High temporal resolution Black Blood Vessel Wall Dynamic Contrast Enhanced Magnetic Resonance Imaging (DCE-MRI) using fast imaging methods

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Target audience: Radiologists and researchers who are interested in accelerating vessel wall DCE-MRI.

Purpose: Vessel wall dynamic contrast-enhanced (DCE) MRI is an emerging technique for in-vivo quantification of intra-plaque inflammation, an important mechanism in the initiation, progression and rupture of atherosclerotic plaque [1]. Due to the critical needs of high-spatial resolution and coverage in vessel wall imaging, the temporal-resolution was low [1]. To improve the accuracy of pharmacokinetic analysis, fast imaging is heavily needed in vessel wall DCE-MRI to increase the temporal-resolution. However, unlike in neural imaging and oncology area, the feasibility of the fast imaging method in vessel wall DCE imaging remains uncertain, especially for those utilizing temporal similarity in dynamic acquisition (k-t methods). One possible reason for the lag is that the acceleration of vessel wall DCE is difficult due to the small size of the imaging target, especially for arteries with early lesions (thickness<=1mm). In this study, we investigated the feasibility of a few fast imaging methods (keyhole[2], k-t GRAPPA[3], k-t PCA[4], k-t SPARSE-SENSE[5], k-t SLR[6]) in DCE-MRI of vessel wall with simulated and in-vivo DCE images with thin atherosclerotic vessel walls.

Net R	1	1.8	3.2
ΔT (ms)	3.93	2.46	1.47
Correlation	0.92	0.97	0.99

Table.1. results of numerical simualation. Correlation coefficient with true K^{trans} increases.

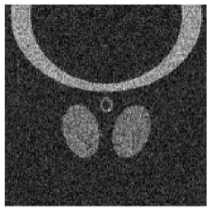

Fig.1. an image of numerical simulation

Methods: Data Simulation/Acquisition: To validate the advantage of higher temporal resolution, a numerical phantom was simulated. The phantom was designed to emulate the vessel wall and adjacent muscles, as showed in Fig.1. The thickness of vessel wall was 1mm and in plane resolution was 0.5mm. Initial T_1 of vessel wall, blood and muscle were 1.15s, 1.55s and 1.15s respectively [7].). Vessel wall intensity curve and arterial input function (AIF) were determined based on the imaging sequence [8], bi-exponential decay AIF model [9], and Patlak model [1]. 10 groups of K^{trans} and v^p value were assumed with K^{trans} varies between 0.05 min^{-1} and 0.2 min^{-1}, v^p varies from 0.04 to 0.24. For a total imaging time of 120 seconds, temporal resolution was 3.93s for fully sampled imaging, 2.46s and 1.47s for accelerated imaging (11 ACS lines, outer reduction factor of 2 and 4) respectively. Blood flow artifacts due to slow flow and pulsatile inflow waveform [10] were also considered. Gaussian noise was added to k-space to simulate the low SNR feature of vessel wall DCE-MRI. K-t GRAPPA was used in this numerical simulation test.

For in-vivo test, full k-space were acquired and undersampled retrospectively. The study used 10 male mature New Zealand White rabbits with balloon-injury-induced lesion in the abdominal aorta. All animals were scanned on a clinical 3.0T MRI scanner (Philips) using a human knee coil to obtain one slice of DCE image, with a QIR-based DCE sequence (cfIBBI) [8]. Only black blood images were used. Imaging parameters were the same as simulation study except that the temporal resolution was 7.9s. Five methods were adopted, keyhole, k-t GRAPPA, k-t PCA, k-t SLR and k-t SPARSE-SENSE. A variable density sampling pattern of 4-fold acceleration and 11 training profile was used for both k-t GRAPPA and k-t PCA to achieve net acceleration factor (R) 3.33. For keyhole, 48 central lines were acquired except for reference scan to achieve the same net acceleration factor 3.33. K-t SLR and k-t SPARSE-SENSE use random undersampling pattern with an acceleration factor of 3.

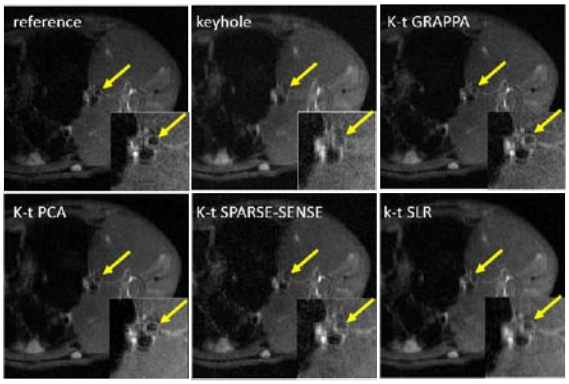

Fig.2 reconstructed images. Arrows indicate abdominal aorta. Reduction factor is 3.

Data Analysis: The vessel wall and reference region are outlined using a customized software (CASCADE [11]) and intensity curves were measured. The reference-region based Patlak model [11] was used to calculate K^{trans} and v^p in both simulation and in-vivo tests. Correlation coefficient was used to evaluate the agreement of the generated parameters between the reference images and the reconstructed images. Image quality was also evaluated blindly using objective scoring (1 (worst) to 4 (best)) by an experienced radiologist.

method	Correlation coefficient	Mean score
keyhole	0.92	2
k-t GRAPPA	0.91	3.1
k-t PCA	0.9	3.2
k-t SPARSE-SENSE	0.87	2.3
k-t SLR	0.92	3.2

Table.2. Evaluation of reconstructed image quality and kinetic parameter estimation for in-vivo data Correlations. All correlation coefficients are statistical significant at the 0.01 level (2-tailed).

Results: The result of the simulated data was shown in Table.1. With higher R (up to 3.2), the correlation between the fitted and the true value become higher. In in-vivo test (Table2), k-t GRAPPA, k-t PCA, and k-t SLR have relative good image quality (>=3.1) reserved with R=3 in early lesions (the maximum thickness from histology analysis of all rabbits was 0.69 ± 0.16 mm). Although keyhole and k-t SPARSE-SENSE have high correlation, but the image quality is quite low (Fig 2, Table 2).

Discussion and conclusion: For the first time, this study validated the feasibility of fast imaging method used in DCE-MRI of thin vessel wall with $R \approx 3$. The results proved that, it is worth well to increase the temporal resolution by fast imaging, leading a higher accuracy of fitted K^{trans} in thin vessel walls (thickness $\leq$ 1mm). K-t GRAPPA, k-t PCA and k-t SLR are good candidates to accelerate the acquisition for high temporal resolution vessel wall black blood DCE-MRI.

References

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