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Pulse sequence considerations for simulation and postimplant dosimetry of prostate brachytherapy

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ABSTRACT

PURPOSE: The purpose of this work is to present a brief review of MRI physics principles pertinent to prostate brachytherapy, and a summary of our experience in optimizing protocols for prostate brachytherapy applications.

METHODS AND MATERIALS: We summarized essential MR imaging characteristics and their interplays that need to be considered for prostate brachytherapy applications. These include spatial resolution, signal-to-noise ratio, image contrast, artifacts, geometric distortion, specific absorption rate, and total scan time. We further described the optimization of the protocols for three pulse sequences: three-dimensional (3D) fast-spoiled gradient echo sequence for T1-weighted imaging, 3D fast-spin echo sequence for T2-weighted imaging, and 3D fast imaging in steady-state precession sequence for combined T1 and T2-weighed imaging. The utilization of an endorectal coil was also described.

RESULTS: Using the optimized protocols, we acquired high-quality images of the entire prostate within 3−5 minutes for each sequence. These images display the desired image contrasts and a spatial resolution that is equal to or better than 0.59 mm × 0.73 mm × 1.2 mm. While 3D fast-spoiled gradient echo sequence and 3D fast-spin echo sequence depict radioactive seed markers and anatomic structures separately, 3D fast imaging in steady-state precession sequence demonstrates great promise for imaging both seed markers and prostate anatomy simultaneously in a single acquisition.

CONCLUSIONS: We have optimized current MRI protocols and demonstrated that the anatomic structures and positive contrast radioactive seed markers for prostate post-implant dosimetry can be adequately imaged either separately or simultaneously using different pulse sequences within a total scan time of 3−5 minutes each. © 2016 American Brachytherapy Society. Published by Elsevier Inc. All rights reserved.

Keywords: Prostate brachytherapy; Radioactive seeds; MRI protocols; Pulse sequences

Introduction

In low-dose-rate (LDR) brachytherapy (1), a high dose of radiation is applied at a low dose rate to cure or control prostate cancer with radioactive seeds that are needle placed under ultrasound guidance. To deliver the desired dose to the tumor and minimize the damage to surrounding normal tissue, the geometric boundaries of the anatomic structures around the prostate must be determined in treatment planning. After brachytherapy, it is necessary to accurately verify that all the radioactive seeds have been implanted in the targeted locations to ensure that the appropriate dose is applied to the cancer in the prostate to achieve curative intent.

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Imaging plays an important role during both simulation and postimplant dosimetry verification in brachytherapy (2). X-ray CT is the most widely used imaging method and is the current standard of practice because of its ease of use, wide accessibility, and positive contrast of the radio-opaque markers inside the seed casings (3, 4). However, disadvantages of CT include the radiation dose imparted and lack of soft tissue contrast and resulting difficulty in delineation of the prostate and its subglandular and surrounding anatomy (5–7). In comparison, MRI has the potential to overcome both of these limitations and has been actively explored for its use in prostate LDR brachytherapy (8–13). To this aim, two different approaches have been reported. The first is the combined use of CT and MRI, taking advantage of CT seed localization and MRI target delineation. Unfortunately, such an approach requires separate CT and MRI studies for a patient, as well as a need for coregistering the two different types of the images, which is a challenging task. The second approach is MRI only, using either different pulse sequences or a single sequence for both seed localization and target tissue delineation (14). Because the radioactive seeds do not produce MRI signals themselves, it can be difficult to conclusively identify their locations on MRI, especially in the presence of other similar image features, such as needle tracks. In this regard, positive MRI contrast marker technology, in which an MRI signal-producing solution is encapsulated and placed as the spacers between the adjacent radioactive seeds, can be especially helpful and has been shown to be feasible (15–17).

Unlike with CT, optimization of MRI images often necessitates a thorough understanding of MRI physics and careful consideration of the different acquisition parameters that are intimately intertwined. The optimization of the acquisition parameters is also dependent on the desired image property and application. For diagnostic purposes, MRI has become widely used and is now fairly standard for the detection and evaluation of prostate cancer (18). However, the technical requirement and protocol can be quite different when MRI is used for postbrachytherapy implant evaluation. The complications and confusion are further exacerbated by the variations in vendors and platforms (hardware and software). We provide a brief overview of the basic underlying MRI physics as well as a summary of our experience with using MRI for simulation and postimplant dosimetry of prostate brachytherapy.

Basic physics considerations

MRI relies on creating a longitudinal magnetization of the mobile protons in the tissue by placing the body in a strong homogeneous magnetic field; creating signal-producing transverse magnetization by applying a radiofrequency (RF) excitation pulse with a transmit RF coil; encoding the spatial locations of the signals with a set of linear magnetic field gradients; detecting the encoded signals with a receiver RF coil; and decoding the signals into an image of the body with image reconstruction (19). Each of these components can separately or jointly affect the quality of the images that are generated. Some of the main image characteristics that need to be considered in prostate brachytherapy applications include spatial resolution, signal-to-noise ratio (SNR), image contrast, artifacts, geometric distortion, specific absorption rate (SAR), and total scan time.

Spatial resolution

Spatial resolution is a measure of the smallest high-contrast object that can be detected without substantial partial voluming (i.e., a mixture of different types of tissues falling into a single voxel). In MRI, the spatial resolution along a given direction is often calculated as the field of view (FOV) along the direction divided by the corresponding matrix step. In the case of two-dimensional (2D) imaging, the slice resolution is simply the slice thickness defined by the slice selection RF pulse and gradient. Note that the reconstructed matrix size may be different from the acquired matrix size when certain user-selectable options are used, such as partial Fourier (e.g., fractional echo along the frequency-encoding direction and a fractional number of excitations along the phase- or slice-encoding directions) or zero padding. The corresponding reconstructed voxel size can only be considered a measure of the apparent spatial resolution. Further, there are other factors that may change the true spatial resolution of an image, including any k-space signal modulations (e.g., the T2* decay during an acquisition window, T2 signal decay along the echo train in a fast or turbo spin echo, or any k-space filters that may be applied for image reconstruction).

Signal-to-noise ratio

SNR is a measure of the graininess of an image relative to the true signal. Figuratively, SNR is, to some extent, the ‘‘currency’’ in MRI; it can be traded for many other desirable attributes of an image, such as the spatial resolution and imaging speed. The SNR in MRI can have different expressions, but fundamentally, it can be expressed as follows (19):

$$\mathrm{SNR} = k\,(\text{voxel volume})\sqrt{T} \qquad [1]$$

Here, T is the total sampling time. k is a proportionality constant that accounts for factors such as the receiver coil sensitivity, mobile proton density, and available magnetization that is used for imaging. The latter is dependent on the static magnetic field strength B0, and the imaging sequence that is used to manipulate the magnetization. For example, a long TR (repetition time) and short TE (echo time) spin echo sequence uses up nearly all the available magnetization for producing proton density-weighted images,

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