Renal Perfusion Imaging with Two-Dimensional Navigator Gated Arterial Spin Labeling

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Purpose: To develop a navigator technique enabling free-breathing acquisition to afford sufficient signal averaging for quantitative renal perfusion measurement using arterial spin labeling MRI.

Methods: A novel two-dimensional (2D) navigator technique was implemented in concert with flow-sensitive alternating inversion recovery (FAIR) preparation and true fast imaging with steady precession (True-FISP) readout. The navigator images were obtained with a low-resolution fast low angle shot readout at end of each arterial spin labeling acquisition. A retrospective algorithm was developed to automatically detect respiratory motion for selective signal averaging. The 2D navigator-gated FAIR True-FISP sequence was performed in ten healthy volunteers and five patients with chronic kidney disease.

Results: Excellent image quality and comparable cortical perfusion rates (healthy: 276 ± 28 mL/100 g/min, patients: 155 ± 25 mL/100 g/min) to literature values were obtained. An average of 3-fold signal-to-noise ratio improvement was obtained in the 2D navigator-gated approach compared with the breath-hold acquisition in healthy volunteers. Good image quality was achieved in patients while the results from breath-hold acquisition were unusable. The quantitative perfusion rates were significantly lower in chronic kidney disease patients compared with the healthy volunteers.

Conclusion: 2D navigator-gated free breathing arterial spin labeling is feasible and is a noninvasive method to evaluate renal perfusion both in healthy subjects and those with chronic kidney disease. Magn Reson Med 000:000–000, 2013. © 2013 Wiley Periodicals, Inc.

Key words: 2D navigator; renal perfusion; FAIR True-FISP; respiratory motion

Quantitative change in tissue perfusion can provide valuable information regarding organ function in different pathological conditions. The kidney is a highly perfused organ, and measurement of renal perfusion can directly indicate functional impairments including inflammatory and degenerative renal disease, renal artery stenosis, renal transplant nephropathy, chronic ischemic nephropathy, and drug-induced nephropathy (1). The use of contrast agents has been pursued to evaluate first pass perfusion MRI and glomerular filtration rate measurements (1). However, given the risk of nephrogenic systemic fibrosis (2,3), the use of gadolinium-based contrast agents is contraindicated in patients with compromised renal function.

Arterial spin labeling (ASL) is a noninvasive technique that uses water in the blood as an endogenous tracer to measure perfusion (4). The principle of ASL involves that uses water in the blood as an endogenous tracer to measure perfusion (4). The principle of ASL involves inflowing spins to be corrected via postprocessing, it is difficult to retrospectively correct for through-plane motion. Several acquisition strategies have been proposed to solve this problem, including single/multiple breath-holds, synchronized breathing, and respiratory triggering (9,10). However, those strategies are still prone to residual motion and impractical for routine clinical use. In particular, breath-holding can be difficult to tolerate for patients.

As a subtraction-based technique, ASL is inherently prone to motion induced errors and artifacts. A mismatch between control and label image can result in artifacts and inaccurate perfusion estimation. The major challenge in performing ASL in the kidneys is the abdominal motion caused by respiration. While in-plane motion may be corrected via postprocessing, it is difficult to retrospectively correct for through-plane motion. Several acquisition strategies have been proposed to solve this problem, including single/multiple breath-holds, synchronized breathing, and respiratory triggering (9,10). However, those strategies are still prone to residual motion and impractical for routine clinical use. In particular, breath-holding can be difficult to tolerate for patients.

In this study, we implemented a novel 2D navigator with flow-sensitive alternating inversion recovery (FAIR) true fast imaging with steady precession (True-FISP) sequence (8,11) to measure renal perfusion in a free-breathing acquisition. An automated postprocessing algorithm was developed to estimate the diaphragm position on the 2D navigator images as a reference to select the final control and label images for perfusion.
calculation. The purpose of this study was to minimize respiration-induced kidney motion via selective signal averaging according to the navigator data. The preliminary results were obtained in healthy volunteers and a small number of patients with chronic kidney diseases (CKDs). The SNR and perfusion quantification results from free-breathing acquisitions were compared with the breath-hold acquisitions.

METHODS

Pulse Sequence Design

The renal perfusion measurement was performed using the FAIR True-FISP sequence (8). The pulse sequence diagram is illustrated in Figure 1. Details of each pulse sequence component are described as follows.

2D FLASH Navigator

The 2D navigator was used immediately after the True-FISP acquisition. The aim was to acquire a low-resolution 2D image to estimate respiratory motion based on the diaphragm position throughout the respiration cycle. The final perfusion-weighted image was computed by only using control and label images considered to be at the same diaphragmatic position. We assumed there was no significant motion during the True-FISP and the navigator readout. The 2D navigator comprised a 2D fast low angle shot sequence with the following parameters: flip angle of 5°, pulse repetition time/echo time = 2.2/1.2 ms, field of view = 400 mm, image matrix size = 96 × 96, receiver bandwidth = 1000 Hz/Pixel, 5/8 partial Fourier encoding along the phase encoding direction, parallel acceleration = 2 with 10 integrated reference lines for calibrating coil sensitivities. The navigator was prescribed graphically to cover the right hemidiaphragm in either sagittal or coronal plane. The total duration for the 2D navigator was 75 ms.

FAIR True-FISP Sequence

In FAIR, perfusion weighting was achieved by a pair of slice selective (control) and global inversion (label) pulses used in an interleaved manner (11). A 10.24 ms adiabatic frequency offset corrected inversion (FOCI) pulse \( \mu = 6, \beta = 1078 \) was used for inversions (12). The slice selective inversion band was positioned carefully to avoid intersection with major arteries. The inversion slab thickness was set to 30 mm to accommodate expected superior-inferior motion (8,13). True-FISP imaging was performed after the inversion pulse following a brief delay (TI) to allow tagged blood to perfuse into the tissue. A TI of 1.5 s was used for healthy volunteers. A TI of 2 s was used for patients in consideration of potentially comprised flow. The imaging slice was positioned in an oblique coronal orientation to match the longitudinal axis of both kidneys. The following imaging parameters were used: flip angle = 60°, echo spacing/echo time = 4/2.02 ms, pulse repetition time = 3 s, receiver bandwidth = 651 Hz/Pixel, image matrix = 128 × 128, field of view = 360–400 mm, imaging slice thickness = 8 mm. To minimize artifacts from the oscillatory transient effects in the True-FISP acquisition, a ramp of 10 radiofrequency pulses with linearly increasing flip angles was used at the beginning of the acquisition without signal recording (14). Centric ordered phase encoding was used to maximize perfusion sensitivity.

In Vivo Experiments

All experiments were performed on a 3 T MRI system (MAGNETOM Verio, Siemens Healthcare, Erlangen, Germany) equipped with high performance gradient coils (45 mT/m maximum gradient strength, 200 mT/m/ms slew rate). The body coil was used as the transmitter, and the combination of spine and body array coils was used as the receiver. Subjects were positioned feet first and all imaging was performed at the magnet isocenter. Ten healthy volunteers (5 male, 5 female, 42 ± 16 years of age) and five patients with CKD (2 male and 3 female, 64 ± 12 years of age) were recruited with the approval from our Institutional Review Board. Written informed consent was obtained from each participant before the study.

Healthy Volunteer Experiment 1

The purpose of this experiment was to test the feasibility of the 2D navigator technique. Eight subjects participated in this study. The volunteers were instructed to relax and breathe normally during the scans. FAIR True-FISP sequence was repeated with the 2D navigator positioned in the sagittal and coronal plane, respectively. Fifty control/label pairs were acquired with a total scan time of 5 min per free-breathing scan. An 18 s breath-hold scan was performed at expiration with 3 control/label pairs. A proton density weighted (PD) image was acquired using an identical True-FISP readout with a pulse repetition time of 10 s and no inversion pulses. In addition, a high-resolution renal perfusion image (matrix size = 256 × 256, slice thickness = 5 mm) was acquired in healthy volunteers. An iPAT acceleration factor of 2 was used in these high-resolution scans to reduce readout duration.

Healthy Volunteer Experiment 2

The purpose of this experiment was to test the robustness of the navigator technique in the presence of different inversion band thicknesses, imaging planes, and respiratory patterns. Three subjects participated in this study. A total of 12 perfusion scans were included in this experiment: three labeling inversion band thicknesses of 10, 20, and 30 mm were used; the imaging plane was prescribed in both transverse and oblique coronal plane; each scan was performed twice where the
subject was instructed to breathe normally on one and deeply on the other. The remaining imaging parameters were identical to experiment 1. Due to time constraints, the 2D navigator was prescribed only in the coronal plane.

**Patient Experiment**

FAIR True-FISP with the 2D navigator was tested in five patients with CKD stage 2–4. The imaging protocol was identical to that used in healthy volunteer experiment 1, except for using a TI = 2 s. Two patients were scanned for 5 min and three patients were scanned for 10 min. A breath-hold scan was performed for comparison.

**Data Processing**

Raw perfusion and navigator data were transferred to a local workstation (Intel i7 2.7 GHz Quad Core, 8 GB of RAM) for the image reconstruction. Control and label images were reconstructed with the 2D inverse Fourier transform followed by the square root of the sum of the squares to combine the data from different coils. The navigator images and the high-resolution FAIR True-FISP images were reconstructed using GRAPPA (Generalized Autocalibrating Partially Parallel Acquisition) (15) when parallel acceleration was used. The first control/label pair in the temporal series for all perfusion scans was considered to be acquired before a stable longitudinal steady state was achieved and was discarded from the subsequent calculation. All data reconstruction and image analysis were performed using MATLAB (MathWorks, Natick, MA).

Automated Retrospective Signal Averaging with the 2D Navigator

The kidneys were assumed to be in the same position on the 2D navigator image and its corresponding control/label image. To minimize through-plane motion, an automated strategy for retrospective signal averaging was developed in the following steps:

1. Sub-navigator identification: The longitudinal kidney motions can be approximated by the diaphragmatic movement at the border of lung and liver. A rectangular subregion in the navigator images (referred to as the sub-navigator) along right hemidiaphragm would be an ideal location to detect the superior-inferior motion. To locate the sub-navigator automatically, the navigator images from the entire temporal series were first averaged into a single image. A general area where the right hemidiaphragm resides was estimated and extracted from the averaged navigator image. Given the center of the imaging plane was often positioned medially on the spine with one kidney on each side, the area containing the right hemidiaphragm was estimated as the top left quadrant in the coronal navigator image or the top middle half in the sagittal navigator image. The sub-navigator position was then determined empirically. The region with the peak correlation value was identified as the sub-navigator. An illustration of the 2D navigator images (Fig. 2a) and the sub-navigator identification process is shown in Figure 2.

2. Diaphragmatic motion detection: A 2 × 21 (row × column) matrix was extracted from each navigator image using the sub-navigator as a reference. The matrix was then averaged along the row to obtain a single vector, containing the diaphragm position for each image acquisition. The concatenation of vectors from all navigator images exhibited a full respiration profile during the entire scan, shown in Figure 2c. The diaphragm position in each vector (red line in Fig. 2c) was subsequently determined using an edge detection algorithm (17).

3. Control/label images selection: Final images were chosen during expiration. An acceptance window width of 8 mm (2 pixel width in the navigator profile) was used. Control and label images of which diaphragm position was contained within the acceptance window were selected for the perfusion calculation. Note the final number of selected control and label images may differ from each other.

Images selected for the final perfusion calculation were realigned using the FMRIB’s Linear Image Registration Tool (FLIRT, FMRIB, Oxford, United Kingdom) (18). The first image in the selected image series was used as the reference. The corrected images were then averaged to obtain a single control and label image. The perfusion weighted image was computed by subtracting the averaged control from the averaged label image.

**Perfusion Quantification**

Neglecting the transit effects and the exchange times between blood water molecule and the kidney tissue, the quantitative renal blood flow can be calculated pixel-by-pixel using a single compartment model (8)

\[
f = \frac{\chi}{2aT1} \frac{\Delta M(T1)}{M0} \exp \left( \frac{T1}{T1} \right)\]

where \( f \) is the perfusion rate (in the unit of mL/100 g/min), \( \chi \) is the blood–tissue–water partition coefficient, which is assumed to be 80 mL/100 g (19), \( z \) is the inversion efficiency which is assumed to be 0.95, \( \Delta M(T1) \) is the perfusion weighted image, and \( M0 \) is the equilibrium magnetization of the tissue (proton density). The \( T1 \) value of 1.15 s for renal cortex (20) is assumed to be the \( T1 \) of the blood. Pixels with exceptionally high perfusion values (\( f > 600 \text{ mL/100 g/min} \)) corresponding to macroscopic blood vessels (8) were all set to 600. In addition, perfusion values between 400 \( \text{ mL/100 g/min} \) and 600...
mL/100 g/min were set to 400 mL/100 g/min according to the maximal possible value (8).

Image Analysis

Mean perfusion values were measured in renal cortex and medulla in both kidneys. Regions near renal arteries with extremely high perfusion values (f > 600 mL/100 g/min) that correspond to macroscopic flow were carefully avoided in the region of interest (ROI). An illustration of the ROI outlines is shown in Figure 2d. SNR was calculated as $\text{SNR} = \frac{S_{\text{control}} - S_{\text{label}}}{\sigma_{\text{noise}}}$, where $S_{\text{control}}$ and $S_{\text{label}}$ were the mean signals in the renal cortex of the averaged control and label images, and $\sigma_{\text{noise}}$ was the standard deviation of the noise in the subtraction image. The acquisition efficiency for the free-breathing scans was defined as $(N_{\text{control}} + N_{\text{label}})/N_{\text{total}}$, where $N_{\text{control}}$ and $N_{\text{label}}$ were the number of navigator-selected control and label images, and $N_{\text{total}}$ is the total number of images acquired. The acquisition efficiency for breath-hold scans was 0.67 because the first control/label pair was discarded in the perfusion averaging. The relative SNR (rSNR), defined as the product of the SNR and the acquisition efficiency, was also calculated.

To estimate the optimal scan time, the imaging data was resorted into sets of 10, 20, 30, 40, and 50 control/label pairs in the original acquisition order, corresponding to 1, 2, 3, 4, and 5 min of acquisition time. Each set then underwent the same retrospective signal averaging process as previously described. The acquisition efficiency for each individual set was calculated. Pairwise rSNR difference among free-breathing and breath-hold scans was tested using a paired two-tail $t$-test. A nonparametric Friedman’s test was used for comparing set acquisition efficiency. A $P$ value less than 0.05 was considered to indicate significance.

RESULTS

Healthy Volunteer Experiment 1

The FAIR True-FISP sequence with the 2D navigator produced renal perfusion images of excellent quality in volunteers. An example of perfusion maps is illustrated in Figure 3. The renal cortex and medulla can be clearly depicted. Subtraction errors and reduced SNR were apparent in the cortex in the breath-hold images (white arrow, Fig. 3). Despite the longer readout duration, excellent image quality with improved delineation of cortex and medulla was obtained with the high resolution protocol (Fig. 4). The realignment correction values are plotted in Figure 5a. The largest kidney motion after the retrospective selection was in the superior-inferior direction with a maximum displacement of 4 mm among all subjects. The average displacement was 1.1 mm along superior-inferior direction, and less than 1 mm along the medial-lateral direction. In-plane rotation variation was less than 1 degree.

One volunteer had a solitary kidney and the result was excluded from the group analysis. The mean perfusion values measured in the healthy volunteers are reported in Table 1. The quantitative perfusion rate in the renal cortex ranged from 196 to 308 mL/100 g/min with the
coronal navigator, 219 to 328 mL/100 g/min with the sagittal navigator, and 182 to 368 mL/100 g/min for the breath-hold scans. These values were in good agreement with other renal ASL studies in the literature (7,8,19,21). The perfusion rates measured in the solitary kidney was higher than the average (cortex: 364 mL/100 g/min, medulla: 61 mL/100 g/min), which was consistent with the known increase in size and function to compensate for the missing kidney (22,23).

The SNR, acquisition efficiency, and rSNR for each healthy volunteer are reported in Table 2. The mean SNR was 36.65 for the coronal navigator, 31.36 for the sagittal navigator, and 12.15 for breath-hold scans. Overall, an average acquisition efficiency of 50 ± 13% and...
41 ± 9% was observed for the coronal and sagittal 2D navigators, respectively. Acquisitions with the coronal navigator yielded slightly higher rSNR; however, there was no significant difference in rSNR between the coronal and sagittal navigator-gated acquisitions ($P > 0.05$). The rSNR of the free-breathing acquisitions using 2D navigator was significantly higher than the breath-hold acquisitions ($P < 0.05$).

An example of the retrospective reconstruction result for the multiple time sets is illustrated in Figure 6a. Figure 6b shows the mean SNR and acquisition efficiency across all healthy subjects. No significant difference was found ($P > 0.05$). The image quality gradually improved as the number of averages increased. The qualitative assessments of the image quality suggested the scan time might be reduced to 3 min and still maintain acceptable SNR (an average of 18) as the 5 min acquisition.

### Table 1
Mean Perfusion Rates (mL/100 g/min) for Healthy Volunteers in Experiment 1

<table>
<thead>
<tr>
<th>Acquisition type</th>
<th>Cortex</th>
<th>Medulla</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left kidney</td>
<td>Right kidney</td>
</tr>
<tr>
<td>Coronal 2D navigator</td>
<td>274 ± 28</td>
<td>264 ± 25</td>
</tr>
<tr>
<td>Sagittal 2D navigator</td>
<td>278 ± 29</td>
<td>265 ± 34</td>
</tr>
<tr>
<td>Breath-hold</td>
<td>287 ± 38</td>
<td>281 ± 55</td>
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</table>

### Table 2
SNR, Acquisition Efficiency, and rSNR Measurements of Healthy Volunteers

<table>
<thead>
<tr>
<th>Subj</th>
<th>SNR</th>
<th>Acquisition efficiency ($N_{control}$, $N_{label}$)</th>
<th>rSNR</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>29.53</td>
<td>65% (31, 34)</td>
<td>43% (22, 21)</td>
</tr>
<tr>
<td>2</td>
<td>38.91</td>
<td>61% (29, 32)</td>
<td>33% (16, 17)</td>
</tr>
<tr>
<td>3</td>
<td>23.54</td>
<td>37% (18, 19)</td>
<td>34% (15, 19)</td>
</tr>
<tr>
<td>4</td>
<td>28.67</td>
<td>35% (18, 17)</td>
<td>50% (24, 26)</td>
</tr>
<tr>
<td>5</td>
<td>46.02</td>
<td>56% (32, 24)</td>
<td>24.85</td>
</tr>
<tr>
<td>6</td>
<td>47.28</td>
<td>58% (30, 28)</td>
<td>34% (16, 18)</td>
</tr>
<tr>
<td>7</td>
<td>23.39</td>
<td>38% (19, 19)</td>
<td>38% (17, 21)</td>
</tr>
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</table>

Healthy Volunteer Experiment 2

An example of varying imaging parameters including inversion slab thickness, imaging plane orientation, and respiratory pattern is illustrated in Figure 7. When the imaging orientation was oblique coronal, subtraction errors were highly apparent with the 10 mm inversion band in both normal and deep breathing conditions. Imaging artifact was apparent with the 20 mm inversion band during deep breathing. No apparent artifact was observed with the 30 mm inversion band for both breathing patterns in the oblique coronal orientation.

The imaging result was much more prone to error in the axial imaging orientation. A reasonable perfusion renal image, judged by the cortico-medullary contrast, was only obtained with the 30 mm inversion band thickness and normal breathing in the axial imaging orientation.
The motion correction values are shown in Figure 5b. In general, the motion correction parameters had larger standard deviations for deep breathing in comparison with normal breathing.

Patient Experiment
The mean renal blood flow values in our patient study \( (n = 5) \) were \( 158 \pm 27 \) and \( 151 \pm 23 \) mL/100 g/min for the left and right renal cortex as well as \( 34 \pm 6 \) and \( 35 \pm 6 \) mL/100 g/min for the left and right renal medulla (Table 3), respectively. An average of 35% acquisition efficiency was achieved in patients. No significant difference in acquisition efficiency between patients and healthy volunteers in experiment 1 was observed \( (P > 0.05) \). When compared with the healthy volunteers in experiment 1, both cortical and medullary perfusion rates were significantly lower in patients \( (P < 0.01) \). The perfusion map of a 58-year-old male patient with stage 2 CKD is shown in Figure 8.

**DISCUSSION**
Like other abdominal MRI techniques, respiratory motion presents a major challenge in renal ASL studies. A variety of acquisition strategies have been proposed to overcome this problem. Breath-holding is commonly used to reduce respiratory motion; however, it still suffers from some degrees of residual motion \( (13) \) and often does not accommodate for sufficient signal averaging required by ASL. In addition, breath-holding can be difficult to tolerate for patients. Synchronized breathing or multiple breath-hold acquisitions \( (10) \) may offer sufficient signal averaging; however, it is impossible to

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**Table 3**
Perfusion Rates (mL/100 g/min) for CKD Patients

<table>
<thead>
<tr>
<th>Patient</th>
<th>Cortex</th>
<th>Medulla</th>
<th>Acquisition Efficiency (%)</th>
<th>Sex/age</th>
<th>CKD stage</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
<td>Right</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>160</td>
<td>158</td>
<td>36</td>
<td>31</td>
<td>38</td>
</tr>
<tr>
<td>2</td>
<td>149</td>
<td>161</td>
<td>40</td>
<td>45</td>
<td>39</td>
</tr>
<tr>
<td>3</td>
<td>170</td>
<td>177</td>
<td>34</td>
<td>36</td>
<td>32</td>
</tr>
<tr>
<td>4</td>
<td>119</td>
<td>116</td>
<td>36</td>
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<tr>
<td>5</td>
<td>191</td>
<td>147</td>
<td>24</td>
<td>29</td>
<td>39</td>
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</table>
guarantee the subjects will hold their breath at exact one position in the respiratory cycle. Those acquisition schemes usually require additional patient training and can be time consuming and impractical for routine clinical use. Respiratory triggering (13) can yield significantly different results based on the quality of the bellow waveform, which can vary from subject to subject depending on the position of the respiratory belt and the breathing pattern. Alternatively, prospective navigator echo gating technique such as the dual navigator echo (24) can be used to monitor kidney motion.

In our study, we demonstrated the feasibility of a novel 2D navigator to measure kidney perfusion with automated retrospective postprocessing. The technique has effectively minimized the subtraction errors related to control and label image misalignment, and provided excellent image quality. When compared with the breath-hold technique, the 2D navigator-gated technique offered a near 3-fold increase in SNR in healthy volunteers and drastically improved image quality in patients (it was difficult to assess the actual SNR improvement in patients because the actual perfusion signal in the cortical region in the breath-hold acquisitions was too low to be accurately measured using ROI analysis). The measured global renal perfusion was significantly lower in CKD patients than healthy volunteers. This finding is consistent with the recent literature (25), which suggested the importance of renal perfusion assessment as an indicator of compromised renal function. Our preliminary results have confirmed that ASL True-FISP can quantify kidney perfusion in the CKD patients, given sufficient signal averaging. The 2D navigator provided a feasible method to accomplish this, whereas breath-hold did not.

The acquisition efficiency of the 2D navigator ranged from 35 to 65% depending on the respiratory pattern. Although the mean acquisition efficiency was higher in healthy volunteers than patients, the difference was not significant ($P > 0.05$). A larger sample size is required to understand whether a direct correlation between acquisition efficiency and the health state of the subject exists. In our experiments, the 5-min imaging time was sufficient for all healthy volunteers, even with the high resolution protocol (which the nominal voxel size was reduced by 85%). Furthermore, retrospective reconstruction with variable acquisition times suggested the scan time may be further reduced to 3 min without affecting the image quality.

Because the 2D navigator was implemented as an independent module, it could be combined with other ASL preparation (e.g., signal targeting with alternative radio frequency (STAR) (26), pseudo-continuous arterial spin labeling (PCASL) (27) and acquisition (e.g., True-FISP (8), echo-planar imaging (13), single-shot fast spin echo (SSFSE) (10) schemes. We chose the FAIR True-FISP implementation for our ASL imaging due to the high inversion efficiency of the FAIR preparation and SNR advantage of the True-FISP readout (8). The central frequency was offset carefully from 0 to 50 Hz to avoid True FISP banding artifacts affecting the kidney. At 3 T, the True-FISP catalyzation pulses with linearly increasing flip angles provided better suppression of transient signal oscillations compared with the $\pi/2$ catalyzation pulse (data not shown).

The width of the acceptance window can impact the image quality and final acquisition efficiency. Too narrow of an acquisition window may result in insufficient SNR, and too wide of an acquisition window may lead to the inclusion of incorrect imaging slices rendering image artifacts. For illustration purpose, we reprocessed a deep-breathing case from experiment 2 with different window widths. 

![FIG. 8. The renal perfusion map from one of the patients (CKD stage 2, male, 58 years of age). Reduced global perfusion in the cortex was observed compared with a healthy volunteer (subject 6 in Table 2). The SNR and image quality of the breath-hold acquisition for the patient was too poor to be useful. The navigator-gated sequence allowed sufficient signal averaging to provide a reasonable perfusion map in the patient. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

![FIG. 9. Perfusion weighted images reconstructed with different acceptance window widths.](image-url)
acceptance window widths, as shown in Figure 9. The degradation of the image quality was clear as the acceptance window width increased. Subjects with smaller fluctuations of the diaphragmatic position during normal breathing may tolerate a wider acceptance window. In addition, imaging position can also affect the decision on the acceptance window width. Imaging in the axial or oblique coronal position, especially with a larger tilting angle would be more susceptible to subtraction errors with a wider acceptance window than imaging in the true coronal position. In our studies, because the imaging plane was in the oblique coronal position to best match the longitudinal axis of both kidneys for maximum coverage, we found an acceptance window of 8 mm was optimal for both volunteers and patients. Because the decision on the acceptance window width is a part of the postprocessing, one may trade between SNR and image quality to determine the optimal acceptance window width for each individual case.

The selective averaging algorithm for retrospective reconstruction was effective, easy to implement, and computationally inexpensive. The sub-navigator position can be manually adjusted as needed. In our study, the algorithm to auto-identify the sub-navigator position provided satisfactory results for all case except one patient. In that particular case, the patient had a large liver such that the motion detected by the navigator did not correlate to the motion of the kidney. The sub-navigator was manually adjusted to gate off the spleen, and the perfusion map quality was greatly improved.

Our preliminary results showed the 2D navigator approach in its current implementation did not enable perfusion assessment in the axial imaging orientation when the subject exhibited a deep breathing pattern. Large motion along the superior-inferior direction can cause mismatch of the imaging slice leading to artifactual perfusion signal. Caution should be taken when imaging in the axial plane, especially when large fluctuations in the navigator profiles are observed. The 2D navigator approach worked well with the oblique coronal imaging position, even with a lower than recommended labeling inversion thickness. During the normal breathing exercise, we have observed the perfusion signal in the oblique coronal plane with the 20 mm inversion thickness was slightly higher than the 30 mm inversion thickness across all three subjects. While we do not know the exact cause, it may be the result of a shorter inflow time due to the thinner inversion band. This observation will be investigated in future studies. Currently, we recommend measuring renal perfusion with the 2D navigator technique applied in the coronal or oblique coronal orientation and using an inversion band at least 3.5 times thicker than the acquired slice thickness.

A few limitations of the study should be noted. Only retrospective navigator gating was used in this preliminary study due to the freedom of postprocessing suitable for preliminary testing. Future implementations could include prospective gating which may aid more widespread utility. We are currently limited to single-slice imaging with the 2D navigator technique. Multislice imaging may be possible with parallel imaging of high acceleration factors and partial Fourier encoding without compromising scan time albeit at a cost of SNR and/or spatial resolution. Spatial variation in the transit delays of the labeled arterial spins, known as the transit time effect, was neglected in this preliminary feasibility study. While the use FOCI inversion pulses partially compensates for this effect (8), the final perfusion rates could be underestimated as a consequence. Alternatively, Q2TIPS (28) can be implemented to improve the accuracy of the perfusion quantification by employing additional saturation pulses. In our study, we only prescribed the 2D navigator in the coronary and sagittal plane because the kidney motion is predominately superior to inferior. Therefore, only in-plane rigid correction was applied. However, it is worth noting that imperfect in-plane motion correction can lead to misalignment between control and label images resulting in subtraction artifacts and inaccurate estimation of renal perfusion. Because conventional signal averaging in the presence of rotation and other nonrigid motion in the abdominal area would likely produce blurring, a more robust realignment method such as elastic model may be more appropriate.

Our experiments were carried out on a 3 T scanner where the perfusion measurements benefited from an intrinsically higher SNR due to longer $T_1$ values. The perfusion sensitivity can be further improved with background suppression (29) and PCASL labeling pulses (10,27). However, $T_1$ inhomogeneity and off-resonance effects can compromise the labeling efficiencies of PCASL, especially in the abdomen. The specific absorption rate may pose as another limitation for PCASL at 3 T, although a recent study (9) has shown feasibility.

In summary, the 2D navigator gated ASL imaging was found to be a feasible method for quantifying renal perfusion without external contrast agent and compromising patient comfort. It provides a potential clinical solution for patients with impaired renal function who may not tolerate breath-holding. In patients with reduced renal flow, perfusion sensitivity may be improved by increasing the number of signal averages at a cost of increased scan time. A study with sufficiently large number of patients is warranted to test the effectiveness of this technique for routine clinical applications.

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