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Journal of Magnetic Resonance 173 (2005) 97–115
JMR Journal of Magnetic Resonance
www.elsevier.com/locate/jmr

The SIMRI project: a versatile and interactive MRI simulator

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Received 6 April 2004; revised 10 September 2004
Available online 22 December 2004

Abstract

This paper gives an overview of SIMRI, a new 3D MRI simulator based on the Bloch equation. This simulator proposes an efficient management of the T_2^* effect, and in a unique simulator integrates most of the simulation features that are offered in different simulators. It takes into account the main static field value and enables realistic simulations of the chemical shift artifact, including off-resonance phenomena. It also simulates the artifacts linked to the static field inhomogeneity like those induced by susceptibility variation within an object. It is implemented in the C language and the MRI sequence programming is done using high level C functions with a simple programming interface. To manage large simulations, the magnetization kernel is implemented in a parallelized way that enables simulation on PC grid architecture. Furthermore, this simulator includes a 1D interactive interface for pedagogic purpose illustrating the magnetization vector motion as well as the MRI contrasts.

Keywords: MRI simulation; Bloch equation; Artefacts; Field inhomogeneity; Software

1. Introduction

The simulation of magnetic resonance imaging (MRI) is an important counterpart to MRI acquisitions. Simulation is naturally suited to acquire understanding of the complex MR phenomena [1]. It is used as an educational tool in medical and technical environments [2,3]. MRI simulation enables the investigation of artifact causes and effects [4,5]. Also, MRI simulation may help the development and optimization of MR sequences [4].

With the increased interest in computer-aided MRI image analysis methods (segmentation, data fusion, quantization...), there is greater need for objective methods of algorithm evaluation. Validation of in vivo MRI studies is complicated by lack of reference data (gold standard) and the difficulty of constructing anatomical realistic physical phantoms. In this context, an MRI simulator provides an interesting assessment tool [6] as it generates 3D realistic images from medical virtual objects that are perfectly known.

The simulators previously developed use different approaches and thus differ in closeness to the reality, extent of applicability, and necessary computation effort.

The first category of simulators use proton density, T_1 and T_2 maps computed from a set of images acquired using different repetition and echo times. Using these maps and equations of the image intensity for different pulse sequences, new images are synthesized [7–10]. The simulator proposed in [11] is based on the same approach but provides more realistic images as it includes phenomena such as noise, tissue heterogeneity, correlation between T_1, T_2, and proton density, influence of the field strength on the relaxation times, partial volume effect, and spatial non-uniformity of the signal.

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1090-7807/$ - see front matter
doi:10.1016/j.jmr.2004.09.027

However, this approach does not closely simulate the whole process of MR images formation and thus is not able to simulate all the artifacts encountered in MR images such as chemical-shift, intra-voxel dephasing, imperfection of slice selection, Gibbs phenomenon, aliasing, non-linear gradients, B_0 inhomogeneity, and radio-frequency (RF) inhomogeneity and susceptibility artifacts.

A second category of simulators uses the k-space formalism. The inverse Fourier transform of the spin density image is computed to create the k-space amplitudes [12]. Then, amplitudes are corrected to simulate the pulse sequence and the relaxation phenomena as well as stimulated echoes, a motion or transverse magnetization that propagates through several periods. The same approach was used in [13,14] to simulate tagging. One drawback of this approach is that each tissue type must be treated separately making the simulation of non-uniform tissue characteristics difficult. Moreover, the use of the transformation of the spin density maps implies strict relationships between gradient strength, sampling frequency, and field of view (FOV), thus disenabling the simulation of non-linear gradients on inhomogeneous magnetic field.

A third approach, "hybrid," is proposed in [15]. Each tissue type is associated with a spin model (defined by T_1, T_2, T_2^*, and proton density) that simulates intra-voxel heterogeneity by replacing a spin by a distribution of spins having different frequencies. The NMR intensities of each spin model are computed using the signal equation or a discrete-event Bloch equation computation. Then, images are formed by weighing the tissue distribution for each voxel with the signal of each tissue. Lastly, noise and partial volume effects are introduced in the images. This approach is very interesting regarding the simulation of intra-voxel heterogeneities. Yet, the simulation of non-uniform tissue characteristics (i.e., different intra-voxel heterogeneities conducting to different T_2^*) would require the simulation of a large number of spin models. Moreover, as the simulators of the first category, the hybrid approach can not simulate the whole MR image formation process and consequently the associated artifacts like those linked to the coding gradients neither.

The fourth category of simulators is based on a discrete-event Bloch equation [16] resolution applied on a spin system [1,4,5,17–21]. This approach is the closest to reality and is not limited except by the assumptions of the Bloch equation (no diffusion) and by the computation time. Most of the phenomena encountered during the MR image formation can thus be simulated but one should take care of the number of isochromats used to describe the object. As underlined in [1,15], the use of high number of isochromats per voxel associated to a frequency distribution provides a frame to simulate the intra-voxel dephasing as well as the corresponding spin echoes. However, using insufficient frequency spacing of the isochromats leads to spurious spin echoes, and a too small number of isochromats leads to truncation artifact [15]. The number of isochromats required is also linked to the T_2 constant or to the acquisition bandwidth and reaches 400 [15]. Such a number is much too high as it multiplies the simulation time accordingly. That is why several authors proposed alternate solutions. In [5], a special scheme based on the separation of each magnetization vector in two parts is proposed. In [19], a random spacing of the object points is used to simulate the echo formation properly. In [18,20,21], a linear change of B_0 across the voxel during the application of the gradient is assumed. However, these approaches do not consider voxel isochromat distribution and consequently only simulate an echo in presence of a gradient. The simulator presented in this paper includes an original approach to simulate the intra-voxel dephasing and a specific signal management for spin echo simulations.

This paper proposes an overview of a new 3D MRI simulator named *SIMRI* that is based on the Bloch equation resolution. *SIMRI* includes an efficient T_2^* management simulating properly spin echoes. It takes into account the main static field value and accepts a 3D map of the main field inhomogeneities to simulate the main MRI artifacts (chemical shift, susceptibility artifact, ...). It enables 2D slice selection with different kinds of RF pulse. Also, a parallel implementation adapted to grid technology [22] was developed to overcome the problem of computation time and to achieve the performances required by the targeted applications. Moreover, the kernel used for simulation of 2D or 3D images is also accessible through a highly interactive graphic interface for a better understanding of the MRI contrast phenomena and the spin magnetization vector evolution.

Section 2 of the paper gives an overview of the simulator through the presentation of its main components, i.e., virtual object description, sequence implementation, magnetization computation kernel, RF pulse shaping, and T_2^* modeling. Section 3 introduces simulation results. 1D, 2D, and 3D results are given, including artifact simulations. We focus on the simulator implementation in Section 4 by introducing the sequence programming strategy, the interactive simulation tool and the distributed implementation.

2. Simulator overview

2.1. Simulator overview

The simulator overview is given in Fig. 1. From a 3D virtual object, the static field definition and an MRI sequence, the magnetization kernel computes a set of RF signals, i.e., the k-space. To simulate realistic images,