RF shield parallel-plate waveguide for travelling-wave MRI experiments at 3 T

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Abstract. Ultra high field magnetic resonance imaging systems suffer from the standing wave problems produced by conventional radiofrequency coils. This causes inhomogeneous fields degrading the image quality. To overcome this problems, propagation of RF waves by antennas inside a waveguide is suggested. We assessed the feasibility of using the magnet RF shield with aluminium strips to form a parallel-plate waveguide for travelling-wave magnetic resonance imaging at 3 T. Imaging results showed that a number of constraints must be solved before good image quality can be obtained. A brief discussion is presented on the possible sources or error and interference. Despite these limitations, results are really encouraging to continue investigating the implementation of this approach.

1. Introduction

Ultra high field Magnetic Resonance Imaging (UHF-MRI) offers an increase in signal-to-noise ratio (SNR), higher sensitivity, greater spectroscopic resolution, and improved contrast. The RF wavelength and the dimension of the human body are comparable in size, causing standing wave patterns generated by standard RF coils. This effect degrades the MR image and the appearing as regional signal losses and perturbing the tissue contrast.

The traveling-wave MRI (twMRI) approach offers the potential to overcome some of these constraints. The remote excitation of the RF signal requires a waveguide to guide the RF energy to the object to be imaged. At conventional clinical field strength (< 4 T), because of the relatively small bore diameter the twMRI faces various constraints: critical frequency requirements for wave propagation prevent the usage of a typical scanner at low field. This has implications on the type of waveguide and its dimensions to adequately conduct the RF signal. Various types of waveguides have been successfully used to generate MR images using whole-body MR imagers [1-2].

It is well known that twMRI is best suited for high and ultra high fields. However, we have previously shown that twMR with a parallel-plate waveguide (PPWG) is feasible at magnetic field intensities lower than 7 T and whole-body systems [3]. These results showed that the RF shield of the imager and the embedded whole-body RF coil did not cause any major interference, and good quality images of a human leg were acquired. The PPWG has two important advantages: a) a very simple configuration and
b) its critical frequency is zero for the dominant mode. This implies that all frequencies propagate regardless the separation of the conducting plates, so it can be used in any MR imager. We have also demonstrated that twMRI is feasible at different resonant frequencies with small-bore MR imagers [4-5].

Initial twMRI experiments mainly and the RF shield of the gradient system and surrounding cryoshield of the magnet for propagation of the RF signal [1-2, 6]. These shields have a circular cross section forming a cylinder. Our group made previously acquired images of healthy leg using these shields and two commercial surface coils at 3 T [7]. In this paper, we studied the use of a PPWG using the cylindrical RF shield of a clinical MR imager to acquire twMR images of a phantom. We considere that the shield and the aluminium strip form a PPWG with critical frequency equal to zero and allowing the propagation of all frequencies.

2. Methods
A parallel-plate waveguide was formed using aluminium foil strips with a thickness of 0.016 mm with a 30 cm width. The two strips were 1.65 m long to fully cover the entire magnet length. Extra precautions were taken to avoid blocking those temperature sensors inside the magnet. Then, the strips were mounted on the tunnel at the same height, as shown in Fig. 1.b). Imager sensors controlling the temperature inside the magnet tunnel were not blocked by the strips, to avoid the interruption of the image acquisition. To transmit the RF signal, a flat coil array with 8 elements and a volume 64-element coil array were used for transmission (Tx) and reception (Rx), respectively. The coil arrays were 1.75 m away from each other and were placed on the patient bed at opposite sides, as shown in Fig. 1.a). It is worth noting that the maximum field of view (FOV) is 50 cm x 50 cm x 45 cm, and the coil arrays were placed outside this FOV.

![Figure 1](image-url)

**Figure 1.** Experimental setup of the twMRI experiments showing coil arrays and their location (a), and the aluminium strips forming the PPWG inside the magnet (b).
A cylindrical phantom was located at the middle point of these coil arrays to obtain images. We used a provider vendor cylindrical MR phantom 5300 ml (10606530 K2305) and (1P) Model number: 10606530: 1000 g H2O distilled: 3.75 g NiSO4 x 5 H2O + 5 g NaCl. Fig. 1.a) shows the experimental setup. The patient bed had an embedded coil array for spine imaging and stayed in the magnet. The imager was not modified in any way and all imaging experiments were conducted according to the scanner-provider protocols for image acquisition.

All phantom imaging experiments were run at 3 T in a clinical MR imager (Magneton Skyra, Siemens Healthcare GmbH, Erlangen, Germany). Phantom images were acquired with gradient echo (GE) sequences and the following parameters: TR/TE=3.15/1.37 ms, Flip Angle = 90°, FOV = 240x120 mm², matrix = 160 x 160, thickness = 1.6 mm, NEX=1.

3. Results and Discussion

Fig. 2 shows a phantom axial and sagittal images acquired with the gradient echo sequences optimized for brain imaging. Images show low signal-to-noise ratio and granular noise.

![Figure 2. Axial (left) and sagittal (right) images of a cylindrical phantom acquired with the PPWG mounted on the magnet tunnel.](image_url)

This is probably due to: a) the Rx coil arrays was located right outside the maximum FOV, b) no dielectric material was added to better conduct the RF waves inside the PPWG, c) the image reconstruction scheme was not working correctly, d) the phantom size was not the adequate one, e) the Tx coil array was designed to couple the reactive near field of the sample, but it was use for remote excitation and, f) the coil array in the patient bed may cause interference with the RF signal.

The MRI system was built to obtain images within a specific region of the scanner. This location is usually the isocenter of the magnet. This means that the coil arrays and the object to be imaged must be at the exact same point. Both coil arrays were at least 45 cm away from the isocentre. This may cause that the image reconstruction scheme to fail, despite the fact that the cylindrical phantom was at the isocentre. The optimized acquisition protocol of the image demands an specific voltage and current for imaging an sample, which is not particularly the same for remote excitation.

As reported by Black and Richie [8], dielectric materials improve the RF wave propagation and signal stability. But in this study, the phantom size was far too small to positively contribute to the good RF signal transmission. In addition, no dielectric materials were included in the experiments. Poynting vector simulations of a PPWG filled with air at 3 T shows a standing wave pattern [9], causing low uniformity and stability of the principal mode magnetic field. This has also been observed in twMRI experiments at 15.2 T with a small-bore MR imager [5].
This PPWG was formed with two conductors, the aluminium strips and the RF shield, which is not a standard design for a waveguide. The interaction of the electromagnetic energy guided in the tunnel with the both the aluminium strips and the metallic cylinder may show a distinctly pattern compared to the one observed in previous experiments [3-5].

Another important source of interference is the patient bed and the spinal coil array inside as reported in [1]. In our experiments, we kept the bed in the magnet tunnel, because of the RF unit to interface the coil arrays is mounted on it. So, there is no other way to actually connect them. Clinical MRI systems are designed to optimally operate to meet the demands in the clinical environment.

The interaction of the travelling wave with the different metallic objects such as patient bed, and the RF shield, and the spinal coil array require a more detail study to propose alternative ways to attenuate these effects. The RF coil arrays for remote excitation of the RF fields also need a further investigation to look into those important aspects to extend their capabilities. Griffith and Pan have derived analytical expressions for the far-zone using circular coils [10-11], that it may be useful to develop coil arrays with characteristics allowing to achieve the double purpose of acquiring imaging close to the sample and remote excitation. A number of important aspects have to be taken into account to improve the quality of the images acquired with this approach.

4. Conclusions
We have experimentally demonstrated that the PPWG made out of aluminium strips mounted on the RF shield of the imager can transmit the RF signal for relatively large fields of view. Although, imaging results show low SNR and noise, it is encouraging to continue testing this idea about acquiring images with larger FOVs at lower fields. These preliminary results showed the feasibility of this waveguide to obtain images at 3 T with a clinical MR imager. We intend to run twMRI experiments with a clinical MR imager with minimal modifications to its present configuration. These results pave the way to a further implementation of the travelling-wave approach for clinical applications.

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5. References
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