Estimation of attenuation maps from UTE derived R_2 images

V. Keereman[1], Y. De Deene[1,2], T. Broux[3], I. Lemahieu[1], and S. Vandenberghe[1]

[1]MEDISIP, Ghent University - IBBT - IBiTech, Ghent, Belgium, [2]Laboratory for Quantitative Nuclear Magnetic Resonance in Medicine and Biology, ECNURAD, Ghent University, Ghent, Belgium, [3]Radiology/BEFY, UZ Brussel, Brussels, Belgium

Introduction

One of the difficulties in the development of multimodality MR-PET scanners is the necessity of attenuation correction to obtain quantitative PET images. Several methods have recently been developed for deriving the attenuation map from a magnetic resonance image [1]. Most of them use a two step approach: first the tissue class of the voxels in the image is defined (bone, soft tissue or air), then the pixels belonging to a certain class are assigned a corresponding linear attenuation coefficient. The segmentation can be performed based on MR signal intensities but many methods also use some sort of template-based anatomic precondition to aid the segmentation. This is necessary for conventional MR sequences, since the very low signal intensity of cortical bone in these sequences makes it difficult to distinguish this tissue type from air. The low signal intensity is caused by fast relaxation of the protons in cortical bone (short T_2 relaxation time). The visualization of cortical bone is however possible using an ultrashort echo time (UTE) sequence, and the segmentation of UTE images into bone, soft tissue and air using a combination of thresholds has been proposed recently [2].

We investigate the use of a quantitative parameter (R_2) that correlates well with the density images obtained with CT. This parameter is derived from UTE images acquired at different echo times. The R_2 relaxation constant is the inverse of the T_2 relaxation time and is high in cortical bone and low in soft tissue.

Materials & Methods

A.   Image acquisition

Part of the shaft of a bovine femur bone was used as a phantom for imaging to investigate the relationship between the R_2 value of a voxel and its density in a CT image. In the phantom large volumes of both hard (cortical) bone and soft bone were present. UTE MR images were acquired on a 3.0T Philips Achieva system (Philips Medical Systems) using a radial UTE sequence and an 8-channel T/R head coil. The UTE sequence samples the FID (TE=0.09 ms) and a gradient echo (TE=1.7 ms). TR was 4.8 ms and the acquisition matrix was 152x152x152 with 1 mm isotropic voxel dimension. The acquisition time was 3 min 44 s. CT images were acquired on a CT scanner (Philips Gemini PET/CT)(16 slice, 200 mAs, 120 kV) with a voxel dimension of 0.4x0.4x0.5mm³.

MR UTE images of the head of a human volunteer were also acquired. The same sequence parameters were used except for a larger isotropic voxel dimension (1.5 mm) and matrix size (172x172x172). The acquisition time was 4 min 47 s.

B.   Image processing

All image processing was performed with methods implemented in the Insight Toolkit (www.itk.org). The images were first smoothed with an edge-preserving gradient anisotropic diffusion smoothing filter to reduce the influence of noise. The R_2-map was derived from the first ($i = 1$) and second ($i = 2$) echo MR image by a voxel-by-voxel least squares fitting:

$$R_2 = \frac{\sum_{i=1}^{N}(TE_i lnS_i - \overline{TE\,lnS})}{\sum_{i=1}^{N}(TE_i - \overline{TE})}$$

Where TE_i is the echo time and S_i the voxel intensity of image i. $\overline{TE}$ and $\overline{lnS}$ denominate the average values. This formula can also be used to fit the R_2 value to more than two images. Voxels which have a very low or zero signal intensity in the short echo image are assumed to contain air and are set to zero in the resulting image regardless of their calculated R_2 value.

The CT image of the bovine bone was registered to the short echo image using a multi-resolution registration method based on a mutual information metric. A joint histogram was then calculated to compare the amplitude of the CT (in HU) and R_2 images.

To illustrate the applicability to human tissue, a transverse slice was selected from the human dataset and segmented using simple thresholds on the R_2-map.

Results

A.   Relationship between R_2 and HU values

In Fig. 1 the joint histogram of the R_2-map and CT image is shown. Two distinct clusters are visible: soft bone, which has HU values comparable with soft tissue, and cortical bone, which has much higher HU values. Both tissue types have respectively a low and high R_2 value range. The clear separation of both clusters implies that the R_2 map can be used to detect cortical bone in MR images.

B.   Segmentation of human images

The R_2-map (Fig. 2) of a transverse slice of the human dataset is shown together with the segmentation based on this R_2-map (Fig. 3). Bone is depicted as white, soft tissue as grey and air as black. The skull (bone) is clearly visible as well as the frontal sinus (air).

Discussion

We have demonstrated the feasibility of using the R_2-map derived from MR UTE images for making the distinction between bone and soft tissue. Segmentation of human MR images into bone, soft tissue and air was also shown. This method could be used to derive the attenuation map needed for PET correction from MR images. Because the R2-map provides a clear distinction between bone, soft tissue and air based only on signal intensities and not on prior anatomic information, this method can even be used in cases where the patient has non-standard anatomic features. This is not possible with template-based approaches. The use of R2 also minimizes any variability caused by the receiver gain.

References

[1] H. Zaidi, "Is MR-guided attenuation correction a viable option for dualmodality PET/MR imaging?," Radiology, vol. 244, no. 3, pp. 639–642, September 2007.

[2] V. Keereman et al., "MR-based attenuation correction for PET using ultrashort echo time (UTE) sequences," Poster presented at IEEE NSS/MIC 2008, October 2008.

Fig. 1

Fig. 2

Fig. 3