A 128-channel Helium-3 Phased Array at 3T for Highly Accelerated Parallel Imaging in Hyperpolarized Gas MRI

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INTRODUCTION Tremendous signal-to-Noise (SNR) gain can be achieved by using hyperpolarized He-3 (HH) gas for imaging human lung in MRI. Meanwhile, due to the different behavior between non-equilibrium state of HH magnetization and equilibrium state of proton magnetization during a RF pulse train, parallel imaging could have additional SNR advantage in HH imaging over proton imaging, when flip angle is optimal [1,2]. The goal of this work is: first, to develop a 128-ch phased array coil system for in vivo HH lung MRI; second, to explore that at which acceleration factor in parallel imaging, the SNR benefits from both hyper-polarization and under-sampling in non-equilibrium state are neutralized by the rapidly increasing g-factor.

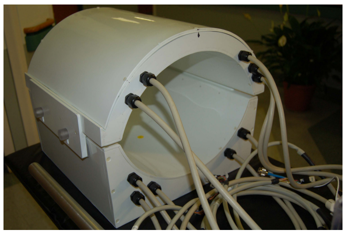

Figure 1

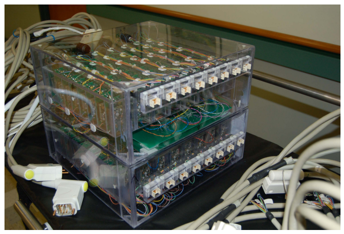

Figure 2

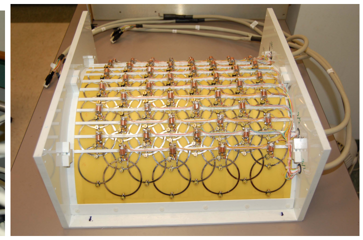

Figure 3

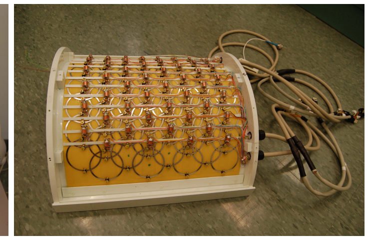

Figure 4

METHOD A 128-ch 3T He-3 coil system includes coil unit (Fig. 1) and preamplifier unit (Fig. 2). The coil unit includes two shells: bottom (Fig. 3) and top shell (Fig. 4). Each shell has 64 receive loop coils and one large spare loop as transmit coil. The design of the receive array is based on the principle in Ref. [3] and the practical concerns in Ref. [4, 5]. The inner and outer diameter of the receive loop are 68mm and 74mm. Any center-to-center distance between two adjacent loops is 53.25mm. Each loop was tuned to 93.9MHz and matched with three capacitors (33pF and 150pF at coil port, 27pF at the other end of loop) and two trimmers. To quantitatively estimate the coil loss caused by both large number of tuned loops and preamplifier decoupling, we have measured the Q-factor of a loop in two different scenarios: one is when the loop is the only resonant circuit (Fig. 5), another is when all the loops are tuned to 93.9MHz, the measured loop is the only one that is not detuned by external power sources (Fig. 6). The Q-factor is dropped from 184 to 110, which suggests that coil loss increases 67%. Note that this number varies when the tested loop is in different location, outer loops have less loss than the middle ones due to they experience less coupling. We also notice that when the preamplifier decoupling mechanism is removed, the loop is strongly coupled with its near neighbors (Fig. 7), which is a different situation from the one in Ref. [3]. The transmit coil is a 38x35cm square loop which is 2cm above receive array. It is tuned and matched by two 10pF, two 22pF, one 3.3pF and one trimmer. All 128 preamplifers have gain 26dB, noise figure <0.5dB, and input impedance <2Ω. To avoid the preamplifier oscillation, we kept all preamplifiers away from the coil array (Fig. 2), which also causes some appreciable additional loss at 93.9MHz.

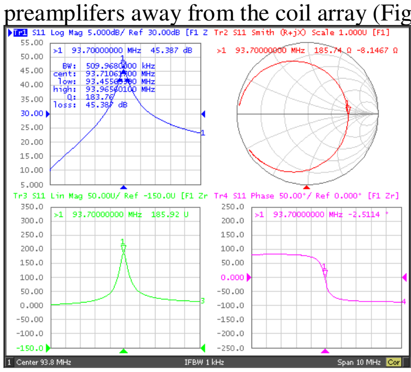

Fig. 5

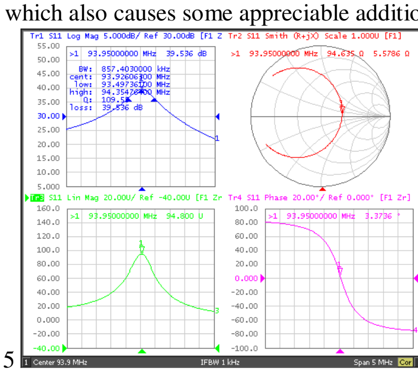

Fig. 6

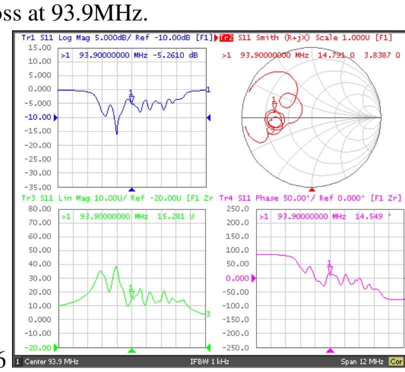

Fig. 7

RESULTS The coil system was tested on a modified Siemens 32-ch TIM Trio MRI scanner which has additional 96 receive channels installed [6]. Five volunteers were scanned when they inhaled HH gas and held breath. HH gas was polarized by spin exchange with an optically pumped rubidium vapor to the level of 35-45% using GE Healthcare helium polarizer. Helium diluted with N_2 to a net polarization level of 10% was transferred to 1 liter Tedlar plastic bags and delivered to healthy human subject. The 2D gradient recalled echo sequence was used for this study, TR 118ms, TE 2.2ms, resolution 128x128, slice thickness 1cm, FOV 40cm. Fig. 8 (a-d) are results of iPAT=1 and iPAT=4 for volunteer two. Here scan time is 15s and 6s, reference line 24, no SNR difference is observed. Fig. 8 (e, f) are results of iPAT=8 for volunteer five, here scan time is 4s, reference line 16. This is the first evidence that the image integrity can be well preserved at such high acceleration factor.

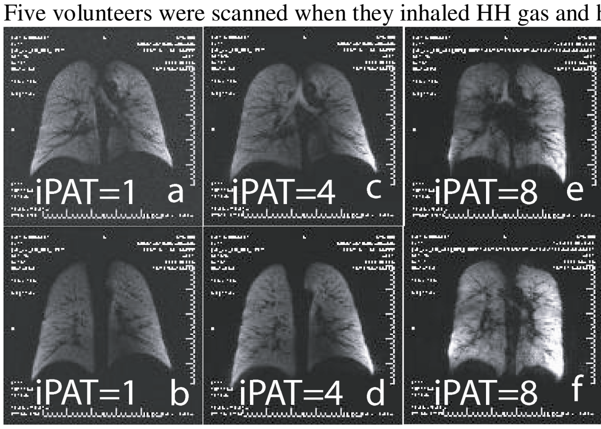

Figure 8

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