MR based electric properties imaging for hyperthermia treatment planning and MR safety purposes
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Citation for published version (APA):
Balidemaj, E. (2016). MR based electric properties imaging for hyperthermia treatment planning and MR safety purposes

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2 Electric properties imaging

2.1 Electric properties

Tissue electric properties determine the behavior of electromagnetic fields in biological tissue. Electric properties are tissue dependent and are described by the magnetic permeability ($\mu$), the permittivity ($\varepsilon$) and the electric conductivity ($\sigma$). Magnetic permeability describes the degree of magnetization that a material obtains in response to an applied magnetic field. As the magnetization of human tissue is negligible [28], the permeability of free space ($\mu_0$) is assumed for all biological tissue types. Tissue conductivity and permittivity are frequency and temperature dependent and are determined by blood and water content, ionic concentrations [29] and ionic mobility in tissue. Various studies have shown that tumor tissue generally has a higher conductivity than normal tissue due to physiological differences between tumor and normal tissue. These differences have been shown for breast tumors and normal breast [31–33], normal liver, malignant liver tumors and cirrhotic liver [34,30], bladder tumors [35] and gliomas and the normal brain [36].

Currently used values in patient models are mostly based on ex vivo measurements of animal and human tissues [37,38]. Furthermore, there is a large variation in reported values between the different studies shown in a review of the literature [39]. This variation can be explained by the inclusion of tissues of various species and differences in measuring conditions (tissue temperature, in vivo, in vitro and ex vivo). Due to practical and ethical reasons, human in vivo electric property (EP) measurements are scarce. Only easily accessible tissue types (e.g. skin, tongue) [37] and liver [30] have been measured in vivo. Various studies have shown that in vivo conductivity values are higher than ex vivo [40,41]. Hence, determination of in vivo electric properties has recently received increasing attention since these properties are essential for more accurate SAR assessment and subsequent computation of the temperature distribution. In particular the use of an MR system is preferred for this purpose as it is a non-invasive technique and can be easily integrated in the current clinical workflow.

2.2 Magnetic Resonance Imaging

Magnetic Resonance Imaging (MRI) is a non-invasive medical imaging technique which is based on the interaction between radiofrequency (RF) fields and certain atomic nuclei (i.e. 1H hydrogen) in the body when they are subjected to a strong magnetic field, which is referred to as B0 field. In the presence of this magnetic field, the nuclear spins will
precess around the B0 field at the Larmor frequency which depends on the strength of the magnetic field as

$$\omega = \gamma B_0$$

with $\omega$ the Larmor frequency and $\gamma$ the gyromagnetic ratio ($\gamma_{1H} = 42.576$ MHz/T). To create an MR image the transmitter RF coil generates a pulse at the Larmor frequency of a certain nucleus corresponding to the present magnetic field strength. For instance, the Larmor frequency of a $1H$ hydrogen nucleus is 64, 128, and 298 MHz when subjected to a magnetic field strength of 1.5, 3.0, and 7.0 Tesla, respectively. The RF pulse is an electromagnetic pulse consisting of an electric and magnetic field component. The magnetic field of the RF, which is referred to as the $B_1$ field, is perpendicular to the B0 field and, therefore, flips the alignment of the nuclear spins towards the $B_1$ field. The degree of the flip angle is dependent on the shape of the applied RF pulse. For a rectangular RF pulse the flip angle is computed by

$$\alpha(x) = \gamma B_{1+}^+(x)\tau$$

where $B_{1+}^+$ the magnetic field of the transmit RF coil and $\tau$ is the pulse duration.

After the pulse, the nuclear spins return to their original state through the longitudinal and transverse planes. This process is called relaxation. The relaxation times in each plane, referred to as T1 and T2, are tissue dependent and determine the intensity and contrast of MR images. The signal generated by the nuclear spins during relaxation is received by a receive RF coil, which in some cases is the same RF coil as used for transmitting. The received magnetic field is referred to as $B_{1-}^-$. To obtain high quality MR images, a homogeneous B1 field in the volume of interest is required. A birdcage coil can produce a homogeneous field over a large volume within the coil. The homogeneity of the B1 field decreases with field strength and antenna arrays are therefore used to increase the homogenization at higher field strengths. Phase and amplitude steering of the RF pulses is used aiming at increasing the B1 field homogenization, however, care should be taken to prevent high electric fields that might lead to unwanted tissue heating. Therefore, the acquisition of patient-specific electric tissue properties is valuable not only for hyperthermia, but also for MR safety purposes at high field strengths.

### 2.3 Electric Properties Tomography

The interaction between the magnetic component of the RF field and tissue can be exploited to determine the electric properties. Electric Properties Tomography (EPT) is an MR-based method that uses B1 maps to derive the electric properties at the Larmor frequency [42–45]. EPT requires both the amplitude and phase of the $B_{1+}^+$ field. In MR systems the amplitude of the B1 field can be measured by various B1 mapping techniques [46–48]. The phase of the $B_{1+}^+$ field is not directly measurable. However, the
measurable MR signal phase ($\varphi^\pm$) contains contributions from the transmit phase ($\varphi^+ = \arg(B_1^+)$), receive phase ($\varphi^- = \arg(B_1^-)$) and off-resonance effects. The off-resonance effects are reduced by using spin echo acquisition [49]. Using the modern multi-transmit MR systems it is possible to separate the transmit and receive phases [50,51]. Most clinical systems use single or double channel quadrature coil, therefore, separation of the transmit and the receive phase is limited. In general, when using such systems, the contributions of transmit and receive phases are assumed identical. This assumption was investigated for the head at various field strength [43], and was shown to hold up to 3T. The central equation of the EPT method is the homogeneous Helmholtz equation

$$\nabla^2 \frac{B_1^+}{B_1^+} = -\mu_0 \varepsilon_0 \varepsilon_r \omega^2 - i\mu_0 \sigma \omega$$

where $B_1^+$ is the complex transmit field ($B_1^+ = |B_1^+|e^{i\varphi^+}$), $\omega$ is the Larmor angular frequency, $\mu_0$ and $\varepsilon_0$ are the permeability and permittivity of vacuum, respectively, and $\varepsilon_r$ and $\sigma$ are the unknown relative permittivity and conductivity of the object of interest, respectively. Using the measured $|B_1^+|$ and the $\varphi^\pm$ distribution the conductivity can be reconstructed by

$$\sigma = Im \left( \frac{\nabla^2 (|B_1^+|e^{i\varphi^+})}{|B_1^+|e^{i\varphi^+}} \right) \frac{1}{-\mu_0 \omega}$$

and the relative permittivity by

$$\varepsilon_r = Re \left( \frac{\nabla^2 (|B_1^+|e^{i\varphi^+})}{|B_1^+|e^{i\varphi^+}} \right) \frac{1}{-\mu_0 \varepsilon_0 \omega^2}.$$
2.4 Contrast Source Inversion – Electric Properties Tomography (CSI-EPT)

CSI-EPT reconstructs electric properties in an iterative manner and is based on the Contrast Source Inversion method by Van den Berg and Kleinman (1997) [54]. This method was later applied for oil exploration purposes [55] and tissue properties mapping [56] where EM measurements were performed outside the object of interest. The unique situation that MR systems are able to measure fields inside the object of interest brought us to the idea to use CSI in a completely new MRI inversion setting. In this new constellation every measured voxel represents a virtual receiving antenna. The large number of “receiving antennas” combined with measured data inside the object of interest leads to a “less” ill-posed problem compared to the measuring conditions the CSI method is currently applied upon.

CSI-EPT is based on the global integral representations for the EM field quantities and therefore is less sensitive to noise since integral operators act on the measured field data. Finally, this method is assumption-free regarding the local variations of electric properties. The reader is referred to 6.2 for a more detailed description of the implemented algorithm in this study. A more general description of CSI-EPT can be found in 7.2.

References


[40] Schmidt G, Neubauer G, Illievich UM. Dielectric Properties of Porcine Brain Tissue in the Transition From Life to Death at Frequencies From 800 to 1900 MHz 2003;422.


