Centre-Accelerated Twisted Projection Imaging for ^{23}Na MRI

R.W. Stobbe, C. Beaulieu
Department of Biomedical Engineering, University of Alberta, Edmonton, Alberta, Canada

Introduction: One of the primary advantages of Twisted Projection Imaging (TPI) [1] is its increased radial trajectory evolution at the centre of k-space, which reduces the initial signal decay rate across k-space from the centre out (i.e. beneficial transfer function reshaping). Rapid signal decay around the centre of k-space otherwise smears signal into the peripheral regions of the point-spread-function, resulting in image intensity values in small objects that may not reflect the true signal value [2]. This transfer function reshaping advantage was actually put forward in the context of Density-Adapted Projection-Reconstruction (DA-PR) [3] which includes the radial evolution of TPI (without twisting for trajectory number reduction). Beneficial radial evolution alteration (and twisting) begins at the radial fraction $r = \rho$, but within this fraction (around the centre of k-space) evolution remains constant. Initiating alteration sooner (i.e. ρ value reduction) increases transfer function reshaping, but small ρ values lead to greater twisting and are constrained by gradient slew limitations. Here a new technique, Centre-Accelerated TPI (CA-TPI), is introduced to further reduce the impact of rapid signal decay.

Methods: The differential equations defining CA-TPI are similar to those of standard TPI. However, they are solved not only outward but also inward from radial fraction ρ (see [**Eqs. 1 – 3**], where $\Gamma(r)$ is the sampling density design function with requirement that $\Gamma(\rho) = 1/\rho^2$). For standard TPI these equations differ in that $w = (1 - (\rho/r)^4)$. This yields trajectory twisting (or azimuthal-angle evolution, $\dot{\theta}$) with constant gradient magnitude, but only for $r > \rho$. Within ρ, constant gradient magnitude is maintained with $\dot{r} = 1$ and $\dot{\theta} = 0$. The alternative w term of **Eq. 3** yields very similar rates of twisting $(\dot{\theta}/\dot{r})$, but unlike standard TPI these rates do not decay to zero at ρ. This facilitates inward solution without discontinuity for the purpose of implementation with initial gradient slew constraint. The goal of inward solution is to increase $\dot{r}$ beyond 1 within ρ.

$$\dot{r} = \frac{1}{\Gamma(r) \cdot r^2} \qquad \textbf{Eq. 1}$$

$$\dot{\theta} = \sqrt{\frac{w}{r^2 sin^2(\phi)}} \qquad \textbf{Eq. 2}$$

$$w = 1 - exp(-1.5 \cdot r/\rho) \qquad \textbf{Eq. 3}$$

CA-TPI was implemented on a Siemens Prisma 3T, using a dual-tuned birdcage head-coil (Rapid Biomedical). Two trajectory sets were implemented that fully sampled a 280 mm isotropic FOV (the extend excited by the coil) with 2500 trajectories and produced a β = 3 Kaiser filtering shape with sampling density beyond ρ. Readout durations were (5 ms, 20 ms) with isotropic voxel dimensions (5.5 mm, 4.35 mm) respectively. Gradient slew limited minimum ρ values were selected (0.25, 0.15). The theoretical advantage of CA-TPI was assessed with Minimum Evaluable Object Volume (MEOV) metric calculation [2], using T_{2f}* = 2.5 ms and T_{2s}* = 25 ms (from 3T brain measurement). MEOV is the minimum volume (e.g. lesion) such that image intensity measurement is at least a given percentage of the full actual signal produced (80%, 90% considered), and potential error with respect to noise is less than a given percentage of reference signal (5%, 2.5% considered). Images were acquired from a 41 year male volunteer (TR 100ms, TE 0.2 ms, flip-angle 90°, 3 averages, 12.5 minutes).

Results: Larger gradient waveform magnitudes at the start of each CA-TPI trajectory **(Figure 1b)** are associated with increased initial rates of radial evolution (**Figure 2a**), resulting in beneficial transfer function reshaping (**Figure 2b**). This in turn yields 7-15% smaller MEOV values for the CA-TPI trajectories when compared to standard TPI. In addition, MEOV values are 4-27% less for the 20 ms (vs. 5 ms) trajectories, with greater advantage for higher accuracy measurement requirements. Representative CA-TPI images of human brain demonstrate good image quality for both short and long readouts with an SNR ~18 **(Figure 3).** The 20 ms readout also shows reduced smearing of the slow-decaying CSF, which is a result of smaller voxels for the same SNR and scan time as the 5 ms readout.

Discussion: Centre-Accelerated TPI yields good quality ^{23}Na images of human brain on a Siemens 3T, and its advantage is the facilitation of signal quantification in smaller objects than standard TPI. This advantage, described the MEOV metric, will be greater for tissues with quicker signal decay (e.g. cartilage) and at higher field strengths where minimum ρ values are further constrained by smaller voxel dimensions (i.e. gradient slew limitations). Further TPI optimization is also warranted, particularly within the context of readout duration.

References: [1] Boada, MRM, 1997; [2] Stobbe, MRM, 2017; [3] Nagel, MRM, 2009

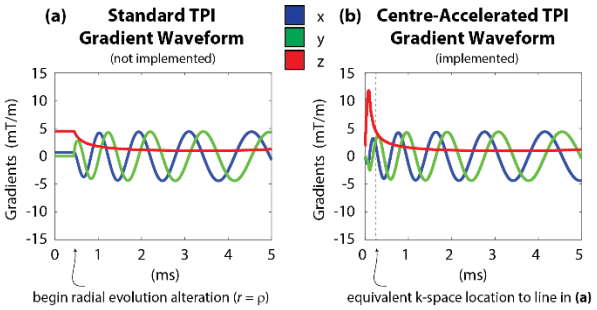

Figure 1: Representative **(a)** standard and **(b)** Centre-Accelerated TPI gradient waveforms for the 5 ms (ρ = 0.25) readout duration.

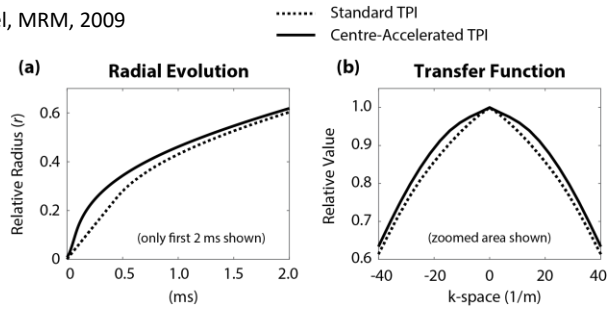

Figure 2: (a) CA-TPI trajectories advance from the centre of k-space more rapidly that standard TPI. **(b)** This pushes the rapid biexponential signal loss to greater k-space locations, reducing the signal decay rate across k-space at the centre of the transfer function.

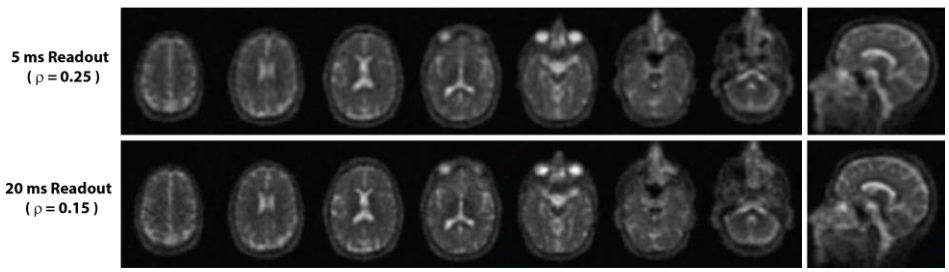

Figure 3: Centre-Accelerated TPI images acquired on the Siemens Prisma 3T in 12.5 min per scan. The SNR in white matter is ~18 for both. Smaller bottom row voxels (4.35 mm vs. 5.5 mm isotropic) are facilitated by longer readouts (for constant SNR), yielding reduced CSF signal smearing.