On-line Prospective Registration of Trabecular Bone MR Images for Longitudinal Examinations

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Introduction: In longitudinal micro-MRI studies of trabecular bone (TB) designed to evaluate structural changes in response to intervention, follow-up scan volumes do not typically align exactly with the baseline scan volumes due to inaccuracies in patient positioning and scan prescription. Failure of accurate registration of the follow-up to the baseline images introduces errors due to the inherent heterogeneity of the trabecular network and anisotropic voxel size. Misaligned image volumes can be matched using retrospective registration techniques [1, 2]. However, this approach often results in blurring due to interpolation and image transformations. These limitations can be overcome by incorporating on-line prospective registration into the data acquisition protocol.

Theory and Methods: The method is based on registering 3D localizer images acquired first at baseline to those obtained in repeat examinations, using an algorithm relying on maximizing the correlation between datasets obtained at different time points. The 2D slice-based cross-correlation

$$\mathbf{C} = \left| F^{-1}([F(\mathbf{L}_b)]^* F(\mathbf{L}_f)) \right|$$

has been used to estimate the 3D translational displacements (Δx, Δy and Δz,) between baseline, $\mathbf{L_B}$, and follow-up, $\mathbf{L_f}$, localizer with F representing the Fourier operator. The rotation around the z-axis of the scanner coordinate system has been calculated by stepping though a series of angles and locating the maximum of **C**. Rotations around x- and y-axes (α and β) can be estimated using the variation in Δy and by Δx across the slices.

A 3D phantom, and the distal tibia of a volunteer, were scanned fifteen and eight times, respectively, at 1.5T (Sonata, Siemens Medical Solutions) with a custom-designed RF surface coil [3]. For the human subject study, prior to acquisition of the high-resolution 3D baseline image (TR = 80ms; TE = 10.5 ms; voxel size = 137x137x410 μm^3) a 3D localizer (TR = 20 msec; TE = 4.2 msec; FOV = 80x80x256 mm^3) was acquired, both with the same patient positioning and landmarking. During each follow-up examination a 3D localizer was acquired again, followed by execution of the registration program that takes the baseline and follow-up localizers as input and returns the translational and rotational correction parameters needed for the high-resolution follow-up scan. The transformation parameters are input by the scanner operator to prescribe the follow-up scan volume via the scanner's graphical user interface. For the purpose of evaluation, a second follow-up 3D localizer has also been acquired (not required in the clinical mode). Thus, if the alignment procedure performs as expected, the transformation parameters produced by registering the baseline and the second localizer images should be equal or close to zero. For the phantom study, before acquiring each of the 15 follow-up localizer images the location and the orientation of the imaging slab was changed in all six degrees of freedom by known amounts relative to the baseline data. This operation was performed with the graphical user interface of the scanner software. The transformation parameters generated by the registration program were then compared with those originally prescribed to calculate the error in registration for each follow-up scan.

Results and Conclusions: Figure 1 shows the correlation between the applied and detected transformation parameters from the phantom study, Figure 2 the corrected misalignments correlated with those present before registration for the volunteer data. The performance of the registration method is illustrated in Figure 3 in which baseline and follow-up high-resolution images of a subject are shown after prospective registration. The relative alignment between a partially segmented baseline and follow-up localizer image of a tibia before and after registration is shown in Figure 4 in which the close alignment is visually evident. Whereas most 3D registration procedures are not fast enough to incorporate on-line registration within the pre-scan portion of a clinical scan sessions, the fast Fourier correlation method presented here takes advantage of the fact that α and β (rotations around x- and y-axes) are always smaller than γ (rotation around z-axis) thereby significantly reducing computation time. As a result, the additional protocol time for the technique is on the order of three minutes, including acquisition time for the 3D localizer, image processing, on-line registration and operator interaction.

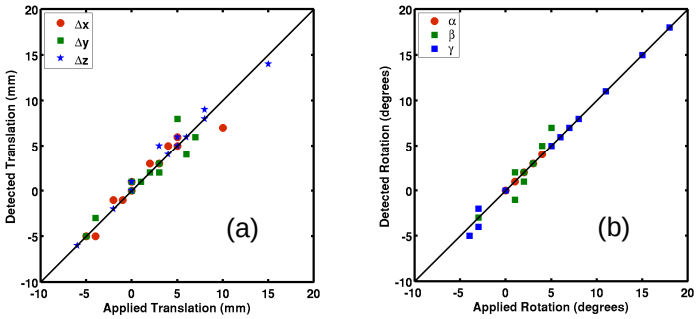

Figure 1: Applied versus detected a) translational and b) rotational displacements between baseline and follow-up phantom experiments. Solid line is line of identity.

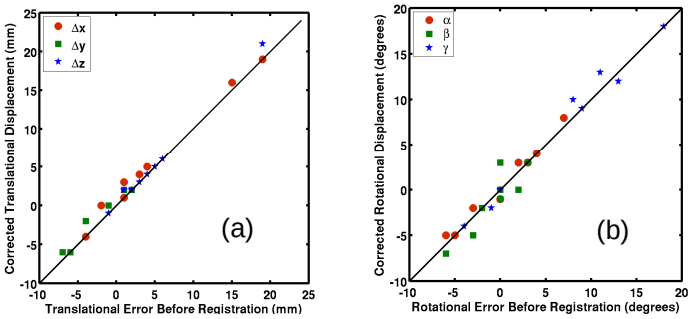

Figure 2: Correlation between detected and corrected follow-up volume misalignment for *in-vivo* studies: a) translational, b) rotational misalignment.

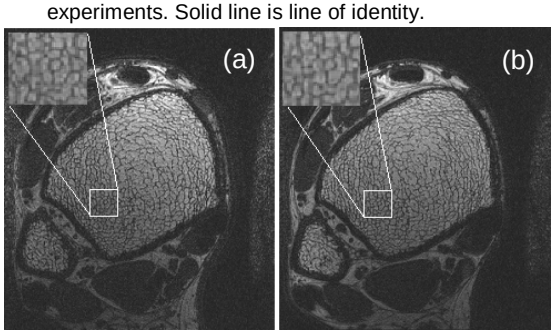

Figure 3: Section of high-resolution *in-vivo* a) baseline and b) follow-up 3D FLASE image data sets after prospective registration. The similarity of the trabecular patterns is clearly recognizable in the zoomed insets. Voxel size was 137x137 μm^2 in the xy-plane and 410 μm in the z-direction.

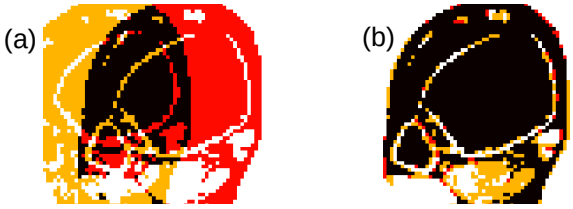

Figure 4: Axial slice of a partially segmented tibia from baseline and follow-up localizer images, a) before and b) after registration. Red, yellow, and black correspond to baseline, follow-up, and regions of overlap.

References [1] Magland et al., Proc. ISMRM, 2006 [2] Vasilic et al., Proc. SPIE MI, 2006 [3] Gomberg et al., Bone, 2004.

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