Optimisation of velocity encoding gradients for phase contrast gas velocity taking diffusion into account.

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Introduction
Phase contrast MRI which is commonly used to map blood [1] has been applied more recently to map gas velocities using hyperpolarized tracers[2, 3]. The bipolar gradient used to induce a phase shift proportional to velocity also introduces diffusion weighting. This can be safely neglected considering diffusion coefficients of liquids. However, for a gas, diffusion can no longer be neglected [3]. As an illustration on helium, considering an encoding time of 10 ms for spins moving at 20 cm.s⁻¹, the displacement (2 mm) becomes the same as the mean diffusion length. Blurring is introduced and a loss of precision in the phase measurement is obtained. While the precision on the velocity measurement can be enhanced by increasing the gradient 1st moment, there is a competition with the loss introduced by diffusion weighting. Here, considering both phenomena, we derive theoretically and verify experimentally the bipolar gradient characteristics for an optimized velocity measurement.

Theory
After a bipolar gradient, velocity error is given by: \( \sigma = \frac{\sigma}{2\pi FOS} \) [4] where \( \sigma \) is the noise standard deviation, \( I_0 \) the signal intensity, FOS is the field of speed characterizing the velocity encoding and \( A \) the attenuation induced by diffusion within the bipolar gradient[5]. For a fixed TE, we can determine velocity error as a function of the FOS. The optimal FOS is thus:

\[
FOS_{opt} = \frac{\sqrt{2D}}{\pi}
\]

and depends on the length of the bipolar gradient (T) and the diffusion coefficient (D) of the buffer gas [6]. Figure 2, presents these theoretical curves.

Material and methods
Data were acquired at 1.5 T. \(^3\)He was hyperpolarized on site and mixed within different buffer gases to change the diffusion coefficient in a controlled way. Three vector gases were used, \(^4\)He, \(N_2\) and \(SF_6\). The mixture flowed through a 3 m long straight tube (Ø = 34 mm) at a constant rate (controlled by a volumetric pump) with a mean velocity of 20 cm.s⁻¹ ensuring a laminar and parabolic flow. The 2D gradient echo sequence was flow-encoded through plane with the following parameters, FOV = 50*60*50 mm, pixel = 2.5*2.5 mm, \( \alpha = 20^\circ \) and TR/TE = 16/6.0 ms. The sequence was repeated 6 times varying the bipolar gradients amplitudes with a constant time corresponding to FOS from 50 cm.s⁻¹ to 200 cm.s⁻¹. Data were reconstructed and processed using Matlab®. A paraboloid was fitted to each velocity map using the experimental mean velocity (\( V_m \)). The SD of velocity (\( \sigma_o \)) was calculated on the central 85% of the tube section from the difference between the experimental values and the model values.

Results
The theoretical model, the measured values and the differences are shown in Figure 1 for \(^4\)He with FOS=200 cm.s⁻¹. The bias between theoretical \( V_m \) and the measured \( V_m \) were: for \(^4\)He, \( \Delta V_m = 10 \) cm.s⁻¹; for \( N_2 \), \( \Delta V_m = 0.5 \) cm.s⁻¹; for \( SF_6 \), \( \Delta V_m = 7.4 \) cm.s⁻¹. Figure 2 shows, for each gas, the theoretical curve of velocity error (\( \sigma_o \)) as a function of FOS with the theoretical optimal point and the 6 experimental points (each experimental value was corrected for the amount of polarized gas that was used).

Discussion
The observed biases between flows measured by MR and by external flowmeter may be due to the difficulty of measuring flows with various gas mixtures. This does not hinder the study, which focuses on velocity measurement precision. Figure 1 shows that we can measure and reconstruct a velocity map very close to a theoretical paraboloid. In Figure 2, big FOS imply an important velocity error because the velocity encoding is less efficient and reducing the FOS corrects it. It would be the same with blood. However, in our case, in spite of continuing to diminish while the FOS diminishes, this effect is due to the diffusion weight growing with little FOS. These results validate our theoretical computations and present experimental minima not so far from theoretical ones. For better results, these experiments should be continued with more experimental points.

Conclusion
We have derived a theoretical expression to adjust the FOS and obtain an optimal velocity measurement for phase contrast MRI on a gas. This optimal value is proportional to the square root of the ratio of the diffusion coefficient to the gradient application time. While traditional phase contrast on liquids will decrease the FOS to increase the precision with virtually no limit, theoretical and experimental data show here that for gas, diffusion becomes rapidly dominant and the corresponding signal attenuation then limits the expected velocity precision. Even though preliminary, the presented results are readily usable to define the optimal bipolar gradient parameters for velocity measurement phase contrast on hyperpolarized gas.