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Original Research Article

Patient-induced susceptibility effects simulation in magnetic resonance imaging

Josef Axel Lundman [a,*], Mikael Bylund [a], Anders Garpebring [a], Camilla Thellenberg Karlsson [a], Tufve Nyholm [a,b]

[a] Department of Radiation Sciences, Umeå University, Umeå, Sweden
[b] Medical Radiation Physics, Department of Immunology, Genetics and Pathology, Uppsala University, Uppsala, Sweden

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ABSTRACT

Background and purpose: A fundamental requirement for safe use of magnetic resonance imaging (MRI) in radiotherapy is geometrical accuracy. One factor that can introduce geometrical distortion is patient-induced susceptibility effects. This work aims at developing a method for simulating these distortions. The specific goal being to help objectively identifying a balanced acquisition bandwidth, keeping these distortions within acceptable limits for radiotherapy.

Materials and methods: A simulation algorithm was implemented in Medical Interactive Creative Environment (MICE). The algorithm was validated by comparison between simulations and analytical solutions for a cylinder and a sphere. Simulations were performed for four body regions; neck, lungs, thorax with the lungs excluded, and the pelvic region. This was done using digital phantoms created from patient CT images, after converting the CT Hounsfield units to magnetic susceptibility values through interpolation between known values.

Results: The simulations showed good agreement with analytical solutions, with only small discrepancies due to pixelation of the phantoms. The calculated distortions in digital phantoms based on patient CT data showed maximal 95th percentile distortions of 39%, 32%, 28%, and 25% of the fat-water shift for the neck, lungs, thorax with the lungs excluded, and pelvic region, respectively.

Conclusions: The presented results show the expected pixel distortions for various body parts, and how they scale with bandwidth and field strength. This information can be used to determine which bandwidth is required to keep the patient-induced susceptibility distortions within an acceptable range for a given field strength.

1. Introduction

The role of magnetic resonance imaging (MRI) is increasing in radiotherapy. MRI has been shown superior to computed tomography (CT) for target volume definition for several diagnoses [1], and is today frequent in routine clinical use. The superior soft tissue contrast compared to CT and the possibilities for functional imaging using diffusion weighted imaging (DWI) and dynamic contrast enhanced (DCE) imaging are contributing factors to a belief that MRI will be the primary imaging modality for delineation in radiotherapy in the future. The development of integrated MRI and treatment units can push this development even further, and can make MRI the sole imaging modality for large groups of patients, that is, both for definition of target and organs at risk, and for positioning of the patient at treatment [2,3].

Image distortions are a well-known and well characterized problem with MRI [4,5]. The spatial encoding in the MRI acquisition is based on a well-defined relation between the magnetic field strength and the spatial location in the patient. In modern scanners, where the homogeneity of the main magnetic field (B_0) is typically very high, the non-linearity of the gradients is often the dominating source of distortions. These distortions are deterministic, and can be corrected for with good results using distortion correction algorithms provided by all major vendors. The corrections are, however, not perfect and some distortion remains. The accuracy of the corrections can be verified using distortion phantoms [6–8]. However, the patient itself also introduces variations in the main magnetic field. These distortions are not corrected for when applying the distortion correction algorithms of the scanner,

* Corresponding author.
E-mail address: josef.lundman@umu.se (J.A. Lundman).

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and cannot be determined in advance using a phantom due to the dependence on the individual patient anatomy. While the shifts in the magnetic field can be calculated for arbitrary objects with known susceptibilities [9], the distortions are difficult to quantify for the individual patient. The patient-induced magnetic susceptibility (χ) effects on the image can be reduced by increasing the bandwidth, but increasing the bandwidth also leads to a reduction of the signal-to-noise ratio. Shim coils are often used to reduce the effects of susceptibility, but the ability to remove local rapidly changing fields with this method is limited. Methods have been suggested for determining and correcting B_0 inhomogeneities [10–12]. All methods have their own advantages and disadvantages, for example in how they handle the phase wrapping and the requirement of extra scan time [13].

The aim of the current work was to develop a method for simulation of the introduced variations in the magnetic field caused by the patient for different imaging sites in the body. The specific goal was to help objectively identify a balanced acquisition bandwidth for different anatomical regions, such that the patient-induced distortions stay within acceptable limits for use in radiotherapy.

2. Materials and methods

2.1. Method

From previous work by de Rochefort et al. [14], it is known that the local magnetic shifts after Lorentz sphere correction in the Fourier domain, $\boldsymbol{b}_{\text{local}}$, is given by

$$\boldsymbol{b}_{\text{local}} = \mu_0\left(\frac{1}{3}\boldsymbol{m} - \frac{\boldsymbol{k}}{k^2}\boldsymbol{k}\cdot\boldsymbol{m}\right), \tag{1}$$

where $\boldsymbol{k}$ is the position in the Fourier domain, and $\boldsymbol{m}$ is the Fourier transform of the magnetisation ($\boldsymbol{M}$), given by

$$\boldsymbol{M} = \frac{\chi\boldsymbol{B_0}}{\mu_0(\chi+1)}. \tag{2}$$

As demonstrated earlier, [15] a displacement map can be calculated from this through

$$\vec{\boldsymbol{D}}(x,y,z) = \bar{\gamma}P(x,y,z)\cdot\left[\frac{\hat{\boldsymbol{x}}}{\text{BW}_x},\frac{\hat{\boldsymbol{y}}}{\text{BW}_y},\frac{\hat{\boldsymbol{z}}}{\text{BW}_z}\right] \tag{3}$$

where $\bar{\gamma}$ is the resonance frequency, the perturbation map $P(x,y,z)$ is the inverse Fourier transform of $\boldsymbol{b}_{\text{local}}$, BW_i is the bandwidth (Hz) per voxel in the i-direction, and $\hat{\boldsymbol{x}}$, $\hat{\boldsymbol{y}}$ and $\hat{\boldsymbol{z}}$ are the position vectors.

The theory above was implemented in the image data analysis application Medical Interactive Creative Environment (MICE). The application, with our implementation included, can be downloaded from the Gentle Radiotherapy homepage (http://gentleradiotherapy.se/downloads/mice/).

2.2. Validation

Validation of the software was done through comparison of simulations and analytical calculations for a homogeneous sphere as well as a homogeneous cylinder transverse to B_0. The analytical expressions for the shift in magnetic field for these geometrical shapes have been previously derived [16]. For both the sphere and the cylinder, the susceptibility was set to that of muscle, and the susceptibility of the surrounding volume was set to zero. A $441 \times 441 \times 441$ image matrix with a voxel size of $1.37 \times 1.37 \times 3.75$ mm was used for the simulations. A Gaussian filter with a standard deviation of 1 was applied in order to smooth the sharp edges introduced by the discretization of the objects. This approximates the partial volume effects that would occur in a scanned image.

For a sphere with radius a centred at the origin, we have the expressions

$$\Delta B_z = \frac{2\Delta\chi B_0}{3} \tag{4}$$

$$\Delta B_z = \frac{\Delta\chi B_0 a^3}{3}\frac{2z^2 - x^2 - y^2}{(x^2+y^2+z^2)^{5/2}} \tag{5}$$

where Eq. (4) applies inside the sphere, and Eq. (5) outside the sphere.

For a circular cylinder with radius a and transverse to B_0, we have the expressions

$$\Delta B_z = \frac{\Delta\chi B_0}{2} \tag{6}$$

$$\Delta B_z = \frac{\Delta\chi B_0 a^2}{2}\frac{z^2 - x^2}{(x^2+z^2)^2} \tag{7}$$

where Eq. (6) applies inside the cylinder, and Eq. (7) outside the cylinder.

The analytical expression for the maximum field perturbation, ΔB_{max}, of the sphere is

$$\Delta B_{\text{max}} = \frac{2\chi B_0}{3}. \tag{8}$$

2.3. Simulation

Simulations of patient-induced susceptibility effects were performed using digital phantoms based on patient CT images. From these simulations the median, 95th percentile and maximum distortion effect was determined for four body regions; neck, lungs, thorax excluding the lungs, and the pelvic region. The neck region was defined as covering the area between the top of the lower jaw to one slice above the lungs. The lungs and thorax excluding the lungs covered the slices containing lung tissue. The pelvic region was defined as the top of the iliac crest to the perineum. These regions were selected as they cover common areas for radiotherapy.

Eight prostate cancer patients, with body mass index ranging from 24.4 to 31.9, were included in the study after giving informed consent. Acquisition and reconstruction of CT images was performed on a Discovery 690 PET/CT (General Electric, WI, US) at the Nuclear Medicine Department, Umeå University Hospital. The system was equipped with a 64-slice CT scanner. For the CT acquisition, a helical thorax abdomen protocol with 120 kV, Auto-mA, Smart-mA (noise index 35), and 0.625 mm slice thickness was used. The images were reconstructed in 512×512 pixel matrices to a field of view (FoV) of 70 cm with 3.75 mm slice thickness and 3.27 mm slice spacing. The patients were chosen because they, in the course of their treatment, undergo PET/CT examinations which include CT scan protocols covering all the body regions included in this study.

In the simulations, one Hounsfield unit (HU) value was specified for each of four material/tissue types; air, fat, muscle and bone. A corresponding susceptibility was specified for each of these four HUs. The values used can be seen in Table 1. Between the specified

Table 1
Four tissues and the values used for converting HU to magnetic susceptibility.

Tissue	HU	χ (10^{-6})
Air	−1000	0.36[16]
Fat	−120	−7.79[17]
Muscle	100	−9.05[16]
Bone	700	−11.30[17]

Numbers in square brackets are references.