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Effect of gradient field nonlinearity distortions in MRI-based attenuation maps for PET reconstruction

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ABSTRACT

Purpose: Attenuation correction is a requirement for quantification of the activity distribution in PET. The need to base attenuation correction on MRI instead of CT has arisen with the introduction of integrated PET/MRI systems. The aim was to describe the effect of residual gradient field nonlinearity distortions on PET attenuation correction.

Methods: MRI distortions caused by gradient field nonlinearity were simulated in CT images used for attenuation correction in PET reconstructions. The simulations yielded radial distortion of up to ±2.3 mm at 15 cm from the scanner isocentre for distortion corrected images. The mean radial distortion of uncorrected images were 6.3 mm at the same distance. Reconstructions of PET data were performed using the distortion corrected images as well as the images where no correction had been applied.

Results: The mean relative difference in reconstructed PET uptake intensity due to incomplete distortion correction was less than ±5%. The magnitude of this difference varied between patients and the size of the distortions remaining after distortion correction.

Conclusions: Radial distortions of 2 mm at 15 cm radius from the scanner isocentre lead to PET attenuation correction errors smaller than 5%. Keeping the gradient field nonlinearity distortions below this limit can be a reasonable goal for MRI systems used for attenuation correction in PET for quantification purposes. A higher geometrical accuracy may, however, be warranted for quantification of peripheral lesions. These distortions can, e.g., be controlled at acceptance testing and subsequent quality assurance intervals.

1. Introduction

The two imaging techniques positron emission tomography (PET) and magnetic resonance imaging (MRI) both have applications within clinical diagnostics as well as a variety of research fields. Recently, the interest in combining the two modalities has risen [1]. Integrated PET/MRI systems have been developed, despite challenges such as interference between the two systems, and a number of such integrated systems are now available [2]. With these systems, the need has arisen to base the attenuation correction (AC) required for quantification of the PET activity distribution on MRI data instead of computed tomography (CT) data [3].

To build an attenuation map based on MRI data, the MRI intensities, which are dependent on proton density and spin relaxation, have to be converted to electron density information. This conversion is challenging as there is no straightforward physical way to correlate the MRI intensity with electron density. Methods to create attenuation maps from MRI data have been developed based on segmentation [4–6], atlas and machine-learning [3,7–9], and the joint estimation of emission and transmission images [10].

Aside from the challenges of data conversion, MRI images suffer from spatial distortions. These distortions can be caused by gradient field nonlinearities, susceptibility effects, chemical shift artefacts, and B_0 inhomogeneity. Keller et al. [11] recently pointed out metal-induced susceptibility artefacts as one of the pitfalls in MRI-based AC, and correction methods for these artefacts have been proposed [12,13]. Gradient field nonlinearities cause spatial encoding errors, manifested as spatial distortions. The nonlinearities are camera dependent and affect all MRI sequences. Several methods have been developed to correct for distortions arising from gradient field nonlinearities [14–16]. Despite the efforts to correct for spatial distortions in the acquired images, parts of the distortions remain, as shown by Walker et al. [17]. These distortions can be quantified using phantom measurements. Several

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studies have been published based on either 2D [17,18] or 3D geometrical phantoms [19–21]. In phantom experiments, the residual absolute positioning deviations after application of corrections have been shown to be less than 2 mm and 0.7 mm at 15 cm and 10 cm radius from the centre of the field of view (FOV), respectively [17,19,20]. To our knowledge, there are at present no studies providing information on the effect of MRI distortions caused by gradient field nonlinearity when used in AC for PET reconstruction. Such information could for instance be of value during acceptance testing.

This study isolates the effect of gradient field nonlinearity distortions on MRI-based AC for PET reconstruction. The evaluation of induced PET uptake quantification errors resulting from these distortions is performed using patient images of prostate tumours. Attenuation maps based on CT images with simulated MRI spatial distortions caused by gradient field nonlinearities are used for PET reconstruction. The aim of this work is solely to describe the effect of gradient field nonlinearity induced distortions on PET AC.

2. Materials and methods

Eight patients diagnosed with prostate cancer were chosen for analysis. All patients had previously undergone a PET/CT examination. Images were anonymised using the Dicom2usb "one-click anonymization" hardware (http://dicom-port.com/). The patients had body mass indices (*BMI*s) ranging from 24.4 to 31.9.

2.1. Imaging

PET/CT acquisition and reconstruction was performed on a Discovery 690 PET/CT (General Electric, WI, US) at the Nuclear Medicine Department, Umeå University Hospital. The system is equipped with a 64-slice CT scanner and 24 rings of lutetium-yttrium-orthosilicate (LYSO) PET detectors with an axial length of the FOV of 15 cm.

All patients fasted at least 6 h before image acquisition. The patients were given an intravenous injection of ^{11}C acetate (ACE), 5.0 MBq/kg body weight. Image acquisition started 10 min after the injection. The acquisition of PET scans was performed in 3D as static scans over the abdomen to neck region. These scans included 8 bed positions with a scan time of 2 min each.

For the acquisition of CT images, a helical thorax abdomen protocol based on the following parameters was used: 120 kV, Auto-mA, Smart-mA (noise index 35), and 0.625 mm slice thickness. The CT images intended for AC were reconstructed in 512×512 pixel matrices to an FOV of 70 cm with a slice thickness of 3.75 mm and a slice spacing of 3.27 mm, using the reconstruction protocol "PET AC Wide View".

The PET images were reconstructed for all manipulated CT images, to an FOV of 70 cm in 128×128 pixel matrices with a slice thickness of 3.27 mm. VUE Point FX (General Electric, WI, US), a fully 3D iterative reconstruction technique based on ordered subset expectation maximisation (OSEM) including a time-of-flight correction, was used for the reconstruction. The number of OSEM iterations was set to 2 (24 subsets) and a Gaussian post-filter with 6.4 mm full width at half maximum (FWHM) was used.

2.2. Simulation of distortions

Spatial distortions induced by gradient field nonlinearities were introduced by simulation onto CT-images. The use of CT images was chosen to isolate the effect of gradient field nonlinearity distortions, and avoid uncertainties from the conversion of MRI intensities to electron density equivalent information. Before distorting the patient CT images, the couch was removed and the area outside the patient volume set to −1000 HU. The latter was done to avoid artefacts outside the patient, originating from the CT reconstruction, that would not be present in MRI images. To simulate the limited axial FOV of MRI scanners, each patient image volume was divided into subvolumes of 20 cm axial length before distortion and correction were applied, as illustrated in Fig. 1. The patient was centred in the subvolume FOV. An 8 slice overlap was added to the top and bottom of each subvolume to avoid edge effects arising from the simulated distortions.

In MRI, distortions induced by gradient field nonlinearities can be approximated using spherical harmonics [14]. One method used in clinical systems for the correction of these distortions is based on such an approximation to determine the correction. In this study, distorted images were created using the spherical harmonics-based method presented by Janke et al. [14] which uses a camera specific model of the gradients. Based on this model, the distortion field (η_v) is mapped as

$$\eta_v(x, y, z) = \frac{B_v^{\mathrm{N}}(r, \theta, \phi)}{G_v^{\mathrm{L}}} \tag{1}$$

where the subscript v denotes the spatial dimension (*x*, *y* or *z*), $B_{\mathrm{v}}^{\mathrm{N}}$ and $G_{\mathrm{v}}^{\mathrm{L}}$ are the nonlinear and linear gradient field components, respectively, and r, θ and ϕ are given by

$$\begin{cases} r = \sqrt{x^2 + y^2 + z^2} \\ \theta = \tan^{-1}\left(\frac{\sqrt{x^2+y^2}}{z}\right). \\ \phi = \tan^{-1}\left(\frac{y}{x}\right) \end{cases} \tag{2}$$

In this study, distortions were simulated for a Siemens Espree 1.5 T MRI scanner. Spherical harmonics coefficients describing the gradient field were provided by the vendor. In–house developed MATLAB software was used for the mapping of the distortion field. The mapping was performed iteratively until the measurement of convergence was lower than 0.01, or the number of iterations was larger than 200. The convergence was evaluated in a cylindrical volume with a diameter of 45 cm in the transverse plane, and the cylindrical axis centred in the patient volume. The axial length of the evaluation covered the length of the entire subvolume.

The subvolumes with simulated distortions were distortion corrected using the same spherical harmonics-based method as for the distortion. The magnitude of the correction was altered by applying changes in the interval of −30% to 30%, in steps of 10%, to the camera model's lowest order coefficient of each of the *x*-, *y*-, and *z*-directions, of the associated Legendre polynomials. This was done by multiplying all three coefficients with the factors

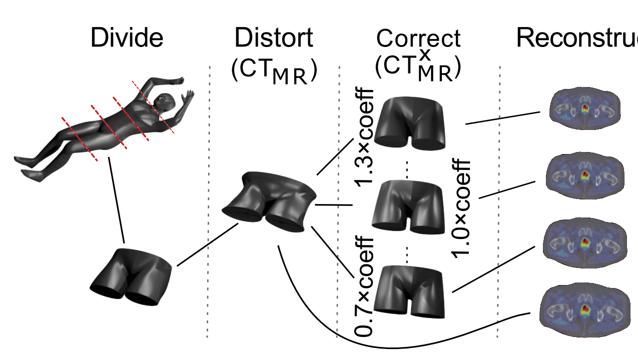

Fig. 1. The CT image distortion process. The patient images are divided into subvolumes with the patient centred in the subvolumes. Each subvolume is then distorted and corrected before reconstructing PET images with AC based on the distorted images ($\mathrm{CT_{MR}}$) as well as on the distortion corrected images ($\mathrm{CT^{x}_{MR}}$).

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