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A 32-Channel Phased-Array Receive with Asymmetric Birdcage Transmit RF Coil for Hyperpolarized Xenon-129 Lung Imaging

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Abstract

Hyperpolarized xenon-129 (HP Xe) has the potential to become a non-invasive contrast agent for lung MRI. In addition to its utility for imaging of ventilated airspaces, the property of xenon to dissolve in lung tissue and blood upon inhalation provides the opportunity to study gas exchange. Implementations of imaging protocols for obtaining regional parameters that exploit the dissolved phase are limited by the available signal-to-noise ratio (SNR), excitation homogeneity, and length of acquisition times. To address these challenges, a 32-channel receive-array coil complemented by an asymmetric birdcage transmit coil tuned to the HP Xe resonance at 3T was developed. First results of spin-density imaging in healthy subjects and subjects with obstructive lung disease demonstrated the improvements in image quality by high resolution ventilation images with high SNR. Parallel imaging performance of the phased-array coil was demonstrated by acceleration factors up to three in 2D acquisitions and up to six in 3D acquisitions. Transmit-field maps showed a regional variation of only 8% across the whole lung. The newly developed phased-array receive coil with the birdcage transmit coil will lead to an improvement in existing imaging protocols, but moreover enable the development of new, functional lung imaging protocols based on the improvements in excitation homogeneity, SNR, and acquisition speed.

Introduction

Hyperpolarized-gas MRI employs the noble gases helium-3 (He-3) or xenon-129 (Xe-129) as non-invasive contrast agents for imaging of lung structure and function [1-3]. The technical implementation of hyperpolarized-gas MRI brings unique challenges. The magnetization available for image encoding is in a non-equilibrium state which is created externally in a laser-based polarization system rather than, as is the case for conventional proton MRI, in the scanner magnet itself. The hyperpolarized gas is inhaled by the subject and subsequently imaging is performed, typically during a breath hold. The available magnetization decays in the lung, mainly due to interactions with the oxygen gas, with a T1 of 16 - 22 s, depending on the local oxygen concentration [2]. RF excitation pulses further contribute to the irreversible depletion of the magnetization. To preserve magnetization over the course of an image acquisition, excitation pulses typically use low and optimized flip

angles. Initially, research in hyperpolarized-gas lung MRI was focused on He-3 rather than Xe-129 given the more advanced polarizing technology available at the time. However, Xe-129, which is an atmospheric gas and hence naturally abundant, recently regained interest due to the restricted supply of He-3. Xe-129 compared to He-3 brings opportunities and challenges. As for the opportunities, we note that Xe-129 is lipophilic and therefore, unlike He-3, dissolves in lung tissue structures and blood. Moreover, in this so-called “dissolved phase,” hyperpolarized Xe-129 (HP Xe) experiences a large chemical shift of approximately 200ppm in its NMR frequency relative to its gas-phase resonance [4]. Frequency-selective manipulation of HP Xe magnetization residing in blood, tissue or air space compartments [2, 5] allows information about lung function, such as gas exchange [6-9] and uptake [8, 10, 11], to be obtained. However, compared to hyperpolarized He-3 (HP He), HP Xe has been more challenging to image. This has been mainly due to two factors: First, the gyromagnetic ratio of Xe-129 is approximately three times lower than that of He-3. Second, lower polarization values were achieved with Xe-129. Both factors contributed to a lower image SNR. While the first is an intrinsic property of the gas, the second is related to hyperpolarization technology, where significant improvements were demonstrated recently [12, 13], overcoming polarization and production rates differences between HP He and HP Xe.

In recent years, progress in hyperpolarized-gas MR imaging was made in the optimization of pulse sequences [14-16], development of new contrast mechanisms [7-11, 17-19] and theoretical modeling [20]. Only little work was done towards dedicated imaging hardware, specifically RF coils. The few reports of dedicated RF coils for HP He showed large improvements in image quality and acquisition speed [21-25]. Specifically, parallel imaging, which employs the sensitivity profiles of phased-array receiver surface coils to replace image encoding RF excitations, allows accelerated acquisition speeds. This is important in hyperpolarized lung imaging, as with the recent improvements in hyperpolarization technology, the SNR becomes sufficient for high resolution imaging. This however requires additional image encoding and therefore is then limited by acquisition length, since imaging has to be completed within a breath hold, sometimes of only a few-seconds duration in the case of severe lung disease. In hyperpolarized-gas MRI, an accelerated acquisition with fewer k-space excitations combined with the use of higher flip angles partially compensates for the SNR loss typically accompanying accelerated proton imaging [26, 27]. This makes parallel imaging especially attractive in hyperpolarized-gas MRI.

A phased-array receiver coil for parallel imaging must be combined with a volume transmit coil. Since the resonance frequency of hyperpolarized gases is different from the proton frequency, the MRI scanner integrated body coil cannot be used for X nuclei, but a dedicated transmit coil must be built instead. While for the receive coil sensitivity and a high number of channels for parallel imaging capability is important, for the transmit coil a homogeneous and, in the case of limited transmit power, high efficiency is the design goal. Homogeneity is especially important for imaging studies involving precise contrast pulses such as Xenon polarization Transfer Contrast (XTC) [6-9], where a regionally varying excitation field would introduce signal variations that mask the desired physiological contrast.

The first phased-array receive coils built for HP He were presented by Lee et. al., who also demonstrated accelerated hyperpolarized gas imaging [26, 27]. Later, Meise et al presented significant improvements in image quality and acquisition speed using a 32-channel phased-array receive coil for HP He [22]. An asymmetric transmit/receive birdcage coil for HP He at 1.5 T designed by De Zanche et al showed superior transmit homogeneity compared to previously used volume coil designs for hyperpolarized-gas imaging [23]. Recently, a combined approach of an asymmetric birdcage and an 8-channel receive array was presented

by Porea et al for HP Xe at 1.5 T [24] and using a 32-channel receive array for HP He at 1.5T by Deppe et al [25].

The goal of this work was to develop a chest coil to address challenges in HP Xe imaging by providing a homogeneous transmit field and the ability to speed up image acquisitions by enabling parallel imaging using a large number of receiver channels. To this end we built and characterized a 32-channel phased-array receive array coil combined with an asymmetric birdcage transmit coil for HP Xe at 3T. We acquired B1 maps to assess the transmit homogeneity and high resolution accelerated HP Xe spin-density images to demonstrate parallel imaging performance and receiver sensitivity of the combined HP Xe chest coil.

Methods

Asymmetric Birdcage Design

Birdcage coil designs are well-known to provide superior transmit field homogeneity. While birdcage coils typically have a circular cross section, in our case an asymmetric design was implemented to maximize the space available inside the magnet bore given also the patient table occupying the lower side of the bore. This way we were able to fit both coils, receive array and volume transmit coil, while still providing fit space for the human subject. The asymmetric design brings challenges in tuning the birdcage to achieve a homogeneous transmit field. This problem had previously been solved theoretically and experimentally by De Zanche et al for a HP He coil [23]. Our asymmetric transmit coil consisted of a circular top half with a radius of 28.2 cm and an elliptical bottom half with axes $a=28.2$ cm and $b=14.2$ cm (Fig. 1). The top half was detachable for easy patient access (Fig.2). The two halves were joined together via a mechanical latch and electrically connected with banana-style plugs. The birdcage employed a high-pass design by placing capacitors in the endrings. The HP Xe birdcage incorporated 12 rungs. Due to the asymmetric design the tuning capacitors were not of equal value. High-voltage capacitors (25-series, Voltronics Corporations, Denville, NJ) were used in the transmit birdcage with values approximately 20% higher (an estimate to account for differences in HP Xe and HP He resonance frequency as well as overall coil size) than the values theoretically derived by De Zanche et al for a HP He coil [23]. While fine tuning the coil, we tried to maintain the symmetry by adjusting capacitors on all positions simultaneously.

Unlike the coil of De Zanche et al, our HP Xe birdcage did not incorporate an RF shield wrapped around the coil. We decided to omit the shield to be able to easily upgrade the existing design in the future by implementing proton traps to allow for simultaneous acquisition of proton images using the proton body coil.

The transmit coil needs to be detuned during the receive phase of the imaging sequence in order to avoid coupling to the receive array during signal acquisition. This was achieved by incorporating high power PIN diodes (HUM2020 series, Microsemi Corporations, Irvine, CA) in series in the endrings, which are switched using DC-bias lines connected through a choke. Finally, in order to switch the signal pathway between transmit and receive states, a self-made quadrature hybrid was implemented. The quadrature hybrid also enabled excitation of a circularly polarized transmit field by introducing a 90-degree phase difference between the two driving ports of the birdcage, which were located at geometrically orthogonal positions. The two modes are coupled vertically and horizontally. (Since the two orthogonal modes of the ellipse are not equally efficient with amplifier power, we note that higher efficiencies can be achieved with single mode excitation for this transmit-only application).

Array Design

The 32 surface-coil elements of the receive array were placed in a hexagonal tiling pattern [28] on a two-component rigid shell, shaped to closely fit an adult human chest. The components consisted of a chest plate containing 15 receive elements and a back plate containing 17 elements (Fig. 3). Neighboring receiver coils were arranged such that the center-center distance was kept at 0.75 times the diameter to cancel mutual inductance and obtain signal decoupling for the respective elements [29]. The 12-cm diameter, circular elements were shaped from copper wire (AWG 12). Preamplifiers were tuned to the xenon resonance at 34.1 MHz and had a gain of 32 dB at the resonance frequency (Stark Contrast, Erlangen). They were mounted directly above the surface-coil element. A circuit diagram for the surface-coil receiver element is shown in Fig. 3b. The implemented circuitry on the receiver element was adjusted to provide tuning, matching, preamplifier decoupling and detuning during the transmit phase. All capacitors were non-magnetic low voltage capacitors (11-series, Voltronics Corporations, Denville, NJ). The capacitor C_t and the capacitors in the active and passive detuning traps are for tuning the coil to the resonance frequency of HP Xe at the 3-T field of the whole-body MRI system (TIM Trio, Siemens Healthcare, Erlangen, Germany). To detune the receive elements during the transmit phase of the imaging sequence, a resonant trap circuit was added over one of the tuning capacitors, which was activated using a PIN diode (MA4P4002B-402, Macom, Lowell, MA). The bias voltage for the PIN diode was applied through a choke in order to separate RF- and DC-current pathways. The ratio of all tuning caps and C_p , C_s is adjusted for matching the coil to the 50 Ohm impedance of preamplifier input and cables. The second purpose of C_p and C_s is to provide preamplifier decoupling by forming a resonant circuit together with the preamplifier input impedance. The high impedance achieved within the coil loop prevented current flow for effective signal decoupling, which added to the geometric overlap decoupling for nearest neighbor coils and it was the only method of signal decoupling for non-overlapping pairs of coil elements.

After the preamplifier, the coaxial cable transmitting the amplified signal was combined with cables from other preamplifiers leading to plugs which connected the coil to the MRI scanner. The coaxial cables were led in groups of four, six, or eight shielded by a copper braid and wound into a balun to prevent common mode currents. The cable lengths between the preamplifier output and balun were ranging from 20 to 100 cm depending on the individual element location. The balun consisted of the cable assembly (coaxes and braid) wound into an inductor and tuned to the resonance frequency with a parallel capacitor to form a resonant trap circuit.

Bench testing

The coil quality factor (Q-factor) of single, isolated array coil elements was determined by measuring the S21 between two broadband decoupled (< 70 dB) inductive probes and taking the ratio of the response curve width at the 3 dB points to the resonance frequency. Matching was determined by measuring the reflection on the output cable and was adjusted to be at least -25 dB under loaded (human load) conditions. Active PIN-diode detuning was determined by comparing diode “on”- and “off”-states in an S21 measurement with the above described decoupled probes. After the receiver coils were assembled in the array, geometric overlap decoupling between next neighbors was measured using a direct S21 measurement between pairs of elements using coaxial cables directly connected to the preamplifier sockets of the two elements under test. While all other elements in the array were detuned, the nearest neighbor coil distance was adjusted empirically to optimize the decoupling to less than -12 dB. Preamplifier decoupling of a given loop with all other loops detuned was measured as the change in the decoupled probes S21, when the coil drive coax

was terminated with either the powered low-impedance preamplifier or a 50 Ohm dummy load.

Similarly, the Q-factors of the two orthogonal modes of the asymmetric birdcage were determined with (slightly larger diameter) decoupled probes. With the matching networks in place the return loss (S11) was measured. The isolation between the two driving ports was measured in a direct S21 measurement. A copper shield simulator was used to tune and match the coil on the bench. Final measurements were performed with the coil inside the MRI system bore.

Parameters for Imaging Protocols

The transmit coil homogeneity was assessed by acquiring a flip-angle (FA) map in a healthy volunteer. Human subject rather than phantom imaging was chosen, because this provides a more realistic situation to test the coil transmit homogeneity with regard to loading. To obtain a flip-angle (FA) map, a low resolution 3D-gradient echo acquisition with two repetitions was acquired. The imaging parameters were TR/TE 5.8/2.7 ms, matrix 32×28×12, resolution 11×11×22 mm³. The ratio of two consecutive acquisitions (I₁ and I₂), corrected for the number of RF excitations (N) to the center of k-space, yielded a FA map according to:

$$FA = \arccos \left[\left(\frac{I_2}{I_1} \right)^{1/N} \right] \quad [1]$$

Parallel imaging performance was demonstrated comparing an accelerated 3D steady-state free-precession (TrueFISP) acquisition with a fully sampled acquisition in the same healthy volunteer. Imaging parameters for the R=1 (R=2×3) acquisition were: TR/TE 2.9/1.29 ms (3.06/1.37 ms), matrix 64×48×60, FA 10° (12°), resolution isotropic 6.4 mm, total scan time 8 s (2 s). Accelerated images were reconstructed using GRAPPA [30].

A high resolution spin-density scan in a healthy volunteer (and a subject with COPD, FEV1 = 60% predicted, FEV1/FVC = 0.53) was acquired using a 2D-, multi-slice gradient echo sequence with TR/TE 8.9/2.5 ms (8.5/2.4 ms), matrix 192×144×12 (256×244×12), FA 9.5° (12°), resolution 2.1×2.1×10 mm³ (1.6×1.6×15 mm³), acceleration factor 2 (3), reference lines 24 (24), total scan time 14 s (8 s).

An image set using a spiral pulse sequence was acquired in an asthmatic subject (FEV1 = 64%, FEV1/FVC = 0.56). Imaging parameters for the fully sampled acquisition were TR/TE 11.1/1.0 ms, matrix 128×128, FA 18°, resolution 3.6×3.6×15 mm³, total scan time 1.7 s. For this data set, accelerated imaging was simulated by removing k-space trajectories before image reconstruction. Accelerated images were reconstructed using SPIRiT [31].

Results

Bench Testing

The unloaded Q-factor of a single wire-surface coil, without decoupling circuitry or other elements present, was found to be 400. The Q-ratio, which is the ratio of unloaded to loaded (with a human load) Q-factor, was 5.75. Active PIN diode detuning resulted in better than 30 dB isolation between the “on” and “off” states. The S21 measurement of preamplifier decoupling showed a change of at least 35 dB.

For the asymmetric birdcage without a shield present, the Q-factor of the vertical mode was 220 and for the horizontal mode was 140. When measured inside the scanner and under the

influence of the scanner's RF shield, the Q decreased to 205 for the vertical mode and 130 for the horizontal mode. The Q-ratio of our birdcage was approximately 2 if loaded with a human volunteer. Measurements inside the scanner bore, with the actively detuned array in place and with a human volunteer loading the coil, yielded for the two birdcage resonant modes a return loss (S11) of -20 dB and -12 dB at the HP Xe resonance, while the isolation between the two driving ports was -12 dB. Imaging a phantom using the birdcage with and without the array showed that the presence of the array inside the birdcage decreased its B1-efficiency by approximately 15%.

Imaging results

The HP Xe human chest coil consisting of a 32-channel phased-array receive and the asymmetric birdcage transmit coil is shown in Fig. 2. To assess the performance of the coil regarding the transmit homogeneity, FA maps were acquired (Fig. 4a). The FA distribution appeared homogeneous within each coronal image slice, however a comparison of slices showed an anterior-posterior variation. The mean (standard deviation) of the whole lung flip angle distribution was 5.2° (0.4°). From the most anterior to the most posterior coronal imaging slice, the flip angle varied by 10%, with higher flip angles towards the posterior image slices (Fig. 4b).

The decoupling between individual array elements is assessed by the noise correlation matrix shown in Fig. 5. The data were collected while a volunteer was lying in the coil. The mean and maximum correlation coefficients were 0.25 and 0.73, respectively. The sensitivity of the receive array coil was demonstrated by high resolution spin-density imaging in a healthy volunteer (Fig. 6a). In spin-density images of a subject with COPD, numerous ventilation defects could be clearly identified (arrows in Fig. 6b). Acceleration factors of up to 6 (3×2) were feasible as demonstrated in a 3D-TrueFISP acquisition in a healthy subject (Fig. 7a). The acceleration enabled by the 32-channel receive array reduced the total acquisition time from 8 s to 2 s.

The parallel-imaging performance of the array was further demonstrated in a simulated accelerated spiral acquisition. Accelerated images were reconstructed using SPIRiT [31]. Using only 4 out of 11 spiral interleaves for the image reconstruction the achieved image quality was comparable to the fully sampled reconstruction (Fig. 7b). The reconstruction algorithm incorporated Principal Component Analysis (PCA)-based channel compression [32] before the parallel image reconstruction. By setting a 5% tolerance, the algorithm used 14 (out of 32) virtual receive coils, a relatively high ratio that indicates that the individual channels are fairly independent. Related channel compression techniques have compressed data from 32-channel proton cardiac coil arrays down to 12-16 virtual coils [32] and 4 virtual coils [33].

Discussion

We have shown the implementation of an RF coil dedicated for 3T HP Xe studies consisting of an asymmetric birdcage coil combined with a 32 channel receive array. For the transmit coil, we adapted the asymmetric birdcage coil design previously described by DeZanche et al [23], however we chose not to implement a RF shield. This way, proton imaging, which uses the scanner body coil, will be possible if additional proton traps will be implemented. This has recently been recognized as an important addition to hyperpolarized-gas imaging protocols [34]. Transmit field maps of our asymmetric birdcage showed regional variation (standard deviation /mean) of less than 8 % across the whole lung. To a large part this remaining variation was along the anterior-posterior direction, which could easily be corrected for in quantitative imaging studies. The excitation field homogeneity of the asymmetric birdcage HP Xe transmit coil showed a significant improvement over wrap coils

previously used at our institute. For example, a quadrature transmit/receive wrap coil (Clinical MR Solutions, Brookfield, WI) showed regional variations of 24% in some subjects, and a larger size vest coil of similar design showed 14% variations. The birdcage transmit coil we describe was actually an upgrade from our initially-implemented two-loop Helmholtz design, similar to the one described by Meise et al [22]. Using the Helmholtz-coil the transmit field varied by up to 50%, which was also found by Meise et al. However, especially in functional imaging protocols, which depend on contrast-generating RF pulses, a homogeneous transmit-field profile throughout the lung is crucial. The resulting image contrast depends on both lung physiology and transmit field amplitude, and can only be uniquely attributed to lung function if the transmit field is sufficiently homogeneous and/or inhomogeneities can be sufficiently corrected. Further, due to the non-equilibrium state of the magnetization, an inhomogeneous transmit field would yield a regionally varying magnetization decay, which would prevent optimum SNR parameter settings for the whole lung. The homogeneous transmit field provided by the asymmetric birdcage not only enabled new functional-imaging protocols [9], but also improved image quality of existing applications such as spin-density and diffusion imaging by enabling an efficient use of the available magnetization.

Parallel-imaging performance of the phased-array receive coil was demonstrated by achieving acceleration factors up to three in 2D acquisitions and up to six in 3D acquisitions without visual detectable loss of image quality. The total acquisition time for a whole-lung 3D-scan was reduced from 8 s to 2 s. The reduction in acquisition time is substantial and of great impact, since it allowed the implementation of novel functional-imaging studies, which facilitated multiple whole-lung 3D-acquisitions during a single breath hold. Without the use of parallel imaging, such a protocol would not have been feasible [9]. In a 2D coronal image slice, coil intensity profile variations in two dimensions could be exploited in a spiral trajectory, as opposed to only one dimension in a Cartesian trajectory. In future studies spiral acquisitions, which are already inherently fast, could be combined with parallel imaging for real time spin-density imaging with applications to depict dynamic ventilation distribution or gas trapping.

One limitation found in the current implementation was the low efficiency of the transmit coil. An important factor to determine the efficiency of an RF coil is the filling factor, which is the ratio of the volume of interest (patient body) to the sensitive volume of the RF coil. Because the asymmetric birdcage described in this paper had to be relatively large to accommodate the receive array coil, this resulted in a low filling factor. Previously used transmit/receive coils incorporated a flexible design allowing these coils to be wrapped tightly around the patient body. We found that the transmit voltage required to achieve a certain flip angle, which is used as a measure of efficiency, using the HP Xe asymmetric birdcage was about a factor of three higher compared to a tight wrap coil. These efficiency considerations can be a limitation for pulse sequence design. In most applications short pulses are desired to keep TE short to minimize SNR losses. The relatively low transmit-coil efficiency was a disadvantage of this coil design, while advantages were transmit field homogeneity of the birdcage and parallel imaging capabilities of the receive array. On the Siemens Tim Trio scanner used for these studies the broadband amplifier available for hyperpolarized gas studies provides only 6.5 kW output power compared to the 35 kW available for proton studies. In the future more powerful RF amplifiers could be implemented for hyperpolarized gas studies, which would resolve the efficiency limitation.

Conclusion

In this work we developed a chest coil, which combined an asymmetric birdcage volume coil with a 32-channel receive array for HP Xe studies at 3T. Initial results from transmit-

field mapping and spin-density imaging successfully demonstrated transmit homogeneity, receive sensitivity and parallel imaging capability. The combined chest coil presented in this study will enable the implementation of novel 3D functional-imaging studies, which depend on homogeneous excitation throughout the lung for contrast-generating RF pulses, as well as on speed to encode multiple 3D images within a single breath hold. Our initial results demonstrate that a dedicated RF coil together with recent improvements in hyperpolarization techniques make high quality imaging using HP Xe possible that rivals image quality previously only seen in HP He studies.

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Figure 1.

Design of the asymmetric birdcage: a) cross sectional view; b) overall view. Drawings show the mechanical shape of the asymmetric birdcage with the bottom half of the receive-array in place. Also shown is the horizontal plane split of the birdcage together with the latches to open and close the top half of the birdcage coil for easy subject access. The box inside which the patient's head lies is used to mount cables and also contains the T/R switch.

Figure 2. Human volunteer inside the finished combined chest coil. The coil is placed on the patient table. The top parts of birdcage and array coils can be opened for easy subject access (a). In the closed position the coil is ready to be moved inside the magnet bore (b).

Figure 3. 32-channel phased-array receive coil. The array layout contains 15 elements on the top chest plate and 17 elements on the bottom plate (a). The array elements are placed in critical overlap with each neighbor in a hexagonal tiling pattern. The circuit diagram of the individual element incorporates tuning and matching capacitors, a passive detuning trap circuit, a preamplifier decoupling parallel resonant circuit, and an active detuning bias line with chokes to separate RF from DC lines (b). The top array is pictured at an early manufacturing stage illustrating the element layout with solder pads for components (c). The finished array is shown with preamplifiers directly mounted above surface coil receivers (d).

Figure 4.

Flip-angle (FA) map for the birdcage transmit coil (a). The map shows a homogeneous FA distribution within each coronal image slice, but a slight gradient along the anterior-posterior direction (b). The whole-lung flip-angle mean (standard deviation) is 5.2° (0.4°). The total anterior-posterior difference was $\sim 10\%$, with higher flip angles towards the posterior image slices.

Figure 5. Noise correlation coefficient matrix. The rows and columns correspond to the receiver channel number (only 31 channels are shown since one channel was disconnected at the time of the measurement). The mean (maximum) correlation coefficient of all 31 channels was 0.25 (0.73).

Figure 6.

Ventilation scans from a healthy volunteer using the xenon chest coil (a) and a COPD subject (b). Image acquisition was a 2D FLASH sequence, with resolution $2.1 \times 2.1 \times 10 \text{ mm}^3$ ($3.3 \times 3.3 \times 15 \text{ mm}^3$) and acceleration factor 2 (3) for the healthy volunteer (and the COPD patient). White arrows in b indicate poor ventilation mainly in the right upper lobe for the COPD patient. These images are from subjects shown in Figure 1 of reference [35].

Figure 7. 3D-TrueFISP spin-density acquisition fully sampled (a) and accelerated in 2D by $R=3 \times 2$ (b). The acceleration reduced the total 3D acquisition time from 8 s to 2 s. Also shown is a spiral acquisition of an asthmatic subject with a fully sampled reconstruction (c) and an undersampled reconstruction from simulated parallel imaging using only 4 out of 11 spiral interleaves (d). These images are from the subject with asthma shown in Figure 1 of reference [35].