In vivo measurement of cortical bone bulk susceptibility with ultrashort TE (UTE) pulse sequences

J-T. Chiang[1], J. Du[1], A. Takahashi[2], R. F. Mattrey[1], and G. Bydder[1]

[1]Radiology, University of California San Diego, San Diego, CA, United States, [2]Applied Science Laboratory, GE Health Care

Introduction

Direct MR imaging of cortical bone can be accomplished by ultrashort TE (UTE) pulse sequences that overcome cortical bone's extremely short T2 (1). The bulk susceptibility of bone is a parameter of biological interest, and previously it has been measured *in vitro* in powder form (2,3) and indirectly *in vivo* (4). We present here a simple, direct approach for *in vivo* measurement of cortical bone bulk susceptibility using phase differences obtained by UTE imaging at two different submillisecond TEs.

Method

Magnitude and phase images were acquired from the tibias of two normal, healthy volunteers using a GE 3.0T scanner with a 3 inch receive only surface coil, and UTE sequences with parameters TE 8 and 200 μs, TR 300μs, BW ±62.5kHz, FOV 10cm, FA 30°, Matrix 512 points x 599 spokes. Phase maps were complex high pass filtered (5) to remove low spatial frequency inhomogeneities in the main field B_0. The off resonance frequency was calculated as $\Delta\omega=\Delta\phi/\Delta TE$ using the phase difference $\Delta\phi$ between two TEs.

Note that it is unreliable to calculate frequency $\omega=\phi/TE$ using the phase ϕ from a single UTE scan because off resonance sources accrue more phase than expected for a given TE as conventionally defined for UTE sequences. However, the phase difference $\Delta\phi$ obtained by subtraction from two separate TEs accurately reflects the phase gained during the period ΔTE. Further, subtraction of ϕ from two different TEs removes any constant offsets in ϕ from the coil.

The slice profile in UTE is often nonideal due to constraints on the RF waveform by speed requirements. We therefore also performed UTE imaging with an adiabatic preparation inversion recovery pulse (UTE-IR) of TI=60ms on one volunteer, to reduce possible contamination of the cortical bone signal with water signal from nearby tissues.

Results and Discussion

Figure 1A shows a frequency (Hz) map calculated using the phase difference $\Delta\phi$ between images with TEs of 200us and 8us. Figure 1B shows a frequency (Hz) map using the UTE-IR acquired phase difference between TEs of 200us and 8us. Table 1 shows the mean frequencies obtained for various tissues using ROI analysis.

We expected muscle to be comprised largely of water signal and marrow to be largely of fat signal. Accordingly, using the two echo UTE approach, we measured muscle in the two volunteers to be close to on resonance at 4 ± 32.4 Hz and 18.1 ± 18.4 Hz, and marrow appropriately shifted off resonance at -326 ± 20.4 Hz and -333 ± 19.7 Hz. The frequency shift in marrow reflects a combination of chemical shift from fat and susceptibility effects from trabecular bone (3,4).

Cortical bone was measured to have corresponding frequency shifts of -85.4 ± 69 Hz and -78.3 ± 62.2 Hz. The relatively large standard deviations were due to large inter-pixel variations within the ROIs, and may reflect underlying structural heterogeneity within cortical bone. Assuming that the frequency shift observed in cortical bone is only due to bulk susceptibility differences, we calculated the bulk susceptibility $\Delta\chi_{bone\text{-}water}$ to be 0.65ppm using the equation $\Delta\chi = \Delta\omega/\gamma B_0$. Note that our mean frequency shift and bulk susceptibility values are ~30% of the reported values in the literature (2-4), which we suspect may be related to the observed spatial heterogeneity in cortical bone or out-of-slice contamination from nearby water signals.

To account for the possibility of out-of slice-contamination on cortical bone measurements, we also obtained the phase difference and derived a frequency map using the two echo UTE-IR approach to reduce water signal (Figure 1B). Muscle and cortical bone signals became generally noisier, as reflected by the larger standard deviations in their frequency measurements (Table 1 Column 3). Nonetheless, the measured cortical bone frequency was similar to that from UTE without inversion recovery, suggesting that out-of-slice contamination may not have been significant. However, we are planning additional experiments to minimize the possibility of out-of-slice contamination on cortical bone without sacrificing the accuracy of frequency measurements.

Conclusion

We measured cortical bone bulk susceptibility using a two echo phase difference method via UTE imaging, and estimated $\Delta\chi_{bone\text{-}water}$ to be ~0.65ppm. This simple method permits direct *in vivo* measurement of cortical bone bulk susceptibility, and may be useful for recognition and localization of pathological changes in bone.

References

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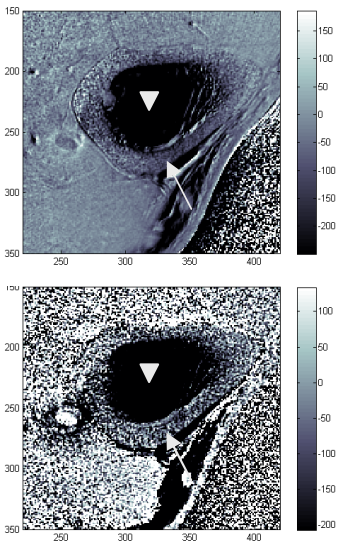

Figure 1A - Frequency map generated by the phase difference from UTE scans of TE 200 and 8μs. Figure 1B - Frequency map generated by the phase difference from UTE-IR scans of TE 200 and 8μs. Both images are windowed to highlight cortical bone frequency shifts (thus marrow frequencies are outside of the depicted range). Arrow indicates cortical bone, arrowhead indicates bone marrow, and surrounding tissue is largely muscle.

	Frequency (Hz) from UTE (Volunteer 1)	Frequency (Hz) from UTE (Volunteer 2)	Frequency (Hz) from UTE-IR
Cortical Bone	-85.4 ± 69 (5893)	-78.3 ± 62.2 (3493)	-75.2 ±155 (3175)
Marrow	-326 ± 20.4 (1891)	-333 ± 19.7 (2001)	-443 ±58.3 (2143)
Muscle	4.0 ± 32.4 (3674)	18.1 ± 18.4 (4159)	85.6 ± 153 (1344)

Table 1 – Mean offset frequencies and standard deviations using ROI (region of interest) analysis. Numbers of pixels in ROI are given in parentheses.