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Magnetization Spoiling in Radial FLASH Contrast Enhanced MR Digital Subtraction Angiography

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Abstract

Purpose—To increase the in-plane spatial resolution and image update rates of 2D MR Digital Subtraction Angiography (DSA) pulse sequences to 0.57×0.57 mm and 6 frames/s, respectively, for intracranial vascular disease applications by developing a radial FLASH protocol and to characterize a new artifact, not previously described in literature, which arises in the presence of such pulse sequences.

Materials/Methods—The pulse sequence was optimized and artifacts were characterized using simulation and phantom studies. With IRB approval, the pulse sequence was used to acquire time-resolved images from healthy human volunteers and patients with X-ray DSA-confirmed intracranial vascular disease.

Results—Artifacts were shown to derive from inhomogenous spoiling due to the nature of radial waveforms. Gradient spoiling strategies were proposed to eliminate the observed artifact by balancing gradient moments across TR intervals. The resulting radial 2D MR DSA sequence (2.6s temporal footprint, 6 frames/s with sliding window factor 16, 0.57×0.57 mm in-plane) demonstrated small vessel detail and corroborated X-ray DSA findings in intracranial vascular imaging studies.

Conclusion—Appropriate gradient spoiling in radial 2D MR DSA pulse sequences improves intracranial vascular depiction by eliminating circular banding artifacts. The proposed pulse sequence may provide a useful addition to clinically applied 2D MR DSA scans.

Keywords

MR DSA; Radial Imaging; View Ordering; Spoiling

Introduction

The successful diagnosis of intracranial vascular disease using CE MRA requires time-resolved datasets that are simultaneously high in SNR, temporal and spatial resolution. However, for a given FOV, these parameters have an inherent tradeoff. This can be especially troublesome for pathologies such as arteriovenous malformations (AVMs), where complex flow patterns require frame rates greater than 1 frame/s and spatial resolution

greater than 1 mm to visualize the anatomy of the nidus and the dynamic properties of feeding and draining vessels(1). For these reasons, X-ray DSA remains the clinical standard for intracranial applications providing in-plane spatial resolutions of up to 0.2mm and 5 frames/s temporal resolution. However, the intraprocedural thromboembolic risk as well as utilization of ionizing radiation and nephrotoxic contrast media makes it a less than preferable diagnostic modality.

There have been many successful time-resolved 3D MR techniques for first pass bolus tracking(2-9). Much of the research in 3D CE MRA has been directed towards increasing the spatial resolution and reducing temporal footprint of 3D techniques at the expense of SNR. For example, the “keyhole” technique (10) acquires an initial high-resolution dataset and updates only low-resolution information in subsequent acquisitions. The TRICKS method (3) acquires central k -space lines at higher sampling rates than peripheral lines. However, little attention has been given to 2D techniques such as MR DSA – also known as MR projection angiography. MR DSA improves spatial and reduces temporal footprint by eliminating 3D slice encoding providing close to sub-millimeter spatial resolution at frame rates near 1 frame/s (11,12).

Thus far, these improvements have been too modest to justify the tradeoff of 3D information; however when combined with acceleration strategies like angular undersampling and sliding window reconstruction, the added improvements in temporal and spatial resolution may justify its use as a comprehensive clinical exam. Existing MR DSA protocols utilize Cartesian sampling and suffer from signal loss in the excitation edges from poor RF flip angle profiles. We aim to address these issues by incorporating angular undersampling and non-selective RF pulses into our protocol. Angular undersampling, also known as radial sampling, has well-known advantages in speed and motion insensitivity. It is also more compatible with sliding window reconstruction (2) because low frequency data (central k -space) is updated regularly and frequently offering superior continuity in bolus dynamics than under a Cartesian scheme. In addition simulations have shown that Cartesian acquisitions can distort the bolus profile during the vascular time course depending on the direction of travel relative to the gradient read out direction, while radial acquisitions produce more uniform bolus profiles regardless of bolus direction (6).

In this article we report the development of a new MRA protocol which incorporates a radial FLASH acquisition into an MR DSA pulse sequence with sliding window reconstruction. Initially, our radial MR DSA protocol produced images with concentric circular banding artifacts. To our knowledge neither this pulse sequence nor the specified artifact, which we observed to arise in the presence of certain RF-spoiled radial trajectory FLASH sequences, have been described in literature. We investigate the nature of these artifacts and present a spoiling mechanism to eliminate them. This spoiling strategy is not included in off the shelf 2D radial FLASH pulse sequences from at least two major MRI sequence vendors. The nature of these artifacts will especially affect highly undersampled radial acquisitions, including 3D radial acquisitions, such as VIPR (13). Our subsequent MR DSA sequence is capable of acquiring a 0.57×0.57 mm in-plane spatial resolution projection angiogram with sub-second image update rates (~ 6 frames/s) using sliding window reconstruction. We evaluate the utility of this new pulse sequence in simulation, phantom, and healthy volunteer studies as well as in patients with X-ray DSA-confirmed neurovascular disease.

Theory

T_1 -weighted contrast in FLASH imaging is predicated on image acquisition using short-TR ($TR \ll T_2$) pulse sequences and spoiling residual transverse magnetization between

successive RF pulses. It is well known that the transverse magnetization of a spin, excited by a train of RF pulses, will accumulate a phase shift ϕ between subsequent pulses, given by

$$\phi = \int_0^{TR} \omega(t) dt \quad [1]$$

where $\omega(t)$ is the resonance offset and TR the pulse repetition time. Zur et al (14) demonstrated that if ϕ is the same during each TR interval, then spin subpopulations created by previous RF-pulses will add coherently and contribute to newly formed echo signals, distorting the FLASH contrast.

Eliminating, or spoiling, these coherences is achieved by varying ϕ between each TR interval such that previously created transverse magnetization will cancel prior to each subsequent RF-pulse. The $\Delta\phi(n)$, where $n = 1, \dots, N$ for N radial views, is commonly induced through RF-phase cycling or the application of gradients with variable moments. An example of the latter is an unbalanced Cartesian FLASH sequence where $\Delta\phi(n)$ occurs through the nominal phase-to-phase increments of the linear phase-encoding gradient, G_y . However, Crawley et al. (15), showed that unbalanced phase gradients in this pulse sequence confer spatial dependence to $\Delta\phi(n)$ resulting in banded regions of signal intensity signifying spatially dependent spoiling. For example, at the center of the image where $\Delta\phi(n) = 0$, spoiling does not occur, and the artifact is the greatest. Conversely, RF-spoiling creates $\Delta\phi(n)$ independently of position by adjusting the phase of the RF-pulse prior to excitation and is thus preferred. Utilizing a radial k-space trajectory in FLASH acquisitions confers spatial dependence to $\Delta\phi(n)$ in spite of RF-spoiling; simply rewinding all gradients undermines T_1 contrast. As a result, RF-spoiled radial FLASH acquisitions can produce spoiling inhomogeneity artifacts under certain conditions.

Materials and Methods

Our initial pulse sequence utilized a typical 2D FLASH protocol to acquire sagittal plane projection images using a radial trajectory. An analytical expression developed in a previous study (6) was used to order angular projections pseudo-randomly between 0 and π . This ordering method has been observed to minimize ghosting artifacts associated with bunched sampling in sequential radial ordered acquisitions (6,16,17). Non-selective RF-pulses were utilized to apply a uniform flip angle across the image volume in a minimal amount of time. RF-spoiling was applied to eliminate coherences from residual magnetization. Asymmetric echo was used to reduce TR and TE. This sequence produced images with signal artifacts in the pattern of concentric rings. The appearance of this annular artifact was not static on a frame-to-frame basis and thus could not be removed with simple mask subtraction.

We conducted simulation and phantom studies to predict and explain the appearance of these artifacts in 2D radial FLASH acquisitions under various view ordering strategies. MATLAB (Mathworks, Natick, MA) simulations modeled the evolution of spin magnetization using the Bloch equations. Our radial FLASH simulations were adapted from a previous study by Sekihara (18) that modeled the steady-state transverse magnetization in rapid Cartesian-trajectory MR pulse sequences. Phantom imaging studies were conducted to verify our simulations and quantify artifacts associated with different radial acquisition strategies. The results of our studies were used to adapt a spoiling strategy to be used in conjunction with RF-spoiling in radial sampling trajectories.

Simulations and Phantoms

Parameters chosen for our simulation and phantom studies are based on our original 2D MR DSA pulse sequence which is used in a typical human acquisition: field of view FOV =

220×220 mm, number of radial projections (N_θ) = 384, number of samples in the radial direction (N_{RO})=384, receiver bandwidth = 651 Hz/Pixel, Flip Angle = 30°, TR/TE = 4.55/1.85 ms. Simulation studies assumed T_1 = 300 ms and T_2 = 50 ms. RF-spoiling was utilized in all phantom and human imaging studies.

Artifact Calculation—Image artifacts in non-Cartesian phantoms were determined as the energy in the error of the signal. That is, if each Cartesian acquisition produces a “truth” signal $x(t)$ and each non-Cartesian acquisition produces an approximation to this signal, $a(t)$, then the error is defined as

$$e(t) = |x(t) - a(t)|, \quad [2]$$

and thus the artifact level is calculated as

$$\text{Artifact Level} = \frac{1}{N} \sum [e(t)]^2. \quad [3]$$

For each radial and Cartesian spherical phantom imaging study, 100 consecutive projection images were acquired and averaged to increase SNR. The averaged image was normalized by its mean signal. Cartesian phantoms, as defined, have zero artifact level.

Sampling Trajectory Study—Sampling trajectories investigated were Cartesian and sequential, non-sequential and pseudorandom radial view orderings. In sequential ordering, the angular spacing between consecutive views is Δk_θ such that, $k_\theta(n) = n\Delta k_\theta$ $n=1, \dots, N$. Non-sequential view ordering increments the line number by an integer μ so that the angular spacing between consecutive views is $\mu\Delta k_\theta$ until all lines are acquired, i.e. $k_\theta(n) = n\mu\Delta k_\theta$. We chose μ such that $\mu\Delta k_\theta = 111.246^\circ$ which is based on the Golden Ratio (19). Phantom images of a uniform sphere were acquired to confirm our simulation results.

N_θ , TR, and T_2 Study—We investigated the dependence of artifact level on N_θ , TR, and T_2 using phantom imaging studies. The effect of N_θ was investigated in sequential and non-sequential ordered radial schemes through phantom studies with 128, 256, and 512 views in a 220 mm FOV. In a separate study, TR was varied from 3.8-8.5 ms during pseudorandom ordered radial acquisition of spherical phantoms, and the corresponding artifact levels were calculated. Furthermore, nickel chloride and agarose solutions were used to create separate phantoms with T_2 = 20, 100, and 200 ms with T_1 = 300ms, (20,21) and the corresponding artifact levels were calculated after pseudorandom ordered radial 2D MR DSA acquisition.

Spoiling

Spoiling schemes were adapted from a Cartesian trajectory to a radial trajectory in order to eliminate gradient-induced, spatially dependent $\Delta\phi$ between each TR interval. Figure 1 illustrates two proposed spoiling strategies for radial FLASH acquisitions with pseudorandom ordering. Figure 1a illustrates a typical radial FLASH acquisition with unaltered gradient spoiling strategies. Figure 1b shows a ‘rewind and spoil’ strategy which utilizes rewinder gradients along G_x and G_y axes after each echo acquisition and a constant spoiler along G_x -axis at the end of each TR interval. In Fig. 1c, a ‘variable spoiler’ strategy is utilized in which spoilers along both the G_x and G_y axes vary their moments as a function of projection angle. That is, let the imaging gradients G_x and G_y be described as $G_x(n) = G_x \cos\theta(n)$, $G_y(n) = G_y \cos\theta(n)$ and $\theta(n)$ the radial projection angle, varies linearly from 0 to π such that $k_\theta(n) = \tan^{-1}[G_y(n)/G_x(n)]$ for $n=0, \dots, N-1$ radial views. Then spoilers $G_{s,x}$ and $G_{s,y}$ are calculated as

$$G_{s,x}(n) = G - G \cos\theta(n) \quad [4]$$

$$G_{s,y}(n) = G - G \sin\theta(n). \quad [5]$$

Each proposed gradient spoiling strategy also utilizes RF-spoiling. Figure 1 demonstrates that each spoiling method balances gradient moments, the time integral of gradient amplitudes, along the physical axes between TR intervals (Fig. 1b-c). The net gradient moments must be constant between TR intervals along each physical axis in order to spoil residual magnetization in a spatially homogeneous way. As Fig 1a shows, in a typical radial FLASH acquisition, gradient moments will be unbalanced between consecutive TR intervals. However, in the modified acquisitions (Fig 1b-c) gradient moments are balanced. We hypothesize this will eliminate the observed artifacts. Artifact level was calculated in phantom imaging studies for each proposed spoiling methodology to confirm reduction in artifact level.

Volunteer/Patient Studies

With IRB approval, we thus far tested the utility of this noninvasive, high resolution and frame rate 2D MR DSA technique on eighty healthy volunteers (6 men, 2 women, mean age 38.1 ± 11.2 years) and four patients (2 men, 2 women, mean age 68.8 ± 17.2 years) with X-ray DSA-confirmed intracranial vascular malformations (iVMs). Patients with DSA-confirmed iVMs in various stages of treatment with endovascular embolization underwent our MR DSA sequence with modified spoiling to assess treatment efficacy and residual lesions.

MR DSA—The same imaging parameters used in simulation studies were applied in our 2D MR DSA protocol for *in vivo* studies. Image resolution was varied using a 384×384 to 512×512 matrix within a 220-250 mm FOV on a study-by-study basis resulting in 0.49-0.57 mm in-plane resolution. Each new projection data set was acquired approximately every 2.6-3.4 s depending on the desired spatial resolution. Sliding window reconstruction was performed in MATLAB with sliding window factor (SWF) (6) of 16. Thus for $N_\theta = N_{RO} = 384$, each updated image contained 24 new projections, increasing the image update rate to ~6 frames/s. The radial undersampling factor was $\pi/2$. Radially acquired k-space data was re-gridded (22) and combined across each image coil using a sum of squares. Complex mask data subtraction was chosen because it emphasizes the difference in phase between stationary and moving magnetization as well as reduces the signal cancellation that occurs in magnitude subtraction (11). RF-spoiling was utilized in all human studies.

MR acquisitions were performed using a 3T Trio (Siemens Medical Solutions, Erlangen, Germany) MR imaging system and a 12-channel head coil. A power injector (Spectris Solaris; MEDRAD, Indianola, PA, USA) was used to administer a single dose (0.1 mmol/kg) of Gd-DTPA (Magnevist; Berlex, Wayne, NJ, USA) at 4 mL/s.

Results

Simulations and Phantoms

Figure 2 shows simulations of the evolving magnetization of two representative spins at spatial coordinates $(1,0)$ and $(2,0)$ over the first 8 projections of a 384 projection radial FLASH acquisition. A spin under a pseudorandom ordered acquisition, shown in Fig. 2a, evolves through a seemingly random pathway when compared to the sequential ordered acquisition shown in Fig. 2b. The pseudorandom pathways confer a 26% difference in the

net signal between the two spatial coordinates after just 8 projections. In the sequential pathways, the difference is just 0.0078 %.

In Fig. 3 a spherical phantom is imaged under an RF-spoiled MR DSA acquisition with four different sampling trajectories (Cartesian and pseudorandom, sequential, and non-sequential radial view ordering). Phantoms imaged with radial trajectories (Fig. 3b-d) all demonstrate the circular artifact stemming from spatially dependent $\Delta\phi(n)$ and non-uniform spoiling. Calculated artifact levels demonstrate that pseudorandom and non-sequential view ordered phantoms (Fig. 3c-d) produce the highest artifact levels at 0.0464 and 0.0581, respectively, while the sequential view ordered radial phantom has a lower artifact level of 0.0322.

Since the spatial variation in $\Delta\phi(n)$ is representative of the spatially varying spoiling due to gradients, it can serve as a metric for determining artifact level in simulation studies. We simulated the expected $\Delta\phi(n)$ in sequential view ordered radial acquisitions with different projection numbers, N_θ . In Fig. 4 the difference in $\Delta\phi(n)$ between representative coordinates $(1,0)$ and $(2,0)$ is shown as n varies over N_θ . That is, we observed $\Delta\phi_{(2,0)}(n) - \Delta\phi_{(1,0)}(n)$ over n radial acquisitions for $N_\theta = 128, 256$, and 512 radial views as a method of measuring the spatial variation of $\Delta\phi(n)$. For a linearly ordered acquisition with higher N_θ , the difference in $\Delta\phi(n)$ (in radians) between the two adjacent points decreases, resulting in greater spatial homogeneity in spoiling. This is confirmed in phantom studies in Fig. 5. Figure 5, row **N θ** shows spheres imaged with a sequential radial 2D MR DSA acquisition using $N_\theta = 128, 256$, and 512 projections have decreasing artifact levels of 0.0303, 0.0257, and 0.0177 respectively. Non-sequential radial 2D MR DSA acquisitions with different N_θ maintain the same angular spacing between consecutive projections and therefore demonstrate similar artifact levels. In imaging studies, for $N_\theta = 128, 256$, and 512 non-sequential radial acquisitions had 0.0378, 0.0332, and 0.0355, artifact level, respectively.

The **TR** and **T₂** rows in Fig. 5 demonstrate the dependence of the observed artifacts on TR and T₂. Artifact level in a radial pseudorandom view ordered MR DSA phantom is observed to decrease as larger TR intervals more effectively spoil residual transverse magnetization and mitigate the effects of spatial non-uniformity in $\Delta\phi(n)$. Similarly, shorter T₂ values reduce the artifact. Spheres in Fig. 5, row **TR** show a spherical phantom imaged with a pseudorandom radial acquisition with separate 3.8, 6.0, and 8.5 ms TR intervals demonstrating artifact levels of 0.0700, 0.0307, and 0.0218 respectively. Figure 5 row **T₂** shows T₂ phantoms with T₂ values of 20, 100, and 200ms imaged with a pseudorandom radial acquisition and demonstrating artifact levels of 0.0259, 0.0317, and 0.0403 respectively.

Spoiling

Figure 6 demonstrates the success of the two aforementioned gradient spoiling techniques combined with RF spoiling in eliminating the circular banding artifacts. In both techniques additional gradients are utilized to cancel the $\Delta\phi(n)$, due to the imaging gradients, between successive TR intervals. The resulting $\Delta\phi(n)$ is solely due to RF-spoiling. Therefore, it is spatially homogeneous and does not produce artifacts. The calculated artifact level is reduced from 0.0464 in the unmodified pseudorandom radial FLASH phantom in Fig. 6a to 0.0197 and 0.0153 for the ‘rewind and spoil’ (Fig. 6b) and ‘variable spoiler’ (Fig. 6c) phantoms, respectively. Both spoiling strategies increase the TR by 1.57 ms to 6.12 ms. However, we can exclude the possibility that the artifact reduction is purely an increased-TR effect because the reduced artifact level is lower than that of the 8.5 ms TR phantom.

Figure 7 demonstrates the improvement in vascular depiction between modified and unmodified sequences in a phantom study. Here a contrast enhanced tube, representing an enhanced vessel, is imaged with a Cartesian, unmodified pseudorandom radial, and modified

pseudorandom radial FLASH MR DSA sequences. The image error due to the unmodified radial FLASH sequence is 0.0517 while the artifact computed in the modified radial FLASH sequence is 0.0183 suggesting a significant improvement in vascular depiction with the modified sequence.

Human Volunteer and Patient Studies

Cartesian acquisitions do not require the modified spoiling implemented in our radial acquisition. As a result, the temporal footprint is 2.0s for a Cartesian acquisition of 0.57mm in-plane spatial resolution as compared to a 2.6s footprint radial acquisition of the same resolution. Figure 8 shows the implementation of a first-pass, time-resolved cranial Cartesian acquisition acquired from a healthy volunteer. The first four consecutive frames that visualized the bolus are shown, with each spaced ~2s apart and having in-plane spatial resolution of 0.57 mm. Cashen et al (6) showed that sliding window reconstruction cannot be implemented in a Cartesian scheme without compromising the continuity of bolus dynamics. As a result, image update rates beyond ~2s/frame are not achieved, and we are limited to a single frame for arterial and parenchymal phases.

Two first-pass, cranial, time-series radial MR DSAs are shown in healthy volunteers with a 'rewind and spoil' spoiling methodology in Fig. 9. Concentric ring artifacts are clearly eliminated. Each subsequent acquisition was spaced ~3.4s apart and had an in-plane spatial resolution of 0.49 mm. Sliding window reconstruction increased the image update rate to ~6 frames/s for improved arterial- and venous-phase dynamic information. Figure 9a-b show three separate frames of the arterial and venous phases from one volunteer at a 45° oblique projection angle. Figure 9c-d show images from a second volunteer taken at a sagittal projection. Selected ROIs to the right of each row show small vessel detail.

Figure 10 compares our MR DSA technique with X-ray DSA in a patient volunteer who had received stage 1 embolization treatment for a dural arteriovenous fistula (AVF). Spatial resolution here is 0.57 mm in plane. Both post-treatment MR and X-ray DSA images display persistent arteriovenous shunting and early venous drainage into the sigmoid sinus clearly indicating residual lesions requiring further intervention. Following successful stage 2 embolization, X-ray DSA corroborates MR DSA findings with elimination of arteriovenous shunting or early venous drainage through the sigmoid sinus.

Discussion

We have developed a pulse sequence which, for the first time, to our knowledge, applies a pseudorandom radial, FLASH protocol to acquire 2D MR DSA projection images. Thus far, this protocol has acquired angiograms with 0.49×0.49mm in-plane spatial resolution and 3.4s temporal footprint. In addition, sliding window reconstruction with SWF=16 was utilized to increase the image update rate to 6 frames/s.

We initially observed image artifacts in the form of concentric rings. We determined the nature of these artifacts to be spatial inhomogeneities in transverse signal arising from TR-TR variations in gradient moments inherent to certain radial FLASH acquisitions. In particular, 2D FLASH sequences utilizing pseudorandom sampling, large opening angles (such as golden angle) or high angular undersampling as well as 3D VIPR sequences will be more severely affected by these artifacts. Conversely, 3D FLASH sequences where through-plane phase encoding is looped within each radial projection have little-to-no TR-TR variation in gradient moments and consequently do not display these artifacts (6,23,24). We proposed a spoiling strategy to eliminate artifacts in our sequence based on balancing gradient moments within each TR interval; however, this is not necessary in 3D FLASH

sequences with the aforementioned phase encoding strategies. Applying our spoiling strategy in our sequence produced purely T_1 -weighted projection angiograms.

Our studies demonstrated that 2D radial FLASH sequences with unbalanced gradients lead to spatially dependent $\Delta\phi(n)$. Such $\Delta\phi(n)$ confer non-uniform spoiling across the FOV which leads to the observed concentric circular band artifacts in radial acquisitions. Our simulation and phantom studies showed that these artifacts are less apparent in sequential ordered acquisitions than in pseudorandom acquisitions due to the smaller spatial variation in $\Delta\phi(n)$ for sequential ordered trajectories. Similarly within a sequential ordering scheme, larger N_θ yields smaller artifact levels for the same reason. These artifacts can be eliminated by utilizing a spoiling strategy that balances gradient moments across the physical axes. This spoiling strategy ensures $\Delta\phi(n)$ is not spatially dependent.

Our spoiling strategies induce a TR prolongation of 1.57 ms, increasing the total TR of our sequence from 4.55 to 6.12 ms. At 384 projections, this increases the temporal footprint by ~ 0.6 s. This TR penalty may potentially be reduced in the ‘rewind and spoil’ scheme by applying spoiling gradients along the through-plane gradient (G_z). An important note is that although prolonged TR intervals increase temporal footprint, SNR is also increased with the square root of imaging time allowing potentially improved delineation of small vessels. Furthermore, sliding window reconstruction allows us to maintain a high frame update rate – 6 frames/s – which was not possible with previous Cartesian MR DSA sequences.

We believe the proposed 2D MR DSA pulse sequence and highly constrained projection reconstruction (HYPR) (5) or parallel imaging (9) are not competing techniques. Rather they can complement one another. It is known that sliding window reconstruction effectively low pass filters temporal data. However, with the proposed 2D technique a smaller acceleration factor is required (4x acceleration) in order to achieve sub-second image updates as opposed to similar 3D techniques (6,23,24) which require nearly (12x – 20x acceleration). Therefore, additional acceleration may be provided to the proposed sequence by utilizing parallel imaging or HYPR reconstruction in order to prevent greater temporal blurring effects with large sliding window factors.

Phase contrast highly constrained projection reconstruction (PC HYPR) flow and other hybrid MRA techniques (25,26) offer isotropic 0.69 mm spatial resolution and a temporal footprint of time resolved data of 0.75 seconds while providing both PC and time resolved MRA data sets. These time-resolved techniques, however, are prone to suffer from motion artifacts as their successful reconstruction depends on the acquisition of an additional high resolution PC exam unlike the proposed 2D MR DSA exam. Furthermore, techniques that utilize HYPR-LR (25,26) are computationally intensive and require offline reconstruction. The proposed MR DSA sequence requires little post-processing computation and can thus be implemented on hospital scanners to provide near instantaneous results for high clinical throughput.

Applications of 2D MR DSA

We believe that the proposed 2D MR DSA protocol may find application in certain clinical settings where higher spatial resolution and image update rates in 2D MR DSA techniques is lacking. In Yoshikawa et al. (27), 2D CE MR DSA was used to evaluate the intracranial hemodynamics in the setting of brain tumors. Mori et al. (28) reported the use of 2D MR DSA to observe flow dynamics in feeding vessels, nidi, and draining vessels of small and medium sized AVMs. Aoki et al (29) reported on the usage of 2D MR DSA in evaluating cerebrovascular occlusive diseases. However, these studies reported obscured visualization of small vessels which could be explained by the use of anisotropic in-plane spatial resolution ($0.5\text{mm} \times 1.5\text{mm}$), higher injection rates ($\sim 8\text{mL/s}$), and magnitude subtraction

instead of complex subtraction. Furthermore image update rates for intracranial applications were limited to ~1 frame/s as view-sharing was not desirable due to the use of Cartesian sampling.

The higher, in-plane spatial resolution and increase in image update rate in our protocol may improve the use of 2D MR DSA techniques in these areas. In a recent study by Edelman et al, a 4D-CE MRA sequence was successfully applied for the grade classification of intracranial AVMs (1); however the in-plane spatial resolution was a limiting factor ($1.1 \text{ mm} \times 1.1 \text{ mm}$ – nearly an order of magnitude behind X-ray DSA). By achieving higher in-plane spatial resolution (0.49 mm) and image update rates (6 frames/s) than existing protocols, our sequence has the potential to advance 2D MR DSA as a diagnostic and reduce the need for certain X-ray applications. For instance, the 2D MR DSA from the subject in Fig. 10 provided sufficient hemodynamic information to confirm the persistence of the dural AVF and plan further treatment without the need for DSA, fundamentally altering the management of this patient.

Our 2D MR DSA protocol could also see clinical application in imaging selective contrast injections through an established vascular access. Selective injections have the advantage of applying smaller doses of contrast material, allowing evaluation of singular vessels, and mitigating vessel overlap in projection MRAs. Through plane vessels and overlapping branches of the same vessel origin may be resolved with multiple projections through repeated acquisitions. Further studies are necessary to assess the potential and clinical utility of our 2D MR DSA protocol under these settings.

Limitations

The reduction in scan time at the expense of 3D information also causes a loss in SNR with the square root of imaging time as well as difficulty in resolving overlapping vessels. Currently, our use of non-selective RF-pulses reduces scan time and applies uniform flip angles across the volume; however, it prevents isolation of data to one hemisphere. X-ray DSA resolves these issues through the use of selective injections and multiple projections. Our method would significantly benefit from a similar, albeit less invasive, approach.

Conclusion

We have developed a novel pulse sequence to acquire 2D MR DSA through a thick volume using a radially sampled FLASH sequence with pseudorandom view ordering, sliding window reconstruction, and complex subtraction. This protocol allows us to maintain a high in-plane spatial resolution of $0.49 \text{ mm} \times 0.49 \text{ mm}$ and sub-second image update rates with minimal acceleration (4x) from sliding window reconstruction. We further utilized a spoiling strategy based on balancing gradient moments to eliminate circular banding artifacts, which we observed to occur with certain radial FLASH acquisitions. The development of a radial FLASH MR DSA sequence allows for the utilization of sliding window reconstruction and subsequent acceleration of image update rates in MR scans. Further testing of our sequence on patients with vascular pathologies is necessary to explore the potential of our protocol in a clinical setting.

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Figure 1.

Pulse sequence diagrams of consecutive projections in pseudorandom radial FLASH acquisitions with (a) unmodified and (b-c) modified spoiling. Unmodified spoiling results in unbalanced gradient moments while each modified spoiling method results in balanced, or equal, gradient moments among TR intervals eliminating the artifact. (b) The 'rewind and spoil' method uses rewinder gradients along each axis and a constant spoiler along G_x , and (c) 'variable spoiler' incorporates a projection-angle-dependent gradient spoiler moments along each axis. Gradient moments are shown along each axis at the bottom of each pulse sequence.

Figure 2.

Transverse magnetization evolution of spins in two view order schemes. Each graph outlines the pathways of two spins at (1,0) – solid line – and (2,0) – dashed lines – through the first 8 projections of a 384 view radial FLASH acquisition. In (a) the view order is pseudorandom and in (b) the view order is sequential. The two spins in (b) evolve through similar pathways – as a result the signal from each varies only slightly (0.0078%) between the two positions. However, the two spins in (a) have very different pathways and as result differ in signal by 26%.

Figure 3. RF-spoiled radial FLASH phantom images under (a) Cartesian (b) sequential radial (c) pseudorandom radial and (d) non-sequential radial view ordering schemes. Cartesian FLASH phantom shows a perfectly spoiled T1-weighted image; ring artifacts in radial FLASH phantoms distort T1-contrast. Artifacts are shown to change with different view orders.

Figure 4.

Plots of the difference in $\Delta\phi(n)$ between two spatial coordinates $(1,0)$ and $(2,0)$, i.e. $\Delta\phi_{(2,0)}(n) - \Delta\phi_{(1,0)}(n)$, is shown at increasing $N_\theta = 128$ (solid), 256 (small dash), and 512 (large dash) projections. At larger N_θ , $\Delta\phi_{(2,0)}(n) - \Delta\phi_{(1,0)}(n)$ approaches 0. As a result, $\Delta\phi$ is more spatially uniform diminishing inhomogeneities in spoiling.

Figure 5. Effect of N_θ , T2, and TR on artifact level in RF-spoiled radial FLASH phantoms. N_θ row shows increased spoiling uniformity and smaller artifacts with larger numbers of projections. TR row shows less artifact at larger TR, and T2 row shows larger artifact with larger T2.

Figure 6. Success of two gradient spoiling schemes in eliminating circular banding artifacts. (a) Shows an RF-spoiled, pseudorandom radial 2D MR DSA acquisition with unmodified gradient spoiling. Two modified gradient spoiling strategies are (b) ‘rewind and spoil’ and (c) ‘variable spoiler’ which combine with RF-spoiling to eliminate the artifacts in (a).

Figure 7. Simulated improvement in vascular depiction due radial FLASH 2D MR DSA with modified spoiling sequence in static tube phantom. (a) Cartesian FLASH acquisition – image artifact is 0 by definition. (b) Unmodified pseudorandom radial FLASH acquisition clearly showing ring artifacts with calculated artifact of 0.0517. (c) ‘Rewinder Spoiler’ modified radial FLASH acquisition eliminating ring artifacts and with reduced artifact of 0.0183.

Figure 8. Time-series of a 0.57×0.57 mm in-plane spatial resolution Cartesian 2D MR DSA acquisition in a healthy volunteer. The temporal footprint is smaller than the proposed radial acquisition by ~ 0.6 s. Only four temporal samples of the bolus can be visualized as Cartesian sliding window reconstruction distorts bolus continuity.

Figure 9.

2D MR DSA time-series in two volunteers. The ‘rewind and spoil’ methodology is utilized with complex subtraction. Resolution is 0.49×0.49 mm and images are spaced 0.5s apart. Enlarged ROIs to the right of each row demonstrate small vessel detail. Three frames of the arterial phase (a) and venous phase (b) of the first volunteer are shown at -45° from the sagittal to coronal projection. In the second volunteer, a sagittal projection is acquired and three frames of the (c) arterial and (d) venous phases are shown with enlarged ROI.

Figure 10.

Comparison of 2D MR DSA technique with X-ray DSA. (a) MR DSA and X-ray DSA of patient volunteer each demonstrating residual AVF lesion post 1st embolization treatment (arrows). (b) MR DSA and X-ray DSA post 2nd embolization treatment demonstrating elimination of venous shunt into sigmoid sinus (box).