An On-the-Fly Radiofrequency Pulse for Bilateral Excitation with Independently Modulated Phase

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INTRODUCTION: Numerous applications of MRI can benefit from imaging multiple 3D volumes, such as breast, kidney, and lower extremity imaging. As with 3D imaging compared to multislice imaging, simultaneous excitation of all volumes can improve the signal-to-noise ratio (SNR) of the acquisition compared to sequential imaging of each volume [1], is particularly useful for dynamic imaging of contrast applications where synchronous timing for all slabs is desired, and allows parallel imaging across volumes. In many applications the volumes are separated in the phase-encode direction, requiring encoding across both the volumes of interest and the space between them.

Independent modulation of slab phase (IMP) has been demonstrated for dual-slab imaging to avoid encoding the space between volumes [2], resulting in a more time-efficient approach. In this work, we present a method that achieves IMP with short slab-selective RF pulses, rather than the spectral-spatial pulses used in [2], enabling the use of faster imaging techniques, magnetization preparation pulses, and multiecho imaging.

THEORY: Ordinarily, to shift FOV in a phase-encode direction, a linear phase is added to the acquired k-space data. The same linear phase can instead be added during excitation. Independent excitation of multiple slabs allows a different linear phase to be added to each excited slab, so that the slabs appear together, and the FOV can be reduced. The calculation of the appropriate value for this added phase for each slab *n* and for each TR, $\varphi_{n,TR}$, necessary to achieve the desired shift, is straight-forward, and is described in detail in [2].

ALGORITHM: For small tip angles, consider RF_n as the RF waveform for the *n*th slab: in order to shift it to its appropriate spatial location Δz_n (in Hz) we need to modulate it with a complex exponential. This is easily calculated for each slab after the scan prescription. For each TR, we need to add a phase of $\varphi_{n,TR}$ to RF_n, and the RF waveform used for excitation is the sum of the RF waveforms corresponding to all slabs:

$$RF = \sum_n RF_n \cdot e^{-i(2\pi\cdot\Delta z_n\cdot t+\varphi_{n,TR})}$$

This RF pulse is calculated and generated on-the-fly during each TR. For larger tip angles the β polynomial is modulated instead of the RF pulse, and the RF pulse is calculated using the inverse SLR transform [3].

METHODS: We implemented the algorithm described above on GE scanners at 1.5T and 3T. We scanned both phantoms and the breasts of healthy volunteers using a breast coil and obtained sagittal images with an A/P frequency direction. The imaging sequence was a flyback echo-planar trajectory [4] with 4 echoes. The scan parameters were a 20 cm FOV with a 256x256 matrix, and 2 mm slice thickness. The number of phase-encodes per slab covered the entire volume, and was 48 for our breast images and 68 for our phantom.

RESULTS: Simulations verified that the resulting excitation profiles had the expected shape and phase for each TR. We tested a pulse design of particular interest: exciting one slab with a minimum phase pulse and the other slab with a maximum phase pulse, both with a time-bandwidth product (TBW) of 8. For this example, the peak RF amplitude of the resultant RF waveform is within 1% of the peak for either component waveform. Although the pulses have non-linear phase, this is completely acceptable for a 3D acquisition [5] since the worst-case phase within a section is 9° for 32 section, leading to a negligible signal loss.

The *in vivo* and phantom images were acquired with the RF pulse combination described above and shown in figures 2 and 3, and neither shows any visible degradation when compared to images acquired with a single slab and with a linear-phase pulse. The images acquired with on-the-fly IMP pulse were all obtained in a time proportional to the phase-encode reduction fraction. The breast image shown in Fig. 1 a) and b) was reformatted from 96 phase encodes and acquired in 3:36 min, while the single slab image with the same left and right endpoints would require 122 phase encodes and 4:34 min acquisition time. The phantom image showed in Fig. 1 c) and d) was reformatted from 136 phase encodes and acquired in 5:06 min, while the single slab image with the same left and right endpoints would require 172 phase encodes and 6:26 min to be acquired.

DISCUSSION: We have developed an algorithm that allows imaging of multiple slabs with independently modulated phase, and with arbitrary RF pulses for each slab. This in turn allows the simultaneous acquisition of multiple volumes without the need to image the non-excited volume. This algorithm is extendable to an arbitrary number of slabs, and the most important limitations are SAR and peak B_1 of the resulting pulse. An important example with two slabs is shown where the peak B_1 amplitude of an RF pulse used to excite two slabs is only minimally larger than that of the RF pulse used for a single slab, and where the resulting loss in signal is negligible.

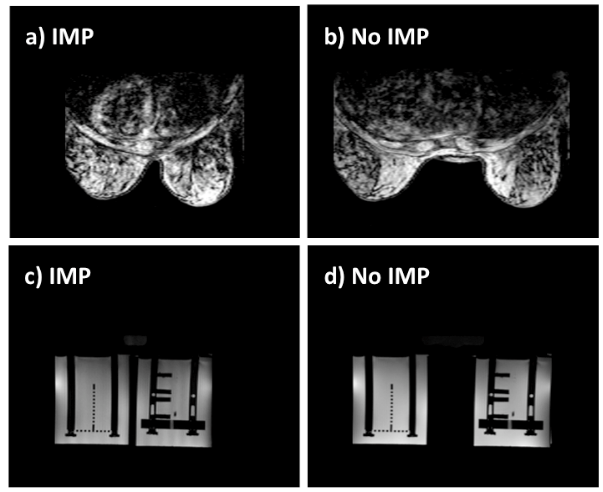

Figure 1: Axial reformats of sagittal images acquired with an RF pulse using IMP (**a** and **c**) and without IMP (**b** and **d**). Image **a** was acquired in 1:20 min shorter time than **b** (saving 26% of the time) and image **d** took 58 sec more to acquire than image **c** (27% savings)

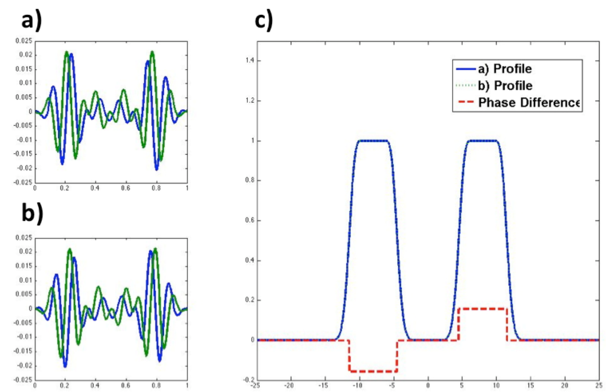

Figure 2: **a** and **b** show the combination of a minimum- and a maximum-phase pulses, one for each slab. The "fast" modulation is due to the location of the slabs. **a** and **b** each have a different additional phase for each slab, creating a subtle, yet important, difference in the waveform. **c** shows the overlapping magnitude slab profiles from playing pulses **a** and **b**, and the phase difference for the two simulation, showing the intended effect.

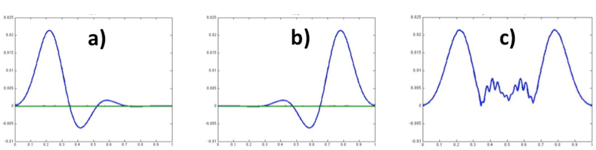

Figure 3: **a** shows a maximum phase pulse as used for the presented results. **b** shows a minimum phase pulse as used for the presented results. **c** shows the sum of the absolute values of these two pulses in the same scale, showing that even for the worse case the peak B_1 is negligibly larger.

REF.: [1] Glover GH, J Magn Reson Imaging 1:457-461, 1991. [2] Hargreaves, BA, Magn Reson Med 57:798-802, 2007. [3] Pauly J, IEEE Trans Med Imag 10:53-65, 1991 [4] Feinberg DA, Magn Reson Med 1:162-169, 1990 [5] Pauly J, Proc ISMRM 9:688, 2001

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