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:
\[
\phi(r) = \mathbf{m}_1 \cdot \mathbf{v}(r)
\]
This can be expressed as
\[
\phi(r) = \phi_0(r) + \Delta \phi(r)
\]
where \(\phi_0 = \gamma \mathbf{m}_1 \cdot \mathbf{v}(r)\) is the phase change due to the mean velocity in a voxel and \(\Delta \phi(r) = \mathcal{J} [\mathbf{m}_1 \cdot \mathbf{v}(r)] \Delta r\) 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° pulse, the net signal is given by
\[
S = \text{Real} [\mathcal{I} (e^{i \phi(r)} + e^{i \phi_0})] \tag{1}
\]
This can be expressed as
\[
S = \cos(\phi_0 + \Delta \phi) = \cos(\phi_0) \cos(\Delta \phi) - \sin(\phi_0) \sin(\Delta \phi) \tag{1}
\]
where \(\phi_0 = \gamma \mathbf{m}_1 \cdot \mathbf{v}(r)\) represents the phase dispersion in x, and similar expressions apply for \(\phi_1\) and \(\phi_2\); the voxel dimensions are \(\Delta x\), \(\Delta y\), and \(\Delta z\).

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.

Materials/Methods
Following ethical approval and informed consent, the lower legs of 6 healthy volunteers were imaged with a 1.5 T Sigma 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 x2, 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 with evenly-spaced different gradient moments from 0–2.0 μT/m (a–c) and combined image shown in Fig. 4d. Such improved vessel uniformity was seen in all subjects.

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.

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

Acknowledgements
We thank Addenbrookes Charitable Trust and the NIHR Cambridge Biomedical Research Centre for funding.