Medical Engineering and Physics 43 (2017) 30–38

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Electric field estimation of deep transcranial magnetic stimulation clinically used for the treatment of neuropsychiatric disorders in anatomical head models

Marta Parazzini[a,*], Serena Fiocchi[a], Emma Chiaramello[a], Yiftach Roth[b], Abraham Zangen[b], Paolo Ravazzani[a]

[a] Consiglio Nazionale delle Ricerche - Istituto di Elettronica e di Ingegneria dell'Informazione e delle Telecomunicazioni, Piazza Leonardo da Vinci 32 - 20133 Milano, Italy
[b] Department of Life Sciences, Ben-Gurion University of the Negev, Be'er Sheva, Israel

ARTICLE INFO

Article history:
Received 20 June 2016
Revised 11 January 2017
Accepted 5 February 2017

Keywords:
Deep transcranial magnetic stimulation
Computational modeling
High-resolution head modeling
Neuropsychiatric disorders

ABSTRACT

Literature studies showed the ability to treat neuropsychiatric disorders using H1 coil, developed for the deep Transcranial Magnetic Stimulation (dTMS). Despite the positive results of the clinical studies, the electric field (**E**) distributions inside the brain induced by this coil when it is positioned on the scalp according to the clinical studies themselves are not yet precisely estimated. This study aims to characterize the **E** distributions due to the H1 coil in the brain of two realistic human models by computational electromagnetic techniques and to compare them with the ones due to the figure-of-8 coil, traditionally used in TMS and positioned as such to simulate the clinical experiments.

Despite inter-individual differences, our results show that the dorsolateral prefrontal cortex is the region preferentially stimulated by both H1 and figure-of-8 coil when they are placed in the position on the scalp according to the clinical studies, with a more broad and non-focal distribution in the case of H1 coil. Moreover, the H1 coil spreads more than the figure-of-8 coil both in the prefrontal cortex and medial prefrontal cortex and towards some deeper brain structures and it is characterized by a higher penetration depth in the frontal lobe.

This work highlights the importance of the knowledge of the electric field distribution in the brain tissues to interpret the outcomes of the experimental studies and to optimize the treatments.

1. Introduction

In recent years, the advent of deep transcranial magnetic stimulation (dTMS) has allowed the stimulation of deep brain regions through the induction of an electric field (**E**) in the brain, to safely and non-invasively modulate the activity of cerebral targets [1]. In particular, the induced **E** can modify the cortical excitability, increasing or decreasing it, depending on the parameters of stimulation [2]. dTMS requires specific coil configurations that have been recently developed, e.g., the family of coils called Hesed (H) coils that are based on certain design principles and have larger dimension compared to conventional circular and figure-of-8 coils [1,3,4,5]. For its ability to reach deeper brain structures than traditional transcranial magnetic stimulation (TMS), dTMS is gaining the interest of several clinical researchers as potential tool for diagnostic and therapeutic purposes [6]. Indeed, previous experiences showed that the regions involved in most neuropsychiatric disorders are placed in non-superficial brain areas and, therefore, the modulation of these deep structures rather than the cortex could be more effective in the treatment of the patients [7]. For example, in the major depressive disorder (MDD), the presence of alterations in the prefrontal cortex (PFC) and in other structures, such as the nucleus accumbens and the ventral tegmental area, interconnected with the dorsal and ventral lateral part of the PFC has been demonstrated by some functional neuroimaging studies [8,9]. These structures belong to the mesolimbic dopaminergic pathway of the reward circuit and are located at a depth of 6–8 cm from the scalp [10].

Several recent clinical studies have reported the use of dTMS delivered by H coils for the treatment of a very wide range of neurological and psychiatric conditions [6], e.g., major depressive disorder [11–16], schizophrenia [17], bipolar depression [18,19], post-traumatic stress disorder [20] and dysthymic disorders [21].

* Corresponding author.
E-mail address: marta.parazzini@ieiit.cnr.it (M. Parazzini).

http://dx.doi.org/10.1016/j.medengphy.2017.02.003
1350-4533/

In summary, all these studies were aimed to modulate the activity of the PFC and of some deeper neuronal regions such as reward-related pathways in order to treat these different neuropsychiatric disorders using in particular the H1 coil (Brainsway Ltd., Israel), i.e., one belonging to the H coil family. The positive outcomes of all these experimental studies encourage further exploration of dTMS for the treatment of these neuropsychiatric disorders; however, more effort should be made to better understand the mechanisms lying behind the results themselves. One fundamental step in that direction is the knowledge of which brain regions are actually stimulated by dTMS and how. More specifically, there is still a gap of knowledge on **E** spatial distributions induced by dTMS in the different cerebral structures and, in particular, in the brain areas which are the targets of the stimulation. Indeed, so far, the only attempts to quantify these distributions were performed both experimentally and computationally, in both cases with some limitations. In the experimental approach, the field distributions were measured in realistically shaped head phantoms [i.e. 3,7,22] in which, however, the brain tissues heterogeneity is not accounted for and with a spatial resolution limited by the size of the probe used for the measurements [23]. As to the computational studies, the most of them used either simplified head geometric models, such as spheres [see 10,24,25] or "human head-like" approximate geometries [see 26,27] or MRI-derived human head models with very few tissues and without any distinction among the different cerebral structures [see 28]. Only a couple of recent studies have used a detailed head anatomical model [29,30], but they were focused on the comparison of different dTMS coils all arbitrarily centred near the vertex (Cz) and they have considered just one head model, disregarding therefore the role of the human anatomical variability on the **E** field distributions. These last indeed depend on both the heterogeneous conductivity of the tissues [23] and on the coil position and orientation on the scalp due to the variable shape of the head surface [31,32].

For these reasons, anatomical human models capable to properly account for the dielectric properties of the various brain tissues, and more realistic coil positions should be used to estimate the **E** distributions in the brain tissues to evaluate the experimental results and to optimize the therapeutic treatments.

This study estimates, by computational electromagnetics techniques, the **E** distributions induced in two anatomical human head models by the H1 coil realistically positioned on the scalp, according to what is done in the experimental studies described above for the treatment of neuropsychiatric disorders. The purpose is to investigate and quantify these distributions in the brain regions of two realistic head models, comparing them with the ones due to the figure-of-8 coils, traditionally used for the TMS treatment of the same pathological conditions, with particular attention to the cerebral targets of the stimulation.

2. Methods

The methodology here applied is similar to the one used in our previous papers [29,30]. In summary, the **E** distributions were estimated using the simulation software SEMCAD X (by SPEAG, www.speag.com), in particular using its magnetic quasi-static solver, based on the scalar potential finite element (SPFE) method and the Biot–Savart's law.

We utilized two anatomical models of the Virtual Family [33] (Ella, a 26 year-old female adult and Duke, a 34 year-old male adult). In the whole body, the models differentiate up to 77 tissues, whereas at brain level, they distinguish the cortex, the white matter, the cerebellum, and some deep brain structures for instance the pons, the midbrain, the hippocampus, the hypothalamus and the thalamus.

Table 1
Conductivities assigned to the different tissues.

Human tissues	Conductivity (S/m)
Esophagus, hypophysis, hypothalamus, pineal body, thyroid gland	0.52811
Air internal, pharynx	0
Amygdala, grey matter, hippocampus, nucleus accumbens, thalamus	0.10954
Artery, blood vessel, vein	0.70001
Bone, mandible, teeth, vertebrae	0.020349
Commissura anterior, commissura posterior, white matter	0.066717
Breast	0.024564
Bronchi	0.31145
Cartilage, ear cartilage, intervertebral disk, larynx	0.17554
Cerebellum	0.12954
Cerebro spinal fluid	2
Connective tissue	0.38575
Cornea	0.43128
Muscle	0.33669
Ear skin, skin	0.1
Eye lens	0.33369
Eye sclera	0.50865
Eye vitreous humor	1.5
Fat, lung, subcutaneous adipose tissues (SAT)	0.08
Marrow red	0.0025837
Medulla oblongata, midbrain, pons, ventral tegmental area	0.0881285
Mucosa	0.0013611
Nerve, spinal cord	0.034567
Skull	0.32
Tongue	0.27812
Trachea, trachea lumen	0.30507

Since the present study aims to evaluate the induced **E** amplitude distributions in the brain regions recognised as the target of the dTMS, in both head models we identified by means of a brain atlas [34] the prefrontal cortex (PFC), the dorsolateral prefrontal cortex (DLPFC) and the medial prefrontal cortex (MPFC), regions that were not specifically segmented in the used models. The same was for the locations of deeper neuronal regions of the reward-related pathways such as the amygdala, the nucleus accumbens and the ventral tegmental area. For the analysis of the **E** amplitude distributions, a geometric shape of appropriate dimensions (ellipsoids) has been modeled and inserted in correspondence of each of these deep brain regions.

For both models, the computational domain was discretized with a mesh step of 1 mm and limited just below the shoulders. Table 1 reports the electrical conductivities assigned to each tissue based on literature data at low frequency [35,36].

The H1 coil is modeled according to the available manufacturer specifications as current paths. The coil is composed of 12 windings which form a base portion (α), a protruding return portion (β) and two contacting return portions (γ), as indicated in Fig. 1a [5]. The base portion has an arcuate shape, which provides a flow of electric current tangential at the orbitofrontal and prefrontal head regions and accommodates the curved shape of the subject's skull [4]. The protruding return portion carries returning current in opposite direction to the one in the brain target area and is put at about 7 cm above the scalp. The γ portion is located over the right and left temporal cortex at the same level as the base portion and connects portions α and β. The current flows clockwise in each winding. The coils was positioned on both head models mimicking the procedure currently done in the clinical experiments that stimulate the PFC through the H1 coil [11–15,17–19,21]. Firstly, the H1 coil was placed on the head model to find its position able to induce the maximum **E** amplitude in the motor cortex region of the abductor pollicis brevis of the right hand (APB). Then, the coil was advanced of 5.5 cm in a line parallel to the sagittal suture