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Treatment planning of intracranial targets on MRI derived substitute CT data

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ABSTRACT

Background: The use of magnetic resonance imaging (MRI) as a complement to computed tomography (CT) in the target definition procedure for radiotherapy is increasing. To eliminate systematic uncertainties due to image registration, a workflow based entirely on MRI may be preferable. In the present pilot study, we investigate dose calculation accuracy for automatically generated substitute CT (s-CT) images of the head based on MRI. We also produce digitally reconstructed radiographs (DRRs) from s-CT data to evaluate the feasibility of patient positioning based on MR images.

Methods and materials: Five patients were included in the study. The dose calculation was performed on CT, s-CT, s-CT data without inhomogeneity correction and bulk density assigned MRI images. Evaluation of the results was performed using point dose and dose volume histogram (DVH) comparisons, and gamma index evaluation.

Results: The results demonstrate that the s-CT images improve the dose calculation accuracy compared to the method of non-inhomogeneity corrected dose calculations (mean improvement 2.0% points) and that it performs almost identically to the method of bulk density assignment. The s-CT based DRRs appear to be adequate for patient positioning of intra-cranial targets, although further investigation is needed on this subject.

Conclusion: The s-CT method is very fast and yields data that can be used for treatment planning without sacrificing accuracy.

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Computed tomography (CT) has many advantages when it comes to radiotherapy treatment planning. It is highly available, provides excellent geometrical accuracy and enables accurate dose calculations. However, the limited soft tissue contrast makes accurate tumor volume definitions difficult. It has been demonstrated that the addition of MRI guidance adds significant value to the target definition of prostate [1,2]. Although multimodal image acquisition is clinically feasible, the extra costs associated with multiple imaging have motivated several groups to investigate the possibility of excluding the CT from the radiotherapy workflow [3,4]. The additional geometrical uncertainty that is introduced with the registration between CT and MRI images has led others to examine the same possibility [5–7].

A precondition for excluding CT from treatment planning is finding a method that will allow accurate dose calculations on MRI images. Some have investigated the approach of disregarding electron density inhomogeneities and calculating the dose distribution in a water equivalent geometry shaped like the patient [8,9]. Results from these studies have ranged from dose differences of 0.9–2.5% compared to CT based calculations. Others have manually segmented the anatomy into different classes, such as soft tissue, bone and air, and assigned them bulk densities [10], yielding dose differences lower than 1% compared to CT based calculations. Although the bulk density approach shows promising results, it is not easily implemented into clinical practice due to the highly time consuming task of manually delineating the tissue classes.

There are two methodologies available when considering automatic conversion of MRI data to electron density data; anatomy and voxel based methods. The anatomy based methods typically employ deformable registration to match a reference CT dataset to a new patient. The disadvantage with this method is the geometric uncertainty that is introduced, especially concerning patients with atypical anatomy [11]. Voxel based methods avoid this complication by directly characterizing individual voxels into tissue classes, which can be assigned bulk densities [12]. The trouble with voxel based segmentation in MRI is the difficulty in discriminating between air and bone, due to the extremely short T2 relaxation time of bone of 0.4–0.5 ms. Therefore, ultrashort echo time (UTE) sequences, which sample the free induction decay as opposed to a spin or gradient echo, are used to acquire signal from bone [13,14]. Recently, a study that employed a voxel based

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Gaussian mixture regression model based on two dual echo UTE sequences and one high resolution T2 weighted image to automatically generate substitute CT (s-CT) data for treatment planning was presented by Johansson et al. [15].

In addition to performing accurate dose calculations, it is also necessary to be able to position the patient at the treatment unit with sufficient accuracy. CT based digitally reconstructed radiographs (DRRs) are routinely used in clinical practice for patient set-up verification. MR images, however, do not contain enough information regarding the bony anatomy of the patient to enable DRR generation [5]. The s-CT images derived from the MRI data on the other hand may be used for this purpose.

In this pilot study, we aim to use the s-CT data generated from the UTE MRI sequences using the method presented by Johansson et al. [15] in a treatment planning workflow. In previous studies, two methods have been presented as possible candidates for MRI based treatment planning; bulk density assigned MRI data and non-inhomogeneity corrected MRI data. We mean to compare the dosimetric accuracy of the s-CT method, as well as the previously mentioned methods to normal CT based calculations. We also aim to generate DRR images from s-CT data to examine the feasibility of an entirely MR based workflow.

Methods and materials

Patients

Datasets with images of five patients, three women and two men, were used in this study. The patient age at examination ranged from 47 to 73 years. The study was limited to targets within the brain.

Imaging

CT images were acquired using a Siemens Emotion 6 (three patients) and a GE Discovery 690 (two patients). Tube voltage was 130 kV for the Emotion 6 and 120 kV for Discovery 690. Slice thicknesses varied between 2.5 and 3.75 mm and in-plane resolution between 0.5 and 1.4 mm. For MR imaging, all patients were imaged with two dual echo UTE sequences and one T2 weighted 3D turbo spin echo (SPACE) sequence using a Siemens Espree 1.5T scanner. All MR images were corrected for geometrical distortions and inhomogeneous coil-sensitivity in the MR scanner using clinical software provided by Siemens [16,17]. The distortion has previously been measured, and although the residual distortion becomes greater toward the edges, we see no indication of it causing problems for treatment planning purposes. Details regarding MR imaging parameters are presented in Table 1.

Image registration

For each patient, all CT and MR images were registered to the first UTE image using a rigid Mattes mutual information image registration algorithm from the Insight Toolkit (ITK). The result of each registration was verified by visual inspection. The SPACE images were resampled using linear interpolation to match the resolution of the UTE images.

Substitute CT generation

Substitute CT images were created using the method described by Johansson et al. [15], where the voxel intensities in the UTE and SPACE images are converted to HUs using a previously estimated Gaussian regression model. In this study, the Gaussian regression model was constructed using MR and CT data from four patients, and was then used to generate the s-CT image for the fifth patient. This procedure was repeated for each patient. In this way the CT image of a patient is not in any way involved in the creation of the s-CT image of the same patient. The resulting s-CT images were resampled to the same resolution as the CT images before they were imported into the treatment planning system.

Bulk density image generation

Bulk density images generated by manual delineation of MR images are time consuming to generate. Therefore, manual delineation was mimicked by extracting the contours of bone and internal air cavities from the CT images using thresholding. Voxels with a CT number below −600 HU were classified as air and voxels with a CT number above 400 HU were classified as bone. The external contour in the bulk density images was derived from the UTE images as described in Johansson et al. [15]. The bulk density images were constructed using mass densities of air 0.001 g/cm^3, soft tissue 1.02 g/cm^3 and cranial bone 1.64 g/cm^3.

Treatment planning

Three treatment plans were generated for every image set, each conforming to a spherical target with a radius of 2.0 cm (Fig. 1). The first target, denoted PTV_{inf}, was centered in the pituitary gland area. The second target, denoted PTV_{med}, was centered in the cerebrum. The PTV_{sup} target volume was placed in the upper part of the brain and was positioned so that it included cranial bone. The multi leaf collimator (MLC) was automatically conformed to the target with 0.5 cm out-of-field margin. Details on the plan setups can be seen in Table 2. Treatment planning was performed using Masterplan (Nucletron BV, Veenendaal, Netherlands), and doses were calculated using pencil beam kernels according to our clinical prac-

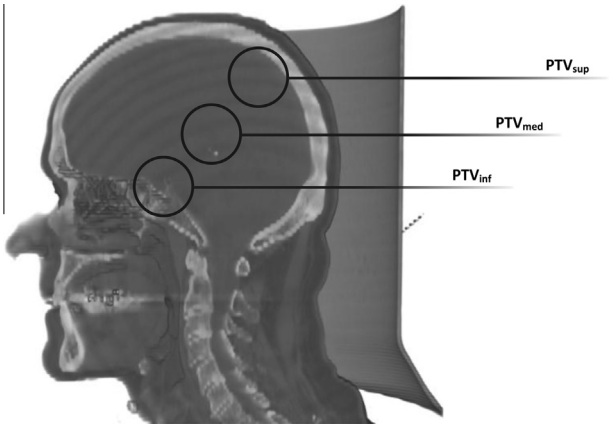

Fig. 1. The placement of the 3 spherical target volumes used in the study.

Table 1
The MR imaging parameters used in the study. *Abbreviations*: TE, echo time, TR, repetition time, FOV, field of view, FA, flip angle.

Sequence	TE (ms)	TR (ms)	Voxel size (mm)	FOV (mm)	FA (°)	Acquisition time (min)
UTE 1	0.07, 3.76	6	1.33 × 1.33 × 1.33	256 × 256 × 256	10	3
UTE 2	0.07, 3.76	6	1.33 × 1.33 × 1.33	256 × 256 × 256	60	3
SPACE	100	1500	0.76 × 0.76 × 1.70	240 × 300 × 204	150	5