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Pulse sequence considerations for simulation and post-implant dosimetry of prostate brachytherapy

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Abstract

High-resolution, high-contrast images help facilitate target delineation and provide accurate localization of the radioactive seeds in the simulation and post-implant dosimetry of prostate brachytherapy. Magnetic resonance imaging (MRI) offers many potential advantages for this application over the current standard of practice of computed tomography (CT). However, MRI is technically complex and has many more user-selectable scan parameters, which can impact image quality, than does CT. In this paper, we provide a brief review of some basic MRI characteristics and the relationships among them that are relevant for its application in prostate brachytherapy. We further present our experience and considerations in optimizing the protocols of three different commercially-available pulse sequences for acquiring images with T1-weighted, T2-weighted, and combined T1- and T2-weighted contrasts: three-dimensional (3D) fast-spoiled gradient echo (FSPGR), 3D fast-spin echo (FSE), and 3D fast imaging in steady-state (trueFISP). We demonstrate that while 3D FSPGR and 3D FSE can be used for imaging radioactive seed markers and anatomic structures separately, 3D trueFISP holds promise for imaging both simultaneously in a single acquisition.

Keywords

prostate brachytherapy; radioactive seeds; MRI protocols; pulse sequences

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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.

Imaging plays an important role during both simulation and post-implant dosimetry verification in brachytherapy (2). X-ray computed tomography (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 co-registering 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 post-brachytherapy 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 post-implant 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 [NEX] 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).

SNR

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 of an MRI can have different expressions, but fundamentally, it can be expressed as follows (19):

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

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 B_0 , 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, while a short TR and

short TE spin echo sequence saturates tissues with long T1 and produces T1-weighted images for which the signal intensity is proportional to $(1 - e^{-\frac{TR}{T1}})$.

The second term in Eq. [1] characterizes the size of an image voxel and is given by:

$$(\text{voxel volume}) = \Delta_x \Delta_y \Delta_z = \frac{(FOV)_x (FOV)_y (FOV)_z}{N_x N_y N_z} \quad [2]$$

For 2D imaging, the voxel size along the slice direction is simply the slice thickness. Note that N_x , N_y , and N_z may not be identical to the number of acquired data points or lines. For example, when partially parallel imaging along the phase-encoding direction with an acceleration factor of 2 is used, N_y in Eq. [2] will be two times the acquired number of phase-encoding lines.

Because it is random and incoherent, noise decreases as a square root of T relative to the signal, as expressed by the last term in Eq. [1]. T can be expressed as a function of the other imaging parameters, as follows:

$$T = \frac{N_x N_y N_z (NEX)}{(bw)S} \quad [3]$$

in which bw is the receiver bandwidth per frequency-encoding or readout point, and S is a sparsity factor that accounts for the scan time reduction when certain acceleration strategies, such as partially parallel imaging or partial Fourier acquisitions, are used.

Total scan time

The total time for an MRI scan can be expressed as:

$$T_s = \frac{N_y N_z (NEX)(TR)}{(ETL)S} \quad [4]$$

in which ETL is the echo train length that accounts for the scan time reduction when multiple echoes are acquired after one RF excitation, such as in a fast/turbo spin echo sequence or an echo planar imaging (EPI) sequence. S is the same sparsity factor as in Eq. [3] that may be used for scan time reduction. For 2D imaging, N_z is equal to 1, but the number of slices that can be acquired within the same time, as given by Eq. [4], is determined by the ratio of TR over the sequence time (defined as the time to play out all the gradient and RF pulses for each slice).

Image contrast

Image contrast relates to the signal difference between two different types of objects (e.g., tumor relative to the surrounding normal tissue). In MRI, many tissue properties (e.g., T1,

T2, proton density, diffusion, and blood flow) can be exploited to generate images of a desired contrast. In general, image contrast is not only determined by the intrinsic property of the tissues (e.g., T1 and T2) but also by the extrinsic set-up and the parameters used for imaging (e.g., pulse sequence type and scan protocol).

Artifacts

Artifacts are any image abnormalities that deviate from or do not represent the actual underlying structure of the object. While MRI is well-known for its rich and excellent soft tissue contrast, it is also highly prone to many types of artifacts. Artifacts in MRI can arise from a patient's body (e.g., motion, chemical shift of fat, or magnetic susceptibility caused by a metal implant (20) or tissue-air interface), imperfections or failures in hardware and software, or even improper selection of a scanning protocol. Involuntary rectal peristalsis in the setting of endorectal coil use for prostate MRI can generate strong repetitive circular artifacts along the boundaries of the coil, which can limit the ability to detect and adequately delineate the boundaries of tumor in the adjacent peripheral zone. The severity of the image artifacts often depends on the severity of the underlying causes but may be mitigated with proper imaging setup and careful selection of the imaging technique and scanning protocol. For example, respiratory motion artifacts can be minimized with respiratory triggering, navigators (21), or non-Cartesian sampling, such as Propeller/Blade sequences (22). They can also be minimized by patient breath hold and fast imaging acquisitions. Another common MRI artifact is the signal wraparound that occurs when the FOV selected is smaller than the actual dimension of the body along the phase-encoding direction.

Geometric distortion

Compared to CT, MRI may be more susceptible to geometric distortion and spatial inaccuracy (2,23). Most conventional MRI techniques rely on the assumption of a perfectly homogeneous magnetic field and spatially linear magnetic field gradients. Non-linearity of the gradients is unavoidable due to the physical constraints of the gradient coil design and can lead to geometric distortion, especially at locations far away from the scanner iso-center. Fortunately, the spatial distribution of the non-linearity of the gradients is preset for a given system design, and most of the resulting geometric distortion errors can be corrected in image reconstruction software provided by the vendors. Magnetic field inhomogeneity, on the other hand, can be patient and anatomy dependent. As a result, the related geometric distortion requires determination of the field map or the use of special pulse sequences. The most noticeable effect of geometric distortion due to magnetic field inhomogeneity is along the frequency-encoding direction. The amount of spatial distortion can be expressed as:

$$\delta = \frac{\Delta B_0}{\gamma G_x} \quad [5]$$

in which B_0 is the local magnetic field inhomogeneity, γ is the gyromagnetic ratio, and G_x is the magnitude of the frequency-encoding gradient. Therefore, geometric distortion can be effectively reduced by using a higher readout gradient for a given amount of magnetic field inhomogeneity. However, the maximum gradient that can be used is limited by a system's

hardware and an increased readout gradient requires increased receiver bandwidth, which will result in a lower SNR (see Eqs. [1] and [3]). Another factor that can contribute to geometric distortion is eddy currents that are induced by gradient switching, especially in images from certain types of pulse sequences, such as EPI.

SAR

SAR is defined as the RF power absorbed by a unit mass of tissue. It is measured in watts per kilogram (W/kg). The International Electrotechnical Commission and the US Food and Drug Administration have defined several levels of SAR limits to prevent tissue from heating during MRI of different anatomic areas (e.g., the current limits for the normal operation mode and the first operating mode of whole body imaging are 2 W/kg and 4 W/kg, respectively). It should be noted that the value of SAR increases with the square of the magnetic field strength and with the square of the RF pulse flip angles. As a result, SAR can be a limiting factor in the performance of certain fast or RF-intensive pulse sequences, especially at a higher field strength such as 3 Tesla.

SCANNING PROTOCOL CONSIDERATIONS

Ideally, MRI should have the highest SNR and spatial resolution, the highest image contrast, the fewest artifacts, the least geometric distortion, and the shortest total scan time. However, the different parameters of a scanning protocol that determine these imaging characteristics often conflict each other in basic physics, as described above, and need to be carefully traded off in each application. For example, an increased number of NEX may be necessary to compensate for the loss of SNR with high spatial resolution imaging. However, the corresponding increase in scan time may exacerbate motion-related ghosting and image blurring, leading to poor image quality and loss of spatial resolution. Here, we present our experience in optimizing the different pulse sequences and scan parameters that are used for simulation and post-implant dosimetry of LDR prostate brachytherapy. We note that in our practice of LDR prostate brachytherapy, we use stranded seeds with the positive MRI-signal markers placed between every two adjacent seeds (Sirius; C4 Imaging, Houston, TX) (15,16). Visualizing the markers in the images help identify the locations of the seeds with confidence. We also note that we use the two-dimensional (2D) formulism as proposed by the AAPM Task Group 43 for dose calculation (24).

Considerations in imaging set-up

High spatial resolution is required in MRI of prostate brachytherapy; thus, the intrinsic SNR (i.e., the proportionality constant k in Eq. [1]) must be maximized in the imaging set-up. Presently, the magnetic field strengths of the two most widely available clinical MRI scanners are 1.5 Tesla and 3 Tesla. While the latter provides a higher inherent SNR, these scanners are less commonly available than are 1.5 Tesla scanners. In addition, higher magnetic field strength increases the T1 of the tissues (the relaxation time constant of the longitudinal magnetization), the sequence SAR, and the off-resonance effects due to field inhomogeneity and chemical shifts. As a result, additional trade-offs in scan parameters may be needed for image quality optimization. Therefore, we have focused our development of the sequences and scanning parameters on 1.5 Tesla scanners. Further, we have chosen to

use a 70-cm wide-bore system because it can better accommodate patients with a large body habitus or claustrophobia than can the more conventional 60-cm bore systems. All the images presented below were acquired on a Siemens Aera scanner operating under the VE11A software (Siemens AG, Erlangen, Germany), although we believe that similar considerations can be taken on scanners of other vendors or platforms to achieve similar image quality.

Besides magnetic field strength, another critical element that affects the image SNR is the receiver coils. Compared to the surface coils that are used for pelvic imaging, endorectal coils have been in wide clinical use for diagnostic prostate imaging because they provide much higher SNR and improved image quality. The most commonly used endorectal coil in prostate imaging is a self-tuned single-channel loop that is enclosed in an inflatable balloon. We have been using an endorectal coil with a rigid two-receiver channel design (25), which has been shown to further improve the SNR compared to the more traditional single-channel balloon coil. In addition, the rigid endorectal coil deforms the prostate similarly to the ultrasound probe that is used to guide brachytherapy seed implantation. Because of its small size, the endorectal coil must be properly placed relative to the prostate for optimal image quality. On the basis of the localizer images, the coil position may need to be adjusted to ensure that the coil covers the entire prostate gland along the superior-to-inferior (S/I) direction and is centered along the left-to-right (L/R) direction (Fig. 1).

T1- and T2-weighted imaging

A myriad of pulse sequences in MRI are available for different applications. The primary interest in prostate brachytherapy imaging is to accurately localize the radioactive seeds and visualize the anatomy. Anatomic definition of the prostate and its surrounding structures is best achieved with T2-weighted images with their excellent image contrast. Since the radioactive seeds do not produce MR signals themselves, direct visualization of the seeds requires that they are tagged with MR signal-producing markers. For this purpose, Frank et al successfully designed markers with cobalt chloride:N-acetylcysteine encapsulated in a polymer capsule (Sirius; C4 Imaging, Houston, TX) (15,16). The C4 markers have a markedly shortened T1 of approximately 100 ms at 1.5 Tesla (26). Therefore, they are best visualized on T1-weighted images.

Traditionally, high-resolution T1-weighted and T2-weighted images are acquired with gradient echo- and spin echo-based pulse sequences, respectively. Most of the sequences can also be performed in either 2D or 3D mode. For brachytherapy, 3D imaging is preferable because it uses phase encoding for slice definition and provides contiguous slices that are thinner than what is achievable by 2D imaging. The latter is limited by the non-ideal slice profile of a slice-selective RF excitation pulse. Because of these considerations, we have chosen to use a 3D fast-spoiled gradient echo (FSPGR) sequence to acquire T1-weighted images for the brachytherapy seed makers and a 3D fast spin echo (FSE) sequence to acquire T2-weighted images for anatomical definition.

The 3D FSPGR sequence is an incoherent steady-state technique that uses short TR and short TE for fast imaging and RF phase spoiling to achieve T1-weighted contrast. The steady-state transverse magnetization in the 3D FSPGR sequence is dependent on the TR,

flip angle, and tissue T1. The signal is maximized when the flip angle used is equal to what has been referred to as the Ernst angle (19):

$$\theta_E = \cos^{-1}(e^{-\frac{TR}{T1}}) \quad [6]$$

Of note, the Ernst angle is dependent on T1. Therefore, it should be born in mind that the signal can be maximized with a select scanning protocol of TR and flip angle for only one specific tissue when multiple types of tissues with different T1s are present.

The 3D FSE sequence (27) is a 3D implementation of the widely used 2D FSE sequence (28), in which a train of refocusing RF pulses is used after each RF excitation to encode multiple k-space lines and thus shorten the scan time. Like the 2D implementation, 3D FSE images can be T2 weighted and have the major advantage of being insensitive to B0 and B1 field inhomogeneity compared to gradient echo-based sequences. Some major improvements that have been made in the 3D FSE sequence during the last few years include the use of non-selective refocusing pulses (which helps shorten the inter-echo spacing) and variable flip angles of the refocusing pulses (which help to substantially increase the duration of the usable signal, minimize blurring due to signal modulation during the echo train, and reduce the sequence SAR) (29).

Regardless of the sequence type, the total scan time for each acquisition should be kept between 3–5 minutes. Our experience in scanning the brachytherapy patients is that image quality can be highly inconsistent and severely compromised with longer scan times because of the spatial resolution requirement and patient motion. From a spatial resolution and SNR perspective, we have chosen an FOV of 15 cm and a frequency-encoding step of 256, yielding a pixel size of 0.59 mm along the frequency-encoding direction. The frequency-encoding direction is placed along the anterior/posterior direction for all imaging sequences to prevent the motion artifacts along the phase-encoding direction from smearing onto the prostate. The readout bandwidth is selected as the maximum that is allowed by the scanner's gradient performance to minimize the potential geometric distortion. On the 1.5 Tesla Siemens Aera scanner, the bandwidth we use is approximately 560 Hz per pixel.

Because the phase-encoding direction is along left/right, phase oversampling needs to be used to avoid signal wraparound artifacts. Note that phase oversampling effectively increases the FOV along the phase-encoding direction and the associated phase-encoding step N_y . Therefore, it has a penalty of a corresponding increase in the total scan time (Eq. [4]). The coverage in the slice direction is dictated by the size of the prostate along the superior/inferior direction. On the basis of our experience, we have selected a slice thickness of 1.2 mm for adequate SNR, and the number of slices is between 72–88 and adjusted on a patient basis to provide adequate coverage. A small percentage of minimum slice oversampling (9.1% in our protocol) is also necessary because of the non-ideal slab selection profile along the slice direction.

In addition to the above general considerations, we use a flip angle of 25° in 3D FSPGR to optimize the conspicuity of the radioactive seed markers and the prostate gland. Our protocol for the 3D FSE sequence employs a scheme for varying the refocusing flip angles and strategies for slice and phase encoding that are particularly optimized for T2-weighted imaging (27,29). As a result, an ETL of 120 can be used without reaching the sequence SAR limit or incurring noticeable image blurring. Images of the 3D FSPGR and 3D FSE from a patient are shown in Fig. 2 and Fig. 3, respectively.

Combined T1- and T2-weighted imaging

The ability to visualize the radioactive seed markers on T1-weighted images and the anatomy on T2-weighted images may allow for MRI-only post-brachytherapy dosimetry workflow. However, such an approach still requires the fusion of separate T1- and T2-weighted images from two different acquisitions. Practically, we have also found that it is difficult to achieve the maximum signal for the seed markers and maintain adequate SNR for the prostate (especially towards the anterior portion of the gland, away from the surface of the endorectal coil) with a single T1-weighted protocol.

We investigated the possibility of using a single pulse sequence of fast imaging in steady-state precession (trueFISP) (30) for both seed marker localization and target delineation. TrueFISP is a coherent steady-state imaging sequence with a very short TR (a few milliseconds) and a maximum flip angle. Compared to incoherent steady-state imaging by FSPGR, the gradient moments along each gradient axis are fully balanced within each TR. As a result, signals from all the echo pathways by all the prior RF pulses are coherently superimposed for a maximum SNR. When the TR is much shorter than the T1 and T2 of the tissues, the contrast of a trueFISP image is determined by T2/T1. A disadvantage of the trueFISP sequence is that it is prone to “banding” artifacts in the presence of B0 inhomogeneity. Several variations of the trueFISP sequence, such as the Constructive Interference in Steady State (CISS) for Siemens Healthcare and Fast Imaging Employing Steady-State Acquisition (FIESTA-C) for GE Healthcare (31), have been developed to effectively reduce the artifacts through phase cycling. Fig. 4 shows the CISS images that were acquired from a patient using the protocol shown in Table 1. It can be appreciated that both the seed markers and anatomic structures are well visualized in the same images. Overall, the image SNR is also visibly superior to the images in Figs. 2 and 3, with almost identical spatial resolution.

DISCUSSION

High-quality images are essential for prostate brachytherapy planning and post-implant dosimetry. With some recent development, it has become possible to generate synthetic CT images required for treatment planning from MRI images (32). An MRI-only approach would eliminate the need for the current standard of practice, CT, which involves radiation exposure, necessitates two separate scans and inter-modality image co-registration when used in conjunction with MRI, and demonstrates poor target delineation and high inter-observer variation when used alone (5,7,13). With an understanding of basic MRI physics, our experience shows that images of different contrasts and of sufficient quality can be

obtained using different pulse sequences and optimized scan parameters for prostate LDR brachytherapy planning and treatment assessment.

Several additional MRI techniques may be incorporated to further improve the image quality that has been demonstrated. Partially parallel imaging (33–35), which employs the different coil sensitivity profiles of a multi-channel coil to shorten the acquisition time, has been increasingly used in MRI. Unfortunately, images acquired with partially parallel imaging have reduced SNR due to the reduced sampling time (Eq. [3]) and the coil geometric factor. We use partially parallel imaging only for the 3D FSE sequence—not to reduce the scan time but to reduce the echo train duration and the associated image blurring. Another MRI technique that can be used to reduce the scan time is compressed sensing (36). Unlike partially parallel imaging, compressed sensing achieves scan time reduction by exploiting the sparsity of the imaged objects in a transformed domain, such as that of a wavelet. It does not suffer from the SNR penalty of partially parallel imaging and may be particularly applicable for imaging radioactive seed markers because of the potential data sparsity due to their well-defined shape and distribution.

All our current protocols are based on the use of an endorectal coil, which is crucial for its intrinsically higher SNR. However, endorectal coils require careful preparation and set-up before patients can be placed on the scanner table, and multiple adjustments are sometimes needed to achieve optimal coil positioning. In our experience, the average time for set-up and adjustment is 20–30 minutes. Imaging without an endorectal coil could substantially improve study efficiency and patient tolerance. For post-treatment assessment, the images acquired without an endorectal coil would be likely to more closely represent the anatomy in its natural position and therefore provide a more accurate dosimetry estimation. Fig. 5 shows a comparison of the CISS images from a patient with an endorectal coil (Fig. 5a–c) and with the endorectal coil deactivated (Fig. 5d–f), using the same protocol. Clearly, the images in Fig. 5a–c demonstrate higher SNR and better anatomic definition. However, the images in Fig. 5d–f also provide a reasonable visualization of the implanted seeds and markers, as well as the anatomy. The seeds and seed markers are also visualized on the corresponding coronal and sagittal reformatted images with the endorectal coil on (Fig. 5g–h) and with the endorectal coil deactivated (Fig. 5i–j). Further optimization in the pulse sequence and scanning parameters may allow an MRI-only dosimetry assessment, even without an endorectal coil, after prostate brachytherapy.

The C4 markers used in our study produce a positive MRI signal and are therefore essential to providing a positive identification of the radioactive seeds on MRI. Without the markers, the seeds' appearance may be difficult to distinguish from other dark signal regions, such as needle tracks, spacers, and vasculature. Presently, the concentration of solution in the C4 markers has been optimized for T1-weighted imaging (26,37,38). It is possible that additional optimization of the MR pulse sequences, concentration of the seed markers, or different types of markers will further improve the signals in the different types of images (such as CISS images), providing even better localization of the seed markers and visualization of the anatomy. Other pulse sequences (39,40) under development may also be able to provide accurate and direct localization of the seeds without relying on the signal-generating seed markers.

In summary, MRI has many intrinsic advantages over CT for simulation and post-implant dosimetry of prostate brachytherapy. With a good understanding of MRI physics and related image characteristics, we believe that an optimal imaging protocol can be developed for different pulse sequences. Our results have shown that high-resolution T1- and T2-weighted images for prostate brachytherapy can be readily acquired with the 3D FSPGR and 3D FSE sequences, respectively. Further, we have demonstrated that high quality images of an entire prostate that help visualize simultaneously both anatomy and radioactive seed markers can be generated with an optimized 3D trueFISP sequence in 3–5 minutes of scan time and potentially even without the use of an endorectal coil.

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Figure 1.

Proper positioning of the endorectal coil is verified and adjusted on the basis of the localizer images. a) The coil is placed too high superiorly to the prostate (white line); b) after the adjustment, the coil covers the prostate along the S/I direction; c) the coil positioning was affected by the presence of feces (arrow); and d) repositioning helped center the coil in the left/right direction.

Figure 2.

T1-weighted post-implantation images by the 3D FSPGR sequence, using the protocol in Table 1 at the levels of the apex (a), of the midgland (b), and of the base (c). Note that while the brachytherapy seeds (yellow arrows) appear as foci of low signal themselves, the active markers (white arrows) placed along the strands between the radioactive seeds yield bright signals to help with seed localization. The seeds and active markers are also visualized on the coronal (d) and sagittal reformatted images. Note also that because the protocol can be adjusted to maximize the signal from only one tissue type or material (in this case, the seed markers), the SNR of the anatomy is not optimized and decreases noticeably towards the anterior direction and away from the endorectal coil. Nevertheless, different anatomical structures can be identified (e.g., pubococcygeus muscles as yellow asterisks, obturator internus as red asterisks, anterior fibromuscular stroma at yellow arrowhead on b); anterior rectal wall at white arrowhead on a); and neurovascular bundles at the yellow arrowhead on c).

Figure 3.

T2-weighted post-implantation images using the 3D FSE sequence and the protocol in Table 1 at the levels of the genitourinary diaphragm, which extends from the apex to the penile bulb (a), the midgland (b), and the base (c). Compared to the T1-weighted image in Fig. 2, the image contrast is better for target delineation of the prostate and its surrounding anatomy. However, both the radioactive seeds and the active markers appear as signal voids. Among the different structures that are well-delineated in a) are external urethral sphincter (white arrow), pubococcygeus muscles (white asterisks), and anterior rectal wall (white arrowhead). The normal zonal anatomy is also much better seen than on T1-weighted images, with T2 hyperintensity of the peripheral zone (white arrows on b) and heterogeneous T2 hyper- and hypointensity of the central gland (white arrowhead on b). The latter is the result of co-existent stromal and glandular foci of benign prostatic hypertrophy. The right peripheral cancer in this case can be identified as a region of focally reduced T2 signal (red arrow on b). Additional pelvic floor anatomy well delineated includes obturator internus as white asterisks on b), the neurovascular bundles as white arrows, urinary bladder lumen as white asterisk, and urinary bladder wall as white arrowhead on c).

Figure 4.

Post-implantation images by the 3D trueFISP sequence, using the protocol in Table 1, at the levels of the genitourinary diaphragm, which extends from the apex to the penile bulb (a), the midgland (b), and the base (c). Unlike the images in Fig. 2 and Fig. 3, the trueFISP images display a combined T1/T2 contrast and clearly show both the anatomy and the seed markers (e.g. white arrow on b). In addition, the images show excellent SNR because the trueFISP sequence has fully-balanced gradients on all the axes and uses all the available magnetization. Among the anatomic structures well delineated are external urethral sphincter (white arrow), pubococcygeus muscles (white asterisks), anterior rectal wall (white arrowhead), and obturator internus (yellow asterisk) on a); obturator internus (white asterisks) on b); and the neurovascular bundles (white arrows), anterior rectal wall (yellow arrowhead), urinary bladder lumen (white asterisk), and urinary bladder wall (white arrowhead) on c). Both the radioactive seeds (yellow arrow on b) and the seed markers as the contours of the anatomy are visible on the coronal (d) and sagittal (e) reformatted images.

Figure 5.

A comparison of the images with the endorectal coil activated (a–c) and the images with the endorectal coil deactivated (d–e) from the same patient after implantation using the 3D trueFISP sequence and the protocol described in Table 1. The images were selected from the levels of the genitourinary diaphragm (a and d), of the midgland (b and e), and of the base (c and f). While the images with the active endorectal coil clearly show better SNR and image quality, the seed markers and the anatomy are also visible on the images with the endorectal coil deactivated. Among the anatomic structures well delineated on both sets of the images are the distal prostatic urethra (white arrowhead) on a) and d); and anterior rectal wall (yellow arrowhead) on c) and f). The dominant tumor can be seen at the level of the right anterior central gland, abutting the anterior fibromuscular stroma (red arrows) on (b–c) and (d–f). The seeds (e.g., yellow arrow on a) and d)) and seed markers (e.g., white arrow on a) and d)) are also visualized on the coronal and sagittal reformatted images with the endorectal coil on (g and h) and with the endorectal coil deactivated (i and j). Further improvement in the pulse sequence and scanning protocol such as by incorporating compressed sensing and using optimized seed markers may allow a single sequence dosimetry evaluation of prostate brachytherapy without using an endorectal coil.

Table 1

Scan parameters for the 3D FSPGR, 3D FSE, and 3D trueFISP pulse sequences that were used to acquire the images in Figs. 1–5.

Pulse sequence	FOV (cm)	Slice thickness (mm)	Slice resolution, oversample (%)	Slice number	Nx	Phase resolution, oversample (%)	TR/TE (ms)	Flip angle	Bandwidth (Hz/pixel)	ETL	NEX	Acceleration factor	Scan time (min:s)
FSPGR	15	1.2	50, 9.1	88	256	85, 100	5.5/2.4	25°	560	N/A	2	None	3:50
FSE	15	1.2	71, 9.1	88	256	80, 85	1500/278	Variable	488	120	1.4	2	3:32
TrueFISP	15	1.2	50, 9.1	88	256	100, 100	5.3/2.3	~50°	558	N/A	1	None	4:02