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Considering low-rank, sparse and gas-inflow effects constraints for accelerated pulmonary dynamic hyperpolarized ^{129}Xe MRI

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Dynamic hyperpolarized (HP) ^{129}Xe MRI is able to visualize the process of lung ventilation, which potentially provides unique information about lung physiology and pathophysiology. However, the longitudinal magnetization of HP ^{129}Xe is nonrenewable, making it difficult to achieve high image quality while maintaining high temporal-spatial resolution in the pulmonary dynamic MRI. In this paper, we propose a new accelerated dynamic HP ^{129}Xe MRI scheme incorporating the low-rank, sparse and gas-inflow effects (L + S + G) constraints. According to the gas-inflow effects of HP gas during the lung inspiratory process, a variable-flip-angle (VFA) strategy is designed to compensate for the rapid attenuation of the magnetization. After undersampling k-space data, an effective reconstruction algorithm considering the low-rank, sparse and gas-inflow effects constraints is developed to reconstruct dynamic MR images. In this way, the temporal and spatial resolution of dynamic MR images is improved and the artifacts are lessened. Simulation and *in vivo* experiments implemented on the phantom and healthy volunteers demonstrate that the proposed method is not only feasible and effective to compensate for the decay of the magnetization, but also has a significant improvement compared with the conventional reconstruction algorithms (*P*-values are less than 0.05). This confirms the superior performance of the proposed designs and their ability to maintain high quality and temporal-spatial resolution.

1. Introduction

Magnetic resonance imaging (MRI) is widely used for diagnostic and therapeutic purposes because of its clinical safety and superior soft tissue contrast to conventional imaging methods (such as computed tomography, X-ray). Although conventional ^{1}H MRI has been applied for most human organ systems, it is rarely utilized for the lungs due to the very low density of ^{1}H in the lung tissue. Through the spin-exchange optical pumping (SEOP) technique, hyperpolarized (HP) gas (such as ^{3}He and ^{129}Xe) MRI makes imaging of both pulmonary ventilation [1] and diffusion [2] feasible. This can provide the structural and functional information about the lungs, such as the microstructure, gas–gas and/or gas-blood exchange functions [3–5]. Accordingly, HP gas MRI offers great potential in the early detection of lung diseases, for example chronic obstructive pulmonary disease (COPD), emphysema and cystic fibrosis [2,6].

Many studies using HP gas MRI (such as the ventilation imaging [5], dissolved-phase ^{129}Xe MRI [7], diffusion-weighted imaging [8]) acquire lung images during the time interval of the breath-hold (denoted as static HP gas MRI). However, such static HP gas MRI does not visualize the dynamics of the pulmonary structure and/or function during the ventilation process, which potentially masks some abnormal regions, such as collateral ventilation and pulmonary air leaks [9,10]. It is known that the dynamic imaging of ventilation process has great potential to provide unique information about lung physiology and pathophysiology, which is valuable for clinical diagnosis and prognosis evaluation. Therefore, these findings warrant the development of dynamic HP gas MRI.

In 1996, MacFall et al. first obtained dynamic images of human lungs using a fast gradient echo sequence, at a temporal resolution of 1800 ms [11]. In 2000, Gierada et al. utilized echo-planar imaging with a temporal resolution of 40 ms in dynamic HP gas MRI of the lungs, but the spatial resolution was seriously constrained [12]. In the same year, Viallon et al. performed dynamic HP gas MRI with

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interleaved spiral cine sequences [13]. Although the acquired images had a high temporal and spatial resolution, the temporal resolution was pseudo due to the combination with the sliding window reconstruction [14]. In 2001, Salerno et al. performed dynamic HP gas MRI in humans using interleaved spiral pulse sequences, which visualized regional ventilatory patterns with high temporal and spatial resolution [15]. However, large number of interleaves in spiral pulse sequences limited the SNR of the image [16,17]. In 2003, Wild et al. demonstrated the feasibility of the radial projection method in dynamic HP gas MRI using sliding window reconstruction [18]. High temporal and spatial resolution can be achieved in this work, but with a loss of SNR and increased radial streaking artifacts [14]. In 2008, Holmes et al. presented 3D dynamic imaging of the lung using a multiple echo vastly undersampled isotropic projection reconstruction sequence [19]. This method resulted in high spatial resolution, but blurred the image details and produced streaking artifacts. It can be found that there exists a trade-off between image quality and temporal-spatial resolution for dynamic HP gas MRI because of the nonrenewable properties of HP longitudinal magnetization. This necessitates an effective dynamic HP gas MRI technique, aiming to simultaneously achieve high image quality and temporal-spatial resolution.

To reduce the disadvantages due to the nonrenewable longitudinal magnetization in dynamic HP gas MRI, we adopt compressed sensing (CS) strategy in this paper because CS can recover signals and images accurately using significantly fewer measurements than the number required by traditional Nyquist sampling [20–22]. CS has been widely applied to accelerate the MRI acquisition by undersampling k-space data [23–25]. Moreover, CS has also successfully been used in the dynamic MRI, such as the dynamic total variation [26], blind CS [27], and low-rank plus sparse matrix decomposition (L + S) methods [28]. However, in dynamic HP gas MRI, when HP gas is inhaled into the lung, considerable effects on the MR signals are generated (i.e., gas-inflow effects), which trigger complicate changes of signal intensity [16,29,30]. These changes will bring some unfavorable factors (such as noise and artifacts) in both sampling and reconstruction processes, limiting the image quality. Consequently, the gas-inflow effects should be considered in the pulmonary dynamic HP gas MRI.

In this work, we develop a new variable-flip-angle (VFA) scheme considering the gas-inflow effects to compensate for the decay of HP longitudinal magnetization, and then design an effective reconstruction algorithm incorporating the low-rank (L), sparse (S) and gas-inflow effects (G) constraints. This aims to acquire pulmonary dynamic images with high quality and temporal-spatial resolution. Meanwhile, HP ^{129}Xe is preferred in this work due to its lower cost and wider availability in regional lung function assessment compared with HP ^{3}He [30,31].

2. Methods

2.1. Data acquisition

In static HP gas MRI with a constant-flip-angle (CFA) scheme, the HP gas signal intensity after the n-th radiofrequency (RF) excitation is described as follows if the longitudinal relaxation time (T_1) is ignored [16,32],

$$S_1(n) = M_0(\cos\theta)^{n-1}\sin\theta,\ M_0 = \mu\cdot V, \tag{1}$$

where θ is the flip angle, M_0 is the initial longitudinal magnetization, μ is a constant relating to the initial polarization and spin density of HP gas, and V is the volume of inhaled HP gas. Eq. (1) shows the decay of longitudinal magnetization of HP gas due to the RF excitations, which suggests that a smooth k-space filter is imposed on the data and results in blurring the image details [14]. VFA schemes were developed to compensate for the losses of magnetization by gradually increasing flip angles [33–35]. However, the gas-inflow effects potentially limit the performance of conventional VFA schemes in pulmonary dynamic HP gas MRI. The inhaled HP gas will bring in more available magnetization during the dynamic imaging, which may increase the weight of high frequency part of k-space data. Consequently, a new VFA scheme is developed that does not only compensate for the depolarization caused by the preceding RF pulses (as is the case with conventional VFA schemes) but also takes into account the continuous gas inflow. Moreover, undersampling strategy is adopted to accelerate the imaging speed of dynamic HP gas MRI. Fig. 1 shows the timing diagram of the data acquisition process, where M, N and K are the number of phase encoding (k_y), frequency encoding (k_x) and frames, a is the acceleration factor (AF), and θ_n, $n = 1,\ldots, (M/a)\times K$ are the variable flip angles. Variable density undersampling patterns along the phase-encoding direction are generated using a Monte Carlo algorithm and extended to minimize the peak interference of the transform point spread function (TPSF) [23].

To obtain favorable variable flip angles, we introduce a concept of average signal intensity (ASI). Assuming that $S_2(n)$ and $V(n)$ denote the HP gas signal intensity and the volume of inhaled HP gas after the n-th excitation in dynamic MRI, the ASI of HP gas [$S_p(n)$] is

$$S_p(n) = S_2(n)/V(n),\ n = 1,\ldots,(M/a)\times K, \tag{2}$$

Moreover, we assume that the inhaled flow rate (r) is a constant. Then, $V(n)$ is equal to the product of r and inspiratory time T ($T = n\cdot$TR, where TR is the pulse repetition time). Owing to the continuous inspiration, $V(n+1) = V(n) + r\cdot$TR. Accordingly, the signal intensity after ($n+1$)-th excitation is

$$S_2(n+1) = \frac{S_2(n)}{\sin\theta_n}\cdot\cos\theta_n\cdot\sin\theta_{n+1} + \mu\cdot r\cdot \mathrm{TR}\cdot\sin\theta_{n+1}, \tag{3}$$

where θ_n and θ_{n+1} are the flip angles used in the n-th and ($n+1$)-th excitations. In this way, $S_2(n)$ can be considered as the sum of the signal intensity of the HP gas inhaled in each pulse TR, that is

$$S_2(n) = r\cdot\mathrm{TR}\cdot\mu\cdot\sin\theta_n\cdot\left(1+\sum_{i=1}^{n-1}\prod_{j=i}^{n-1}\cos\theta_j\right), \tag{4}$$

Thus, the ASI after the n-th excitation can be expressed as

$$S_p(n) = S_2(n)/V(n) = \frac{1}{n}\cdot\mu\cdot\sin\theta_n\cdot\left(1+\sum_{i=1}^{n-1}\prod_{j=1}^{n-1}\cos\theta_j\right), \tag{5}$$

The gas-inflow effects make the decay of the magnetization more complex (16), which results in difficulties in depicting the dynamics of the HP gas signal. It can be seen from Eq. (5) that with a constant flip angle the ASI is monotonically decreased with the excitation number, which indicates that the ASI can reflect the filter imposed on the k-space data. Therefore, it may be a feasible way to compensate for the filter by maintaining a constant ASI. We intend to compensate for the decay by maintaining a constant ASI, namely, $S_p(n) = S_p(n\text{-}1)$, $n = 2, \ldots, (M/a)\times K$. Therefore, the flip angles used in the proposed scheme are [See Appendix A],

$$\theta_n = \arcsin\left\{\frac{n\cdot\sin\theta_1}{(n-1)\cdot\sin\theta_1\cdot A(n)+1}\right\},\ \text{where } A(n) = \frac{\cos\theta_{n-1}}{\sin\theta_{n-1}}. \tag{6}$$

If the initial flip angle θ_1 is known under a given n (or if the θ_n is defined), a series of flip angles are determined. For example, when the excitation number is 960 and the initial flip angle is 2.42°, $\theta_{960} = 31.81°$.