Quasi-static traveling wave imaging on a clinical 3T MRI system

Alexey A Tonyushkin[1,2], Norman B Konyer[3,4], Michael D Noseworthy[3,4], and Andrew JM Kiruluta[1,2]

[1]Radiology Dept., Massachusetts General Hospital, Harvard Medical School, Boston, MA, United States, [2]Physics Dept., Harvard University, Cambridge, MA, United States, [3]Imaging Research Centre, St. Joseph's Healthcare, Hamilton, Ontario, Canada, [4]Electrical and Computer Engineering Dept., McMaster University, Hamilton, Ontario, Canada

Introduction

Conventional MRI relies on near-field inductive coils that require complicated design and most importantly high RF deposition power at the surface to achieve usable, relatively uniform field pattern in the deep region of interest. Currently, multichannel arrays of near field receive-only coils are offered for such imaging for clinical 3T MR scanners. In contrast, we describe a new MRI modality, namely quasi-static field (QSF) MRI that relies on RF transmission and reception by coupling to a metal waveguide (the scanner bore) in a far-field regime. Initially a traveling wave (TW) mode is excited in the bore but, since 127.8 MHz (3T) is below the frequency cut-off for TW propagation, this mode decays relatively quickly [1]. An empty bore RF wavelength at 3T is 2.34m, and a decay constant of such a TW is on the order of a meter, far in excess of currently accessible maximum FOV with conventional near-field implementations. TW MRI can be beneficial as it avoids large RF energy deposition at the body surface and instead channels RF power into a waveguide and then radiatively couples it into the region of interest. While TWs at field strengths greater than 7T were reported for an empty bore system [2], most clinical scanners are not generally accessible to TW due to cut-off requirements at field strengths up to 3T. To satisfy TW cut-off requirements at fields less than 4T one has to incorporate a coaxial waveguide [3], which would obstruct the field of view, or use dielectric structures [4]. Here we present results of QSF TW propagation and its coupling to biological materials in an unmodified 3T clinical scanner (GE Waukesha, Wisconsin).

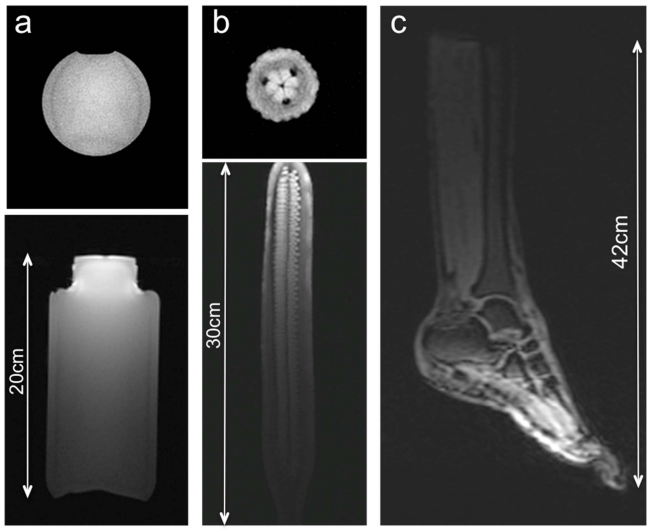

Figure 1: *Quasi-static RF imaging at 3T: a) Axial and coronal slices of 0.9% saline solution bottle; b) axial and coronal slices of cucumber; c) coronal image of a human leg (coils ~5 cm from the foot).*

Methods and Materials

In our imaging experiments we used an RF shield (inside a 60 cm diameter bore), incorporated into the bore as a cylindrical metal waveguide, and a quadrature T/R dual loop coil (D=15 cm) as a probe for a mode coupling into/from the guide. Such an under-moded cylindrical waveguide allows propagation of a complex TE11 mode inside a scanner's bore [1]. The B_1 field of a mode couples into a dielectric phantom or sample, which is placed into the isocenter and few cm away from the probe. Due to the large RF wavelength (λ=234cm at 127.8 MHz) a resonance patch antenna [6] is not practical to excite such a waveguide. Therefore, we used two orthogonal loop-coils [7] as a probe to excite and receive TW modes in a waveguide. The coils were tuned inside the bore and driven in quadrature to provide the required circularly polarized B_1 field that couples into the waveguide.

Results and Discussion

To demonstrate imaging capabilities of our approach we used various samples as shown in Fig. 1: (a) 2 liter bottle filled with 0.9% saline solution, (b) cucumber, and (c) volunteer's leg. The RF probe was placed at the entrance to the bore and at least 5 cm away from the imaging volume. We also tested larger (up to ~50cm) separation of the bottle phantom from the probe that is limited by the imaging gradients size and still obtained images with reduced SNR. Data were acquired using a standard GRE pulse sequence: TR=100ms, TE=6.9ms, flip angle=45^0, 256x256, rbw=15.63kHz, 5mm slice, 8NEX, 3:28 min scan time (7 slices) for images in Fig. 1 a, b and 3D-FGRE, TR=10ms, TE=4.7ms, flip angle=30^0, rbw=15.63kHz, 2mm slice, FOV=480x240, 256x128 matrix, 6NEX, 4:30 min scan time (66 slices) for Fig. 1 c. The T/R coils were carefully tuned in the bore to 127.8 MHz, 50Ω impedance-matched, with a return loss of approximately -17dB and -26dB for the two coils. As can be observed from the MR image of the leg (Fig. 1 c) there is a star-like brightening artifact at the foot. It is hypothesized that one, or both, of the following, cause this artifact: aliasing of the spatially unencoded upper part of the leg that is lying beyond the gradient coils, or the proximity of the coils to the foot. Most of the volunteer's body is outside of the 480mm FOV and not encoded, but still generates significant signal from the guided RF excitation. This artifact diminishes with the increased probe-to-foot distance since the upper part of the leg is exposed to a more significantly decayed TW RF excitation. Further experiments are required to fully explain this effect. The other feature of the images is non-uniformity in SNR along z-axis, which arises due to the decaying nature of the complex waveguide mode. To remedy this effect one could apply a spatial filter during the processing stage. The more advanced and promising direction for RF field control, however is to use specially designed dielectrics inserted into the bore that modifies the B_1 field profile to make it more uniform [5].

Conclusions

We have demonstrated imaging with QSF concept using an unmodified clinical 3T MRI scanner as an under-moded RF waveguide. The MRI images of a vegetable and human leg are comparable to the ones obtained with TW concept at 7T MRI scanners [2]. Unlike real TW propagation at ultra-high fields, here we do not have a standing wave due to the natural damping of RF along the bore. This work may result in more robust and fast imaging in clinical settings as well as new applications in deep tissue imaging of hard to reach areas. Future directions include designing probes for high efficiency excitation of QSF in the bore, and RF field profile control with (meta-materials) dielectrics.

References: [1] Tonyushkin A. and Kiruluta AJM, Proc. ISMRM 19, 2011; [2] Brunner D.O. et al., Nature 2009, 457: 994-999; [3] Alt S, et al., Proc. ISMRM 18, 2010; [4] Brunner D.O. et al., MRM 66:290–300, 2011; [5] Tonyushkin A., Konyer N., Noseworthy M., and Kiruluta A., Proc. ISMRM 20, 2012; [6] B. Zhang, G. Wiggins, Q. Duan, D. Sodickson, Proc. ISMRM 17, 2009; [7] Webb A. G., et al., MRM 63:297–302, 2010.