# Imaging without Gradients: First In Vivo MR Images using the TRASE RF Imaging Method

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**Introduction:** TRASE (Transmit Array Spatial Encoding) is a novel MRI acquisition scheme in which high-resolution k-space encoding is performed by the transmit RF fields, thus eliminating the need for the entire switched B0-gradient system (1). The electronics becomes simplified and, since only a single transmit channel is used, there is only 1 power amplifier in the whole system (the RF power amplifier), unlike some other RF encoding methods that require multiple transmit channels (2). The aim of the project is to increase the accessibility of MRI to wider patient populations by developing a simplified and lower-cost imaging technology. The technique is well-suited to low- and mid-field MRI using permanent magnets as: 1) SAR regulations are not restrictive, and 2) magnet operating and maintenance costs are negligible.

*Imaging Principle*: A linear B1 phase-gradient field can be represented by a fixed point in k-space (the 'field focus'). Spin-refocusing by a 180deg pulse, generated using the phase-gradient field, results in a k-space point-reflection about the coil focus. By alternately refocusing with 2 different phase-gradients, a 1D k-space trajectory is played out. In this way k-space trajectories can be built-up using an echo-train. Two different phase-gradient fields are required per spatial encoding axis (1).

*FOV & Spatial Resolution:* In 1D the k-space point spacing is $2\Delta k$, where $\Delta k$ is the k-space separation of the 2 field foci. So FOV = $1/(2\Delta k)$. For an echo train with NE echoes, the pixel size is FOV/NE, i.e. $\Delta x = 1/(2\Delta k\ NE)$. Maximum spatial resolution is therefore much higher than other RF imaging methods as there is a "Resolution Amplification Effect" due to spins being repeatedly refocused down the echo train. For the coil shown here (4.1deg/cm), NE=128 corresponds to $\Delta x$=1.7mm.

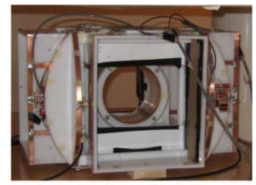
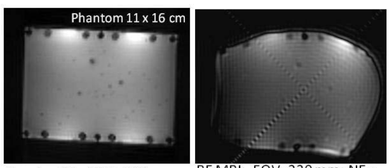


FIG 1. 2-axis RF phase gradient coil array

**Methods:** Experiments were performed at 0.2 Tesla using an NRC TMX MRI Console, adapted to provide PIN diode control. The transmit array (Fig.1) was designed to generate 4 different B1 phase gradients (+X, -X, +Y, -Y) to allow 2-axis encoding. The phase gradient is selected by PIN diode controlled circuitry. The X & Y encoding arrays are identical, but just rotated 90deg with respect to one another. The coil design was optimized for uniformity of |B1|, linearity and strength of phase gradient. Aspects of the design have been described previously (3). Signal reception was achieved using a single-channel solenoid.

*Sequence:* The 2D-encoding sequence consists of (2NE-1) echo trains, with each train containing NE echoes. Refocusing pulses were 500us square pulses, with echo spacing 1.1ms or 1ms. Acquistion duration = (ntrains*npartitions*TR). For Figs.3&4 B0-phase-encoding was used to encode the 3rd dimension.



Regular GE: FOV=250mm; matrix=128; pixel= 1.95mm, (projection)

RF MRI: FOV=220mm; NE= 128; pixel=1.7mm, Phase Gradient= 4.1deg/cm (projection)

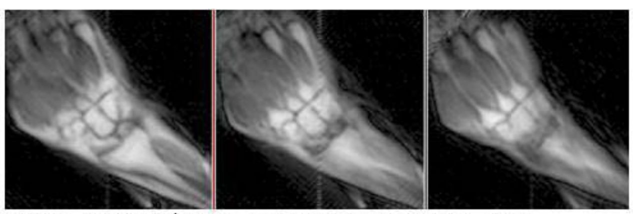


**FIG.3A** RF Wrist/Hand: NE=128; echo spacing 1.1ms; TR=500ms; partitions=8, width=7.5mm

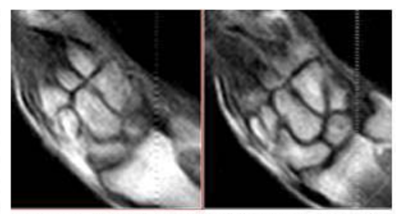


**FIG.3B** RF Wrist: NE=200; TR=500ms; partitions=8, width=5mm

**Results:** For all RF images shown, in-plane encoding is performed by RF encoding. Figure 2 shows a 2D projection GE image, compared with a 2D TRASE image. For this long T2 sample the resolution of the RF image is only limited by the pixel size. Figure 3 shows several partitions from two RF-encoded 3D datasets. Figure 4 shows one partition of an RF-encoded coronal knee image. All RF images show geometric distortions at the limits of the RF coil useable imaging volume, due to |B1| inhomogeneity and phase gradient non-linearity. No wrapping artifacts occur in the RF-encoded directions, (even though TRASE RF imaging is a form of phase-encoding). Un-encoded signal does not smear, but occurs as line artifacts at the FOV edge (visible in the top-left of Fig.4).

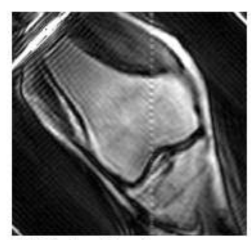


**FIG 4**. In vivo knee. NE=128, partitions=16, width= 6.25mm

**Discussion:** Phantom results show the resolution expected, based on the k-space coverage, thus validating the RF encoding technique and the potential for high resolution. In these in vivo experiments the spatial resolution is primarily limited by T2 decay down the echo train. Results (not shown) indicate that shorter RF pulses are obtainable, which will increase in vivo resolution. The self-limiting FOV behavior of the RF coils is beneficial in preventing the occurrence of aliasing artifacts (see knee images). So both spatial resolution and FOV are controllable through RF coil design. Although in these experiments the 3rd dimension ('slice direction') was encoded using conventional B0-phase-encoding, the presented results are reasonably representative of what may be expected from future full 3-axis RF encoding. This is because the behavior of the 1D B0-phase-encoding is fairly similar to 1D B1-phase-encoding (in terms of acquisition time, resolution, FOV). Acquisition times can be long for this phase-encoded imaging technique (e.g. 27mins for Fig.4); however the method is expected to be compatible with standard parallel imaging acceleration techniques in all 3 axes, which would reduce acquisition times by an order of magnitude.

**Conclusion:** We have demonstrated the first in vivo 2D imaging with the TRASE RF imaging technique. Results are encouraging. Next steps are the extension to full 3D RF-encoding and coil design improvements to further increase FOV.

**References**: (1) Sharp-JC, King-SB MRM 63, p.151 (2010); (2) Katscher-U MRM 63 p.1463 (2010), (3) Deng-Q ISMRM (2010);