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Original contribution

Motion-compensated reconstruction of magnetic resonance images from undersampled data

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ABSTRACT

Magnetic resonance imaging of patients who find difficulty lying still or holding their breath can be challenging. Unresolved intra-frame motion yields blurring artifacts and limits spatial resolution. To correct for intra-frame non-rigid motion, such as in pediatric body imaging, this paper describes a multi-scale technique for joint estimation of the motion occurring during the acquisition and of the desired uncorrupted image. This technique regularizes the motion coefficients to enforce invertibility and minimize numerical instability. This multi-scale approach takes advantage of variable-density sampling patterns used in accelerated imaging to resolve large motion from a coarse scale. The resulting method improves image quality for a set of two-dimensional reconstructions from data simulated with independently generated deformations, with statistically significant increases in both peak signal to error ratio and structural similarity index. These improvements are consistent across varying undersampling factors and severities of motion and take advantage of the variable density sampling pattern.

1. Introduction

Motion affects a wide range of magnetic resonance imaging (MRI) acquisitions. For instance, breathing or respiratory motion can move internal organs like the heart, liver, or kidneys during a scan. Cardiac motion can distort images of the heart as well. Gating or triggering (with a respiratory belt, electrocardiogram, or navigator) can suppress these motion effects, but at the cost of less efficient acquisitions. Inconsistent breathing, arrhythmia, or inaccurate physiological monitoring also can diminish the effectiveness of such methods. Additionally, such methods do not address other sources of motion, such as patient movement (bulk motion). Certain patient populations, including young children, cancer patients, and those with neurological or neurodegenerative disorders have difficulty lying still for extended periods and would be expected to move during longer scans. To deal with all these types of motion, for fast, high resolution MRI, motion-corrected reconstructions are leading the way.

In this work, we are interested in suppressing bulk and respiratory motion that can affect high-resolution abdominal images, such as those used to monitor and plan treatment for liver cancer (e.g., hepatoblastoma) in young children. These images can suffer from intra-frame motion blur due to both bulk motion and respiratory motion. While sedation can address the former, doing so precludes breath-holds, poses risks for young children, and requires an anesthesiologist to be present. The need for an anesthesiologist makes scheduling such exams challenging. Even without sedation, repeated breath-holds may be difficult for these children, limiting the achievable resolution and image quality of abdominal images. Alternative imaging techniques such as X-ray computed tomography pose radiation-related risks to these subjects, motivating un-sedated imaging techniques for pediatric MRI. This work, focusing on intra-image (or intra-frame) motion, complements other recent developments addressing inter-frame motion in body imaging, including in dynamic imaging [1–3], multishot imaging [4], and in 4D flow imaging [5–7].

To handle both bulk and respiratory motion, this work focuses on non-rigid correction of intra-frame motion. This contrasts with rigid or translational motion correction that is common for intra-frame motion using k-space navigators (e.g., autofocus [8]). Parallel imaging and sharpness criteria like gradient entropy extend the autofocus approach to non-rigid motion [9,10]. However, these navigators have limited compatibility with spiral-out pulse sequences that can be used for fast imaging. Another approach, MOCCA [11], senses motion without

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navigator sequences, but is primarily used for gating. Inter-frame non-rigid motion correction methods avoiding navigators include non-rigid registration [2–4], k-t FOCUSS [12], MASTeR [13], and BLOSM [14], rank constraints [15,16], and the distance-sensitive regularization in PRICE [17] and other methods [18]. But, using these methods to correct intra-frame motion unnecessarily reduces the amount of data per reconstructed image by a factor equal to the number of groups or "frames", and these methods do not specify how to combine the reconstructed time series into a final image. Alternatively, low-resolution image frames can be used for rigid or non-rigid registration, to resolve motion before reconstructing high-resolution video that takes advantage of spatiotemporal redundancy [19].

Incorporating motion estimation into image reconstruction is a highly flexible motion compensation solution. K-space pre-processing is not flexible enough to correct non-rigid motion [20], and non-rigid image registration post-reconstruction [21,22] cannot address intra-frame motion. Methods such as GRICS [23–25] correct respiratory motion via a parametric model of deformable motion in the reconstruction. GRICS requires specifying respiratory phase or other pseudo-navigational information, but avoids traditional navigation. Instead of constraining our deformable motion to a fixed parametric model, we use multi-scale motion estimation, which was previously applied to image registration [26]. This approach is geared to address arbitrary motion without specific tracking. This model-based reconstruction technique relies on variable sampling densities present in non-Cartesian acquisitions (e.g., spirals) used for highly accelerated imaging. Other methods take advantage of radial sampling to aid motion estimation in time-series imaging [27–31]. Many other works also address motion correction and are described in reviews on medical image registration [32] and motion correction methods [33].

A conference abstract [34] described a single-scale version of our motion-compensated reconstruction. Fig. 1 depicts the reconstruction of an image corrupted by such motion by grouping the undersampled k-space data acquired sequentially in time, and jointly estimating the non-rigid motion present for each group along with the reconstructed image. That framework is extended here to use multi-scale motion estimation and parallel imaging for undersampled non-Cartesian k-space data. This paper also includes a comparison against an existing intra-frame motion correction technique (non-rigid autofocus with locally linear translations [9]) and an illustration of the multi-scale motion estimation technique. This paper also illustrates the advantage of non-constant sampling density for this method. These experiments operate with synthetic motion, so a ground truth image with identical contrast and resolution is available for comparison. An additional experiment with real motion indicates that the proposed method can suppress some motion even when correcting motion for each spiral readout individually.

The next section describes the multi-scale non-rigid intra-frame motion compensation method in detail. Section 3 describes the data and evaluation criteria used to validate and compare this method. Experimental results follow in Section 4 and are discussed in Section 5.

2. Theory

When acquiring a single image (or a frame of a video or time series), if the time-scale of intra-frame motion is comparable to the acquisition time for that image, and if the spatial-scale of that motion is comparable to the resolution of the image, that motion can introduce blur or other artifacts. These artifacts are caused by the image content (e.g., organs or tissue) being in different positions for each k-space readout. To reconstruct images without these artifacts, this method adjusts the forward model to include these intra-frame displacements, modeled as non-rigid motion. Ideally, we would estimate such a deformation for every single k-space readout (like autofocus can via navigation). In practice, individual readouts do not contain sufficient data to calculate these deformations. Instead, we group readouts consecutive in time into a single subset and assign all the readouts in a given subset the same non-rigid motion. The group size should be selected to reflect the time scale of discernable motion; using too many groups would unnecessarily trade off data availability for minimal motion reduction.

Defining an unknown "source" image as I to be the target of the reconstruction, we let the transformation $T(w_g)$ map the source image to the deformed state corresponding to the gth group of k-space m_g. This transformation $T(w_g)$ corresponds to a b-spline-based local interpolation of an image onto a new grid that corresponds to adding the displacement vectors w_g to the original (Cartesian) grid locations. Each displacement vector contains coordinate offsets in each direction, nominally on a pixel-by-pixel basis. However, to constrain the motion to be smooth, and limit the number of unknowns, we use b-splines on a lower-resolution grid to describe the coordinate offsets with a reduced set of coefficients. So,

$$[w_g]_x(x,y) = \sum_{i,j} c_{g,x}[i,j]\beta\left(\frac{x}{m_x}-i\right)\beta\left(\frac{y}{m_y}-j\right), \tag{1}$$

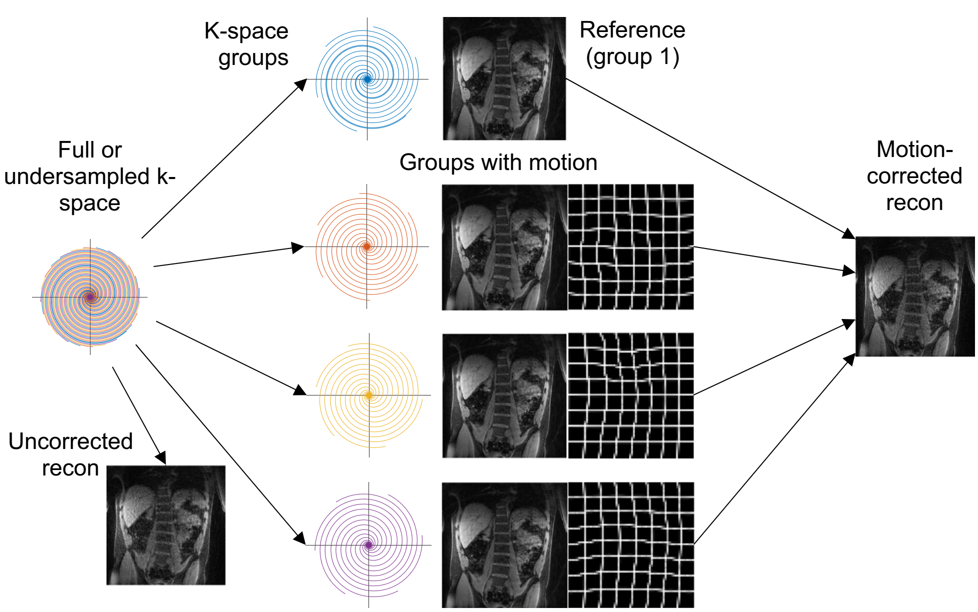

Fig. 1. In this work, full or undersampled k-space corresponding to a single image (or frame) is simulated with non-rigid motion in each of four subgroups (different color spiral readouts). Without motion correction, an uncorrected reconstruction produces the blurred image depicted in the bottom left. The motion-corrected reconstruction accounts for different non-rigid motion in each group and produces an image consistent with the reference group (nominally group 1) from all acquired k-space. The deformation maps and deformed images are shown for groups 2–4 to depict the extent of non-rigid motion in this example; this prior knowledge is not used in the reconstruction. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)