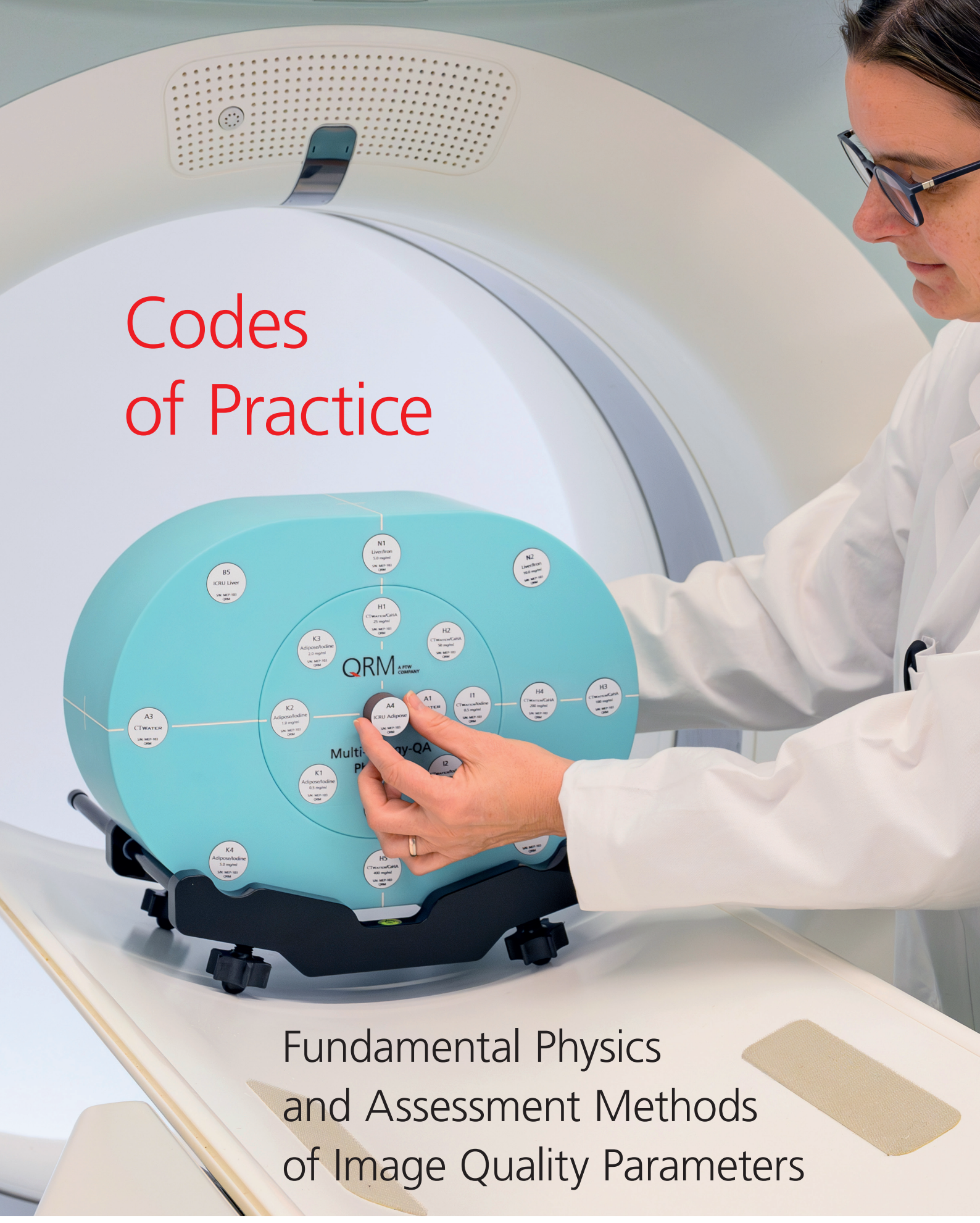
Codes of Practice



Fundamental Physics and Assessment Methods of Image Quality Parameters

QRM
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QRM - A PTW Company

A milestone in the history of both companies

In 1922, PTW started being a pioneer in medical radiation measurements. Since then, PTW has always been and will continue to be at the forefront of advancing patient safety through innovation and cutting-edge measurement technology.
In 1994, QRM (Quality Assurance in Radiology and Medicine) was founded by Professor Willi A. Kalender as a university spin-off of the Institute of Medical Physics in Erlangen, Germany, for the development, design, construction, and production of phantoms for quality assurance in X-ray imaging and associated procedures.
Since April 2020 QRM is a subsidiary of PTW Freiburg GmbH.

Phantoms for your needs

We look back on more than 25 years of experience in designing and manufacturing phantoms for medical imaging modalities. Our first products were designed for lung imaging in Computed Tomography (CT) and bone densitometry in DXA and CT. Since starting manufacturing phantoms in the early '90s, our range of phantoms has increased substantially over the years. Today, we offer products for different imaging modalities in clinical routine, research, development, and science, as well as customized and OEM products.

Our engineers design dedicated phantoms for many applications in the field of medical imaging. Our customers are physicians, scientists, and manufacturers all over the world. Our main focus is on developing products for diagnostic X-ray, CT, and micro-CT imaging. We offer phantoms to analyze image quality (IQ), calibrate HU-levels as well as for dosimetry issues.

QRM phantoms are professionally designed and manufactured with highest precision. The components are carefully selected and adapted to the specific imaging modality. Our machinery uses the latest technology to produce and manufacture phantoms with high accuracy.

Standard as well as custom-made phantoms undergo a quality control test according to their use. Therefore, different types of medical imaging devices (e.g. DXA, CT, C-arm, Micro-CT, MRI, etc.) are used for our test setups.

Our core competence is the development and production of customized phantoms in cooperation with our customers. We successfully collaborate with manufacturers in medical and industrial X-ray markets as well as with scientists and physicians working on research projects and studies worldwide. All standard phantoms can be modified according to your needs. We also offer customized phantoms for: PET, SPECT, radiation therapy, and for other modalities.

Are you interested in a completely new, specially designed phantom for a specific study, project, or new application?

Contact us - we provide phantoms for your needs.

Codes of Practice

1 Introduction

This chapter aims to briefly explain the physics and the assessment methods of image quality parameters for multiple QRM-phantoms dedicated to various X-ray applications. For a more detailed description we recommend relevant literature, e.g. (book) '*Computed Tomography – Fundamentals, System Technology, Image Quality, Applications – Willi A. Kalender, Publicis 2011*'

Although this chapter provides the reader with a concise overview for the evaluation of the X-ray system's imaging performance, it should be noted that **QRM phantoms are NOT registered as medical devices**. Hence, the procedures outlined in this chapter are recommended only for research and development purposes and do not replace relevant protocols and publications and shall not be used for the acceptance tests and constancy tests of X-ray imaging devices used for the diagnosis or treatment of patients.

Safety warnings and instructions
Please read these safety warnings and instructions carefully before use and keep them well for your future reference.

1. Do not place the phantoms in direct sunlight or near chemicals. Make sure that the environment does not experience abrupt changes in temperature or humidity.
2. Do not put the product on a cart, table, or desk, which is not stable, to avoid falling off.
3. Do not clean the phantom with solvents. Just use a mild soap and rinse with clear water.
4. Transportation is only recommended if a transport case is available.

2 CT Image Quality

Image quality parameters are of utmost importance for evaluating the performance of imaging systems and are affected by different factors. The physics and the assessment techniques of the common CT image quality checks defined in the literature, guidelines, and reports are briefly described below. The models name of the appropriate QRM phantoms for which these tests can be performed are indicated in each section.

1. Spatial Resolution and Modulation Transfer Function (MTF)

The MTF describes the capability of an imaging system to resolve fine structures, i.e. its ability to show small details. It is specifically defined for high-contrast structures. Basically, the measurements are carried out at high dose levels, such that noise in the image can be eliminated, allowing for low-noise assessments. Here, one has to evaluate spatial resolution in the axial plane and in the longitudinal direction (z-axis) as both quantities are affected by different quantities. The spatial resolution in the axial plane depends on the number of projections, reconstruction filters, pixel size, focal spot size, and detector size whereas the spatial resolution in the longitudinal direction depends on slice thickness, focal spot size and noise.

The spatial resolution can be measured either directly by visually determining the smallest resolvable pattern/structure, or indirectly by calculating the point spread function (PSF) and modulation transfer function (MTF). The PSF is a common metric used for the indirect measurement of spatial resolution. This two-dimensional (2D) mathematical function is the response of the imaging system to a point source input. While the PSF is defined in the spatial domain, the MTF is a way to quantify spatial resolution in the spatial frequency domain. Mathematically, the MTF is the Fourier Transform of the PSF and has the unit line pairs per cm (lp/cm). When resolving fine structures there are several blurring mechanisms (not discussed in detail here) which cause a loss of contrast in the image. Generally, the contrast reduces with decreasing structure size. The highest spatial resolution that can be achieved for a given system is often specified/indicated by the 10% value of the MTF, i.e. the spatial frequency, where the achieved contrast has dropped to 10% of the maximum value. This value serves as an objective quantity and can be used to compare the spatial resolution of different imaging modes as well as comparing the performance of different imaging systems.

For the acquisition of the spatial resolution using the PSF and MTF as described above, phantoms with thin wires are commonly used.
The spatial resolution along the longitudinal direction (z-axis) can be investigated using slice sensitivity profiles employing phantoms with thin high absorbing metal inserts.
Alternatively/Similarly, instead of using the PSF, the 3D MTF can also be calculated using the edge spread function (ESF) obtained from measurements of phantoms including sharp edges or spheres. Phantoms including such inserts hence allow to assess spatial resolution in all planes by evaluating the 3D MTF.

Visual spatial resolution evaluation with QRM Phantoms

- Use thin slices (approx. 1 mm slice thickness) and a suitable kernel, e.g. `standard` for regular scan protocols and `hi-res` for high-resolution scan protocols.
- Several line patterns shall be seen with distinguishable bright bars and dark spacing between the bars. Read out the corresponding spatial resolution (in lp/cm) of the smallest resolvable structure from the phantom datasheet.
- The smallest visible pattern determines the highest in-plane spatial resolution.

Example:
30 lp/cm corresponds to $\frac{1\ \text{cm}}{2 \cdot 30} = 0.016\ \text{cm}$

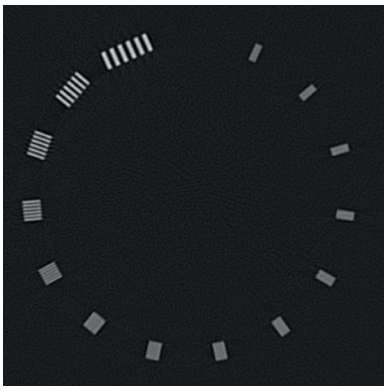

Figure 1: Cone-Beam Phantom CT Image (Test pattern section)

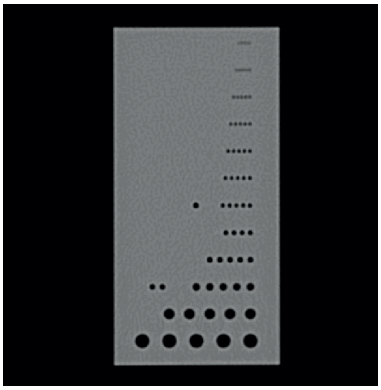

Figure 2: 3D Spatial-Resolution Phantom CT Image (Hole pattern section)

Suitable Phantoms
- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic
- QRM-10131 Dental CBCT QA Phantom, Expert
- QRM-10111 3D Spatial Resolution Phantom
- QRM-10101 3D Spatial Resolution Phantom, D100
- QRM-10140 High Contrast Resolution Phantom, D100

MTF evaluation with QRM Phantoms
- Use thin slices (approx. 1 mm slice thickness) and a suitable kernel (e.g., `standard` for regular scan protocols and `hi-res` for high-resolution scan protocols). Further, use the maximum available dose to reduce noise.
- The ESF or PSF should be measured to calculate the MTF of the imaging system. Typical values for describing the performance of scanners are 10% and 50% of the maximum value of the MFT.

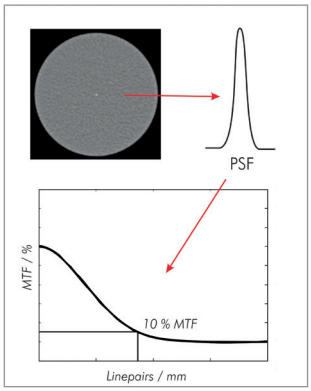

Figure 3: Wire Phantom CT Image - Resin, PSF and MTF

Suitable Phantoms
- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic
- QRM-10131 Dental CBCT QA Phantom, Expert
- QRM-10130 Dental CBCT QA Phantom, Basic
- QRM-10136 Breast CT QA Phantom, Expert
- QRM-10104 Wire Phantom, resin
- QRM-10105 Wire Phantom, D100
- QRM-10138 Wire Phantom, air
- QRM-10114 Slice Sensitivity Phantom

2. Low-Contrast Resolution and Contrast-to-Noise Ratio (CNR)

The low-contrast resolution describes how well an imaging system can differentiate adjacent tissues/materials that have similar X-ray attenuation. The capability to resolve low contrast features primarily depends on the noise level in the image. A high level of noise prevents two objects from being distinguishable resulting in poor contrast differentiation. The main X-ray-tissue interactions are mainly Photoelectric effect and Compton interactions, except for high-contrast structures such as bone. Primary photons lose their energy through scattering or absorption as they pass through tissue. Differences in image contrast are the result of atomic number, electron density, physical density, or differential attenuation of X-rays through tissues of different thicknesses. In imaging systems, windowing allows image contrasts to be visualized in grayscale. Low-contrast resolution is affected by different parameters such as tube current, tube voltage, beam energy, and contrast medium.

One important quantity to describe the low contrast detectability is the contrast-to-noise ratio (CNR). The CNR depends on the differences in the mean HU values, $CT\#_{\text{contrast insert}}$ and $CT\#_{\text{background}}$, and the background signal noise, $\sigma_{\text{background}}$:

$$\text{CNR} = \frac{CT\#_{\text{contrast insert}} - CT\#_{\text{background}}}{\sigma_{\text{background}}}$$

There are two implications of the CNR:
1. For features/structures of same size: the higher the CNR the lower the discernable contrast.
2. For a given/fixed contrast value of different sized features: the higher the CNR, the smaller the detectable structures.

The CNR can be assessed with phantoms containing several spherical (3D measurements) or cylindrical low contrast inserts (2D measurements) which provide different contrast values and insert diameters.

Evaluation of the low-contrast resolution with QRM Phantoms

We describe the assessment/evaluation of the low-contrast resolution here using the QRM Cone Beam Phantom (Low Contrast Section).
For the assessment of the CNRs of different contrast values provided in the phantom only the largest inserts (seen as circles/disks) should be considered. Smaller inserts should only be evaluated according to their visibility. Regions of Interest (ROIs) smaller than the diameter of the insert should be used to exclude cupping/edge effects from the analysis.

- Draw circular ROIs of appropriate size and place them within the large inserts.
- For reference, place similar ROIs in the proximity of the inserts, but at some distance from the phantom edge.
- Read out mean CT values within the ROIs and the signal noise (standard deviation of the fluctuating CT values) of the background and calculate the CNR.

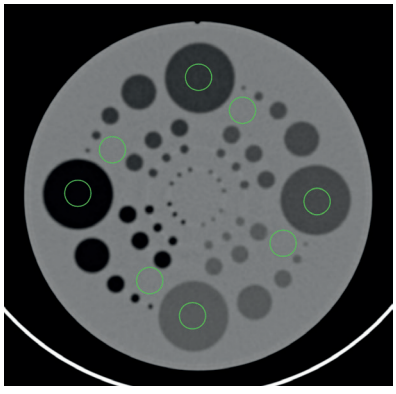

Figure 4: Cone Beam Phantom Low-Contrast section CT Image

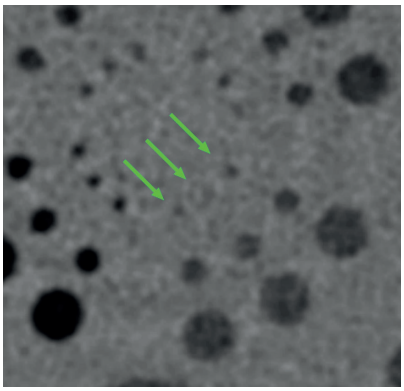

Figure 5: Detection of small low-contrast targets (arrows)

Suitable Phantoms

- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic
- QRM-10112 2D Low Contrast Phantom
- QRM-10100 2D Medium Contrast Phantom
- QRM-10109 3D Low Contrast Phantom (-10 HU)
- QRM-10110 3D Low Contrast Phantom (-20 HU)

3. CT Value Accuracy (Linearity)

After collecting/obtaining the projection data the 3D image matrix is calculated/reconstructed. The physical quantity affecting the intensity of the projection data and hence, creating the image contrast is the linear attenuation coefficient, $\mu_{material}$, of the irradiated material or patient. However, $\mu_{material}$ is an energy- dependent quantity and, hence, the intensity of the projection data depends not only on the material composition, but also on multiple other parameters as X-ray spectrum (i.e. tube voltage), filtration, detector configuration, beam hardening etc. Further, the image contrast is affected by the data processing, i.e. reconstruction kernel, pre-filtering etc. In order to ensure the comparability between different scanners and protocols in the clinical routine, CT values (grey values, CT#) are commonly quantified in Hounsfield Units (HU). The Hounsfield scale is defined by setting the grey value of water to 0 HU and the grey value of air to -1,000 HU, regardless of the used tube voltage or scan protocol. The CT value of an arbitrary material is calculated as follows:

$$CT\# = \frac{\mu_{material} - \mu_{water}}{\mu_{water}} \cdot 1000\ HU$$

The HU-scale is used in all clinical applications and guarantees comparable results regardless of tube voltage, filtration, CT manufacturer etc.

Typical CT values are -300 HU to 100 HU for soft tissues, 300 HU to 2000 HU and higher for compact bone, and 100 HU to 600 HU for iodinated contrast media.

The stability/constancy of the CT values should be monitored regularly as part of the quality assurance. Consecutively acquired values should not deviate more than ±4 HU of the mean value.

Homogeneous phantoms containing different materials (e.g. bone, water, air) are standard tools for this purpose. CT value linearity should be assessed separately for each protocol used in the clinical routine.

CT Value Accuracy (Linearity) evaluation with QRM Phantoms

Thick slices (approx. 5 mm slice thickness) and a soft kernel shall be used to reduce the image noise. In order to exclude cupping/edge effects (in particular in the highly absorbing bone insert) ROIs smaller than the inserts' diameter should be used to evaluate the HU values. Within this limit, the ROIs should be chosen as large as possible to average over many image pixels.

- Draw circular ROIs of appropriate size and place them in the scaling inserts
- Read out mean CT values within the ROIs and compare them to data acquired in prior quality assurance checks/ measurements.

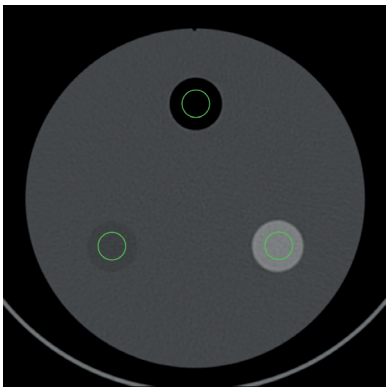

Figure 6: CT Image showing CT Value analysis

Suitable Phantoms

- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic
- QRM-10131 Dental CBCT QA Phantom, Expert
- QRM-10130 Dental CBCT QA Phantom, Basic
- QRM-10136 Breast CT QA Phantom, Expert

4. Signal-to-Noise Ratio (SNR)

In imaging systems, there are several sources of noise as, for example, thermal noise in the detector or scattered photons that do not pass straight through the tissue due to the Compton effect, but still contribute to the signal.

The Signal-to-Noise Ratio (SNR) is a good metric to quantify the image noise and hence, is particularly useful to specify/assess the system's performance. It can be calculated by comparing the mean signal amplitude, (*CT#*) (mean CT value), to the image noise, $\sigma_{background}$ (standard deviation of the fluctuating CT values) in a specific ROI:

$$SNR = \frac{(CT\#)}{\sigma_{background}}$$

The higher the SNR, the lower the noise in the image. The SNR is directly related to the (low) contrast separation: the higher the SNR, the better the low contrast separation (see section 2.2)
The SNR is affected by various parameters. Generally, the SNR is scales with the number of X-ray photons contributing to the image. Therefore, a high dose favors a high value of the SNR:

$$SNR \ \alpha \ \sqrt{\text{Radiation Dose}}$$

Further, high tube current (mAs), high tube voltage (kV), increased slice thickness or voxel size facilitate a high SNR while large patients or strong absorbing materials lead to a reduced SNR.

Also the reconstruction kernel in imaging systems affects the SNR and hence, the contrast separation. Bone filters provide a low SNR while soft tissue filters provide a high SNR. For this reason, smooth filters are used in soft tissue images, while sharp filters are used in imaging high-contrast structures such as bone.

SNR evaluation with QRM Phantoms
For an evaluation of the SNR uniform phantoms are used.
- Place the homogeneous section of the phantom at the isocenter and scan the phantom with the scan protocol for which the SNR should be evaluated using a slice thickness of at least 2 mm.
- Draw a circular ROI of sufficient size (e.g. 30 % of the phantom diameter) in the center of the phantom and examine the mean CT value and the noise level (standard deviation of CT values within the ROI).
- If necessary, repeat the measurement at several positions within the homogeneous phantom, with different ROIs or with different scan protocols, filters, reconstruction kernels etc.

Suitable Phantoms
QRM-10103 Cone-Beam Phantom, Expert
QRM-10120 Cone-Beam Phantom, Basic
QRM-10131 Dental CBCT QA Phantom, Expert
QRM-10130 Dental CBCT QA Phantom, Basic
QRM-10136 Breast CT QA Phantom, Expert

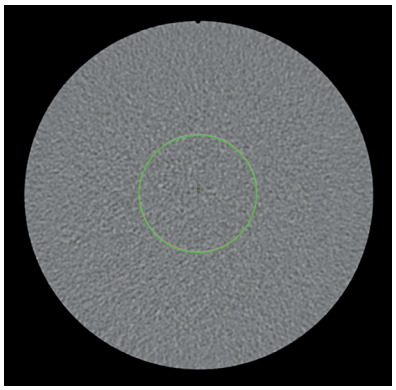

Figure 7: CT Image showing SNR analysis

5. CT Value Uniformity (Homogeneity)

Uniformity is an important image quality parameter that determines how uniform the CT values of a homogeneous test object are depicted in an image.Due to artifacts such as beam hardening, cupping artifacts or detector-specific issues that often arise in clinical practice, uniformity should be inspected regularly to ensure that the CT value deviation between the center and periphery of the image is not higher than the tolerances. The tolerance value is defined according to international and national guidelines, and the difference in the mean CT values between the center and the periphery of the image should not exceed ± 4 HU. The deviation from the baseline values should not exceed ± 2 HU.

CT Value Uniformity (Homogeneity) evaluation with QRM Phantoms

- Place the homogeneous phantom at the isocenter and scan the phantom with the largest dose (mAs) and most frequently used tube voltage. Use relative thick slices of at least 2 mm thickness.
- Insert several ROIs (central, top, right, bottom and left) of roughly 20 % of the phantom diameter in the axial image of the homogeneous phantom and examine the mean CT values to assess the homogeneity of the CT values within the image.
- Avoid placing the peripheral ROIs too close to the edge of the phantom.
- Evaluate how much the mean CT values of the peripheral ROIs deviate from the value at the center.
- If necessary, repeat the analysis using different ROI sizes and locations, as well as slice thicknesses, scan protocols and reconstruction kernels.

Suitable Phantoms

- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic
- QRM-10131 Dental CBCT QA Phantom, Expert
- QRM-10130 Dental CBCT QA Phantom, Basic
- QRM-10136 Breast CT QA Phantom, Expert

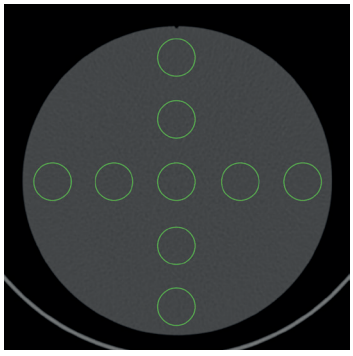

Figure 8: CT Image showing homogeneity analysis

6. Geometric Accuracy

In radiotherapy or interventional applications, treatments are based on the accuracy of the patient's anatomy obtained by CBCT, i.e. sensitively depend on the exact localization/spatial positions of the internal structures. The treatment isocenter and the imaging isocenter must match each other, and the reproducibility of the spatial positioning of movable internal structures should be checked regularly. Different parameters such as reconstruction algorithms and spatial resolution affect the geometric accuracy. While monthly geometric accuracy tests are recommended in Radiotherapy Units, annual controls are sufficient for dental and interventional CBCT.

Geometric accuracy evaluation with QRM Phantoms

- Use thin slices (approx. 1 mm thickness) and a suitable reconstruction kernel, e.g. a regularly/frequently used scan protocol and high-resolution kernel.
- Measure the dimensions of the holes at each position using a ruler or equivalent tool of your DICOM Viewer to determine a distortion of the hole matrix. An overestimation of hole size depends on scanner settings and beam hardening in the image. The holes may appear larger than they are. Measure the positions of the holes within the regular hole grid to uncover a distortion of the hole matrix in the periphery.

Suitable Phantoms

- QRM-10103 Cone-Beam Phantom, Expert
- QRM-10120 Cone-Beam Phantom, Basic

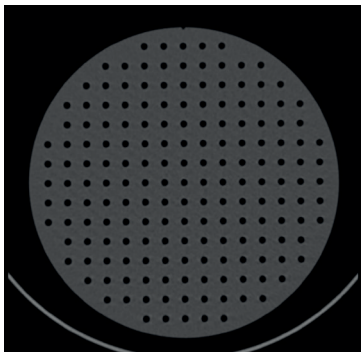

Figure 9: CT Image showing image distortion analysis

3 CT Dose Index (CTDI)

The Computed Tomography Dose Index ($CTDI_{100}$) indicates the air-kerma readings in mGy across a 100 mm long calibrated pencil-shaped ionization chamber in CT systems. The $CTDI_{100}$ is defined as the integral of the dose profile (z axis) along a line perpendicular to the tomographic plane (x-y plane).

$$CTDI_{100} = \frac{1}{N \cdot T} \int_{-50\,mm}^{+50\,mm} D(z)\, dz$$

D (z): dose profile along the axis perpendicular to the axial/scan plane (z-direction)
T: slice thickness
N: number of slices acquired per single axial rotation

Although the $CTDI_{100}$ does not represent the patient dose and does not yield information about the radiation attenuation within real tissue it is a important parameter for dose comparison between different CT systems and scan protocols. By default cylindrical PMMA phantoms (usually with a diameter of 32 cm/16 cm for the adult body/head section and 16 cm/10 cm for the pediatric body/head section) are used in clinical routine during the acceptance and constancy tests. However, for a more realistic investigation of the radiation dose , semi-anthropomorphic water-equivalent phantoms that exhibit the same x-ray attenuation properties as liquid water in for tube voltages of 80 kV to 140 kV can also be a useful tool for dose evaluation.

Since the chamber has a homogeneous response along its axis, it not only integrates over radiation within the slice, but also detects scattered radiation along its entire length. The weighted CTDI ($CTDI_{weight}$) more reflects the real absorption of x-rays in the examined body and is an established metric to quantify the dose. It is the weighted sum of the central and peripheral CTDI values:

$$CTDI_{100,weight} = \frac{1}{3} CTDI_{100,center} + \frac{2}{3} CTDI_{100,periphery}$$

where $CTDI_{100,c}$ is the CTDI measured in the center hole of the phantom and $CTDI_{100,p}$ is the average value of the four peripheral CTDIs.
Even though the CTDI characterizes the particular CT system it is not a direct measure of the real patient dose as it depends on scan range, patient size and anatomy etc.

Therefore, the dose-length-product (DLP) is commonly considered to estimate the patient dose resulting from a complete examination:

$$DLP = \Sigma_i\, CDTI_{100,weight,i} \cdot N_i \cdot N_{rot,i} \cdot T_i \cdot C_i$$

The summation with index i accounts for all scanning sequences of the examination. N_i indicates the number of simultaneously acquired slices, $N_{rot,i}$ is the number of sequential scans or spiral rotations and T_i is the collimation (slice thickness) in examination i. C_i is the mAs value of the i^{th} scan. Typically, the standard value of 100 mAs is used for the determination of the DLP.

The volumetric CTDI ($CTDI_{vol}$) is another important metric used for dose evaluations particularly for helical CT scans which takes the pitch factor into account. Each CT scan protocol indicates the $CTDI_{vol}$ as part of the scan report. It is an important indicator for comparing patient scans and protocols during clinical routine.
It is calculated from the ratio of weighted CTDI over pitch:

$$CTDI_{volume} = CTDI_{100,weight} / pitch$$

CTDI and DLP assessment with QRM Phantoms

- After placing the phantom on the couch, it should be positioned accurately at the isocenter by using the internal alignment lasers.
- Insert the 100 mm pencil-shaped ion chamber in the phantom (using the adapter) and connect it to the calibrated electrometer.
- Acquire the values first of from the central position and then from the peripheral positions (top, right, bottom, left).
- Scanning parameters (kVp, mAs, collimation, pitch etc.) are recorded for each scan protocol.
- During the measurement of one position, all other bore holes must be filled with plugs

Suitable Phantoms

- QRM-40100 CTDI CTwater Phantom
- QRM-40101 Oval CTDI CTwater Phantom
- QRM-40104 Thorax Dosimetry Phantom
- QRM-40105 Abdomen Dosimetry Phantom
- QRM-20120 Pediatric Thorax Phantom, newborn
- QRM-20137 Pediatric Thorax Phantom, 1 year
- QRM-20138 Pediatric Thorax Phantom, 3 years
- QRM-20121 Pediatric Thorax Phantom, 6 years
- QRM-20123 Pediatric Thorax Phantom, 12 years
- QRM-20139 Pediatric Thorax Phantom, 15 years
- QRM-20125 Pediatric Abdomen Phantom, newborn
- QRM-20140 Pediatric Abdomen Phantom, 1 year
- QRM-20141 Pediatric Abdomen Phantom, 3 years
- QRM-20142 Pediatric Abdomen Phantom, 6 years
- QRM-20143 Pediatric Abdomen Phantom, 12 years
- QRM-20144 Pediatric Abdomen Phantom, 15 years

4 Multi-Energy and Photon Counting CT

Multi-Energy CT (MECT)

The linear attenuation coefficient and, hence, the CT value depends on the material's effective atomic number, effective X-ray energy (poly-energetic spectrum) and mass density. Therefore, in conventional CT imaging materials with different atomic numbers might appear with similar CT values at a particular tube voltage making material differentiation difficult. As the linear attenuation coefficient depends on the X-ray energy, this limitation can be circumvented by measuring at different tube voltages, i.e., different (poly-energetic) spectra and the materials can be separated. This approach is called Dual- or Multi-energy imaging. The most common example is the separation of Calcium and Iodine, but recently also other materials and contrast media gain importance (e.g. Gd, Fe, etc.).

The basic principle behind dual- or multi-energy imaging is the energy-dependence of the linear attenuation coefficient: the dominant interactions between X-ray radiation and tissue occurring at the energies used in diagnostic imaging are Compton scattering and Photoelectric absorption. As the tube voltage, and hence, the X-ray energy increases, the cross-section of photoelectric effect decreases, while the Compton interaction gains importance.

In clinical practice, there are different technical approaches to multi-energy imaging invlolving photon counting detectors , dual-layer CT, kV-switching or dual-source CT. All these material-specific imaging techniques provide valuable information regarding the contribution, concentration, and chemical composition of the materials in specific tissues and enables separate visualization these materials.

Photon-Counting CT

Among all established approaches of Multi- or Dual-energy imaging, the new generation of CT scanners using photo-counting detectors is particularly promising.
X-ray detectors used in conventional CTs integrate over the energy of all radiation deposited in the detector, i.e. the signal comprises all photons absorbed in the detector. By contrast, X-rays absorbed in photon counting detectors are directly converted into electrical signals and hence, counted individually. Especially, it is possible to detect the energy of the individually detected photons as the amplitude of the electrical signal in the detector scales with the energy of the incident photon. This new approach offers multiple benefits with respect to conventional detectors.
Due to the way the new detectors are constructed, PC detectors exhibit smaller pixels compared to conventional detectors leading to enhanced spatial resolution. Further, as PC detectors are not susceptible to electronic noise, high contrast-to-noise ratio, low dose and reduced image artifacts can be achieved. The main advantage of PC technology is that the energy of the individual X-ray photons can be detected. Therefore, photons can be sorted according to their energy. Selecting appropriate energy thresholds possibly yields the separation of multiple contrast media or elements (e.g. Ca, I, Fe, Gd) with a single scan. Being challenging at the moment, this new technique paves the way to various new/future applications. For material decomposition, the CT values of different materials at low tube voltage (e.g. 80 kV) is plotted against the CT value at high tube voltage (e.g. 140 kV). With this method, materials can be differentiated from each other according to their position on the graph.
Modern dual- or multi-energy or PC post-processing protocols offer the possibility to create so-called virtual non-contrast images or iodine/calcium maps which automatically disentangle tissues rich in Iodine (or other contrast agents) or calcium. These images can be combined to color-maps with different colors indicating the presence of a specific element in the tissue (calcium, iodine or other contrast agents).

Multi-energy CT postprocessing and material decomposition with QRM Phantoms

- Position the phantom accurately at the isocenter by using the internal alignment lasers.
- Measure the phantom with standard scan protocols and evaluate different post-processing techniques and algorithms regarding the correct material separation, concentration etc.
- If necessary, repeat the measurement with varied positions of the test rods (possible for QRM-10150, QRM-10147 and QRM-10139).

Suitable Phantoms

- QRM-10150 Multi-Energy QA Phantom
- QRM-10147 Spectral CT Phantom II
- QRM-10139 Spectral CT Phantom
- QRM-10123 Dual Energy CT Phantom, V5
- QRM-10107 Dual Energy CT Phantom, V2

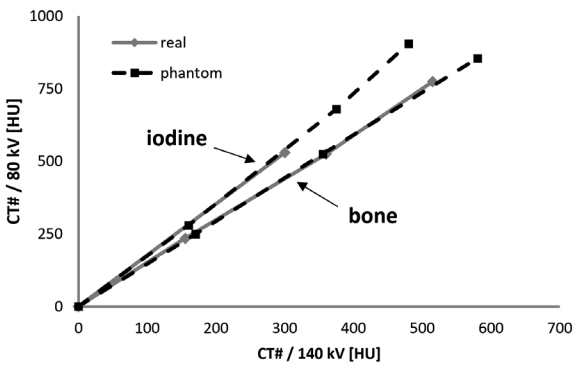

Figure 10: CT# of Iodine and CaHA-enriched material

5 CT Value Calibration in RT Planning Systems

CT Value calibration in Radiotherapy Treatment Planning Systems

In radiotherapy, it is crucial to deliver the prescribed dose to the target volume with high precision. Therefore, different algorithms are used in treatment planning systems (TPSs), which take elemental composition, stopping power, and electron density relative to water (ρ_e^w) into account when determining tumor and OAR (organs at risk) dose distribution. In order to perform pixel-by-pixel heterogeneity correction based on the estimated attenuation within the non-uniform tissue, an electron density map is generated corresponding to the CT values for all clinically used photon energies. Since the predominant tissue-photon interaction at typical X-ray energies in CT is Compton scattering and Photoelectric effect, the linear attenuation coefficient is considered to be proportional to the relative electron density of the tissues. The electron densities are calculated from the mass density and the elemental composition of each material. By scanning a phantom containing different tissue-equivalent materials with known electron densities on the tomography device, the CT values of each tissue-equivalent material can be determined and an electron density map (calibration curve) is created, which visualizes the relation between the relative electron densities and the corresponding CT values. Treatment planning systems use the linear fitting equation for heterogeneity corrections. The linear fitting equation is adapted to the CT models and acquisition parameters such as tube voltage, mAs, reconstruction algorithms, filters etc.

Suitable Phantoms

- QRM-90114 Comprehensive Electron Density Phantom
- QRM-90110 Electron Density Phantom, D100

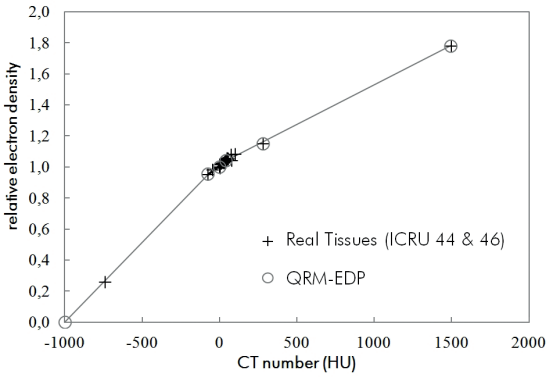

Figure 11: Relative Electron Density vs CT# (at 120 kV)

Electron Density Calibration Curve assessment with QRM Phantoms

- Although the tissue-equivalent electron density rods can be arranged in various ways depending on the research purpose, it is recommended to be place them according to the manufacturer's recommendation to minimize artifacts and achieve the best possible results.
- It is possible to compare results obtained in body geometry (phantom body and head insert) and using the head insert as stand-alone phantom.
- Place the phantom on the couch and insert the tissue-equivalent electron density rods according to the manufacturer's recommendation.
- Position the phantom accurately at the isocenter by using the internal alignment lasers.
- Acquire CT images within 3 different slices, one at centrally within the phantom and two slightly off-centered (approx. ± 1 cm). To have a good SNR, a 5+ mm slice thickness should be appropriate
- Place ROIs withing the tissue-equivalent inserts and evaluate the mean CT values. Use ROIs smaller than the inserts' diameter in order to exclude cupping/edge effects (in particular in the highly absorbing bone insert).
- Create a calibration curve using the measured CT values (average of three slices) and the corresponding electron density relative to water (ρ_e^w) provided by the manufacturer.

6 Bone Mineral Density

Osteoporosis is a condition that develops more rapidly in women after menopause than in men and increases fracture risk due to bone mineral loss and decrease in bone structure. Dual energy X-ray absorptiometry (DXA, DEXA) is the most often used method to measure bone mineral density. The method has gained acceptance for screening osteoporosis patients. DXA systems are widely used in hospitals and medical practices.
Quantitative CT (qCT) is beside DXA a standard method in bone densitometry. Due to its ability to acquire 3-dimensional images and its higher image quality, bone mineral density assessment is more accurate than DXA.

Calcium hydroxyapatite (CaHA) is the main component of bone. Because of its relatively high effective atomic number (Z_{eff}) and high mass density CaHA has a high linear attenuation coefficient compared to soft tissues. This can be exploited to measure the bone mineral density by methods such as dual energy X-ray absorptiometry (DXA) or quantitative CT (qCT).

Dual energy absorptiometry is a planar X-ray technique to assess bone mineral density. Considering the X-ray attenuation using two different poly-energetic spectra allows to separate the attenuation by bone from that of soft tissue.

While DXA gives information about the area bone mineral density (aBMD) in g CaHA/cm^2 , the volumetric density (vBMD) in g CaHA/cm^3 can be acquired with qCT. Further, as 3D cross-sectional images are obtained, this approach facilitates to evaluate the vBMD of cortical bone and trabecular bone separately.

In BMD examinations, there are specific criteria determined by the World Health Organization to assess the BMD which are the T score and the Z score. The T score is the standard deviation of the individual's BMD value to the average BMD of the young population of the same sex. This quantity serves for the diagnosis of osteoporosis: a value above -1 implicates a BMD in the normal range, a T score between -1 to -2.5 indicates osteopenia (low BMD) and a value below -2.5 implies osteoporosis. The Z score is the standard deviation relative to the average BMD of the same age group providing a comparison to the average population of the same age.

As the accuracy of the analysis, for both, DXA systems and CT systems, is very prone to beam hardening effects, amount of adipose tissue around the investigated area, the exact spectrum used, etc.
Hence, the actual value of the BMD can drastically vary between systems. In addition, different manufacturers use their own calibration methods which also lead to varying BMD values.
Therefore, it is crucial to calibrate the CT or DXA systems using phantoms containing well-defined BMDs.

Different body regions are commonly used for BMD analysis. Regions of highest interest are the lumbar spine, and the distal forearm. Other areas of interest are the Hip-region, the femoral neck and the distal legs (heel).

There are different phantoms available satisfying the requirements of DXA or qCT (or both) for the different body regions mentioned above. In qCT, one can create a calibration between CT values and BMD using specifically designed phantoms. DXA systems can be calibrated by phantoms specially designed for DXA applications. Furthermore, there are phantoms that are suitable for a cross-calibration between different DXA systems (e.g. for multi-center studies).

In addition to the assessment of Bone Mineral Density in female or male population, it is also possible to evaluate BMD of small animals using Micro-CT systems. For more information, please read the Micro-CT Section.

Bone Mineral Density assessment with QRM Phantoms in qCT

- Position the phantom accurately on the couch in the desired configuration.
- Acquire CT images in the region(s) of interest (e.g. in the different vertebrae of the ESP). To have a good SNR, use at least 5 mm slice thickness.
- Place several ROIs within the bone inserts (spongious/trabecular part) and measure the mean CT values. Importantly, use ROIs smaller than the inserts' diameter in order to exclude edge effects.
- Create a calibration curve (linear fit) using the measured CT values and the corresponding electron vBMD values provided by the manufacturer.
- This calibration curve can be used to determine the vBMD value of any bone under investigation.

Figure 12: Cross-calibration example (ESP with BDC Phantom)

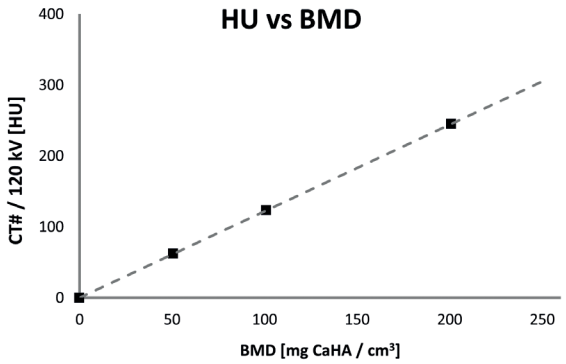

Figure 13: Example for BMD calibration curve

Bone Mineral Density assessment with QRM Phantoms in DXA

- Position the phantom accurately on the couch in the desired configuration.
- Acquire the aBMD using a standard examination.
- Compare the value of the a BMD to the one provided by the phantom manufacturer

Suitable Phantoms for DXA and qCT

- QRM-50100: European Spine Phantom (ESP)

Suitable Phantoms for DXA

- QRM-50100 European Spine Phantom (ESP)
- QRM-50110 DXA Spine QA Phantom, 3 HA
- QRM-50146 DXA Femur Phantom
- QRM-50139 JIS Forearm Phantom
- QRM-50143 JIS Heel Phantom
- QRM-50140 JIS Lumbar Spine Phantom
- QRM-50144 JIS Uniform Lumbar Phantom

Suitable Phantoms for qCT

- QRM-50100 European Spine Phantom (ESP)
- QRM-50111 European Forearm Phantom (EFP)
- QRM-50112 Forearm Phantom (BMF)
- QRM-50113 HIP Calibration Phantom, V2
- QRM-50131 HIP-QC120 Phantom
- QRM-50121 Spine-QC120 Phantom
- QRM-50137 Knee-QC Phantom
- QRM-50115 Bone Density Calibration Phantom, 3 H200
- QRM-50118 Bone Density Calibration Phantom, 3 H300
- QRM-50120 Bone Density Calibration Phantom, 3 H400
- QRM-50116 Bone Density Calibration Phantom, 3 H500
- QRM-50117 Bone Density Calibration Phantom, 3 H600
- QRM-50119 Bone Density Calibration Phantom, 3 H700
- QRM-50124 Bone Density Calibration Phantom, 6 H200
- QRM-50125 Bone Density Calibration Phantom, 6 H300
- QRM-50129 Bone Density Calibration Phantom, 6 H400
- QRM-50126 Bone Density Calibration Phantom, 6 H500
- QRM-50127 Bone Density Calibration Phantom, 6 H600
- QRM-50128 Bone Density Calibration Phantom, 6 H700

7 Pre-clinical Imaging (Micro-CT)

Micro-CT enables non-destructive material testing (NDT) at industrial sites and visualization and analysis of the internal structures of small animals (mouse, rat, etc.) for preclinical research purposes. Depending on the desired application, there are various systems offering a wide range of possible sample size and resolution. Nowadays spatial resolution in the sub-micrometer range can be reached.

In clinical CT scanners the gantry with X-ray tube and detector rotates around the patient. In micro CT, there are also other setups that can be found. E.g. the sample can be directly rotated around its axis (z-axis) between a fixed X-ray tube and detector (mostly for industrial applications). By varying the distances between sample, detector and X-ray tube the magnification can be adjusted. While rotating the sample, multiple projections at different angles are obtained and reconstructed in order to get a 3D image of the sample.

Compared to clinical CT, the X-ray beam originates from a very small focal spot (microfocus) enabling the superior spatial resolution.

Micro-CT systems are usually no medical devices and are not calibrated to the Hounsfield scale. There are not strictly pre-defined and calibrated scan protocols as in clinical systems, but rather various scan geometries as the user can adjust multiple settings (tube voltage, tube current, focal spot, distance between detector, sample and X-ray tube, acquisition time per projection, number of projections, etc.) in order to get the best imaging result. Unlike in clinical devices, there are no prescribed daily, weekly or monthly standard quality assurance test procedures. Especially, due to the specific scan geometry (Cone-beam) and detector type (usually flat-panel detectors are used), Micro-CTs are very prone to image artifacts.

Therefore, QA test procedures are essential to ensure constant quality and comparability of results obtained at a specific system or at several systems.
In principle, the image quality parameters/metrics and calibration procedures recommended for Micro-CT systems are the same as for clinical CT devices and can be assessed using high-tech phantoms specially developed for Micro-CT.
There are various phantoms available addressing different tasks: spatial resolution (directly by line patterns or indirectly by evaluation of the MTF), low-contrast resolution (CNR), signal-to-noise ratio (SNR), uniformity, grey value accuracy (linearity), geometric accuracy, in-vivo or in-vitro bone mineral density calibration and dose evaluation (CTDI and DLP).

Assessment with QRM Phantoms
The assessment methods explained above in the clinical CT part of this document can be applied to Micro-CT phantoms as well. Therefore, please read the corresponding section for clinical CT for more details of these different measurement tasks. Instead of quantifying the CT values in Hounsfield units, use the grey value scale of your Micro-CT.

Suitable Phantoms for Micro-CT
Spatial resolution:
- QRM-70109 Micro-CT Slice Sensitivity Phantom
- QRM-70113 Micro-CT Bar Pattern Phantom, air
- QRM-70114 Micro-CT Bar Pattern Phantom, resin
- QRM-70119 Micro-CT Bar Pattern Phantom, NANO
- QRM-70100 Micro-CT Wire Phantom air, D20, 10 micron
- QRM-70101 Micro-CT Wire Phantom air, D20, 25 micron
- QRM-70102 Micro-CT Wire Phantom air, D32, 10 micron
- QRM-70103 Micro-CT Wire Phantom air, D32, 25 micron
- QRM-70117 Micro-CT Wire Phantom air, D20, 3 micron
- QRM-70118 Micro-CT Wire Phantom air, D32, 3 micron
- QRM-70130 Micro-CT Wire Phantom resin, D32, 25 micron
- QRM-70131 Micro-CT Wire Phantom resin, D32, 10 micron
- QRM-70133 Micro-CT Wire Phantom resin, D20, 25 micron

Grey value accuracy (linearity):
- QRM-70105 Micro-CT Contrast Scale Phantom

Low-contrast resolution:
- QRM-70108 Micro-CT Low Contrast Phantom, V1
- QRM-70124 Micro-CT Low Contrast Phantom, V2

Signal-to-noise ratio, Uniformity (homogeneity):
- QRM-70110 Micro-CT Water Phantom, D60
- QRM-70111 Micro-CT Water Phantom, D32
- QRM-70112 Micro-CT Water Phantom, D20

Geometric accuracy (distortion) and post-processing techniques:
- QRM-70104 Micro-CT Multi Disk Phantom (Defrise)
- QRM-70137 Micro-CT Mouse Phantom

Bone mineral density calibration:
- QRM-70107 Micro-CT HA Phantom D32
- QRM-70129 Micro-CT HA Phantom D25
- QRM-70126 Micro-CT HA Phantom D20
- QRM-70127 Micro-CT HA Phantom D10
- QRM-70128 Micro-CT HA Phantom D4.5
- QRM-70134 Micro-CT HA set of 5 single rods

Dose phantoms (CTDI and DLP):
- QRM-70106 Micro-CT Dose Phantom

Notes

Phantoms for your needs.

PTW is a global market leader for dosimetry and quality control solutions in radiation medicine, serving the needs of medical radiation experts in more than 160 countries worldwide. Starting with the famous Hammer dosemeter in 1922, the German manufacturer is the pioneer in medical radiation measurement, known for its unparalleled quality and precision.

Since April 2020 QRM is a subsidiary of PTW Freiburg GmbH. QRM looks back on more than 25 years of experience in designing and manufacturing phantoms for medical imaging modalities. QRM phantoms are professionally designed and manufactured with highest precision.

For more information on QRM Phantoms visit qrm.de or contact your local PTW representative: ptwdosimetry.com/en/contact-us/local-contact

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