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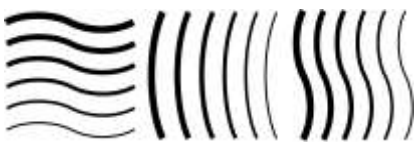
Quality Assurance of MRI for Radiotherapy

NEDERLANDSE COMMISSIE VOOR STRALINGSDOSIMETRIE

Report 36 of the Netherlands Commission on Radiation Dosimetry

March 2023



Netherlands Commission on Radiation Dosimetry

Subcommittee 'MRI QA for RT'

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Preface

The Nederlandse Commissie voor Stralingsdosimetrie (NCS, Netherlands Commission on Radiation Dosimetry, http://www.radiationdosimetry.org) was officially established on 3rd September, 1982 with the aim of promoting the appropriate use of dosimetry of ionising radiation both for scientific research and for practical applications. The NCS is chaired by a board of scientists, made up via recommendations from the supporting societies, including the Nederlandse Vereniging voor Radiotherapie en Oncologie (Dutch Society for Radiotherapy and Oncology), the Nederlandse Vereniging voor Nucleaire Geneeskunde (Dutch Society of Nuclear Medicine), the Nederlandse Vereniging voor Klinische Fysica (Society for Medical Physics of the Netherlands), the Nederlandse Vereniging voor Radiobiologie (Netherlands Radiobiological Society), the Nederlandse Vereniging voor Stralingshygiëne (Netherlands Society for Radiological Protection), the Nederlandse Vereniging voor Medische Beeldvorming en Radiotherapie (Dutch Society for Medical Imaging and Radiotherapy), the Nederlandse Vereniging van Klinisch Fysisch Medewerkers (Dutch Society for Medical Physics Engineers), the Nederlandse Vereniging voor Radiologie (Radiological Society of the Netherlands) and the Belgische Vereniging voor Ziekenhuisfysici/Société Belge des Physiciens des Hôpitaux (Belgian Hospital Physicists Association) and expanded with a representative from the Dutch Metrology Institute VSL. To achieve its aims, the NCS carries out the following tasks: participation in dosimetry standardisation, promotion of mutual comparisons of dosimetry, drafting of dosimetry protocols and the collection and evaluation of physical data related to dosimetry. Furthermore, the commission shall establish or maintain links with national and international organisations concerned with ionising radiation and promulgate information on new developments in the field of radiation dosimetry.

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Quality assurance of magnetic resonance imaging for use in radiotherapy

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Summary

In Spring 2017, the NCS installed a subcommittee to develop guidelines on the quality assurance of magnetic resonance imaging (MRI) for radiation therapy. The subcommittee had its kick-off meeting on June 30, 2017.

MRI is increasingly used for different applications within the radiation therapy workflow. This report describes four of these applications and their requirements for quality control (QC), namely: (i) MRI not registered to computed tomography (CT); (ii) MRI co-registered to CT; (iii) MR-only treatment planning for external beam radiotherapy; and (iv) online MR-guided radiotherapy.

After the definition of this scope in Chapter 1, the basic principles of MR image formation required for the proposed QC tests are covered in Chapter 2. In Chapter 3, recommendations are given for radiotherapy-specific MRI acquisition protocols. Subsequently in Chapter 4, the specific QC tests including relevant tolerances are described for the four different applications. The report concludes with Chapter 5 on basic MR safety concepts and the introduction of devices and equipment into the scanner room.

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Abbreviations and acronyms

AAPM	The American Association of Physicists in Medicine
ACR	American College of Radiology
ASTM	American Society for Testing and Materials
a.u.	arbitrary units
B_0	main magnetic field of MRI scanner
bSSFP	balanced Steady-State Free-Precession
CBCT	cone-beam CT
CF	centre(resonance) frequency
CNR	contrast-to-noise ratio
CT	computed tomography
CT-sim	refers to when the CT is used for treatment simulation, i.e. treatment planning
DAC	digital analogue converter
DICOM	Digital Imaging and Communications in Medicine
DSC	dynamic susceptibility-weighted contrast
DSV	diameter spherical volume
DWI	diffusion-weighted imaging
EBRT	external beam radiotherapy
EPI	echo-planar imaging
ERE	electron return effect
FDA	Food and Drug Administration
FFE	fast-field echo
FFP	feet-first prone
FFS	feet-first supine
FID	free induction decay
fMRI	functional MRI
FoV	field-of-view
FWHM	full-width-at-half-maximum
GE	gradient-echo (also FFE)
GNL	gradient non-linearity
HFP	head-first prone
HFS	head-first supine

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HU	Hounsfield units
IPEM	Institute of Physics and Engineering in Medicine
MAE	mean absolute error
ME	mean error
MR	magnetic resonance
MRI	magnetic resonance imaging
MRL	MR-linac
MRS	magnetic resonance spectroscopy
MR-sim	refers to when the MRI is used for treatment simulation, i.e. treatment planning
NEMA	National Electrical Manufacturers Association
NVKF	Society for Medical Physics of the Netherlands (Ned. Vereniging voor Klin. Fysica)
OAR	organ at risk
PET	positron emission tomography
PIQT	periodic image quality test
ppm	parts per million
PRF	pulse repetition frequency
PTV	planning target volume
QA	quality assurance
QC	quality control
RD	radiology
RF	radiofrequency
RMS	root-mean-square
ROI	region of interest
RT	radiotherapy
RTT	radiation therapy technologist
SAR	specific absorption rate
SBRT	stereotactic body radiation therapy
sCT	synthetic CT
SD	standard deviation
SE	spin-echo
SNR	signal-to-noise ratio
SSIM	structural similarity index measure
SRS	stereotactic radiosurgery

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STE	stimulated echo
TE	echo time
TG	transmitter gain
TPS	treatment planning system
TR	repetition time
TRUFI	true fast imaging (with bSSFP)
UTE	ultrashort echo time
WFS	water-fat shift

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1 Introduction

1.1 Background

The aim of radiotherapy (RT) for cancer treatment is to irradiate the tumour with a sufficiently high dose to eradicate it, whilst sparing (vital) healthy tissue in order to minimise radiation-induced toxicities. The radiotherapy preparation consists of acquisition of a computed tomography (CT) scan of the patient in treatment position and often a magnetic resonance imaging (MRI) scan and/or a positron emission tomography (PET)/CT scan. Typically, target volume definition is performed on a CT simulation scan, which is needed for accurate treatment planning as well. Due to the superior soft tissue contrast of MRI compared to CT, the use of MRI scans to guide delineation of target and organs at risk (OAR) has increased over the last decades and is considered standard clinical practice for various treatment sites such as prostate cancer and brain metastases [1,2]. However, in order to fully benefit from the potential of MRI, the MR imaging needs to be of high quality in terms of image contrast, signal-to-noise ratio (SNR) and geometric fidelity. Mediocre MR image quality can lead to suboptimal radiation treatments. The aim of the current NCS report is to give recommendations for quality assurance (QA) tailored to the various applications of MRI in radiotherapy. It is the best practice as defined by a panel of expert users.

1.2 MRI – a diagnostic modality applied in radiotherapy

The primary technical requirements for diagnostic use of MRI scanners are not necessarily the same as the requirements for radiotherapy. Historically, radiologists are less interested in exact geometrical information compared to radiation oncologists, and even today many MRI scanners do not have manufacturer specifications for geometrical accuracy. The use of MR images for target volume delineation has made requirements for geometric fidelity of MRI scanners more important. Radiology (RD) departments should all have basic QC (quality control) protocols in place to check the quality of MRI for diagnostic use as proposed in several guidelines [3–5]. This NCS report takes the QC protocol for diagnostic MRI, as formulated in the ”Leidraad Kwaliteitscontrole Radiologische Apparatuur” by the Society for Medical Physics of the Netherlands (NVKF), as a basis and introduces additional QC tests that are required for application in radiotherapy.

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1.3 Applications of MRI in radiotherapy

The report describes four different applications of MRI in RT, each with their own requirements for MR image quality and hence for MRI QC:

1. ***MRI not registered to CT***

   Visual assessment of diagnostic imaging in addition to the utilisation of a (primary) CT for radiation treatment planning. In this case the MRI and CT simulation (CT-sim) scans are not registered, since patient positioning is different between MRI and CT. It is not recommended to perform target volume definition on this MRI; the MRI scan can be used only to facilitate tumour definition in a rough, general fashion. For this application the diagnostic-specific QC is deemed sufficient, under the assumption that the MRI quality is appropriate for tumour definition.

2. ***MRI co-registered to CT***

   MR images are registered to the CT images, to aid the delineation of target volumes and/or organs at risk. For this application, patient positioning between the MRI and CT images should be similar. Therefore, RT-specific measures (for example, flat couch top, RT-specific scanning protocols) should be in place. It is advised to implement additional QC to monitor geometrical fidelity and ensure that the location of target and OAR are not displaced in MR images.

3. ***MR-only treatment planning for external beam radiotherapy***

   An MR-only simulation workflow does not require a CT scan but uses only the MRI scan for delineation, treatment planning and position verification. In practice, MR images are converted to a synthetic CT (sCT) that is used for both treatment planning and as reference for position verification. This application requires an RT-specific configuration of the MRI scanner (for example external lasers, coil bridge) and adds to the requirements on geometric fidelity. The MRI scan needs to be geometrically correct over the full body contour within the field-of-view (FoV). Moreover, QA is required to verify the Hounsfield Units (HU) of the sCT. As the generation of sCTs is still quite new, QA on this topic has not been fully established yet.

4. ***Online MR-guided radiotherapy***

   An MR-guided adaptive radiotherapy workflow uses an MR-linac (MRL); a linear accelerator (linac) integrated with an MR system. Such a workflow requires additional QC of the MR system and its interplay with the linac to ensure accurate delivery.

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The type of MRI QC and the tolerance levels depend on the application of MRI in the RT workflow. In this document, we will present RT-specific QC, tailored to the type of application of MRI in RT with application-specific tolerances.

1.4 Scope of this report

This report aims to provide the reader with basic principles of MR imaging required to understand image formation and related image artefacts that are relevant for RT. It is, however, not a full and comprehensive explanation of MR physics. Furthermore, this report describes current practice of MRI in RT at the date of writing. Future applications of MRI in RT planning may impact MRI QC requirements. This report covers recommended QC for typical clinical use of MRI in RT for the four applications mentioned in the previous section. Functional MRI [6], 4D-MRI [7], quantitative MRI [8] and use of MRI for brachytherapy are beyond the scope of this report.

1.5 Contents of this report

The chapters of this report are organised as follows:

2. ***Theoretical background of MRI***

   This chapter covers the basic principles of MR image formation that are required to understand the subsequent chapters of the report.

3. ***Recommendations for radiotherapy-specific MRI protocols***

   This chapter covers RT-specific considerations when designing MRI acquisition protocols for radiotherapeutic applications. It will provide a starting point for departments considering utilising MRI for radiotherapy purposes.

4. ***Recommendations for system QC***

   Here, the specific QC tests are described for the four different MRI applications, as well as the accompanying tolerances on the QC parameters.

5. ***MR safety and introducing devices in the scanner room***

   This chapter explains basic MR safety concepts with focus on the introduction of (RT) equipment and devices in the scanner room.

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2 Theoretical background of MRI

2.1 Scope of this chapter

MRI is often considered a challenging subject to master, due to the diversity of physics and engineering principles that are necessary to understand many of the concepts in MRI. Because it is beyond the scope of this report to cover all the fundamentals of MR physics, knowledge of the basics of MRI is assumed.

Readers who would like to refresh their knowledge on basic MR physics principles are kindly referred to the excellent literature that is available [9–13]. This chapter focusses only on those aspects that are most important to the applications of MRI for radiotherapy and the corresponding QC: MR system components, spatial encoding, and geometric accuracy.

2.2 MR system overview

MRI is an imaging technique usually based on magnetic resonance of hydrogen nuclei, also referred to as 'spins'. The principles of MR physics can be divided into three different aspects: signal formation, contrast formation, and image formation (spatial encoding). These three basic aspects of MR physics translate into the various system components of the MRI scanner:

- An (often superconducting) magnet produces a strong, static magnetic field. The magnetic field is required for polarisation of the proton spins.
- Radiofrequency (RF) transmit and receive coils excite the proton spins and detect the MR signal, respectively. The transmitted RF signal induces the emission of an RF signal by hydrogen nuclei, while the receive coils are highly sensitive to pick up this emitted signal. Since the gyromagnetic ratio of hydrogen nuclei is 42.58 MHz/Tesla, the MR systems that are currently used for radiotherapy purposes operate between 15 MHz (0.35 T) and 128 MHz (3.0 T).
- Gradient coils are used to produce dynamic magnetic field variations that are superimposed onto the static magnetic field. Each MRI system contains three sets of gradient coils, which are used to spatially encode the MR signals. The gradient coils are of specific interest for RT planning because they have a direct impact on the spatial fidelity of the MR images.
- Patient couch and immobilisation equipment support the patient during the exam. It is necessary to ensure that all these items and do not interfere with the MRI acquisition (i.e. do not have a negative impact on the image quality and spatial integrity of the images, nor on patient safety).

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2.3 Image formation: basic theory

The Larmor frequency (or: resonance frequency) of the nuclei has a linear relationship with the magnetic field strength. For slice selection, a weak magnetic field that changes linearly with position is superimposed on the static magnetic field by the gradient coil. For transverse imaging the z-axis is used, which is oriented from head to toe. By applying a gradient along the z-axis the Larmor frequency varies linearly in the caudal-cranial direction. If the patient is then exposed to an RF excitation pulse with a narrow range of frequencies, only those nuclei in the caudal-cranial position where the Larmor frequency matches the frequency of the RF excitation pulse will actually be excited and subsequently produce an MR signal. As a result, a transverse plane is excited. After the RF excitation pulse, the gradient used for slice selection is switched off.

After slice excitation, the origin of the MR signal still has to be encoded along the remaining two dimensions. In order to achieve this, a second magnetic field gradient is applied orthogonally to the slice selection gradient *during* signal reception. The spins at each position along this gradient axis now have a unique resonance frequency, which allows us to decode the positions along the gradient axis by using Fourier analysis. The direction of this second gradient is referred to as the frequency-encoding direction or readout-gradient direction.

The remaining (third) dimension is encoded by the so-called phase-encoding gradient. The phase-encoding gradient is a gradient of short duration that is applied between the slice excitation and frequency-encoding gradient. The goal of the phase-encoding gradient is to prepare the magnetisation with a certain phase profile just prior to the collection of each readout line. By varying the area of the phase-encoding gradient, a different amount of phase encoding is applied to each readout line, which allows reconstruction of the eventual image via a 2D Fast Fourier transform (FFT).

As the Fourier transform implies, the MRI signals are acquired in the (spatial) frequency domain, referred to as k-space. K-space consists of complex signals, which are transformed into image space via the FFT. While the resulting image is also complex, often only the magnitude of the image is stored and displayed.

2.4 2D vs. 3D imaging

In 2D multi-slice imaging, the spins are excited for each slice sequentially. A gap between slices is commonly used to avoid crosstalk between the slices due to imperfections in the RF slice profile. A small gap may lead to signal overlap between consecutive slices. On the other hand, a large gap increases chances of missing small pathological features, since the tissue within the gap is not imaged

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at all. 3D volume imaging avoids these issues: instead of only a thin slice, the entire imaging volume is excited by the RF pulse. The spatial localisation in the third dimension is created through application of a second phase-encoding gradient, usually referred to as a 3D- or partition-encoding direction.

The advantages of volume imaging include a higher intrinsic signal-to-noise ratio, since the MR signal is derived from the entire volume rather than from a single slice. In contrast, for 2D imaging the signal arises only from the selected slice while noise is still detected from the same volume. In addition, in 3D imaging, slices can be very thin and closely spaced without suffering from crosstalk, and isotropic voxels can be obtained, permitting the 3D data set to be analysed via successive reformatting in any plane with identical image quality. The main disadvantage of 3D imaging vs. 2D multi-slice imaging is a longer acquisition time.

2.5 Geometric accuracy of MR images

When a linear gradient field is superimposed on the main magnetic field, the result is a linear relationship between the resonance frequency and the spatial position. However, imperfections in the main magnetic field (B_0) homogeneity and gradient fidelity lead to geometric distortions of MR images [14]. These imperfections in the B_0 field may arise from imperfections in the magnet design or perturbations due to spatial variation in magnetic susceptibility in the patients themselves. Figure 1 illustrates how local magnetic field perturbations relate to spatial infidelity. Both system-related and patient-related causes of geometric distortion are undesirable for RT planning. By selecting a high receiver bandwidth (steep gradient in Figure 1), the distortions due to B_0 field homogeneity imperfections are reduced and the gradient imperfections become the most relevant source of distortion. Details of practical trade-offs in protocol optimisation are discussed in Chapter 3. The technical aspects of these trade-offs are briefly presented in paragraphs 2.6 to 22.8.

2.6 System-related geometric distortion

2.6.1 Gradient amplitude

The measured geometry scales linearly with the applied gradient amplitude. Therefore, a deviation in amplitude results in an identical deviation in the scale of the image. When the gradient amplitude is lower than indicated, the object will appear smaller on the image than in reality. The absolute scaling of the gradient amplitude needs to be calibrated for each gradient coil. Furthermore, as the amplitude

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depends on (stability of) gain factors in the signal chain, the calibration of the gradient amplitude needs to be verified regularly.

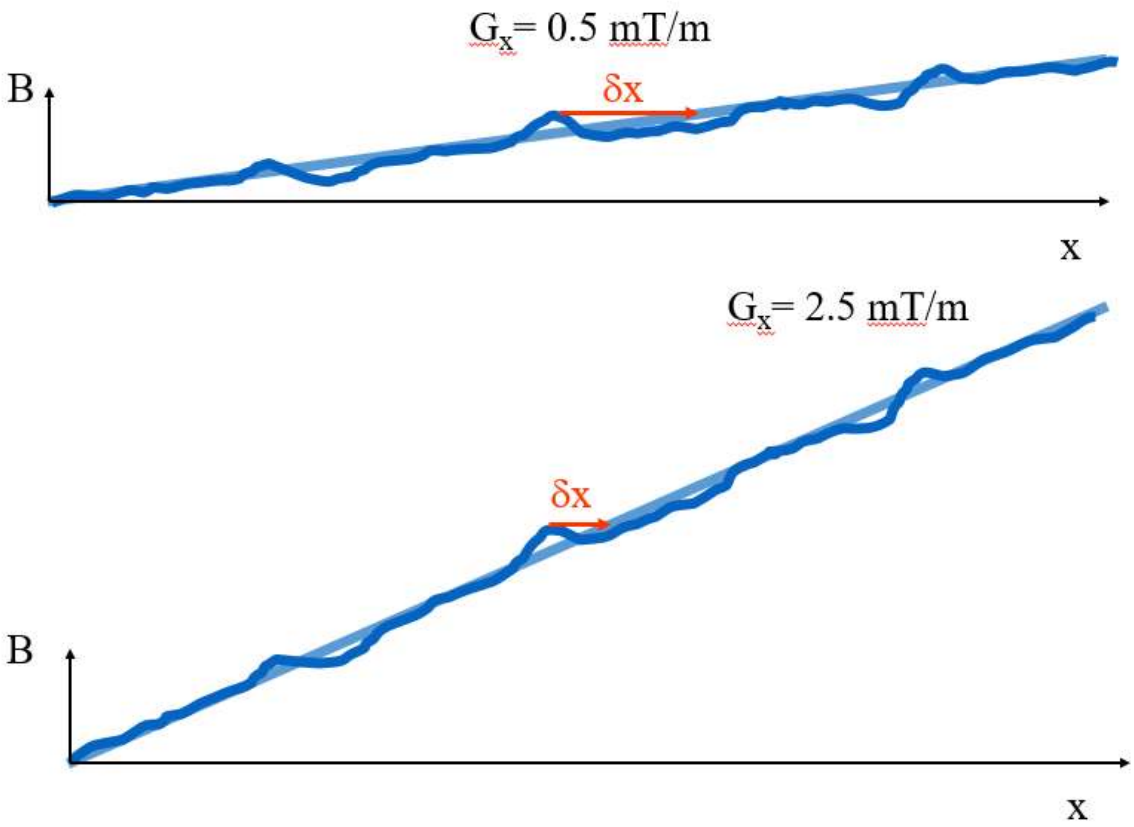

Figure 1: Schematic graphical explanation of geometrical infidelity due to local B_0 field inhomogeneity. The ideal linear gradient G_x (light blue) creates a linear relation between resonance frequency and position (x). If any perturbations of the B_0 field exist, e.g. resulting from the presence of air-tissue interfaces, the linear relation between position and B field strength is also broken. Let us assume this gradient is used for spatial encoding during signal readout (frequency encoding). The top graph shows a low frequency-encoding gradient and the bottom graph a higher gradient. The perturbations of the B_0 field result in a certain frequency shift, which translates into a large spatial error in the top graph, while the error is 5 times smaller for the high gradient in the bottom graph. On an MRI scanner, the amplitude of the readout gradient is controlled by setting the readout bandwidth (GE, Siemens, Canon) or the water-fat shift (Philips).

2.6.2 Gradient linearity

In practice, magnetic field gradients are not perfectly spatially linear in amplitude over the full image field of view, but flatten off towards the edges of the scanner (cf. Figure 2), resulting in a non-linear spatial encoding. This gradient non-linearity is a design feature of high-end gradients that reduces their peripheral nerve stimulation and thereby allows them to operate at higher slew rate and amplitude compared to spatially linear gradients. Additionally, physics laws (Maxwell's equations) dictate that non-linear concomitant fields necessarily exist. The geometric distortion resulting from these non-

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linear gradients increases with distance from the isocentre and can reach a few centimetres at the edges of the scanner's maximum FoV [15,16]. Modern MRI scanners have integrated software to correct for these gradient non-linearities. Implementation details differ between vendors.

Gradient linearity and its effect on slice excitation

When the applied gradient field is not exactly linear, the excited slice will not be a true plane, but will be warped instead. This is also known as the "potato chip" effect. The effect increases with distance from isocentre and may result in errors up to 4 mm on slices with a FoV of 20 x 20 cm^2 that are located 10 cm away from isocentre (when no correction is applied) [17]. While these distortions are easily corrected for in 3D and multi-slice 2D acquisitions, they are very difficult to correct for single slice acquisitions and should therefore be considered for single-slice motion monitoring sequences planned far from the isocentre on MRL systems.

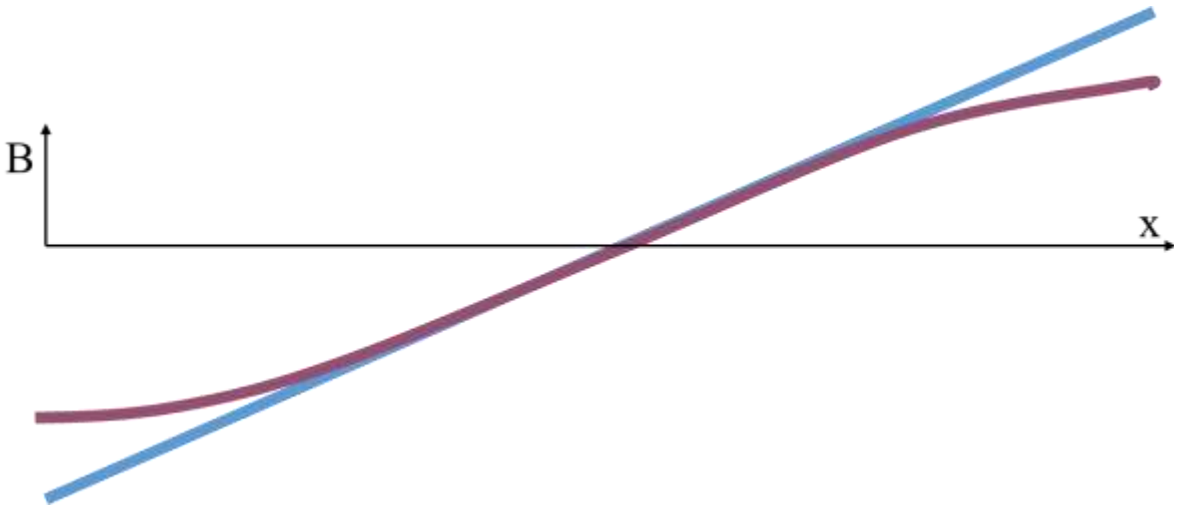

Figure 2: Schematic graphical explanation of geometrical infidelity due to gradient non-linearity. The ideal linear gradient is shown in blue. The purple line shows a non-linear gradient that flattens off away from the isocentre, resulting in geometrical distortion.

2.6.3 Homogeneity of the static magnetic field

Modern MRI magnets deliver a highly homogeneous magnetic field at the isocentre, with typical specifications below 0.1 ppm over a small spherical volume of approximately the size of a head. However, in the outer regions of a large FoV as used in body imaging, the homogeneity of the B_0 field can have a relevant impact on geometrical accuracy.

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Distortions arising from gradient non-linearity and magnetic field inhomogeneity are machine-specific, constant over time, and can be measured [18]. For MR-linac systems, the rotating gantry adds non-static perturbation of the field homogeneity, depending on the gantry position. Field homogeneity may be affected by foreign metallic objects in the scanner magnet (coins, keys etc), which can be monitored. Gradient non-linearity is a design property of the gradient coil. The correction is software-based and can be verified upon software changes. Gradient amplitude is also considered stable over time [19], but may need closer monitoring by regular measurements resulting in occasional recalibration.

2.7 Patient-related geometric distortion

Since patients contain air spaces and paramagnetic materials (e.g. implants) in various locations and concentrations, inserting a patient into an MRI scanner locally disturbs the static magnetic field due to changes in susceptibility, which can cause artefacts. Local field perturbations result in geometrical infidelity in frequency-encoding direction, but not in the phase-encoding direction. Note that in case of 3D imaging, the “phase encoding” technique is used for spatial localisation in both phase and partition (also referred to as slice or 3D) directions, thus, for 3D imaging geometrical distortion occurs only in one direction (the frequency-encoding direction).

2.8 Echo-planar imaging (EPI)

In contrast to regular sequences, for echo-planar imaging (EPI) the phase-encoding direction is effectively a frequency-encoding direction with exceptionally low readout-gradient amplitude [20]. In general, the geometrical distortion in the phase-encoding direction is therefore unacceptably high for radiotherapy planning purposes. EPI is widely used for diffusion-weighted imaging (DWI), neuro perfusion (dynamic susceptibility-weighted contrast, DSC) and functional MRI (fMRI). With EPI, areas of signal loss due to dephasing of MR signals in the presence of field inhomogeneity and by bright areas due to signal pile-up are common.

2.9 Magnetic field strength

With respect to magnetic field strength: each field strength has its own pros and cons. At a higher field strength, it is possible to acquire more signal (higher SNR) given the acquisition time. Theoretically, the MR signal scales linearly with the field strength, but in practice the SNR gain is often lower. As an example, we compare a 1.5 T with a 3 T scanner. The 1.5 T scanner suffer less from susceptibility

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artefacts than high field scanners (less local perturbation of the magnetic field), therefore, on 3 T systems it may be necessary to increase the readout bandwidth to compensate for the increased associated geometrical distortion, thereby reducing the gain in SNR. This limits the SNR gain of higher field strength in practice. For MR-sim 1.5 T may also be preferred over 3 T for patients with metallic implants to minimise geometric distortions around the implant. The most important factor to take into account when choosing field strength and protocol is the intended use of the MR image.

2.10 Sequence optimisation

Optimising the quality of an MRI scan is a complex iterative process, generally involving a trade-off between SNR, image contrast, spatial resolution, geometric fidelity and sensitivity to artefacts (e.g. flow or motion artefacts) and scan duration, as illustrated in Figure 3. The scan duration also affects image quality in the sense that a long scan time increases chance of artefacts due to movement of the patient or his internal organs, besides discomfort for the patient. Most sequence parameters (e.g. repetition time, echo time, FoV, acquisition matrix size, readout bandwidth/water-fat shift) have a direct impact on these three acquisition properties.

The main difference between MRI for radiotherapy and diagnostic imaging lies in the fact that the trade-off between resolution, SNR, and time (scan duration) is made differently. In radiology, the contrast-to-noise ratio (CNR) determines whether a lesion can be detected or not. Diagnostic MRI protocols are therefore optimised to produce high SNR images in order to detect lesions even if the contrast difference with the surrounding tissue is low. Geometric fidelity is less important in radiology than detectability. In radiotherapy, however, geometric fidelity is key. Scans optimised for radiotherapy put more emphasis on high resolution and spatial integrity as both are essential for accurate delineation and treatment guidance. As a result, radiotherapy scans usually have lower SNR and often take longer than diagnostic scans. The practical considerations of MRI scan protocol optimisation for RT applications are discussed in Chapter 3.

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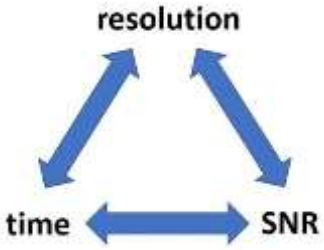

Figure 3: MRI pulse sequence parameter optimisation involves trade-off between spatial resolution, scan time and SNR.

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3 Recommendations for radiotherapy-specific MRI protocols

Each of the four different applications of MRI in RT has its own requirements for MRI quality and geometric fidelity. For each use, specific recommendations are outlined below.

3.1 MRI not registered to CT

For the work-up of radiotherapy patients, diagnostic images are used regularly as a visual aid to define the target volume.

3.1.1 Challenges

Typically, patients have not been scanned in RT treatment position, such that reliable accurate registration to the planning CT scan is often not feasible. In addition, there is neither sufficient knowledge nor control over the QA on the MR images. It is therefore highly recommended to always verify whether the diagnostic centres where the MRI scans originate from, use the QC protocols of the NVKF recommendations [5] or similar.

3.1.2 Recommendations

In this case, it is recommended to use the CT simulation and the MR images side-by-side only, without performing image registration. Diagnostic images generally have relatively thick slices and high SNR is prioritised over geometric accuracy. Geometric inaccuracies may therefore be too large to be suitable for delineation [21].

3.2 MRI co-registered to CT

In the case that delineation of the target volume or OAR is directly done on MR images, the MR images need to be registered to the CT scan. In most cases, the registration may be more accurate when the MR scan is acquired in RT treatment position. This facilitates target volume delineation with smaller inter-observer variation, as has been found in prostate, breast and brain [22].

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3.2.1 Challenges

For this application it is important that the geometric fidelity and resolution of the MR images are sufficiently accurate for delineation. In general, standard QC for diagnostic imaging is not sufficient to verify the geometric accuracy of the MRI scans for target volume delineation. Diagnostic sequences are usually optimised for SNR, which can be at the expense of geometric accuracy. The FoV is sometimes small and focused on the target area (e.g. one vertebra), while for a good registration with the planning CT scan, more information from the surrounding anatomy is needed.

In order to facilitate delineation on MR images, it is necessary to register the MRI images to the CT scan. If the accuracy of non-rigid registration methods cannot be guaranteed due to lack of QC tools, which is at the time of writing in general the case in current practice, rigid registration method is advised.

Standard ("diagnostic") positioning of the patient in the MRI scanner, for example on a non-flat couch top, with MRI coils placed on top of the patient touching the skin and ad-hoc immobilisation accessories, will result in a patient position different from the position on the CT simulation, which makes an accurate co-registration between CT and MR difficult. Also, the internal anatomy may vary which can make registration challenging.

3.2.2 Recommendations

When using MRI co-registered to CT, the following steps are recommended:

- Perform additional QC measurements on the MRI scanner to quantify geometric accuracy;
- Implement specific MRI scan protocols for radiotherapy patients;
- Check for each patient that the MRI acquisition was done according to protocol;
- Take remaining registration inaccuracies into account in the PTV margin.

These recommendations are described in the following sections.

Perform additional QC measurements on the scanner for geometric accuracy

To guarantee the geometric accuracy needed for accurate delineation of the target volume or OARs for RT, it is not sufficient to comply with the NVKF recommendations for QC of diagnostic MRI scanners [5]. In contrast to the first application, when the MRI scan is not registered to the CT scan, it is recommended to perform additional QC measurements on the MRI scanners for the application where

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the MRI is co-registered to the CT scan. The suggested measurements, tolerances and frequencies for these measurements are presented in Chapter 4. Some QC measurements need to be performed after maintenance and upgrades of the MRI scanner. Therefore, it is recommended to make clear agreements with the departments/centres where the MRI scans are performed about the procedure for clinical release of the MRI scanners after maintenance and upgrades.

Implement specific MRI protocols for radiotherapy patients

Similar patient positioning and geometry between MRI and CT-sim will facilitate the image registration. Therefore, it is recommended to reproduce the patient's treatment position as accurately as possible. For this purpose, flat couch tops and overlays, coil bridges and MR-compatible immobilisation devices are available. It is essential that the positioning of the patient with these devices is performed by trained professionals. Carbon fibre materials can be subject to RF heating and therefore it is recommended to avoid these materials. A laser-based patient positioning system is not necessary for this application but can help to reduce any rotation between the patient anatomy in the CT and MRI scan. Since the scans are performed at different time points, there will be differences in anatomy such as bladder and bowel filling. These differences can be mitigated by using the same patient instructions for the MRI scan as for the planning CT scan and radiotherapy treatment, such as diet instruction or bladder and bowel filling protocols. Any remaining uncertainties should be accounted for in PTV margins.

As treatment planning for this application will still be CT-based, the body contour on the MRI scan does not have to be accurate or even visible, as long as image registration is feasible. However, for precise delineation, geometric accuracy of the MRI scan in the region of interest is essential. It is important to consider carefully what this region of interest comprises (i.e. target volume, organs at risk, structures used for co-registration, or a combination of these regions) and what its size is. This also dictates whether the image registration needs to be done over the full FoV or only locally.

When scanning head and neck patients in RT position, another practical issue arises: the use of diagnostic receiver coils having a fixed geometry is not possible anymore. These coils have a fixed construction, leaving no space for positioning the patient in a fixation mask. This fixation mask is necessary to achieve good image registration results [23]. The use of flexible receiver coils allows for this fixation mask, while still placing a receiver coil close to the scanned anatomy. All vendors offer coil

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bridges/holders to facilitate the positioning of flexible coils. Verduijn et al. showed that the resulting images can be successfully used for target volume delineation [24].

In order to obtain the optimal MRI protocol for a specific RT application, it is important to pay attention to the following image properties:

- *In-plane resolution (voxel size), slice thickness and slice gap*

  In-plane resolution, slice thickness and the slice gap should be appropriate for the application, enabling detailed visualisation of tumour and organs at risk, and the registration of the MRI with the CT-sim (involving resampling of the MRI volume in CT coordinates). When a large field of view in feet-head direction is required to include the entire treatment region, multi-station acquisition is preferred to reduce displacements due to gradient non-linearity.

  A stereotactic treatment will require smaller, preferably isotropic (same size in all three directions), voxels than other treatments. Isotropic images can be acquired using 3D acquisition applying two phase encoding directions. In multi-slice imaging, it is recommended to keep the gap between slices small, for instance 10% of the slice thickness, because a larger gap increases chances of missing small pathological features, but without any gap there could be crosstalk between slices in 2D acquisitions. An option to scan without a gap is to acquire 2D slices in an interleaved way. However, crosstalk may still occur between slices, resulting in contrast differences between the slices. Therefore, if it is essential to scan without gap, a 3D acquisition is advised (see next paragraph). Note that the user of image viewing and treatment planning software typically observes the voxel size as reported in the DICOM (Digital Imaging and Communications in Medicine, a data interchange protocol) header of MR images (in-plane pixel spacing, slice thickness). This voxel size is often not equal to the acquisition spatial resolution, because interpolation is widely used in MRI scan protocols. The acquisition resolution is visible on the MRI scanner, or may be obtained from other DICOM header information such as FoV and acquisition matrix size.

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- *2D versus 3D acquisition.*

  The use of a 3D sequence may improve the low SNR associated with acquisition of 2D thin slices while at the same time covering a sufficiently large FoV. Furthermore, acquiring a single 3D acquisition may reduce scan time compared to the acquisition of three separate 2D acquisitions in transverse, sagittal and coronal orientations. However, the contrast may be different in a 3D acquisition compared to a 2D acquisition. Additionally, behaviour in relation to motion will be different for different sequences. In general, a longer scan time will lead to more motion during the acquisition.

- *Image orientation and angulation*

  In radiotherapy, the CT-sim and CBCT acquisition protocols are always with orthogonal axes and transverse slices. Therefore, it is recommended for accurate delineation and registration to use similar orientations for MRI for radiotherapy purposes. Additionally, the requirements of the treatment planning software need to be considered (historically, some treatment planning systems require transverse slices). Dependent on the application and workflow, it is necessary to exercise caution with oblique volumes.

- *Geometric accuracy*

  It is advisable to position the area of interest in the isocentre of the MRI scanner, because that is where the magnetic field is the most homogeneous and the geometric distortion is smallest. The receiver bandwidth is recommended to be chosen such that the water-fat shift, which depends on the B_0 field strength, is smaller than 1 mm. The WFS (water-fat shift) in mm can be calculated by:

$$\text{WFS [mm]} = f_0 * \sigma * \text{resolution [mm/pixel]} / \text{BW [Hz/pixel]}$$

  This formula uses the Larmor frequency $f_0 = \gamma * B_0$, with $\gamma = 42.85$ MHz/T for protons, and σ, the chemical shift between fat and water approximately 3.5 ppm. For the resolution, the acquisition spatial resolution in the frequency encoding direction should be used. For example, at 3 T an acquisition readout bandwidth of 225 Hz/pixel with an acquisition resolution of 1 mm results in a WFS of approximately 2 mm.

  Some readout protocols are more sensitive to induce geometric inaccuracies than others. For example, single-shot EPI is widely used in functional and diffusion-weighted MRI, but suffers

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from significant geometric distortions in the phase-encoding direction due to the very low effective readout bandwidth in the phase-encode direction. Therefore, it is recommended to not use EPI sequences to obtain images for delineation.

In order to reduce distortion due to field inhomogeneities, extra attention could be given to shimming, for instance by application of higher order shimming (if available on the MRI scanner). Extreme caution needs to be exercised when applying higher order shimming, because moving gas in the rectum within the volume used for shimming could cause poor shimming results and in fact deteriorate image quality instead of improve it.

In addition, the use of a 3D acquisition technique is generally preferred over 2D acquisitions, since options for 3D distortion correction of gradient non-linearities may be limited or might not perform well for 2D acquisitions. It is highly recommended to enable the scanner's distortion correction options suitable for the application.

Even when taking all these precautions, it should be noted that there will still be residual distortions and deformations, due to both technical (inhomogeneities) and physiological causes, such as bladder filling.

Check for each patient that the MRI acquisition was done according to protocol

For diagnostic purposes, MRI technicians are used to adapt sequence parameters, FoV settings, slice angulation, etc. to obtain a visually good scan quality for a specific patient. However, for radiotherapeutic applications it is important to not change any parameters that can influence geometric accuracy or are important for accurate image registration and delineation. To guarantee that all MR images which are used for delineation in the RT workflow adhere to the described recommendations for the additional QC measurements and MRI protocols, it is necessary to check the parameters of every scan. This can be done by checking the DICOM tags of all MRI scans that will be used for delineation. One option is to include a manual check of the DICOM tags in the RT workflow. Another option is to develop an automatic check, e.g. by writing and running a script in the contouring software. A selection of DICOM tags can be evaluated to check that a scanner approved for RT simulation is used (on which the additional QC measurements have been performed) and that the correct sequences and parameters are used, e.g. spatial resolution, slice thickness, interslice gap, readout bandwidth and distortion correction.

Take remaining registration inaccuracies into account in the PTV margin

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Even after mitigation using the previous recommendations, inaccuracies related to MR image formation and image registration may still exist. These inaccuracies should be estimated and taken into account in the PTV margin. Important factors are: the geometric accuracy (expected distortion) of the MRI sequence used for delineation, the expected delineation accuracy (dependent on contrast and resolution), and the expected accuracy of image registration. Since different MRI series in a study are made sequentially, registration issues due to anatomy changes or patient movement can occur between different acquisitions of the same imaging session. It is necessary to verify that the MRI series are inherently registered, otherwise they need to be registered manually.

3.3 MRI-only treatment planning for external beam radiotherapy

In an MR-only workflow MRI scans need to be acquired for both delineation and, after conversion to synthetic CT, for treatment planning. Several methods have been proposed to convert MRI scans to sCT scans. In general, the main challenge is to distinguish air and bone in the MR images. Some methods include dedicated MRI sequences to discriminate air and bone such as ultrashort echo time sequence (UTE) followed by post-processing and conversion to Hounsfield units (HU). Other methods converted conventional MRI sequences (mostly T1-weighted MRI or T1 with Dixon fat separation) directly to HUs using either, manual bulk density assignments, atlas based or voxel-wise HU assignments e.g. using deep learning strategies [25]. The required accuracy of the HU in the sCT scan for accurate dose calculations is dependent on the treatment modality. For instance, proton therapy requires a higher accuracy than photon therapy [26].

To enable an MR-only workflow, also at least one MR image series in the protocol is required to include the body contour and thus comprise a large FoV and scan volume. A coil bridge can be used to avoid deformation of the body contour due to coils lying directly on the patient's skin.

3.3.1 Challenges

External laser systems for patient positioning are available for installation at RT-dedicated MRI scanners. This makes it possible to mark the skin of the patient with lines in order to easily reposition the patient on the treatment couch. Marking a reference point in the scan by placing three lipid beads on the skin does not allow for precise repositioning of the patient, since geometric accuracy is not guaranteed at the skin, due to 1) distortions further from MR isocentre, and 2) susceptibility effects at the tissue-air interface. It is therefore important to implement a suitable IGRT protocol at the treatment machine, mitigating the inaccuracies of skin markers with an MR-only workflow. An MR-

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only workflow thus dictates an online IGRT protocol, at least in the first few fractions, in order to correct the set-up error introduced by the inaccurate reference point definition.

The patient couch alignment is usually not an issue for MRI scanning for RT when the scan is registered to the CT scan. However, in an MR-only workflow, the flatness of the couch is more of an issue since any rotation can result in a set-up error if the treatment couch is able to correct translations only. In an online adaptive workflow on an MR-linac or a CBCT, rotation can be corrected during treatment planning. During scanning, the couch is moved between scans if the centre of FoV differs too much. Errors in couch movement can result in displacement between scans.

3.3.2 Recommendations

Automatic generation of sCTs is a recent development that currently has been released by a number of industrial vendors of MRI and radiotherapy image processing software. The sCT scan is additional to the other MRI sequences in the MR-RT protocol, and is designed for dose calculation. In addition, the sCT can be used for position verification and delineation of organs at risk, provided that the sCT is checked carefully with the source MR images. As with an MR-CT fused workflow, since the MRI scans are made sequentially, registration issues due to anatomy changes or patient movement can be present. It is recommended to verify that no motion is present between the MRI scans; otherwise, they need to be registered.

Furthermore, if the sCT is used as reference scan for position verification, it is advised to thoroughly verify whether the accuracy of registration of the CBCT to sCT (or CBCT to MRI) is sufficient as the grey values of the sCT may result in different registration results. In case that the IGRT protocols rely on the visibility of fiducial markers (e.g. gold markers used for position verification for prostate cancer treatment), it is important to account for the poor visibility of the markers in the sCT scans [27–30]. In addition, the gold fiducials cause field inhomogeneities resulting in local artifacts in MR images [31,32]. The fiducial marker appearance depends on the pulse sequence used, the imaging parameters and the shape and orientation with respect to the direction of the main magnetic field [33].

3.4 Online MR-guided radiotherapy

The use of hybrid MRL systems to perform online MR-guided external beam radiotherapy (EBRT) is increasing rapidly. According to a survey in 2019, 60% of the Dutch radiotherapy departments already has installed or is planning to install an MRL within the next five years [1]. Two commercial systems

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are currently available: the 0.35 T MRIdian by ViewRay (ViewRay Inc, Oakwood Village, OH, USA) and the 1.5 T Unity by Elekta (Elekta AB, Stockholm, Sweden). Both systems offer the ability to adapt to inter-fraction anatomy changes by daily adaptation of the treatment plan and mitigate intra-fraction motion by means of gated treatment delivery.

On both systems a 3D volumetric MRI scan is acquired directly after patient positioning, which serves as the daily MRI scan on which the treatment plan is based. In order to speed up the online workflow a previously acquired reference image (either CT-sim or MRI-sim) is registered non-rigidly to the daily MRI to propagate the contours. Contour propagation based on MRI-sim has been shown to be more accurate than based on CT-sim (at least for prostate treatments) [34]. After contour propagation, the reference (pre-treatment) plan is evaluated to check if clinical goals for target coverage and OAR sparing are still within tolerance. If the requirements are not met, a new plan can be generated online. During irradiation, a fast 2D cine MRI sequence can be acquired to monitor tumour motion and perform gated treatment delivery.

3.4.1 Challenges

Although both ViewRay and Elekta only offer a select set of fixed imaging protocols to be run during clinical workflow, there are several potential pitfalls one has to be aware of.

First, since the treatment plan is recalculated (and possibly adapted) based on the daily MRI acquisition, the same requirements with regards to FoV, geometric accuracy, and electron density accuracy, apply as described in the MRI-only section (Section 3.3) above. Second, due to the adaptive nature of MR-guided RT, the average duration of a single treatment fraction on the MRL is considerably longer than on a conventional (CBCT-based) system. Internal organ motion, such as peristalsis and bladder filling, is therefore much more pronounced. For these reasons, as well as patient comfort, all processes – including image acquisition – need to be performed as fast as possible. Finally, when performing gated treatment delivery, it is important that the image quality (resolution and CNR) of the 2D motion monitoring is high enough for the tracking algorithm to robustly track the tumour (or OAR in case of avoidance gating).

3.4.2 Recommendations

In terms of patient positioning, it is advisable to reproduce the set-up as used during the preparatory phase (CT-sim and/or MR-sim) as much as possible, as this will aid the registration process. This also increases the chance that the reference plan is valid at the time of treatment, which in turn shortens

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the treatment time and thus minimises intra-fraction motion (bladder and bowel filling) and increases patient comfort. It is therefore recommended that the patient positioning devices are matched between the CT-sim, MR-sim and MRL. Lasers are not mandatory (and not part of the standard equipment on the Elekta Unity), but will help in removing any rotational discrepancy between the reference images and the daily set-up. Imaging time is more restricted for reasons described above, but no data is yet available on the optimal trade-off between image quality (resolution and CNR) and speed. Regardless of the imaging time, however, the daily volumetric MRI should have high geometric fidelity. It is advised to check the readout bandwidth of each imaging protocol to ensure that the water-fat shift does not exceed 1 mm.

When using 2D single-slice acquisitions for real-time tumour tracking or gated delivery, the required frame rate is dependent on the type of motion and the PTV margins used. A higher frame rate, however, limits the achievable resolution and achievable SNR. It is therefore advisable to perform simulation scans on the MRL in order to optimise the 2D cine protocol and test the robustness of the tracking algorithm, particularly for small lesions or lesions with poor CNR. In addition, the geometric accuracy should be determined and preferably visualised during slice positioning, as the geometric fidelity varies depending on slice location (i.e. distance to isocentre). 2D single-slice images are not corrected for gradient non-linearity through-plane, and therefore demonstrate larger geometric distortions than volumetric acquisitions.

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4 Recommendations for system QC

This chapter describes the system QC tests and phantom tests recommended for the use of MRI in radiation therapy. The recommendations are presented separately for the different applications of MRI in RT:

1. MRI not registered to CT
2. MRI co-registered to CT
3. MR-only treatment planning for external beam radiotherapy
4. Online MR-guided radiotherapy

The chapter starts with an overview of all QC tests presented in Table 1. This guideline builds on the recommendations for general MRI-QC, as published by the Society for Medical Physics of the Netherlands (NVKF) [5]. The NVKF recommendations were developed for diagnostic imaging and describe a variety of tests to ensure adequate imaging performance and geometric accuracy near the isocentre of the scanner. The use of MR images without registering them to CT images, to aid the delineation done on the CT scan (Table 1, application 1) needs to meet at least the recommendations set by the NVKF. More elaborate testing is required depending on the RT-specific application. The imaging requirements in terms of geometric fidelity are higher when the MR image is co-registered to the CT image and used for the delineation (Table 1, application 2), or when the MRI is used as a sole reference image in the TPS, the so-called MR-only treatment planning (Table 1, application 3). Finally, hybrid MR-linac systems require additional tests to characterise the potential interplay between the MRI and linac subsystems for online MR-guided radiotherapy (Table 1, application 4). Each QC test as listed in Table 1 is described in more detail in Section 4.2, together with the suggested frequency of testing and the tolerance/action limits. Detailed instructions on how to perform the tests can be found in Appendix A, Section A.2. Practical aspects of the implementation of MRI QC are discussed in Section 4.3. The chapter concludes by listing a number of unmet needs in Section 4.4.

4.1 MRI QC measurements overview

All QC tests are summarised in Table 1. For each test, a short explanation and the method of measurement are given, as well as the applications for which the test is recommended. Tests that are based on the recommendations by the NVKF [5], and thus apply to all four applications, are listed first. The explanation of these tests in Appendix A are translations from the Dutch text in the NVKF guidelines. Additional QC tests that apply to applications 2 to 4, *MRI co-registered to C*T, *MR-only*

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treatment planning, and *online MR-guided radiotherap*y are listed next in grey, blue, and green, respectively.

Table 1: Overview of QC tests described in this document. Top section describes QC tests recommended by NVKF (minimum requirement for all applications) [5]. Additional tests, specific to each application are listed below and coloured per application. White: application 1 - MRI not registered to CT; grey: application 2 - MRI co-registered to CT; blue: application 3 - MR-only treatment planning; green: application 4: online MR-guided radiotherapy.

	Parameter	**Explanation**	**Method**	**Application**
1	SNR	Sensitive parameter, affected by many system parameters	1. Repeated measurements [5,35] 2. Separate noise measurement [5]	1,2,3,4
2	Image uniformity	Depends on homogeneity of RF transmit and receive coils. May also be dependent on B_0 homogeneity	Percent signal change in flood field phantom [5,36]	1,2,3,4
3	Ghosting	Relates to system stability for both RF and gradients	Amount of background ghosting relative to object signal [5,36]	1,2,3,4
4	Image artefacts	A-specific indication for a variety of problems	Visual inspection	1,2,3,4
5	Gradient-related geometric distortion	Check for amplitude calibration of magnetic field gradients and residual errors due to gradient non-linearity	3D acquisition of head-sized or body-sized phantom, depending on the clinically relevant volume	1,2,3,4
6	Resonance frequency	Indicates changes in B_0 field strength	Interactive f_0 measurement. Logging of prep scan result [5]	1,2,3,4
7	RF transmit amplitude	Indicates changes in transmit signal path and RF power amplifier	Logging from DICOM header or logfile [5]	1,2,3,4
8	Shim (B_0 homogeneity)	Homogeneity of B_0 field	GRE based B_0 field map or interference image (SE/TSE) of head-sized or body-sized phantom, depending on application	1,2,3,4
9	Couch positioning	Relative couch positioning accuracy between scans for multi-station image acquistion	Ruler based method	-,2,3,4 If FoV comprises > 1 couch position
10	Connectivity and orientation	To test connectivity, SW compatibility, orientation	Visual inspection on asymmetric phantom	-,2,3,4
11	External laser position	Determines accuracy of patient positioning in MR-only workflow	Marker phantom aligned to external lasers	-,-,3,4 If installed

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	Parameter	Explanation	Method	Application
12	Synthetic CT generation for dose calculation	Assessment of geometric and HU accuracy	Image similarity and dosimetric evaluation	-,-,3,- If used
13	Testing the use of the synthetic CT as reference for position verification	To test workflow, including patient positioning	Evaluation of image registration performance	-,-,3,-
14	MR-MV coincidence	Determine systematic offset between MRI and MV isocentre	Phantom containing MRI and EPID visible landmarks or dosimetric film	-,-,-,4
15	B_0 direction	B_0 direction determines direction of dose kernel tilt	Loop wire method	-,-,-,4
16	Gantry-angle dependent B_0 homogeneity & MR isocentre shift	Poorly shimmed gantry may introduce angle dependent B_0 field inhomogeneities	GRE based B_0 field maps at various gantry angles	-,-,-,4
17	Linac-induced RF interference	RF producing components (e.g. magnetron) may introduce spurious noise	Noise only acquisition	-,-,-,4
18	Radiation-induced artefacts	Radiation induced currents in receiver coils may introduce image artefacts (spiking)	Beam-on protocol	-,-,-,4
19	Temporal stability test	Gradient heating may introduce artefacts and/or geometric drifts	Visual inspection and image registration of time-series images	-,-,-,4
20	Real-time feedback latency	To test the total latency of the beam gating functionality	4D phantom with trackable object	-,-,-,4

4.2 MRI QC measurements descriptions

For each test presented in Table 1, a short motivation and explanation is provided in this section as well as the recommended action limits. Two limits are described: the 'Acceptable' action limit, which, if exceeded, is meant as a trigger to investigate the system, and a 'Critical' action limit, which indicates that corrective maintenance is required. Where possible, the action levels are predefined and fixed for all systems. Some tests, however, are defined as constancy tests. The action limits for these tests refer to the maximum permissible deviation (%) relative to the average value obtained during the initial reference period (e.g. the first 10 measurements). An example of a constancy test is the SNR, which is highly dependent on the combination of field strength, receiver coil, and phantom setup, and thus unique to each system.

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It is advised to perform all the tests during commissioning of the system. After commissioning, it is advised to implement a reference period, during which the tests are performed at a high testing frequency to establish the reference values for the constancy tests. Once these values are established the testing frequency can be reduced. The minimum interval per test as specified below refers to the minimum interval *after* the initial reference period.

For a number of tests, the minimum interval is set to the frequency of periodic maintenance. For some tests, a fixed time-based periodicity (monthly, quarterly, or annually) is stated. It is important to realise, however, that these tests will also have to be conducted when (corrective) maintenance or system updates are carried out that may affect the parameter of interest. It is therefore important to always be aware of the type of maintenance that is carried out by the vendor and to be extra cautious when software or hardware updates or upgrades have been carried out.

4.2.1 Signal-to-noise ratio

Description

The signal-to-noise ratio (SNR) is a generic parameter determined by several system components. In MR imaging it is a sensitive parameter; deviations of the order of 5% can be measured systematically. The absolute value of the SNR is highly dependent on a variety of scan parameters, the phantom used and its set-up, the receive coil (position), and the specific MR system used. As SNR is an overall measurement, it reflects the RF transmission and receive chain, and interference of other RF sources. Besides phantom and coil set-up and hardware performance, also image reconstruction and intensity correction filters can affect the SNR. Generally, a flood field phantom section is used to measure the system SNR (referred to as “SNR combined image”).

As the RF receive chain is most vulnerable, SNR measurements are a crucial test for coil performance. As the sensitivity for a single coil deterioration in an image acquired by a multicoil array is limited, analysis of the single coil element is recommended as well. For this test, a multicoil array measurement is performed, but the individual coil images are reconstructed and analysed (“SNR uncombined image”).

Limits

Criteria (constancy)	**Acceptable**[1]	**Critical**[1]	**Applications**	**Minimum interval**
SNR combined imaged [a.u.]	10%	20% Vendor specs	1,2,3,- -,-,-,4	Maintenance Monthly

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SNR uncombined image [a.u.]	15%	30% Vendor specs	1,2,3,- -,-,-,4	Maintenance Monthly

[1]Action limits for SNR refer to the maximum permissible deviation (%) relative to the average value obtained during a reference period (e.g. 10 measurements). For MR-linac applications only the vendor specifications are currently listed as a critical action level as long-term QA data is currently still scarce. Tolerance levels may be set in the future once these data become available.

4.2.2 Image uniformity

Description

Image uniformity is a measure of the MRI system's ability to produce a constant signal over the entire scanned volume of a homogeneous object. It depends on the homogeneity of RF transmit and receive coils and may be affected by B_0 homogeneity. The image uniformity is dependent on the coil, the phantom and the position relative to each other. All modern scanners, equipped with multi-channel receive arrays employ image filters to improve image uniformity. The uniformity in MRI scanners with a higher magnetic field strength is generally lower than in MRI scanners with a lower magnetic field strength. The test determines baseline values that are then re-tested in a constancy test.

Limits

Criteria (constancy)	**Acceptable**[2]	**Critical**[2]	**Applications**	**Minimum interval**
Uniformity [%]	3%	6%	1,2,3,4	Maintenance

[2]The degree of uniformity is a design parameter of the RF coil, and can vary widely for different RF coils. The typical uniformity for a head coil is above 80% [2]. Upon acceptance, it is recommended to compare with the specification as stated at the time of purchase or by the manufacturer. For the current recommendations, the action limit is set on the constancy of the uniformity measurement. Rationale: to a certain extent the limits are arbitrary. In principle, uniformity may of course improve. However, in the case of major changes, it may be relevant to find the cause as it may be a result of SNR loss in high signal areas. These constancy values have proved to be well achievable in practice with a number of MRI systems.

4.2.3 Ghosting

Description

Ghosting refers to repeated (parts of) the object at image locations from which it does not originate. These can be very clear copies of the original image, but also signal with much less structure. Ghosting can be found in the phase and frequency encoding direction, originating from system (RF and

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gradients) and frequency gradient instability, respectively. Ghosting is expressed as a percentage of the magnitude signal of the measured object. This is determined by measuring the signal of the ghosts in the image background and relating this to the mean signal of a homogeneous phantom (see Appendix A.1 for detailed information).

Limits

Criteria	Acceptable[3]	Critical[3]	Applications	Minimum interval
Ghosting frequency encoding direction [%]	1	3	1,2,3,4	Maintenance
Ghosting phase encoding direction [%]	1	3	1,2,3,4	Maintenance

[3]The action limits listed here are the action limits as specified by the NVKF [5]. Other guidelines use comparable action limits: ACR gives an action limit of 2.5% for ghosting [4]. AAPM [2] specifies a specific limit for quadrature ghosting of 2%. A 1% change is therefore a reasonable trigger to investigate the system ("acceptable" action limit). Some MR manufacturers apply a stricter "critical" action limit of 1%.

4.2.4 Image artefacts

Description

Artefacts may be caused by a variety of problems, including relatively common issues such as defects in RF receiver coils, cables and connectors. Therefore, this is an important test to monitor changes in image quality.

Limits

Criteria	Acceptable[4]	Critical[4]	Applications	Minimum interval
Image artefacts	No artefacts present	Artefacts present	1,2,3,4	Maintenance

[4]For various reasons, other artefacts than ghosting may arise in MRI images. The image should be free of artefacts, except for generic artefacts that are intrinsic to MRI. Other artefacts (to be assessed visually) require further investigation to what extent they disrupt the diagnostic quality of an MRI examination. No reference value is specified because the relevance of the artefact is determined by the nature of the artefact more than by the intensity.

4.2.5 Gradient-related geometric distortion

Description

Geometric errors occur when the actual gradient fields that are produced by the scanner deviate from the intended gradient fields. Two main sources are responsible for these potential deviations: 1) errors in the calibration of the gradient amplitude, resulting in a global scaling error (i.e. linear component),

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and 2) errors due to gradient non-linearities, resulting in local distortions that typically increase with distance from the isocentre.

The test described in this section can either be performed on a small head-sized phantom (20-25 cm diameter), or a large FoV body-sized phantom (40-50 cm diameter) depending on the clinically relevant FoV. The clinically relevant FoV is defined as the volume in which the delineations are performed on MRI and/or the volume that is used for image registration and dose calculations (application 3 and 4).

- *Small FoV gradient fidelity (non-SRS)*

  Head-sized phantoms are commonly used in diagnostic QC guidelines. The measured diameter of the phantom in the x, y and z directions gives an indication of the geometrical accuracy of the system. For non-stereotactic treatments with a small relevant FoV, a check on just the linear components (i.e. a check on the dimensions of a rigid phantom, such as the ACR phantom), is sufficient as displacements due to gradient non-linearities are general small (< 1 mm) close to isocentre.
- *Small FoV gradient fidelity (SRS)*

  For stereotactic treatments (e.g. brain SRS) it is advised to check the linearity and quantify the displacements due to linearity errors within the phantom volume (e.g. by using the Magphan EMR128 phantom, which is also known as the ADNI phantom).
- *Large FoV gradient fidelity (non-SBR and SBRT)*

  This test characterises the MRI gradient fidelity on a large, body-sized, geometric fidelity phantom, which contains MRI visible markers at known marker locations (e.g. the Quasar MRID3D phantom or the CIRS Model 604-GS). The displacements are calculated by subtracting the marker locations found in the MRI dataset from the known marker locations. Such phantom quantifies the net error of both linear (i.e. gradient amplitude calibration) and non-linear gradient distortions. In order to minimise B_0 inhomogeneity effects it is recommended to either use a large readout bandwidth or, preferably, use the reverse gradient method. In the latter approach the measurement is performed twice with gradient polarities reversed [37,38], which allows an independent quantification of both the gradient-induced distortions and B_0-induced distortions. It is necessary to perform the test with the vendor 3D distortion correction turned on, as the residual errors, after correction, are of interest in this test. The action levels in the table below are given as a percentage of the phantom size, or in millimeters (mm). For large FoV measurements the 99th percentile (p99) and the 95th percentile (p95) of

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the displacement within a volume of a certain size are specified. The size of the volume is indicated by the diameter spherical volume (DSV). Note that, when a small DSV is included in the analyses, no separate gradient fidelity test on the small sized phantom is required.

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Small FoV phantom[5]: deviation from actual size	1% 1%	3% 1.5%	1,-,-,- -,2,-,-	Maintenance Maintenance
Small FoV phantom[5]: local geometric displacement	1 mm within relevant FoV	1.5 mm within relevant FoV 1 mm for brain stereotactic RT	-,2,3,4	Maintenance
Large FoV phantom[6]: local geometric displacement	p99 1.0 mm within DSV 200 mm p99 2.0 mm within DSV 350 mm p99 2.5 mm within DSV 400 mm	p95 1.0 mm within DSV 200 mm p95 2.0 mm within DSV 350 mm p95 2.5 mm within DSV 400 mm	-,-,3,4	Maintenance[7]

[5]These action limits assume a phantom the size of a human head (20-25 cm).

[6]For non-stereotactic RT applications, the local geometric displacements need to be < 2 mm within the relevant clinical volume, or < 1 mm for intracranial SRS. If the displacement within a 400 mm DSV cannot be accurately determined it is recommended to document the volume where displacements are > 1 mm and > 2 mm during acceptance & commissioning as suggested by the IPEM [39].

[7] A large FoV check should be performed at least after relevant maintenance and hard- or software updates. It is further recommended to perform a large FoV gradient fidelity test quarterly to monitor the stability of the gradient performance. Gradient performance does not tend to drift over time, but changes to the gradient calibration or gradient non-linearity (GNL) correction tables may have considerable effect on the geometric fidelity of the system.

4.2.6 Resonance frequency

Description

The resonance frequency of the signal is determined by the strength of the B_0 field; a drift in resonance frequency over time is therefore an indication of a drift in the B_0 field. The test determines baseline values that are then re-tested in a constancy test.

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Limits

Criteria (constancy)	Acceptable[8]	Critical[8]	Applications	Minimum interval
Resonance frequency [MHz]	0.04% per month	0.1% per month	1,2,3,4	Maintenance

[8]The action limits are relative compared to an average value of a reference period. The action limits are derived from MR system specifications from manufacturers. Large deviations in the resonance frequency result in suboptimal coil function. Strong variability in resonance frequency may indicate problems in the RF transmit subsystem.

4.2.7 RF transmit amplitude

Description

The RF transmit amplitude is a measure for the stability of the RF transmit chain of the MRI scanner. Usually, the reference amplitude is defined as the RF amplitude required to create a certain flip angle (e.g. 90° or 180°) with a rectangular pulse with a fixed duration (e.g. 1 ms). The test determines baseline values that are then re-tested in a constancy test. Some vendors will perform active monitoring on this parameter. If that is the case, manual logging may be omitted.

Limits

Criteria (constancy)	Acceptable[9]	Critical[9]	Applications	Minimum interval
RF transmit amplitude	15%	30%	1,2,3,4	Maintenance

[9]There is no strict requirement for the action limit, but this value is based on experience with a number of MR systems.

4.2.8 Shim (B_0 homogeneity)

Description

Shimming the system is to optimise the homogeneity of the B_0 field, i.e. the uniformity of the B_0 field over the scanned volume. The homogeneity is usually expressed in parts per million (ppm) of the magnetic field or in Hz. The B_0 homogeneity is best in a small volume around the isocentre and becomes worse as the volume increases. Similar to the gradient-related geometric distortion test, the action limits are given for a range of volumes that vary in size, indicated by the diameter spherical volume (DSV). The B_0 homogeneity can be measured without application of shimming gradients or with linear or second order shims.

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A drop in B_0 homogeneity may be caused by small ferromagnetic objects in the magnet (hairpins, coins, etc.) or by large ferromagnetic materials in the vicinity of the magnet. Ideally, the measurements are performed with the default shim settings (i.e. automatic shim calibration on a per scan basis turned off) to be more sensitive to these effects. However, this is not possible on every scanner.

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Shim Peak-to-Peak[10] in phantom with head size (20-25 cm)	1.5 ppm	2 ppm	1,2,-,-	Maintenance
Shim Peak-to-Peak[10] In body size phantom (≥ 40 cm) DSV 300 DSV 350	 1.0 ppm 1.5 ppm	 1.5 ppm 2.0 ppm	-,-,3,-	Quarterly
Shim Peak-to-Peak[11] In body size phantom (≥ 40 cm) DSV 300 DSV 350	 64 Hz 96 Hz	 96 Hz 128 Hz	-,-,-,4	Quarterly

[10]The deviations are specified in parts per million (ppm) relative to the B_0 field. Rationale: chemical shift between water and fat is 3.5 ppm, and disruptions of the field by the patient are also in the order of 3-4 ppm. In addition, a 2 ppm disturbance of B_0 is significant. In many clinical protocols, the linear term of the shim is still optimised in vivo. The AAPM specifies a B_0 homogeneity of 10 ppm within a 30-40 cm diameter sphere [2], although most modern systems achieve a much higher homogeneity.

[11]For online MR-guided applications the absolute geometric accuracy is of importance, which is directly linked to the readout bandwidth [in Hz/pixel]. Therefore, the tolerances are also stated in Hz. For 1.5 T the tolerances in Hz equal the tolerances stated in ppm. At 0.345 T the tolerances correspond to 4.35 ppm, 6.5 ppm, and 8.7 ppm.

4.2.9 Couch positioning

Description

Absolute couch position is of interest for certain use cases of MR-only treatment planning. However, when online position verification is used, absolute couch position errors during the treatment planning phase are mitigated. Couch rotations are more often of concern as they often cannot be corrected for on the conventional linac when a MR-only workflow is used.

Another application that needs accurate couch positioning is multi-station MRI acquisition, which is used in case of long FoV in craniocaudal direction to avoid large geometric distortions due to gradient

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non-linearities. Accurate stitching of the images acquired in the different stations is dependent on the accurate determination of the couch position before and after couch movement.

For online MR-guided RT, the consequence of the couch inaccuracy depends on the method of correction of the patient position. For systems that use online couch movement, it is essential that the couch translations are as accurate as on a conventional linac. For online MRI-guided treatment adaptation without couch translation, using a virtual couch shift or full replanning, the couch positioning accuracy is of less relevance than for systems that use online couch movement.

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Couch angulation	0.15 deg (1 mm per 400 mm)	0.3 deg between couch and MR	-, -, 3,-	Maintenance
Couch position	< 0.5 mm discrepancy in vertical offset	1 mm discrepancy in vertical offset	-, -,3,- -,-,-,4	Quarterly Monthly
Couch position	< 0.5 mm discrepancy in lateral offset	1 mm discrepancy in lateral offset	-, -, -,4	Monthly

4.2.10 Connectivity and orientation

Description

As MRI is used for treatment planning and position verification, the consistency between the scan parameters at the MRI scanner and in the treatment planning system (TPS) need to be verified. Important parameters include pixel spacing, slice thickness and scan orientation. These can be checked by scanning an asymmetric phantom with different pixel spacing in the left-right and anterior-posterior directions and with various orientation scan settings: head-first supine (HFS), feet-first supine (FFS), head-first prone (HFP) and feet-first prone (FFP).

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Scan orientation	NA	Correct in DICOM header and properly visualised in TPS	-,2,3,4	Software updates and relevant hardware upgrades

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Pixel spacing	NA	Correct in DICOM header and properly visualised in TPS	-,2,3,4	Software updates and relevant hardware upgrades

4.2.11 External laser position

Description

Similar to the absolute couch position, the required accuracy of the external laser position depends on the use of the MR images in combination with the treatment delivery workflow. The orthogonality, vertical laser position and laser-MR isocentre coincidence should be tested regularly as the lasers are susceptible to mechanical interactions (e.g. collisions), which may affect their alignment. The action limits are based on tolerance levels as described for CT simulators in NCS report 11 [40].

For online MR-guided treatment using daily adaptation of the target and organ at risk, the positioning in the pre-treatment phase is not crucial. Not all MR-linac systems have in-room lasers (e.g. Elekta Unity). However, when the laser system is used to calibrate and QC the MR-MV isocentre coincidence (e.g. ViewRay MRIdian), the action limits as described in Section 4.2.14 (MR-MV coincidence) apply.

Limits

Criteria	**Acceptable**	**Critical**	**Applications**	**Minimum interval**
Orthogonality	< 0.3 deg	0.5 deg	-,-,3,4[12]	Monthly
Vertical position	< 1 mm	1 mm	-,-,3,4[12]	Monthly
Consistency isocentre of laser and MRI in x, y and z	1 mm	2 mm	-,-,3,4[12]	Monthly

[12]If installed. When used to calibrate and QC the MR-MV isocentre coincidence stricter action limits apply (see 4.2.14)

4.2.12 Synthetic CT generation for dose calculation

Description

For synthetic CT (sCT) generation, the geometry has to be accurate, as is described in Sections 4.2.5 and 4.2.8. In addition, the conversion of MRI arbitrary units to Hounsfield units has to be correct. At this moment, there is no phantom on the market to test this. Moreover, the appropriate method for QC of the sCT scan depends on the method that is used to convert MRI scans into a sCT scan. Different methods that have been proposed include automated contouring of high- and low-density structures followed by bulk density overrides, zero TE sequences that distinguish bone and air, followed by

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conventional or deep-learning-based post-processing and deep-learning-based conversion from conventional MRI scans (e.g. T1 or T2 weighted) to sCTs. Specifying the best method per application is outside the scope of this report and is a topic of ongoing research. Here we present some considerations, without aiming to describe complete QA of sCT scans.

First an image similarity comparison of patient CTs with the sCTs is recommended. In addition, the dose on the sCT can be compared to the dose on the pretreatment CT (CT-sim) or the CBCT. For the image similarity comparison, the mean absolute error (MAE) and the structural similarity index measure (SSIM) between the CT-sim and the sCT can be calculated to assess the image similarity and, if possible, a comparison of electron densities in the treatment planning system. For the dose, the relative dose difference and the gamma analysis can be used to compare the dose calculated on the CT-sim and the sCT. However, one needs to take into account that the CT-sim and the MR images for the sCT are not acquired simultaneously. The criteria for acceptance will therefore also depend on changes in the anatomy (e.g., air in bowel) and the quality of the image registration between CT-sim and sCT. Further, the dose threshold (D10, D90, D100), the target volume or OAR assessed also play a role in the acceptance of the dose difference and gamma pass rate. Once the system for sCT generation is accepted, the following limits could be used to evaluate the system after software updates and maintenance or changes in imaging acquisition that may affect the conversion from MR sequence to sCT.

Limits

As sCT generation has relatively recently been introduced in the clinical practice of radiotherapy, the QA procedures and their target values are still under development and the critical limits have not been established yet. Therefore, the recommendations are loosely defined and might change in future reports when more experience has been gained.

Criteria for mean (in group of 10 patients)	**Target value**	**Critical**	**Applications**	**Minimum interval**
Image similarity - $MAE = \sum_1^n \frac{\lvert CT_i - sCT_i \rvert}{n}$ in the intersection of the body contours	55 HU	NA	-, -, 3, -	Maintenance and software updates
- $SSIM = \frac{(2\mu_{CT}\mu_{sCT}+c_1)(\sigma_{CTsCT}+c_2)}{(\mu_{CT}^2+\mu_{sCT}^2+c_1)+(\sigma_{CT}^2+\sigma_{sCT}^2+c_2)}$ #	0.90	NA	-, -, 3, -	
- Dose (D > 90%)	+/-0.5%	NA	-, -, 3, -	
- Dose difference $(D_{CT}-D_{sCT})/D_{CT}$ *100%	1%	NA	-, -, 3, -	

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[#]μ_{CT} en μ_{sCT}: mean pixel value of CT and sCT resp., σ_{CT}^2 en σ_{sCT}^2: variance of pixel value of CT and sCT resp., σ_{CT} is covariance of CT and sCT, $c_1 = (k_1L)^2$, $c_2 = (k_2L)^2$ two variables to stabilise the division with weak denominator; L the dynamic range of the pixel-values ($2^{\#bits/pixel}$-1) and the default values for $k_1 = 0.01$ and $k_2 = 0.03$ [41]

4.2.13 Testing the use of the synthetic CT as reference for position verification

Description

When no CT-sim is used in the treatment chain, the sCT (or MRI directly) needs to be suitable for position verification. When introducing MR-only for a tumour site or specific position verification protocol (e.g. bony anatomy match, MRI to CBCT match), it is advised to add for the first group of patients the MR-only workflow to the existing workflow (e.g. adding a pseudo-CT acquisition to the existing clinical protocol) and simulate the MR-only workflow for these patients next to their clinical workup. Validate whether the variation of registration results (translations/rotations) of the MR-only simulation is in the order of the existing CT-based workflow. Exact comparison of registration results is challenging since the reference CT (CT-sim) and MRI (for sCT) are acquired at different time points; hence a comparison of registration result variations is recommended. The number of patients is required to be sufficient to generate enough statistics for a proper comparison, for example ten patients or hundred CBCT registrations. Example analyses are given in literature [27–29].

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Variation of registration results	NA	Results need to be in the same order as position verification based on the CT-sim	-,-,3,-	Maintenance and updates; introduction new tumour site or position verification protocol

4.2.14 MR-MV isocentre coincidence

Description

In MR-linac systems, the MRI scanner and the linac each have their own isocentre. The MR isocentre is determined by the image gradient system. For treatment planning, the geometrical shift between the image isocentre and the linac isocentre needs to be determined to correctly position the image and contours in the linac frame of reference. After initial TPS calibration, the consistency of the vector between the isocentres should be assessed periodically. Two tests are described to compare the alignment of the isocentres. The mechanical isocentre of the gantry can either be determined by the

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integrated megavoltage portal imager using a phantom with multiple ceramic ball bearings mounted in a $CuSO_4$-filled framework, or using dosimetric film. For the latter, a star shot procedure can be used. Both the film or ceramic ball bearings are placed in an MRI-visible phantom to connect the gantry isocentre to the MR isocentre. For both tests it is important to use high imaging bandwidths to prevent B_0 induced displacements. The action limits are based on the minimum requirements as set by several guidelines, including AAPM TG104, AAPM TG 142, and AAPM TG179. An overview is presented in NCS 32 [42]. For SBRT or SRS treatments it is advised to aim for the acceptable tolerance of 0.5 mm or less. The full MR-MV coincidence test as described here should be performed at least monthly. However, a quick check (e.g. MRI isocenter constancy check added to the morning startup routine or weekly QA) is recommended.

Limits

Criteria	Acceptable	Critical	Applications	Minimum interval
Δ MR-MV isocentre distance (3D vector)	0.5 mm	1.0 mm	-,-,-,4	Monthly

4.2.15 B_0 direction

Description

The direction of the main magnetic field should be verified on MRI-Linac systems as it determines the direction of the dose kernel tilt, the electron return effect (ERE) [43] and the electron streaming effect (ESE) [44]. In order to ensure a correct dose calculation, it is therefore essential that the direction of the B_0 field is correctly modeled. After installation, the direction of the B_0 field should be measured and compared to the direction as modeled by the TPS. The measurement should be performed at least once during commissioning and after relevant maintenance (i.e. magnet ramp-up).

Limits

Criteria (pass/fail)	Acceptable	Critical	Applications	Minimum interval
Agreement between actual B_0 direction and TPS definition	Pass	Fail	-,-,-,4	Magnet ramp-up

4.2.16 Gantry-dependent B_0 homogeneity and MR isocentre shift

Description

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Because the gantry on an MR-linac contains large amounts of ferromagnetic material, the gantry may introduce spatially varying offsets to the B_0 field, which could lead to image artefacts or even a shift of the imaged object of up to 1 mm [45]. Potential shifts in the daily MR image (i.e. the reference image) are mitigated by using a consistent gantry angle for calibration, QC, and daily MRI. Nevertheless, gantry dependent shifts will still have an effect on the motion monitoring imaging *during* the treatment, since the gantry is moved to different angular positions during irradiation. These shifts, when not accounted for, can lead to small errors in the gating window.

For this test, B_0 field maps are obtained for gantry angles between 0° and 360° with 30° increments. The magnitude images are analysed for relative geometric shifts, while the field maps (expressed in Hz, ppm, or nT) provide information on the B_0 homogeneity.

Limits

Criteria[13]	**Acceptable**	**Critical**	**Applications**	**Minimum interval**
Apparent shift (max abs. deviation from location at reference angle)	0.5 mm	1.0 mm	-,-,-,4	Annually
Shim Peak-to-Peak	96 Hz	128 Hz	-,-,-,4	Annually

[13]Deviations are specified as a peak-to-peak values measured in a body-size phantom (45-50 cm), with DSV 350 mm and no active shimming.

4.2.17 Linac-induced RF interference

Description

The magnetron on both the ViewRay MRIdian and the Elekta Unity system is mounted on the rotating gantry unit and is a potential source of RF noise when not shielded properly. In order to test whether RF noise is picked up by the MR subsystem, images may be acquired with and without the magnetron turned on and visually inspected for image artefacts. In addition, noise-only scans (i.e. images acquired without RF excitation) can be acquired, which makes a more sensitive scan to detect unwanted RF signals. The standard procedure offered by the MR vendor can be followed here. At the time of writing no quantitative limits have been established within the RT community, which is why the limits are specified qualitatively.

Limits

Criteria (pass/fail)	**Acceptable**	**Critical**	**Applications**	**Minimum interval**

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No observable image degradation during linac operation on scans that are clinically used	Pass	Fail	-,-,-,4	Annually

4.2.18 Radiation-induced RF artefacts

Description

In addition to the RF generated by the magnetron, the interaction of high energy photons with the electronics in the receive coils can introduce electronic noise in the RF receive chain of the MR-linac [46,47]. The effect of radiation will appear as short bursts of discharges that are delivered with the pulse repetition frequency (PRF) of the linac. The signal bursts will appear as small spikes in k-space. Because the location of the spikes is predetermined (by the PRF), their appearance in image space will be dependent on the acquisition parameters, like the repetition time TR, echo spacing, and readout order, as well as the actual field size.

To test the clinical relevance, the standard imaging protocols may be tested with and without radiation and inspected for image artefacts. It is important that no additional noise is observed when a 10x10 cm^2 field is used during irradiation (some additional noise may be observed when using the maximum field size as scattered photons may reach the sensitive components on the circuit boards of the RF receive coils).

Limits

Criteria (pass/fail)	Acceptable	Critical	Applications	Minimum interval
No observable image degradation while irradiating a 10x10 cm^2 field	Pass	Fail	-,-,-,4	Annually

4.2.19 Temporal stability test

Description

During long measurements, such as cine during beam delivery, temporal drift can cause geometric offsets and changes in banding artefacts in balanced Steady-State Free-Precession (bSSFP) [14], images due to B_0 drift. Therefore, it is recommended to assess geometric drifts and image degradation in clinically relevant cine images during a clinically relevant acquisition time.

A clinical time series (cine) acquisition is acquired for a duration that is representative of a treatment delivery time (e.g. 10 minutes). During the acquisition bSSFP banding artefacts should not drift through the image such that they affect the tracking algorithm. To ensure no geometric drifts have occurred,

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the time series data can be analysed offline by registering the data to a single reference image or a selection of the images can be compared at the console using the ROI and/or measurement tools available.

Limits

Criteria [15]	**Acceptable**	**Critical**	**Applications**	**Minimum interval**
Maximum drift	0.5 mm	1 mm	-,-,-,4	Annually

[14]bSSFP is referred to as TruFi (MRIdian) or bFFE (Unity)

[15]Additionally, the entire image time series is free of artefacts that could influence the tracking algorithm.

4.2.20 Real-time feedback latency

Description

The latency (i.e. delay) between the actual motion of the object and the correctional action by the linac determines the efficacy of the motion management. The gating latency depends on duration of the entire feedback chain, which includes: image acquisition, image reconstruction, image processing (i.e. the tracking algorithm), and radiation beam triggering. It has been shown that the image acquisition component can make up 75% of the total latency, and depends on the type of readout that is chosen [48]. Typical gating latencies between 200 ms and 400 ms have been reported for both the MRIdian and Elekta system [49,50]. In these studies, the measured latency was dependent on the acquisition frame rate as well as the tracking algorithm that was used. To measure the latency, an MR-compatible 4D motion phantom with MR-visible moving target can be used. Imaging latency is highly dependent on the sequence and timing parameters so it is essential to use clinically relevant sequences. By recording both the true position of the object (output of the 4D phantom) and the electronic trigger of the linac, the total latency can be measured. One experiment typically consists of a series of gating cycles from which the average beam-on and beam-off latency are recorded. Since the total latency depends strongly on the imaging sequence, it is important to test all cine sequences that are employed clinically. The limits given in the table below are constancy limits that can be used once a baseline for each sequence has been established. In absolute terms, AAPM Task Group 76 states that latencies > 500 ms, should be accounted for when setting the gating interval for respiratory gating. To improve gating accuracy, the tumor position can be predicted using adaptive filters, but the accuracy of the prediction decreases rapidly with delays longer than 200 ms [51,52].

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Limits

Criteria (constancy)	Acceptable	Critical	Applications	Minimum interval
Total latency	±50 ms	±100 ms[16]	-,-,-,4	Annually

[16] the critical value of 100 ms is in line with AAPM reports [51,53].

4.3 Practical considerations

QC testing of *diagnostic* MR systems is largely covered by system tests of MR manufacturers. If these tests are performed during periodic maintenance, and the institution verifies all tests are within manufacturer specifications, a minimum but sufficient level of QC is maintained for diagnostic applications. For the radiotherapy specific applications (2-4), it is advised to implement a QC program independent of the manufacturer tests in which additional tests are performed by the institution. To some extent the test phantoms as provided by the vendor may be used for this, although for certain tests (e.g. large FoV geometric distortion) a third-party phantom may be required as the phantom provided by the vendor may be too small. Another reason to use a third-party phantom is when scanner calibration and testing are performed on the same phantom by the vendor. Testing of gradient amplitude calibration (geometric accuracy) by manufacturer QA, for example, has a potential pitfall that both calibration and testing are performed on the same phantom and the same (edge) detection algorithm. This may result in an internally consistent but incorrect calibration. If this is the case it is recommended to test the gradient amplitude using another (rigid) phantom, e.g. the ACR phantom or dedicated (large FoV) geometric fidelity phantoms.

More details on the phantoms and detailed descriptions on how to perform each of the tests as described above are provided in Appendix A.

4.3.1 Frequency of QC tests

Regular QC may be implemented with an adaptive periodicity. For example, start with biweekly measurements of the parameters for which you want to build up reference values (SNR, image uniformity, resonance frequency and RF transmitter amplitude), and double the interval after about ten tests if there are no or few actions. Increase the interval to match the minimum interval of periodic maintenance as specified for each test. For other parameters, the initial interval may be directly set to the frequency of periodic maintenance.

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For application 1 (MRI not registered to CT) the periodic maintenance is the recommended minimum. For the other applications a time-based periodicity is set. In addition, however, QC should be performed:

- after relevant periodic maintenance (e.g. any re-calibration of the gradient subsystem or configuration files that influence the image reconstruction are particularly important as they may directly affect the geometric fidelity of the MR images);
- after repairs (to a component in the imaging chain);
- after software and hardware upgrades or updates;
- in the event abnormalities are suspected.

4.3.2 Coil testing

The MRI system is equipped with various RF coils. These all need to be tested; however, the interval can be varied per RF coil:

- The head coil or the torso coil (depending on local use) needs to be checked at each test interval;
- In addition, it is recommended to test SNR, image uniformity and image artefacts for all RF coils on a yearly basis. This would typically be tested by the manufacturer after periodic maintenance;
- Depending on local use, it is advisable to regularly check other RF coils for SNR and uniformity in addition to the regular checks on head or body coil. In this context, it is important to pay attention to quantitative parameters that can detect failing receive coil elements or channels.

4.4 Unmet needs

The list of QC tests proposed in this chapter cover all subsystems of the MRI system and is the minimum requirement to ensure adequate QA of MRI scanners that are used for radiotherapy workflows. The reader is, however, encouraged to include other tests during acceptance testing and commissioning if deemed necessary. Additional tests may include for example, spatial resolution, slice thickness, and slice position, which are included in most vendor provided QA protocols and well described by the ACR guideline [54]. QA specific to functional and quantitative imaging sequences (e.g. DWI, DCE, or relaxometry) are outside the scope of this report. For more information on QA specific to these types of functionalities the reader is referred to other guidelines (e.g. QIBA [55]).

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4D-MRI functionality and geometric checks on physiologically triggered scans are also excluded from this report, as the vendor offerings are currently limited and the clinical implementation not widespread. Some examples on how to perform QA for 4D-MRI are described in literature [56].

At the time of writing synthetic CT is not an option on the commercial MR-linac systems. If added in the future, similar tests as described in Sections 4.2.11 to 4.2.13 may be used, but tolerances will need to be reviewed.

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5 MR safety and introducing equipment in the scanner room

5.1 Introduction

An MRI scanner generates a strong static magnetic field, typically of the strength of 1.5 - 3.0 T. This static magnetic field is always present. During image acquisition, time-varying magnetic field gradients (gradient fields) and radiofrequency (RF) fields are applied. At the levels of the static magnetic fields, gradient fields and RF fields, no clear long-term harmful biological effects have been reported for patients and staff. Yet, the MR environment requires many control measures for safe operation. This chapter gives a brief summary of various risks specific to the MR environment, which RT personnel may not be familiar with. It is quite likely for instance, that for RT-dedicated scans additional equipment needs to be introduced in the MRI scanner room.

One may divide risks related to MR into two categories:

- *Accidents:*

  Ferromagnetic objects may be attracted by the magnetic field and propel towards the scanner, potentially harming the patient, staff, accompanying persons or equipment. Medical implants and metal objects in the body may be subject to torque/forces when introduced into the static magnetic field. The RF fields may result in heating or malfunction of the implants.
- *Device interference:*

  Objects in the scanner room may deteriorate image quality due to RF interference (RF noise, spikes) and RF shielding. Furthermore, image artefacts may arise such as local signal dropout and image deformation due to magnetic field inhomogeneity (susceptibility artefacts). The static field, time-varying gradient fields or RF field may also influence the functioning of devices brought into the scanner room.

In this chapter, first, a synopsis of general MR safety is given. Second, safety considering implants is discussed, followed by safety aspects when using (RT-specific) equipment during MRI acquisition. Finally, a protocol is proposed for safely introducing equipment into the scanner room. This section is a short summary of some aspects of MR safety, for a more detailed and complete overview of MR safety aspects the reader is referred to the provided references.

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5.2 General MR safety

In order to minimise risks at the MRI scanner, adequate safety measures need to be implemented at the department. These precautions are described in great detail in several MRI safety guidelines [57–61]. In general, it is advised to adhere to the hospital's local (radiology) protocols. These should address at least the following aspects:

- A proper MR safety organisation within the department, with an MR safety officer (MRSO, most often a senior technologist working with the scanner) appointed, and access to an MR safety expert (MRSE, typically a medical physicist or an MR physicist).
- A Risk Inventory and Evaluation should be in place for the MR personnel, according to European legislation. A generic Dutch inventory has been published by the NVKF [62]
- Adequate training of employees. Everyone working in the MRI scanner room must undergo safety training. This also applies to external staff: e.g. anaesthetists, cleaning personnel and property service staff. A registry of trained personnel should be present.
- Screening of people entering the scanner room. Staff members must be checked in accordance with the relevant screening form before they enter the scanner room for the first time. All patients (and volunteers) must be checked in accordance with the relevant screening form before entering the scanner room.
- Signs at the MRI scanner. All rooms with field strength over 0.5 mT should have a warning sign posted [57], with texts such as "*Warning! Strong magnetic field*" and "*Do not enter if you have a pacemaker or an electrical/battery powered implant*". Furthermore, the entrance to the scanner room should have additional signs stating: "*Warning! Magnetic field is always on*", "*Do not enter if you have an implanted metal object*" and "*No loose metal objects*". See Figure **4** for examples.
- Labels on equipment and devices. All equipment that enters the scanner room should be identified according to the FDA labelling criteria developed by the ASTM [60] (see Figure 5):
    - **MR Safe:** Devices that do not pose any risk in the scanner room, e.g. non-metallic devices.
    - **MR Conditional**: Devices that do not pose any risks in the MR environment under certain specific conditions. Conditions should be part of the labelling on the device (e.g. MR Conditional: Maximum 3.0 Tesla, only use outside the 0.5 mT region).
    - **MR Unsafe:** Devices that pose a risk in the scanner room, e.g. devices with ferromagnetic parts. These devices may only be brought into the scanner room under

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the direct supervision of specifically designated MR personnel who are thoroughly familiar with the device, its function and the reason for its introduction to the room.

- **Not known:** Never assume MR compatibility if the device is not clearly labelled and documented. A device is considered MR Unsafe until it has been tested and deemed otherwise.

Figure 4: Example signs for an MRI scanner room, indicating the various hazards. The IPEM offers free downloads of MRI safety notices [63].

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Figure 5: U.S. FDA labelling criteria (developed by ASTM and included in IEC-62570) for devices taken into the scanner room [60]. These figures are redrawings from Wikimedia Commons [64]. Green square: MR Safe; yellow triangle: MR Conditional; red circle: MR Unsafe.

5.3 MR safety: implants

All people entering the scanner room must first pass an MR safety screening process. This is the case for technicians and patients, non-MR personnel (e.g. anaesthetist, cleaning staff) and for accompanying persons (e.g. parents of a paediatric patient). Only trained MR personnel are authorised to perform an MR safety screening. Any individual undergoing an MR procedure should remove all readily removable metallic items on them. Careful assessment should be performed when an individual has a medical implant that is not readily removable. Most radiology departments have a dedicated MR safety protocol in place, and it is advised to adhere to the same (local) safety regulations.

Medical implants may interact with RF fields and/or magnetic fields of the MRI scanner. Different restrictions and conditions may apply to different models of the same manufacturer; therefore, it is necessary to know both the type and model of the implant to make a proper assessment. Shellock compiles data on the safety of implants and other metallic objects in an MR environment [65]. The database is updated regularly and reports conditions at which the implant can be safely scanned. Furthermore, commercial solutions for MR safety databases are available. Local procedures should be adhered to, and national guidelines have been compiled for the usage of MRI for patients with implants [61].

5.4 MR safety: RT specific equipment

For the purpose of radiation therapy, additional equipment may be needed during MR acquisition, for instance for patient positioning and stabilisation. Much equipment for radiation therapy consists of carbon fibre and may contain (ferromagnetic) metal parts and are thus not suitable for usage at the MRI scanner. Carbon fibre may be heated during the scan if conductive, and introduce RF-shielding artefacts, reducing signal-to-noise ratio for spin-echo, turbo spin-echo and gradient-echo sequences

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[66]. However, most manufacturers provide MR Safe alternatives for radiation therapy positioning equipment, which enable exact reproducibility of the patient at the MRI scanner, CT and PET-CT scanner, and treatment machines. Of course, the equipment should be labelled and personnel should be properly instructed for use.

5.5 Bringing unlabelled equipment into the scanner room

In some cases, equipment which is not yet labelled needs to be introduced into the scanner room. Equipment is considered MR Unsafe until it is tested and labelled by the local MR safety expert (MRSE), according to ACR recommendations [57]. Some equipment needs to be brought into the scanner room, but does not have to be positioned in the magnet bore. This equipment then requires testing for interactions with the static magnetic field, switching gradient fields (if close to the magnet) and RF interference. Equipment positioned in the scanner bore during imaging needs to be tested for interactions with the gradient fields and RF fields as well.

MR safety testing for equipment inside the MR room or MRI scanner bore generally consists of the following consecutive steps, depending on the application of the equipment (e.g. usage in the scanner room outside of the MRI bore, or usage of the equipment in the MRI bore during acquisition):

1. **Checking for ferromagnetic parts:** This step needs to be done outside the scanner room, by using a handheld magnet with considerable field strength (> 0.1 T). Other metallic parts can be detected by making a CT scan or an X-ray image of the equipment. If the equipment contains any ferromagnetic parts, it is MR Unsafe or MR Conditional and may not be introduced in the scanner room without supervision or regulations of the MRSE. If no ferromagnetic parts are present, the equipment is MR Conditional and further testing is needed [57]. It is only when the composition of the equipment and its components are known to be non-magnetic and not electrically conductive (not carbon or metal), the equipment is MR Safe.
2. **Bringing the equipment in the scanner room:** Enter the scanner room holding the equipment, slowly approaching the entrance of the scanner bore. Take care that nobody is between the scanner and the equipment. Make sure that no attractive force is observed. If substantial attractive force is observed, the equipment is MR Unsafe and should not be used in the scanner room. It is common practice to compare attractive force with gravity to decide on safety.

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3. **Bringing the equipment in the scanner bore:** By carefully moving the equipment at high speed through the magnetic field in the scanner bore entrance, validate that no Lenz forces and torque are observed, caused by electrical conductivity of non-ferrous material. Furthermore, testing for heating is essential, and strict limits apply for heating if the equipment is in contact with the patient [67,68]. Note that carbon materials may not produce strong Lenz forces, but can become hot during scanning. If Lenz forces on (or heating of) the equipment in the scanner bore is observed, the device is generally MR Conditional and should not be used inside the scanner bore during image acquisition.
4. **Interference of the MRI scanner on the equipment:** test whether the equipment is still functioning within specifications inside the scanner room, and during image acquisition. This especially holds for equipment with electronic parts. Test sequences with worst-case scenario gradient fields and RF field, e.g. maximum specific absorption rate (or SAR) values, may be used to validate that the equipment works properly during clinical scanning protocols. If the equipment is not operational during scanning under the worst-case scenario acquisition parameters, the equipment is MR Conditional, and conditions apply.
5. **Interference of the equipment on the MRI scanner**: test whether the image quality is not negatively affected. It is recommended to scan a phantom together with the equipment with appropriate QC protocols. Image quality can then be assessed, taking into account signal-to-noise levels, occurrence of RF spikes or other RF interferences, and geometrical distortions, amongst others. Compare MR images acquired with and without equipment. For equipment inside the scanner bore, geometrical distortions due to the equipment device can be assessed visually by repeating scanning the object submerged in a water tank, taking an MRI acquisition twice, one with opposite read-out gradient polarity. The resulting images can be subtracted revealing the difference in equipment position due to image deformations [69]. Another option is to acquire a B_0 map and calculate the expected image distortions. Note that the equipment needs to be positioned in the centre of the phantom, and that a square phantom geometry induces B_0 inhomogeneities as well, which results in additional image deformations. Preferably use phantoms with rounded shapes (balls, ovals, cylinder).

The equipment should be labelled according to the ASTM guidelines (Figure 5, [60]). Equipment may only be labelled as MR Safe if the equipment is safe in the whole scanner room, and in the scanner bore during scanning. If equipment is MR Conditional, the specific conditions should be specified, e.g. MR Conditional having been tested to be safe at 3.0 T at gradient strengths of less than 40mT/cm and

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a SAR of less than 2 W/kg. Or: MR Conditional, equipment may be present in the scanner room fixated at the wall and outside the 20 mT field line.

The results of testing should be documented by the MRSE. A test report should contain at least the date of testing, the name of the person who performed the test and the test results. The department should have an archiving system in place where such reports are accessible for employees to consult.

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6 Bibliography

[1] Speight R, Tyyger M, Schmidt MA, Liney G, Johnstone R, Eccles CL, et al. IPEM Topical Report: an international IPEM survey of MRI use for external beam radiotherapy treatment planning. Phys Med Biol 2021;66:075007. https://doi.org/10.1088/1361-6560/abe9f7.

[2] Glide-Hurst CK, Paulson ES, McGee K, Tyagi N, Hu Y, Balter J, et al. Task group 284 report: magnetic resonance imaging simulation in radiotherapy: considerations for clinical implementation, optimization, and quality assurance. Med Phys 2021;48. https://doi.org/10.1002/mp.14695.

[3] Jackson E, Bronskill M, Drost D, Och J, Pooley R, Sobol W, et al. Acceptance Testing and Quality Assurance Procedures for Magnetic Resonance Imaging Facilities. 2010. https://doi.org/10.37206/101.

[4] Price R, Allison J, Clarke G, Dennis M, Hendrick RE, Keener C, et al. Magnetic Resonance Imaging - Quality Control Manual. 2015.

[5] Jeukens C, Hofman M, Schilham A, van Dalen J. Leidraad Kwaliteitscontrole Radiologische Apparatuur - NVKF. 2019.

[6] Glover GH. Overview of Functional Magnetic Resonance Imaging. Neurosurg Clin N Am 2011;22:133–9. https://doi.org/10.1016/j.nec.2010.11.001.

[7] Stemkens B, Paulson ES, Tijssen RHN. Nuts and bolts of 4D-MRI for radiotherapy. Phys Med Biol 2018;63:21TR01. https://doi.org/10.1088/1361-6560/aae56d.

[8] Raunig DL, McShane LM, Pennello G, Gatsonis C, Carson PL, Voyvodic JT, et al. Quantitative imaging biomarkers: A review of statistical methods for technical performance assessment. Stat Methods Med Res 2015;24:27–67. https://doi.org/10.1177/0962280214537344.

[9] Bernstein MA, King KF, Zhou XJ. Handbook of MRI Pulse Sequences. Elsevier; 2004. https://doi.org/10.1016/B978-0-12-092861-3.X5000-6.

[10] Brown RW, Cheng Y-CN, Haacke EM, Thompson MR, Venkatesan R. Magnetic Resonance Imaging: Physical Principles and Sequence Design. Second. Wiley-Blackwell; 2014.

[11] McRobbie DW, Moore EA, Graves MJ, Prince MR. MRI From Picture to Proton. Second. Cambridge: Cambridge University Press; 2006. https://doi.org/10.1017/CBO9780511545405.

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

[12] Runge VM, Nitz WR, Schmeets SH. The Physics of Clinical MR Taught Through Images. Second. Stuttgart: Georg Thieme Verlag; 2009. https://doi.org/10.1055/b-002-59228.

[13] Plewes DB, Kucharczyk W. Physics of MRI: A primer. Journal of Magnetic Resonance Imaging 2012;35:1038–54. https://doi.org/10.1002/jmri.23642.

[14] Schmidt MA, Payne GS. Radiotherapy planning using MRI. Phys Med Biol 2015;60:R323–61. https://doi.org/10.1088/0031-9155/60/22/R323.

[15] Doran SJ, Charles-Edwards L, Reinsberg SA, Leach MO. A complete distortion correction for MR images: I. Gradient warp correction. Phys Med Biol 2005;50:1343–61. https://doi.org/10.1088/0031-9155/50/7/001.

[16] Wang D, Doddrell D. Geometric Distortion in Structural Magnetic Resonance Imaging. Curr Med Imaging Rev 2005;1:49–60. https://doi.org/10.2174/1573405052953029.

[17] Sumanaweera TS, Adler JR, Napel S, Glover GH. Characterization of Spatial Distortion in Magnetic Resonance Imaging and Its Implications for Stereotactic Surgery. Neurosurgery 1994;35:696–704. https://doi.org/10.1227/00006123-199410000-00016.

[18] Liney GP, Moerland MA. Magnetic Resonance Imaging Acquisition Techniques for Radiotherapy Planning. Semin Radiat Oncol 2014;24:160–8. https://doi.org/10.1016/j.semradonc.2014.02.014.

[19] Gunter JL, Bernstein MA, Borowski BJ, Ward CP, Britson PJ, Felmlee JP, et al. Measurement of MRI scanner performance with the ADNI phantom. Med Phys 2009;36:2193–205. https://doi.org/10.1118/1.3116776.

[20] Jezzard P, Clare S. Sources of distortion in functional MRI data. Hum Brain Mapp 1999;8:80–5. https://doi.org/10.1002/(SICI)1097-0193(1999)8:2/3<80::AID-HBM2>3.0.CO;2-C.

[21] Dawson LA, Ménard C. Imaging in Radiation Oncology: A Perspective. Oncologist 2010;15:338–49. https://doi.org/10.1634/theoncologist.2009-S106.

[22] Vinod SK, Min M, Jameson MG, Holloway LC. A review of interventions to reduce inter-observer variability in volume delineation in radiation oncology. J Med Imaging Radiat Oncol 2016;60:393–406. https://doi.org/10.1111/1754-9485.12462.

[23] Fortunati V, Verhaart RF, Verduijn GM, van der Lugt A, Angeloni F, Niessen WJ, et al. MRI integration into treatment planning of head and neck tumors: Can

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

patient immobilization be avoided? Radiotherapy and Oncology 2015;115:191–4. https://doi.org/10.1016/j.radonc.2015.03.021.

[24] Verduijn GM, Bartels LW, Raaijmakers CPJ, Terhaard CHJ, Pameijer FA, van den Berg CAT. Magnetic Resonance Imaging Protocol Optimization for Delineation of Gross Tumor Volume in Hypopharyngeal and Laryngeal Tumors. International Journal of Radiation Oncology*Biology*Physics 2009;74:630–6. https://doi.org/10.1016/j.ijrobp.2009.01.014.

[25] Nyholm T, Jonsson J. Counterpoint: Opportunities and Challenges of a Magnetic Resonance Imaging–Only Radiotherapy Work Flow. Semin Radiat Oncol 2014;24:175–80. https://doi.org/10.1016/j.semradonc.2014.02.005.

[26] Hoffmann A, Oborn B, Moteabbed M, Yan S, Bortfeld T, Knopf A, et al. MR-guided proton therapy: a review and a preview. Radiation Oncology 2020;15:129. https://doi.org/10.1186/s13014-020-01571-x.

[27] Maspero M, van den Berg CAT, Zijlstra F, Sikkes GG, de Boer HCJ, Meijer GJ, et al. Evaluation of an automatic MR-based gold fiducial marker localisation method for MR-only prostate radiotherapy. Phys Med Biol 2017;62:7981–8002. https://doi.org/10.1088/1361-6560/aa875f.

[28] Goudschaal K, Beeksma F, Boon M, Bijveld M, Visser J, Hinnen K, et al. Accuracy of an MR-only workflow for prostate radiotherapy using semi-automatically burned-in fiducial markers. Radiation Oncology 2021;16:37. https://doi.org/10.1186/s13014-021-01768-8.

[29] Tyagi N, Fontenla S, Zhang J, Cloutier M, Kadbi M, Mechalakos J, et al. Dosimetric and workflow evaluation of first commercial synthetic CT software for clinical use in pelvis. Phys Med Biol 2017;62:2961–75. https://doi.org/10.1088/1361-6560/aa5452.

[30] Tyagi N, Fontenla S, Zelefsky M, Chong-Ton M, Ostergren K, Shah N, et al. Clinical workflow for MR-only simulation and planning in prostate. Radiation Oncology 2017;12:119. https://doi.org/10.1186/s13014-017-0854-4.

[31] Koch KM, Hargreaves BA, Pauly KB, Chen W, Gold GE, King KF. Magnetic resonance imaging near metal implants. Journal of Magnetic Resonance Imaging 2010;32:773–87. https://doi.org/10.1002/jmri.22313.

[32] Hargreaves BA, Worters PW, Pauly KB, Pauly JM, Koch KM, Gold GE. Metal-Induced Artifacts in MRI. American Journal of Roentgenology 2011;197:547–55. https://doi.org/10.2214/AJR.11.7364.

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

[33] Jonsson JH, Garpebring A, Karlsson MG, Nyholm T. Internal Fiducial Markers and Susceptibility Effects in MRI—Simulation and Measurement of Spatial Accuracy. International Journal of Radiation Oncology*Biology*Physics 2012;82:1612–8. https://doi.org/10.1016/j.ijrobp.2011.01.046.

[34] Christiansen RL, Johansen J, Zukauskaite R, Hansen CR, Bertelsen AS, Hansen O, et al. Accuracy of automatic structure propagation for daily magnetic resonance image-guided head and neck radiotherapy. Acta Oncol (Madr) 2021;60:589–97. https://doi.org/10.1080/0284186X.2021.1891282.

[35] National Electrical Manufacturers Association (NEMA). MS1:2008: Determination of Signal-to-Noise Ratio (SNR) in Diagnostic Magnetic Resonance Imaging. 2012.

[36] Phantom Test Guidance for Use of the Large MRI Phantom for the ACR MRI Accreditation Program. 2018.

[37] Baldwin LN, Wachowicz K, Thomas SD, Rivest R, Fallone BG. Characterization, prediction, and correction of geometric distortion in 3T MR images. Med Phys 2007;34:388–99. https://doi.org/10.1118/1.2402331.

[38] Moerland MA, Beersma R, Bhagwandien R, Wijrdeman HK, Bakker CJG. Analysis and correction of geometric distortions in 1.5 T magnetic resonance images for use in radiotherapy treatment planning. Phys Med Biol 1995;40:1651–64. https://doi.org/10.1088/0031-9155/40/10/007.

[39] Speight R, Dubec M, Eccles CL, George B, Henry A, Herbert T, et al. IPEM topical report: guidance on the use of MRI for external beam radiotherapy treatment planning *. Phys Med Biol 2021;66:055025. https://doi.org/10.1088/1361-6560/abdc30.

[40] Meijer GJ, van Kleffens HJ, Mijnheer BJ. NCS Report 11: Quality control (QC) of simulators and CT scanners and some basic QC methods for treatment planning systems. Delft: 1997. https://doi.org/10.25030/ncs-011.

[41] Spadea MF, Maspero M, Zaffino P, Seco J. Deep learning based synthetic-CT generation in radiotherapy and PET: A review. Med Phys 2021;48:6537–66. https://doi.org/10.1002/mp.15150.

[42] Remeijer P, Deurloo K, Eenink M, Geleijns K, Hermans J, van Herpt H, et al. NCS Report 32: Quality assurance of cone-beam CT. Delft: 2019. https://doi.org/10.25030/ncs-032.

[43] Raaijmakers AJE, Raaymakers BW, Meer S van der, Lagendijk JJW. Integrating a MRI scanner with a 6 MV radiotherapy accelerator: impact of the surface

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

orientation on the entrance and exit dose due to the transverse magnetic field. Phys Med Biol 2007;52:929–39. https://doi.org/10.1088/0031-9155/52/4/005.

[44] Hackett SL, van Asselen B, Wolthaus JWH, Bluemink JJ, Ishakoglu K, Kok J, et al. Spiraling contaminant electrons increase doses to surfaces outside the photon beam of an MRI-linac with a perpendicular magnetic field. Phys Med Biol 2018;63:095001. https://doi.org/10.1088/1361-6560/aaba8f.

[45] Latifi K, Moros EG, Zhang G, Harrison L, Feygelman V. A Method to Determine the Coincidence of MRI-Guided Linac Radiation and Magnetic Isocenters. Technol Cancer Res Treat 2019;18:153303381987798. https://doi.org/10.1177/1533033819877986.

[46] Burke B, Wachowicz K, Fallone BG, Rathee S. Effect of radiation induced current on the quality of MR images in an integrated linac-MR system. Med Phys 2012;39:6139–47. https://doi.org/10.1118/1.4752422.

[47] Hoogcarspel SJ, Zijlema SE, Tijssen RHN, Kerkmeijer LGW, Jürgenliemk-Schulz IM, Lagendijk JJW, et al. Characterization of the first RF coil dedicated to 1.5 T MR guided radiotherapy. Phys Med Biol 2018;63:025014. https://doi.org/10.1088/1361-6560/aaa303.

[48] Borman PTS, Tijssen RHN, Bos C, Moonen CTW, Raaymakers BW, Glitzner M. Characterization of imaging latency for real-time MRI-guided radiotherapy. Phys Med Biol 2018;63:155023. https://doi.org/10.1088/1361-6560/aad2b7.

[49] Kim T, Lewis B, Lotey R, Barberi E, Green O. Clinical experience of MRI 4D QUASAR motion phantom for latency measurements in 0.35T MR-LINAC. J Appl Clin Med Phys 2021;22:128–36. https://doi.org/10.1002/acm2.13118.

[50] Glitzner M, Woodhead PL, Borman PTS, Lagendijk JJW, Raaymakers BW. Technical note: MLC-tracking performance on the Elekta unity MRI-linac. Phys Med Biol 2019;64:15NT02. https://doi.org/10.1088/1361-6560/ab2667.

[51] Keall PJ, Mageras GS, Balter JM, Emery RS, Forster KM, Jiang SB, et al. AAPM Task Group 76: The management of respiratory motion in radiation oncology. Med Phys 2006;33:3874–900. https://doi.org/10.1118/1.2349696.

[52] Keall PJ, Sawant A, Berbeco RI, Booth JT, Cho B, Cerviño LI, et al. AAPM Task Group 264: The safe clinical implementation of MLC tracking in radiotherapy. Med Phys 2021;48. https://doi.org/10.1002/mp.14625.

[53] Klein EE, Hanley J, Bayouth J, Yin F-F, Simon W, Dresser S, et al. AAPM Task Group 142: Quality assurance of medical accelerators. Med Phys 2009;36:4197–212. https://doi.org/10.1118/1.3190392.

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

[54] Price R, Allison J, Clarke G, Dennis M, Hendrick RE, Keener C, et al. ACR - Magnetic Resonance Imaging Quality Control Manual. 2015.

[55] Shukla-Dave A, Obuchowski NA, Chenevert TL, Jambawalikar S, Schwartz LH, Malyarenko D, et al. Quantitative imaging biomarkers alliance (QIBA) recommendations for improved precision of DWI and DCE-MRI derived biomarkers in multicenter oncology trials. Journal of Magnetic Resonance Imaging 2019;49:e101–21. https://doi.org/10.1002/jmri.26518.

[56] van Kesteren Z, van der Horst A, Gurney-Champion OJ, Bones I, Tekelenburg D, Alderliesten T, et al. A novel amplitude binning strategy to handle irregular breathing during 4DMRI acquisition: improved imaging for radiotherapy purposes. Radiation Oncology 2019;14:80. https://doi.org/10.1186/s13014-019-1279-z.

[57] Kanal E, Barkovich AJ, Bell C, Borgstede JP, Bradley WG, Froelich JW, et al. ACR guidance document on MR safe practices: 2013. Journal of Magnetic Resonance Imaging 2013;37:501–30. https://doi.org/10.1002/jmri.24011.

[58] Netherlands Working Group on MR Safety. Using MRI safely: practical rules for employees. 2008.

[59] IEC 60601-2-33. Medical electrical equipment – Part 2-33: Particular requirements for the basic safety and essential performance of magnetic resonance equipment for medical diagnosis. 2010.

[60] American Society for Testing and Materials (ASTM). F2503-13: Standard Practice for Marking Medical Devices and Other Items for Safety in the Magnetic Resonance Environment. 2013.

[61] Society for Medical Physics of the Netherlands (NVKF). Guideline Use of MRI in patients with implants. 2019.

[62] Hofman M, van Straten M, Kappert P, van de Berg N, Jennekens W, Kuijer J, et al. Generieke verdiepende RI&E MRI voor werken met MRI - NVKF. 2019.

[63] IPEM MRI Safety Notices n.d. https://www.ipem.ac.uk/resources/mri-safety-notices-magnetic-resonance-imaging/ (accessed September 6, 2022).

[64] Wikimedia Commons n.d. https://commons.wikimedia.org/wiki/Main_Page (accessed September 6, 2022).

[65] Shellock FG. MRISafety.com 2022. http://www.mrisafety.com/ (accessed March 9, 2022).

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

[66] Jafar MM, Reeves J, Ruthven MA, Dean CJ, MacDougall ND, Tucker AT, et al. Assessment of a carbon fibre MRI flatbed insert for radiotherapy treatment planning. Br J Radiol 2016;89:20160108. https://doi.org/10.1259/bjr.20160108.

[67] American Society for Testing and Materials (ASTM). F2182-19: Standard Test Method for Measurement of Radio Frequency Induced Heating On or Near Passive Implants During Magnetic Resonance Imaging. 2019.

[68] Winter L, Seifert F, Zilberti L, Murbach M, Ittermann B. MRI-Related Heating of Implants and Devices: A Review. Journal of Magnetic Resonance Imaging 2021;53:1646–65. https://doi.org/10.1002/jmri.27194.

[69] van Heerden LE, Gurney-Champion OJ, van Kesteren Z, Houweling AC, Koedooder C, Rasch CRN, et al. Quantification of image distortions on the Utrecht interstitial CT/MR brachytherapy applicator at 3T MRI. Brachytherapy 2016;15:118–26. https://doi.org/10.1016/j.brachy.2015.10.008.

[70] ACR large MRI accreditation phantom n.d. https://cspmedical.com/acr-large-mri-accreditation-phantom/ (accessed September 4, 2022).

[71] Henkelman RM. Measurement of signal intensities in the presence of noise in MR images. Med Phys 1985;12:232–3. https://doi.org/10.1118/1.595711.

[72] Constantinides CD, Atalar E, McVeigh ER. Signal-to-noise measurements in magnitude images from NMR phased arrays. Magn Reson Med 1997;38:852–7. https://doi.org/10.1002/mrm.1910380524.

[73] Och JG, Clarke GD, Sobol WT, Rosen CW, Mun SK. AAPM Task Group 6: Acceptance testing of magnetic resonance imaging systems. Med Phys 1992;19:217–29. https://doi.org/10.1118/1.596903.

[74] Tijssen RHN, Philippens MEP, Paulson ES, Glitzner M, Chugh B, Wetscherek A, et al. MRI commissioning of 1.5T MR-linac systems – a multi-institutional study. Radiotherapy and Oncology 2019;132:114–20. https://doi.org/10.1016/j.radonc.2018.12.011.

[75] Pauly J. Field Maps n.d. https://ece-classes.usc.edu/ee591/library/Pauly-FieldMaps.pdf (accessed September 4, 2022).

[76] Lewis BC, Gu B, Klett R, Lotey R, Green OL, Kim T. Characterization of radiotherapy component impact on MR imaging quality for an MRgRT system. J Appl Clin Med Phys 2020;21:20–6. https://doi.org/10.1002/acm2.13054.

[77] SPM n.d. https://www.fil.ion.ucl.ac.uk/spm/ (accessed September 4, 2022).

[78] FSL n.d. https://fsl.fmrib.ox.ac.uk/fsl/fslwiki/FSL (accessed September 4, 2022).

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

[79] ImageJ n.d. https://imagej.net/imaging/registration (accessed September 4, 2022).

[80] Elastix n.d. https://elastix.lumc.nl/doxygen/index.html (accessed September 4, 2022).

https://doi.org/10.25030/ncs-036 The NCS report has been downloaded on downloaded on 18 May 2024

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Appendix A: Detailed test description per QC test

A.1 Prerequisites

A.1.1 Phantoms

For the generic QC tests (test 1 - 8), a phantom for MRI quality control is needed, with the following characteristics:

- It is recommended that the phantom is approximately the size of a head and that it is positioned in a reproducible way inside the radio frequency (RF) coil;
- It is advisable that the phantom contains water or gel with an additive resulting in T1, T2 and T2* relaxation times similar to tissue. Additionally, a stabiliser is important to prevent bacterial/fungal growth in the phantom;
- It should be possible to accurately determine the diameter of the phantom in different orientations (a cylindrical phantom is preferred). It is recommended that the phantom contains a section with homogeneous MR signal intensity (for homogeneity and SNR measurements);
- Finally, it is advised that the phantom is electrically conductive, in order to provide RF loading for the RF coil, similar to the body part for which the RF receiving coil is intended.

Example of a generic MRI phantom:

- large ACR phantom (JM Specialty Parts Inc. US, American College of Radiology (ACR) [36]) including holder [70]

The other tests (9 - 20) that are set up to test specific radiotherapy applications require dedicated phantoms in some cases. Examples include, but are not limited to:

- Laser alignment (e.g., Aquarius phantom);
- MR-MV (currently no 3rd party available, MRL vendor phantoms available);
- Large FoV geometric distortion (e.g., CIRS Large Field MR Image Distortion phantom, Modus Quasar MRID3D phantom, Gamma Gurus GRADE phantom);
- Real-time feedback latency (e.g., Modus Quasar MRID4D phantom, Zeus MRgRT Motion Management QA phantom);
- Motion phantoms to test 4DMRI or other dynamic applications (e.g., Modus Quasar MRID4D phantom, Zeus MRgRT Motion Management QA phantom);

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A.1.2 RF coils

The head coil or the body coil (depending on local use) needs to be checked at each test. In addition, it is recommended to test SNR, image uniformity and image artefacts for all RF coils on a yearly basis. Typically, the manufacturer would perform these tests after periodic maintenance. Depending on local use, it is advisable to regularly check other RF coils for SNR and uniformity in addition to the regular checks on head or body coil. In this context, it is recommended to pay attention to quantitative parameters that can detect failing receive coil elements or channels.

Flex-coils can be considered especially if they are used for measurements with fixations masks, e.g. for head-neck. However, coil dependent parameters such as SNR and uniformity are highly sensitive to phantom and coil positioning. However, the added value of QC is limited for these coils since relevant deviations will immediately be visible on clinical scans as well.

A.1.3 Tests performed on the MR-linac

The MR-linac tests specified in this appendix are specific to the hardware modifications of such a hybrid system and assess the possible interaction of the linac and gantry components on the image quality, the geometric accuracy of MRI, and the latency of the real-time feedback loop. The instructions are based on the system designs of the current offerings that are commercially available. At the time of writing these are: The 1.5 T Elekta Unity and the 0.35 T ViewRay MRIdian. Unless stated otherwise, it is recommended to conduct *all* QC tests on the MR-linac with the following consistent setup:

- Magnetron turned off;
- Consistent Gantry position (ideally at which the daily clinical images are also acquired);
- Anterior and posterior receive array in place.

A.2 Methods description per test

A.2.1 Signal-to-noise ratio

Introduction

The signal-to-noise ratio (SNR) is a generic parameter determined by several system components. In MR imaging it is a sensitive parameter; deviations of the order of 5% can be measured systematically. The absolute value of the SNR is highly dependent on a variety of scan parameters, the phantom used (especially its RF load) and the specific MR system used. System parameters that affect the SNR are:

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- The RF transmission chain. If the B_1 field (RF field) deviates from its target, the RF pulse has effectively a different flip angle, which will change the signal strength. Thus, SNR can be linked to the RF transmitter amplitude;
- The RF receive chain. Problems in the RF receive coil, preamplifiers and ADC can result in SNR loss;
- Errors in the gradient system can lead to different effective voxel sizes that also affect the SNR, either lower or higher.
- Interference from other RF sources can increase noise, which also lowers SNR. A similar effect occurs in the event of a Faraday cage leak (RF sources outside the cage);
- Finally, all kinds of reconstruction parameters also influence the SNR, such as filtering of the raw data.

Because errors in the RF receive coil in particular occur relatively often, the SNR determination is an important test to detect quality deterioration. Limited image uniformity makes this measurement sensitive to precise phantom placement.

Clinical relevance

For medical imaging it is important to realise that this parameter directly affects image quality, especially in those MRI protocols with intrinsically low SNR, such as: spectroscopy applications, fMRI, dynamic scans for perfusion. The SNR determines to a large extent the detectability of small low-contrast details.

Goal

- This test checks a specification from a manufacturer;
- The test determines a baseline value that can then be re-tested in a constancy test.

Method

The SNR of an MR system in combination with a specific coil can be measured in several ways:

a) By repeated measurement and signal variation per voxel (NEMA procedure [35]);
b) By determining the signal in the phantom and the noise in the background outside the phantom;
c) By making a separate noise scan without RF excitation.

Depending on the MRI system, certain methods may or may not be practically applicable.

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- *a) SNR noise measurement by repeated measurement (NEMA procedure)*

The SNR is determined by immediately repeating a scan (one or more times). The difference between the two shots is a measure of the noise, so image subtraction creates a noise image obtained. The NEMA definition of SNR is [35]:

$$SNR\_NEMA = (S1 + S2) / (noise * \sqrt{2})$$

with S1 and S2: mean pixel value of the ROI positioned in the centre of the phantom of image 1 and image 2 respectively. Noise is the standard deviation of the same ROI in the noise image. The advantage of this method is that it also works with parallel imaging. A disadvantage is that this measurement can be affected by the short-term system instability.

- *b) SNR noise measurement with noise determination in the background*

With this method, illustrated in Figure 6, the SNR is calculated by the following expression:

$$SNR = 0.655\ S_{phantom}/SD_{background}$$

with $S_{phantom}$ the mean pixel value of the ROI positioned in the centre of the phantom and $SD_{background}$ the standard deviation of a ROI in the noise image (background).

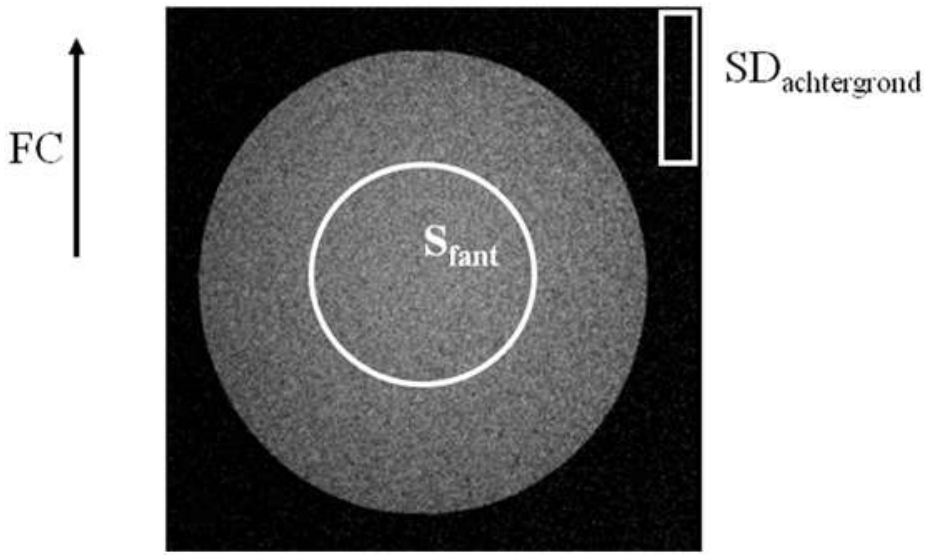

Figure 6: SNR determination on homogeneous phantom image. FC = phase encoding direction, S_{fant} = $S_{phantom}$ = mean pixel value of ROI in phantom, $SD_{achtergrond}$ = $SD_{background}$ = standard deviation of ROI in noise image.

Necessary steps to calculate the SNR in this manner:

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- Determine ROI in phantom at identical position each time, in order to prevent influence of signal inhomogeneity;
- Place background ROI such that ghosting of the phantom in the phase encoding direction (FC) does not contribute to the noise in the ROI. Do not choose this ROI too small, at least 40 pixels;
- Determine the standard deviation (SD) of the background ROI;
- The factor 0.655 is used because the noise is determined in a magnitude image, so that the actual noise is underestimated [71].

A limitation of this method is that in the MRI reconstruction, the noise in the background can be filtered, which makes noise estimations potentially inaccurate. This can be checked for by calculating the ratio of the signal and the SD in the background ROI: f_{mean} / SD = 1.91 (in areas where there is noise only, the Rician distribution converges to a Rayleigh distribution for which the mean / SD is equal to 1.91). The factor of 0.655 applies only to single-element RF coils and not to combined images of phased-array RF receive coils [72]. With parallel imaging, this method no longer works because the noise is inhomogeneous across the image.

- *c) SNR measurement using a separate noise image*

The SNR can also be determined by making a separate noise image in addition to a standard phantom image. The noise image needs to be scanned using exactly the same protocol, but using a 0° flip angle, thus obtaining a noise image. The SNR is simply calculated by:

SNR = signal/noise

with signal: mean pixel value of the ROI positioned in the centre of the phantom of the image. Noise is the standard deviation of the same ROI in the noise image.

The limitation of this method is that there needs to be a possibility on the scanner to easily record the RF pulse 0° without changing any other parameter. This is possible on certain scanners.

Also important for SNR measurements is the "loading" (load) of the coil through the phantom, which determines the noise level. Some MR systems use an extra part, a "loading-ring", to achieve this. By looking at the RF transmitter amplitude, a "loading" can be obtained comparable to a clinical study. This is especially important to the effect of external noise sources. In case a phantom with a very low "loading" is used, the external noise source can be largely overestimated.

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- *Additional and alternative checks*
    - Absolute SNR measurements are difficult, requiring phantom standardization. The NEMA protocol [35] provides guidelines. However, this goes beyond a constancy test;
    - As an additional measurement, the SNR measurement per individual coil element applies for a "phased array" receiving coil. Depending on the MR system, the MRI images can be obtained for individual elements of a "phased array" receive coil. This does not require additional acquisition time. The advantage of SNR measurement in separate coil elements is that it is more sensitive to finding errors in a single RF receive channel. Especially in case of a large number of elements, a deviation in a single element cannot immediately be visible in the global SNR, but cause a detectable SNR drop in a single element image. Larger element defects may be detected in the combined SNR or as inhomogeneity effects. Note that ROI selection may need to be adapted to the coil geometry for individual coil elements.

Pitfalls

- It is not recommended to measure the noise level in the foreground at the position where the signal level is determined, because the standard deviation of the pixel values partly depends on signal inhomogeneities;
- When determining the noise in the background, it is important to set the "windowing" in such a way that the background noise can be clearly distinguished from any filled black area outside the field of view;
- Especially when using a flexible coil, accurate positioning is important for a reliable measurement result.

A.2.2 Image uniformity

Introduction

Image uniformity is a measure of the MRI system's ability to produce a constant signal over the entire scanned volume of a homogeneous object. System parameters that affect image uniformity are:

- imbalance between the elements of the RF receiver coil, or failure of an RF receiver coil;
- homogeneity of the B_0 field (for Gradient Echo (GE, also known as (Fast) Field Echo, FFE) techniques, not for Spin Echo (SE) techniques);
- homogeneity of the RF excitation field (B1) of the RF transmit coil.

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Many scanners have filters to improve image uniformity in post-processing. The image uniformity is strongly dependent on the coil design and therefore varies greatly from coil to coil. Absolute determinations are typically part of the acceptance protocol.

Because errors in the RF receiver coil in particular are relatively common, the determination of the image uniformity is an important test to observe quality stability. In combination with SNR measurement of independent coil elements, this problem can be easily determined.

Clinical relevance

Due to poor image uniformity, the SNR can become too low in parts of the image, making the image partially not suitable for diagnostics.

Goal

- This test checks a supplier's specification;
- The test determines a baseline which can then be redetermined with a constancy test.

Method

Image uniformity can be measured in the same recording as the SNR recording; a phantom with homogeneous signal is needed. The image uniformity is an indication of the homogeneity of the MRI signal over the FoV and this parameter is expressed as the difference in signal intensity compared to an average pixel value.

The image uniformity can be calculated in different ways.

percentual image uniformity = 100% * [1-(maximum signal - minimum signal)/(maximum signal + minimum signal)] (see ACR manual p 106)

A large ROI is chosen depending on the phantom and the RF coil used. To reduce SNR influences, the maximum and minimum value of a certain ROI (size in the order of 1 cm) can also be selected, or the image can be filtered with a low-pass filter to reduce noise.

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Additional and alternative controls

If the uniformity is not good, it makes sense to determine the SNR for the different coil elements to check if a coil element is defective. NEMA uses a slightly different definition and calls it integral uniformity [35]:

Integral Uniformity = (max pixel value - min pixel value) / (max pixel value + min pixel value)

with:

- max pixel value: maximum pixel value in the specification ROI;
- min pixel value: minimum pixel value in the specification ROI.

The image is filtered to reduce the influence of noise on the maximum and minimum pixel values. The specification ROI is a circle.

In addition, there is another procedure by means of a histogram (called Flood field uniformity at Philips):

- The average pixel values in the centre (C) and an artefact-free background (B) of ROIs at a reference position R are calculated (where R depends on the type of coil);
- The threshold T is selected as 10 * B;
- Depending on the coil, a histogram ROI has been defined in which the number of pixels N_{tot} greater than T is determined;
- For each grey value in the histogram ROI, the percentage ratio is calculated as the number of pixels with that grey value in the ROI divided by N_{tot};
- The final image uniformity is calculated using the percentage ratios.

Pitfalls

Especially with surface coils and when the phantom is close to the coil, this test is sensitive to phantom positioning.

A.2.3 Ghosting

Introduction

Ghosting refers to 'ghost images'; MRI signal in the image at locations from which it does not originate. These can be very clear copy images of the original image, but also signal with much less structure. Because this process already occurs in the raw complex data, the ghosting can be expressed in the

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standard magnitude MRI image as a higher or lower signal depending on the phase of the ghost signal with respect to the actual MRI signal at the location.

- Ghosting is measured in a projection line of the image beyond the phantom boundaries. Because of this, the ghost signal is always positive in the magnitude image.
- Ghosting in the phase encoding direction is, among other things, an indication of system instability in terms of both RF and gradients.
- Ghosting in the frequency encoding direction is an indication of problems with constancy of the frequency encoding gradient. So, it is a measure of short-term stability. The stability could have been be reduced by a poor eddy current compensation adjustment.
- Finally, specific ghosts can arise due to incorrect adjustment of the phase of the quadrature elements in the RF transmitter or receiver coils.
- Also, movement of ferromagnetic objects at the time of the MRI scan, such as an elevator near the magnet, can lead to ghosting in the phase encoding direction.

Because the ghosting level can be very sensitive to small deviations in the tuning of system components such as gradient timing, eddy-current compensation, and RF transmitter and receiver coils, this is an important test to notice quality degradation. Since ghosting in the phase encoding direction is due to instability across different excitations and ghosting in the frequency encoding direction is due to instability relative to adjacent data samples, in general ghosting in the phase encoding direction is greater than ghosting in the frequency encoding direction.

Clinical relevance

With ghosting MRI signal is present at locations from which it does not originate. Therefore, ghosting artefacts can influence the interpretation of images and/or diagnosis.

Goal

- This test checks a supplier's specification.

Method

Ghosting can be measured in the same recording as the SNR recording. Though a prerequisite is that the FoV of the recording is larger than the phantom, such that ROIs can be set outside the phantom. The intensity of the ghosting signal is measured in the (rectangular) ROIs to the left and right of the phantom and above and below the phantom to determine the individual ghosting levels in both frequency and phase encoding direction.

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ghosting signal [%] = 100% * | ghost signal - noise level| / (2 * average signal phantom),

with:

- average signal: signal in a large ROI in central part of the phantom;
- ghost signal: background intensity in the ROI next to the phantom in the frequency respectively phase encoding direction;
- noise level: intensity in the background outside the projection directions of the phantom, i.e. in a corner of the image.

This definition is taken from the ACR MRI Quality Control Manual [54].

Pitfalls

When determining the background noise, it is important to set the windowing in such a way that the background noise can be clearly distinguished from the black area outside the image.

A.2.4 Image artefacts

Introduction

Because artefacts may be caused by a variety of problems, including relatively common issues such as defects in an RF receiver coil, this is an important test to monitor changes in image quality. Especially in a constancy test, a reference acquired during acceptance testing provides a reference for future QC measurements.

Clinical relevance

Image artefacts can influence the interpretation of images which can lead to incorrect diagnosis and can negatively affect treatment accuracy.

Goals

- This test checks a supplier's specification;
- The test determines a baseline which can then be redetermined with a constancy test.

Method

The images need to be visually assessed for image artefacts. Some artefacts will also be visible in the SNR or ghosting measurements, depending on the type of artefact. For some artefacts it may be useful to determine the intensity of the artefact. This can be expressed as the intensity (artefact in the

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background) or the intensity deviation (artefact in the phantom) divided by twice the signal intensity in the phantom.

Percent artefact level = 100% *| artefact intensity - surrounding intensity | / (2 * average phantom signal),

with:

- average signal: signal in large ROI in central part of the phantom;
- artefact level: intensity in ROI where the artefact is;
- surrounding intensity: intensity around or nearby the artefact.

Pitfalls

Automatic analysis of these artefacts is difficult. Some image artefacts will lead to deviations in SNR, homogeneity or ghost measurement, while other artefacts are visually evident, but will not be reflected by these measurements.

When determining the background noise, it is important to set the windowing in such that the background noise can be clearly distinguished from any black (padded) area outside the image.

A.2.5 Gradient-related geometric distortion

Introduction

The measured diameter of the phantom in x, y and z directions gives an indication of the geometrical accuracy of the system; in particular the gradient amplifier output. For geometric accuracy, the spatial linearity of the gradient coil is also important; this is a design specification of the gradient coil.

Clinical relevance

This parameter is important for imaging for stereotactic applications and radiotherapy, where accurate positioning is crucial. In addition, this parameter is of importance in morphological scientific studies where small volumetric changes of anatomical structures are measured longitudinally to detect changes.

Goals

- This test checks a supplier's specification.

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Method

The phantom diameter is a simple measure to check for errors in the calibration of the gradient amplitude, resulting in a global scaling error (linear component). The diameter of a phantom can be determined in the three orientations, to test the three orthogonal gradient coils.

In order to assess local distortions from gradient non-linearities, a phantom with multiple markers as shown in Figure 7Figure 7 is required. If the markers are positioned in a single plane (2D), the phantom should be rotated in the magnet to assess local distortions on the third axis. For a 3D grid of markers, a 3D volume scan is generally recommended. [35]

According to the Philips quality procedure:

- For each marker in the image, the horizontal and vertical shift relative to the known position in the phantom is measured;
- The differential linearity of each pair of adjacent markers is determined;
- The size of the phantom in horizontal and vertical direction is determined.

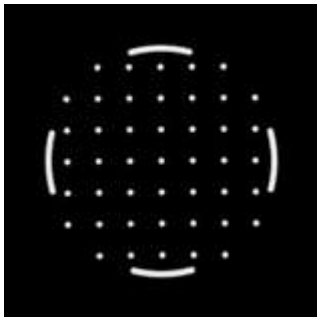

Figure 7: Part of the Philips PIQT phantom for determination of geometric accuracy.

Pitfalls

- When analysing the images on a system other than the scanner, it is important to check whether the pixel size of the scan is correctly incorporated in the measured distance;
- This test is sensitive to phantom positioning. Depending on the shape of the phantom, the result can be more or less sensitive to phantom positioning. The advantage of a spherical phantom is that it cannot be positioned incorrectly, on the other hand, the image plane needs to be exactly at the right level. A cylindric phantom has the advantage that if it is placed parallel to the B_0 magnetic field, there is less distortion due to inhomogeneity of the magnetic field. In

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our experience, in order to achieve sufficient accuracy, a cylindric phantom with a good phantom and/or RF coil holder is required.

A.2.6 Resonance frequency

Introduction

The resonance frequency of the signal is determined by the strength of the B_0-field; so, a drift in resonance frequency over time is therefore an indication of a drift in the B_0-field. A number of processes can cause the strength of the B_0-field to change:

- cryogen evaporation;
- thermal or mechanical changes that lead to changes in current density in the magnet;
- shim changes;
- energy loss due to movement of ferromagnetic objects in the vicinity of the magnet (long-term drift);
- large ferromagnetic objects in the vicinity of the MRI system.

Clinical relevance

A change in resonance frequency changes affect the sensitivity of the system. In case the resonance frequency drifts outside of the optimal range of the RF receiving coils, the SNR reduces. With short-term changes, during an MRI acquisition, it can lead to ghosting. The variation of the resonance frequency is not great in practice. A check of the resonance frequency does not require extra acquisition time and can easily be included in automatic analysis of QC results, since the resonance frequency is indicated in the DICOM header. However, in case of manual analysis of the QC results, the cost/benefit ratio is not advantageous.

Goals

- This test checks a supplier's specification;
- The test defines a baseline which can then be reevaluated over time as a constancy test.

Method

This parameter is determined by the system by default for each examination. Different measurement methods can be used, such as: the peak frequency from a spectral MRI measurement and frequency determination by means of minimal RF reflection at an RF transmitter frequency sweep. In the DICOM

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header the resonance frequency is indicated in field (0018,0084). This parameter can be determined from the SNR recording.

A.2.7 RF transmit amplitude

Introduction

The RF transmit amplitude is a measure for the stability of the RF transmit chain of the MRI scanner. The signal path roughly consists of small signal formation in the time domain, Digital Analog Converter (DAC), RF amplifier, cables, PIN diodes, and connections to the transmit RF coil. The RF transmit amplitude is generally shown on the scanner as the amplitude of the RF output to achieve a 90 degrees excitation angle (for a system-specific, fixed RF pulse duration). This parameter depends on the RF loading by the phantom. Changes of this parameter are initially compensated because the RF transmit amplitude is calibrated for each patient. When this calibration goes out of range of the RF transmit amplifier, this influences the effective excitation angle. This affects SNR (gradient echo techniques are more sensitive than SE techniques), but it may also affect slice thickness and position. In case the RF power amplifier contains a vacuum tube, it is expected to degrade over time. The amplitude requires recalibration over regular intervals. A rapid change of the RF transmit amplitude over time most likely indicates near end-of-life of the vacuum tube. Modern RF amplifiers built with semiconductors do not show this behavior anymore.

Clinical relevance

A large deviation of this parameter affects the SNR, which largely determines the detectability of small layer-contrast details. At the same time problems may occur, especially in heavy patients, and artefacts may be visible that do not reproduce on a phantom. Sudden large changes of the transmit amplitude are not expected, but because monitoring it is little effort on most systems, it is included in the constancy tests. For Siemens and GE systems the transmitter amplitude can be found in the DICOM header, and does not require extra acquisition time. For systems where this parameter cannot easily be monitored (e.g. Philips), monitoring of this parameter can be omitted, as in manual analysis the cost/benefit ratio is disadvantageous.

Goals

- The test is purely informative for the user.

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Method

The transmit amplitude is determined by the system by default for each examination. Usually, the reference amplitude is defined as the RF amplitude required for a 90° excitation hard pulse. Sometimes this value is mentioned in a private DICOM field, but it is not a mandatory DICOM parameter. It may be taken from the DICOM header of an SNR image. If the value is not stored in the DICOM header, it can often be found in the log file or in a user menu on the scanner.

A.2.8 Shim (B_0 homogeneity)

Introduction

Shimming the system is to optimise the homogeneity of the B_0 field, i.e. the uniformity of the B_0 field over the scanned volume. The homogeneity is usually expressed in parts per million (ppm) of the magnetic field within a volume (often a sphere, sometimes a cylinder).

System parameters that affect the shimming:

- accuracy of the B_0 field of the magnet;
- presence of larger ferromagnetic materials in the vicinity of the magnet, or small ferromagnetic materials in the magnet;
- effectiveness of active shim-coils or passive shim strips.

The determined value is shim Peak-to-Peak:

$$\text{Shim Peak-to-Peak [ppm]} = 10^6 * (\max B_0 - \min B_0) / \text{mean } B_0$$

This parameter can be strongly influenced by a local disturbance. The homogeneity of the shim expressed in a Root-Mean-Square (RMS) value of multiple measurements over a volume is less sensitive to this:

$$\text{Shim Root-Mean-Square [ppm]} = 10^6 * \sqrt{\sum_{x,y,z} B_0(x,y,z)^2 / \text{Volume}}$$

Specifications of MRI systems are usually RMS values expressed. However, for constancy measurements, the Peak-to-Peak value is sufficient; in addition, a relatively small phantom (compared to the image FoV) is measured, so that the requirement that the Peak-to-Peak variation needs to be small as well is realistic.

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Clinical relevance

When a system is not properly timed, geometric distortions can occur, image uniformity and SNR becomes worse with GE techniques, and fat suppression using spectral techniques becomes more difficult. MR spectroscopy is still the most sensitive technique for this; bad shim leads to widening of the spectral peaks.

For systems where this parameter is not easily self-monitored (e.g. Philips), the user needs to make a cost/benefit assessment to monitor this parameter. Or he can have this parameter monitored by the manufacturer.

Goals

- This test checks a supplier's specification.

Method

The homogeneity of the B_0 field, the shim, can be determined in different ways [73]:

a) a spoiled GE measurement with different echo times;
b) by an interference image of spin echo (SE) and stimulated echo (STE);
c) by line width of the total signal

With shim measurements, be aware that the scanner can perform an extra shim at the patient level, both the linear and sometimes a higher order shim. Depending on the manufacturer and the protocol, this may or may not be activated. If possible, it is recommended to turn off the shim parameter in the scanning protocol, but this is not possible on all systems.

- *a) A spoiled GE measurement with different echo times*

  With a spoiled GE scan, the signal phase depends on the local B_0 field. A map of the B_0-field can be made by making 2 scans with a different echo time in the order of 5 ms, with reconstruction of phase images. Based on the phase difference between the 2 images, a B_0-map can be calculated. For further explanation see AAPM report 34 of Task Group No 6 [73].

  The advantage is that a quantitative B_0 map is obtained, which can easily be analysed automatically. The disadvantage is that reconstruction of phase images is not available on all scanners. Furthermore, if any phase corrections have been applied during image reconstruction, the images may not be suitable for B_0-mapping.

- *b) Interference image of SE and STE*

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An image of the B_0 inhomogeneity can also be obtained by making a scan where the signal is a combination of the Spin Echo (SE) signal and the stimulated echo (STE) signal. Depending on the local field, these 2 signals are in phase and add up, or they are in counter-phase, causing signal cancellation. Due to variation in the B_0-field, interference lines appear in the image (see Figure 8). By varying the time between the RF pulses (TE is used for this), the sensitivity of the measurement can vary.

A measure of the variation of the B_0 field across the phantom is obtained by counting the number of white-white transitions in the image. The number of ppm's of one colour change (e.g. white to white) in the image stands for:

$$\delta B_{\text{white-white distance}} = \frac{1}{\gamma B_0 \text{TE}}$$

Here γ stands for the gyromagnetic ratio (MHz/T), B_0 for the strength of the magnetic field (Tesla), δB for the number of ppm's per line and TE for the echo time used in seconds. By taking different TEs, images with different line densities can be taken.

The advantage is that rapid visual assessment is possible. Disadvantage: automatic analysis is more difficult, and it requires a special pulse sequence that is not available on all MR systems.

Figure 8: Shim assessment using an interference image of SE and STE.

- *c) Line width of the total signal*

This method uses a spectroscopic measure of field homogeneity. It uses the FWHM (Full Width at Half Maximum) of a non-selective FID (Free Induction Decay). The obtained FID is valid for the entire volume of the phantom used, and thus corresponds more to the Shim RMS value than to the Shim Peak-to-Peak value.

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Pitfalls

- To make the measurement sensitive to B_0 variations of the system, it is important to disable the shim parameter of the scanning protocol so that no specific shim correction is performed for the scan, and to measure with the default shim settings.
- When analysing the B_0 map, the system can report a B_0 value that does not take phase wrap into account. In that case, the phase wrap needs to be manually added in the calculation of the Peak-to-Peak value.
- For a B_0-map based on 2 phase images with different echo times, it is desirable that the system does not apply phase correction per echo. Sometimes such phase correction, or echo shift correction, is implemented and cannot be turned off.

A.2.9 Couch positioning

Introduction

For MRI couch position the situation in the simulation phase and the treatment phase are different. At this moment no phantoms are available to measure the accuracy of the couch translation.

Clinical relevance

Couch position accuracies are important in two situations:

1 – the couch moves during the acquisition of the image; e.g. in multi-station imaging

2 – the couch is used to shift the patient to the correct position before treatment based on the online images.

Goals

- This test checks a supplier's specification.

Method

No independent method has been established, but an external laser measurement positioned at the feet and of the table can be performed to compare the requested and the performed couch translation. Also external lasers in combination with a MR compatible ruler can be used.

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A.2.10 Connectivity and orientation

Introduction

As MRI is used for treatment planning and position verification, it is important that scan orientation is properly handled. This is checked scanning an asymmetric phantom with various orientation scan settings: head first supine (HFS), feet first supine (FFS), head first prone (HFP) and feet first prone (FFP). An example phantom is shown in Figure 9.

Figure 9: Asymmetric phantom displaying patient orientation explicitly.

Clinical relevance

The scan orientation of the MRI is relevant for treatment planning and position verification. Improper handling of scan orientation may result in mis treatment.

Goals

- This test checks a supplier's specification;

Method

An asymmetric phantom is scanned with various settings of patient orientation (HFS, FFS, HFP and FFP). It helps to take a phantom with orientation explicitly discernible, e.g. the phantom in Figure 9. The phantom is put in desired orientation in the MR bore and the corresponding setting is selected at the scanner console:

- Check whether the patient orientation is properly given in the DICOM header of the scan;
- Load the image in a DICOM viewer and check whether the orientation of the phantom is properly depicted (e.g. left corresponding to left, feet to feet, anterior side with anterior). Ideally this DICOM viewer is the software package used in the radiotherapy workflow of your

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department (e.g. image registration package, delineation software package, treatment planning system);

- Measure the phantom size in the software package to check whether it corresponds to the physical size of the phantom.
- The scan orientation should be visible in the DICOM header and properly visualised in the DICOM viewer. The phantom size measured in the DICOM viewer needs to be within 1 mm of the physical size.

A.2.11 External laser position

Introduction

External laser position can be used for absolute positioning of the patient or phantoms. Quality measures are meant to check the consistency between laser alignment and scanner isocenter.

Clinical relevance

The relevance depends on the situation.

1 - In an MR-only workflow: rotation of the horizontal lasers will result in patient rotation on conventional linacs if tattooing is based on the external lasers. This cannot be corrected for on a conventional linac couch.

2 – In an online MR guided workflow using table translations to correct the patient position: incorrect positioning of the patient will give a displaced dose delivery.

Goals

This test checks a supplier's specification.

Method

External Laser Positioning System (ELPS) QA test is performed, which consists of an acquisition protocol and laser alignment phantom. This phantom has external markings to align the phantom with the lasers. The crosshairs in the sagittal and coronal image plane are used to determine the offset of the lasers.

Pitfalls

There will always be dependency on the precision of the couch motion. Lasers are tested using a phantom that is set-up on the couch outside of the MRI bore followed by a couch translation into the

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isocentre of the MRI. Improper couch translation, couch rotation during shift, or incorrect predefined travel distance to the isocentre will all result in deviations of the centre and other markers in the image of the phantom.

Another pitfall is that the phantom is light weighted. To mimic the travel properties to clinical situation it is advised to put extra weight on the table.

A.2.12 Synthetic CT generation for dose calculation

Introduction

For the synthetic CT (sCT) generation, dedicated phantoms have not become available yet. For the commissioning, both a CT and a sCT have to be generated and compared. The workflow described below, is what is in generally used at this moment when commissioning sCT implementation in the clinic. If imaging, hardware and sCT generation algorithm do not change, periodic maintenance is not necessary. If changes have been applied, part of the checks have to performed again. In case of new software release, a comparison (image similarity and dose) of the sCTs generated with the old and new release might be the easiest test.

As the field has not yet established a consensus on the sCT QA, the tests are briefly outlined.

Clinical relevance

The dose calculation is dependent on the electron density (ED) that are retrieved from the Hounsfield Unit (HU) of the sCT. Deviations of the electron densities could result in an incorrect dose.

Goals

This test checks the specifications set in this guideline.

Method

First, the HU of the sCT and the conventional CT are compared using image similarities for group of patients. Subsequently, the dose as calculated on both CTs is compared voxel-by-voxel using the gamma index.

Pitfalls

For the dose calculation, the accurate conversion table from HU to ED (HURED table) has to be used.

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Since the CT and the MR sequence used for sCT generation are not acquired at the same time, differences in patient anatomy may affect the dosimetric comparison. For instance, differences in anatomy during imaging (e.g. air in rectum) could give small deviations in the dose.
More recent deep-learning based methods may yield different types of errors than more traditional methods based on bulk density assignments.

A.2.13 Testing the use of the synthetic CT as reference for position verification

Introduction

When no CT is used in the treatment chain, the synthetic CT (sCT), or MRI directly, is required to be suitable for position verification. When introducing MR-only for a tumour site or specific position verification protocol (e.g. bony anatomy match, MRI to CBCT match), it is advised to add the MR-only workflow to the existing work flow for the first group of patients (e.g. adding an sCT acquisition to the existing clinical protocol) and simulate the MR-only workflow for these patients next to their original clinical workup.

Clinical relevance

The use of the sCT, having benefits of not needing a registration between planning CT and MRI, should not introduce other uncertainties in the radiotherapy workflow, most notably in the position verification procedure.

Goals

This test checks the specifications set in this guideline.

Method

For a group of patients (e.g. ten), acquire both the CT-sim from the standard clinical workflow and the sCT as additional imaging. In an offline setting, perform the position verification with daily imaging (e.g. MRI, CBCT, …) which is used in the clinical workflow, but now based on the sCT as a reference instead of the CT-sim. Perform two comparisons [28]:

- Compare the variation of the registration results (translations, rotations) of both workflows. These need to be in the same order of magnitude.
- Evaluate the difference of registration results. Per treatment session, subtract the registration results based on the CT-sim reference from the registration result based on the sCT. It is

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desirable that the population mean of this difference is close to zero. The variation of the registration difference needs to be in the order of the differences due to organ deformation between the CT-sim and the sCT.

Pitfalls

The CT-sim and the sCT are made on a different time point and anatomical changes will occur. This means that the registration results are not entirely comparable.

A.2.14 MR-MV coincidence

Introduction

For accurate image guidance it is important that the MR isocentre coincides with the MV isocentre. On both the MRIdian and the Unity the small mechanical offset, as determined during installation, is entered into the system such that the software can correct for this. After commissioning it is important to monitor that the offset (after software correction) remains close to zero. If the offset does not pass the critical action level, a re-calibration of the offset value needs to be performed.

Clinical relevance

Any discrepancies will result in a population-wide systematic geometric error. It is therefore crucial to accurately determine the offset between the imaging (MRI) and treatment (MV) isocentres.

Goals

- This test checks a supplier's specification.

Method

On the Unity system, which has an onboard MV imager (EPID panel), a phantom with MR-visible solution and ceramic ball bearings is used to determine both the MRI and MV isocentre in a single phantom setup. On the MRIdian, no EPID panel is present. For this reason, the MR-MV coincidence test is performed on a phantom, which contains an MR-visible solution and a radiographic film. At the time of writing, no third-party MR-MV phantoms, with corresponding analysis software, exist. It is therefore advised to start with the procedure as specified by the MR-linac vendor. Once a baseline on the vendor-specified procedure is established, one could choose to refine the procedure if deemed necessary.

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Pitfalls

- No independent, third-party, phantoms currently exist. For an independent check one would have to rely on an in-house developed measurement procedure.
- Make sure the imaging volume is adequately shimmed and a high readout bandwidth is used to ensure the geometric distortion of the MR image is minimized.

A.2.15 B_0 direction

Introduction

The direction of the main magnetic field determines the direction of the Lorentz force on the secondary electrons (i.e. the electron return effect or ERE) and should therefore be checked.

Clinical relevance

An incorrectly defined B_0 direction would result in an incorrect calculation of the deposited dose as the calculated dose kernel would be skewed in the opposite direction compared to reality.

Goals

- This test checks the manufacturer's configuration.

Method

The magnet field direction can simply be determined using a standard (MR-safe) compass. The compass is simply placed inside the bore to determine the polarity. An alternative method to check the B_0 direction is by placing a direct current onto a loop wire inside the bore (i.e. a conducting loop attached to a low voltage battery) as described in [74] (Appendix A). The Lorentz force that is applied on the current-carrying wire is directly visualised. Here, $\vec{F} = q\vec{v} * \vec{B}$, where the vector $\vec{F}$, the mechanic force, is the literal one, $q\vec{v}$, is the vector of the positive moving charge, and the vector $\vec{B}$ the magnetic field. The direction of the main magnetic field can be deduced by applying the right-hand rule taking into account the negative charge of the electrons. The direction of the magnetic field is reported according to IEC61712 and should agree with the treatment planning system.

Pitfalls

It is important that the compass is placed *inside* the bore. A measurement near the edge, but slightly outside the bore, may be corrupted by the influence of the shim coils, which have opposed polarity to minimise the magnetic stray field.

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A.2.16 Gantry-dependent B_0 homogeneity and MR isocentre shift

Introduction

Both the ViewRay MRIdian and the Elekta Unity are designed with a ring gantry, which holds all the beam generating components. For the MRIdian the gantry is positioned just around the gradient coils, while for the Unity system the ring gantry is positioned slightly further away, around the cryostat. Because the gantry contains large amounts of ferromagnetic material, the gantry can introduce spatially varying offsets to the B_0 field, or eddy currents which vary per gantry angle. It is shown that this leads to gantry angle dependent B_0 fields and even a small shift in the magnetic isocentre.

Clinical relevance

Gantry dependent isocentre shifts can affect the effective gating window. Changes in B_0 homogeneity may affect sequences that are sensitive to B_0 fluctuations, such as DWI or quantitative MRI.

Goals

- This test checks the specifications set in this guideline.

Method

Depending on the phantom that is used, these acquisitions may be performed with either a receive array (e.g. Torso coil) or the transmit receive (TX) body coil.

- *B_0 homogeneity test*

  A large diameter phantom with flood field is positioned in transverse orientation at isocentre. It is advised to acquire B_0 field maps at increasing gantry angles with 30° intervals. If the B_0 field mapping option is not available, one could a) acquire two phase images acquired at different echo time and calculate a B_0 field offline [75], or b) use an alternative method like ring counting on a spin echo stimulated - echo (SE-STE) interference image (see Figure 8)[1].

[1] When an SE-STE interference image is used, a peak-to-peak value over all gantry angles cannot be determined. In that case, the B_0 homogeneity of each gantry angle would be assessed individually.

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Per gantry angle two transverse scans are acquired. One scan with and one scan without electronic shimming. The scans without shimming are used to quantify the effect of the gantry. The scans with shimming are acquired to test whether standard shimming mitigates the field offsets effectively. Make sure to verify whether the vendor performs a reshim for each gantry angle in clinical mode to determine which acquisition corresponds to the clinical situation. Analysis: the mean B_0 map over all shimmed measurements is subtracted from each individual image to remove the static B_0 contribution. The peak-to-peak values are calculated from each difference image and are reported in nT as a quantitative metric for the added contribution of the gantry.

- *MR isocentre shift*

For this test any rigid phantom, which allows repeated images to be registered to one another suffices. Similar to the B_0 homogeneity test, the phantom is placed at isocentre and images are acquired or gantry angles with 30° intervals. Per gantry angle a single transverse magnitude image is acquired.

Analysis: the images are registered to the image that is acquired at the reference angle (i.e. the gantry angle at which the daily MRI is acquired during clinical operation). More details on this test can be found in [76].

Pitfalls

It is important that the readout bandwidth in this test is matched to the readout bandwidth of the clinical MRI sequences.

A.2.17 Linac-induced RF interference

Introduction

The linac subsystem contains a lot of electronics, in particular the magnetron, that produce RF signals that interfere with the MRI acquisition when not shielded properly. This test is indicative of a potential fault in the RF shielding by the Faraday cage.

Clinical relevance

RF interference may introduce image artefacts that obscure anatomical details or affect the tracking algorithm during irradiation.

Goals

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- The test is purely informative for the user.

Method

Any image quality phantom may be used for this test. The phantom is scanned multiple times using a clinical imaging protocol (e.g. a single 2D cine image is sufficient, as the analysis is relative to the baseline image when the linac is turned off), with different stages of linac operation. For example:

- LINAC OFF, *baseline image*
- Magnetron ON, but no radiation *to test the influence of the magnetron only*
- LINAC OFF, but moving MLCs *to test the influence of the MLC motors*
- LINAC ON (smallest field size) *to test the influence in full operational mode*

Analysis: the amount of noise needs to be similar between all the images that are acquired.

In addition to the test described above, noise only images may be acquired during different levels of operation. Both the MRIdian as well as the Unity allow the acquisition of noise-only images in service mode. It is advised to perform this test with your service engineer.

A.2.18 Radiation-induced RF artefacts

Introduction

In addition to the introduction of spurious RF signals, the radiation itself may produce artefacts, if the electronics in the RF receive chain are irradiated during image acquisition. Radiation induced artefacts may introduce noise-like image artefacts that reduce the SNR and thereby the image quality during irradiation.

Goals

- The test is purely informative for the user.

Method

Any standard imaging phantom can be used for this test. The phantom is positioned at isocentre with standard receive array setup. Similar to the previous test, multiple scans are acquired with different forms of linac operation. While test A.2.17 focusses on the effect of the magnetron, this test investigates the image quality when the phantom (and, more importantly the receive array) are

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irradiated with different field sizes. This test is therefore an extension of the previous test and can easily be combined into a single experiment. Like the previous test, the analysis is performed on magnitude images of a clinical scan protocol. For a more quantitative analysis noise only images can be acquired in which elevated levels of noise are more easily observed (see [74] for example images) With these settings, the last dynamic in a dynamic series will be acquired without RF and gradients (i.e. noise scan). The exam card contains a set of time-series measurements, which are ran with and without radiation. The image stacks do not need to be adjusted (acquired with no offsets). All measurements are performed at gantry angle 0°. Make sure to set up the treatment field with enough MU to irradiate for one minute. Start the scan immediately after beam-on.

Any image quality phantom may be used for this test. The phantom is scanned multiple times, at different levels of linac operation. For example:

- LINAC OFF, *baseline image*
- LINAC ON (10x10 cm field size) *to assess the image during typical clinical field sizes*
- LINAC ON (largest field size) *to assess the image during worst case scenario*

Analysis: the amount of noise needs to be similar between all the images that are acquired. The images are required to be free of artefacts.

A.2.19 Temporal stability test

Introduction

During long measurements, such as cine during treatment, temporal drift can cause geometric offsets. 2D cine acquisitions that are used for motion monitoring during irradiation are often acquired with balanced Steady-State Free-Precession (bSSFP). The typical banding artefact of bSSPF, which are often observed at the edge of the FoV, may move across the image due to B_0 drift. This may impact the tracking algorithm.

Clinical relevance

Geometric drifts can cause a disposition of the effective gating window. Banding artefacts that move across the image may cause the tracking algorithm to break depending on the implementation.

Goals

- This test checks the specifications set in this guideline.

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Method

To test the geometric drifts, any phantom containing a homogeneous section (e.g. the ACR phantom or bottle phantom) can be used. The phantom is placed at isocenter. A clinical 2D time-series (cine) acquisition is acquired for a duration that is representative of a treatment delivery time (e.g. 10 minutes) using standard receive array setup. The images are exported offline and analysed by registering the time-series images to each other. Examples of open-source image registration tools are: FSL [77], SPM [78], ImageJ [79], and Elastix [80]. Inspection can either be done offline or at the MRI console.

Pitfalls

Image registration of time-series is currently not offered by the vendors. The registration tool for this analysis needs to be validated first if an open-source registration tool is used.

A.2.20 Real-time feedback latency

Introduction

For systems that perform active gating during irradiation it is important that the latency of the real-time control system is known and within specification. Latency is defined as the time required by the control system to turn the beam off (or on) when the target moves beyond a preset boundary.

Clinical relevance

The latency affects the efficacy of the gating procedure. A high latency results in a slow response of the machine to the displacement of the target. A high latency requires larger PTV margins to mitigate the lag in the system.

Goals

- This test checks a supplier's specification.

Method

This test requires an MR-safe motion phantom and access to the gating trigger signal of the linac. The motion phantom is programmed to perform a set motion trajectory (e.g. a sinusoid or a cos6 waveform that resembles respiratory motion). The phantom is set up in such a way that an MR-visible marker

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will move in and out of the gating window. Both the actual position of the phantom and the trigger signal are simultaneously recorded. The time between the moment when the phantom actually crossed the gating boundary and the moment that the trigger signal is given, is defined as the latency. By comparing the two, the latency can be determined for each respiratory cycle. It is advised to run a series of gating cycles (e.g. > 20 cycles) and determine the average latency.