Development of a Quality Assurance Strategy for Magnetic Resonance Imaging in Radiotherapy

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Abstract

Radiotherapy is a common treatment option for malignant tumors. The use of magnetic resonance imaging (MRI) in radiotherapy is increasing world-wide and sets additional requirements on the image quality compared to diagnostics applications. Thereby, the overall goal of the present project was to develop and implement a structured image quality assurance approach for MRI in radiotherapy. Specific aspects studied was the image quality, spatial accuracy and spin-lattice relaxation time quantification. Measurements using three different phantoms were performed at the positron emission tomography MRI (PET/MRI) scanner at Norrlands University Hospital. The image quality was evaluated using the American College of Radiology (ACR) phantom and its accompanying tests. ACR phantom measurements were performed once or twice daily throughout the project. The results from the ACR tests were implemented in an automatic analysis program and displayed on a website, accessible from the hospital network. The image quality was found to be constant over time and within acceptance values. Furthermore, the spatial distortion was quantified using a large field of view (FOV) phantom and a spatial analysis program supplied with the phantom. A method to study the reproducibility of the distortion was developed and the distortion in the PET/MR scanner was mapped in different images planes. The image distortion was shown to be reproducible and less than 1 mm inside a volume of $20 \times 20 \times 20 \text{ cm}^3$ around the magnetic isocenter, which corresponds to a FOV used in target delineations images. However, outside this volume larger distortions than 2 mm was observed. In order to evaluate the quantification of the spin-lattice relaxation time two methods were evaluated, variable flip angle (VFA) and two point inversion recovery spin echo (IRSE) using the Test object 5 (TO5) phantom. Large variations in the quantification was observed using the VFA method compared to the two point IRSE method when performing phantom measurements. Unfortunately, the two point IRSE method was not adequate when used in head exams in the clinical practice.
### Commonly Used Acronyms

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Description</th>
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<tbody>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>RT</td>
<td>Radiotherapy</td>
</tr>
<tr>
<td>PET</td>
<td>Positron Emission Tomography</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>BW</td>
<td>Bandwidth</td>
</tr>
<tr>
<td>QC</td>
<td>Quality Control</td>
</tr>
<tr>
<td>QA</td>
<td>Quality Assurance</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal-to-Noise Ratio</td>
</tr>
<tr>
<td>FOV</td>
<td>Field of View</td>
</tr>
<tr>
<td>ACR</td>
<td>American College of Radiology</td>
</tr>
<tr>
<td>TO5</td>
<td>Test Object 5</td>
</tr>
<tr>
<td>NUS</td>
<td>Norrlands Universitetssjukhus</td>
</tr>
<tr>
<td>RF</td>
<td>Radio Frequency</td>
</tr>
<tr>
<td>SE</td>
<td>Spin Echo</td>
</tr>
<tr>
<td>GRE</td>
<td>Gradient Echo</td>
</tr>
<tr>
<td>SPGR</td>
<td>Spoiled Gradient Echo</td>
</tr>
<tr>
<td>FSPGR</td>
<td>Fast Spoiled Gradient Echo</td>
</tr>
<tr>
<td>CSF</td>
<td>Cerebrospinal Fluid</td>
</tr>
<tr>
<td>NEX</td>
<td>Number of Excitations</td>
</tr>
<tr>
<td>TI</td>
<td>Inversion Time</td>
</tr>
<tr>
<td>IRSE</td>
<td>Inversion Recovery Spin Echo</td>
</tr>
<tr>
<td>TR</td>
<td>Repetition Time</td>
</tr>
<tr>
<td>TE</td>
<td>Echo Time</td>
</tr>
<tr>
<td>CNR</td>
<td>Contrast-to-Noise Ratio</td>
</tr>
<tr>
<td>ROI</td>
<td>Region of Interest</td>
</tr>
<tr>
<td>PIU</td>
<td>Percent Integral Uniformity</td>
</tr>
<tr>
<td>HNU</td>
<td>Head-and-Neck Coil</td>
</tr>
<tr>
<td>RMRT</td>
<td>Large Flex Coil</td>
</tr>
<tr>
<td>CMA</td>
<td>Built-in Table Coil</td>
</tr>
<tr>
<td>UAA</td>
<td>Upper Anterior Array</td>
</tr>
<tr>
<td>8CH</td>
<td>Eight Channel Head Coil</td>
</tr>
</tbody>
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1. Background

1.1 Magnetic Resonance Imaging in Radiotherapy

Magnetic resonance imaging (MRI) is a medical imaging technique used for over 40 years as a clinical modality. It produces images with excellent soft tissue contrast and high spatial resolution. MRI has the advantage of low induced risks to the patients since no ionizing radiation is used, in contrast to other common diagnostic imaging methods. MRI has mainly been used as a diagnostic technique, for localization and classification of tumors and other diseases. However, the use of MRI in radiotherapy (RT) is today increasing world-wide and today integrated RT-MRIs are commonly seen in RT departments. This opens the possibility for personalized RT and improved tumor treatment through MRI tumor response assessments during treatment [1]. RT is a common treatment method for tumors; ionizing radiation is used to control or kill malignant cells by damaging the DNA in the cancerous tissue. In order to optimize the treatment, shaped beams are often used, radiating from several angles around the patient. This results in a three-dimensional dose distribution that enables high absorbed dose to the tumor (target) while minimizing radiation to healthy tissue. To achieve this, accurate definition of the target, organs at risk (OAR) and healthy tissue is necessary, as well as assessment of their spatial relationship [2]. Today, different imaging techniques are used in the RT workflow; computed tomography (CT), positron emission tomography (PET) and MRI. CT imaging is commonly used for the technical aspects of treatment planning, i.e. dose calculations and generation of reference images i.e. digital reconstructed radiograph (DRR). MRI and PET are often used for the medical aspects, such as defining the target volumes and OAR [3]. Image registration is used to define the spatial relationship between the CT and magnetic resonance (MR) images. Today’s RT workflow, illustrated in figure 1.1, is relying on the image registration between MR and CT images. Unfortunately, the correct alignment between two images are unknown and small variations in patient set-up, internal movement or small changes in the anatomy highly affects the result of the registration. As a consequence, image registration introduces systematic geometric uncertainties to dose plans which are present throughout the entire treatment and no adequate quality confirmation of the image registrations exists [3]. These uncertainties could be avoided by using only MRI data in the RT workflow, and could soon to be implemented for some common diseases [1].
Magnetic Resonance Imaging in Radiotherapy

1 Background

Figure 1.1: RT workflow in common practice; images of the patient are acquired using different modalities (CT, MRI and PET). MRI and PET data is usually used to define target, image registration between CT and MRI data is performed to define the spatial relationship, CT data is mainly used for dose calculations and follow-up under and after RT is done using MRI, PET and CT [3].

1.1.1 MRI Challenges

MRI in RT applications sets additional requirements on the image quality compared to diagnostics application, some of them are listed in table 1.1. It also entails some other technical challenges, especially when an MRI-only RT workflow is implemented. CT is an X-ray based imaging technique where the image contrast reflects the electron density in the patient which is used in the RT dose calculations [5]. MR images have no information of the electron density. They are created from magnetic properties of the hydrogen nuclei - MR image contrast is due to different magnetic properties of different tissues, i.e. the relaxation characteristics of the protons in the body [6]. There are many methods for calculating RT doses using MRI data, all of them uses synthetic CT images produced from MR-images. While all medical imaging modalities can suffer from anomalies, or artifacts, in the visual representation of a patient or object, MR is especially prone to such phenomena. Patient and machine induced geometrical distortion are for example a common and troublesome image artifact. Many artifacts can be avoided with proper imaging parameters, or in the image reconstruction. Nevertheless, knowledge about occurring artifacts is necessary to set up proper imaging protocols and avoid interpretation of artifacts as pathology. It is important that a uniform and consistent quality assurance is performed, especially in multi-unit imaging centers [7]. Image quality parameters like high- and low-contrast resolution, ghosting and uniformity of the images should be comparable in images from different MRI scanners. The integration of MRI in RT also requires special MRI-compatible RT equipment such as immobilization devices similar to those used during RT treatment. Figure 1.2 shows the MRI RT-setup for head and neck patients. A flex coil support accommodates the fixation mask for head and neck patients used for RT treatment. Flex coil supports are also designed for other anatomical regions, for example the pelvic area. Usually flex coils are placed on the patient in pelvic exams which can distort the outer contour of the patient. The flex coil support removes this distortion. Furthermore, a flat table top (identical to RT treatment tables)
Table 1.1: Summary of differences between MRI used for diagnostic applications and RTP applications [4]. Abbreviations in the table: digital reconstructed radiograph (DRR), image guided RT (IGRT) and dose volume histogram (DVH).

<table>
<thead>
<tr>
<th>Field of view (FOV)</th>
<th>MRI in RT</th>
<th>Diagnostic MRI</th>
</tr>
</thead>
<tbody>
<tr>
<td>One scan with the full body contour is required</td>
<td>Images can be acquired with reduced FOV</td>
<td></td>
</tr>
<tr>
<td>Readout bandwidth (BW)</td>
<td>High BW is often used to minimize image distortions</td>
<td>Usually set as a tradeoff between fat/water shift and SNR</td>
</tr>
<tr>
<td>Slice thickness and spacing</td>
<td>Thinner slices will improve the DRR image quality, no spacing is often used</td>
<td>Slice thickness of 4-5 mm and slice gaps of 0-2 mm is normally used</td>
</tr>
<tr>
<td>Slice coverage</td>
<td>Larger coverage is required for target and OAR delineation, DVHs and IGRT etc.</td>
<td>Only the volume of interest is required</td>
</tr>
<tr>
<td>Geometric distortion</td>
<td>&lt; 2 mm distortion required in all planes over the volume of interest</td>
<td>Diagnostic capability sets the acceptance limit</td>
</tr>
<tr>
<td>Uniformity</td>
<td>Increased uniformity is required for intensity based image registration and segmentation accuracy</td>
<td>Diagnostic capability sets the acceptance limit</td>
</tr>
</tbody>
</table>

indexing trays and lock bars (used to lock the fixation mask to the table), in non-magnetic material must be used [4].

### 1.1.2 Quantitative MRI

The intensities in MR images are in arbitrary units, and depend on e.g. scanner settings, signal processing and image postprocessing, which prevents quantitative analysis of standard MR images. Quantitative MRI is a collection of methods that measure physical quantities such as relaxation times and diffusion, a highly valuable tool since it combines quantitative information with the favorable MRI properties [8]. For example, in dynamic contrast enhanced MRI, changes in longitudinal relaxation times induced by a contrast agent is used to quantify perfusion and vascular integrity in order to find active tumor tissue [8]. The working potential of quantitative MRI is enormous but not yet standardized in the clinical routine and clinical trials [1]. In quantitative MRI, as well as the classical ways of using MRI, the reliability and reproducibility of the images are critical.
Quality Assurance in MRI

1.2 Quality Assurance in MRI

As mentioned above, there are technical challenges using MRI in RT especially when implementing a MRI-only RT workflow. MR images must be reliable, in terms of image quality, distortion, quantification etc and comparable between different MR scanners. This implies the need for routine quality controls (QC). Routine QC gives indications of the system’s operational status and when system calibration is necessary. Routine QC also gives information about how to interpret the images, i.e. the magnitude of distortion and artifacts occurring in the images. Knowledge about the degree of distortion and image artifacts (and their origins), implies that methods for correction can be developed in a secondary step.

1.2.1 MRI Quality Control and Phantoms

QC for MRI is often a non-optimized process that involves a lot of manual work for the MR physicist and may be sparsely done. There are few detailed national and international QA guidelines for MRI compared to imagining methods that uses ionizing radiation. Vendors provide built-in QC-programs that monitors the systems performance e.g. measures signal to noise ratio (SNR). These
QC-programs do not cover all parameters of interest and are often vendor specific. The National Electrical Manufacturers Association, American College of Radiology (ACR), American Association of Physics in Medicine and Intersocietal Accreditation Comission provides guidelines for acceptance testing of scanners and QC programs with appropriate tolerance levels. QC using the ACR MRI large phantom and its accompanying tests [9] has been done and evaluated in several reports [6, 7, 10–14]. Fitzpatrick [10] for instance, developed a semi-automatic analysis program for the ACR QC program. An semi-automatic analysis program is almost an prerequisite for developing a fast and reliable QC. ACR MRI QC test includes seven quantitative tests, described in detail in the Phantom Test Guidance [9]. Since the introduction of MRI in RT, more concern has been taken to the spatial accuracy since larger FOV is required and the spatial accuracy is critical. Larger FOVs introduces more geometrical errors, especially in the periphery of the FOV. As the spatial accuracy gets worse with the distance from isocenter, a distortion QC phantom of suitable size is necessary. With the purpose of assessing the spatial accuracy in MR-images GE Healthcare has developed a large FOV spatial accuracy phantom [15]. Furthermore, specially designed phantoms for evaluation of the quantitative images are today developed, for example Test object 5 (TO5) phantom developed by Eurospin II.

1.2.2 RT Dedicated PET/MR at Norrlands University Hospital

Norrlands University Hospital (NUS), Umeå has recently received a dedicated PET/MR (SIGNA PET/MR (3T) GE Healthcare, Milwaukee, WI USA) for clinical cancer research in the RT department. The scanner was donated from the Cancer Research Foundation in Norrland. This scanner will be used in in the daily RT workflow as well as in different cancer research programs. There is therefore a need to assure the quality of the scanner performance and its stability over time. Aspects that are necessary to assure as a first step is the image quality, spatial accuracy and the accuracy of quantitative imaging and the following questions needs to be answered: Is the image quality within reasonable acceptance levels and constant over time? How large are the image distortions, and are they clinically relevant? Is there a method for image distortion comparison between different scans and different MR scanners? Which quantification method gives the most accurate estimations and how stable is it? In order to answer these questions assurance methods for each parameter needs to be developed.

1.3 Aim

The overall goal of the present project is to develop and implement a structured image quality assurance approach for MRI in radiotherapy. The specific aim is to develop a semi automatic quality assurance strategy for MRI usage in radio-
therapy where the image quality, spatial accuracy, and methods for verifications of the quantitative images are included. The quality assurance strategy should be able to show the stability over time and identify deviations from the regular scanner performance.

1.3.1 Limitations

The project is limited to the PET/MR scanner in the radiotherapy clinic at NUS and will only concern the MR part of the scanner. Phantoms used for the measurements are limited to the ACR large phantom, TO5 phantom and the spatial accuracy large FOV phantom. $T_2$ quantification and dynamic contrast enhanced MRI sequences will not be studied.
2. Introduction

2.1 Basic Physics of MRI

2.1.1 Magnetic Resonance and Excitation Pulses

MRI is an imaging technique that uses the properties of the hydrogen nucleus to image the human body. Both water molecules and lipids consist of hydrogen atoms, which makes up 70% - 90% of the human body [16]. The hydrogen nuclei consist of one charged particle, a proton, which has the intrinsic property of spin and an associated magnetic moment [17]. When protons are placed in a strong external magnetic field they will initially align with the magnetic field due to the intrinsic magnetic moment. The protons will not align perfectly due to quantum mechanical effects which makes the proton precess around the magnetic field. Summing the magnetic moments from all protons gives a vector pointing parallel to the magnetic field. This vector is called the net magnetization vector and is the origin to the MRI signal. The proton precession frequency or resonance frequency, often called the Larmor frequency ($\omega_0$), is proportional to the main magnetic field ($B_0$) and nucleus specific according to the Larmor equation

$$\omega_0 = \gamma B_0$$

where $\gamma$ is the gyromagnetic ratio. The gyromagnetic ratio for protons is equal to $2.7 \times 10^8$ rad/sT, and in terms of frequency $\gamma = 42.57$ MHz/T [16].

2.1.2 MRI Signal

Compared to $B_0$, the net magnetization vector is extremely small. However, by applying a radio frequency (RF) field ($B_1$) the net magnetization vector tilts into the plane perpendicular to $B_0$ (the transverse plane). In order to obtain this, the RF-field must have the same frequency as the precession frequency of the hydrogen nuclei [17]. The net magnetization vector will now precess around the axis of the main magnetic field at the same time as it is tilted away from the field axis to the transverse plane. This phenomenon is quite hard to visualize, but using a rotating frame of reference that excludes the influence of $B_0$ it becomes more clear. Figure 2.1 shows the translation from the ordinary frame to a rotating frame of reference. As the net magnetization vector precesses around $B_0$ an RF signal is emitted [17]. The transverse component of the net magnetization is proportional to the signal strength [17]. Immediately after the $B_1$ field is turned off, all protons rotate in phase. Due to local magnetic field variations and magnetic interactions between different hydrogen atoms, some spins will rotate...
faster, other slower, and the coherence will gradually be lost. This is called $T_2^*$ relaxation, and it causes the signal to gradually decay in 50-200 ms. The solution to this is to create an echo from the signal and detect this signal instead. The echo will occur at the time (TE) after the initial signal. There are two different branches of MRI sequences that differently creates echoes of the initial signal, spin echo (SE) and gradient echo (GRE). In SE, a 90 degree-excitation pulse is used and the echo is created using a 180 degree-refocusing pulse [16]. In GRE the excitation pulse is is usually smaller (less than 90 degree) and denoted as $\alpha$, and the echo is formed using rephasing and dephasing gradients [16]. Receive and/or transmit coils are used to detect the signal and produce $B_1$.

### 2.1.3 Signal Detection and Localization

In order to detect the signal and localize the origin of the signal, variations in the main magnetic field are created. This is done through three sets of gradient coils in the MRI system that produce magnetic field variations, often called gradient fields. Two-dimensional MR images can be acquired in different image planes; axial, sagittal and coronal planes are common choices, shown in figure 2.2. In order to localize signal only from the volume of interest, $B_0$ is made non-uniform by applying a slice selection gradient. The precession frequency of the protons becomes a function of position, for example along the z-axis (depending on the choice of image plane). Phase encoding gradient works in a similar way but is applied perpendicular to the slice direction, creating a phase shift of the excited protons. Depending on the position along the phase encoding direction, the protons will be more and more out of phase from each other. This causes different spin patterns of the magnetization that varies over the object [17]. The spatial position of the protons is localized by a frequency encoding gradient, applied perpendicular to both the slice and phase encoding gradients. It forces the protons in the slice to precess with different frequencies [16]. The precession frequency of the protons will change, depending on their position along the
frequency encoding direction. The signal now contains a range of frequencies, corresponding to each unique position in the patient. These frequencies are collected in a raw data matrix called $k$-space. They represent the spatial frequency information in the image and by taking the Fourier transform the final image is reconstructed.

2.2 Relaxation

Each atom and molecule have a molecular motion; they vibrate, rotate and translate in random directions. These particle interactions will affect the excited protons in different relaxation processes \[16\]. In MRI there are two main components that contribute to the relaxation of the magnetization. These are called $T_1$-relaxation (spin-lattice relaxation) and $T_2$-relaxation (spin-spin relaxation). The image contrast in MRI depends on these tissue specific relaxation times, and by adjustments of the sequence acquisition parameters, specific tissues can be enhanced or suppressed in the image. Contrast difference between a $T_1$ weighted image and a $T_2$ weighted image is shown in figure 2.3. Relaxation can be described mathematically using the Bloch equation

\[
\frac{d\vec{M}(t)}{dt} = \gamma \vec{M}(t) \times \vec{B}(t)
\]

where $\vec{M}(t)$ is the magnetization vector at time $t$ and $\vec{B}(t)$ is the magnetic field experienced by the nuclei \[16\]. By separating $\vec{M}(t)$ and the $\vec{B}(t)$ into its three orthogonal components, a set of equations are obtained that describes the behavior of adding RF pulses to protons inside a magnetic field. Furthermore, relaxation times depends both on the magnetic field strength and the sample temperature \[18\]. Higher temperatures and magnetic fields increases the relaxation time and it is therefore necessary to monitor the temperature when performing relaxation time estimations.
2.2.1 Spin-lattice Relaxation

Spin-lattice relaxation time, $T_1$, is the amount of time it takes for the magnetization to realign with the main magnetic field after the excitation pulse. $T_1$ is defined as a recovery of 63% of the equilibrium value of the initial magnetization. In this process protons will lose their absorbed energy from the excitation pulse to the surrounding lattice and establish thermal equilibrium [19]. $T_1$ relaxation can be seen as the net magnetization vector recovering from the transverse plane and aligning against the axis of $B_0$. To calculate $T_1$ the following equation can be used, derived from the Bloch equation (2.2):

$$M_z(t) = M_0 + [M_z(0) - M_0] e^{-t/T_1}.$$  \hspace{1cm} (2.3)

$M_z(t)$ is the magnetization at time $t$ and $M_0$ is the equilibrium magnetization [20]. Examples on $T_1$-values for different tissues can be seen in table 2.1.

2.2.2 Spin-spin Relaxation

Spin-spin relaxation time, $T_2$, is defined as the amount of time it takes for 37% of the transverse magnetization ($M_{xy}$) to dephase. Intrinsic inhomogeneities in the magnetic field, created by neighboring spins which depend on the molecular motion of the proton, causes the dephasing. Assuming a 90°-excitation pulse was applied along the x-axis in the rotating frame of reference and switched off. Further $M_x(0) = M_z(0)$ is assumed to be zero, $T_2$ can be calculated with the following equation:

$$M_{xy}(t) = M_y(0) e^{i\omega_0 t} e^{-t/T_2}$$  \hspace{1cm} (2.4)

where $M_{xy}(t)$ is the transverse magnetization at time $t$, and $M_y(0)$ is the initial transverse magnetization [16].
Table 2.1: Examples on $T_1$-relaxation times in different tissues. These values are valid for in a magnetic field strength of 3T and the temperature of 37 °C.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_1$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cerebrospinal fluid (CSF)</td>
<td>$\sim 4350$</td>
</tr>
<tr>
<td>Prostate</td>
<td>$\sim 1600$</td>
</tr>
<tr>
<td>Gray matter</td>
<td>$\sim 1450$</td>
</tr>
<tr>
<td>White matter</td>
<td>$\sim 850$</td>
</tr>
</tbody>
</table>

2.3 $T_1$ Estimation Methods

2.3.1 Inversion Recovery Sequence

$T_1$ can be estimated using an inversion recovery SE (IRSE) sequence. In IRSE sequences the magnetization is initially inverted through an 180°-inversion pulse applied along the z-axis. After the inversion pulse the magnetization starts to relax back to equilibrium and an 90 degree pulse is applied. The time between the inversion pulse and the 90 degree pulse is called the inversion time (TI). The acquired signal will depend on the inversion time and the $T_1$ relaxation time. By acquiring a set of images with varying TI, $T_1$ can be calculated. The signal from an IRSE can be calculated according to:

$$S_{IR}(TI) = S_0 \left[1 - (1 - k)e^{-TI/T_1} + e^{-TR/T_1}\right]$$

where $S_0$ is the initial magnetization, $k$ is a constant compensating for imperfect inversion pulses and TR is the repetition time of the sequence [21]. $T_1$ can be measured by either varying the repetition time (TR) or TI, using this method. A minimum of two different TI must be used and full relaxation can be assumed when $TR \gg T_1$ [21]. The assumption of full relaxation implies that equation 2.5 can be reduced to:

$$S_{IR}(TI) = S_0 \left[1 - (1 - k)e^{-TI/T_1}\right].$$

Furthermore, equation (2.6) can be simplified by using a reference signal obtained from a SE sequence, $S_{SE} = S_0e^{-TE/T_2}$ [16], and two different TI (TI$_1$ and TI$_2$). Then, the following signal ratio is obtained [21]:

$$\frac{S_{SE} + S_{IR}(TI_1)}{S_{SE} - S_{IR}(TI_2)} = \frac{e^{-TI_1/T_1}}{e^{-TI_2/T_1}} = e^{(TI_2-TI_1)/T_1}.$$ (2.7)

Rearranging equation 2.7, an expression for the estimated $T_1$ can be found:

$$T_{1,est} = \frac{T_{I1} - T_{I2}}{\ln \left[ \frac{S_{SE} + S_{IR}(TI_1)}{S_{SE} - S_{IR}(TI_2)} \right]}.$$ (2.8)
This estimation method is from now referred to as "tow point IRSE". When using this method no consideration needs to be taken to imperfect inversion pulses and zero-bouncing [21].

### 2.3.2 Variable Flip Angle

Another commonly used sequence for $T_1$ estimation is spoiled gradient echo (SPGR) used with different flip angles, $\alpha$. The signal equation for SPGR sequences is:

$$S_{\text{SPGR}}(\alpha) = S_0 \sin(\alpha) \left[ \frac{1 - e^{-TR/T_1}}{1 - \cos(\alpha)e^{-TR/T_1}} \right] e^{-T_2^*/TE} \quad (2.9)$$

where $\alpha$ is the flip angle and TE is the echo time [22]. A spoiled sequence implies that all transverse components of the steady state magnetization is removed before each new RF-pulse. This removes all disruptions from previous excitation pulses. Holding TR constant and increasing $\alpha$ gives a curve which can be written in linear form ($y = kx + m$) [22] as follows:

$$\frac{S_{\text{SPGR}}(\alpha)}{\sin(\alpha)} = e^{-TR/T_1} \frac{S_{\text{SPGR}}(\alpha)}{\tan(\alpha)} + S_0(1 - e^{-TR/T_1}). \quad (2.10)$$

Linear regression of equation 2.10 yields the slope and the intercept of the curve which gives an expression for $T_1$:

$$T_1 = -\frac{TR}{\ln(k)} \quad (2.11)$$

where $k$ is the slope of the curve [22]. Unfortunately, the transmitted RF field will not be homogeneous over the imaging object if the size of the object is in the order of the wavelength of the field [23]. This implies that the flip angle will not be homogeneous over the imaging object when using VFA to estimate $T_1$. A more precise estimation can be obtained by calculating the actual flip angle, which is done by mapping the RF-field, $B_1$. $B_1$-maps can be calculated in several ways and one common method is to use a double angle method [24].

### 2.4 Image Quality in MRI

The image quality of MRI depends on different scanning acquisition parameters, for example the echo time, repetition time, number of excitations and the size of the sampling matrix. It also depends on e.g. the gradient fields, image artifacts, type of coil used, patient movement (external and internal) and the magnetic field strength etc. All scanning protocol parameters must be optimized for the anatomical region and the subject of interest. In many parameters there is a give-and-take relationship where improvement of one aspect causes
2 Introduction

Image Quality in MRI

No Artifacts Aliasing Flow Artifacts

Figure 2.4: Examples of image artifacts in MRI. The images are obtained from a healthy volunteer.

deterioration of another. For example, high magnetic field strength ($\geq 3T$) increases signal to noise ratio, SNR, which enables improved spatial resolution or allows faster imaging. It also enhances the sensitivity to local magnetic field inhomogeneities [25] and improves spectral fat suppression and spectroscopy but increases the non desirable chemical shift and dielectric effects [16]. Furthermore the scanning time increases the image quality since it scales to the SNR as the square root of the number of excitations (NEX). However, the scan time is often a limiting factor since a long scan time is inconvenient for the patient and increases the risk for patient movement.

2.4.1 Image Artifacts

An image artifact is an incorrect depiction of the anatomy in the imaged object. It is caused by either the scanner, scanner operator or for example motion of (or inside) the imaged object. Many artifacts can be avoided by using correct scanning parameters, image post-processing and short scan times, to minimize patient motion. Figure 2.4 shows one MR image with correct anatomical depiction, one with an aliasing artifact and one image with a flow artifact. Aliasing artifacts occurs when there are too few sampling steps in the phase encoding direction and can be avoided by oversampling in this direction. Flow artifact occur due to a mismatch in the signal detection caused by the blood flow, excited protons have changed position between excitation and signal detection. Another common artifact is ghosting; appearance of a copy of the imaged object displaced from its true location. Sometimes, if there are many low-level ghosts, they are not clear copies of the object. Instead the artifact is seen as a smear of brighter signal in the phase encoding direction. The source of the smear is the brighter areas of the true image. Ghosting occurs due to signal instability between pulse cycle repetitions [9].
2.4.2 Spatial Resolution

Image spatial resolution influences the quality of the image. High spatial resolution gives sharp images and enables detection of small objects while low spatial resolution makes the image look blurry. The spatial resolution depends on the voxel size, which in turn is determined by the size of the sampling matrix, FOV and the slice thickness. Low contrast object detectability is defined as the ability to distinguish different objects with similar signal intensity. Likewise, high contrast detectability is the ability to resolve high contrast objects in the image. Both depends on the contrast-to-noise ratio (CNR) achieved by the scanner in the image. CNR is calculated by:

\[ CNR = \frac{|S_{ROI,A} - S_{ROI,B}|}{\sigma_{bkgd}} \]  \hspace{1cm} (2.12)

where \( \sigma_{bkgd} \) is the standard deviation of the noise in the image, obtained from a region of interest (ROI) outside the object. \( S_{ROI,A} \) and \( S_{ROI,B} \) are mean signal intensities from different structures (A and B). Artifacts in the image, such as ghosting, lowers CNR.

2.4.3 SNR

Another important image quality parameter is the image SNR. SNR gives a measure of how much of the image intensity is due to true signal from the object in the scanner versus random signals (background noise). SNR is calculated as the ratio of the mean pixel value within a ROI, \( S_{ROI} \), to the standard deviation of the noise, \( \sigma_{bkgd} \), in the following way:

\[ SNR = \frac{S_{ROI}}{\sigma_{bkgd}} \]  \hspace{1cm} (2.13)

SNR is an important indication of the image quality and high SNR is preferable. Image SNR can be optimized by choosing optimum sequence acquisition parameters for the specific pulse sequence used and the relaxation time of the tissue studied. In two-dimensional sequences SNR has the following relations:

\[ SNR \propto \frac{FOV_x FOV_y \Delta z \sqrt{NEX}}{\sqrt{N_x N_y} \sqrt{BW}} \]  \hspace{1cm} (2.14)

where \( N_x \) and \( N_y \) is the frequency and phase encoding steps. \( \Delta z \) is the slice thickness and \( FOV_x \) and \( FOV_y \) is the FOV size in the x- and y-direction, respectively, and \( BW \) is the readout bandwidth.
2.4.4 Readout Bandwidth

The readout bandwidth, BW, sets the range of frequencies used in the MRI signal detection. It is related to the number of sampled points in the frequency encoding direction, $N_x$, and the sampling time, $t_f$, as follows:

$$BW = \frac{N_x}{t_f}. \quad (2.15)$$

Using a high BW decreases the image SNR, while a low BW increases SNR. Unfortunately, a low BW also increases the chemical shift artifacts in the image [16]. Chemical shift occurs due to the slightly different resonance frequency of fat and water, characterized by miss-mapping of fat. This is visually noticeable as bright and dark bands located at the interface of fat and soft tissue. MR manufacturers defines BW in different ways. Common definitions are; the total BW across the image, half BW over the image (denoted as ±62.5 kHz which implies 125 kHz across the total image), BW/mm and BW/pixel.

2.4.5 Magnetic Field Inhomogeneity

The magnetic field is not perfectly homogeneous, there exists small variations of the magnetic field strength over the imaging volume, which scales to the distance from the scanner core, isocenter. The inhomogeneity is usually expressed in parts per million inside a given spherical volume. Magnetic field inhomogeneity causes geometric distortions in the images due to spatial shifts during the spatial encoding. As mentioned before, spatial encoding of the signal is done using phase and frequency encoding gradients. Before the signal readout, e.g. the frequency encoding, the phase of the protons are made non-uniform and each protons position becomes a function of its phase. In each phase encoding step the same amount of time is spent which results in a constant phase shift over the image due to the magnetic field inhomogeneity. A constant phase shift will not affect the resulting image. Unfortunately, during the frequency encoding the phase shift will vary over the readout, resulting in geometrical distortions in the image. Geometrical distortions due to magnetic field inhomogeneity are hence observed in the direction of the frequency encoding gradient. These distortions are related to the sampling time, i.e. the BW and the size of the FOV used. Using a high BW, the sampling time gets faster and smaller phase shifts are observed between the first and last row in k-space which decreases the image distortions. In the same way, using a smaller FOV the sampling time gets shorter and the geometrical distortions decreases. Patient-induced inhomogeneity of the magnetic field occurs through susceptibility effects, but are generally small [26]. Poor magnetic field homogeneity can sometimes be noticed only if a phantom of proper size is used, phantom volumes smaller than 24 cm is not adequate for homogeneity testing of whole body scanners [13]. Shimming is a method used for compensating for the magnetic field inhomogeneity, which corrects for some of the geometrical distortions. It also corrects for drift in the center frequency [13]. Figure 2.5 illustrates the effect of image distortion.
Figure 2.5: Illustration of image distortion in MRI. The image to the right is acquired without using 3D-correction which results in an inaccurate geometrical depiction.

2.4.6 Gradient Field Non-Linearity

In the ideal case, the gradient fields should be linearly increasing, where the magnitude of the field is linearly correlated to the distance from the isocenter of the magnetic field. In reality the gradient fields are not linear over the entire FOV. The linear deviation is increasing with the distance from isocenter and does not scale with the strength of the gradient field. Similar to the static magnetic field inhomogeneity this causes geometric distortions in the images. However, distortions due to gradient non-linearities are constant can occur in any arbitrary direction. Image post-processing (GE Healthcare, for example, uses Gradwarp) corrects for the gradient induced spatial distortions.
3. Image Quality

3.1 Methods

3.1.1 ACR MRI Large Phantom

The image quality control was done using the ACR MRI large phantom. The ACR phantom is a hollow cylinder of acrylic plastic with an inside diameter of 190 mm and inside length of 148 mm. The phantom is filled with a solution of nickel chloride and sodium chloride [27]. Inside the phantom there are several structures that are used when performing the image quality tests and scanner performance tests. The ACR phantom can be seen in figure 3.1.

Figure 3.1: ACR phantom used to study the MR image quality.

3.1.2 Phantom Measurements

The ACR phantom measurements were performed at the PET/MR scanner, located in the RT department at NUS. The scanning setup and pulse sequence
Table 3.1: Pulse sequence acquisition parameters used for the ACR Phantom scanning according to ACR Site Scanning Instruction [27]. The axial $T_2$ weighted sequence is a double echo and one NEX was used in all sequences. Abbreviations in the table: axial (ax), weighted (w), pulse sequence (PS), repetition time (TR), echo time (TE), slice thickness (ST), slice gap (SG).

<table>
<thead>
<tr>
<th>Sequence</th>
<th>PS</th>
<th>TR (ms)</th>
<th>TE (ms)</th>
<th>FOV (cm)</th>
<th>Slices</th>
<th>ST (mm)</th>
<th>SG (mm)</th>
<th>Matrix</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sagittal-locator ax $T_1$ w</td>
<td>SE</td>
<td>200</td>
<td>20</td>
<td>25</td>
<td>1</td>
<td>20</td>
<td>N/A</td>
<td>$256\times256$</td>
</tr>
<tr>
<td>ax $T_2$ w</td>
<td>SE</td>
<td>500</td>
<td>20</td>
<td>25</td>
<td>11</td>
<td>5</td>
<td>5</td>
<td>$256\times256$</td>
</tr>
<tr>
<td>ax $T_2$ w</td>
<td>SE</td>
<td>2000</td>
<td>20/80</td>
<td>25</td>
<td>11</td>
<td>5</td>
<td>5</td>
<td>$256\times256$</td>
</tr>
</tbody>
</table>

parameters used were taken from the recommendations from the ACR Site Scanning Instruction [27]. One exception from the recommendations was that a clinically used head protocol was not included in the ACR test protocol. Pulse sequence parameters used can be seen in table 3.1 and acquired ACR images can be seen in figure 3.2. The phantom was scanned daily (sometimes twice daily) by the technicians working at the PET/MR scanner. The phantom was scanned in a one-channel transmit/receive head coil, only used for quality control. Precise position of the ACR phantom was done by aligning the coil with the scanner laser beam. The phantom was placed in the center of the coil with the help of the laser beam and the alignment marker on the phantom and a stack of papers inside the coil that corrected the height of the phantom. The imaging protocol and coordinates for slice positioning was saved to simplify the scanning process and to ensure reproducible scans.
Figure 3.2: ACR sagittal locator and axial slices one to eleven used in the ACR image quality test. These images are acquired from the $T_1$-weighted pulse sequence.
3.2 ACR Analysis Program

The ACR phantom test guidance [9] recommends evaluation of seven different image quality and scanner performance tests. The image analysis of each test is a manual procedure, which is time consuming and the results are operator dependent. To simplify the process to a more time efficient and operator independent quality control, an automatic ACR analysis program was designed in Matlab (The MathWorks Inc., Natick MA, USA). Further reasons for the development of this program was to obtain total control of the image quality assurance process and flexibility in presenting the results.

3.2.1 Phantom Center and Rotation

The ACR analysis program uses the Hough transform to calculate the phantom radius and center position. All measurements were calculated in reference to the center of the phantom, making the analysis insensitive to positional and rotational variations.

3.2.2 Geometric Accuracy Test

The phantom diameter was measured in slice 1 and 5, in the x- and y-directions, and two diagonals ($\theta = 45^\circ$ and $\theta = 135^\circ$). The phantom edges were found by taking the discrete derivative of the intensity profile and using a suitable intensity threshold. The threshold was set to 12.5 % of the maximum intensity, which gave length measurements comparable to manual phantom length measurements.

3.2.3 High Contrast Spatial Resolution Test

The high-contrast spatial resolution test was performed by calculating the amount of resolved high contrast object in a matrix was calculated. Twelve rows and columns of high contrast objects of different sizes must be detected and resolved to pass the test. Threshold values was used to define the peak as resolved compared to the neighboring peaks and the background noise.

3.2.4 Slice Thickness Test

The slice thickness test is designed to verify that the scanner produces slices with the desired thicknesses. A slice thickness-insert, located in slice 1, with two crossed ramps was used to measure the slice thickness. Slice thickness was calculated by measuring the length of the two ramps in the slice [9].
3.2.5 Slice Position Accuracy Test

Slice position accuracy was verified by measuring the length difference in two opposite ramps located in slice 1 and 11. This test verifies the scanner's ability to correctly position the slices at the specific location chosen by the user. The absolute difference should not be larger than 5 mm [9]. The length difference in each slice was obtained by subtracting the length of the left ramp from the right ramp, so that the sign of the measurement indicated the direction of the position error [9].

3.2.6 Image Intensity Uniformity Test

The uniformity of the images was evaluated in a homogenous slice in the middle of the phantom, in slice 7. Inside 80% of the phantom area in the slice two smaller ROIs with an area of 1 cm$^2$ encompassing the lowest and highest intensities were placed. The uniformity was calculated by comparing the difference in the ROIs with the sum, in percentage units, labeled percent integral uniformity (PIU) [9].

3.2.7 Percent-Signal Ghosting Test

The ghosting ratio was calculated by measuring the mean values in four ROIs outside the phantom, in slice 7. Two rectangular ROIs were placed in the phase encoding direction, measuring the ghost signal. Additionally, two ROIs were placed in the frequency encoding direction, sampling the ghost free signal (background signal and noise). The difference between the mean values in the phase encoding direction and the frequency encoding directions were compared to the mean value in central 80% of the area in the phantom. [9]

3.2.8 SNR Test

Since the image SNR is a good indication of the stability of the scanner, this additional test were added to the original ACR tests. Using equation (2.13) with the mean signal value in 1 cm$^2$ ROI encompassing the highest value in the phantom (used in the uniformity test) and a background ROI placed in the upper left corner in slice 7 (to avoid ghost signals).
3.3 Results

3.3.1 Image Data Management and Presentation

After each phantom measurement, images were sent from the MRI scanner to a local research database called MIQA from where the analysis software can access the images and perform the QC measurements. When the analysis is completed the result from each test is written to a database and can be accessed from an internal website (vs0044.vll.se/mrqa). Plots from the seven ACR tests and the additional SNR test and noise level measurement in slice 7 are presented on the website. The user can choose to study data from different MRI-scanners, different RF coils and specify the sequences. The user can show data in one month, three month or six months intervals. Figure 3.3 shows the interface of the website and the possible user choices.

3.3.2 ACR Test Results

Figures 3.4 to 3.7 shows the result from four of the ACR tests evaluated by the ACR analysis program. The remaining plots can be seen in appendix A, figures A.1 to A.6. In tests where only one result per sequence was obtained (figures 3.5 to 3.7 and A.4 to A.6) the $T_1$-weighted sequence were symbolized by blue dots and the $T_2$-weighted double-echo sequence with TE = 20 ms with black dots and TE = 80 ms with green dots. In tests where multiple results per sequence is obtained, each result is shown by a different symbol, described in the label under the figure. Axis rescaling is used to visualize outliers. Acceptance levels used in the following figures are the recommended from the ACR phantom test manual [9]. A summary of reasons for failure of the seven ACR test can be seen in appendix B.
Figure 3.3: The MR QA web page where the results are displayed. The user can choose time interval, modality, RF coil and sequences.
Figure 3.4: Results from the ACR geometry test with phantom diameter measurement in slice 1. The green band indicates the acceptance level of the diameter measurement (19±0.2 cm). The diameter is measured in x- and y-direction, and two diagonals (45° and 135°). The different diameter measurement directions are plotted with different symbols.

Figure 3.5: Results from the ACR SNR measurement with the ACR phantom.
Figure 3.6: Results from the ACR percent uniformity integral test. All pulse sequences must at least have a PIU result of 82% to pass this test.

Figure 3.7: Results from the ACR slice thickness test. The green band corresponds to the acceptance values (5±0.7 mm).
4. Spatial accuracy

4.1 Methods

4.1.1 Large FOV Phantom

The image QA includes a geometrical accuracy test, where the diameter of the ACR phantom is measured. It has a diameter of 19 cm, i.e. the measurement was performed at a distance of ±9.5 cm from isocenter. A normal-sized patient, imaged over the pelvic area has a diameter of about 40 cm, which yields a distance of ±20 cm from isocenter. Hence, a phantom larger than the ACR is required to estimate geometrical distortions relevant in clinical applications, since the geometrical distortions increases with the distance from isocenter. Regarding this, the phantom used in this project was a large FOV phantom made of foamed polyvinyl chloride with a size of 55×55×35 cm³ designed by GE Healthcare, shown in figure 4.1. This phantom is composed of 14 plates, each 25 mm thick. In each plate, paintballs are distributed in well-defined geometrical patterns. A total number of 2056 paintballs (6 mm in diameter) are located inside the phantom, which covers 45×45 cm coronal FOV around isocenter. The paintball markers yield a positive contrast in the MR image and can be used to measure the spatial accuracy of the scanner. [15]

4.1.2 Spatial Accuracy Analysis Program

The distortion was calculated using a spatial accuracy analysis program developed by GE Healthcare. The program is designed in Matlab and uses an intensity threshold to find the locations of the markers in the image data. It subtracts the $i$:th theoretical reference position, $(x_i, y_i, z_i)_{\text{ref}}$, from the $i$:th measured position, $(x_i, y_i, z_i)_{\text{meas}}$ to obtain the error vector, $\vec{E}_i$ [15]:

$$
\vec{E}_i = (x_i, y_i, z_i)_{\text{meas}} - (x_i, y_i, z_i)_{\text{ref}}.
$$

(4.1)

The theoretical reference used was a binary template file describing the locations of the markers inside the phantom. Alternatively, CT data of the phantom could be used as a reference. The analysis program also excludes false markers (image artifacts) that does not correspond to a reference marker. The output figures from the analysis program are shown in figure 4.2 where the distortion errors in the x, y and z direction for each paintball marker are shown as a function of the distance from isocenter. Figure 4.2 also shows the absolute error for all paintballs. As can be seen in this figure, the distortion is larger in the
x-direction, which is the frequency direction. This is a consequence of magnetic field inhomogeneities.

4.1.3 Phantom Measurements

All scans were performed on the 3T PET/MR scanner using the built-in RF body coil, due to the size of the phantom. An acquisition protocol was recommended by the manufacturer with the following sequence acquisition parameters; 3DFSPGR with flip angle $\alpha = 12^\circ$, BW = $\pm 125$ kHz, FOV = $61.4 \times 61.4 \text{ cm}^2$ and matrix size $512 \times 512$. The frequency encoding direction was set to right-left (R-L). Ideally, the phantom should be aligned with the gradient coil coordinate system and the center marker, declared in the template file, positioned in the isocenter of the magnet. The center marker was placed in the isocenter manually and its position verified by a 3-plane localizer. The center marker position was accepted when it was within $\pm 1$ mm from isocenter in R-L and S-I direction and $\pm 3$ mm in anterior-posterior (A-P) direction. [15]

4.1.4 Reproducibility and Distortion Analysis

Using the data from the spatial analysis program, color coded maps of the distortions, were produced in Matlab. The distortion maps were generated by interpolating the distortion vector field obtained by equation (4.1) to a regularly spaced grid, using triangulation-based natural neighbor interpolation.
Figure 4.2: The top graph shows the distortion error in the x, y and z directions obtained from the spatial analysis program are plotted against the distance from the isocenter. The bottom graph shows the absolute distortion error plotted against the distance from isocenter.

To illustrate the magnitude and direction of the error, surface plots over the distortions were created. The distortion errors were illustrated by both using the absolute error and total error for different phantom planes. When using the total error the absolute error was used to color code the error.

4.2 Results

4.2.1 Reproducibility

The spatial analysis program identified different numbers of paintballs at different scans, depending on the image quality. This made it hard to study variations of individual markers. By creating interpolated maps of the distortions it was possible to compare distortions between scans and opened the possibility to study the constancy of the distortion. Figure 4.3 shows comparisons between four different scan occasions. All scans were acquired using the reference protocol and positioned identically. Figure 4.3 shows axial slices of the phantom. The maximum color range was set to 2 mm equally to the requirement for MRI image distortions in RT [4].
4.2.2 Distortion Surface Plots

Figure 4.4 shows axial, sagittal and coronal slices of the image distortion in varying distances from isocenter. At the distance of $z = -225$ mm from isocenter large distortions were observed in each image plane. Patient images are rarely obtained at such distances from the isocenter. A more representative distance for target delineation is $\pm 10$ cm from isocenter (marked with dashed lines). Figures 4.5 and 4.6 shows the impact of using different BW ($\pm 62.5$ kHz, $\pm 125$ kHz and $\pm 250$ kHz) on the distortions, at isocenter (figure 4.5) and $z = -50$ mm (figure 4.6). The image distortion at $z = 50$ mm, $z = \pm 100$ mm and $z = 225$ mm can be seen in appendix C in figures C.1 to C.4. Increasing the BW lowers the image distortion in the periphery of the FOV, which is due to magnetic field inhomogeneities (see section 2.4.5). One scan with no 3D correction and BW $= \pm 250$ kHz were acquired, seen to the right in figures 4.5, 4.6 and C.1 to C.4. Only the periphery of the FOV is affected of the changes BW. Inside $\pm 10$ cm from isocenter (marked with dashed lines) the distortion is not significantly affected. Even the images obtained without 3D correction shows reasonably result in this area, but as expected severe distortions is observed outside the marked area. The axial plane shows no impact when the 3D correction is turned off. The reason for this is due to a default 2D correction that always is applied.

4.2.3 Distortion Visualization on Patient Images

Another use of the quantified distortion is the possibility to visualize the distortions in patient images, see figures 4.7 and 4.8. This can be useful when there is a need to verify that the distortions are within tolerance, for example when designing RT plans for large patients.
Image Distortion, BW = 488.28 Hz/pixel, Axial Slices

Figure 4.3: Distortion map comparison between identical scans performed at different occasions. All scans were acquired using the reference sequence acquisition.
**Spatial accuracy Results**

Image Distortion, BW = 976.56 Hz/pixel

<table>
<thead>
<tr>
<th>Axial</th>
<th>Sagittal</th>
<th>Coronal</th>
</tr>
</thead>
<tbody>
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<td><img src="image1" alt="Axial Image" /></td>
<td><img src="image2" alt="Sagittal Image" /></td>
<td><img src="image3" alt="Coronal Image" /></td>
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<td><img src="image6" alt="Coronal Image" /></td>
</tr>
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<td><img src="image8" alt="Sagittal Image" /></td>
<td><img src="image9" alt="Coronal Image" /></td>
</tr>
<tr>
<td><img src="image10" alt="Axial Image" /></td>
<td><img src="image11" alt="Sagittal Image" /></td>
<td><img src="image12" alt="Coronal Image" /></td>
</tr>
</tbody>
</table>

**Figure 4.4:** Spatial distortion illustrated at different positions from isocenter and in different image planes. The total error is used to create the meshed surfaces and the absolute error is used as the color coding.
Results 4 Spatial accuracy

Image Distortion, z = 0 mm, Different BW

Axial  | Sagittal  | Coronal
--- | --- | ---
244.14 Hz/pixel | 488.28 Hz/pixel | 976.56 Hz/pixel (No 3D-corr)
-150 | -150 | -200
0  | 0 | 0
200 | 200 | 200
-150 | -150 | -200

Distortion (mm)

Figure 4.5: Image distortion using different BW and no 3D correction. The absolute error is used to illustrate the distortion in different image planes at isocenter.

Image Distortion, z = −50 mm, Different BW

Axial  | Sagittal  | Coronal
--- | --- | ---
244.14 Hz/pixel | 488.28 Hz/pixel | 976.56 Hz/pixel (No 3D-corr)
-150 | -150 | -200
0  | 0 | 0
200 | 200 | 200
-150 | -150 | -200

Distortion (mm)

Figure 4.6: Image distortion using different BW and no 3D correction. The absolute error is used to illustrate the distortion in different image planes, at z = 100 mm.
**Figure 4.7:** A sagittal distortion map fused with a sagittal prostate image. In this figure the distortion covers the FOV = 32.5×45.0 cm² and the image FOV = 20×20 cm². The distortion range is limited to 2.5 mm.

**Figure 4.8:** Magnification of figure 4.7. As can be seen in this figure, abdomen distortions are above 1.5 mm.
5. $T_1$ Estimation

5.1 Methods

5.1.1 TO5 Phantom

To assess the accuracy of the $T_1$ quantification, a phantom with known relaxation times are required. For this purpose the test object 5 (TO5) phantom (Eurospin II Test System, Diagnostic Sonar Ltd., Edinburgh, UK) was used. This phantom is constructed for relaxation time measurements and supplied with 18 gel-filled test tubes. All tubes are filled with different substances with specific relaxation times, given in the phantom manual for three different temperatures (292 K, 294 K and 296 K) and at different magnetic field strengths. At 292 K using a field strength of 3T, the range of $T_1$ is from 200 ms to 1445 ms. The phantom is filled with water and supports the use of 12 different tubes, seen in figure 5.1.

![Figure 5.1: TO5 phantom used to estimate the relaxation time $T_1$. The phantom supports the use of 12 tubes with different relaxation times.](image-url)
5.1.2 Phantom Measurements

Phantom scans using VFA and "two point IRSE" quantification methods were performed at the PET/MR scanner at different occasions. Different phantom test tubes and sequence acquisition parameters was evaluated for each method. Furthermore, in the VFA method different combinations of flip angles and parameter settings were evaluated. Pulse sequence used for the VFA method was a 3D FSPGR with a 160×160 matrix. Different combinations of TI were used in the two point IRSE method: $T1_1 = 50$ ms, $T1_2 = 700$ ms and $T1_1 = 50$ ms,$T1_2 = 1100$ ms. In this method a matrix size of 128×128 was used. Appendix D, tables D.1 and D.3 show the sequence acquisition parameters used in both methods. The room temperature was monitored using a an alcohol thermometer since the $T1$ estimation depends on the temperature (see section 2.2). The thermometer was placed next to the phantom in the scanner room between phantom scans. The temperature was recorded before the each scan in order to select the reference $T1$. $T1$-values and maps was calculated using the clinical used method, i.e. a program called MICE (Department of Radiation Sciences, Umeå University, SWE). This program allows complex image calculations in a neat and easily managed interface. Figure 5.2 shows an example on a work schedule in MICE.

5.1.3 Coil Dependence

Coil dependency of the $T1$ estimation was also investigated. A good quantification method must be independent of which coil is used. The anatomical regions were
$T_1$ estimations are performed at NUS are in prostate and brain examinations. Coils used in brain exams are a 24 channel head (HNU) coil and a large flex (RMRT) coil together with the built-in posterior (CMA) coil. Prostate exams are mainly done using an upper anterior array (UAA) coil and the CMA coil. Appendix D, table D.2 shows the sequence parameter settings used in this test.

5.2 Results

Figures 5.3 and 5.4 shows the quantified $T_1$ values estimated using VFA and two point IRSE. The quantified values are plotted together with the known $T_1$-values at 292 K (green dashed line). The room temperature was 292 – 293 K at all scan occasions. The accuracy of the true $T_1$-values is ±3% plotted with red dashed lines in figure 5.3a and figure 5.4a. The relative error in each measurement can be seen in figure 5.3b and figure 5.4b.

5.2.1 Assessment of the Coil Dependence

Figures 5.5 and 5.6 show the result of $T_1$ quantification using different coils. The VFA method, figure 5.5, was tested using four different coils; a eight channel head coil (8CH), HNU coil, RMRT coil and the UAA coil. When the flex coils (RMRT and UAA) are used the CMA coil is also activated. Figure 5.5a shows the quantified $T_1$-values and figure 5.5b the relative error. Figure 5.6 shows the same test using two point IRSE with the following coils; RMRT, UAA and HNU. The quantified $T_1$-values is shown in figure 5.6 and the relative error in figure 5.6b.
Figure 5.3: Quantified $T_1$-values using VFA with different angle combinations. The label in the lower graph is valid for both graphs.

Figure 5.4: Quantified $T_1$-values using two point IRSE, with two different combinations of $TI$. The label in the lower graph is valid for both graphs.
Results 5

**$T_1$ Estimation**

**VFA FSPGR, Different Coils, flip angles: 3°, 6°, 10°, 20°, 30°**

![Graph](image1)

**Figure 5.5:** $T_1$ measurements using VFA FSPGR with different coils acquired at different occasions. The label in the lower graph is valid for both graphs.

**Two Point IRSE, Different Coils, $T_{I1} = 50$ ms, $T_{I2} = 1100$ ms**

![Graph](image2)

**Figure 5.6:** $T_1$ measurements using two point IRSE with different coils. The label in the lower graph is valid for both graphs.
6. Discussion

The aim of this project was to develop a semi-automatic quality assurance strategy for MRI in RT. The project was divided into three different parts where methods to assess the image quality, spatial accuracy and the quantitative images were included. One method to assess the scanner performance regarding the image quality was developed and clinically implemented. The spatial accuracy was quantified and a method to study the reproducibility of the distortions was developed. Using the clinical method for $T_1$-mapping the accuracy of two $T_1$ quantification methods were evaluated. The assurance strategy was showed to be adequate, but unfortunately both quantification methods used gave unsatisfactory $T_1$ estimations.

6.1 Image Quality

Almost all ACR tests performed at the PET/MR scanner were within the ACR acceptance levels. Occasional measurements were outside the acceptance levels in the; phantom diameter test, high-contrast resolution test, slice position accuracy test and slice thickness test. The reason for some extreme outliers in the $T_1$ weighted sequence was due to operational errors. The coil was not inserted correctly making the phantom images very noisy, which affected the length measurement, PIU, noise and SNR test. The high-contrast resolution measurements, figure A.2, outside tolerance is measurements from the $T_2$ double echo with longer echo time, 80 ms. This sequence has the lowest signal, which is the reason for not resolving as many object. Furthermore, some other deviations are observed, for example in the slice thickness test, see figure 3.7, where the measured slice thickness is sometimes closer to 6 mm than 5 mm. The reason for this deviation is unknown. The slice position accuracy test, figure A.3, shows a distinctive pattern where the measurements fluctuates the first months and the stabilizes. The reason for the varying results in the beginning is probably due to operational errors, that reduced when the slice position coordinates were saved in the protocol. All PIU measurements, figure 3.6, were within the accepted minimum of 82%. However, the measured PIU for the $T_1$-weighted sequence fluctuates more than the $T_2$-weighted double echoes. The sequence that has the highest PIU is the $T_2$-weighted double echo with longer echo time, 80 ms while the $T_1$-weighted sequence has the lowest. In the ghosting ratio test, all measurements were below the maximum accepted ratio of 0.025, see figure A.4. However, ghosting artifacts were seen occasionally in slice 5 and slice 1. The magnitude of the artifact were not constant and always more distinct in slice 5. The artifact was seen in all scanners where the ACR protocol was used. To keep track of this artifact an additional plot was added to the test, see figure A.5.
The SNR test, figure 3.5, showed a trend, especially in the $T_2$ weighted double echo sequence were the measurements separated in to two groups. One group with a higher SNR recorded and one with slightly lower SNR, see figure 3.5. The same trend was seen in the other two sequences but with less separation. This was traced back to the noise which has the same trend. Why this occurs is unclear and the question has been raised to the manufacturer. Both echoes are affected by this unexplained phenomenon, see figure A.6.

### 6.1.1 Future Work

A possible cause of failure on some of the ACR test is due to the coil performance, see appendix B. Therefore, a clinically used coil should be of more interest in ACR tests. Unfortunately, the head coil mostly used in the clinic, HNU, does not fit the ACR phantom. An eight-channel coil sparsely used in the clinic has been used for some measurements. The phantom is positioned tilted due to the coil design and oblique slices must be used. Adding this uncertainty makes this coil inappropriate to use as a daily QC coil. However, as MRI is more and more integrated in RT-planning a dedicated table top, a flex coil support to accommodate the coil, a plastic index tray and a lock bar is purchased. To use this setup with the ACR phantom would require a holder to stabilize the phantom and place the axial center of the phantom centered in the coil. A CAD model of the holder was designed and will soon be printed by the 3D printer available at NUS. This would also reduce a possible source of systematic errors i.e. operational error in the phantom positioning which would reduce some of the variations in the measured data. The low-contrast resolution ACR test was not satisfactory implemented in the ACR analysis program. Different approaches were tested and rejected since the results differed too much from the results obtained using the human eye. In the approach with best results ROIs covering each low-contrast dot were compared to the closest region outside the ROI. An intensity threshold was used to declare the dot as resolved or unresolved. This approach gave good results, comparable to results obtained by visual investigation, except in images with higher noise. Another future implementation is to couple the acceptance values of the test results to system that gives an alert when measurements are observed outside tolerance.

### 6.2 Spatial Accuracy

The spatial distortion in the PET/MR scanner was quantified using the large FOV phantom and the constancy of the distortion can be seen in figure 4.3, were minor changes was seen when comparing difference scans. However, in some of the difference images one spot was identified where the difference is 2 mm (or more). This spot is located at the same place in each scans, in the upper left corner at $z = -225$ mm. A possible reason for this is operational error
i.e. that the phantom was mispositioned slightly or interpolation errors. When studying the BW comparison, figures 4.5, 4.6 and C.1 to C.4 some difference is seen in the periphery of the FOV altering the BW, but inside a FOV of 20×20 cm² the difference is not significant. Furthermore, the distortion became much more significant without using 3D correction, hence one conclusion is that 3D correction is a necessary factor using a large FOV. Since 3D correction only can be used in 3D sequences and 2D sequences with zero slice gap, this is the obvious choice when acquiring large FOV MR images. Furthermore, figures 4.4, 4.7 and 4.8 are useful images in the discussion of image distortion in MR, informing technicians and radiologist.

6.2.1 Future Work

An improvement when studying the geometrical distortion is to correct the phantom geometry to fit the scan table. The paintball markers inside the phantom is only distributed in the periphery in some of the phantom images. For example at distances $z = \pm 25$ mm and $z = \pm 75$ mm. These planes are of high interest since the patient images often are acquired at these distances from isocenter. Interpolation of these planes became unsatisfactory, and an improvement of the phantom would be to include more markers at these positions. Furthermore, distortion maps for sequences other than FGRE should be produced, especially the clinically used sequences for studying the abdomen or pelvic area since these anatomical areas often requires a large FOV. Using the obtained information about the distortions patient image correction is possible. The can be done using the vector field obtained by the spatial analysis program. In order to correct patient images the distortion is required to be constant in all scans, or distortion maps created at the same occasion as the patient scan. To verify that the distortion is consistent, two identical phantom scans can be performed at the same occasion without repositioning the phantom.

6.3 $T_1$ estimation

As can be seen in figure 5.3 the quantified $T_1$ values using VFA (FSPGR) varied from the true $T_1$ values, especially at the higher values. According to the relative error in figure 5.3 the absolute error increases with increasing $T_1$. Using more than three flip angles yield quantifications closer to the true values, consistent with the theory. Unfortunately, the measurement time gets very long and a maximum of five angles can be used in the clinic due to time constraints. Possible reasons for the poor performance of the VFA metod is that the spoiling in the sequence works is insufficient, the SNR gets too low in regions with high $T_1$ and that no $B_1$ corrections were used. The effect of additional sequence acquisition parameters such as, 3D correction, no phase wrap (NPW) and using SPGR instead of FSPGR was evaluated. Unfortunately, no improvement of the method
was seen. Furthermore, no significant difference was seen when using other coils, figure 5.5. Two point IRSE, figure 5.4, showed a more stable quantification of $T_1$ over the available $T_1$ range compared to the VFA method. However, using $TI_2 = 700$ ms gave less accurate quantification compared to $TI_2 = 1100$ ms, at high $T_1$-values. Similarity, the quantification of low $T_1$ was less accurate using $TI_2 = 1100$ ms. This is consistent with the theory and implies that $TI_2$ must be of the same order as the estimated $T_1$. Two point IRSE was evaluated in one head-and-neck patient where the estimated value of CSF unfortunately was nowhere close to the true CFS $T_1$-value (400 ms compared to 4350 ms). CSF occupies the ventricular system, a series of cavities filled with CSF inside the brain. It can exist in tumors and such bad quantification affects the $T_1$-map negatively. Using a $TI_2 = 4$ s may improve the clinical result but requires a scan time not clinically acceptable. This method probably gives a more reasonable result used in prostate exams since $T_1$-values in this region are closer to the ones available in the TO5 phantom ($\sim 1600$ ms).

### 6.3.1 Future Work

An investigation about the clinical functionality of two point IRSE in prostates exams should been done. It is possible that this method gives better results in this anatomical area. If this is the case, anatomical specific quantification protocols should be used. Furthermore, the reason for the incorrect quantification using VFA must be found. A fist step of doing this could be to map the $B_1$ field, and verify that the spoiling works correctly.
7. Conclusions

The automatic ACR analysis program developed to study the image quality of MRI facilitates the process of image quality verification. The test results are easily obtained through a website where both technicians, engineers and physicists working with the scanner can access them. As seen from the ACR test results, the image quality parameters studied are constant over time and within reasonable acceptance levels.

The developed method for assessing the spatial accuracy showed that the distortion is reproducible and thus fairly constant over time. It also showed that inside a 20×20×20 cm³ volume the distortions were less the 1 mm, while outside this volume distortions larger than 2 mm was observed. In conclusion, using a larger FOV the body contour becomes distorted (>2 mm) and the distortions become clinical relevant, while in target and OAR delineations the distortion are small and not clinically significant.

Using the clinical $T_1$-mapping method for quantifying $T_1$ was feasible. However, the $T_1$ estimations using VFA and two point IRSE method gave estimations with unaccepted uncertainties. Two point IRSE was shown to be a suitable method when quantified $T_1$ using the TO5 phantom, i.e. in a $T_1$ range of 200 ms to 1445 ms. The clinical measurement using the same method gave however unsatisfactory results. The VFA method gave estimations with large relative errors during phantom measurements. Since no satisfactory method was found, the constancy of the $T_1$ quantification was not evaluated.
8. Bibliography


A. ACR Tests Results

**Figure A.1:** Phantom diameter measurement with the ACR phantom in slice 5. The green band indicates the acceptance level of the diameter measurement (19±0.2 cm). The diameter is measured in x and y directions, and two diagonals (45° and 135°). All length measurements are plotted with different symbols.

**Figure A.2:** Results from the ACR high-contrast resolution test. A score of 12 for each pulse sequence is required to pass the test.
Figure A.3: Results of ACR slice position accuracy test, evaluated in slice 1 and 11. Each pulse sequence has two measurements, symbolized with different marker types. The acceptance level, 0±5 mm, is illustrated with the green band and applies to all measurements.

Figure A.4: Results of the ACR ghosting ratio test measured in slice 7 according to the test manual [9]. The maximum ghosting ratio accepted according to the test manual is 0.025.
Figure A.5: Additional ACR ghosting ratio test in slice 5 due to visible ghosting artifacts in this slice.

Figure A.6: Measured noise values from a small ROI in ACR slice 7, used in the SNR test. Image noise was added since fluctuation in the SNR test was observed. This pattern can be deduced to the noise as can be seen in this figure.
B. Reasons for Failure of the ACR Tests

Table B.1: Common causes of failure for the seven ACR tests. [9]

<table>
<thead>
<tr>
<th>Test</th>
<th>Most common cause of failure</th>
</tr>
</thead>
</table>
| Geometry                    | Miscalibrated gradients\ 
|                             | $B_0$ inhomogeneities\ 
|                             | Too low acquisition bandwidth                                                             |
| High-contrast resolution    | Excessive image filtering\ 
|                             | Poor eddy current compensation\ 
|                             | Excessive image ghosting (*due to instability of the measured signal between pulse-cycles, origin from the receiver, transmitter or gradients*)\ 
|                             | Gradient amplifier instability\ 
|                             | Phantom motion or vibration\ 
|                             | Geometric errors                                                                          |
| Slice thickness accuracy    | RF amplifier nonlinearity (*can cause distorted RF pulse shapes*)\ 
|                             | Extremely bad gradient calibration\ 
|                             | Poor gradient switching performance                                                        |
| Slice position accuracy     | Error by the scanner operator\ 
|                             | Bad gradient calibration\ 
|                             | Magnetic field inhomogeneity\ 
|                             | Error in the table positioning mechanism                                                  |
| Image intensity uniformity  | Poor positioning or centering of the phantom\ 
|                             | Image ghosting\ 
|                             | Mechanical failure of components in the head coil\ 
|                             | Failure of coupling between coils\ 
|                             | Phantom motion or vibration                                                               |
| Percent-signal ghosting     | Phantom motion or vibration\ 
|                             | Hardware problem\ 
|                             | Instability of the measured signal from pulse cycle to pulse cycle (*origin from receiver, transmitter or gradients*) |
| Low-contrast object detectability | Incorrectly positioned slices\ 
|                             | The phantom is tilted\ 
|                             | Ghosting artifacts\ 
|                             | Inadequate SNR                                                                            |
C. Spatial Distortion Maps

Figure C.1: Image distortion using different BW and no 3D correction. The total error is used to illustrate the distortion in different planes, at \( z = 50 \) mm.

Figure C.2: Image distortion using different BW and no 3D correction. The total error is used to illustrate the distortion in different planes, at \( z = -100 \) mm.
C Spatial Distortion Maps

Image Distortion, \( z = 100 \text{ mm} \), Different BW

Table of values:
- Axial: 244.14 Hz/pixel, 488.28 Hz/pixel, 976.56 Hz/pixel, 976.56 Hz/pixel (No 3D-corr)
- Sagittal
- Coronal

**Figure C.3:** Image distortion using different BW and no 3D correction. The total error is used to illustrate the distortion in different planes, at \( z = 100 \text{ mm} \).

Image Distortion, \( z = -200 \text{ mm} \), Different BW

Table of values:
- Axial: 244.14 Hz/pixel, 488.28 Hz/pixel, 976.56 Hz/pixel, 976.56 Hz/pixel (No 3D-corr)
- Sagittal
- Coronal

**Figure C.4:** Image distortion using different BW and no 3D correction. The total error is used to illustrate the distortion in different planes, at \( z = -200 \text{ mm} \).
D. $T_1$ estimation  
Parameters Settings

**Table D.1:** Pulse sequence acquisition parameters used for studying the different angles used for $T_1$ estimation by the VFA method. TE min full implies that the scanner optimizes TE as short as possible.

<table>
<thead>
<tr>
<th>Method</th>
<th>TR (ms)</th>
<th>TE (ms)</th>
<th>Flip angles</th>
<th>Coil</th>
</tr>
</thead>
<tbody>
<tr>
<td>VFA</td>
<td>5</td>
<td>min full</td>
<td>$2^\circ, 7^\circ, 14^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$2^\circ, 7^\circ, 21^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$2^\circ, 14^\circ, 21^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$7^\circ, 14^\circ, 21^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$17^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$1^\circ$-$30^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$5^\circ, 10^\circ, 50^\circ, 20^\circ$</td>
<td>Quadrature (no 3Dcorr)</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>min full</td>
<td>$5^\circ, 10^\circ, 50^\circ, 20^\circ$</td>
<td>Quadrature (no turbo mode)</td>
</tr>
</tbody>
</table>

**Table D.2:** Pulse sequence acquisition parameters used for investigating the coil dependence for $T_1$ estimation using the VFA method.

<table>
<thead>
<tr>
<th>Method</th>
<th>TR (ms)</th>
<th>TE (ms)</th>
<th>Flip angles</th>
<th>Coil</th>
</tr>
</thead>
<tbody>
<tr>
<td>VFA</td>
<td>6.1</td>
<td>min full</td>
<td>$3^\circ, 6^\circ, 10^\circ, 20^\circ, 30^\circ$</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>6.1</td>
<td>min full</td>
<td>$3^\circ, 6^\circ, 10^\circ, 20^\circ, 30^\circ$</td>
<td>UAA + CMA</td>
</tr>
<tr>
<td></td>
<td>6.1</td>
<td>min full</td>
<td>$3^\circ, 6^\circ, 10^\circ, 20^\circ, 30^\circ$</td>
<td>Flex small + CMA</td>
</tr>
<tr>
<td></td>
<td>6.1</td>
<td>min full</td>
<td>$3^\circ, 6^\circ, 10^\circ, 20^\circ, 30^\circ$</td>
<td>8CH</td>
</tr>
</tbody>
</table>

**Table D.3:** Pulse sequence acquisition parameters used for comparing the different method for $T_1$ estimation. Abbreviation in the table: ETL = Echo Train Length.

<table>
<thead>
<tr>
<th>Method</th>
<th>TE</th>
<th>TR</th>
<th>$T_1$</th>
<th>$T_1$</th>
<th>ETL</th>
<th>Flip angle</th>
<th>Coil</th>
</tr>
</thead>
<tbody>
<tr>
<td>Two point IRSE</td>
<td>45</td>
<td>7000</td>
<td>50</td>
<td>700</td>
<td>24</td>
<td>180°</td>
<td>HNU</td>
</tr>
<tr>
<td></td>
<td>45</td>
<td>7000</td>
<td>50</td>
<td>1100</td>
<td>24</td>
<td>180°</td>
<td>HNU</td>
</tr>
</tbody>
</table>

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