Under-Sampled Dynamic Reconstruction Using Motion Model of the Tissues and Multiple Coils

C. Prieto[1], J. Hajnal[2], M. Guarini[1], and P. Irarrazaval[1]
[1]Departamento de Ingenieria Electrica, Pontificia Universidad Catolica de Chile, Santiago, Chile, [2]Robert Steiner MRI Unit, Imaging Sciences Department, MRC Clinical Sciences Centre, Hammersmith Hospital, Imperial College London, London, United Kingdom

INTRODUCTION. Recently, a new approach based on the motion estimation of *ob*ject *el*ement*s*, *obels*, (i.e. pieces of tissue) to reconstruct undersampling dynamic MRI was presented [1-2]. In contrast with most of the methods, which are based on the modelling of pixel intensity changes [3-5], it requires a static reference frame and the motion model for all the *obels* in this frame. The advantage of this approach is that the displacement of an *obel* is smoother than the pixel intensity changes at a fixed location. The method can successfully reconstruct dynamic images, although there is a trade off between the maximum undersampling factor and the accuracy of the motion estimate. In this work we propose to use additional data from multiple receiver coils to increase the available information by combining the above method based on *obels* with parallel imaging [6]. The idea relies on the fact that the static reference frame and the *obels'* displacements do not change with coil sensitivities, therefore, it is possible to apply the same model to each single-coil image, increasing the data without modifying the unknowns of the reconstruction process.

This abstract describes the method and the results of applying it to cardiac images acquired with multiple receiver coils. Using an array of 4 coils it shows that an undersampling factor of 16 is quite feasible, and that for an undersampling factor of 8 images display better quality than those obtained using only one coil with the same undersampling.

METHOD. Let $\boldsymbol{m}_t$ be a discrete image of a dynamic sequence at time t. Under the assumption that *obels* do not change their intensity over time, if a reference frame $\boldsymbol{m}_0$ and the displacement of each *obel* initially defined in this frame are known, it is feasible to obtain any frame of the sequence using $\boldsymbol{m}_t = \boldsymbol{P}_t(\boldsymbol{e})\cdot\boldsymbol{m}_0$. The parameterised transformation matrix $\boldsymbol{P}_t(\boldsymbol{e})$ describes the spatial displacements over time for each *obel* in $\boldsymbol{m}_0$. In the multiple coils case the relation becomes $\boldsymbol{m}_{t,i} = \boldsymbol{c}_i\cdot\boldsymbol{m}_t = \boldsymbol{c}_i\cdot\boldsymbol{P}_t(\boldsymbol{e})\cdot\boldsymbol{m}_0$ for each coil i, where $\boldsymbol{m}_{t,i}$ is a single-coil image and $\boldsymbol{c}_i$ the corresponding complex coil sensitivity. The undersampled acquisition is described by $\boldsymbol{b}_{t,i}=\boldsymbol{W}\cdot\boldsymbol{S}_t\cdot\boldsymbol{W}\cdot\boldsymbol{c}_i\cdot\boldsymbol{P}_t(\boldsymbol{e})\cdot\boldsymbol{m}_0$, where $\boldsymbol{b}_{t,i}$ represents aliased single-coil data, $\boldsymbol{S}_t$ is the undersampling pattern and $\mathbf{W}$ the Fourier transform matrix. This equation is a non linear system with unknowns $\boldsymbol{m}_0$ and $\boldsymbol{e}$; which are independent of the coils. The system becomes fully determined if $(N_d\cdot N_t\cdot N_c)/Q>(N_d+N_e)$ where Q is the undersampling factor, N_d the image size, N_t the number of frames, N_c the number of coils and N_e the total number of parameters needed to model the motion. Thus, the higher the number of coils, N_c, the higher the feasible values of Q or N_e. As described in [1-2] for a single coil, the above equation is solved by two nested optimization loops.

The proposed algorithm was used to reconstruct a 2D cardiac sequence acquired using a Philips Intera 1.5T with an array of 4 coils (*B-FFE, 256·154·50, 1.56·2.08 mm*^{2} resolution, *TR/TE=3/1.46 ms*). The raw data was undersampled post acquisition by factors of 8 and 16 using a lattice pattern in the phase encoding direction. The coil sensitivities were estimated from the acquired data. To reduce the computational load we worked on spatial regions of interest of 64·64 and 128·128 with 50 frames. We considered each pixel in $\boldsymbol{m}_0$ to be an *obel*, and B-Splines with 3 coefficients were used to describe their displacement in each Cartesian direction. The reconstruction took approximately 6 hours on a regular PC.

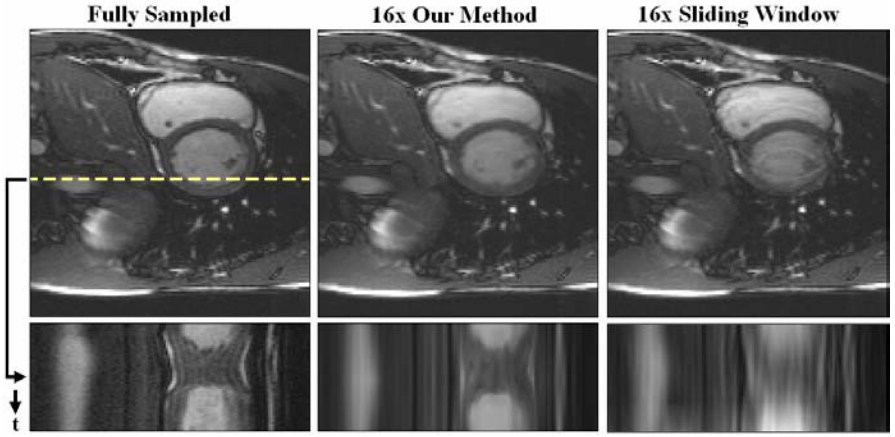

Fig1. Fully sampled (first column), proposed (second column) and sliding window (third column) 16x reconstruction for one cardiac phase and time evolution.

RESULTS. The reconstruction results for an undersampling factor of 16 are shown in Fig.1 for one cardiac phase. The image reconstructed with the proposed method has an RMS error of 1.72% compared to the fully sampled image, while the one reconstructed with Sliding Window has an RMS error of 2.03%. Most of the aliasing was eliminated and only a slight spatial and temporal blurring remains, which is dependent on the quality of the reference frame reconstruction. The results for single and multiple receiver coils with an undersampling factor of 8 are shown in Fig.2. The images are in good agreement with the fully sampled image with RMS errors of 2.65% and 2.45% for a single and multiple receiver coils, respectively.

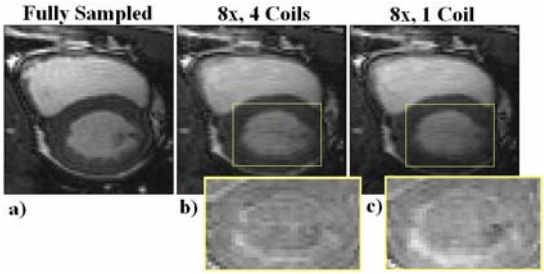

Fig2. a) Fully sampled. b)-c) 8x our reconstruction and difference respect fully sampled, b) multiple coils c) single-coil.

CONCLUSION. A modification to the method based on modelling *obel* displacement to reconstruct undersampling dynamic images is described. Further undersampling and/or improved reconstruction accuracy is possible through the use of parallel imaging data. The method has been tested on cardiac images acquired with an array of 4 receiver coils for an undersampling factor of 16 and to improve the quality of the images previously achieved with undersampling factor 8. This method does not require the motion to be confined to a portion of the field of view or to a portion of the temporal frequency and an approximation of the motion vector field is obtained as an additional result.

REFERENCES. [1] Prieto et al. In Proc.13th ISMRM; [2] Prieto et al. In Proc.14th ISMRM; [3] Madore et al. MRM 42,5(1999); [4] Tsao et al. MRM 50,5(2003); [5] Irarrazaval et al. MRM 54,5(2005); [6] Pruessmann et al. MRM 42,5 (1999).