Projection Imaging of Myocardial Perfusion: Minimizing the Subendocardial Dark-Rim Artifact

Behzad Sharif[1], Rohan Dharmakumar[1], Chrisandra Shufelt[2], Troy LaBounty[2], Louise Thomson[2], Noel Bairey Merz[2], Daniel S. Berman[2], and Debiao Li[1]
[1]*Biomedical Imaging Research Institute, Cedars-Sinai Medical Center, and University of California Los Angeles, Los Angeles, California, United States,* [2]*Heart Institute, Cedars-Sinai Medical Center, Los Angeles, California, United States*

INTRODUCTION Despite significant technical advances over the past decade, myocardial perfusion (MP) MR imaging is one of the most challenging dynamic imaging applications [1,2]. Current clinical MP MR methods suffer from image artifacts that reduce diagnostic accuracy, specifically the so-called subendocardial dark-rim artifact (DRA) [2-4]. DRAs are caused by multiple factors and mimic subendocardial perfusion deficits [1-3], and are especially limiting since they can reduce the sensitivity/specificity of detecting subendocardial MP deficits. Recently, there have been many attempts at minimizing DRAs in specific protocols, yet it remains an active area of ongoing research ([2] and references therein). In this work, we demonstrate that projection imaging of first-pass myocardial perfusion is robust to Gibbs ringing, a major cause of DRAs; and we therefore propose radial k-space sampling as a preferred acquisition scheme for DRA-free perfusion imaging.

THEORY In Cartesian MP MR, Gibbs ringing along the phase-encode (PE) direction leads to DRAs due to an inherent property of the Fourier transform (FT) at sharp signal intensity transitions [3,4]. However, in projection imaging (radial k-space acquisition), the underlying data transform is the Radon transform [5], which does not exhibit the Gibbs ringing in the same form as FT (instead shows streaking artifacts if highly undersampled). This fact is demonstrated in Fig. 1 using a numerical phantom simulation with realistic intensity values (108 PEs in Panel (a) versus 108 projections in Panel (b)). Fig. 1(c) shows a 1D cut along the PE direction, zoomed-in to the blood-pool/myocardium border. It can be seen that, unlike the Gibbs ringing in (a), the streaking artifacts in (b) do not result in a "signal dip" (which causes noticeable DRA in Panel (a)).

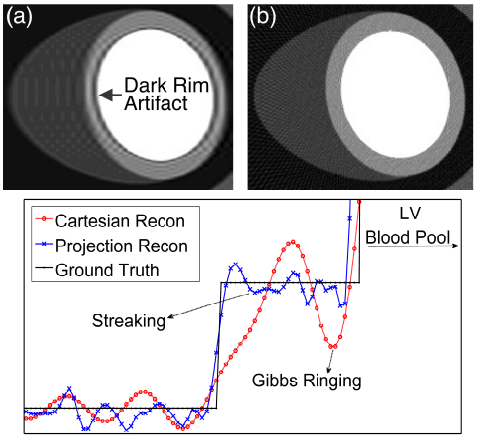

Figure 1. Numerical simulation (with realistic intensity values) demonstrating robustness of projection imaging to Gibbs ringing. (a) Cartesian imaging with 108 PEs (DRA is seen as highlighted); (b) projection imaging with 108 projections and filtered backprojection reconstruction (mild streaking artifacts are present). (c) a 1D cut along the PE direction, zoomed-in to blood-pool/myocardium border.

METHODS To investigate the validity of this observation in-vivo, healthy canines (N=3) and healthy human volunteers (N=4; IRB approved) were imaged on a clinical 3T scanner (Magnetom Verio, Siemens AG Healthcare, Erlangen, Germany) with a 32 channel cardiac-torso coil array. Two first-pass MP scans (SR-prepared FLASH) were performed at rest (>10 minutes gap) using a single-shot radial (customized) pulse sequence followed by a single-shot Cartesian (product) sequence (common parameters: FOV read = 270-300 mm; BW ≈ 800 Hz/pixel; flip angle = 12; TR/TE = 2.4/1.4 ms; TI = 100 ms). Both scans were accelerated using rate 2 self-calibrating parallel imaging (TGRAPPA for Cartesian and self-calibrating non-Cartesian SENSE for radial) with 36-40 readouts. The customized radial sequence incorporated adaptive gradient-delay correction and a 4-fold interleaving scheme combined with KWIC processing [6,7]. Although the readout resolution for the Cartesian scan was higher than the base resolution for the radial scheme (≈1.8 mm vs. ≈2.2 mm), the overall resolution of Cartesian (anisotropic with PE resol. ≈2.6 mm) and radial scans (isotropic ≈2.2 mm in-plane) were matched to within 10%.

RESULTS Figure 2 shows the reconstruction results for representative dog (a,b) and human (c,d) scans. Each pair of Cartesian/radial images correspond to the same phase of the contrast uptake (late-LV/early-myocardial enhancement). The arrows in panels (a) and (c) point to hypo-intensities that were considered DRAs. As is seen in the figure, Cartesian images (a,c) show DRAs (arrows pointing to hypo-intensities) in the subendocardial regions whereas the projection reconstructions (b,d) are free of dark rims (although with slightly reduced contrast-to-noise ratio).

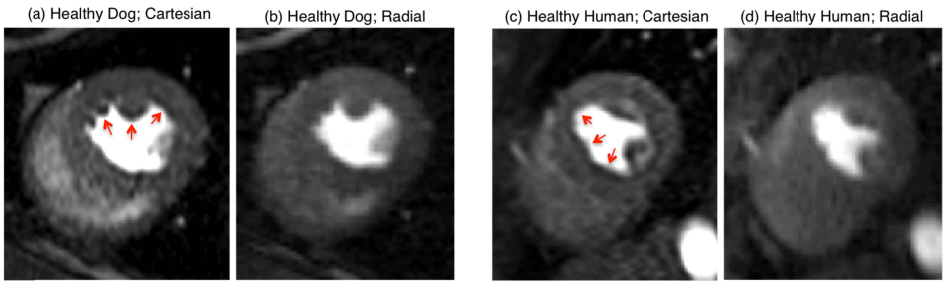

Figure 2. Representative in-vivo results for rest first-pass myocardial perfusion scans: (a,b) healthy canine model; (c,d) healthy human volunteer. Each pair of Cartesian/radial images corresponds to the same phase of the contrast uptake (late-LV or early-myocardial enhancement phase). The arrows in (a) and (c) point to dark-rim artifacts (DRAs). However, DRAs are not seen in the projection-reconstruction (radially sampled) images in (b) and (d).

DISCUSSION Recently, a major approach for eliminating the DRAs in first-pass MP MR has been to improve the spatial resolution (especially along the PE direction) and thereby reducing Gibbs ringing [8,9]. In this work, we demonstrated that projection imaging is inherently robust to Gibbs ringing, a major contributing factor to DRAs. We conclude that an alternative strategy (besides increasing the resolution) is to employ projection imaging (radial trajectories). The presented results were limited to rest scans, although we expect the same properties to hold for stress imaging. However, to match the high resolutions achieved in advanced Cartesian schemes (e.g., [8,9]), rate 2 parallel imaging acceleration is not sufficient and constrained (e.g., using compressed sensing) projection reconstruction will be needed.

REFERENCES [**1**] Wilke NM, et al., JMRI 1999;10:676-685. [**2**] Gerber BL, et al., JCMR 2008, 10(18). [**3**] DiBella EV, et al., MRM 2005;54:1295-59. [**4**] Ferreira P, et al., JCMR 2009;11(17). [**5**] Kak and Slaney, Principles of Computerized Tomographic Imaging, IEEE, 1988. [**6**] Song HK, et al., MRM 2000;44:825-32. [**7**] Peters DC, et al. MRM 2006;55:1150-6. [**8**] Plein S, et al., MRM 2007;58:777-85. [**9**] Otazo R, et al., MRM 2010; 64:767-76.