# Optimization of Gradient-Echo Sequences for Hyperpolarized Noble Gas MRI

JP Mugler III

*Department of Radiology, University of Virginia Health Sciences Center, Charlottesville, VA, USA*

## INTRODUCTION

Conventional gradient-echo pulse sequences have been demonstrated to be a practical and robust method for imaging the lung with hyperpolarized $^3He$ or $^{129}Xe$ [1-4]. Nonetheless, due to the unique characteristics of the hyperpolarized magnetization, the image properties as a function of the pulse sequence parameters are not directly analogous to those in proton MRI. Hence, the optimum pulse configurations for noble gas imaging need to be specifically determined. In the following, we explore the relationship between the signal-to-noise ratio (SNR), flip angle, and number of phase-encoding steps for hyperpolarized gas MR imaging using a conventional (spoiled) gradient-echo pulse sequence with sequential phase-encoding and constant flip angles.

## THEORY

The image SNR can be written:

$$\mathrm{SNR} \propto \sqrt{T_S N}\, \cos^{N/2}\theta \, \sin\theta \, e^{-(N/2)TR/T1} \qquad [1]$$

where $T_S$ is the data sampling period, N is the total number of phase-encoding steps (2D or 3D), $\theta$ is the flip angle, TR is the repetition time, and T1 is the time constant for the hyperpolarized magnetization to relax to thermal equilibrium. (The change in signal level with phase-encoding step imposes a *k*-space filter. The form and effects of this filter will not be discussed here, although we note that a variable-flip-angle acquisition may be used to control the form of the signal evolution [5].) It is assumed that the thermal polarization is negligible compared to that in the hyperpolarized state. Note that the dependence of the SNR on $\theta$ and TR is quite different than in the proton case, and that the T1 value is long (>20 s in the human lung [6]). T1 depends, among other factors, on the oxygen concentration.

The optimum value of $\theta$ involves a trade-off between depletion of the (non-renewable) longitudinal magnetization and generation of transverse magnetization. This value is found by solving $\partial(\mathrm{SNR})/\partial\theta = 0$, which yields:

$$\theta = \tan^{-1}\sqrt{\frac{2}{N}} \qquad [2]$$

Replacing $\theta$ in Eq. [1] with the value given in Eq. [2], and ignoring T1 decay for the moment, since T1 is relatively long, we find the very interesting result plotted in Fig. 1. We see that the SNR very quickly approaches an asymptotic value, such that for practical purposes the SNR (at the optimum value of $\theta$) is independent of the number of phase-encoding steps. Although naively it may seem that the continued depletion of the hyperpolarized longitudinal magnetization would result in decreasing SNR with increasing N, the image noise (averaging) term ($\sqrt{T_S N}$) exactly balances the flip angle term. (Of course for very large N, T1 decay will attenuate the SNR.) Also note that this behavior is contrary to that for proton MRI, wherein for fixed TR the SNR would increase as $\sqrt{N}$. The optimum $\theta$ values corresponding to Fig. 1 are plotted in Fig. 2.

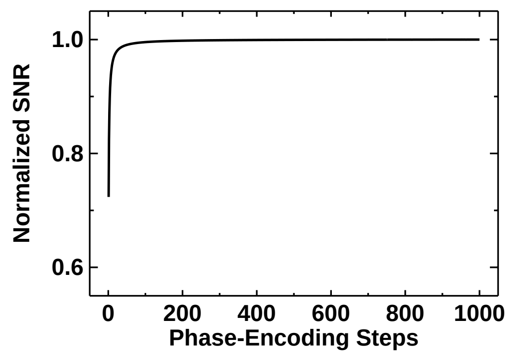


**FIG 1:** SNR versus the number of phase-encoding steps for a spoiled GRE pulse sequence using sequential phase-encoding and the optimum flip angle value in Eq. [2].

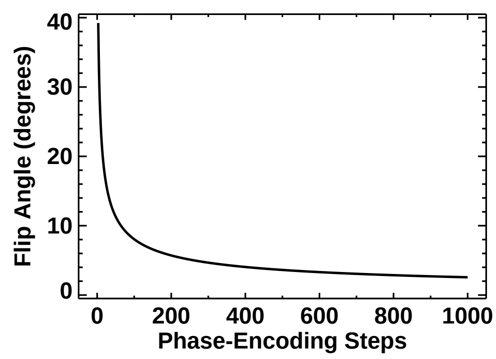


**FIG 2:** The optimum flip angle from Eq. [2] versus the number of phase-encoding steps.

## CONCLUSIONS

For hyperpolarized noble gas MRI with conventional sequentially-encoded gradient-echo pulse sequences, we have derived a general expression for the flip angle value that provides maximum image SNR. This optimum flip angle depends only on the number of phase-encoding steps (Eq. [2]). The unique characteristics of the hyperpolarized magnetization lead to the interesting result that the image SNR at optimum flip angle is independent of the number of phase-encoding steps, aside from a secondary dependence on the usually long T1 value. The SNR advantage typically associated with 3D acquisitions, as found in proton MRI, therefore does not apply to hyperpolarized gas imaging.

## REFERENCES

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