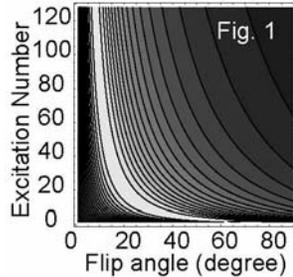


# Applying parallel imaging in hyperpolarized <sup>3</sup>He in vivo MRI to achieve high acceleration factor without the conventional SNR penalty

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**INTRODUCTION** Hyperpolarized (HP) <sup>3</sup>He gas MRI has the potential to assess pulmonary function (1). The non-equilibrium condition of the HP <sup>3</sup>He causes continual depletion of the signal level over the course of excitations, which results in a SNR that can be independent of the number of excitations under certain conditions. This suggests that parallel MRI (2, 3) can enhance both temporal and spatial resolution without the conventional SNR penalty (4, 5). We describe a relationship between the SNR and the number of excitations for any flip angle. Additionally, using parallel imaging with HP gas MRI, we quantitatively demonstrate the SNR by the point-spread-function (PSF). A 24ch phased array was constructed and used to perform parallel MRI. Our *in vivo* experimental results prove that significant temporal resolution can be gained without conventional loss in the SNR.



**METHOD** In the case of the gradient-recalled echo pulse sequence, the *k*<sup>th</sup> excitation is

$$s(k) = M_0 e^{-(k-1)TR/T1} \cos^{k-1} \alpha \sin \alpha, \quad [1]$$

Here, *M*<sub>0</sub> is the initial longitudinal magnetization, *k* is the excitation index, and  $\alpha$  is the flip angle. T1 is the spin-lattice relaxation and TR is the repetition time. Since TR << T1, Equation [1] can be simplified to

$$s(k) \propto \cos^{k-1} \alpha \sin \alpha, \text{ or } S(x) \propto \sin \alpha \left( \frac{1 - \cos^N \alpha e^{-j2\pi x}}{1 - \cos \alpha e^{-j2\pi x/N}} \right), \quad [2]$$

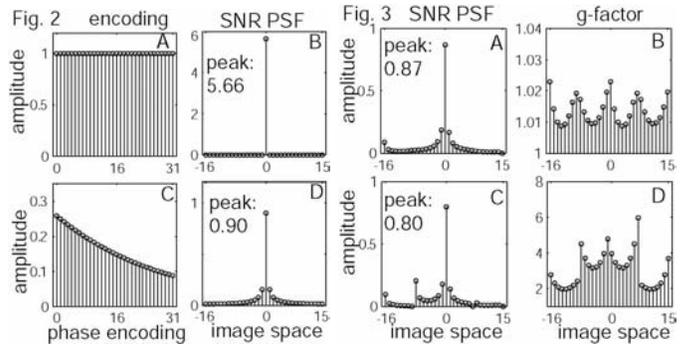
in either *k*-space or image domain *x*. Here, *N* is the total number of excitations, *x* = -*N*/2, -*N*/2+1, ..., *N*/2-1 is the spatial pixel index. If the peak of the S(*x*) is used to estimate the signal, and noise is known to be proportional

to  $\sqrt{N}$ , then the SNR point-spread-function (PSF) is

$$SNR \sim S(0)/\sqrt{N} \propto [\sin \alpha (1 - \cos^N \alpha)/(1 - \cos \alpha)]/\sqrt{N}. \quad [3]$$

Fig. 1 is the contour plot that illustrates the relation between SNR, flip angle, and the number of excitations based on Eq. [3]. The grayscale indicates the relative SNR where the brightest area is the highest SNR that can be achieved by optimizing flip angle and the number of excitations, which suggests that SNR is independent of the excitation number for the optimal flip angle.

Comparisons of the phase encoding and their PSF between the equilibrium (A,B) and non-equilibrium states (C,D) are given in Fig. 2, here the sampling points are 32. Fig. 2C indicates that unlike the case of an equilibrium system, reducing the excitation number in a non-equilibrium system decreases the resolution but not the SNR. Fortunately, parallel imaging can recover the lost resolution caused by reducing the phase encoding steps using phased array coils. Fig. 3 shows the PSF and g-factor of parallel imaging with a reduction factor of two and four. Here, four-element loop array and the weak reconstruction of the SENSE method (3) are used for data acquisition and reconstruction. Note that peaks of SNR PSF in Fig. 2 (D) and Fig. 3 (A) and (C) vary only slightly, not following the  $\sqrt{N}$  rule of an equilibrium system.

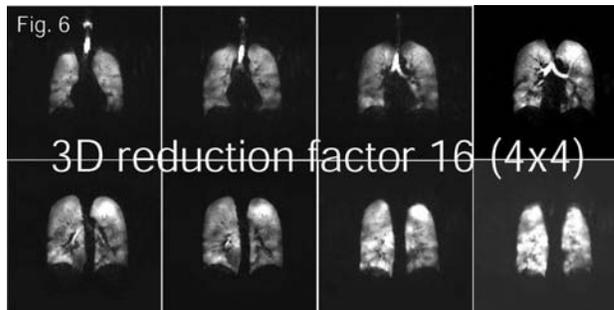
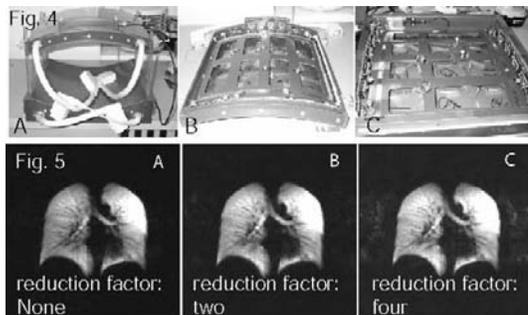


**EXPERIMENTS** To perform the parallel hyperpolarized gas MRI at 1.5T Magnetom Avanto (Siemens, Erlangen, Germany), we built both transmit and receive coils that are tuned to 48.5MHz. A 2-ch transmit and 24-ch receive arrays are housed in the two rigid shells shown in Fig. 4 (A). The insides of both top and bottom shells are shown in Fig 4 (B) and (C). There are two levels inside each shell. The receive arrays are in the inner levels and the transmit coils are in the outer levels of the shells.

A comparison of the parallel imaging with different reduction factors is shown in Fig. 5 where (A) is the full phase encoded image, and (B) and (C) show parallel imaging with a reduction factor of two and four. The pulse sequence used is a 2D gradient echo with TR=79ms TE=2.55ms, slice thickness 10mm, resolution 128x128, number of average 1, and flip angle 10°. Total ten ROI were measured for each image; the average values are (A) 104.3, (B) 113.3, and (C) 96.0, which are basically the same given the unavoidable small uncertainty of gas polarization and administration.

Using a 3D gradient echo sequence, we achieved a reduction factor of 16 (4x4), which allows us to scan the entire lung in 13s and one breath-hold. Some selected images from the 3D data set are shown in Fig. 6, with a TR 15.2ms, TE 1.26ms, slice thickness 5mm, resolution 128x128, 32 slice per slab, 50% over-sampling in slab selection direction, and 30% polarization.

**CONCLUSIONS** Parallel imaging is used in HP <sup>3</sup>He *in vivo* MRI to gain both temporal and spatial resolution without conventional SNR penalty. Using a 24ch phased array and parallel <sup>3</sup>He MRI with a reduction factor of 16 (4x4), an entire lung can be imaged in 13ms and one breath-hold.



## REFERENCES:

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