Identification and Reduction of Residual Signal from Slow Flowing Blood in 3D Volume Selective Turbo Spin Echo Arterial Wall Imaging Using a Velocity Sensitive Phase Reconstruction Method.

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Aim
Slow flowing or recirculating blood can lead to incomplete nulling in 3D vessel wall imaging making it difficult to differentiate between diseased arterial wall and blood pool. The aim of this work is to improve 3D volume selective Turbo Spin Echo (TSE) arterial wall imaging by introducing velocity phase sensitivity to the sequence so that the reconstructed phase images can be used to control the removal of residual blood signal with post-processing.

Introduction
The majority of vessel wall imaging of the carotid artery has focussed on 2D techniques.[1-4] This is partly due to time and coverage constraints and because of the problems with blood suppression over a large 3D slab. Slow recirculating flow, believed to be a feature of carotid bulb geometry,[5] can contribute to the problems with blood suppression making it difficult to differentiate between diseased arterial wall and blood pool.

Methods
Images were acquired on a Siemens Magnetom Sonata 1.5T scanner. 3D volume selective TSE[8] sequence was tested firstly in a pulsatile flow phantom and then in the arterial wall were acquired in healthy volunteers and patients. In vitro validation of the residual blood signal removal was carried out using a pulsatile flow system with blood mimicking fluid and a phantom with realistic carotid artery geometry (Shelley Medical Imaging Technologies). A carotid pulse waveform with peak flow of 40 ml/s and 'R'-interval of 828 ms was used. The sequence was also connected to the phantom to enable ECG gating. For in vivo validation, the study included 6 healthy subjects (average age 35, range 23 - 47 years) and 17 patients with known carotid artery disease (average age 63, range 48 - 76 years). All subjects gave informed consent.

Typical imaging parameters are FOV 120 x 24 mm, matrix size 256x52 (true resolution 0.47x0.47 mm), 18 slices each 2 mm thick, central slice located at the bifurcation, echo train length 11 and a short acquisition window of 65 ms to avoid motion blurring of the vessel wall due to pulsatility. For these T1 weighted images, TE was 11 ms and the acquisition was cardiac gated with dark blood double inversion preparation. The images and navigator were acquired using a 2-element phased-array coil (Machnet BV, The Netherlands). A velocity encoding bipolar gradient pulse was added to the sequence and magnitude and phase image reconstruction was carried out. Some minor velocity sensitivity was observed in the standard sequence and the bipolar pulse, with as high a sensitivity as could be achieved without altering the timing of the sequence or going beyond the gradient limits, was added with the correct polarity to increase sensitivity. A velocity phase sensitivity of ±40 rad/cm/s was used as this was the maximum that could be added without further alterations to the sequence.

Lumen to wall contrast ratio, (S_lumen − S_tissue) / S_lumen, was measured on magnitude images from both the standard and velocity sensitive sequences for quantitative analysis. Values were taken from an average over several slices but only where blood signal was observed.

A MATLAB (The MathWorks, Inc.) program was written to threshold the phase images to identify regions of motion and subsequently remove blood signal from affected magnitude images by multiplication of the magnitude and phase images. Recirculating blood to improve image clarity and assist conclusions about the disease state of the vessel wall.

Results
Post-processing correction using phase reconstruction intensifies removes the artificial signal as shown in the phantom images in Figure 1.

Discussion
Optimisation of the velocity sensitivity is needed to achieve the required phase differences for the threshold depending on the flow rates involved. Fast flow artefacts are seen in 3D images of both healthy and diseased vessels and, in some cases, faster flow in the region of a stenosis will help to reduce this effect. Lumen encroachment by disease is not affected and images remain unchanged when correction is applied. Figure 3 shows removal of blood signal from the internal carotid artery by the threshold correction. The clarity of the lumen can be improved by this method as illustrated in figure 4, where the double inversion blood suppression was adequate but not ideal.

A limitation is, of course, that any truly stationary blood will not be affected by this technique and will always be difficult to remove from such images.

Table 1

<table>
<thead>
<tr>
<th>Mean lumen signal</th>
<th>Mean wall signal</th>
<th>Lumen wall contrast</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standard</td>
<td>Sensitive</td>
</tr>
<tr>
<td></td>
<td>Velocity</td>
<td>Standard</td>
</tr>
<tr>
<td>mean (sd)</td>
<td>0.69 (0.11)</td>
<td>0.73 (0.11)</td>
</tr>
<tr>
<td>mean (ad)</td>
<td>0.63 (0.24)</td>
<td>0.55 (0.24)</td>
</tr>
<tr>
<td>mean (md)</td>
<td>0.52 (0.19)</td>
<td>0.49 (0.19)</td>
</tr>
<tr>
<td>mean (overall)</td>
<td>0.61 (0.24)</td>
<td>0.53 (0.22)</td>
</tr>
</tbody>
</table>

* no significant difference (p > 0.2)
† significant difference (p < 0.01)

References

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