

# Feasibility of in vivo phosphorus imaging of cortical bone at 7T in humans

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## Introduction

Phosphorus concentration of bone plays an important role in the maintenance of bone strength in disorders such as osteomalacia which is characterized by hypomineralization of bone [1]. <sup>31</sup>P MRI potentially provides a noninvasive method to evaluate the degree of bone mineralization. Previous reports have shown that solid-state imaging techniques are suited for acquiring <sup>31</sup>P MR images of specimens and, possibly, in vivo [2-4]. However, since SNR scales as  $\sqrt{(T2/T1)}$  [5], it is difficult to obtain sufficient SNR at desirable resolutions due to the extremely short T2\* (~ 220  $\mu$ s) and long T1 (~ 50 s) of bone phosphorus, as well as its relatively small gyromagnetic ratio. We estimate that <sup>31</sup>P SNR at 7T is intrinsically about 1000 times less than that of muscle water. For this reason, optimization of the imaging hardware, pulse sequence, and post processing is particularly important. In this study, we constructed small surface coils for use in a 7T whole-body MRI scanner and developed a 3D radial concentric-cone imaging sequence to image <sup>31</sup>P of tibial cortical bone in vivo.

## Methods

To estimate the T2\* of cortical bone, we acquired a <sup>31</sup>P spectrum of a lamb cortical bone specimen. For in vivo human imaging, a healthy volunteer was scanned in a supine position in a 7 T MR system (Siemens, Germany) and the cortical bone of the left tibia was chosen as the target at approximately mid-shaft, where cortical bone is thickest. A home-built <sup>1</sup>H transmit-receive surface coil (5x5 cm<sup>2</sup>) first was placed on the tibia to obtain a localizer image. The <sup>1</sup>H coil then was removed and replaced by a <sup>31</sup>P receive-only 2-turn surface coil (3x3 cm<sup>2</sup>). A small reference phantom, consisting of a glass vial filled with synthetic calcium hydroxyapatite, was attached to the top of this <sup>31</sup>P surface coil. Excitation was achieved by a <sup>31</sup>P birdcage head coil, which provided a relatively homogeneous B<sub>1</sub><sup>+</sup> field. For <sup>31</sup>P imaging, a 3D radial cone pulse sequence (Fig. 1) was applied with a non-selective hard pulse, nominal flip angle = 5.6°, pulse duration = 10  $\mu$ s, TR = 250 ms, TE = 30  $\mu$ s, number of cones = 64, total projections = 5332, readouts per projection = 128, dwell time (dw) = 5  $\mu$ s, voxel size = (2.42 mm)<sup>3</sup>, and total acquisition time = 22m13s. Following data acquisition, only 64 points of 128 readouts were used to reconstruct the image. In addition, T1 and T2 modulation correction was applied to compensate for amplitude discontinuities from progressive saturation. The data were reconstructed using a nonuniform fast Fourier transform method combined with density compensation based on the correction of imperfect gradient performance due to eddy currents.

## Results

The <sup>31</sup>P spectrum of lamb cortical bone had a full width half maximum (FWHM) approximately equal to 4.5 kHz corresponding to T2\* ~ 220 $\mu$ s. The central slice of a phosphorus image data set from the volunteer is shown on Fig 2a. The SNR of the cortical bone in this central slice was around 15 at the voxel size of (2.42 mm)<sup>3</sup>. Signal from the upper area of cortical bone was higher than deeper regions due to the restricted sensitive depth of the surface coil. Fig 2b shows a fusion of the central slice <sup>31</sup>P image with the corresponding <sup>1</sup>H localizer image, in which <sup>1</sup>H and <sup>31</sup>P signals are displayed in grayscale and in color, respectively. It can be seen that the shape of the phosphorus signal closely matches the boundary of the cortical bone showing the shape of the endosteal cavity containing fatty marrow. The latter exhibited a small right-left chemical shift artifact along the readout direction in the <sup>1</sup>H localizer image.

## Discussion and Conclusions

The present results demonstrated the feasibility of in-vivo phosphorus imaging of cortical bone at 7T. Although phosphorus in muscle has a much longer T2 compared to that in bone, there was no measurable signal from muscle with the surface coil due to the scan parameters having been optimized for the relaxation properties of bone phosphorus and the greater depth. A previous paper showed an in vivo phosphorus image of cortical bone at 1.5 T using a 3D hybrid UTE sequence with a selective excitation pulse [4]. However, the image slice thickness required was 60 mm due to SNR constraints. Imaging at 7 T somewhat mitigates this problem, along with the use of a 3D hard pulse allowing for an short pulse duration (10 $\mu$ s), which preserves more of the short T2\* signals and permits an "echo time" limited only by the deadline of the coil T/R switch. In summary, our preliminary findings indicate that the proposed method can map the phosphorus content of cortical bone in vivo at relatively high resolution, and may have potential to quantify the change of bone mineralization density during the progression of osteomalacia. Future work will address the necessary steps to achieve quantification of bone phosphorus concentration, which will require correction of the <sup>31</sup>P image intensity for any flip angle variation due to inhomogeneity of the RF transmit field and for the reception sensitivity profile of the <sup>31</sup>P surface coil, as well as for the specific relaxation properties of bone and reference. Following these corrections image intensity can be converted to phosphorus concentration by scaling it with that of the reference phantom of known phosphorus concentration..

## References

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## Acknowledgements

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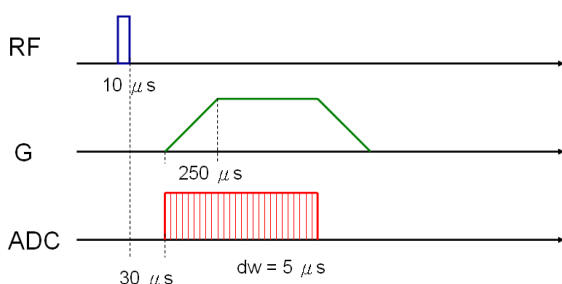


Fig. 1. 3D radial sequence with ramp sampling.

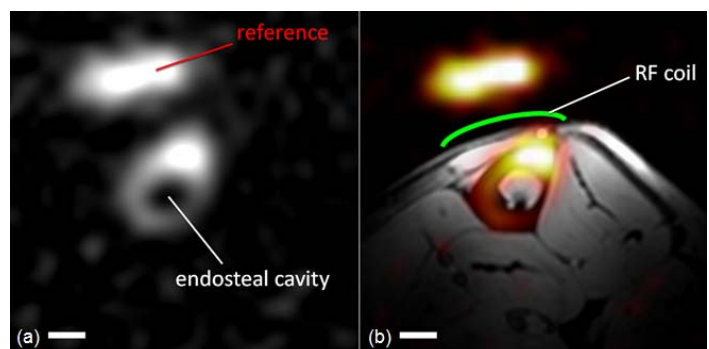


Fig. 2. (a) Central slice of magnified phosphorus image. (b) Fusion image of (a) with the corresponding <sup>1</sup>H image. Scale bars = 1 cm.