'Dielectric properties of mammalian tissues from 0.1 to 100 MHz: a summary of recent data', *Phys. Med. Biol.*, **27**, pp. 501–513

UBAGS, L. H., KALKMAN, C. J., BEEN, H. D., and DRUMMOND, J. C. (1996): 'The use of a circumferential cathode improves amplitude of intraoperative electrical transcranial myogenic motor evoked responses', *Anesth. Analg.*, **82**, pp. 1011–1014

UBAGS, L. H., KALKMAN, C. J., BEEN, H. D., and DRUMMOND, J. C. (1997): 'Differential effects of nitrous oxide and propofol on myogenic transcranial motor evoked responses during sufentanil anaesthesia', *Br. J. Anaesth.*, **79**, pp. 590–594

VAN-DONGEN, E. P., TER-BEEK, H. T., SCHEPENS, M. A., MORSHUIS, W. J., LANGEMEIJER, H. J., KALKMAN, C. J., and BOEZEMAN, E. H. (1999): 'The influence of nitrous oxide to supplement fentanyl/low-dose propofol anesthesia on transcranial myogenic motor-evoked potentials during thoracic aortic surgery', *J. Cardiothorac. Vasc. Anesth.*, **13**, pp. 30–34

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