Magnetic field distribution and signal decay in fMRI using the dipole model and Monte Carlo diffusion modeling.

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Introduction

In functional magnetic resonance imaging (fMRI), the decay of the transverse magnetization is caused by diffusion of proton spins in an inhomogeneous background magnetic field. Especially the effect of vessel radius and deoxygenation of blood outside the vessel is of crucial importance to understand where the Blood Oxygen Level Dependent (BOLD) signal comes from [1]. It is shown in the presented work how the algorithms of the presented 'dipole model' [2], [3] might be used to determine the decay of the transverse magnetization in the presence of diffusion in a background magnetic field distortion. It is shown how the model is implemented onto a computing machine, which algorithms might be used for computing the signal decay in the presence of diffusion in a background magnetic field distortion.

Methods

The use of elementary dipoles without iteration assumes that coupling to the exterior field is weak. This is the case for tissue (blood, bone marrow) and for deoxygenated vessels in brain parenchyma. Monte-Carlo (MC) modeling allows for the diffusion of proton spins. Three methods (from the most simple to the most realistic) with random series initialization have been tested by the authors. The various MC step methods were first tested by computing the radial distribution of the protons after N time steps for M protons. Different time intervals for linear regression of the logarithm of the signal decay for a simple 3-vessel-model were used by the authors and the results were compared. A number of 5 computation series for a vessel model with changing initialization of the random generator underwent, therefore, subsequent statistical evaluation.

The limits of elements can be defined from being impenetrable (probability of transgression equal to 0) to penetrable (probability equal to 1) for diffusing spins. Transgressing of a boundary (a vessel wall) is thereby rejected with the corresponding probability, until a collision free step is obtained. In the vessel model, actually vessel walls are to be regarded as impenetrable, since exchange rates are much longer (500 ms) than TE (100 ms), as discussed in Fisel et al. [4]. To treat diffusing spins passing the limits of our computational domain, a number of 3 different boundary conditions have been used which are based on simple geometrical arguments.

Results

The dipole model allowed for the simulation of different background geometries (susceptibility distributions) and the separation of the contributions of the different spin pools (intravascular-/ extravascular). The combination of elements makes complicated geometries – also those containing multiple vessels – feasible.

Conclusion

Numerical modeling of relaxation processes is possible with the dipole model. The dipole model thereby serves to compute the background field distortion due to the susceptibility distribution brought into the main magnetic field of an MR scanner. Dynamic processes (‘Brownian motion’ caused by diffusion of the water protons) can be modeled using a Monte Carlo method. Results for different MC methods were compared and it was found that all methods tested yield (in a statistical sense) the same result in the temporal and spatial distribution of protons, and hence lead to the same SE and GRE signal decay.

References

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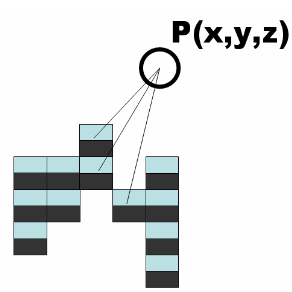

Fig. 1: To the top: Superposition of elementary dipoles is done by computing the field distribution at each point P for each dipole and summing up over all the individual dipoles.

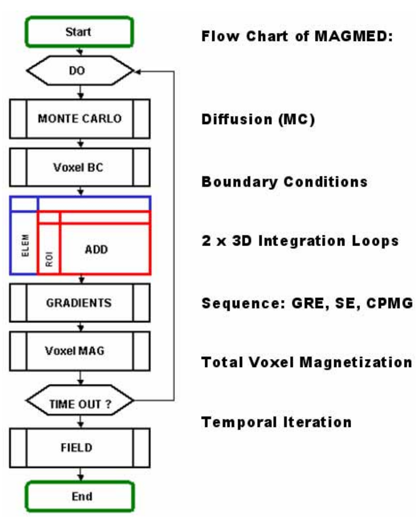

Fig. 2: To the left: Flow chart of the MAGMED algorithm, based on the ‘dipole model’, extended to Monte-Carlo simulation of the diffusion of the individual proton spins.

Fig. 3: To the bottom: Typical decay curves for GRE signal decay rate R2* and SE signal decay rate R2, depending on the vessel radius and for constant volume fraction. The open symbols belong to 500 steps in time; the closed symbols belong to 50 steps in time. The echo time TE = 25 ms is the same for both computations.