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Variable Field of View for Spatial Resolution Improvement in Continuously Moving Table Magnetic Resonance Imaging

Houchun H. Hu, Ananth J. Madhuranthakam, David G. Kruger, James F. Glockner, and Stephen J. Riederer*

Magnetic Resonance Imaging Laboratory, Department of Radiology, Mayo Clinic College of Medicine, Rochester, Minnesota, USA

Abstract

An approach is described in which the field of view (FOV) along the Y (right/left) phase encoding direction can be dynamically altered during a continuously moving table (CMT) coronal acquisition for extended FOV MRI. We hypothesize that with this method, regions of the anatomy exhibiting significantly different lateral widths can be imaged with a matching local FOV _{Y} , thereby improving local lateral spatial resolution. k -space raw data from the variable-FOV CMT acquisition do not allow simple Fourier reconstruction due to the presence of a mixture of phase encodes sampled at different Δk_Y intervals. In this work, we employ spline interpolation to reregister the mixed data set onto a uniformly sampled k -space grid. Using this interpolation scheme, we present phantom and peripheral contrast-enhanced MR angiography results demonstrating an approximate 45% improvement in local lateral spatial resolution for continuously moving table acquisitions.

Keywords

continuously moving table; spatial resolution; variable field of view; peripheral contrast-enhanced MRA; k -space interpolation

Interest in whole-body MRI has been recently growing (1–4). In particular, acquisition using continuously moving table (CMT) has been studied for imaging of the peripheral vasculature in 3D contrast-enhanced MR angiography (CE-MRA) (2,5,6). The motivation to use CMT for “runoff” CE-MRA studies naturally arises when one considers that the peripheral vasculature from the renal artery origins to the ankles spans a distance greater than 100 cm and extends beyond the homogeneous volume of all commercially available MRI scanners. For peripheral CE-MRA, a CMT technique is an alternative to a multistation approach in which a data set is acquired at each of several overlapping stationary table positions after contrast injection (7–14).

Lateral (right/left) and slice (anterior/posterior) spatial resolution in coronal angiograms become critical as vessel diameters in the peripheral vasculature approach several millimeters or less. In CMT, spatial resolution is intrinsically limited by the actively sampled field of view (FOV), the repetition time TR, and the table velocity used (2). Investigators have recently studied this limitation with parallel imaging using stationary (15) and continuously moving (16) coils as well as matching the table velocity to the known variable velocity of the contrast bolus in the peripheral vasculature (17), where slower contrast velocities in the distal vessels allow reduced table speeds and more sampled views, thus improving spatial resolution (18).

*Correspondence to: Stephen J. Riederer, Magnetic Resonance Laboratory, Mayo Clinic College of Medicine, 200 1st Street S.W., Rochester, MN 55905, USA. E-mail: riederer@mayo.edu.

Multistation approaches offer the flexibility of tailoring FOV, slice thickness, and the imaging matrix size to the vasculature of interest at individual stations (12–14). Spatial resolution can be optimized on a station-by-station basis by selecting the appropriate values. On the other hand, CMT FOV values have thus far been held constant over the entire longitudinal FOV, with the lateral and slice FOV typically determined by the widest portion of the peripheral vasculature over the extended anatomy. This approach can be inefficient, as unnecessary scan time is spent in regions where the object extent in either the right/left or the anterior/posterior direction is considerably smaller than the prescribed volume.

This work tests the hypothesis that, compared to a fixed FOV_Y , CMT techniques can benefit from improved lateral spatial resolution by tailoring the FOV_Y to the width of the anatomy along the extended longitudinal FOV. Specifically, we allow for the switching from one lateral FOV to another by introducing a dual-FOV CMT method. The FOV_Y transition occurs dynamically during table motion at an operator-specified location along the longitudinal axis. Improvement in lateral spatial resolution is achieved by maintaining the sampling matrix size while reducing FOV_Y at appropriate locations. Raw data from a dual-FOV CMT technique do not allow for simple Fourier reconstruction, because of the mixing of Δk_Y intervals from two different FOV_Y values. We describe a reconstruction algorithm for such mixed data and examine the hypothesis using hybrid k -space analysis, phantom experiments, and volunteer studies.

Materials and Methods

Hybrid k -Space Description

CMT data reconstruction for 2D and 3D acquisitions with table motion along the readout direction has been previously described (2). Figure 1a illustrates a simplified 2D hybrid k -space (X - k_Y) acquired using this CMT technique. Frequency (X) and phase (Y) encoding axes are along the vertical and horizontal axes, respectively. Assuming constant table velocity and a sequential view order starting at the most negative k_Y location, acquired echoes (green lines) are positioned into hybrid k -space with the diagonal pattern shown in Fig. 1a. FOV_S is the actively sampled FOV with longitudinally oriented readout. Because the table translates continuously along X , after each sampled echo is Fourier transformed (FT) along k_X it is placed at a progressively increasing X -position as shown, corresponding to the table location during the time of measurement. The k_Y location of the echo is determined by the Y -gradient applied for that measurement. After all X -transformed echoes have been appropriately placed into hybrid space and the k_Y - k_Z views for a X -position are acquired, additional FT along k_Y and k_Z yields the final image for that X -position. The table velocity is chosen such that the table travels a distance of exactly one FOV_S after the collection of a cycle of views (N_Y , N_Z). A complete cycle of 12 k_Y phase encodes is used for illustration simplicity in Fig. 1a, with slightly over 2 cycles shown. As in standard MRI, the spacing between successive views along Y is governed by $\Delta k_Y = 1/\text{FOV}_Y$.

Figure 1b shows the effects of altering FOV_Y during a CMT acquisition. The scan starts by using the same FOV_{Y1} as in Fig. 1a for one complete view cycle (green lines). The FOV_Y is then reduced to a smaller value, FOV_{Y2} , at the start of the second cycle, resulting in a larger Δk_Y between subsequent echoes and a farther extension of data along the k_Y axis. If the number of views (N_Y) per cycle is kept constant, as shown, then a greater extent of k_Y -space is sampled, thereby improving spatial resolution along Y . The X -location where the switch from FOV_{Y1} to FOV_{Y2} occurs is shown as an integer multiple of FOV_S for illustration simplicity in this example, but in practice is arbitrary. Views acquired with FOV_{Y2} are represented as blue lines. Several distinct regions of hybrid space can be identified in Fig. 1b. Areas containing data acquired solely from FOV_{Y1} and FOV_{Y2} are highlighted as Region 1 (green) and Region 3 (blue), respectively, and can undergo standard FT. However, in Region 2 (red), data along any

line of fixed X has been acquired by two different Δk_Y sampling intervals, with the specific sampling pattern different for each X -location. FT along k_Y with no accounting for FOV_Y changes in Region 2 leads to artifacts.

Variable-FOV Data Reconstruction

The general idea behind reconstruction in Region 2 is to form evenly sampled data at each X -line through a combination of retaining and interpolating the acquired views. In Fig. 1b, the X -position denoted by the solid red arrow contains three points acquired using the smaller FOV_{Y2} and nine samples obtained using the larger FOV_{Y1} along k_Y . By tracking the k_Y spatial frequency locations of all sampled points, spline interpolation is performed on the nine known FOV_{Y1} -acquired samples to generate new points on the Δk_{Y2} grid. The interpolation continues for all X -positions in Region 2, where data acquired on the Δk_{Y1} grid are replaced by interpolated values at a spacing of Δk_{Y2} . Points originally acquired on the Δk_{Y2} grid are retained. This results in Region 2 having uniform Δk_{Y2} sampling after interpolation and allows subsequent FT along k_Y to yield an artifact-free image corresponding to FOV_{Y2} . Because interpolation is done in hybrid (X - k_Y - k_Z) space only along the transverse (k_Y and/or k_Z) directions, resolution along X is not altered.

Variable-FOV CMT can create nonsampled gaps in k -space. Since the extent of data along k_Y is determined by N_Y and Δk_Y , the absolute k_Y boundaries ($\pm k_{Y\text{MAX}}$) of FOV_{Y2} in this example are greater than those of FOV_{Y1} . Consequently, a nonsampled region of k -space (zeros, box “B”) exists between $+k_{Y\text{MAX}1}$ and $+k_{Y\text{MAX}2}$, as determined by FOV_{Y1} and FOV_{Y2} . Another region of zeros is shown by the box labeled “A.” Here, for the particular X -position denoted by the solid red arrow, the k_Y spatial frequency of the third FOV_{Y2} sample is more than Δk_{Y2} apart from the first FOV_{Y1} sample. Currently, interpolation is not done in A and B regions. The size of A depends on the relative number of data points with which each FOV_Y contributes to the full N_Y -cycle data set. The 2D example in Fig. 1b overestimates the size of A for illustration purposes. In practice for 3D CMT, each acquired slice contains only a few X -lines (ca. 3–6) that include the zero A section, and they account for less than 1% of all sampled points in Region 2. At more inferior X -locations in Region 2, section A becomes gradually smaller and is eventually replaced by a zone where data samples overlap. This is illustrated for the X -location denoted by the open red arrow in Fig. 1b. For such cases of overlap, data from the Δk_{Y2} sampling interval is retained, those from the Δk_{Y1} interval are ignored, and interpolation is not necessary.

To present the final reconstruction from Regions 1–3 on one extended FOV image it is desirable to use the same pixel size and FOV throughout. This is done by zero padding the hybrid k -space data in Region 1 prior to FT, such that the corresponding pixel size matches the smaller pixels of Regions 2 and 3, and by zero padding the image periphery of Regions 2 and 3 after FT to match the width of the lateral FOV_Y of Region 1. Gradient nonlinearity correction is also performed as discussed in Ref. (19).

Phantom Experiments

A phantom experiment was performed to demonstrate the variable-FOV CMT reconstruction algorithm. A tube filled with Gadolinium-enhanced water was scanned on a GE 1.5-T Signa Lx system (GE Medical Systems, Milwaukee, WI, USA) with a 3D spoiled gradient echo using the following parameters: TR = 6 ms, TE = 1.8 ms (full echo), $\text{FOV}_S = 30$ cm, $\text{FOV}_{Y1} = 36$ cm, $\text{FOV}_{Y2} = 26$ cm, $N_X/N_Y/N_Z = 256/128/16$, FA = 30 degrees, BW = ± 62.5 kHz, and coronal 3-mm slices. The extended FOV was 135 cm. Sequential view order was used with slice (Z) and phase (Y) encodings in the outer and inner loops, respectively. The first two complete view cycles were acquired using FOV_{Y1} . At the start of the third view cycle the lateral FOV was reduced to FOV_{Y2} and was maintained for the remainder of the CMT scan.

A second phantom experiment was performed to study the effects of zero regions A and B on Y -spatial resolution in variable-FOV reconstruction. A resolution bar phantom with line pairs oriented along the right/left axis was placed in the region of FOV_Y transition where A and B are present. Acquisition parameters were similar to the first phantom experiment. The reconstructed variable-FOV data were compared to a reference CMT scan where the same resolution phantom was placed in a region of constant FOV_{Y2} with full sampling along the k_Y axis.

Peripheral Vasculature CE-MRA Volunteer

Seven volunteers underwent peripheral CE-MRA exams using the variable-FOV CMT method. Consent and approval for each study was obtained from the subject and the institutional review board. Studies were performed on a 1.5-T GE Signa scanner. Coronal images were obtained with a 3D spoiled gradient echo with sequential view order. Acquisition parameters were determined from localizers. An example (Volunteer 2) is discussed below. Scan parameters were $TR = 4.9$ ms, $TE = 1.4$ ms (partial echo), $FOV_S = 20$ cm, $FOV_{Y1} = 30$ cm, $FOV_{Y2} = 21$ cm, $\Delta z = 7$ mm, $N_X/N_Y/N_Z = 192/128/16$, $FA = 30$ degrees, and a table velocity of 1.99 cm/s. A total of 30 mL of contrast agent (Omniscan, Amersham Health, Inc., Princeton, NJ, USA) followed by equal saline was injected at 2.0 mL/s via an electronic injector (Medrad Spectris Solaris, Medrad, Inc., Indianola, PA, USA). From localizers, a 30-cm FOV_{Y1} was determined adequate for lateral coverage from the renal artery origins to the femoral arteries. A smaller 21 cm FOV_{Y2} was deemed sufficient for imaging from the popliteal trifurcation to the distal vasculature. Longitudinal coverage was approximately 110 cm, and the changeover in FOV_Y occurred after 60 cm of table travel. A mask image of the extended FOV was acquired prior to contrast injection, and for the contrast run, table motion was fluoroscopically triggered upon contrast arrival in the abdominal aorta. Reconstructed images from each study were evaluated for the presence of artifacts and for the continuity of vessels over the region of changing FOV_Y .

Results

Figure 2 shows maximum intensity projections (MIP) from the phantom experiment. Figure 2a illustrates the reconstruction of variable-FOV CMT data without interpolation. The most superior (Region 1) and inferior (Region 3) portions were acquired with single FOV_Y data of 36 and 26 cm, respectively, and are free of artifacts. However, Region 2 of the image (within the two dashed lines) exhibits dual images (arrowheads) of the single tube as a result of data mixture. Figure 2b shows a replication Regions 1 and 3 from Fig. 2a, whereas Region 2 was generated after applying the interpolation scheme described. Note that in Region 2 the tube signals from the larger FOV_{Y1} frame have been removed, and the phantom is displayed purely in the FOV_{Y2} frame. Finally, Fig. 2c shows the result after matching the pixel size and display FOV_Y for all regions.

Figure 3a–d illustrates the effects of zero regions A and B with a set of resolution bars oriented along Y . The enlarged images are part of a larger phantom used to test a variable-FOV CMT acquisition in which FOV_Y transition occurs at the location of the resolution bars shown. For the scan parameters used, three consecutive X -locations with A gap sizes of 14, 9, and 5 units of Δk_{Y2} , respectively, are present. The size of B for all three X -lines is $18 \cdot \Delta k_{Y2}$. For comparison, Fig. 3a illustrates a fully sampled image. In Fig. 3b, the resultant image reconstructed from data with gaps A and B is shown. A 1D line-profile (Fig. 3c) along Y and corresponding point-spread-functions (Fig. 3d) confirm the expected loss in signal intensity and spatial resolution in Fig. 3b as a result of A and B. However, the effects are minimal and are confined to only a small number of X -positions in transition Region 2.

Variable-FOV CMT reconstruction was performed successfully for all seven volunteers, yielding images with good vessel enhancement and continuity. A representative example is shown in Fig. 4. With $\text{FOV}_{Y1} = 30$ cm, adequate lateral coverage from the renal artery origins to the femoral arteries is achieved without undesired aliasing from secondary “runoff” vessels. The selection of $\text{FOV}_{Y2} = 21$ cm is motivated by the widest portion of the vasculature distal to the femoral arteries, which occurs at the popliteal trifurcation. Note that FOV_{Y2} closely circumscribes the anterior tibial arteries. Figure 4a shows the interpolated reconstruction, with a 43% improvement in lateral spatial resolution in the distal vessels had FOV_{Y1} been used to acquire the entire CMT scan. In forming Fig. 4b, two additional k -space data sets were acquired after the table stopped at the inferiormost position and were added to the original to provide further enhancement of the distal arteries in Region 3. In this volunteer, approximately 4100 more views (20 s) were sampled.

Discussion and Conclusion

We have demonstrated an approach to improve lateral spatial resolution in 3D CMT by tailoring the right/left FOV to the local width of the object. By maintaining the sampling matrix size while changing FOV_Y , improvement in lateral spatial resolution is achieved in regions imaged with a smaller FOV_Y . While the method has been demonstrated in peripheral CE-MRA, it is immediately adaptable to other CMT applications, such as whole-body screening.

The implication of variable-FOV CMT on k -space involves a region where mixed data from different Δk_Y sampling intervals are present. In addition, potential gaps in k -space may occur at a small number of X -locations in this region. The reconstruction algorithm described in this work employs 1D interpolation in hybrid k -space to restore uniform sampling. Interpolation over zero regions is not done because they make up less than 1% of all sampled data points and are confined to high spatial frequencies. Homodyne detection (20) is a possible alternative.

With application in peripheral CE-MRA, the transition from a large to a small FOV_Y is more likely than the reverse because the lateral width of the renal/femoral arteries is typically greater than that at the popliteal trifurcation for a majority of the patient population. The algorithm, however, can accommodate small to large FOV_Y transitions. The method has worked successfully for a FOV_Y change as large as 16 cm. In peripheral CE-MRA using CMT, the appropriate selection for phase and slice resolution is determined on a patient-specific basis. The interpolation algorithm has the flexibility to account for multiple matrix sampling sizes and any length of table travel prior to FOV_Y transition.

The proposed technique can be made more flexible in several ways. First, it has only been implemented with the body receiver in this work. Extension to surface coils is straightforward, as each coil is treated separately and reconstructions are later combined in a sum-of-squares manner. Use of multichannel phased-array assemblies can further facilitate the integration of variable-FOV CMT with parallel imaging. Second, the current reconstruction algorithm has been studied using sequential view order. Extension to nonsequential view orders will require additional modifications to the interpolation algorithm, since accurate knowledge of hybrid k -space locations where FOV_{Y1} - and FOV_{Y2} -acquired samples are allocated is critical for an artifact-free reconstruction. Last, a further improvement of the technique is to implement variable-FOV along the slice-select direction, apply 2D interpolation, and with flexible slab thickness allow improved slice resolution.

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FIG. 1. Illustration of X - k_Y hybrid space for a 2D CMT acquisition using (a) a constant FOV_Y and (b) two FOV_Y , where $\text{FOV}_{Y1} > \text{FOV}_{Y2}$. FOV_S is the actively sampled FOV along the extended readout X -axis. In (b), the switch to FOV_{Y2} occurs after one complete view cycle. Regions 1 and 3 identify X -locations in which the lateral FOV is fixed to purely FOV_{Y1} and FOV_{Y2} , respectively, whereas Region 2 contains a mixture of views from both. For illustration clarity, 12 echoes are shown for each complete view cycle. Interpolation is performed in Region 2 to create a uniformly sampled k -space, using samples acquired on the Δk_{Y1} grid to obtain new points on an interval of Δk_{Y2} . Implementation of variable-FOV CMT creates small data gaps in Region 2 that are not interpolated (gray boxes A and B).

FIG. 2.

MIP images of gadolinium-enhanced water phantom acquired with variable-FOV CMT with $FOV_{Y1} = 36$ cm and $FOV_{Y2} = 20$ cm. **(a)** Reconstruction without interpolation. The extent of Region 2 is shown by the dashed lines, containing dual images (arrowheads) of the single tube. **(b)** Artifact-free reconstruction after hybrid k -space interpolation of mixed data in Region 2. **(c)** Subsequent zero padding to match pixel sizes and FOV_Y widths for proper image display. The image is displayed at a window/level to accentuate the zero padding in Regions 2 and 3.

FIG. 3.

Effects of zero regions A and B on resolution are simulated with Y -oriented bars for three consecutive X -lines in Region 2 with A gap sizes of 14, 9, and 5 units of Δk_{Y2} and a B size of $18 \cdot \Delta k_{Y2}$. **(a)** Reference resolution with full-sampling for comparison. **(b)** Image reconstructed from data with k_Y gaps, showing minor loss in resolution and signal intensity for the affected X -locations. **(c)** Normalized 1D line-profiles along Y for the bar set highlighted in (a), and **(d)** corresponding point-spread functions. Legends for (c) and (d) are the same.

FIG. 4.

(a) Reconstructed peripheral CE-MRA MIP using variable-FOV CMT, with $\text{FOV}_{Y1} = 30$ cm at the renal artery origins and subsequent switch to $\text{FOV}_{Y2} = 20$ cm (dashed line) at the femoral arteries. Note that the choice of $\text{FOV}_{Y2} = 20$ cm to image the popliteal trifurcation allows the FOV_Y to tailor more precisely to the lateral width of the vessels. **(b)** An extension of (a) in which data collection continued for approximately 20 more seconds after table motion stopped at the most distal position, permitting additional vascular enhancement in Region 3.