High Precision Translational Motion Correction for Micro-MRI of Trabecular Bone Using Cartesian Navigators

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Introduction

There is broad consensus that the micro-architecture of trabecular bone, where most osteoporotic fractures occur, is an independent determinant of the bone's mechanical competence **[1]**. High-resolution MRI has demonstrated potential for *in-vivo* quantification of trabecular network integrity **[2]**. Since scan time is on the order of 10 minutes displacements due to involuntary subject motion even on a sub-millimeter scale usually occur in spite of tight immobilization, causing substantial errors in the derived structural parameters. Correction via navigator tracking of motion can significantly improve image sharpness and thus reproducibility **[3]**. We show that the accuracy of motion sensing and correction critically hinges on navigator signal-to-noise ratio (SNR), which can be optimized by appropriate temporal positioning of the navigator within the framework of an enhanced version of the 3D fast large-angle spin-echo (FLASE) pulse sequence **[4]**. The improved method is evaluated in phantoms and *in-vivo* on the basis of an objective image sharpness criterion.

Methods

The sensitivity of the navigator echo is a function of timing parameters and imaging constraints. Eq. 1 expresses the navigator SNR as a function of its RF pulse parameters, timing T_N and flip angle α_N, employing imaging parameters, T_1 and T_2, as well as sequence parameters, TR, TE, τ_d and τ_c. The two first components of Eq. 1 express the steady-state magnetization in the transverse plane derived from the Bloch equations, the third and the fourth components represent the effect of acquisition time T_{aqc} and T_2-decay, respectively. The number of readout samples of the navigator, N, also affects navigator performance in terms of the trade-off between resolution and SNR, which was determined empirically.

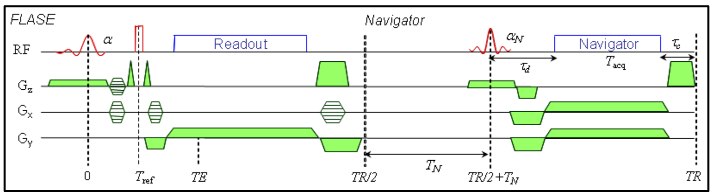

Fig. 1 3D FLASE utilizing a selective RF pulse at t=0 and a non-selective refocusing pulse at time t=*TE*/2. A selective RF pulse of flip angle α_N at time t= $TR/2+T_N$ is applied in a separate slab parallel to the imaging slab, followed by alternating *x-y* gradient echo projection navigators **[4]**.

Navigator SNR was maximized with respect to timing and flip angle of the navigator RF pulse for fatty bone marrow T_2/T_1=60/300*m*sec, with pulse sequence parameters $TR/TE/\tau_c/\tau_d$=80/11/4.7/1.7*m*sec, and in two steps: 1) Maximizing Eq.1 with respect to α_N, the Ernst angle of the navigator is found to be about 140°, 2) Maximizing Eq. 1 considering T_{acq} and T_2-decay, we found T_N=20*m*sec and T_{acq}=13.6*m*sec. Acquisitions with N=16, 32, 64, 128 and 256 were performed with no motion to determine the best N. While there is no significant gain for N larger than 128, we chose N=256. The sequence was designed using SequenceTree4 **[5].**

$$SNR(\alpha_N, T_N) \propto \frac{\sin\alpha_N}{1+\cos\alpha_N \exp(-TR/T_1)} \times \left[1-3\exp\left(-\frac{TR-TE+2T_N}{2T_1}\right)+\exp\left(-\frac{TR+2T_N}{2T_1}\right)+\exp\left(-\frac{TR}{T_1}\right)\right] \times \sqrt{\frac{TR}{2}-T_N-\tau_d-\tau_c} \times \exp\left(-\frac{TR-2T_N-2\tau_c+2\tau_d}{4T_2}\right) \qquad 1)$$

Imaging Experiments: Results obtained with the optimal settings were compared with those used in previous clinical studies, referred to as "standard protocol" (T_N/α_N =10*m*sec/90° and T_{acq}=18.7*m*sec). Two sets of experiments were conducted: Phantom images were obtained for ten different step-based one-dimensional motion trajectories; see Fig. 2a for three of them. Each motion trajectory was intended to mimic abrupt involuntary jerks as they typically occur in patient scans at various time points. Translational displacements were applied manually in a tissue-equivalent phantom that mimics the composition and relaxation properties of muscle and bone marrow as well as the "trabeculae" in the distal tibia and fibula. *In-vivo* images were obtained at the distal tibia in a human subject. All images were acquired at 1.5T (Siemens Sonata™, Erlangen, Germany) using a custom-built two-element receive-only ankle coil, within 8min, FOV=70×64×13mm^3 and matrix size of 512×460×16.

Motion Detection and Correction: 1) 1D navigator projections were obtained by FFT of each *x*- and *y*-readout; 2) *x*- and *y*-projections were summed over slice encoding direction (16 slices) to make *x*- and *y*-gradient projections P_i^x and P_i^y with maximum SNR; 3) P_i^x and P_i^y were correlated with reference projections P_{ref}^x and P_{ref}^y taken from the *k*-space center; 4) translations, Δx and Δy, were chosen to maximize the correlation; and 5) measured displacements were converted to phase shifts applied to the *k*-space data. Performance of the motion detection was then evaluated by the normalized gradient squared (NGS) metric **[6]** as a quantifier of image sharpness.

Results and Conclusions

Figs. 2 and 3 show images of the phantom and the human distal tibia corrected using the two navigator protocols. The *in-vivo* images acquired with the two different navigator settings had NGS values within 0.01% of each other before correction.

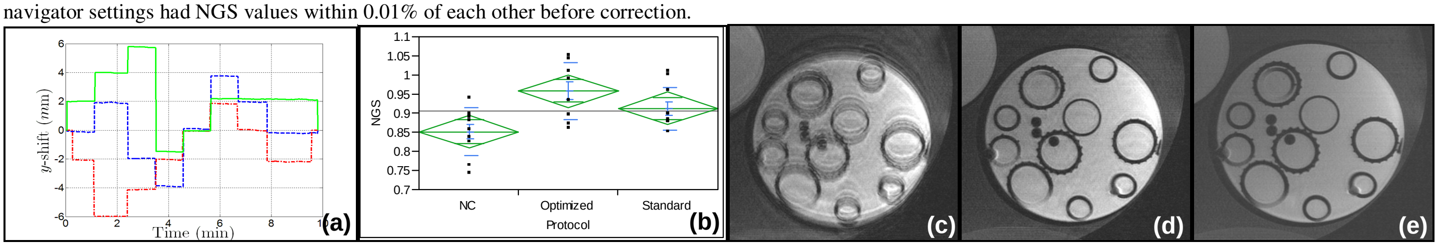

Fig. 2. (a) Motion trajectories, (b) Image sharpness for non-corrected images (NC) and motion corrected images for the two navigator settings (p=0.003 and 0.034, respectively). Corrected phantom images: (c) Standard; (d) Optimized, (e) Reference (still), with the NGS values of 0.8753, 1.0548 and 1.0848, respectively.

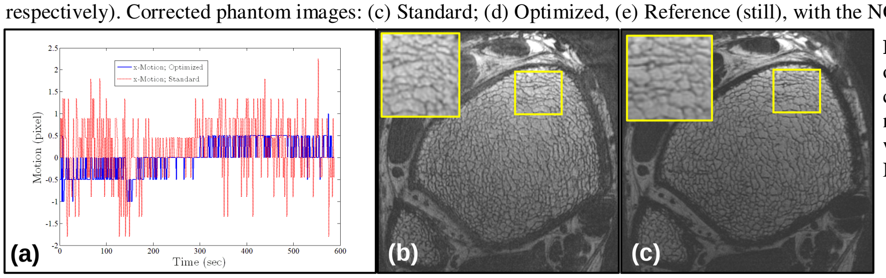

Fig. 3 (a) Motion trajectories estimated with standard and optimized protocols; trajectory detected by the former is considerably noisier than the one detected with the optimized navigator protocol. *In-vivo* Images of the distal tibia corrected with the two navigator p: (a) standard; (b) optimized, yielding NGS values of 0.6548 and 0.8372, respectively.

The data show that navigator SNR is critical for accurate motion tracking and correction in micro-MRI of trabecular bone *in vivo*. Performance improvements achieved are expected to yield improved reproducibility for the study of drug intervention in patients undergoing treatment for osteoporosis.

References: **[1]** Kleerekoper M *et al., Calcif Tissue Int.* 37, 594 (1985). **[2]** Wehrli FW *et al.*, JBMR 23, 730 (2008). Zhang XH *et al.*, *JBMR* 23, 1426 (2008). **[3]** Song HK, Wehrli FW, *MRM* 41, 947 (1999). **[4]** Magland JF *et al.*, *MRM*, in press (2008). **[5]** Magland J, Wehrli FW. *Proc. ISMRM* (2006). **[6]** Lin W, Song HK, *JMRI*, 24 (2006).

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