Chapter 1

Introduction

Magnetic resonance imaging (MRI) of pulmonary ventilation using hyperpolarized noble gases (^{3}He and ^{129}Xe) is a promising new method for assessing and monitoring pulmonary disease [6, 7, 8]. High-quality (high-temporal and high-spatial resolution) MR images of animal and human lung airways and airspaces have been obtained using hyperpolarized ^{3}He, thus enabling identification of chronic pulmonary obstructive disease [9], emphysema [10], asthma [11], cystic fibrosis [12] and apnea [13].

Hyperpolarized ^{3}He gas was first used as a nuclear target in accelerator physics experiments measuring spin composition of neutrons [14, 15]. Soon afterwards the researchers realized the potential of ^{3}He and ^{129}Xe for MR imaging. The non-equilibrium polarization of hyperpolarized noble gases is up to five orders of magnitude larger than the thermal polarization of water [6]. After compensating for the smaller density of gas as compared to that of water one ends up with a tenfold increase in the MR signal. First MR images using polarized noble gas were published in a Nature article in 1994 [16].

The most common method for polarizing noble gases uses a transfer of polarization from an alkali metal (usually Rb) to the noble gas [6]. Rb electrons can be polarized to high values ($\approx 90\%$) using optical pumping of Rb vapor with circularly polarized laser light tuned to the D1 (795 nm) transition in Rb. Polarization of Rb electrons is transferred to the nucleus of the noble gas during collisions between Rb atoms and noble gas atoms [17, 18, 19].

We describe the production of cells containing noble gas and Rb and the optical pumping setup in Chapter 2. Since monitoring the magnetization levels of hyperpolarized gas is important for understanding the physics of hyperpolarized gases, we describe the implementation of NMR (nuclear magnetic resonance) and EPR (electron paramagnetic resonance) polarimetry at Caltech. We present NMR signals of ^{3}He, ^{129}Xe and water, EPR signals of

^{3}He, and preliminary EPR shifts of ^{129}Xe.

An important advantage of hyperpolarized gas MRI over the conventional proton MRI is that the hyperpolarized gas MR signal strength does not depend on the size of the magnetic field used during imaging [20, 21]. Furthermore, if imaging is performed at field strengths at which the sample (body) presents the dominant source of noise, the signal-to-noise ratio of the image is not affected by the field strength [22, 23]. Since low-field systems are easier and cheaper to build, and potentially accessible to a larger sample of population, it may be advantageous to perform hyperpolarized gas imaging at low magnetic field strengths.

Realizing the importance of low-field hyperpolarized gas imaging, we started a collaboration with the group of Dr. Steven Conolly at Stanford University Electrical Engineering Department. Dr. Conolly's group has developed a *pulsed* (or variable) resistive low-field MR scanner for prepolarized MR imaging of water (so-called PMRI). PMRI replaces the static superconductive main field magnet of a conventional MR scanner with two dynamic electromagnets: a polarizing magnet which creates the sample magnetization and thus has to produce a strong but not necessarily homogeneous field, and a readout magnet which needs to produce a homogeneous but not necessarily strong field and which determines the readout frequency [24, 25]. One of the main advantages of the pulsed resistive low-field system is reduction in capital cost. While the superconducting magnets can easily cost $1 million and in addition have high maintenance costs, the two resistive magnets can be built for less than $50,000. This cost reduction could significantly increase the access to MRI and thus enable early detection and regular monitoring of pulmonary disease.

The electronics of the pulsed resistive low-field MR scanner and the pulsed sequence used for PMRI of water are described in Chapter 3. In this chapter we also motivate the construction of a hybrid hyperpolarized gas/prepolarized water MR system by examining the SNR properties of conventional MRI, PMRI and hyperpolarized gas MRI.

While the SNR properties of hyperpolarized gas and prepolarized water are similar, there are also essential differences between the two imaging techniques. In particular, two properties of hyperpolarized gas distinguish hyperpolarized MRI sharply from proton MRI: the nonrenewable nature of the gas polarization and the substantially larger diffusion constant of gases as compared to water (^{3}He, for instance, has five orders of magnitude larger diffusion constant than water) [6]. The nonrenewable polarization, coupled with the long T_2 relaxation times of gases, motivate the use of single-shot sequences, such as

RARE [26] and trueFISP [27], which utilize all the available gas magnetization and can thus produce higher image SNR than small flip-angle sequences, such as FLASH [9, 28]. The large diffusion constant of gases causes rapid signal decay, which, however, can be minimized by proper sequence design.

In Chapter 4 we study, in detail, the T_2 relaxation and diffusion processes of hyperpolarized gases. We make a distinction between the reversible and nonreversible T_2 decay, and further divide the nonreversible decay into diffusion losses in the magnetic field gradients and the decay due to spin-spin interactions. The first half of the chapter gives the theoretical background for all these processes, while the second half presents our experimental results. We use Free-Induction-Decay (FID) signals of hyperpolarized ^{3}He, ^{129}Xe and water to compute the polarization of hyperpolarized gas. Furthermore, we collect spin echo trains using a CPMG sequence [29], which also serves as the basis for measurements of diffusion coefficients and the inherent T_2 relaxation times. In the Appendix A we estimate diffusion coefficients of binary gas mixtures using Lennard-Jones potentials [30].

In Chapter 5 we use the experimental values from Chapter 4 to develop a numerical model of signal decay during gradient echo sequence. We divide the effects which decrease the size of hyperpolarized gas signal into three groups: the effect of the excitation flip-angle; T_1 and T_2 relaxation losses; and diffusion losses. The simulation helps us to obtain a gradient echo image of a 1-inch spherical cell filled with hyperpolarized ^{3}He. In addition, we study, through modelling and experiments, the SNR gain in 1-D spin echo projection images when using centrally ordered phase-encode gradients. Our results show promise for 2-D RARE sequences with central ordering of encoding gradients.