Can flow be measured with a flow-compensated sequence?

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PURPOSE: A new flow velocity mapping method is proposed. As opposed to traditional phase contrast imaging that uses a velocity encoding gradient[1], this method employs a multi-echo pulse sequence that is flow compensated[2,3]. This sequence is a basic pulse sequence method for quantitative susceptibility mapping (QSM)[4] but uses the acquired phase date in a different way. The combination of non-zero flow velocity and spatially varying field inhomogeneity leads to a signal phase that is non-linear in echo time. By modifying the field inhomogeneity in a controlled manner, it is then possible to measure flow.

THEORY: For a moving spin, the phase it accumulates after RF excitation depends on the field it experiences. The phase $\phi(t)$ can be expressed as:

$$\phi(t)=\psi+\gamma\int_0^t \delta B(\mathbf{r}(\tau))d\tau \qquad [1]$$

where ψ is the initial phase after the RF pulse and $\delta B(\mathbf{r})$ is the field inhomogeneity on the spin's travelling path. Assuming that, for a small region around **r**, $\mathbf{v}(\mathbf{r}(\tau))$ is constant in space; and during k-space sampling, for $0<\tau<TE$, the velocity $\mathbf{v}(\mathbf{r}(\tau)) \sim$ constant, we can write $\mathbf{r}(\tau)=\mathbf{r}(TE)+\mathbf{v}(\mathbf{r}(0))(\tau-TE)$. Furthermore assuming that $\delta B(\mathbf{r})$ is linear in a small region, i.e. $\delta B(\mathbf{r})=\delta B(\mathbf{r}_0)+(\mathbf{r}-\mathbf{r}_0)\bullet\nabla\delta B$, we can write:

$$\phi(TE) = \psi+\gamma\cdot\int_0^{TE}\delta B(\mathbf{r}(TE)+\mathbf{v}(\mathbf{r}(0))(\tau-TE))d\tau = \psi+\gamma\cdot\int_0^{TE}[\delta B(\mathbf{r}(TE))+\mathbf{v}(\mathbf{r}(0))(\tau-TE)\bullet\nabla\delta B(\mathbf{r}(TE))]d\tau$$

$$\phi(TE) = \psi+\gamma\cdot\delta B(\mathbf{r}(TE))TE-\tfrac{1}{2}\cdot\gamma\cdot\mathbf{v}(\mathbf{r}(0))\bullet\nabla\delta B(\mathbf{r}(TE))TE^2 = \psi+L\cdot TE+Q\cdot TE^2 \qquad [2]$$

The phase $\phi(t)$, rather than a linear function of TE, becomes a quadratic function of TE. The linear coefficient L is the field inhomogeneity map and the quadratic coefficient Q is proportional to flow velocity and the spatial gradient of field inhomogeneity. By manually changing the shimming gradient we could get another fitting: $\phi(\mathrm{TE})^* = \psi^*+L^*\cdot\mathrm{TE}+Q^*\cdot\mathrm{TE}^2$.Then:

$$Q^*-Q = (-\tfrac{1}{2}\cdot\gamma\cdot\mathbf{v}(\mathbf{r})\bullet\nabla\delta B^*(\mathbf{r}))-(-\tfrac{1}{2}\cdot\gamma\cdot\mathbf{v}(\mathbf{r})\bullet\nabla\delta B(\mathbf{r})) = -\tfrac{1}{2}\cdot\gamma\cdot\mathbf{v}(\mathbf{r})\bullet(\nabla\delta B^*(\mathbf{r})-\nabla\delta B(\mathbf{r})) = -\tfrac{1}{2}\cdot\gamma\cdot\mathbf{v}(\mathbf{r})\bullet\nabla(\delta B^*(\mathbf{r})-\delta B(\mathbf{r})) = -\tfrac{1}{2}\mathbf{v}(\mathbf{r})\bullet\nabla(L^*-L)$$

$$\mathbf{v}(\mathbf{r}) = -2\nabla(L^*-L)^{-1}(Q^*-Q) \qquad [3]$$

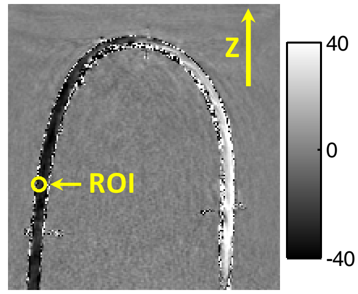

Fig. 1 Velocity map (cm/s) of one experiment with 36 cm/s flow rate

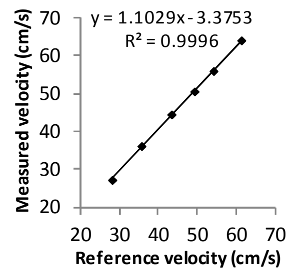

Fig. 2 Comparison between measured and reference flow

METHOD: A 3D multi-echo spoil gradient echo sequence with flow compensation along all axes is used. A flow phantom experiment was carried out to validate the proposed method. In vivo scans were performed and 3D vessel velocity map is generated. ***1) Phantom validation*:** A U-shaped tube connected to a pump was fixed inside a plastic box which was filled with tap water and put inside the scanner bore. Constant flow could be generated by the pump. Water flow with variable velocity from 28cm/s to 62cm/s was imaged and measured with the proposed method. Measured velocity was compared with that measured by a phase contrast scan. ***2) In vivo study*:** $N = 3$ human subjects were scanned. Typical scan parameters were: $TR/TE_{first}/TE_{last}$ = 37.7/2.8/32.5ms, FA = 20°, FOV = 24cm, voxel size = 1×1×1mm^3. Each volunteer was also scanned with a 3D phase contrast sequence, the result of which was compared with that of the proposed method.

RESULTS: Fig. 1 shows the z direction flow velocity map of one phantom experiment at 36cm/s. Fig. 2 shows the flow measured by proposed method verse the reference flow from phase contrast imaging at the ROI indicated in Fig. 1. Linear regression gives a slope of 1.1 and R^2 of 0.99. Fig. 3 shows superior-inferior velocity map of an in vivo study at 3 consecutive slices. Blood flow velocity measured in sagittal sinus and middle cerebral artery are 14.3 cm/s and 32.2 cm/s, compared to 13.4 cm/s and 35.4 cm/s, respectively, when measured using phase contrast.

DISCUSSION AND CONCLUSION: A flow compensated multi-echo acquisition was used to perform flow measurements by modifying the magnetic field. The use of flow compensation gradients allows a straightforward measurement of the signal phase without the confounding pixel displacement artifacts that increase with increasing echo time. As arterial flow is pulsatile, also considering vessels with large curvature, higher order polynomial fit could be used to get more accurate flow estimation.

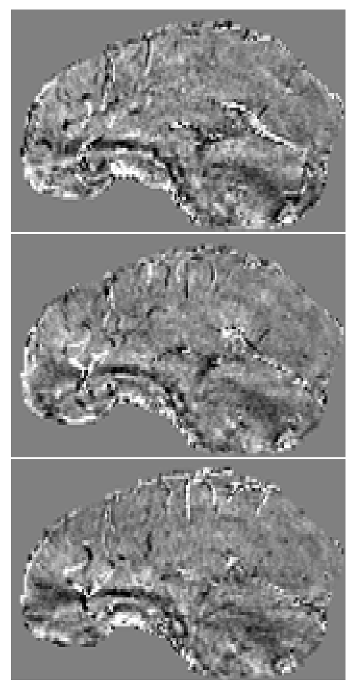

Fig. 3 Superior-inferior velocity map of an in vivo study (shown are 3 consecutive slices

REFERENCES: [1] Pelc et al. JMRI 1991; 1:405-413. [2] Bernstein et al. JMRI 1992; 2:583-588. [3] Slavin et al. MRM 1997; 38:368-377. [4] Liu et al. MRM 2012, DOI: 10.1002/mrm.24272.