High Resolution Image Co-Registration *via* Phase Modulation in the Reciprocal (k) Spatial Domain

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Introduction: Obtaining appropriate temporal resolution in dynamic studies such as fMRI and DCE-MRI often requires sacrifice of spatial resolution and/or signal-to-noise ratio (SNR). Algorithms that interpolate low-resolution images for sub-pixel co-registration can be computationally burdensome and can potentially introduce artifacts (1, 2). Herein, we demonstrate a modified application of subunit coordinate translation (1). This is a fast, fully automated method for high-resolution translational co-registration of low SNR, low-resolution images. This method is applied to k-space data and does not introduce the filter artifacts endemic to finite spatial filters (2). It requires no additional data acquisition or pulse sequence modification. (In the form presented here, this method corrects for only translational motion in one, two, or three dimensions, which is the most common non-physiological motion associated with animal studies. Restraints generally prevent rotational motion.)

Theory: The inverse [two-dimensional] Fourier Transform (IFT) of an image, $A(x_{l+rx},y_m)$, can be normalized to that of a reference image $B(x_l,y_m)$, where r_x represents the number of pixels A is shifted from B in the x spatial dimension. This normalization results in a linear phase modulation in the k-space dimension corresponding to that of the spatial shift seen between the two images of a phantom sample (**Figure 1**). According to the shift theorem (3), the phase modulation in k-space in the direction corresponding to the spatial shift can be described in one dimension as:

$$S\{\omega_i\} = IFT[A(x_{l+ri}, y_m)] / IFT[B(x_l, y_m)] = C \cdot \exp(\mathrm{i} \cdot \omega_i \cdot n_i), \quad [1]$$

where $\omega_i = 2\pi\, r_i / p_i$ for each direction i = x, y, or z. n_i is the point index in the k-space dimension in which the shift, r_i, occurred, and p_i is the number of k-space points acquired in the dimension of the shift. The index n_i, (the phase encoding step or read-out point number), is shifted so that the most intense k-space point (generally the center) is assigned a value of zero. The proportionality C arises if there are amplitude differences between the two images. ω_i, and thus the shift, r_i, may then be determined with an optimized non-linear least squares algorithm. The shifted spatial image A is then co-registered in the x spatial direction with the reference spatial image B by the following operation:

$$A(x_l, y_m) = FT\left[\exp(-\mathrm{i} \cdot \omega_i \cdot n_i) \cdot IFT\left[A(x_{l+ri}, y_m)\right]\right] \quad [2]$$

This operation is repeated for each direction in which motion occurs.

Materials and Methods: This analysis was carried out on rat brain DCE-MRI data. These were acquired with an 11.74 Tesla magnet (Varian) operated with a Bruker Avance II spectrometer. This study was approved by the local IUCAC. T_1-weighted, gradient echo images were acquired every three seconds for 50 minutes, followed by a rest period of 10 minutes. This cycle was repeated for 6.8 hours, during which time the animal was provided intravenous fluid, oxygen and gaseous anesthic (isoflurane), and a warm water bed for body temperature maintenance. Oxygenation and heart rate were monitored over the entire experiment. A bolus dose of 0.1 mmol/kg body weight gadoteridol was delivered *via* a tail vein catheter 5 minutes (or 100 images) into the first 50 minute image acquisition block. This additional 5 minutes of imaging prior to bolus delivery required the first imaging period to be 55 minutes long. After the data were transferred to a personal computer, images acquired after the first 10 minutes of scanning (*i.e.*, the first 200 images) were time-averaged in blocks of 20 images each, resulting in a total of 545 images. The in-plane resolution was 200 microns.

Results: The correction of one representative time-course image is shown in **Figure 2**. **Figure 3** is a graphical representation of the time-dependence (abscissa) of linear k space phase modulation (ordinate) in the direction corresponding to that of the translation, before (top) and after (bottom) co-registration. **Figure 4** is a plot of the estimated spatial translation over the entire data time-course (6.8 hr).

Discussion: After the data were loaded into Matlab, the entire computation for co-registration of 545, (128 x 256) matrix images, took less than one minute. The only input required from the user is to designate the reference image and make an initial guess for ω_i. In addition to these benefits, the greatest advantage of this high resolution coregistration method is that it is that this method is considerably less intrusive on the data than methods involving spatial interpolation filters.

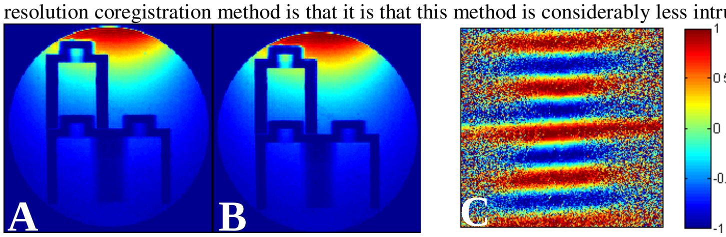

Figure 1. Images of a phantom. After image A was acquired the phantom was physically shifted down before acquiring image B. Dividing, or "normalizing", the k-space image of A by the k-space image of B gives the linear phase modulation shown in C. A and B are absolute value images. C displays only real channel data.

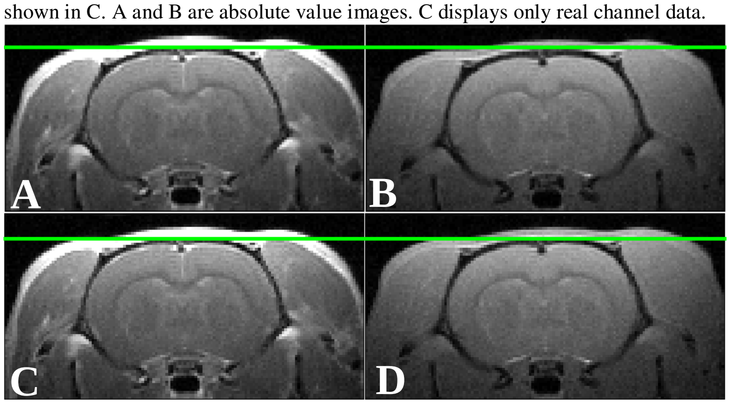

Figure 2. The image in A was acquired 5.5 minutes after delivery of the contrast reagent bolus. The image in B was acquired ~6.5 hours later. It is clear that the image in B has moved down approximately 1 pixel (~200 microns) over that period. The images in B and D are the same, but the k-space of D has been phase modulated from that of B so that it is registered with C (which is the same as A). The green line is to guide the eye to this subtle correction.

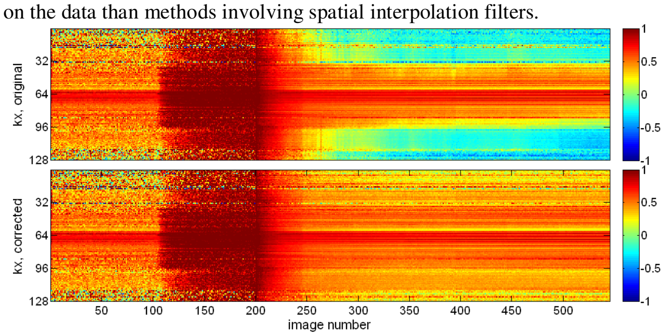

Figure 3. These two plots show the real data of the sum of the central 21 vertical k-space lines for the 545 "normalized" images in the time-course (see eq[1]). The top plot is of unregistered data, and a linear phase in the vertical (shift-corresponding) direction is evident in the right half of the figure. The bottom plot is of registered data and it is clear that the linear phase has been removed.

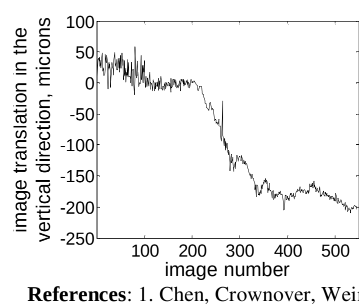

Figure 4. The time-dependence of the image translation magnitude (ordinate). This plot was obtained by solving eq[1] for r. Multiplying r by the in-plane resolution (200 microns) gives the ordinate value. Resolution of the sub-voxel correction is quite high, especially for the time-averaged images (image number > 200).

References: 1. Chen, Crownover, Weinhous, *Med.Phys* **26**:1776(1999).
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This work was supported by RO1-EB00422, and RO1-NS40801.