(RATE) Rapid MRI Acquisition using Tailored Signal Excitation modules: A k-space aliasing method to accelerate MRI scans

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INTRODUCTION

A new MRI acceleration technique, referred to as RATE, which relies on a new concept of k-space aliasing, is introduced. RATE uses tailored excitation modules consisting of RF pulses and gradients to deliberately overlap distinct k-space points. This accelerates a scan as all k-space points in a time frame can now be sampled in a time span shorter than that of a full acquisition. RATE uses the RF excitation pulses to tag the overlapped points that are then resolved through Fourier transformation in time. K-space signal can either have a narrow spectrum because they come from features that are non-dynamic [1] or because they come from structures that have weak signal magnitude. In the latter case, the temporal spectrum may not be narrow but only a small portion remains relevant as the rest remains under noise. RATE exploits these properties through a targeted allocation of temporal bandwidth such that the bandwidth assigned to central k-space is much higher than that assigned to the edges. When used with Parallel Imaging [2,3] (PMRI), the total acceleration will be A_{RATE}*A_{PMRI} where A_{PMRI} is PMRI acceleration and A_{RATE} is RATE acceleration.

THEORY

A tailored signal excitation module designed for an acceleration factor of 2, is given in Fig.1. The RF pulses $P_1(t)$ and $P_2(t)$ are slice selective, the flip angles of both the pulses are equal to θ. $P_1(t)$ and $P_2(t)$ excite the same slice and are accompanied by gradient blips with amplitudes G_{pe1} and G_{pe2} in the primary phase encoding direction. The areas of these gradient blips can either be constant or can be varied from one TR to the next. The resulting signal is given by $S(t)=\iint O(x,y)\{a_1 e^{-jk_{1y}} + a_2 e^{-jk_{2y}}\} e^{-jk_x x}$ $dxdy$. Here, $k_1=\gamma(G_{pe1}+G_{pe2})t_p$ and $k_2=\gamma G_{pe2} t_p$, t_p is the duration of the gradient blips, $O(x,y)$ is a 2D slice and a_1, a_2 are weighting coefficients, that are user defined. The magnitude and phase of the weighting coefficients are dependent on the flip angle and phase of individual RF pulses in the excitation module. Therefore, $a_n = A_n e^{-j\Phi_n}$ where A_n is determined by the flip angles of some/all the RF pulses in the module and Φ_n is the initial phase of the n^{th} RF pulse in the module. Now, consider a first acquisition where A_{RATE}=2, $A_1=A_2=1$, $\Phi_1=\Phi_2=0$, such that two phase encodes k_1 and k_2 overlap onto a single k-space point $k_{alias}=k_1+k_2$. Now, consider a 2nd scenario where all parameters remain the same except for Φ_2, which is set to π. The overlapped point k_{alias} is now k_1-k_2. Now, consider the acquisition of a time series of images where $A_1=A_2=1$, $\Phi_1=\Phi_2=0$ for every even time frame and $A_1=A_2=1$, $\Phi_1=0$, $\Phi_2=\pi$ for every odd time frame. The overlapped point $k(t)_{alias}$ is then given by $k(t)_{alias}=k_1(t)+k_2(t)e^{j\pi t}$ where t is the time frame number. A Fourier transformation in time will give the temporal spectrum of $k(t)_{alias}$. As shown in Fig1(b), this spectrum will consist of $k_2(f)$, centered at Nyquist, and that of $k_1(f)$, centered at the DC, which is extracted using a Fermi filter.

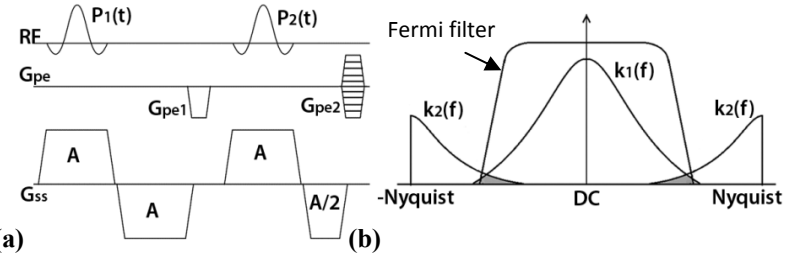

Figure 1: (a) Excitation module for A_{RATE}=2. **(b)** $k_1(f)$ and $k_2(f)$; the Fourier spectrum of $k_1(t)$ and $k_2(t)$ respectively, separated using a Fermi filter.

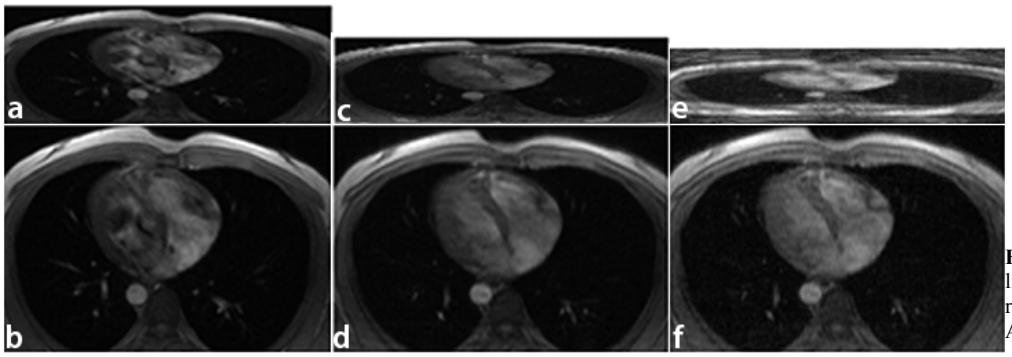

Figure 2: (a) A single cardiac phase from a RATE dataset (A_{RATE}=2) and **(b)** after RATE reconstruction. **(c)** A slice from a different RATE dataset (A_{RATE}=3) and **(d)** after RATE reconstruction. **(e)** RATE-GRAPPA dataset with A_{RATE}=3 and A_{PMRI}=2 for a total acceleration of 5.2. **(f)** Image (e) after RATE and GRAPPA reconstruction.

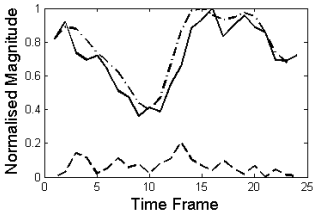

Figure 3: continuous, dashed-dotted and dashed lines show temporal variation for original, reconstructed and error images respectively for A_{RATE}=3.

RESULTS

A cardiac cine feasibility study was conducted on a 1.5T Siemens scanner using a FGRE sequence with FOV: 300*300mm, 256 phase encodes, TR=6ms, 16 cardiac phases, segment size of 10 with an excitation module designed for A_{RATE}= 2. With a module designed for $A_{RATE\text{-}PMRI}$= 5.2, the altered scan parameters were TR=7ms, segment size=6 and 23 cardiac phases were collected. Figure 2 illustrates RATE and RATE-PMRI outputs. Simulations results in Fig.3 confirmed the fidelity of the reconstructed temporal information with an error range of 0.0 to 0.15. For A_{RATE}=3, central k-space was allocated a bandwidth of 13 Hz while the rest of k-space was allocated 5Hz. With PMRI, this bandwidth allocation was nearly doubled.

DISCUSSION

The FGRE sequence was used for feasibility demonstration only and is sub-optimal because as A_{RATE} is increased; the TE and TR get longer, thus negating the benefits of acceleration. However, the excited signal's property is such that the one or more k-space lines acquired in the TR will always be from the desired accelerated k-space. Therefore, the excitation module can be used sparingly, such as in sequences like EPI and Fast Spin Echo (FSE) that can collect multiple aliased k-space lines per excitation, to preserve the benefits of acceleration. With regards to CE-MRA, RATE can ensure not only an increase in the probability of capturing the peak arterial time frame but also sample all k-space points in every accelerated time frame. This eliminates the need for methods like sliding window reconstruction or temporal data interpolation to estimate unacquired data. It is also stated that, while there are some similarities with UNFOLD [4] as far as temporal filtering is concerned, the RATE concept is completely different. As a result, the primary limit on A_{RATE} is the excitation module duration and the current achievable A_{RATE} is in the range of 5-7, with multiple-echo sequences. Future work involves using RATE-PMRI with very large acceleration factors, for high SNR applications such as CE-MRA.

REFERENCES

[1] Korosec et al. MRM. 36:345, 1996. **[2]** Griswold et al. MRM. 47:1202, 2002 **[3]** Pruessmann et al. MRM.42:952, 1999**[4]** Madore et al. MRM.42:813, 1999