Improved visualization of pulmonary parenchyma using SSFP sequence for dynamic MR-studies

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Introduction

Proton based lung imaging is a difficult task in MRI. Predominance of air within the pulmonary tissue implying low proton density reduces the signal intensity. A large number of air-tissue interfaces in lung alveoli leads to local field inhomogeneities inducing high susceptibility differences on intra-voxel scales, which are responsible for rapid signal decay. In lung MRI, due to the fast signal dephasing, respiratory motion, cardiac pulsation, short acquisition times or the application of triggering techniques are required. The Steady-State Free Precession (SSFP) pulse sequence was proposed for morphological and non-contrast-enhanced functional lung imaging at a low magnetic field 0.2 T [1-2] and at a high field of 1.5 T [3-4]. SSFP offers very fast image acquisition and sufficient signal-to-noise ratio (SNR) in the lung tissue. The aim of this work was to numerically simulate and further optimize the SSFP imaging scheme for dynamic MR-studies.

Methods

Measurements were performed on a 1.5 T whole body MR-scanner (Siemens MAGNETOM Avanto, Erlangen, Germany) using a 12-channel thorax/spine coil as receiver and transmitter. Five healthy volunteers were examined. Sets of coronal lung images were acquired using time-resolved scans with a 2D+t balanced SSFP sequence. The imaging scheme is presented in figure 1. Parameter TW characterized the time interval between each image acquisition. Scans were performed for various parameters: TE (0.8 – 1.6 ms), TR (1.9 – 3.7 ms), TW (0 – 400 ms), flip angle α (5-100º) with a 180º phase cycling every RF excitation, bandwidth BW (698 – 1502 Hz/px), centric and linear k-space sampling. The fixed sequence parameters were: FOV = 450^2 mm^2, slice thickness = 15 mm, matrix size = 128 x 128 (bicubic interpolated to 256 x 256), asymmetric echo sampling (factor s=0.4). Parallel imaging (GRAPPA factor 3) and acquisition of 24 auto-calibration signal lines (ACS) were applied prior to each image scan. A ramp of 10 RF pulses increasing linearly from α/10 to α was used to stabilize the magnetization vector. After the end the acquisition, magnetization was refocused using an α/2 RF pulse. To determine optimal SSFP sequence parameters improving the visibility of lung tissue, simulations of the magnetization vector evolution using Bloch equations were performed. The T_1 and T_2 relaxation times of the lung parenchyma required for the simulations were measured using IR-HASTE and T_2-prep SSFP sequences in every volunteer. The SNR dependence on the bandwidth assuming that T_1>>TR can be expressed as:

$$\mathrm{SNR} \propto \frac{\rho\, e^{-\mathrm{TE}/T_2^*}}{\sqrt{\mathrm{BW}}} = \frac{\rho\, e^{-(\tau+s/\mathrm{BW})/T_2^*}}{\sqrt{\mathrm{BW}}} \qquad \text{where} \qquad \mathrm{TE} = \tau + \frac{s}{\mathrm{BW}}$$

Here: ρ - proton density, τ - time between middle of RF and acquisition window, s – asymmetric echo sampling factor. All the dynamic scans were acquired in free breathing, thus non-rigid image registration was mandatory to correct for the respiratory motion [5]. Lungs were manually segmented to calculate the signal intensity in the pulmonary parenchyma.

Results

A dynamic imaging scheme using a combination of the central k-space sampling, parallel imaging, high bandwidth and minimal inter-echo sampling allowed to enhance the lung parenchyma signal. The amplitude of the magnetization vector for different values of α after subsequent RF excitations was simulated and is shown in Fig. 2a. The measured signal intensity in the lung tissue for various α compared to theoretical predictions was depicted in Fig. 2b. The SNR dependence on the applied bandwidth shown in figure 2c was measured and calculated theoretically. Centric k-space sampling and the introduction of intervals TW between each image acquisition increased significantly the signal compared to other sampling schemes (Fig. 2d). The drawback of the centric sampling was the enhancement of the pulsation artifacts in the presence of pulmonary arteries and heart. For the parameters TE/TR/TA/TW=0.8/1.9/116/180 ms, BW=1302 Hz/px and α=75º the average SNR of 37.5 ± 6.8 was achieved. An example of a SSFP lung image in a log-grayscale to depict the pulmonary tissue is presented in Fig. 3.

Discussion

Fast imaging using the SSFP sequence enhances the signal in the pulmonary parenchyma sufficiently for functional lung MRI. The proposed imaging scheme utilizes very short TR which widens the SSFP frequency response function and helps to avoid banding artifacts. Due to the very short acquisition times per image, transient signal behavior was dominant. Parallel imaging combined with asymmetric echo sampling significantly reduced scan time. Minimization of the TW can be used to increase the temporal resolution up to 10 frames/s and can be used for breathing maneuver studies. The theoretical calculations of the transient signal behavior fit well with the experimental data. In future work the implementation of a variable flip angle may further improve the imaging scheme providing higher signal intensity.

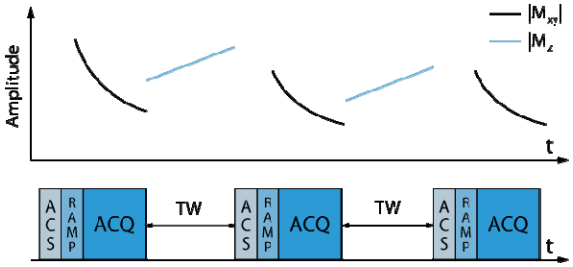

Fig. 1. Imaging scheme used for dynamic MR measurements. TW is defined as a time interval between each image acquisition.

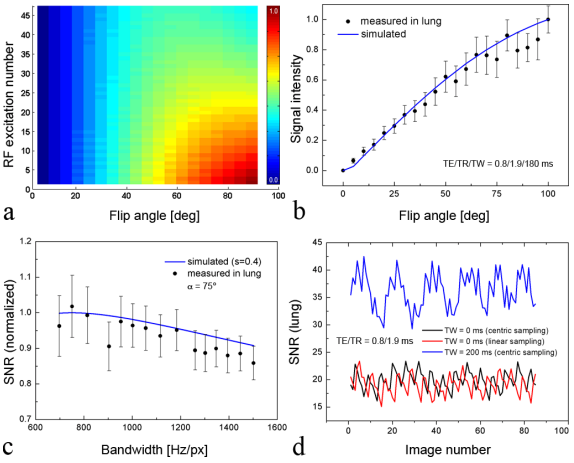

Fig. 2. Diagrams (**a**) and (**b**) present simulations and measurements of signal intensity in the lung tissue for different α values. The SNR as a function of the bandwidth and for various imaging schemes is shown in diagrams (**c**) and (**d**).

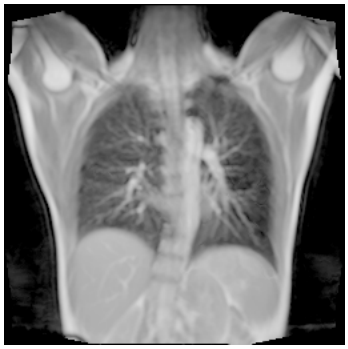

Fig. 3. An example coronal SSFP thorax image from a time-resolved MR-data set. (TE/TR=0.8/1.9 ms, α=75º). The image is presented in the log-grayscale look-up table.

References:

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