Compressed Sensing: Applications in Hyperpolarised 3He lung MRI

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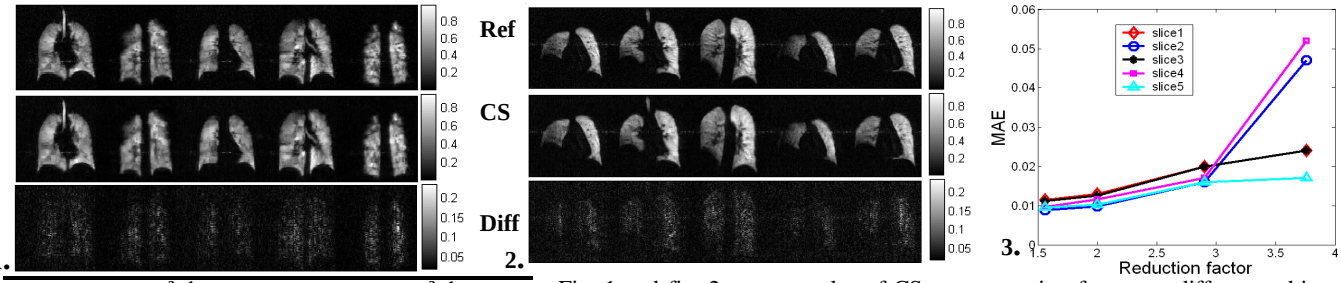
Introduction The polarisation in hyperpolarised (HP) ^{3}He MRI is non-renewable and the scan time of both breath-hold and dynamic lung ventilation images is by necessity short. There is thus a demand to reduce the number of RF excitations needed to fully reconstruct an image and to speed up the acquisition. In previous work, parallel RF encoding [1], radial [2] and spiral [3] under-sampling methods have been used in HP gas MRI to accelerate the acquisition. These methods can be demanding on the hardware in that they require multiple receivers and implementation of robust non-Cartesian sequences. There has been recent interest in Compressed Sensing (CS) techniques in MRI [4]. The idea behind the theory is reconstruction of a sub-set of linear measurements that are smaller than the actual full data set, using non-linear methods. In this work CS methods were applied to reconstruction of Cartesian encoded HP ^{3}He gas images (ventilation and ADC) to investigate the potential advantages and limitations of the method. Simulations were performed in 2D and, under-sampled images are presented with preservation of qualitative and quantitative image information.

Materials and Methods The study was conducted with images collected from a female volunteer with mild emphysema (age 56) and four non-smoking healthy male volunteers (age varying between 24 and 34). Approval from the local research and ethics committee were obtained. ^{3}He gas was polarised on site to 26% by optical pumping with Rb spin exchange apparatus (GE). 300 ml ^{3}He gas mixed with 700 ml N_2 was inhaled from a Tedlar bag and breath-held for ~ 6 s upon full inhalation. A set of ventilation and ADC images were acquired on a 1.5T whole-body MRI system using the hardware and pulse sequences described in full in [5].

Simulations CS in 2D (phase encode k_y) was simulated by under-sampling the k-space of each slice. Four random under-sampling patterns were generated with variable density, as described in [4], with reduction factors of 1.56, 2, 2.9, and 3.79 for the 2D scheme. Images from the under-sampled CS data were reconstructed by minimisation of the following formula [4] using [6]: $\underset{m}{\operatorname{argmin}} \|Fm - y\|_2^2 + \lambda_1 \|\Psi m\|_1 + \lambda_2 TV(m)$ where F is the encoding matrix, m the image we are aiming to reconstruct, y the under-sampled acquired data, Ψ the sparsifying transform (here Daubechies 4 wavelet), TV the total variation regularisation and λ_1, λ_2 weighting parameters to balance data fidelity and artefact reduction. The simulated reconstructions were compared with the images from the fully sampled data. Reconstruction quality was estimated using the mean absolute error (MAE), i.e. mean absolute value of the difference between the reference image and the CS reconstructed image, note that both the lung field and the background noise was used to compute the MAE. Prospective CS acquisition was then simulated by applying the CS undersampled k-space pattern to the previously acquired k-space of 2D ventilation images acquired from the other subjects. For the ADC simulation, both k-spaces of the reference and diffusion weighted images were undersampled in the phase encoding direction as described for the 2D ventilation images.

Results and Discussion



ADC ref (cm^2s^{-1})	ADC CS (cm^2s^{-1})
0.183 ± 0.059	0.184 ± 0.051
0.176 ± 0.058	0.178 ± 0.048
0.156 ± 0.054	0.157 ± 0.045
0.144 ± 0.046	0.144 ± 0.036
0.145 ± 0.044	0.145 ± 0.037

Table.1

Fig. 1 and fig. 2 are examples of CS reconstruction from two different subjects using the same optimised CS pattern. The first row represents the reference (ventilation) images of the five coronal slices; the second row corresponds to the images obtained by reconstructing under-sampled k-space with a CS reduction factor of 2. The lower row illustrates the difference between the reference and the CS images. Good visual replication of the fully sampled images was possible with a MAE value of 1.46%. Fig. 3 emphasises the effect of reduction factor on the quality of the reconstruction. A reduction factor of 2 seems to be the limit for a satisfactory reconstruction in the case of 2D images with this SNR (SNR average = 19). Table.1 shows the mean and s.d. of the ADC map per slice, for fully sampled and 2 fold CS schemes, the means are in agreement, however the s.d. of the ADCs from the CS reconstructed images are lower which could be a result of the inherent spatial smoothing imparted by the CS undersampling algorithm. In conclusion, HP ^{3}He lung images appear to be compatible with the CS method and a good 2D reconstruction with relatively small error was achieved with a reduction factor of 2. Reconstructed CS images from different subjects were obtained using the same undersampling pattern suggesting that once we have a pattern that allows good reconstruction it can be used for prospective acquisition of undersampled lung images in other subjects. Current work is focused on implementation of 2D and 3D CS schemes for acquisition of undersampled k-spaces. With an appropriate pulse sequence acquisition that can sample the prospective CS k-space, an associated reduction in acquisition times could be achieved. The remaining polarisation could then be used to increase spatial resolution by representing a larger fully sampled phase encode matrix or alternatively could be utilised to acquire additional functionally weighted scans such as extra DW scans. Alternatively RF flip angles [1, 2] could be increased with the CS subsampling in the interest of increased SNR/t.

References [1]R.F.Lee et al., MRM, 2006.55(5):1132-41. [2]J.M.Wild et al., MRM, 2003.49(6):991-7. [3]R.O'Halloran et al.Proc. ISMRM. 15 (2007). [4]M.Lustig et al, MRM, 2007.58(6):1182-95. [5]J.M.Wild et al, MRM, 2007. 57(6): 1185-9. [6] http://www.stanford.edu/~mlustig/SparseMRI.html.

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