Quantitative Imaging of Cortical Bone Using Ultrashort TE (UTE) Sequences

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Background

Mineralized cortical bone has a free water component of approximately 15-20% by volume (1). Bone water plays a key role during mineralization, in which collagen-bound water is gradually replaced by calcium apatite-like mineral. Quantification of bone water may be able to capture changes in porosity during aging and progression of osteoporosis, and provide a means of assessing response to treatment. However, bone appears as a signal void with all types of conventional clinical MR sequences. This problem can be resolved by use of Ultrashort TE (UTE) sequences with TEs of 100 μs or shorter which permit direct imaging and quantification of bone (1-3). In this study we report a fast and efficient 2D UTE technique to quantify T1, T2* and bone water using a clinical 3T scanner.

Materials and Methods

A 2D UTE sequence with a minimal TE of 8 μs was developed on a clinical 3T GE scanner. The 2D UTE sequence is sensitive to eddy currents which result in a broadened slice profile (3-5). A major quantification error comes from contamination from long T2 muscle and fat signals from out-of-slice excitation. To address this problem a long adiabatic inversion pulse (8.6 ms in duration) was employed to simultaneously invert and null long T2 muscle and fat signals. Our simulation showed that more than 80% of the long T2 water and fat signals could be suppressed with an appropriate combination of TR and TI. T2* was quantified using UTE acquisitions with a series of different TE delays. T1 was quantified with a saturation recovery UTE technique where a short hard 90° pulse (232 μs) was followed by UTE acquisitions at a series of different saturation recovery time (TSR) to detect the recovery of bone longitudinal magnetization. Bone water content was quantified by comparing UTE signal of cortical bone with a reference phantom, which was a mix of distilled water (20%) and D2O (80%) doped with MnCl2 titrated to match the ~400 μs T2* of cortical bone. A 3-inch receive only coil was used for T1 and T2* quantification. A quadrature knee coil with lower sensitivity but more homogeneity than the 3-inch or 8-channel knee coil was used for bone water quantification. Bone water ρ_{bone} was calculated using equation [1], where ρ_{ref} is the water proton density of reference, I_{bone} and I_{ref} are the corresponding image intensities, η_{bone} and η_{ref} are the corresponding coil sensitivities, Γ_{bone} and Γ_{ref} are the correction factors due to out-of-slice excitation. f_{xy} describes the behavior of the transverse magnetization as a function of RF pulse b1(t), and T1 and T2 of the short T2 species. Since the reference phantom and cortical bone had a similar T2*, f_{xy} is assumed to be similar. T2* decay is neglected since a short TE of 8 μs was used. The reference tube was put close to the mid-shaft of the tibia, with both close to the coil center, resulting in a similar η_{bone} and η_{ref}. As a result equation [1] can be simplified to equation [2] for bone water quantification. However, the short T2 attenuation is fully recovered for the reference phantom (T1~5ms) but not for bone (T1~224 ms) after the adiabatic IR pulse. A separate compensation factor is necessary for this approach. In total 5 healthy volunteers were evaluated for T1, T2* and water content in tibia. Typical imaging parameters included a FOV of 10 cm, a slice thickness of 6 mm, TR of 300 ms, TI of 120 ms, readout of 512, bandwidth of ±62.5 kHz. The imaging protocol for T1 and T2* quantification was shown in Table 1, where the half projections were progressively undersampled to reduce the total scan time and minimize the effects of patient motion. Bone water quantification was performed without and with the adiabatic inversion pulse designed to reduce out-of-slice long T2 contamination.

$$\rho_{bone} = \frac{I_{bone}}{I_{ref}} \times \frac{f_{xy}^{bone}(b_1(t),T_{1,ref},T_{2,ref})}{f_{xy}^{ref}(b_1(t),T_{1,bone},T_{2,bone})} \times \frac{(1-e^{-TR/T_{1,ref}})}{(1-e^{-TR/T_{1,bone}})} \times \frac{(1-\cos\alpha\times e^{-TR/T_{1,bone}})}{(1-\cos\alpha\times e^{-TR/T_{1,ref}})} \times \frac{e^{-TR/T_{2,ref}}}{e^{-TR/T_{2,bone}}} \times \frac{\eta_{ref}}{\eta_{bone}} \times \frac{\Gamma_{ref}}{\Gamma_{bone}} \times \rho_{ref} \quad (1)$$

$$\rho_{bone} \approx \frac{I_{bone}}{I_{ref}} \times \frac{(1-e^{-TR/T_{1,ref}})}{(1-e^{-TR/T_{1,bone}})} \times \frac{(1-\cos\alpha\times e^{-TR/T_{1,bone}})}{(1-\cos\alpha\times e^{-TR/T_{1,ref}})} \times \rho_{ref} \quad (2)$$

T1 Measurement			T2* Measurement		
TSR [ms]	Readout × Projection	Scan time [min:sec]	TE [μs]	Readout × Projection	Scan time [min:sec]
15	512 × 511	0:31	8	512 × 355	3:33
30	512 × 511	0:46	50	512 × 355	3:33
50	512 × 455	0:59	100	512 × 355	3:33
100	512 × 411	1:35	200	512 × 355	3:33
200	512 × 355	2:33	400	512 × 355	3:33
400	512 × 311	4:18	800	512 × 355	3:33
800	512 × 255	6:54	1500	512 × 355	3:33
Total scan time: **17:36**			Total scan time: **24:51**		

Table 1 Imaging protocol for T1 and T2* measurements of cortical bone of volunteers. The projections were undersampled to reduce total scan time to 18 min for T1 measurement and 25 min for T2* measurement.

Results and Discussion

Figure 2 shows the undersampled UTE images of the tibia of a healthy volunteer with TEs ranging from 8 μs to 1.5 ms. High contrast bone images were achieved with scan time under 3.5 minutes and efficient suppression of long T2 muscle and fat signals with a TR of 300 ms and TI of 120 ms. Excellent exponential curve fitting was achieved and showed a short T2* of 412 μs and T1 of 233 ms for tibia. Figure 3 shows UTE imaging without and with the adiabatic IR pulse. Table 2 summarizes the quantitative values, demonstrating a mean T1 of 224 ms and mean T2* of 389 μs for 5 volunteers with a mean age of 30.4 years. A mean bone water of around 19.7±0.7% and 17.0±0.6% was demonstrated without and with the IR pulse, respectively. This difference may be attributed to two factors: 1) out-of-slice long T2 muscle/fat contamination in UTE without IR; 2) long T2 bone water component suppressed in UTE with IR. Further work is needed to investigate this difference.

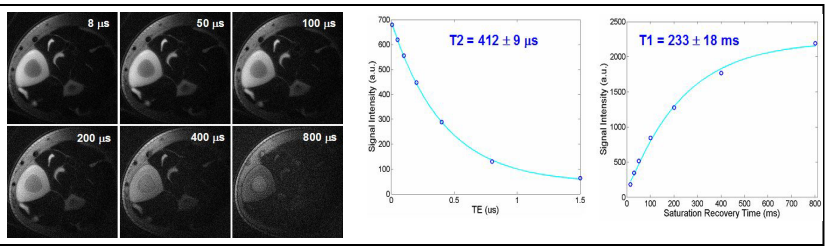

Fig 2 UTE imaging of the tibia of a 31 year old healthy volunteer at progressively increasing TEs within 3.5 minutes required for each TE (a). Long T2 signals from muscle and fat were efficiently suppressed using adiabatic inversion and signal nulling. Mono-exponential fitting demonstrates a short T2* of 412 ± 9 μs (b) and short T1 of 233 ± 18 ms (c).

Conclusion

Quantitative imaging of cortical bone is feasible with undersampled UTE imaging in under 50 min. Long T2 muscle and fat signals can be effectively suppressed with adiabatic inversion and nulling to reduce out-of-slice muscle/fat contamination.

References

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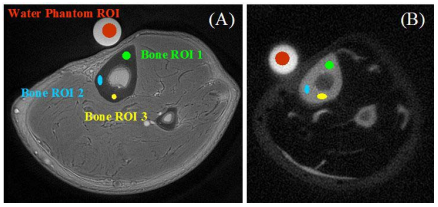

Fig 3 UTE imaging without (A) and with (B) adiabatic IR pulse using a quadrature knee coil. One ROI in reference phantom and three ROIs in bone were drawn for bone water quantification.

Subject	Age	T1 (ms)	T2 (μs)	Bone Water Without IR (% volume)	Bone Water With IR (% volume)
1	26	223 ± 22	376 ± 14	18.8	16.5
2	30	231 ± 17	401 ± 10	19.8	17.3
3	31	233 ± 18	412 ± 9	19.5	16.4
4	35	221 ± 21	387 ± 12	19.8	17.2
5	30	213 ± 20	369 ± 15	20.7	17.8

Table 2 Quantitative measurements of T1, T2* and water content of 5 healthy volunteers show a mean T1 of 224 ms and a mean T2* of 389 μs. A mean bone water content of 19.7% and 17.0% were demonstrated without and with IR pulse.