Non-Contrast-Enhanced Vein Imaging in the Deep Veins: Impact of Velocity Patterns and Improved Image Quality

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Introduction

MRI has the potential for multi-contrast imaging of venous thrombo-embolism (VTE) including thrombus imaging and luminal imaging methods. A candidate luminal imaging method is Acceleration-Dependent Vascular Anatomy for Non-Contrast-Enhanced MR Venography (ADVANCE-MRV) [1] which is based on a flow-dependent angiography method [2–3] but uses a combination of velocity- and acceleration-sensitised [4] images; these respectively suppress arterial and venous signals and are subtracted to give vein images with most arterial and background signals removed. Recent initial work suggested that, particularly in veins, luminal image quality from such methods can be impaired by apparent signal voids related to imperfect suppression of the flowing blood in the pre-subtraction images. This effect could be related to spatial variations in the vascular flow patterns.

This study aims to investigate the impact of flow dispersion on venous signals and thus to generate improved vein images based on combination of multiple images with varying flow suppression.

Theory

Venous signal is suppressed using an iMSDE module [5] as shown in Fig 1a, which has an effective first gradient moment $\mathbf{m}_1$. The phase change during the iMSDE module in the velocity field $\mathbf{v}$ is given by:

$\phi(\mathbf{r}) = \gamma \mathbf{m}_1.\mathbf{v}(\mathbf{r})$. This can be expressed as $\phi(\mathbf{r}) = \phi_0(\mathbf{r}_0) + \delta\phi(\mathbf{r})$ where

$\phi_0 = \gamma \mathbf{m}_1.\mathbf{v}(\mathbf{r}_0)$ is the phase change due to the mean velocity in a voxel and $\delta\phi(\mathbf{r}) = \gamma \nabla(\mathbf{m}_1.\mathbf{v}(\mathbf{r})).\delta\mathbf{r}$ represents the changes due to the (assumed linear) velocity dispersion within the voxel. Remembering that only one component of the transverse magnetisation is rotated back to longitudinal magnetisation by the final 90°$_{-x}$ pulse, the net signal is given by $S \propto \mathrm{Real}\left(\iiint e^{i\phi(\mathbf{r})} d^3\mathbf{r}\right)$. This can be expressed as

$$S \propto \cos\phi_0 \,\mathrm{sinc}(\beta_x \Delta x)\,\mathrm{sinc}(\beta_y \Delta y)\,\mathrm{sinc}(\beta_z \Delta z) \qquad [1]$$

where $\beta_x = \frac{1}{2}\frac{d\phi}{dx} = \frac{1}{2}\gamma\frac{d(\mathbf{m}_1.\mathbf{v})}{dx}$ represents the phase dispersion in x, and similar expressions apply for β_y and β_z; the voxel dimensions are Δx, Δy, and Δz.

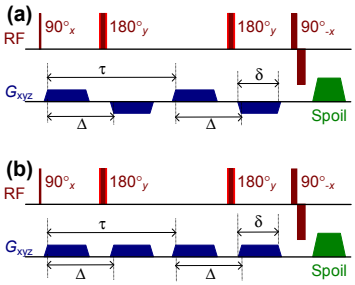

Fig. 1: velocity-sensitive (a) and acceleration-sensitive (b) preparation modules. The motion-sensitisation gradients (MSG) are shown in blue.

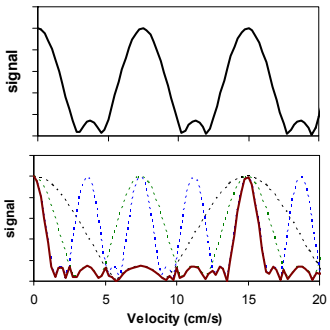

Fig. 2: (a) Signal profile for a single image as a function of velocity; (b) signal combination to achieve low signal over a range of velocities.

The expression for the signal in equation 1 contains a decaying component, given by the sinc functions, which is related to variation of the velocity within the voxel, but also an oscillatory component (cos ϕ_0) which reflects the average velocity. This oscillatory component could be highly relevant in voxels with little velocity dispersion.

In cases where velocity dispersion is low, it is impossible to pick an appropriate gradient moment to guarantee that signal suppression will be achieved for the range of velocities seen *in vivo*. However this can be achieved by combining the signals from multiple acquisitions with different moments. One approach is to acquire several acquisitions with the gradient moment increasing by a factor of 2 each time, and then to take the minimum signal. This is illustrated in Fig. 2, which simulates the signal modulus in the absence of dispersion, followed by a combination of several profiles. An offset has been applied to account for relaxation between preparation and readout.

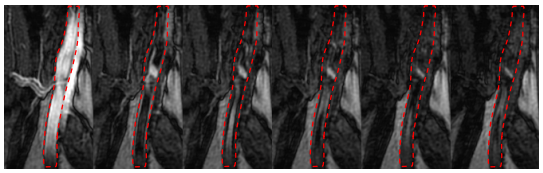

Fig. 2: Slice showing a vein (and artery) with increasing gradient moment (0, 0.4, 0.8, 1.2, 1.6, 2.0 μTs^2/m). The vein signa

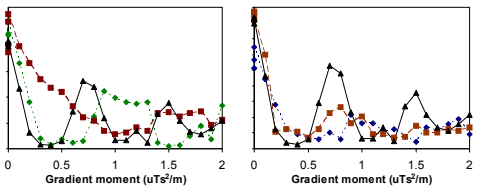

Fig. 3: Graphs of signal variation with gradient moment, showing changes due to with (a) velocity and (b) dispersion, which affect the frequency and amplitude of signal oscillations.

Materials/Methods

Following ethical approval and informed consent, the lower legs of 6 healthy volunteers were imaged with a 1.5 T Signa HDx MR system (GE Healthcare, Waukesha, WI). The iMSDE module was placed at the cardiac trigger delay time corresponding to approximately peak arterial flow, and followed by a 3D balanced-SSFP readout (coronal orientation, flip angle 65°, TE/TR=1.6/3.4 ms, ASSET ×2, acquisition matrix 256×256, FoV 35-40 cm).

In 2 volunteers, a detailed study into the behaviour of the signal in the (unsubtracted) images as a function of gradient moment was carried out: a dataset was acquired using evenly-spaced different gradient moments from 0–2.0 μTs^2/m in steps of 0.1 μTs^2/m; ROIs were drawn in representative locations in the veins and signal was plotted against gradient moment.

In 5 volunteers, datasets were acquired with three gradient moments increasing by successive factors of 2. Additionally, an acceleration-sensitised dataset was acquired to enable the arterial signal to be suppressed by subtraction [1]. (The acceleration-sensitisation module is shown in Fig. 1b). By taking the maximum of the subtracted images, vascular images with reduced signal variation were created.

Results

Fig. 2 shows an example (unsubtracted) images with increasing gradient moment, demonstrating the location-dependent signal fluctuations within the vein. Fig. 3 plots these signal oscillations for different locations and demonstrates a range of oscillation frequencies and amplitudes, corresponding respectively to varying flow velocity and flow dispersion within each voxel.

Fig. 4a–c shows subtraction venograms (all exhibiting good arterial suppression) acquired with successive factor-2 increases in gradient moment. Signal loss (related to the above oscillations) occurs either in the centre or near the edges of the vessel. However this signal loss is greatly reduced in the combined image shown in Fig. 4d. Such improved vessel uniformity was seen in all subjects.

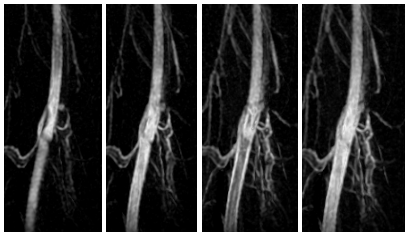

Fig. 4: MIPs of subtraction venograms acquired with gradient moments 0.3, 0.6 and 1.2 μTs^2/m (a–c) and combined image (d) showing improved signal uniformity.

Discussion & Conclusions

For clinical studies of VTE, it is important to be confident that luminal signal voids represent thrombus and not merely signal loss in flowing blood. As demonstrated here, such signal loss can occur but may be overcome by combining multiple images with appropriately chosen flow sensitivities. A similar, though less pronounced signal loss has been seen in arteries, so this approach may also be beneficial for arterial imaging. Since flow dispersion is reduced with smaller voxel sizes, this issue becomes increasingly important as imaging resolution is increased.

References
[1] Priest AN et al. proc ISMRM 2011;19:1289. [2] Priest AN et al. MRM (in press).
[3] Fan Z et al. MRM 2009;62:1523. [4] Priest AN et al. proc ISMRM 2011;19:90. [5] Wang J et al, JMRI 2010; 31:1256.

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