Optimization of Multi-turn Litz Wire Radiofrequency Coils for Hyperpolarized Noble Gas Imaging of Rodent Lungs at 73.5mT

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Introduction

Hyperpolarized noble gases (HNG), ^{3}He and ^{129}Xe, have become a promising tool for lung MR imaging. With hyperpolarization, the available magnetization is independent of the magnetic field strength making imaging feasible at field strengths considerably lower than those used clinically. For small animal imaging, coil noise dominates over a large range of frequencies and SNR improvements are expected by reducing RF coil noise by minimizing electrical resistance especially at low frequencies [1, 2]. It has been shown that multi-strand conductors at room temperature (eg. Litz wire) can be used to reduce resistance compared to conventional (ie. solid) copper conductors up to 10MHz [2]. SNR improvements have previously been demonstrated for HNG imaging of rodent lungs with the use of Litz wire compared to solid copper [3]. More specifically, Litz wire type 2 with AWG 18 (660 strands of AWG 46) has been found to introduce negligible proximity effects between windings suggesting that further improvements in performance would be possible by taking advantage of the large number of turns possible at such low frequencies [3].

The objective of this work was to further improve RF coil performance by increasing the number of turns of Litz wire (type 2 with AWG 18; 660 strands of AWG 46) from a single turn up to 48 turns. Quality factors (Q) and image SNR measurements obtained from ^{1}H phantoms for each coil were compared at 0.866 and 2.385 MHz corresponding to ^{129}Xe and ^{3}He Larmor frequencies at 73.5 mT. In vivo ^{129}Xe and ^{3}He rat lung images at 73.5 mT were also obtained.

Methods

Experiments were performed using a broad-band, variable field (0.01 – 0.15 T) MR imaging system for rodents [4]. Hyperpolarized xenon gas was produced by spin exchange optical pumping using a home-built, continuous-flow polarization system delivering 5mL/s gas mixture of 1 % xenon gas (enriched to 99% ^{129}Xe) with polarizations up to 15%. The polarizer used a 60W diode array laser (λ=794.8 nm, Coherent, Santa Clara, USA). Hyperpolarized ^{3}He was produced by a turn-key polarizer (Helispin®, GEHC) providing polarizations up to 40 %.

The imaging system used a Transmit Only/Receive Only (TORO) coil configuration. For this work the same transmit coil and similar receive coil geometries were used. Receiver coils were constructed on identical plastic formers with an inner diameter of 3.5". Coils with 1, 2, 10, 20, 34 and 48 turns were built using Litz wire AWG 18 type 2 (660 strands of AWG 46) with windings side-by-side (i.e. no winding spacing). No spacing between windings was previously shown [3] to be the optimal when using this type of Litz wire. All coils were separately tuned using appropriate choices of tuning capacitance to 866 kHz or 2.385 MHz corresponding to ^{129}Xe and ^{3}He Larmor frequencies at 73.5 mT. The quality factor (Q) of each coil was measured using a vector impedance analyzer (Bravo MRI II, AEA Wireless, Inc., Vista CA, USA). The SNR was estimated from ^{1}H gradient echo images (TR/TE = 30/7 ms 64 x 64 pixels and 10cm FOV) acquired with a 6cm diameter sphere, filled with saline solution doped with gadolinium (Gadoteridol 0.5mg/mL, Squibb Diagnostics, Montreal, Canada), by changing the magnetic field strength to 20.35 mT and 56 mT respectively. The SNR values were estimated using the mean value of the image signal divided by the standard deviation of the noise in the background. *In-vivo* ^{129}Xe and ^{3}He images were also obtained from the 48-turn and 38-turn Litz coils respectively in Sprague Dawley rats ventilated with a custom ventilation system using a University-approved animal care protocol.

Results and Discussion

Table 1 shows the Q values and SNR estimated for each coil at 0.866 and 2.385 MHz. Due to the very small inductance of the 1- and 2-turn coils, tuning and matching at 0.866 MHz wasn't possible. Higher Q values were obtained for the coils at 0.866 MHz because the size of the Litz strands was optimized for this frequency. The SNR was higher at 2.385 MHz due to the higher field strength. As expected, the SNR increased as the square-root of Q and number of turns for all cases. The maximum SNR at 2.385 MHz was obtained for the 34-turn coil and decreased thereafter as shown in Fig.1. For the 0.866 MHz case, the maximum SNR was obtained for the 48-turn coil (Fig 2), even though at this point the Q factor was lower for this coil than for the 34-turn coil. This decrease in Q for the 48-turn coil at 0.866 MHz, suggests higher numbers of turns would probably result in a decrease in the SNR, as for the 2.385 MHz case shown in Fig.1. Images of ^{129}Xe and ^{3}He of *in vivo* rat lungs at 0.866 and 2.385 MHz respectively are shown in Fig. 3, using the best coil at each frequency based on Figs. 1 and 2 (i.e. 48-turn coil for ^{129}Xe and 34-turn coil for ^{3}He).

	866 kHz		2.385 MHz	
Coil	**Q**	**SNR**	**Q**	**SNR**
1 turn	-	-	78	21
2 turns	-	-	170	44
10 turns	215	33	196	116
20 turns	350	61	282	195
34 turns	377	72	211	230
48 turns	372	87	125	187

Table 1. SNR and Q factor measured in water phantoms for each coil at the frequencies of interest.

Conclusions

Construction of low field RF coils for hyperpolarized gas imaging of rodents lung benefits from the use of Litz wire, particularly with increased number of turns where an SNR improvement of approximately 300% is possible compared to copper wire and nine times compared to single turn Litz coils. In future, the use of multi-turn, multi-layered Litz coils will be investigated.

Acknowledgements: This work was supported in part by NSERC and CIHR. Thanks to Ryan Kraavanger for assistance with animal care.

References

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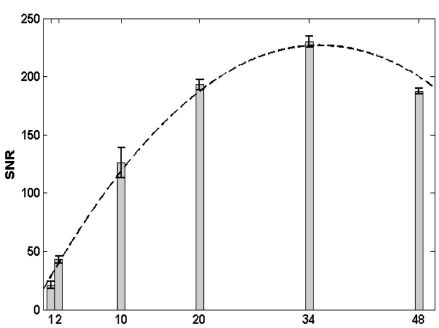

Fig. 1: SNR vs number of turns for coils at 2.385 MHz.

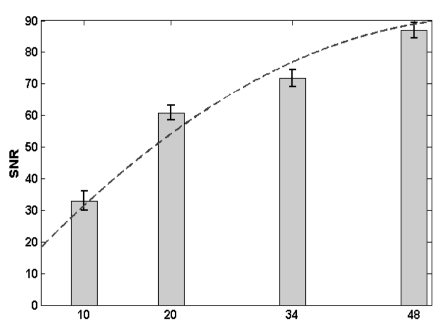

Fig. 2: SNR vs number of turns for coils at 0.866 MHz.

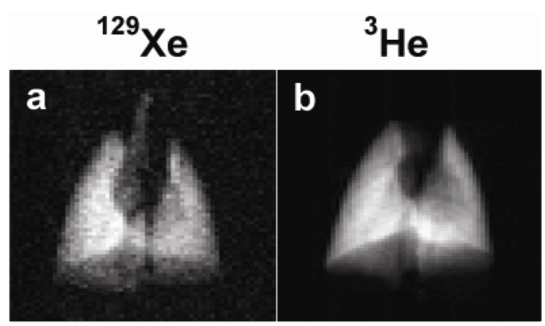

Fig. 3: *In-vivo* rat lung images for ^{129}Xe and ^{3}He. (a) ^{129}Xe image using 48 turns coil at 0.866 MHz. (b) ^{3}He image using 34 turns coil at 2.385 MHz.