Fat Suppression in Combination with Multiple Overlapping Thin-Slab 3-D Acquisition MR Angiography: Proposed Technique for Improved Vessel Visualization

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PURPOSE: To study the parameters of the 1331 fat saturation pulse (saturation pulse flip angle (SPFA) and interpulse delay (τ)) in order to identify values that provide optimal vessel visibility.

METHODS: Carotid bifurcations of three healthy subjects were studied. RESULTS: The optimal values for the SPFA and τ were 60° (7.5°, -22.5°, 22.5°, -7.5°) and 2.0–3.0 msec, respectively. At these values there was a 1.56-fold increase (SD = ±12, P < .01) in the contrast-to-noise ratio between the left internal carotid artery and fat compared to the study without fat saturation. A 3.0-fold increase was noted for the smaller left occipital artery. Additionally, there was no statistically significant reduction in the vessel-muscle contrast-to-noise ratio nor in the absolute vessel signal-to-noise ratio at these values. CONCLUSION: Implementation of fat saturation into carotid MR angiography markedly improves visualization of vascular anatomy without increasing imaging time.

Index terms: Magnetic resonance, technique; Magnetic resonance, fat suppression; Magnetic resonance angiography (MRA); Arteries, carotid


Time-of-flight (TOF) magnetic resonance angiographic (MRA) techniques (1, 2) are highly T1-weighted and produce a relatively bright signal for fat due to the short T1 of fat and the resulting incomplete saturation of the fat signal. Depending on the parameters of the sequence used, the abundant fat frequently present in the neck surrounding the carotid sheath can have a greater signal intensity than flowing blood. This is particularly true for small vessels and in patients that have reduced flow secondary to stenosis. Thus, in maximum intensity projection (MIP) (3) images of TOF MR angiograms, vessel visibility may be limited primarily by the signal intensity of the background fat.

A number of methods to reduce signal intensity from fat in magnetic resonance (MR) imaging have been developed including, spectroscopic (4, 5) and gaussian presaturation (6) techniques. Hore (7) initially described the 1331 fat saturation pulse and several hybrid techniques have been subsequently developed by Szumowski (8). Preliminary results of attempts at combining fat saturation techniques with MRA have been reported (9). We have recently described a novel hybrid TOF MRA technique that utilizes multiple overlapping thin-slab 3-D acquisition (MOTSA) (10, 11). MOTSA combines the noise reduction of 3-D acquisition with the improved vessel signal of multiple thin-section (2-D TOF) techniques. We have implemented a 1331 fat saturation pulse into the MOTSA sequence to enhance the visibility of smaller vessels and improve the robustness of the technique in slow flow states. The
The purpose of this study was to evaluate the effectiveness and determine the optimal parameters for a 1331 fat saturation in MOTSA MRA. We report a significant improvement in vessel visibility (contrast-to-noise ratio (CNR)) and describe the optimal values of the radio frequency (RF) saturation pulse flip angle (SPF $\alpha$) and interpulse delay ($\tau$) of the 1331 fat saturation pulse.

**Methods**

**Subjects**

Multiple cervical carotid artery MRA were performed on three healthy volunteers. All three subjects were young adult men with an average age of 29.

**MRA**

Imaging was performed at 1.5 T using an unmodified GE Signa MR scanner (GE Medical Systems, Milwaukee, WI) with a quadrature head coil and standard patient positioning. MRA images were obtained using the MOTSA TOF technique that has been described in detail by Parker et al (10, 12). Briefly, the technique consists of 3-D acquisition of multiple overlapping thin slabs within which an arbitrary number of sections may be present. To this technique we added a standard 1331 fat saturation pulse described by Hore (7). The fat saturation pulse consists of four RF pulses separated in time such that at each pulse the relative phase of the fat magnetization is shifted by an additional 180° with respect to that of water (see Fig. 1). The waveform timing diagram is shown in Figure 2. The SPF $\alpha$ and interpulse delay ($\tau$) of these four pulses were varied to identify the values of optimal fat saturation. The SPF $\alpha$ ranged from 0° to 200° (expressed as the sum of the flip angles of the four RF pulses) and $\tau$ extended from 1.5 to 3.5 msec.

The three subjects were imaged in the axial plane just above the carotid bifurcation using the SPF $\alpha$ and $\tau$ values described above and the following imaging parameters: 55/5.5/1 (TR/TE/excitations), 24-cm field of view and a 256
The average signal intensity normalized to noise (signal-to-noise ratio (SNR)) of fat, muscle, and vessel (left internal carotid artery (LICA) and left occipital artery (LOA)) subregions were calculated for the subjects at various SPFA and τ values. These two arteries were chosen to cover the interesting range of vessel sizes and flow rates. It is expected that the smaller vessel with slower flow (i.e., less blood contrast) will show the largest improvement from the reduction in fat signal. The CNRs between LICA and fat, LICA and muscle, LOA and fat, and LOA and muscle were calculated for each of the various fat saturation pulse parameters. Vessel regions of interest (ROIs) were circular, centered over the vessel cross-section to include only pixels that were vessel. The fat ROI was an elliptical shape, placed in the subcutaneous fat in the posterior neck region. The muscle ROI was placed in the muscle deep to the subcutaneous fat in the posterior neck. Noise was computed as the standard deviation in a region external to any tissue signal. Computed in this manner, the noise measure is consistently lower than the true image noise but avoids problems associated with signal intensity. The statistical significance of differences in the absolute signal intensity and CNRs following implementation of the fat saturation pulse into the MOTSA technique was based on the Student’s t-test.

**Results**

In an attempt to optimize the 1331 fat saturation pulse for the MOTSA MRA sequence, SNRs for arteries, fat, and muscle were measured as a function of RF saturation pulse flip angle (SPFA) and interpulse delay (τ) of the 1331 fat saturation pulse. Figure 3 shows the effect of SPFA on fat and muscle for all three subjects studied. In each subject, the SNR of both fat and muscle decreases as the SPFA increases. However, as expected, fat shows a more pronounced decline in signal intensity.

Figure 4 demonstrates the effect of SPFA on the SNR of the LICA, LOA, fat, and muscle subregions for one subject. This data is typical of that seen for all three subjects. The signal intensity from fat rapidly declines as the SPFA increases. The fat signal becomes isointense with muscle when the SPFA (expressed as the sum of the four RF pulses) reaches 60° (7.5°, -22.5°, 22.5°, -7.5°). It continues to decline until a signal intensity less than 35% that of muscle is observed at a SPFA of 120°. The average SNRs of the LICA and LOA remain relatively constant until the SPFA reaches 80°, at which time the intensity begins to decrease with increasing SPFA. Additionally, it can be noted that the average SNR of the LICA is larger than that of the smaller LOA. This is due primarily to relative differences in vessel size with resulting partial volume effect and possibly to differences in blood velocity within these vessels.

Figure 5 shows the effect of τ on the SNR of arteries, fat, and muscle intensity for the same subject with the SPFA held constant. The fat signal demonstrates the characteristic trough shape described by Hore (7). The optimal value of τ for fat suppression ranges between 2.0–3.0 msec (167–250 Hz) with the fat signal reaching a nadir at 2.5 msec (200 Hz). The signal intensity of the LICA and LOA fluctuates slightly with the peak noted at 2.5 ms.

A plot of the artery-fat and artery-muscle CNRs are shown in Figure 6. The CNRs are plotted against changes in SPFA. For the subject whose data are shown, a 1.8-fold increase in the CNR between LICA and fat was observed at a SPFA of 80° (10°, -30°, 30°, -10°) compared to no fat.
saturation. At the point where the CNR of LICA-fat crosses the CNR of LICA-muscle (SPFA = 60°) a 1.7-fold increase in the CNR was observed. For the three subjects studied, an average increase in the CNR of LICA-fat of 1.56 (SD ± 0.12, \( P < .01 \)) was observed at this intersection point compared to the same conditions without fat saturation. Although the CNR of LICA-muscle and the absolute SNR of the LICA decreases at large SPFA's, there was no statistically significant reduction in these values at a SPFA of 60° compared with no fat saturation.

The LOA-fat CNR increases 3.0-fold at a SPFA of 80° compared with no fat saturation. The LOA shows a greater increase in the artery-fat CNR compared with the LICA because the signal intensity of this smaller artery is closer to that of fat tissue when fat saturation is not implemented. Once again, at a SPFA of 60° no statistically significant reduction is noted in the LOA-muscle CNR nor in the absolute SNR of the LOA.

The improvement in CNR between arteries-fat can be appreciated in Figure 7 which shows MIP images of the carotid bifurcation. Figure 7A was obtained without the 1331 fat saturation pulse and shows the bifurcation of the left common carotid artery. The same segment of vessel with implementation of the 1331 fat saturation pulse is shown in Figure 7B. Note that the fat signal from the connective tissue planes and carotid sheath surrounding the vessels has been greatly reduced without significant loss of signal from the vessels. The images in Figures 7A and 7B were obtained with a quadrature transmit-and-receive head coil. Figure 7C shows a MIP image obtained using a receive-only anterior neck surface coil. As evident, there is significant degradation of image quality that occurs as a result of using the body coil to transmit the RF pulses. This is due to inhomogeneity of \( B_0 \) and the resulting saturation of blood by the 1331 pulse outside the field of interest.

**Discussion**

TOF MRA techniques are highly T1-weighted and produce a relatively bright signal for tissues with short T1 such as fat. Increasing the excitation flip angle causes an increase in saturation and a reduction in signal from stationary tissues. Thus, 2-D TOF techniques, which use large flip angles (typically > 60°) have less contribution from the bright fat. The use of large flip angles in 3-D TOF techniques results in loss of vessel
Fig. 7. MIP images of the left carotid artery of subject 1 without fat saturation (A), with implementation of the fat saturation pulse (B) (both A and B were obtained using a quadrature transmit-and-receive head coil). Note the improved vessel visibility due to decreased signal from background tissue. For example, marked improvement in the facial artery is apparent (arrows). Figure C is a MIP image of the same left carotid artery using fat saturation, however, a receive only anterior neck surface coil was used. Obvious degradation of image quality is noted when the body coil is used to transmit due to saturation of blood outside the field of interest and inhomogeneities of $B_0$.

signal from saturation of blood that remains in the imaging region for multiple excitation pulses. To minimize signal loss in 3-D TOF techniques, flip angles of $15^\circ$ to $20^\circ$ are typically used. This compromise in flip angle results in a relatively large signal from fat and still results in significant signal loss from blood that has remained in the slab for multiple pulses. The MOTSA technique, which allows the use of thin slabs, still requires the use of smaller flip angles (typically $30^\circ$) to minimize signal saturation in small blood vessels. Thus MOTSA images, which show less vessel signal loss due to saturation than conventional, single-volume, 3-D techniques, still result in a significant signal from fat.

In addition to signal reduction due to partial saturation, the bright fat signal can be reduced by frequency-selective presaturation and also by using frequency-selective timing in echo readout (B). Selecting echo times such that fat and water are out of phase actually provides significant reduction in the fat signal. Unfortunately, such echo times are not always coincident with the minimum achievable echo time required for maximal blood vessel signal. The effects of such out-of-phase techniques were not evaluated in this study.

A variety of fat presaturation techniques have been described in the literature. The 1331 fat saturation pulse has the advantages of being very selective in a homogeneous $B_0$ field and adds relatively little to the imaging time. In fact, we have found that implementation of the 1331 fat saturation pulse into the MOTSA MRA technique has resulted in no significant increase in the imaging time at a TR of 40 msec. Alternatively, the Gaussian presaturation technique is potentially less sensitive to inhomogeneities of $B_0$ but requires a much longer RF pulse duration to achieve the same frequency selectivity as the 1331 pulse, and thus requires a much greater TR and a greater patient imaging time.
We have observed significant improvement in the CNR between vessels and fat of healthy subjects through implementation of the 1331 fat saturation pulse into the MOTSA technique. This improvement is greatest for regions of low blood velocity (ie, distal to a stenosis), small vessels, and where the vessel is surrounded by fat. Based upon our measurements for the three healthy subjects, the optimal values for SPFA and $\tau$ of the 1331 pulse are 60° and 2.0–3.0 msec, respectively. The use of these parameters results in the improvement observed between Figures 7B and 7A. In this figure, there is marked improvement in vessel visibility due to decreased signal from background tissue. For example, we note the improvement in the facial artery (arrows).

As can be seen from Figure 7C, implementation of fat saturation using a large volume coil can result in a decrease in blood signal. This is apparently due to the saturation of blood from outside the imaging region due to variations in $B_0$. We would conclude that implementation of a 1331 fat saturation pulse requires the use of a restricted volume transmit/receive coil such that $B_0$ remains homogeneous in the excited region to avoid saturation of blood outside the field of interest. This may require the development of special purpose cervical and thoracic transmit/receive coils to replace the use of receive-only surface coils for cervical carotid imaging.

The intensity variation (venetian blind artifact) that is seen at each slab boundary interface is due to several complicating factors and does make image interpretation more difficult. Since the conclusion of this study, this artifact has been substantially reduced by optimization of imaging parameters and improvements to the postprocessing algorithm. Specifically, the artifact is now reduced by increasing the slab excitation width relative to the imaged width and by combining image detail from overlapping sections.

In older patients who have an inherently lower CNR between vessel and fat due to reduced flow states, stenosis, or increased fatty tissue in the neck, this technique may provide significant improvement in vessel visibility. Additional work is needed to evaluate the potential utility of this technique in a broad patient population.

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**References**