MRI of hyperpolarized 3He at 3T

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Introduction The optimal B_0 field strength for hyperpolarised (HP) MRI experiments is the subject of some discussion [1-3]. The non-Boltzmann polarisation (achieved here with laser optical pumping) makes the magnetisation independent of B_0, and low-field MRI becomes a realistic possibility [2]. Furthermore the field inhomogeneity at high B_0 in the lungs makes the use of higher fields for lung imaging more challenging. Nevertheless, MRI manufacturers are shifting their multinuclear engineering focus on to higher B_0 systems and engineering quality as well as electromagnetic physics will ultimately determine the SNR achieved in practice. The objective of this work was to investigate the feasibility of HP ^{3}He MRI at 3T on a whole body system. The engineering aspects of upgrading our 3T system for ^{3}He transmit and receive are presented. Preliminary imaging results and SNR comparisons are made with data acquired at 1.5T with non-lossy ^{3}He samples of known polarisation and coils of the same size and geometry.

Methods A whole body 3T system (Philips, Achieva) was modified for transmit-receive at the ^{3}He frequency of 97 MHz with a 4kW RF amplifier (CPC) and second receive chain. An active T-R switch was built (Fig. 1) that allows transmit on one channel with receive on 2 channels.

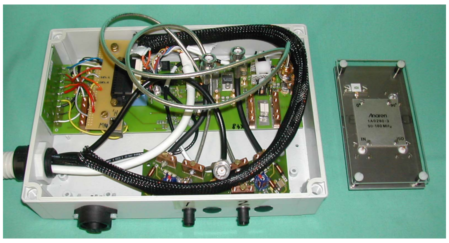

Fig. 1 Interior of T-R switch built for ^{3}He at 97 MHz

Fig. 2 High-pass ^{3}He birdcage coil (97 MHz)

RF coils: Phantom experiments were conducted with a purpose-built high-pass birdcage coil operated in linear mode –Fig.2. The circular coil has 12 rungs of length 22 cm, shield radius R=11cm, and coil former radius r = 7.5cm, the coil is shielded to preclude interaction with the ^{1}H body coil –Fig. 2. High voltage high Q ceramic capacitors (56pF AVX SQCA/CB ±5% tolerance and 500V acceptance) were used on the end ring segments. The Q of the coil was measured inside the magnet using a network analyser : Q= 110 at 97.175 MHz. The coil's filling factor η was measured on the bench using the frequency shift method of Doty [4], whereby a 2 cm aluminium sphere was placed at isocentre causing a frequency shift δf_0=0.1059MHz from the centre frequency f_0=97.175MHz, this gives η = 13% for a sample filling the coil's 3.5 litre inner volume and η_{sample}=3.7% for the 1 litre bag phantom. Comparisons were made with data acquired at 1.5T (^{3}He ω_0 48.6 MHz) with a low-pass birdcage coil [5] of the same radius, and rung topology. This coil was used in quadrature mode and it's Q was measured as 250 inside the magnet. The difference in Q is attributed to the Q of the ATC capacitors used for the 48MHz coil. Filling factor was evaluated as before, giving η_{sample}= 5%. The difference in the measured filling factors of the two coils is attributed to the slightly shorter leg length (19 cm) of the 1.5T coil. **^{3}He** gas was polarised to 24% with spin exchange apparatus (GE). The imaging phantoms for the static imaging experiments consisted of 1 litre Tedlar plastic bags filled with 100 ml ^{3}He and 900 ml, N_2 resulting in a longitudinal magnetisation density ~ 5.4×10^{-5} J T^{-1}m^{-3}. Preliminary NMR tests were done using a 1° pulse –acquire to centre the spectrometer f_0 and to optimize the receiver gain settings. After preliminary imaging experiments it was found that limited preamplifier gain was needed due to the high coil sensitivity and the highly polarised sample. SNR measurements were performed using MRI rather than pulse-acquire spectroscopy as previous observations of the effects of radiation damping in the FIDs of highly polarised ^{3}He samples [5], were found to complicate analysis of the measured signal. MRI tests were performed on the phantoms using an optimised 2D SPGR sequence. High resolution 2D imaging was performed with a 10 mm slice with 140 sequential phase encodes x 140 read samples over a 140 mm FOV (pixel = 1 x 1 x 10 mm) using a flip angle of 6°. A lower resolution SPGR image was also acquired (112 phase encodes, 10 mm slice, 8° flip angle, BW 50 kHz) for direct comparison with data acquired previously on a 1.5T system using similar ^{3}He phantoms with the quadrature low pass birdcage coil.

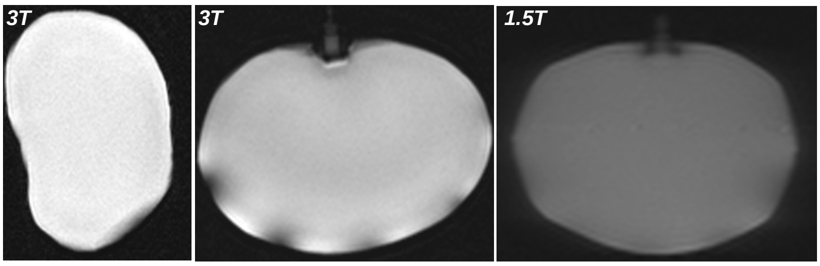

Fig. 2 2D axial SPGR high resolution images from a coronal (left) and axial (centre) slice at 3T. The axial image (1mm x1mm in plane) has been windowed to show the focal hyper intensity seen at the base where the bag phantom is very close (<5 mm) to the birdcage rungs. The remainder of the central region shows very high image uniformity indicative of the high B1$^+$ field homogeneity of the birdcage design, an important factor for HP gas MRI with its inherent sensitivity to flip angle. These images are of higher SNR and clarity than those previously obtained with bandwidth matched sequences on our 1.5T system (right) with a coil of equal or better sensitivity.

Results and Discussion When quantitative SNR comparisons were made from the lower resolution images: at 1.5T the SNR was 110, whilst at 3T SNR was 469. These gas samples (of the same polarisation) are lossless and coil noise dominates. If the relative preamplifier noise and differences due to T2* in the ^{3}He phantom are set aside (these factors are system and B_0 dependent), then theoretical SNR comparisons can be simplified to a comparison of the relation **SNR $\propto(\omega_0\eta Q)^{1/2}$** [1,6]. Thus we might expect the relative SNR from our experimental comparison to be given by $SNR_{3T}/SNR_{1.5T} = (97.3\times0.037\times110)^{1/2} / (48.6\times0.05\times250\times2)^{1/2} =$ 0.68. The additional factor of $\sqrt{2}$ in the 1.5T term numerator accounts for the fact the coil is operated in quadrature. For two coils of identical η and Q the ratio of improvement at 3T would be expected to be $\sqrt{2}$. The experimental observation of ~ 4 x SNR at 3T implies that either : the preamplifier noise is lower on the 3T system and/or that any interference due background noise from equipment present in the scanner or equipment room (intermittent noise lines experienced previously at 1.5T) is less of a problem at 3T. A more detaield analysis of SNR will require measurement of the preamplifier noise figure for both systems. In conclusion, dedicated coil and transmit receive hardware has successfully enabled HP ^{3}He MRI on a 3T clinical whole-body system and the preliminary results indicate an encouraging outlook for the in-vivo studies planned.

References: [1] Med. Phys. 2005 32 (1), 221-9 [2] Proc. ISMRM 2003, p. 1387. [3] Magn Reson Med. 2005;53(1):212-6. [4] J Magn Reson 138 (1999)144–154. [5] J Magn Reson 185 (2007) 94–102 [6] J Magn Reson 24 (1976), 71-85.

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