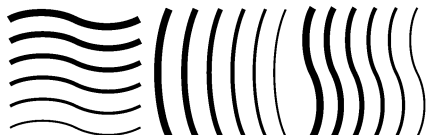


# **Quality assurance of cone-beam CT for radiotherapy**

**NEDERLANDSE COMMISSIE VOOR STRALINGSDOSIMETRIE**

**Report 32 of the Netherlands Commission on Radiation Dosimetry  
March 2019**



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Subcommittee “Quality assurance of cone-beam CT for  
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## **Disclosure of Potential Conflicts of Interest**

The NKI-AvL has a close collaboration with Elekta, Crawley, UK regarding ConeBeam CT and has had a hand in the design of the MIMI<sup>TM</sup> phantom.

## Preface

The Netherlands Commission on Radiation Dosimetry (Nederlandse Commissie voor Stralingsdosimetrie, NCS) was officially established on September 3rd, 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, installed upon the suggestion of the supporting societies, including the Netherlands Society for Radiotherapy and Oncology (Nederlandse Vereniging voor Radiotherapie en Oncologie), the Dutch Society of Nuclear Medicine (Nederlandse Vereniging voor Nucleaire Geneeskunde), the Dutch Society for Medical Physics (Nederlandse Vereniging voor Klinische Fysica), the Netherlands Radiobiological Society (Nederlandse Vereniging voor Radiobiologie), the Netherlands Society for Radiological Protection (Nederlandse Vereniging voor Stralingshygiëne), the Dutch Society for Medical Imaging and Radiotherapy (Nederlandse Vereniging Medische Beeldvorming en Radiotherapie), the Radiological Society of The Netherlands (Nederlandse Vereniging voor Radiologie), the Dutch Society for Medical Physics Engineers (Nederlandse Vereniging van Klinisch Fysisch Medewerkers) and the Belgian Hospital Physicists Association (Belgische Vereniging voor Ziekenhuisfysici / Société Belge des Physiciens des Hôpitaux).

To pursue its aims, the NCS accomplishes the following tasks: participation in dosimetry standardisation and promotion of dosimetry inter-comparisons, drafting of dosimetry protocols, collection and evaluation of physical data related to dosimetry. Furthermore, the commission shall maintain or establish links with national and international organizations concerned with ionising radiation and promulgate information on new developments in the field of radiation dosimetry.

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# Quality assurance of cone-beam CT for radiotherapy

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## Summary

In February 2012, the NCS installed a new subcommittee to develop guidelines for quality assurance of cone-beam computed tomography (CBCT) for radiotherapy. This report was written by Dutch medical physicists and medical physics engineers, all experienced users of CBCT equipped treatment machines, X-ray systems and portal imagers.

The aim of this report is to present uniform guidelines for the commissioning and quality assurance of X-ray based image guidance systems on conventional linacs, i.e. CBCT (XVI, OBI/kV imager), and portal imagers. The guidelines are based on current literature as well as clinical experience from the participating members of the NCS subcommittee on quality assurance of CBCT for radiotherapy.

The introduction describes the need for comprehensive QA of kV CBCT and gives a short overview of the focus of this report. The second chapter covers the major CBCT equipped treatment machines and the QA recommendations by the manufacturers. In Chapter 3, the geometric calibration, accuracy and distortion of the imaging devices are highlighted. Furthermore, an end-to-end test which checks the registration and couch movement accuracy is described. In Chapter 4, the recommendations for regular X-ray tube QA and 2D and 3D kV dosimetry of CBCT are described. Chapter 5 continues with QA checks for image quality and an overview of known artefacts. In Chapter 6, the safety aspects are described. Chapter 7 focuses on 4D CBCT and how this workflow can be implemented into the clinic. In the last Chapter, special attention is paid to software related aspects of the IGRT systems. The report further gives a recommendation on which tests should be added to the regular daily, monthly and annual QA program. In addition, tests which should be performed following system repair or upgrades are described.

The subcommittee has attempted to make a complete and thorough report. This report should be considered as a set of guidelines for proper QA for kV CBCT and MV portal imagers. The user is advised to take the local situation into consideration when implementing the proposed guidelines.

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## Contents

Preface.....	iii
Summary.....	v
Contents.....	vi
Abbreviations .....	ix
1 Introduction .....	11
2 System overview .....	13
2.1 Description of Varian image-guidance components.....	13
2.2 Varian QA recommendations on the OBI.....	14
2.2.1 Introduction .....	14
2.2.2 Routine QA recommendations for kV imaging .....	14
2.2.2.1 Safety and functionality checks .....	14
2.2.2.2 Geometry checks .....	14
2.2.2.3 Image quality checks.....	15
2.2.3 Routine QA recommendations for MV imaging.....	15
2.2.3.1 Safety and functionality checks .....	15
2.2.3.2 Geometry checks .....	15
2.2.3.3 Image quality checks.....	15
2.3 Description of Elekta image-guidance components .....	16
2.4 Elekta QA recommendations on the XVI .....	17
2.4.1 Introduction .....	17
2.4.2 Routine QA recommendations for kV imaging .....	17
2.4.3 Routine QA recommendations for MV imaging.....	17
2.5 Conclusion .....	17
3 Geometry .....	19
3.1 Geometric calibration .....	19
3.1.1 kV imaging .....	19
3.1.2 MV imaging .....	21
3.2 Geometric accuracy .....	21
3.3 Geometric distortions .....	24
3.4 End-to-end test.....	24
3.5 Discussion.....	26
4 X-ray radiation output .....	28

4.1	X-ray tube QA .....	28
4.1.1	Kilovoltage peak accuracy and Half-Value Layer value .....	29
4.1.2	Timer accuracy and linearity.....	30
4.1.3	Tube current accuracy and linearity.....	31
4.2	2D kV imaging output .....	32
4.3	CBCT dosimetry .....	34
4.3.1	Introduction .....	34
4.3.2	3D kV dose measurements in CTDI phantom following the IAEA method .....	35
4.3.3	3D kV CTDI phantom dose measurements following the ICRU/AAPM method 37	
4.3.4	Simplified method to measure CTDI equivalent dose for clinical presets .....	38
4.3.4.1	Correction factor tables for the Elekta XVI and Varian OBI systems.....	39
4.4	Summary.....	42
5	Image quality and artefacts .....	43
5.1	Image quality.....	43
5.1.1	Planar imaging .....	44
5.1.1.1	Spatial resolution.....	44
5.1.1.2	Low contrast detectability .....	44
5.1.1.3	Uniformity.....	45
5.1.1.4	Geometrical features.....	45
5.1.2	kV CBCT imaging.....	45
5.2	Recommendations .....	47
5.2.1	Planar imaging .....	48
5.2.1.1	aSi flat panel imagers (MV and kV) .....	48
5.2.1.2	Camera-based imagers (MV) .....	48
5.2.2	kV CBCT imaging.....	49
5.3	Image artefacts in kV projections and reconstructed CBCT images .....	49
5.3.1	Method/patient related artefacts .....	50
5.3.2	Artefacts caused by technical failures .....	53
5.4	Summary.....	56
6	Safety.....	57
6.1	Mechanical safety .....	57
6.2	Radiation safety .....	58

6.2.1	Interlocks and warning lights .....	58
6.2.2	Radiation leakage .....	58
6.3	Recommendations .....	59
7	4D CBCT.....	60
7.1	Basic principles .....	60
7.2	Image quality.....	60
7.3	Geometric accuracy and distortions .....	61
7.4	Registration accuracy for 4D CBCT.....	62
7.5	Comparison between 3D and 4D IGRT workflow .....	62
7.6	Recommendations .....	65
8	Software.....	66
8.1	Aspects of QA of IGRT software .....	66
8.2	Connectivity .....	66
8.2.1	Reference data import.....	67
8.2.2	Patient selection and saving acquired images.....	69
8.2.3	Remote couch movement .....	69
8.2.4	Data export .....	70
8.2.4.1	Image data sets.....	70
8.2.4.2	Image registration results.....	71
8.3	Image registration .....	71
8.4	Data storage, archival/retrieval and backup.....	75
8.5	Additional functionality and third party software.....	75
8.6	Summary of tests .....	75
	Acknowledgements .....	78
	References.....	79



## Abbreviations

AAPM TG	American Association of Physicists in Medicine Task Group
Al	Aluminium
aSi	Amorphous Silicon
BB	Ball-bearing
CBCT	Cone-beam Computed Tomography
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
DICOM	Digital Imaging and Communications in Medicine
DoF	Degrees of Freedom
DRR	Digitally Reconstructed Radiograph
EPI	Electronic Portal Imaging
EPID	Electronic Portal Imaging Device
FoV	Field of View
GTV	Gross Tumour Volume
HU	Hounsfield Unit
HVL	Half-Value Layer
IAEA	International Atomic Energy Agency
ICRU	International Commission on Radiation Units & Measurements
IGRT	Image-Guided Radiotherapy
kV	kilovoltage
kVp	kilovoltage peak
linac	Linear accelerator
MC	Monte Carlo
MRI	Magnetic Resonance Imaging
MTF	Modulation Transfer Function
MV	megavoltage
NCS	Nederlandse Commissie voor Stralingsdosimetrie
NVKF	Nederlandse Vereniging voor Klinische Fysica
OBI	On-Board Imager <sup>®</sup>
PTV	Planning Target Volume
PV	Portal Vision
QA	Quality Assurance
RF	Radio Frequency
RT	Radiotherapy

RTT	Radiotherapy Technologist
R&V	Record & Verify
SAD	Source-to-Axis Distance
SBRT	Stereotactic Body Radiotherapy
SRS	Stereotactic Radiosurgery
SSD	Source-to-Skin Distance
TPS	Treatment Planning System
VSIM	Virtual Simulation
XVI	X-ray Volume Imaging

## 1 Introduction

In modern radiotherapy, image guidance plays a very important role in minimizing geometrical uncertainties, i.e. setup errors and anatomical variations, thereby allowing a reduction of PTV margins. To this end, a large variety of image guidance equipment has been developed, ranging from ultra sound devices, optical guidance systems, RF beacons, planar X-ray systems, CBCT systems, imagers on rails, to MR guided systems. Of all these systems, CBCT systems have become a standard for image guidance in most modern radiotherapy departments. The basis for its success is the capability to produce 3D images of the region of interest in a relatively short time (1-3 min) in a sufficient quality for the purpose of both patient alignment and the evaluation of target coverage.

Because of the large variety in image-guidance systems on the market, it would be very challenging to develop specific QA guidelines for all systems. Given the high prevalence of CBCT within the community, the subcommittee has therefore opted to restrict the report to quality assurance of CBCT for radiotherapy, but including X-ray and portal imagers, as the procedures involved in the latter are very similar to those necessary for CBCT QA.

Popular kV CBCT systems are the Varian OBI kV imaging system (Varian Medical Systems, Inc., Palo Alto, CA, USA) and the Elekta XVI system (Elekta AB, Stockholm, Sweden). New, linac-vendor independent kV CBCT systems are on the market today (e.g. for proton irradiation) and the same QA program can be used for these systems as for those from the linac-vendors. Other radiation devices like for example the Helical Tomotherapy system use pre-treatment MV CT imaging for patient setup. These systems will not be included in this report. We refer to NCS report 27 [1] and the reports of the AAPM TG 148 [2] and the AAPM TG 179 [3] for more details about those systems.

Since the start of this subcommittee, the AAPM has published the report of TG 179 [3] that also addresses the QA of the IGRT utilized CT-based technology. Moreover, the EFOMP-ESTRO-IAEA published in 2017 a protocol on quality control in cone-beam computed tomography [4]. Finally, the NVKF WAD protocols (<https://nvkf.nl/nl/documenten>) provide useful information on quality assurance of diagnostic devices. Our report has a strong practical focus with QA recommendations to be followed and is more extensive than the recommendations given by the manufacturers.

The aim of the subcommittee is to present uniform guidelines for the commissioning and quality assurance of X-ray based image guidance systems on conventional linacs, i.e. CBCT (XVI system, OBI kV imaging system), and including portal imagers. The guidelines are based on current literature as well as on the clinical experience from the participating members of this subcommittee.

## 2 System overview

There are two major manufacturers of linear accelerators, Varian Medical Systems and Elekta. In this chapter we present an overview of the imaging systems and the quality assurance recommendations of these manufacturers.

### 2.1 Description of Varian image-guidance components.

The Varian systems described in this section are Clinac's and Truebeams equipped with an On-Board Imager®, OBI. The OBI system consists of a kV X-ray source and a kV amorphous silicon detector, which are placed perpendicular to the MV beam axis. In line with the MV-beam a flat panel amorphous silicon MV-detector is placed (aSi500 or aSi1000). The detectors and kV-source are mounted on robotic (Exact™) arms.



Figure 2.1. Varian Truebeam® with On-Board Imager®.

Both the MV-detector and kV-detector can be shifted in all 3 directions. The kV-source has blades to shape a rectangular kV-field. In CBCT-mode a so-called full-fan acquisition is made over a 180° gantry rotation (plus fan angle) in which the detector is centred with respect to the source. To image larger anatomy a half-fan acquisition can be made with the detector shifted laterally. In this half-fan mode, projection images have to be acquired over a full 360° gantry rotation to be able to reconstruct 3D images of the full field of view. In CBCT-mode a bow-tie filter (either half-fan or full-fan) is placed in front of the kV-source. The system also allows for non-isocentric imaging to deal with excentric located isocentres.

The process of image-guidance is assisted by imaging software. Reference images, either DRRs with drawn contours or CT's with RT-structures are automatically loaded from the ARIA database (Varian's record and verify system) when the corresponding plan is used.

Radiation Therapists (RTTs) match the images and can move the couch remotely for image-guided corrections.

## *2.2 Varian QA recommendations on the OBI*

### **2.2.1 Introduction**

In this chapter the recommendations of Varian are presented on the QA of the On-Board Imager® [5]. Varian states that these recommendations are in addition to the QA recommendations given in the scientific literature [6-8].

Varian distinguishes three types of checks in a QA programme for the (On-Board) Imager: safety and functionality checks, geometry checks, and image quality checks. Varian gives recommendations for the frequency of the checks but states that ultimately, the frequency is based on the professional judgment of the person in charge and the requirements of the existing QA program. Varian recommends to keep records of all the performed checks.

### **2.2.2 Routine QA recommendations for kV imaging**

#### **2.2.2.1 Safety and functionality checks**

Varian recommends a daily test to check that the safety features (interlocks with respect to collisions) are functioning properly and the system is ready for clinical use. In practice, this means testing the collision devices on the elbow joint and the cassette cover. The test is described in the OBI manual [5] as ‘the collision detection and override test’.

#### **2.2.2.2 Geometry checks**

Varian recommends a daily test to check the geometric accuracy and stability of the kV source and kV detector arms. A specific test, the OBI isocentre accuracy test, is described in the OBI manual [5]. The test is based on a kV-image of the Cube phantom in the treatment isocentre. The user must compare the centre pixels with the digital graticule. The distance between the ball bearing (BB) and the digital graticule must be below 1.5 mm.

Varian recommends a weekly test to check the vertical position of the kV source. A specific test, the SAD check, is described in the OBI manual [5]. The check explains how to measure the distance between the kV source and the treatment isocentre (tolerance 2 mm) and to measure the magnification in a kV-image (tolerance 1 mm on 20 cm distance).

Varian recommends an annual test to check the positions of the blades which form the kV-field, the 'Blade Position Check' is described in the OBI manual [5]. The tolerance is 2.5 mm for a 10cmx10cm field size.

### **2.2.2.3 Image quality checks**

Varian recommends to monitor the radiographic and CBCT images over time. No specific tests are described. Regular visual inspections of the images over time by experienced operators, will reveal if the imaging systems needs to be calibrated or not. If in doubt of the image quality, the acceptance tests can be re-measured and compared to the results of the original acceptance test to see if image quality has indeed deteriorated. If image quality deteriorates, calibration measurements must be made.

During installation, or after repair, the image quality is checked against manufacturer settings by technical tests on kV and mAs values, the Leeds test (for a description of the phantoms, see chapter 5) and, for CBCT, by scanning and reading out a well-defined inhomogeneous phantom (e.g. the Catphan<sup>®</sup> phantom, see chapter 5). If values are within criteria levels, evaluation of the image quality as experienced during daily clinical use is considered the "most sensitive" check, hence Varian does not advise a specific QA program.

## **2.2.3 Routine QA recommendations for MV imaging**

### **2.2.3.1 Safety and functionality checks**

Varian recommends to perform daily safety checks for MV imaging. The check consists of collision detection and overrides. The check is identical to the safety check of the kV imaging device.

### **2.2.3.2 Geometry checks**

In the OBI manual [5] Varian recommends a daily check of the isocentre accuracy of the MV imager, the so called OBI/PV isocentre accuracy QA test for MV imaging. The check is done with a cube phantom. The procedure is identical to the procedure of the kV imager. The test can be combined with the geometric check of the kV imager due to their 90° orientation.

### **2.2.3.3 Image quality checks**

During installation, or after repair, the image quality is checked against manufacturer settings by a contrast resolution test with the Las Vegas phantom (see figure 5.1).

### 2.3 Description of Elekta image-guidance components

The Elekta systems described in this section are equipped with an X-ray Volume Imager, XVI. The XVI system consists of a kV X-ray source and an amorphous silicon kV detector panel, which are placed perpendicular to the MV beam axis. In line with the MV beam a flat panel amorphous silicon MV detector is placed (iViewGT™).



Figure 2.2. Elekta X-ray volume imager (XVI) and iViewGT™ systems.

Both the MV and kV detector panels can be unfolded, after which the detectors can be moved in two directions (motorized). The distance to the isocentre is fixed. The kV source arm is attached at the end of a pair of shafts and can be pulled out manually to a fixed position. The kV source arm contains one slot for a filter cassette and one slot for a collimator cassette. There is an empty filter cassette (F0) and a bow-tie filter cassette (F1) available.

A lead sheet in the collimator cassettes collimates the axial field length and field of view of the kV beam. In the so-called Small field of view, an acquisition is made in which the detector is centred with respect to the source in combination with an S-collimator. To image larger anatomy an M- or L collimator can be used, in combination with a lateral shift of the detector (half-fan mode). In this situation only half of the patient is scanned with a 180 degrees gantry rotation. A full 360 degrees rotation is necessary to scan the whole patient.

The process of image-guidance is assisted by imaging software. Reference images, either DRRs or CT's (with RT-structures and RT-plan) from the treatment planning system are imported into the XVI database and the region of interest for image registration of the CBCT and the reference image is defined. A CBCT is acquired with predefined settings that are optimized for the target region of the patient. The CBCT image is registered with the



reference image and the necessary table correction is determined. This table correction can be applied remotely for on-line corrections.

## *2.4 Elekta QA recommendations on the XVI*

### **2.4.1 Introduction**

In this chapter the recommendations of Elekta on the QA of the iViewGT™ MV imager [9,10] and the XVI [11] are presented. Elekta states that it is the responsibility of the authority who has control of the equipment to set up a QA program and refers to scientific literature to set up such a program [8,12-15].

### **2.4.2 Routine QA recommendations for kV imaging**

Elekta recommends a number of daily checks for properly functioning safety features. In practice, this means testing the stop button on the handheld controller, the source arm interlock for full extension, room door interlocks, collision interlocks on kV detector and source, radiation warning lamps and the functioning of the terminate buttons. Furthermore, a daily phantom scan is recommended to test the communication between the software (table move assist) and the treatment table.

Elekta recommends to check the coincidence of the kV/MV isocentre monthly. This can be done by using the BB phantom. The difference between kV and MV isocentre should be  $\leq 0.5$  mm. Furthermore, a yearly gain calibration should be performed and the 3D uniformity, 3D low contrast visibility, 3D spatial resolution, 3D transverse scales, 3D sagittal geometry, 3D registration accuracy and table movement assistant accuracy should be checked.

### **2.4.3 Routine QA recommendations for MV imaging**

For the MV imaging Elekta recommends to do a weekly check on the accuracy and acceptability of acquired images by using the Las Vegas phantom. Yearly the gain calibrations should be performed, as well as a check of the image quality with the Las Vegas phantom. Furthermore, the stations settings should be backed up yearly.

## *2.5 Conclusion*

The QA recommendations as stated by the vendors of the equipment and summarized above, can be seen as a basic set of QA requirements. However, there are differences between the requirements of the different vendors and a broader QA programme that

includes all relevant steps in the image guidance workflow is desirable. Therefore, the NCS subcommittee QA CBCT has written this report to give recommendations about the minimum QA procedures that should be used for the complete image guidance workflow on linear accelerators equipped with 2D MV and/or 2D kV imagers and CBCT. With this report we hope to achieve more uniform QA programmes for image guidance workflows in the Netherlands, Belgium and beyond.

### 3 Geometry

Linear accelerators with a kV CBCT and a MV EPID are widely used for image-guided radiotherapy. The coincidence of the kV and MV imaging isocentres and the radiation treatment isocentre is essential for high-precision, image-guided radiotherapy. Therefore, a highly accurate and efficient QA programme is required to calibrate or verify the coincidence of the kV and MV imaging isocentres with the radiation treatment isocentre.

In this chapter an overview will be given of geometric tests that are advised in each phase of the QA programme during acceptance and/or commissioning and quality control. All tests assume that:

- The room lasers coincide with the radiation treatment isocentre.
- The radiation treatment isocentre coincides with the mechanical isocentre for all energies.
- The treatment couch contains no hysteresis.
- The treatment couch motion accuracy is within tolerances [15].
- The reference images are of high resolution (CT and DRRs).

#### 3.1 Geometric calibration

The relationship between the radiation treatment isocentre and the imaging system isocentre (geometric calibration) is typically expressed as a function of gantry angle since the treatment and imaging system components flex during gantry rotation.

##### 3.1.1 kV imaging

The geometric calibration of CBCT and 2D kV systems has been extensively described [3,7,13,16] and the method is derived from the Winston-Lutz procedure [17].

Align a ball-bearing phantom to the radiation treatment isocentre (see figure 3.1) and acquire a CBCT. The apparent travel of the ball-bearing on the projection images, used for reconstruction of volumetric datasets, provides a measurement of the components' flexing as a function of the gantry angle (flexmap). Correction for these flex motions can be done digitally by the reconstruction software (Elekta Synergy<sup>®</sup> system and Varian OBI system C-series platform) or by servos in the robotic imaging arm (Varian OBI system TrueBeam<sup>®</sup> platform).

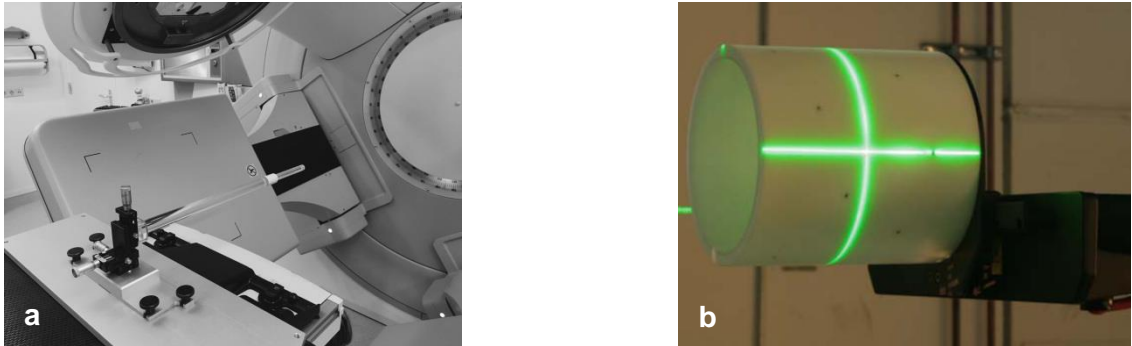


Figure 3.1. (a) Elekta ball-bearing phantom, (b) Varian IsoCal phantom.

Flexmaps should be measured for each field of view, both in clockwise and counter clockwise direction. The calibration procedure is independent of the radiation beam energy as long as you make sure beforehand that the radiation treatment isocentre coincides with the mechanical isocentre.

Flexmaps are typically measured during commissioning time and after system upgrades or service that could potentially invalidate them. AAPM TG 104 [18] and TG 179 [3] also recommend to verify the flexmaps on a monthly basis. Based on our experience of the stability of the system, we recommend to measure the flexmaps only when geometric accuracy QA is out of tolerance (provided that this QA is done on a regular basis, see section 3.2).

Table 3.1. QA recommendations for geometric calibration (<sup>#</sup> A: acceptance; U: upgrade; S: service; M: monthly).

Report	Frequency <sup>#</sup>	Tolerance
NCS “CBCT QA”	AUS and at kV-MV isocentre disagreement	Replace/refresh
AAPM TG 104 [18]	MS	Replace/refresh
AAPM TG 179 [3]	MU	Replace/refresh

After measuring the flexmaps, we recommend to do the following checks:

- Perform routine geometric accuracy QA (see section 3.2).
- Perform routine spatial resolution QA (see section 5.1).
- Examine the flexmaps for mechanical instabilities (e.g. figure 3.2) (can be helpful to diagnose problems).

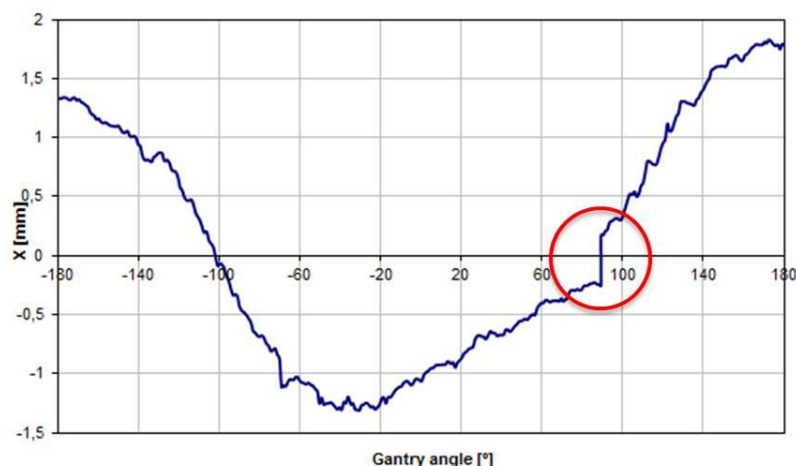


Figure 3.2. Flexmap demonstrating mechanical instability.

### 3.1.2 MV imaging

The geometric calibration procedure described in section 3.1.1 can also be used for correction of the flex of the MV imaging panel. However, if the image registration software uses field edge detection, no geometric calibration procedure is necessary. It is recommended to check the accuracy of this workflow in an end-to-end test.

## 3.2 Geometric accuracy

Testing of the geometric accuracy should be done on a regular basis as the effect of an erroneous geometric calibration is not always obviously apparent on reconstructed images. Although the flex of the system remains stable, small lateral shifts in the panel position can occur. When the displacement is perpendicular to the axis of rotation, it will cause blurring of the image. However, when the displacement is parallel to the axis of rotation, a misalignment of the CBCT image with respect to the MV isocentre will be introduced, without significant reduction in image sharpness [14].

Many different CBCT to MV isocentre coincidence tests exist [3,15,18]. We will describe a rapid as well as a lengthy but more precise procedure.

Rapid procedure:

- Align a phantom to the room lasers (e.g. QUASAR™ Penta-Guide (figure 3.3a), MIMI™ (figure 3.3b) or self-made (any phantom that contains a small well-defined, central radio-opaque structure can be used).

- Acquire 2 orthogonal portal images and match them with the reference DRRs (difference between the isocentre indicated by the room lasers and that of the treatment beam).
- Acquire a CBCT and match it with the reference CT data set (difference between the isocentre indicated by the room lasers and that of the volumetric imaging system).
- The difference between the 2 matches gives an indication of the discrepancy between the volumetric imaging system isocentre and the radiation treatment isocentre (kV-MV coincidence).
- A deviation of both the kV and MV registration in the same direction indicates either a misaligned phantom, or laser misalignment.

### Frequency

Daily (or weekly after proven long-term stability) and after collisions. Use alternating MV energy / CBCT field of view / CBCT scan direction.

### Tolerance

- Translation  $\leq 1.5$  mm in the three principal directions. Gantry sag (typically around 1 mm [19,20]) and potential collimator asymmetries will not allow smaller tolerances. Use the lengthy procedure to achieve higher accuracy ( $\leq 0.5$  mm).
- Rotations  $\leq 1$  degree. Larger rotations indicate match software / hardware problems (e.g. lag in gantry angle readout).

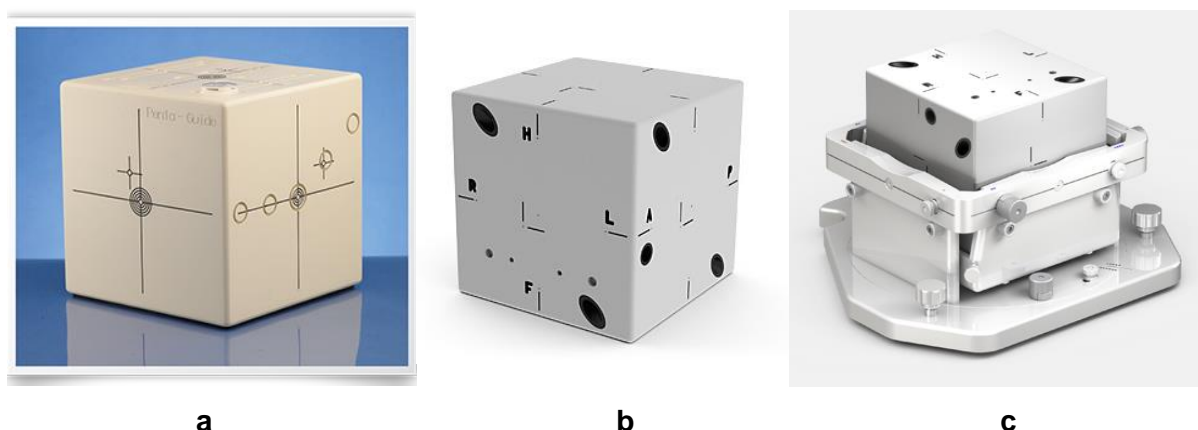


Figure 3.3. (a) QUASAR™ Penta-Guide phantom (Modus Medical Devices Inc.), (b) MIMI™ phantom (Standard Imaging, Inc.), (c) MIMI™ phantom secured to the HexaCheck base (Standard Imaging, Inc.).

Lengthy procedure:

- Align a phantom (e.g. QUASAR™ Penta-Guide, MIMI™ or self-made) to the room lasers.
- Acquire a series of 8 portal images at the 4 cardinal gantry angles (0°, 90°, 180° and 270°), with the collimator angle at both -90° and +90°, and match them with the reference DRRs. Gantry sag and potential collimator asymmetries will be averaged out.
- Acquire a CBCT and match it with the reference CT data set.
- The difference between the 2 matches gives an indication of the discrepancy between the volumetric imaging system isocentre and the radiation treatment isocentre (kV-MV coincidence).

### Frequency

Monthly (provided that a daily or weekly rapid procedure is performed). Use alternating energy / field of view / CBCT scan direction.

### Tolerance

- Translation  $\leq 0.5$  mm in the three principal directions. Note that this strict limit may not be required for all treatment sites.
- Rotations  $\leq 1$  degree. Larger rotations indicate match software / hardware problems (e.g. lag in gantry angle readout).

If the deviation of the volumetric imaging system isocentre from the radiation treatment isocentre is out of tolerance, follow the next steps:

1. Make sure that the alignment of the X-ray tube and the flat panel detector is correct (see manufacturer's maintenance manual for correct procedure).
2. Check whether the radiation treatment isocentre coincides with the mechanical isocentre (spoke shot analysis).
3. Perform a geometric calibration (flexmap measurement), see section 3.1.

*Table 3.2. QA recommendations for geometric accuracy (# D: daily; W: weekly; M: monthly).*

Report	Frequency #	Tolerance
NCS "CBCT QA"	D (or W after proven long-term stability) and after collisions (rapid procedure)	$\leq 1.5$ mm, $\leq 1$ degree
	M (lengthy procedure)	$\leq 0.5$ mm, $\leq 1$ degree
AAPM TG 104 [18]	D	$< 1$ mm
AAPM TG 142 [15]	D	$\leq 2$ mm (non-SRS/SBRT)
		$\leq 1$ mm (SRS/SBRT)
AAPM TG 179 [3]	D	$\pm 2$ mm
	M	$\pm 1$ mm

### 3.3 Geometric distortions

The accurate reconstruction of an object in EPID / CBCT images in terms of dimension and orientation can be tested with a rigid phantom with three or more embedded objects in the x-, y- and z-directions of known dimensions and orientation and a minimum separation of 10 cm (e.g. PTW EPID phantom (see figure 5.1a in chapter 5) [21] or Catphan<sup>®</sup> phantom (see figure 5.2a in chapter 5)).

- Acquire a portal image or CBCT of the phantom.
- Compare distances in 2 or 3 directions and orientation of the phantom as they appear in the portal image or CBCT images to the physical distances and orientation of the phantom.

#### Frequency

During system acceptance testing and after hardware / software upgrades.

#### Tolerance

Distance  $\leq 1$  mm in all directions (over a minimum distance of 10 cm), orientation is correct.

*Table 3.3. QA recommendations for geometric distortions (# A: acceptance; U: upgrade; M: monthly).*

Report	Frequency #	Procedure	Tolerance
NCS "CBCT QA"	AU	Distances	$\leq 1$ mm (over a minimum distance of 10 cm)
	AU	Orientation	Correct
AAPM TG 104 [18]	M	Scale & distances	$\pm 0.5$ mm
AAPM TG 142 [15]	M	Scaling	$\leq 2$ mm (non-SRS/SBRT)
			$\leq 2$ mm (SRS/SBRT, MV)
			$\leq 1$ mm (SRS/SBRT, kV)
AAPM TG 179 [3]	M	Scale, distance, orientation	Baseline

### 3.4 End-to-end test

A variant of the geometric accuracy procedures (section 3.2) can be implemented to measure the accuracy of the match and the remote treatment couch shift.

- Align a phantom with several markers (e.g. QUASAR<sup>™</sup> Penta-Guide or MIMI<sup>™</sup>) to the room lasers.



- Translate the couch at predefined distances in 3 directions. The QUASAR™ Penta-Guide and MIMI™ phantom can be moved by hand, as both phantoms have visible alignment marks not only for the true isocentre but also for an offset isocentre (this would still include a setup error but will generally be very small < 0.5 mm).
- The displacement should be significantly larger than the expected tolerances but less than 2 cm in each direction because remote-controlled couch movements are limited to 2 cm to reduce the likelihood of collisions. Moreover, use different displacements in each direction to be able to detect coordinate axes mismatches between the scan of the imaging system and the reference scan.
- If a 6DoF couch is being used clinically, the phantom should also be rotated 2.5° in 3 directions (for example by securing the MIMI™ phantom to the HexaCheck base (fig. 3.3c)).
- Acquire 2 orthogonal portal images and register them with the reference DRRs / acquire a CBCT and match it with the reference CT data set (calculated shift).
- The difference between the true and calculated shift is a measure for the registration accuracy and should be  $\leq 1$  mm &  $\leq 0.5^\circ$  (in case of kV imaging) /  $\leq 1.5$  mm (in case of MV imaging) in any direction.
- Apply the calculated shift by moving the couch remotely.
- Verify that the phantom is aligned to the room lasers again.
- Again, acquire 2 orthogonal portal images / a CBCT. The registration of these images is a measure for the couch shift accuracy and should be  $\leq 1$  mm &  $\leq 0.5^\circ$  in any direction. Larger deviations indicate an incorrect calibration of the treatment couch.

### Frequency

Weekly and after treatment couch / software upgrades.

### Tolerance

- Registration accuracy  $\leq 1$  mm &  $\leq 0.5^\circ$  (in case of kV imaging) and  $\leq 1.5$  mm (in case of MV imaging) in the three principal directions.
- Couch shift accuracy  $\leq 1$  mm &  $\leq 0.5^\circ$  in the three principal directions.

*Table 3.4. QA recommendations for match and couch shift accuracy (# U: upgrade; D: daily; W: weekly; M: monthly; An: annual).*

<b>Report</b>	<b>Frequency #</b>	<b>Procedure</b>	<b>Tolerance</b>
NCS “CBCT QA”	W and after treatment couch / software U	Match accuracy  Couch shift accuracy	$\leq 1 \text{ mm} \ \& \ \leq 0.5^\circ$ (in case of kV imaging) $\leq 1.5 \text{ mm}$ (in case of MV imaging) $\leq 1 \text{ mm} \ \& \ \leq 0.5^\circ$
AAPM TG 104 [18]	D M  An	Accuracy of shift Couch shifts: accuracy of detection and correction motions Couch scale and motion accuracy	$\pm 2 \text{ mm}$ $\pm 1 \text{ mm}$  $\pm 1 \text{ mm}$
AAPM TG 142 [15]	D	Positioning / repositioning	$\leq 1 \text{ mm}$
AAPM TG 179 [3]	D  M	Phantom localization and repositioning with couch shifts Couch shifts: accuracy of motions	$\pm 2 \text{ mm}$  $\pm 1 \text{ mm}$

### 3.5 Discussion

As QA needs to be performed in short timeslots, for the sake of efficiency users will tend to combine the tests described in sections 3.2 and 3.4 into one quick daily test, at the expense of test precision. Moreover, be aware of the risk that errors in subtests will not be exposed as they can cancel each other. Therefore, in parallel to a quick daily test, we recommend to keep performing the individual tests described in sections 3.2 and 3.4.

At days when SRS/SBRT treatments are performed at the linac, we recommend to check the treatment couch shift accuracy. This can be done beforehand by the end-to-end test described in section 3.4 or at the time of treatment by performing a second CBCT scan to verify the matched & shifted patient position.

All tests suggested in this chapter are summarized in table 3.5.

Table 3.5. Summary of suggested geometric tests (<sup>#</sup> A: acceptance; U: upgrade; S: service; D: daily; W: weekly; M: monthly).

Test	Tolerance	Frequency <sup>#</sup>	See
<i>Geometric calibration</i>			
Measurement of flexmap	Replace / refresh	AUS and at kV-MV isocentre disagreement	§3.1
<i>Geometric accuracy</i>			
Verify kV to MV isocentre coincidence	≤ 1.5 mm (rapid) ≤ 0.5 mm (lengthy)	D <sup>(*)</sup> and after collisions M	§3.2
<i>Geometric distortions</i>			
Verify distances	≤ 1 mm <sup>(**)</sup>	AU	§3.3
Verify orientation	Correct	AU	§3.3
<i>End-to-end test</i>			
Verify registration accuracy	≤ 1 mm & ≤ 0.5° (kV) ≤ 1.5 mm (MV)	WU	§3.4
Verify couch shift accuracy	≤ 1 mm & ≤ 0.5°	WU	§3.4

(\*) W after proven long-term stability

(\*\*) over a minimum distance of 10 cm

## 4 X-ray radiation output

In this chapter an overview will be given of the X-ray radiation output tests that are advised in each phase of the QA programme during acceptance and/or commissioning and quality control.

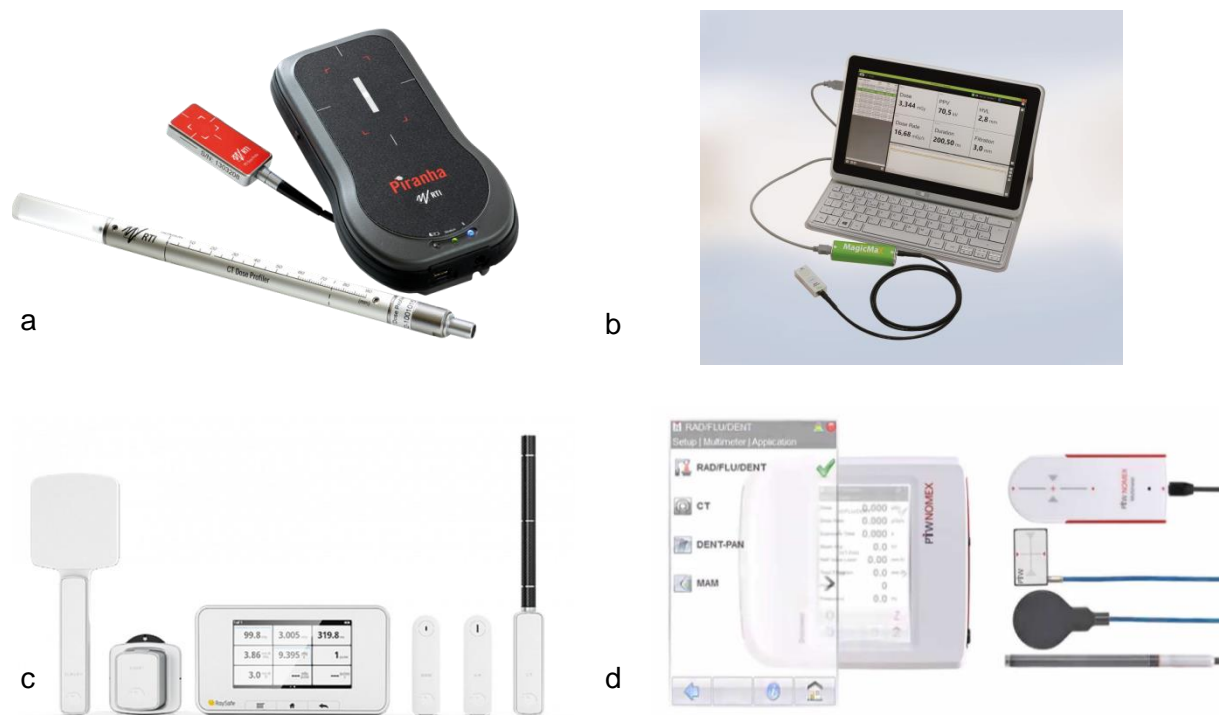
In the following sections we will present procedures on X-ray tube QA (4.1), 2D kV imaging output (4.2), and CBCT dosimetry (4.3).

### 4.1 *X-ray tube QA*

The CBCT imaging chain consists of the X-ray high-voltage generator, X-ray tube, and the digital imaging device. It is important to assure that the high-voltage generator and X-ray tube work properly in a known manner so that imaging technique parameters can be confidently adjusted by the user. For example, if the user wishes a less noisy image, he should be confident that increasing the mA setting by 50% results in 50% more radiation being incident on the patient.

To assure that the generator and X-ray tube are stable, a set of measurements should be performed periodically to reveal major changes and/or malfunction of the system.

In the acceptance phase the following tests have to be measured to check the output of the X-ray tube of the CBCT system. These test results form the baseline for the periodic QA tests. It is recommended to use a dedicated calibrated X-ray detector for the tests, for example the Piranha RTI, MagicMaX IBA, Raysafe X2, and PTW Nomex Multimeter (see figure 4.1).



*Figure 4.1. Examples of available commercial detectors to perform kV QA measurements: (a) Piranha RTI, (b) MagicMaX IBA, (c) Raysafe X2, (d) PTW Nomex Multimeter.*

The general setup for all tests in section 4.1 is:

- Position the detector at a distance of 100 cm from the source and in the centre of the imaging field.
- Rotate the gantry such that the X-ray tube is at zero degrees, straight above the detector.
- Select the desired X-ray settings.
- Perform the measurement three times (to exclude outliers) and calculate the average.

#### **4.1.1 Kilovoltage peak accuracy and Half-Value Layer value**

##### Tube potential

The voltage applied to the X-ray tube determines the energy of the X-ray photons and is a major factor in determining the contrast in the image. Assessment of the tube potential ensures that the delivered kilovoltage peak (kVp) is close to that set on the unit by the operator. Poor agreement between the two will affect clinical image quality, equipment radiation output and patient dose. The exposure kVp of the X-ray tube is set using the control panel, the accuracy of the peak voltage should be within  $\pm 5\%$  kVp of the selected setting [22,23].

Filtration

The filtration of an X-ray tube absorbs the low energy photons that do not contribute to the image but do increase the skin dose. Having adequate filtration is essential to ensure that patient dose is controlled. The total filtration should be marked on the X-ray tube housing. Total filtration can be estimated by measuring the Half-Value Layer (HVL). The HVL is the thickness of the absorber required to reduce the intensity of the incident X-ray beam by half. Unit of this quantity is “mm Aluminium” for low-energy X-rays. The HVL is an estimate of the penetrating power of the X-ray beam which means that the higher the HVL the more penetrating the X-ray beam is.

How to measure HVL

The HVL is usually measured by placing increasing thicknesses of Aluminium (Al) plates between the focus and a dosimeter, and measuring the output while keeping the kV constant as described in AAPM TG 61 [24]. Currently, there are various detectors available that have the ability to directly measure the HVL for each exposure without the need for filters and multiple exposures (e.g. the detectors in figure 4.1).

- Measure the kVp for the clinical protocols and compare these values with the chosen one on the generator.
- Measure the HVL for clinical used tube voltage and filter and compare with the specifications of the manufacturer. Minimum HVL for 70, 100 and 120 kV is 1.8, 3.6 and 4.3 mm Al, respectively [25].

Table 4.1. QA recommendations for kVp accuracy and HVL (# A: acceptance; U: upgrade; S: service; An: annual).

Report	Frequency #	Tolerance
NCS “CBCT QA”	AUSAn	kVp ± (5%) of intended kV HVL ± 5% of mm Al compared with manufacturer’s specifications
AAPM TG 104 [18]	AnS	Baseline
AAPM TG 179 [3]	AnU	Baseline

**4.1.2 Timer accuracy and linearity**

The NCS subcommittee does not give any recommendations on accuracy measurements of the absolute exposure time, as the only relevant parameter is dose/mAs, which is determined

when measuring the dose (section 4.3). As long as periodic QA is performed on dose this parameter is implicitly tested as well. However, since not all presets use the same mAs settings, exposure time linearity will have an impact on expected dose output, hence QA is required.

To measure exposure time linearity, choose a preset that is in the spectrum of the clinical range, select three different timer settings, holding peak kilovoltage and tube current constant. Perform an exposure at each setting, recording dose and pulse duration. Use an ion chamber or a dedicated commercial X-ray detector to measure the relative dose variation in the isocentre. The dose should increase linearly with exposure time.

*Table 4.2. QA recommendations for timer linearity (# A: acceptance; U: upgrade; S: service; An: annual).*

Report	Frequency #	Tolerances
NCS "CBCT QA"	AUSAn	Linearity: $1 \pm 0.05$
AAPM TG 104 [18]	AnS	Baseline
AAPM TG 179 [3]	AnU	Baseline

#### **4.1.3 Tube current accuracy and linearity**

The NCS subcommittee does not give any recommendations on accuracy measurements of the absolute tube current, as the only relevant parameter is dose/mAs, which is determined when measuring the dose (section 4.3). As long as periodic QA is performed on dose these parameters are implicitly tested as well. However, since not all presets use the same mAs settings, tube current linearity will have an impact on expected dose output, hence QA is required.

To measure tube current linearity, choose a clinical relevant preset and hold kilovoltage peak (kVp) and time constant. Select three current settings. Make an exposure at each setting, recording dose and current. Use an ion chamber or dedicated commercial X-ray detector to measure the relative dose variation in the isocentre. The dose should increase linearly with current.

*Table 4.3. QA recommendations for tube current linearity (# A: acceptance; U: upgrade; S: service; An: annual).*

Report	Frequency #	Tolerances
NCS "CBCT QA"	AUSAn	Linearity: $1 \pm 0.05$
AAPM TG 104 [18]	AnS	Baseline
AAPM TG 179 [3]	AnU	Baseline

## 4.2 2D kV imaging output

The output of a kV static/fluoroscopic image depends on the energy, filtration, number of frames, X-ray tube current, and time settings of the acquired image. 2D kV output must be performed by using a suitable calibrated detector, such as an ionisation chamber. More details on the calibration routine are described in the report of the AAPM TG 61 [24].

Some of the dedicated commercial X-ray devices are able to register dose rate as function of time (see figure 4.2), while other detectors, for example the Farmer like chamber (see figure 4.7) and pencil chamber, only provide the integrated dose value. From the dose rate as function of time registration, the dose per frame can be extracted. Note, that since the dose correlates with the number of pulses, it is necessary to normalize the number of pulses given during the acquisition. An additional advantage of using dedicated detectors is registration of parameters such as half-value layer (HVL) and the peak kilovoltage (kVp).

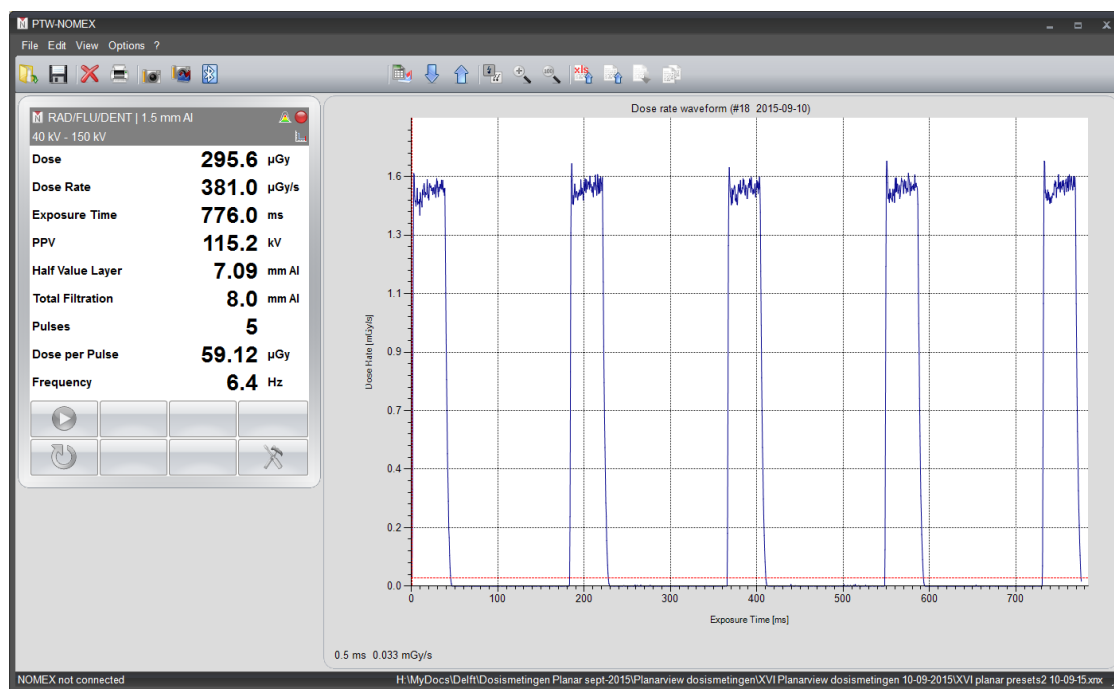


Figure 4.2. Example of dose rate as function of time using the PTW Nomex detector.

The procedure to measure dose in air of a kV static/fluoroscopic image is:

- Position the ionization detector (without build-up) in air at the isocentre as depicted in figure 4.3.



- Rotate the gantry such that the X-ray tube is at zero degrees, straight above the detector.
- Select the desired X-ray settings.
- Record the measurement, correct for temperature and pressure effects, and convert the measurement to dose.
- Perform the measurement three times (to check for outliers) and calculate the average dose per frame/pulse.
- Compare the measured values with manufacturer’s reference.
- Note that when applied to 3D acquisitions, this procedure will yield identical results.

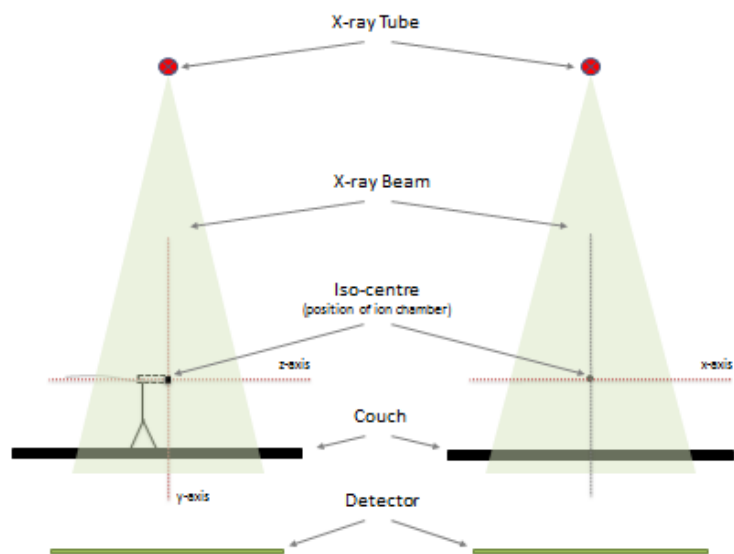


Figure 4.3. Schematic overview of the setup of the ionization detector on the couch.

Not all manufacturers provide reference values for comparison. If this is the case, we suggest to create a baseline after checking if the X-ray parameters such as HVL, kVp, mA, and time per frame are within the manufacturer’s specification.

Table 4.4. QA recommendations for 2D kV imaging output (# A: acceptance; U: upgrade; S: service; An: annual).

Report	Frequency	Tolerances
NCS “CBCT QA”	AUSAn	Within ± 5% of manufacturer’s specifications or baseline
AAPM TG 104 [18]	AnS	Baseline
AAPM TG 179 [3]	AnU	Baseline

### **4.3    *CBCT dosimetry***

The use of CBCT systems for IGRT is rapidly expanding in the developed world. CBCT imaging will yield radiation dose to the patient, both in- and outside the target area. We advise to evaluate the imaging dose, prior to clinical application.

Measurements in standard cylindrical phantoms are useful for quality assurance purposes. However, they should not be interpreted as patient dose. Because these measurements are conducted in Computed Tomography Dose Index (CTDI) phantoms or other non-realistic phantoms, it is difficult to interpret the results in real patient cases, as the individual patient body geometry and compositions greatly alter dose magnitudes and distributions.

A CBCT dose distribution can be of highly heterogeneous nature due to the use of relatively low X-ray beam energy, different CBCT geometrical setups and the half scan CBCT technique. Therefore, dose measurements at a few isolated points may not capture some key information, such as maximum dose, and may, for example, under- or overestimate the mean organ dose.

It is outside the scope of this document to give recommendations for optimizing the IGRT process to manage the additional dose due to imaging (for a complete survey, the reader is referred to the report of the AAPM TG 75 [26]). CBCT is currently used as the standard procedure for IGRT. The use of CBCT imaging during treatment is limited by the fact that CBCT gives also kV dose to the patient. Therefore, quantification of the imaging dose is necessary in order to report, optimize and validate the given CBCT exposures. There are many questions regarding CBCT dosimetry without satisfactory answers, and this has contributed to a lack of consensus among the radiotherapy community regarding appropriate measurement techniques.

The imaging dose given with Electronic Portal Imaging (EPI) can be taken into account in the treatment planning process [27]. The dose delivered by MV imaging can be precisely calculated in the TPS and therefore included in the treatment plan. In this report we therefore only focus on kV dosimetry.

#### **4.3.1    *Introduction***

Dosimetry on a CBCT system has several challenges if the conventional CTDI methodology is applied. The beam width is much larger than the length of the pencil ion chamber which is

used for conventional CT dosimetry. Furthermore, the large beam width yields the application of longer phantoms due to the longer scatter paths. Finally, CBCT systems can be used in an off-axis position with a scan protocol which only needs a scan range of 180 degrees, which leads to large differences in the peripheral measurement points. It is therefore obvious that the CTDI formalism will be inadequate for estimating patient dose.

Different reports (AAPM report 111 [28], ICRU 87 [29] and IAEA no 5 [30]) address the issues of the CTDI concept, adopting new methods of CT dosimetry which are suitable not only for axial scanning, but also for helical scanning and CBCT geometry, but are not trivial to implement.

In the next section we will describe both methods (IAEA and ICRU/AAPM), and present an adapted and simplified ICRU/AAPM methodology to measure the 3D kV dose of the clinical presets with a CTDI phantom and a conventional Farmer type ionisation chamber, aimed at straightforward clinical implementation.

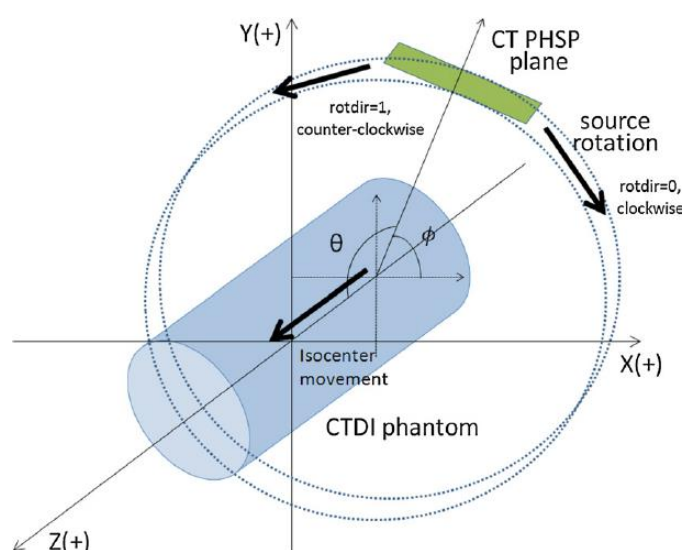


Figure 4.4. Schematic overview of the coordinate system when measuring with a CTDI phantom.

#### 4.3.2 3D kV dose measurements in CTDI phantom following the IAEA method

The IAEA published a report in 2011 [30] on the status of CT dosimetry for wide cone-beam scanners. The report recommends to correct the CTDI measured with a 100 mm pencil chamber in a standard 150 mm long phantom for a reference beam width of typically 20 mm by the ratio of the CTDI free-in-air for the clinical beam width to the CTDI free-in-air for the reference beam width. Note that measurement of the clinical presets following the IAEA method underestimates the dose, because scatter induced in the phantom by the larger field

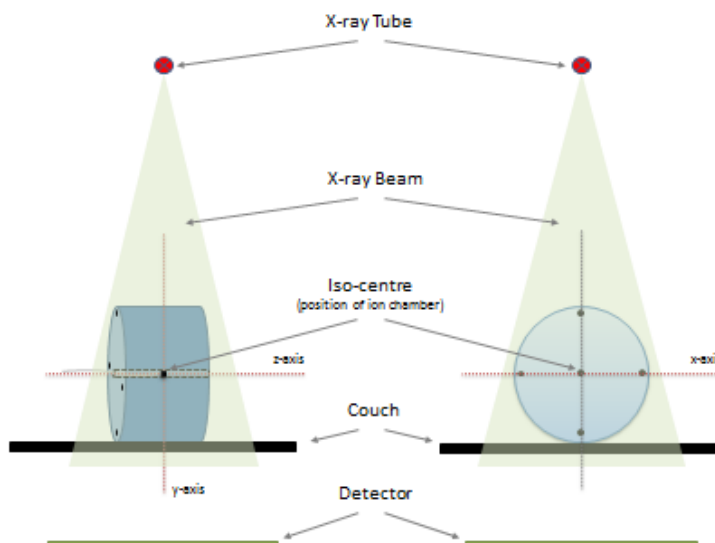
size of the clinical preset is not taken into account. Moreover, for the CTDI free-in-air measurement of the clinical beam, the whole beam width has to be measured. Hence, multiple measurements that include movement of the detector are required if the beam size is larger than the pencil chamber (100mm).

Following the IAEA method, free-in-air measurements as described in section 4.2 have to be performed for the clinical preset under two conditions: 1) with the clinical collimator, and 2) using the 20 mm collimator.

For the CTDI phantom measurements the 20 mm collimator is used. Note, that a 20 mm collimator is currently also available for Elekta XVI. By measuring in the central and peripheral holes of the CTDI phantom, the weighted CTDI can be calculated using:

$$CTDI_w = \frac{1}{3}CTDI_{\text{central}} + \frac{2}{3}CTDI_{\text{peripheral}} \quad (4.1)$$

Figure 4.5 shows the setup of the CTDI phantom and ionization detector on the couch.



*Figure 4.5. Schematic overview of the setup of the ionization detector and CTDI phantom on the couch.*

The equation to calculate the CTDI, after collecting all the measurements (one to three free-in-air measurements for the clinical beam, one free-in-air measurement for the reference

beam, one central CTDI phantom measurement, and four peripheral CTDI phantom measurements), is given below:

$$CTDI_{100,N \times T} = \frac{1}{(N \times T)_{ref}} \times \left( \int_{-50mm}^{+50mm} D_{ref}(z) dz \right) \times \left( \frac{CTDI_{free-in-air,N \times T}}{CTDI_{free-in-air,ref}} \right) \quad (4.2)$$

With  $(N \times T)_{ref}$  the nominal beam width of the reference beam.

#### **4.3.3 3D kV CTDI phantom dose measurements following the ICRU/AAPM method**

The AAPM and ICRU reports describe the usage of a small Farmer type ionisation chamber. AAPM TG 111 explains a theoretical method of measuring dose in axial, helical fan or CBCT systems with and without couch translation. ICRU 87 endorses this method and has added a method of measuring the equilibrium curve by means of real-time detector (a small thimble ionization chamber) which is moved through the phantom during the measurement. This procedure results in a profile of the air-kerma rate as function of the z-position (longitudinal direction). From this profile, the rise-to-dose-equilibrium curve can be derived by the integration over different lengths. Measurements obtained with this method are in excellent agreement with AAPM TG 111 based results [28].

For CBCT systems it is not practical to change the scan lengths or moving the detector. It was demonstrated by Dixon et al. [31] that the dose on the central plane of the cone-beam is the logical and unique choice for a stationary CBCT dose descriptor consistent with the CTDI-based dose of conventional CT.

AAPM TG 111 [28] also notes that the most appropriate method to determine the dose delivered by a CBCT system, with a beam width that is typically much longer than the ionization chamber length, is a measurement of the dose in the axial plane in the centre of the kV field. This is in analogy with CTDI which also expresses the dose as the average dose across the central axial plane.

Furthermore, the ICRU/AAPM method proposes to measure  $D_w$ , which is the weighted dose (see equation 4.3) at midpoint  $z=0$  measured in all holes of the CTDI phantom (see figure 4.6).

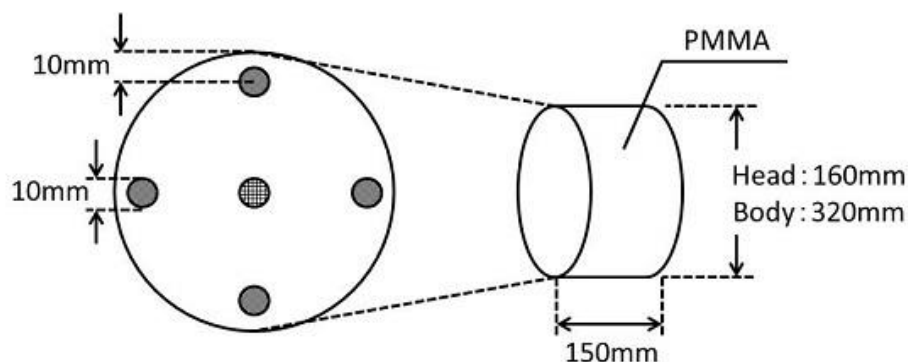


Figure 4.6. Position of the central and peripheral holes in an CTDI phantom.

Similar to  $CTDI_w$ ,  $D_w$  reflects the variation of dose deposition at depth by differentially weighting peripheral ( $D_p$ ) and central ( $D_c$ ) doses measured in standard CTDI phantoms, which is described as:

$$D_w = \frac{1}{3}D_c + \frac{2}{3}D_p \quad (4.3)$$

Since the phantom size is only 15 cm and the beam size of the CBCT can be larger, the measured dose in a standard sized CTDI phantom will be underestimated due to the lack of scatter.

#### 4.3.4 Simplified method to measure CTDI equivalent dose for clinical presets

We propose to measure with a calibrated Farmer chamber (see figure 4.7) as suggested by AAPM TG 111 and ICRU 87, whilst acknowledging that this method will not account for the dose in the penumbral region. Measurements should be performed during commissioning of the CBCT and compared with the reference data of the manufacturers. However, not all manufacturers provide reference values for comparison. If this is the case, we suggest creating a baseline after checking if the X-ray parameters such as HVL, kVp, mA, and time per frame are within the manufacturer's specification.

Note that the large beam width associated with CBCT is typically larger than a standard sized CTDI phantom, hence the actual dose will be underestimated up to 30% due to the lack of scatter during the measurement. To overcome this, a longer CTDI phantom can be used that is comprised of two standard-sized CTDI phantoms or a custom phantom could be manufactured.



*Figure 4.7. Example of Farmer like ionization chamber.*

Procedure for dose measurements in a head/body CTDI phantom:

- Position the CTDI phantom at the isocentre and rotate it to align the cross mark on the surface of the phantom to the horizontal laser cross.
- Set up the electrometer:
  - Select the appropriate settings on the electrometer for the Farmer ion chamber for kV measurements.
  - Select the dose measurement mode.
  - Set the integration time to 120 seconds.
- Set up the Farmer ion chamber:
  - Put the Farmer ion chamber detector (with a PMMA cap) in the central hole of the phantom.
  - Put the supplied acrylic rods in all the other holes in the phantom.
  - Fix the Farmer ion chamber cable with tape to the couch to hold the Farmer ion chamber in position.
- Select the desired preset for the measurement.
- Record the measurement, correct for temperature and pressure effects, and convert the measurement to dose.
- Perform the measurement three times and calculate the average dose.
- Repeat the steps for the peripheral holes.

#### **4.3.4.1 Correction factor tables for the Elekta XVI and Varian OBI systems**

Because of the practical drawbacks of using an elongated phantom, the subcommittee created a table to translate the measurements in a normal CTDI phantom to an estimate of the dose in an elongated phantom. Absolute dose measurements at  $z = 0$  were done with a Farmer chamber in a normal and an elongated CTDI phantom for all possible preset combinations for an Elekta XVI system and a number of representative presets for the Varian OBI. Presets with a full 360° rotation for image acquisition were used to get a symmetric dose distribution in the CTDI phantom.

Measurements were done in both head and body CTDI phantoms, following the dose measurement procedure described in section 4.3.4. From these measurements the correction factors for central dose ( $Cf_{Dc}$ ) and total dose ( $Cf_{Dw}$ ) in a short and long phantom were calculated for 16 different XVI presets for the system in a head (H) and body (B)

phantom. These correction factors can be used to translate dose measurement results from a short phantom to an elongated phantom. Since the difference in dose can be up to 15% it is recommended to do this conversion.

Tables 4.5 and 4.6 show the averaged correction factor ( $C_f$ ) data of two institutes with Elekta XVI systems for all common clinical presets of the XVI. The maximum difference between the data of the two institutes was 5%. The purpose of these measurements is to create a table, which can be used to predict the dose values from a single CTDI phantom dose measurement and distinguish the different preset factors like energy (kV), filter (F), field of view (FoV), and length (L).

The measurements for the head phantom and the body phantom were separately normalized to one (marked grey). The measurements of the correction factor in the centre of the phantom were also added to be able to simply convert from central dose ( $D_c$ ) to total dose ( $D_w$ ) for any preset.

The equation to convert central dose from preset 1 to total dose in any other preset is given by:

$$D_{w\_long\_phantom\_preset\_y} = D_{c(short\_phantom\_preset\_1)} \times \frac{C_{f_{Dw(long\_phantom\_preset\_y)}}}{C_{f_{Dc(short\_phantom\_preset\_1)}}} \quad (4.4)$$

where  $C_{f_{Dc}}$  and  $C_{f_{Dw}}$  present the correction factors for central dose and total dose.

Using this approach, the central dose from preset 1 can be measured in the central position of a short CTDI phantom where all other presets can then be calculated by equation 4.4. For example, a measurement in the central hole for preset 1 can be converted to the total dose ( $D_w$ ) of preset 9 in a long head phantom by using the values  $C_{f_{Dc(short\_phantom\_preset\_1)}} = 0.86$  and  $C_{f_{Dw(long\_phantom\_preset\_9)}} = 0.67$  from table 4.5 in equation 4.4.



*Table 4.5. Correction factors (Cf) between the short and long CTDI head phantoms for all common clinical presets of the Elekta XVI system.*

Preset					Short head phantom		Long head phantom	
No.	FoV	Length (cm)	Energy (kV)	Filter	Cf <sub>Dc</sub>	Cf <sub>Dw</sub>	Cf <sub>Dc</sub>	Cf <sub>Dw</sub>
1	S	10	100	F0	0.86	1.00	0.87	1.00
2	M	10	100	F0	0.84	0.92	0.85	0.92
3	S	20	100	F0	0.91	1.04	1.01	1.11
4	M	20	100	F0	0.88	0.95	0.98	1.01
5	S	10	120	F0	1.47	1.66	1.49	1.68
6	M	10	120	F0	1.43	1.52	1.44	1.53
7	S	20	120	F0	1.55	1.73	1.73	1.85
8	M	20	120	F0	1.50	1.58	1.67	1.69
9	S	10	100	F1	0.63	0.67	0.63	0.67
10	M	10	100	F1	0.62	0.62	0.63	0.62
11	S	20	100	F1	0.66	0.70	0.73	0.75
12	M	20	100	F1	0.65	0.64	0.71	0.69
13	S	10	120	F1	1.12	1.17	1.13	1.17
14	M	10	120	F1	1.09	1.08	1.10	1.09
15	S	20	120	F1	1.18	1.22	1.31	1.31
16	M	20	120	F1	1.14	1.12	1.27	1.21

*Table 4.6. Correction factors (Cf) between the short and long CTDI body phantoms for all common clinical presets of the Elekta XVI system.*

Preset					Short body phantom		Long body phantom	
No.	FoV	Length (cm)	Energy (kV)	Filter	Cf <sub>Dc</sub>	Cf <sub>Dw</sub>	Cf <sub>Dc</sub>	Cf <sub>Dw</sub>
1	S	10	100	F0	0.46	0.99	0.50	1.00
2	M	10	100	F0	0.42	0.83	0.44	0.84
3	S	20	100	F0	0.51	1.03	0.66	1.13
4	M	20	100	F0	0.45	0.87	0.58	0.94
5	S	10	120	F0	0.82	1.63	0.89	1.65
6	M	10	120	F0	0.74	1.36	0.79	1.38
7	S	20	120	F0	0.91	1.71	1.19	1.89
8	M	20	120	F0	0.81	1.42	1.04	1.56
9	S	10	100	F1	0.33	0.55	0.35	0.56
10	M	10	100	F1	0.30	0.47	0.32	0.48
11	S	20	100	F1	0.36	0.58	0.46	0.65
12	M	20	100	F1	0.33	0.50	0.42	0.55
13	S	10	120	F1	0.61	0.96	0.65	0.98
14	M	10	120	F1	0.55	0.82	0.59	0.84
15	S	20	120	F1	0.67	1.02	0.87	1.14
16	M	20	120	F1	0.60	0.87	0.77	0.96

For the Varian OBI system, three preset protocols [32] were measured with a Farmer chamber in a short and long body phantom; pelvis, low-dose thorax and pelvis spot light. The correction factors are given in table 4.7. The Pelvis and Low-Dose Thorax presets are comparable to presets 14 and 12 of the Elekta system, respectively. For all presets, the mAs settings of the pelvis preset were used. The data was normalized to the pelvis data.

*Table 4.7. Correction factors (Cf) between the short and long CTDI body phantom for some clinical presets of the Varian OBI system.*

Varian preset	Energy (kV)	Fan type	Bowtie filter	Short body phantom		Long body phantom	
				Cf <sub>Dc</sub>	Cf <sub>Dw</sub>	Cf <sub>Dc</sub>	Cf <sub>Dw</sub>
Pelvis	125	Half	Half	0.67	0.97	0.71	1.00
Low-Dose Thorax	110	Half	Half	0.49	0.75	0.57	0.76
Pelvis Spot Light	125	Full	Half	0.61	0.87	0.70	0.91

Using these correction factors in combination with measurements in a short phantom, an accuracy to within 10% of the specification of the manufacturer should be attainable.

*Table 4.8. QA recommendations for 3D kV imaging dosimetry (# A: acceptance; U: upgrade; S: service; An: annual).*

Report	Frequency #	Tolerances
NCS “CBCT QA”	AUSAn	Within $\pm 10\%$ of manufacturer’s specifications or baseline
AAPM TG 104 [18]	AnS	baseline
AAPM TG 179 [3]	AnU	baseline

#### 4.4 Summary

The X-ray radiation output and 2D and 3D kV dosimetric quality assurance procedures that were discussed in this chapter are summarized in table 4.9.

*Table 4.9. Summary of suggested tests (# A: acceptance; U: upgrade; S: service; An: annual).*

Test	Tolerance	Frequency #	See
<i>Radiation output</i>			
Kilovoltage peak accuracy and Half-Value Layer value	kVp $\pm (5\%)$ of intended kV HVL $\pm 5\%$ of mm Al compared with manufacturer’s specifications.	AUSAn	§4.1.1
Timer accuracy and linearity	Expected correlation coefficient of $1 \pm 0.05$	AUSAn	§4.1.2
Tube current accuracy and linearity	Expected correlation coefficient of $1 \pm 0.05$	AUSAn	§4.1.3
<i>2D kV imaging output</i>			
2D dose in air	Within $\pm 5\%$ of manufacturer’s specifications or baseline	AUSAn	§4.2
<i>3D kV imaging dosimetry</i>			
3D dose in a head and body CTDI phantom	Within $\pm 10\%$ of manufacturer’s specifications after application of correction factors or baseline	AUSAn	§4.3.4.1

## 5 Image quality and artefacts

### 5.1 Image quality

The image quality of planar imaging and CBCT's is usually measured by means of a comparison to baseline values [3,15,18]. The baseline values may be equal to the values that the vendor defines in its acceptance procedure, but we recommend the user to define baseline values for imaging protocols that are more often used in clinical practice. The purpose of a comparison with baseline values is to detect a possible deterioration of image quality before this hampers clinical practice.

Image quality aspects which can be taken into account are:

- Spatial resolution,
- Low-contrast detectability (signal-to-noise ratio),
- Image uniformity,
- CT-numbers (CBCTs),
- Signal linearity (planar imaging),
- Geometrical features (scaling, orientation, geometric distortion).

As a minimum we recommend to perform subjective tests (visual assessment of image) in comparison with the baseline values. The advantage of such observations is that they are fast and pragmatic. Another advantage of subjective methods is that the performance of the monitor and the viewing environment are taken into account, which is not the case when you perform an image analysis on the digital image itself.

One can extend the QA programme with more quantitative tests. For an objective analysis of the spatial resolution the Modulation Transfer Function (MTF) can be determined [21,33,34]. The low-contrast detectability depends on the signal-to-noise ratio, and an objective measurement consists for example of measuring the signal-to-noise ratio directly [21] or the noise power spectrum [34].

### 5.1.1 Planar imaging

#### 5.1.1.1 Spatial resolution

A subjective test for spatial resolution is to measure the number of line pairs visible on a planar image of a phantom. Figure 5.1 shows different phantoms (a-5,b-1,d-1) which can be used for this purpose.

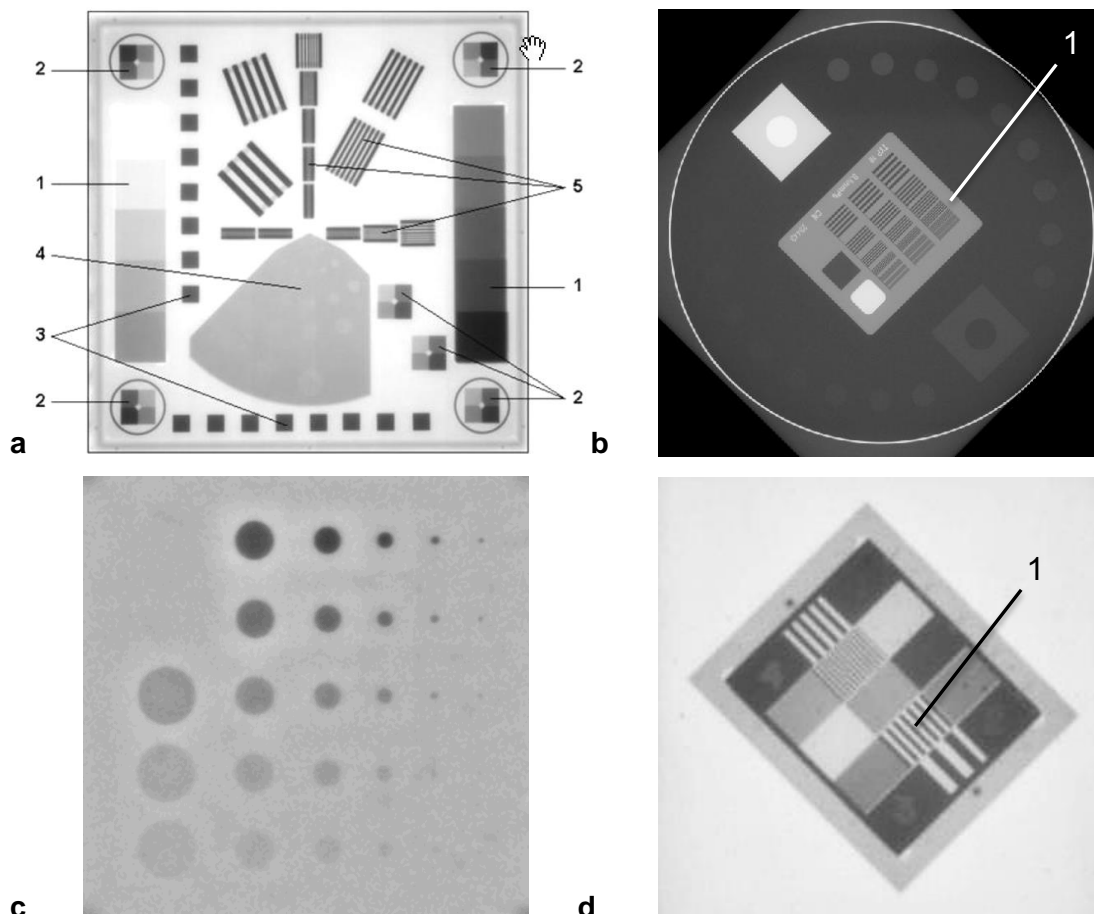


Figure 5.1. Examples of phantoms to be used for image quality QA on planar images: (a) PTW EPID QC, (b) Leeds (kV), (c) Las Vegas (MV), (d) PIP.

#### 5.1.1.2 Low contrast detectability

The low-contrast detectability can be measured by imaging a phantom and determining the just visible object with a given low contrast. Figure 5.1 shows some phantoms which can be used for this purpose (a-4,c) [35-37]. If the holes with a given contrast and size that are just visible are connected by a line, one constructs a contrast-detail curve, which makes it easy to do a comparison with the baseline low contrast detectability contrast-detail curve.

### 5.1.1.3 Uniformity

The uniformity in an image can simply be measured by taking a flood-field image. Qualitatively, this image can be evaluated by judging the uniformity of the image. Quantitatively, one can compare the pixel values in different areas of the image. These images can also be used to determine the noise level in the image by looking at the variation of the pixel values. However, when larger areas are used to determine the noise level, care has to be taken that the image within the area is uniform.

### 5.1.1.4 Geometrical features

Scale and orientation can simply be determined by measuring the size of an object with a known size and orientation in the image. An automatic field edge registration will inherently check scaling and orientation for each image. Errors in orientation will show up in clinical practice and are therefore not needed in a regular QA programme. However, orientation and scaling should be checked after updates to hardware and/or software. Image distortion is only expected in camera based systems and can be determined by imaging a phantom such as in fig. 5.1a item 3.

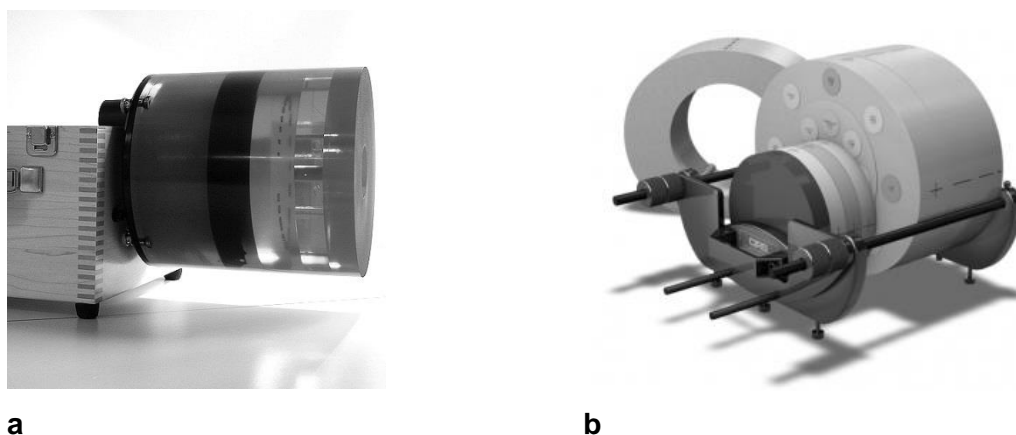
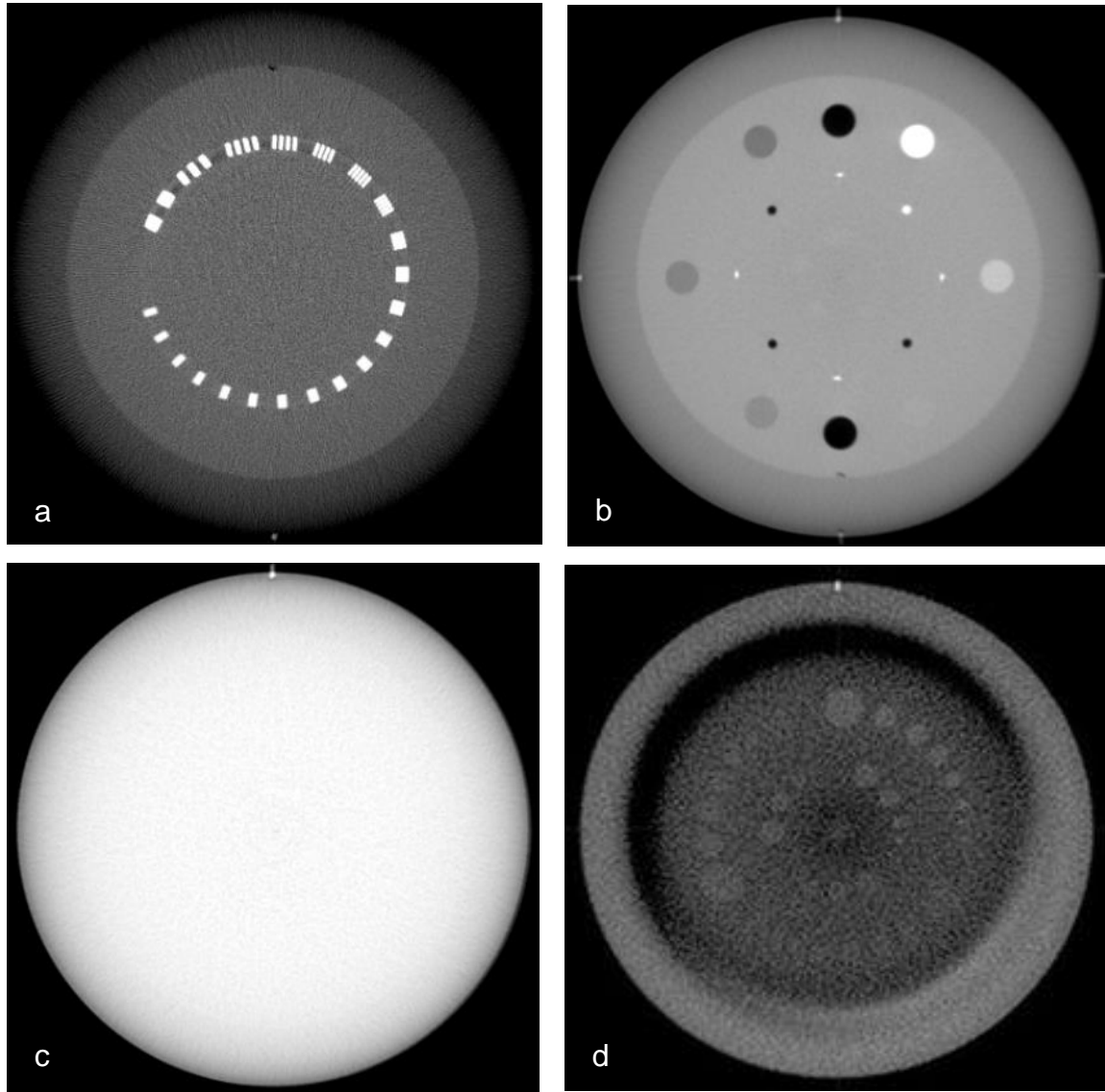


Figure 5.2. The Catphan® 500 series (a) and the CIRS 062MQA (b) phantoms.

### 5.1.2 kV CBCT imaging

Image quality depends on the settings that are used to generate the CBCT, therefore QA should always be performed with representative settings. By making a scan of a phantom such as the Catphan® 500 phantom (The Phantom Laboratory, Salem, NY, figure 5.2a) or the CIRS 062MQA (CIRSINC, Norfolk, VA, figure 5.2b) one can measure the image quality aspects of the CBCT in different sections of the phantom (see figure 5.3). Usually during acceptance testing of the kV CBCT imaging system the phantom is used for image quality

measurements as well [6,8] and therefore the acceptance test results can be used as reference values to detect any alterations in the image quality of the kV CBCT system over time.



*Figure 5.3. Different image quality sections of the Catphan® 500 series phantom: (a) spatial resolution, (b) scaling, and Hounsfield Units, (c) uniformity and noise, (d) low-contrast detectability.*

The kV CBCT provides quantitative CT-numbers (Hounsfield Units), but small deviations from acceptance conditions may affect CT-number significantly due to the large scatter

contributions [3]. Therefore, CT-numbers should be calibrated for representative geometries and acquisition parameters.

When the CBCT is to be used in dose calculations, one should consider the following aspects in a QA programme:

- Check CT-number accuracy for all representative geometries and acquisition parameters, since CT-numbers depend on geometry, acquisition parameters and energy.
- Desired accuracy of dose calculation (and resulting desired accuracy of CT-numbers) depends on the clinical use.

In case that CBCTs are used for treatment planning, we refer to appendix A.2.1 of NCS report 15 [38] for suggestions on how to convert CT-numbers to relative electron densities that are needed for dose calculations in a treatment planning system. The NVKF recommends in their report on QA of CT scanners [39] to check the CT-numbers of conventional CT scanners that are used for radiotherapy treatment planning at acceptance, weekly and after replacement of the X-ray tube. The CT-numbers should be checked with a phantom with a water insert. The calibration of CT-numbers for water and air are, by definition, independent of the kV used. The CT-number of water should not deviate more than 4 HU from 0 HU. Furthermore, CT-numbers of other relevant materials like muscle, fat, lung and bone should be checked. These checks should be performed for all used tube voltages because the CT-numbers depend on the kV.

## 5.2 Recommendations

The frequencies mentioned in this section are based on proven stability of the system at about half a year after commissioning. The frequencies and the tests mentioned are recommendations for periodic QA.

Obvious sudden deterioration of an imager will probably be detected by the clinical users of the panel and not by periodic QA. Complaints about the quality of clinical images should trigger an image quality measurement. QA measurements should be performed after repair, maintenance, or upgrades of hard- and/or software that could affect image quality. After each system adaptation one should consider on a case by case basis the extend of image quality tests needed.

### 5.2.1 Planar imaging

AAPM TG 142 [15] recommends to measure image quality monthly and does not recommend a measurement frequency for the geometric distortion.

#### 5.2.1.1 aSi flat panel imagers (MV and kV)

Almost all problems with image quality in planar MV- and kV-imaging with aSi flat panel imagers can be recognized in flood-field images. Problems caused by bad pixels, inhomogeneity in the response of subpanels, synchronization of the readout of the panel with the linac or radiation induced damage of the electronics result in inhomogeneities in a flood-field image. Therefore, we recommend acquisition of flood-field images (uniformity) at least once every 3 months. Other image quality measurements are recommended annually and after acceptance, service and upgrades.

#### 5.2.1.2 Camera-based imagers (MV)

With camera-based imagers problems related to deterioration of the camera (bad pixels, radiation induced damage) can be recognized in flood-field images. However, problems related to the alignment and focusing of the mirror-lens-camera system will not always result in inhomogeneities in flood-field images. Therefore, for camera-based imagers the tests should be performed more frequently. We recommend to measure uniformity, spatial resolution, geometric isotropy, and low-contrast detectability monthly and after acceptance, service and upgrades.

*Table 5.1. Image quality measurements for planar MV- and/or kV-imaging with its recommended minimum intervals (# A: acceptance; U: upgrade; S: service; M: monthly, An: annual).*

Test	Frequency # AAPM TG142	Frequency # NCS (aSi MV/kV imagers)	Frequency # NCS (camera based MV imagers)	Tolerance NCS
Spatial resolution	M	An and AUS	M and AUS	Baseline**
Low-contrast detectability	M	An and AUS	M and AUS	Baseline**
Uniformity (and noise)	M	3M (MV) and AUS	M and AUS	Baseline**
Scaling	M	An and AUS	M and AUS	See section 3.3
Orientation	M	An and AUS	M and AUS	See section 3.3
Geometric distortion	-	-	M and AUS	Baseline**

\*\* Initial tolerance as specified by vendor



## 5.2.2 kV CBCT imaging

AAPM TG 142 [15] recommends to measure CBCT image quality monthly where AAPM TG 179 [3] recommends that most measurements can be performed on a semi annual basis after stability has been demonstrated (6-12 months after commissioning). This report also recommends to perform measurements on kV CBCT image quality on a semi annual basis. Because a deterioration of the image quality is likely to be noticed by the clinical users and because the geometric accuracy is measured more often (see section 3.2), it is not necessary to measure image quality every month. After maintenance or incidents that could influence panel or X-ray tube positioning, additional image quality QA should be performed.

Because noise will influence the low-contrast detectability, it is strictly not needed to determine it separately. However, when measuring uniformity it is straightforward to determine also the noise in this image.

There is limited clinical experience with using CBCT CT-numbers for dose calculation and the procedures and applications are very diverse and depend on the CBCT vendor and planning system used. Therefore, we do not recommend a specific frequency for QA on CT-numbers. The necessary frequency and tolerances should be determined locally based on a risk analysis of local procedures.

*Table 5.2. Image quality measurements for kV CBCT imaging with its recommended minimal intervals (# A: acceptance; U: upgrade; S: service; An: annual).*

Test	Frequency # AAPM TG179	Frequency # NCS (kV CBCT)	Tolerance NCS
Spatial resolution	½An	½An and AUS	Baseline**
Low-contrast detectability	½An	½An and AUS	Baseline**
Uniformity (and noise)	½An	½An and AUS	Baseline**
CT-numbers	½An	To be determined locally	To be determined locally
Scaling	½An	½An and AUS	See section 3.3
Orientation	½An	½An and AUS	See section 3.3

\*\* Initial tolerance as specified by vendor.

## 5.3 Image artefacts in kV projections and reconstructed CBCT images

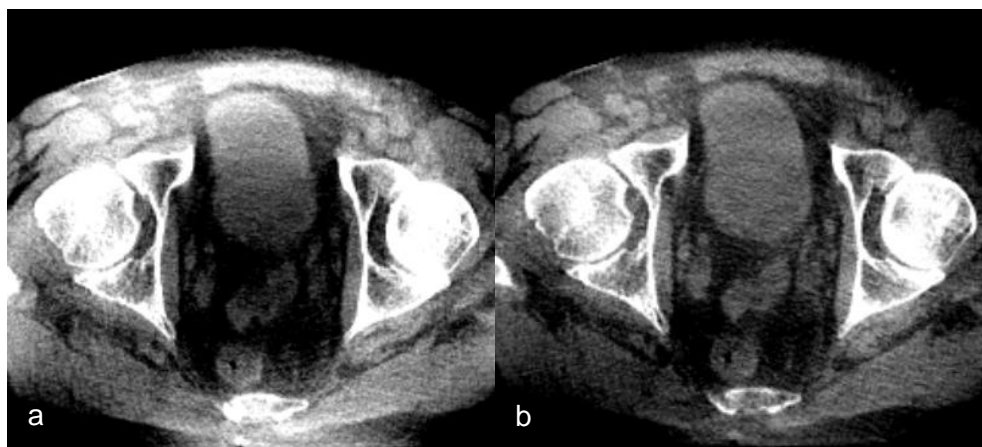
Various artefacts can appear in CBCT images during normal clinical operation or QA procedures. Some of these artefacts are related to the imaging and reconstruction method, properties of the amorphous silicon detectors or features of the object that is being imaged.

These artefacts cannot always be prevented. However, in some cases measures can be taken to minimize the artefacts. Other artefacts occur because of defects in the system hardware. It is important to recognize the origin of the artefacts to decide which action is necessary. In the next two sections we will distinguish the method/patient related artefacts from the artefacts caused by technical failures.

### **5.3.1 Method/patient related artefacts**

#### Cupping artefact

Cupping artefacts are a result of a combination of differences in beam hardening and scatter effects through an object [40,41]. As a result the centre of the image shows an unintended lower CT-number than expected (see figure 5.4). This effect can be minimized by software corrections and hardware solutions (e.g. bowtie filter).



*Figure 5.4. (a) Image with cupping artefact, (b) less perceptible cupping present after software correction.*

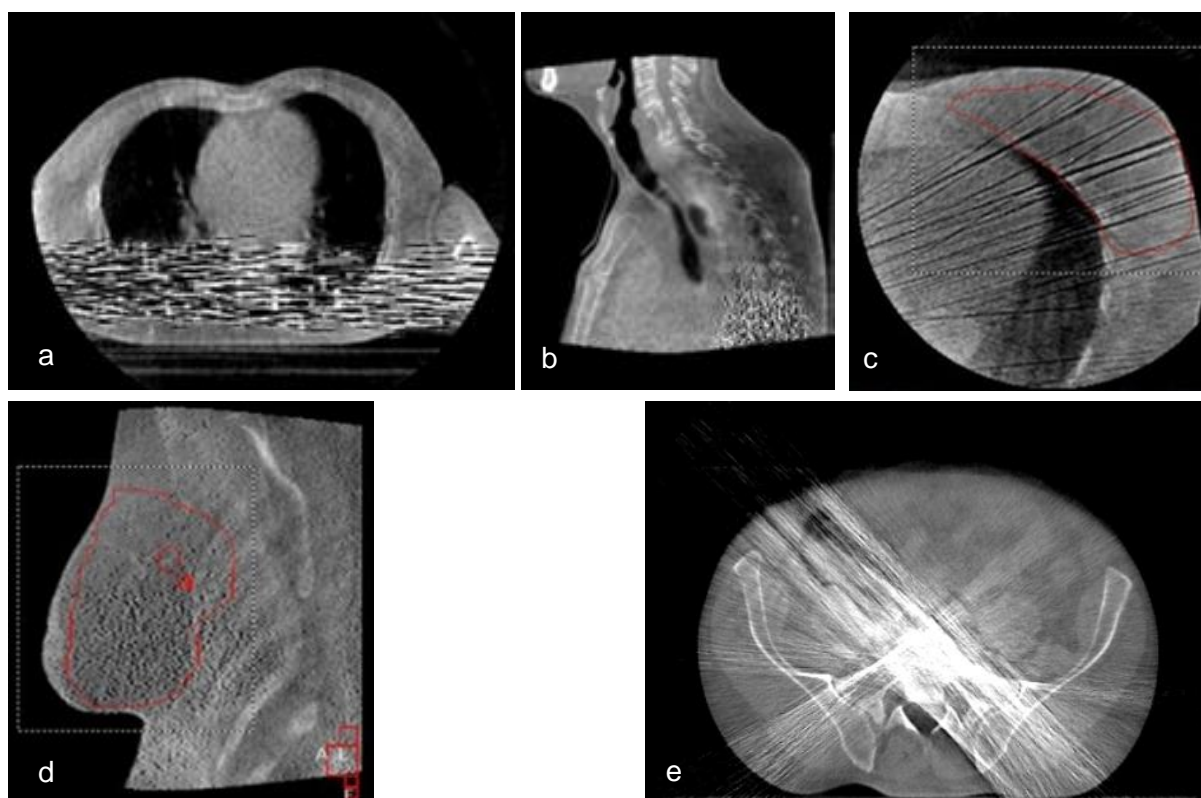
#### Streak artefacts

Streak artefacts in the images can have several causes:

1. Under-sampling/aliasing: A too low number of projection images is available for the reconstruction. For example in 4D images this may occur. By lowering the gantry speed the number of projection images can be increased, which minimizes the streak artefacts due to under-sampling.
2. Objects with high density that are not present in the field-of-view of all the projection images or that move between the different projection images can cause streaking artefacts (for example ribs in 3D CBCT scans that move due to patient breathing). These streaking artefacts can be minimized by increasing the field-of-view so that high-density objects are present in all the projection images or, in case of breathing motion, by

diminishing the breathing motion or combining images of one breathing phase only in the reconstructions.

3. Photon-starvation: in some regions of the detector no or very few photons are detected because of high density objects in the kV-beam. By increasing the kV or mAs of the image acquisition the effect of photon-starvation can be diminished. In figure 5.5 a-d several examples of photon starvation are shown.
4. Overcorrection by the scatter correction algorithm (Elekta): in rare cases the scatter correction algorithm in the Elekta software may correct too much (e.g. a too high Scatter to Primary Ratio in the reconstruction preset), resulting in “zeroed” areas in the projection images used in the reconstruction. These areas will show up as very noisy bands through the 3D image (see figure 5.5e).



*Figure 5.5. (a)-(d) Streak artefacts due to photon-starvation, (e) streak artefact caused by the scatter correction algorithm.*

#### Ghosting artefact

The ghosting artefact (see figure 5.6) is caused by the influence of previous projection image acquisitions on the sensitivity of the scintillator of the detector panel. This effect can cause an additional image visible in the projection images and will result in erroneous reconstructions.

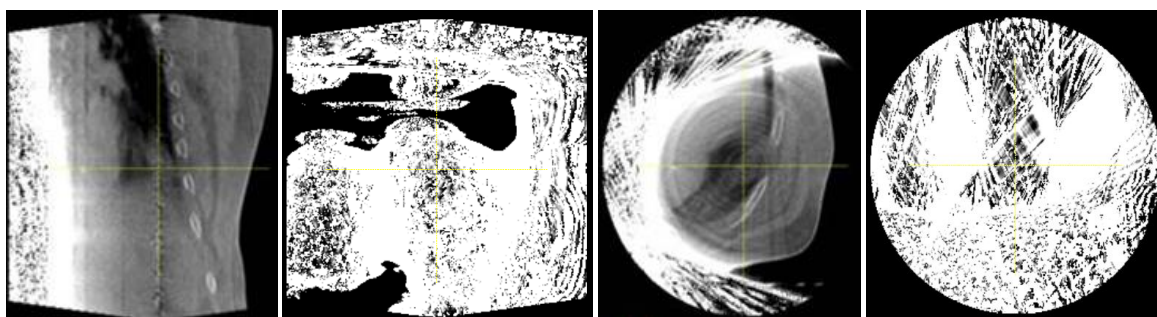
For planar imaging this artefact can be solved by waiting longer between successive images. For CBCT optimizing the acquisition parameters (e.g. changing the rotation direction or the start-stop angles) might reduce the influence of the ghosting artefact.



*Figure 5.6. Example of a ghosting artefact.*

#### Extremely noisy images

If CBCT images are acquired immediately after irradiation with MV-beams, it can occur that the CBCT images are very noisy (see figure 5.7). This noise is caused by the scattered MV-photons that are detected by the detector panel during acquisition of the background (darkfield) image. The solution is to wait one or two minutes between MV-irradiation and kV-image acquisition.



*Figure 5.7. Different examples of extremely noisy images.*

#### Patient outline reduction

High transmission of X-rays through thin and less dense areas of the patient can cause saturation of the panel. These saturated pixels are treated as “air” by the reconstruction algorithm, causing the outline to be reduced (see figure 5.8). Adjustment of the acquisition parameters (kV/bowtie filter) might reduce this effect. This effect is not always distinguishable from an actual patient outline reduction.

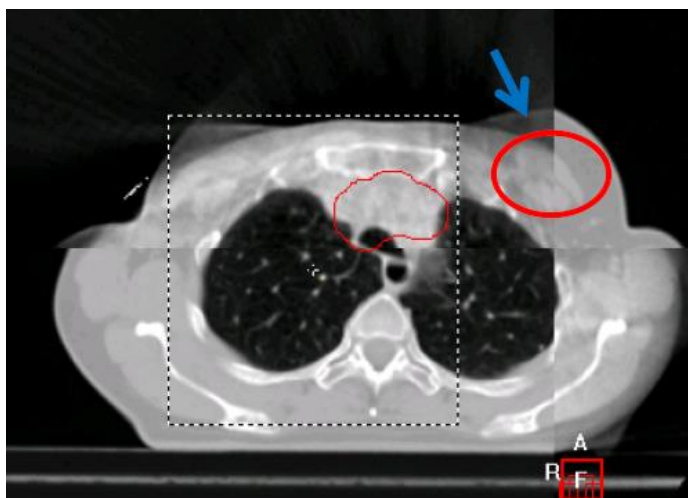


Figure 5.8. Image with patient outline reduction.

### 5.3.2 Artefacts caused by technical failures

#### Ring artefact

If clear ring artefacts are present in almost all reconstructed CBCT images, they are probably caused by bad pixels in the detector panel (see figure 5.9). The problem can be solved by making a new bad pixel map or by replacing the detector panel. Low contrast rings in the reconstructed CBCTs can be caused by ghosting of small structures from previous acquisitions in the projection images. Beware of scanning high contrast objects just before performing a gain calibration of the panel, as this may result in a similar problem for all images that are acquired subsequently using that calibration. A new gain calibration should be performed in that situation.

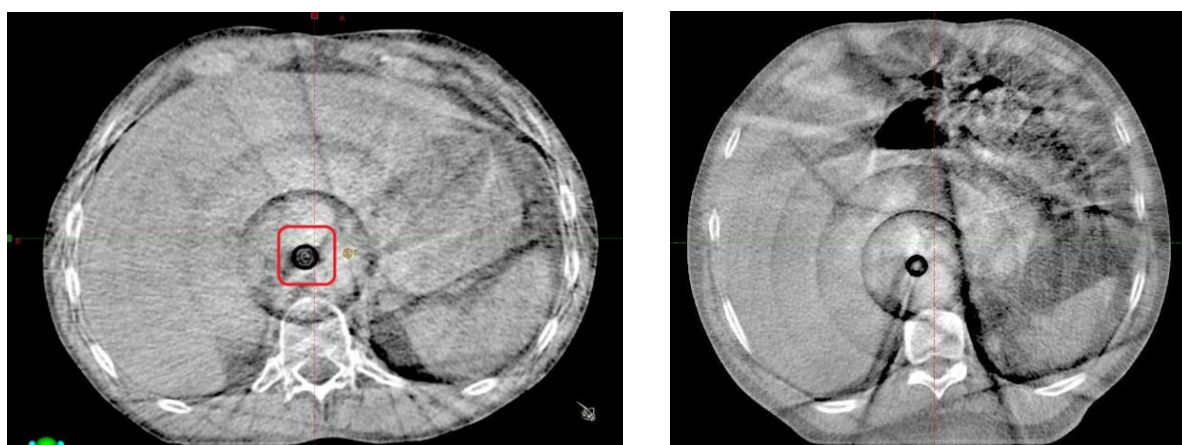


Figure 5.9. Examples of ring artefacts.

#### Bar artefact

The bar artefact (see figure 5.10) is probably caused by readout or correction problems of a subpanel. It is known to appear and disappear now and then. Until now there is no solution, except replacing the hardware.

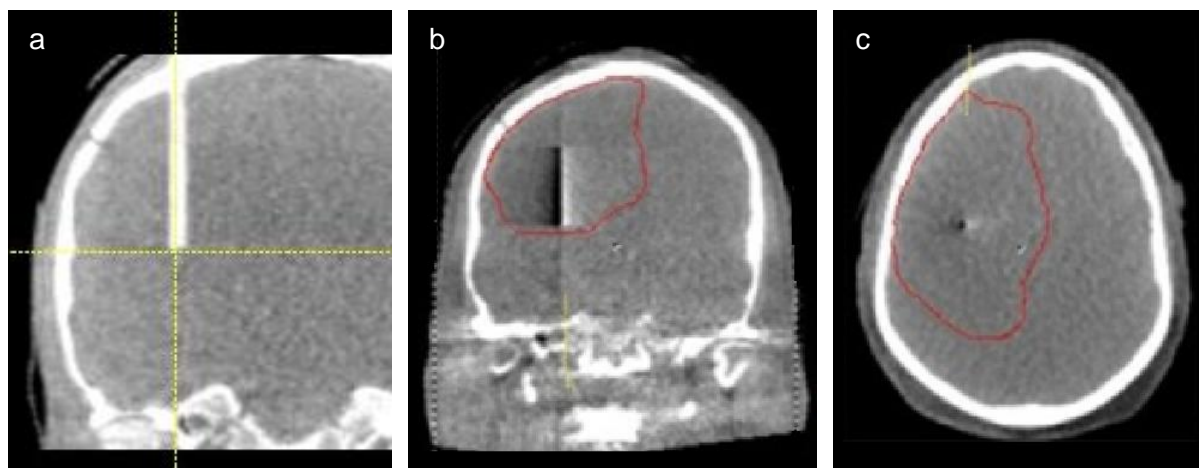


Figure 5.10. Example of a bar artefact: (a),(b) coronal view, (c) transversal view.

#### Panel edge artefact / halo artefact

If the kV panel is not well aligned with the kV tube, the halo artefact can appear (see figure 5.11). Part of the panel is not in the field. This can be solved by correcting the alignment of the panel with respect to the X-ray tube.



Figure 5.11. Transversal view of the halo artefact. Note that the white sphere in the middle is a ball-bearing phantom.



### Dark blot artefact

Air in the cooling oil of the X-ray tube can cause a dark blot, best seen when making an homogeneous image (see figure 5.12). Exchanging the X-ray tube will solve this problem.

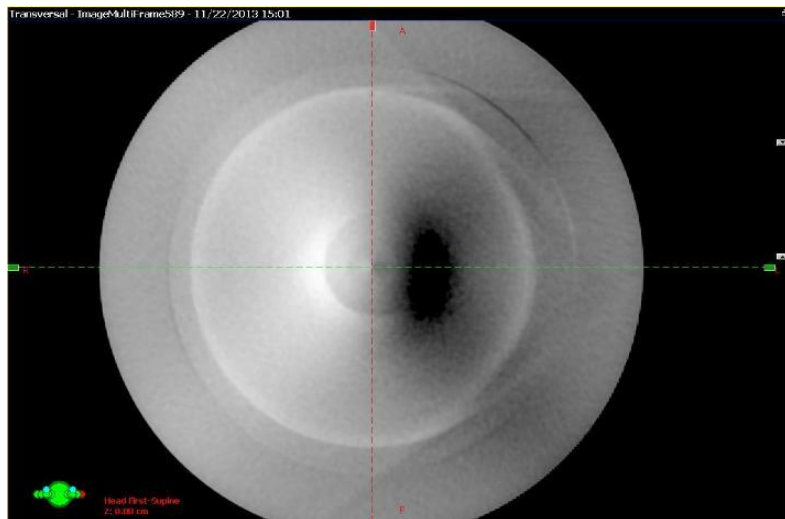


Figure 5.12. A scan of the uniformity part of the Catphan<sup>®</sup> phantom, showing the dark blot artefact.

### Speckle artefact

If the planar image shows a lot of speckles (see figure 5.13) it is most likely that the bad pixel map is not correct or not taken into account. Check whether the system settings are correct for accessing the bad pixel map and perform a new bad pixel map calibration if necessary.

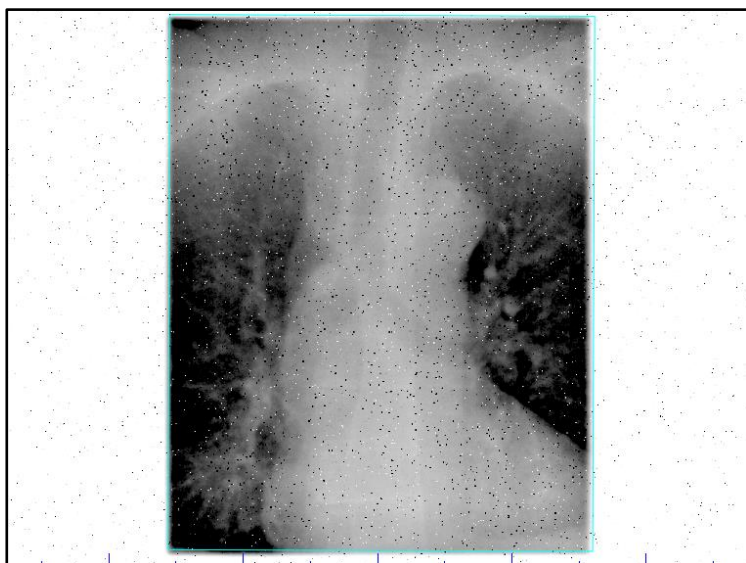


Figure 5.13. Planar image with speckles.

Block artefact

If one subpanel area is broken or not read out, a block artefact will be present (see figure 5.14). A new panel or new panel electronics will eliminate this artefact.

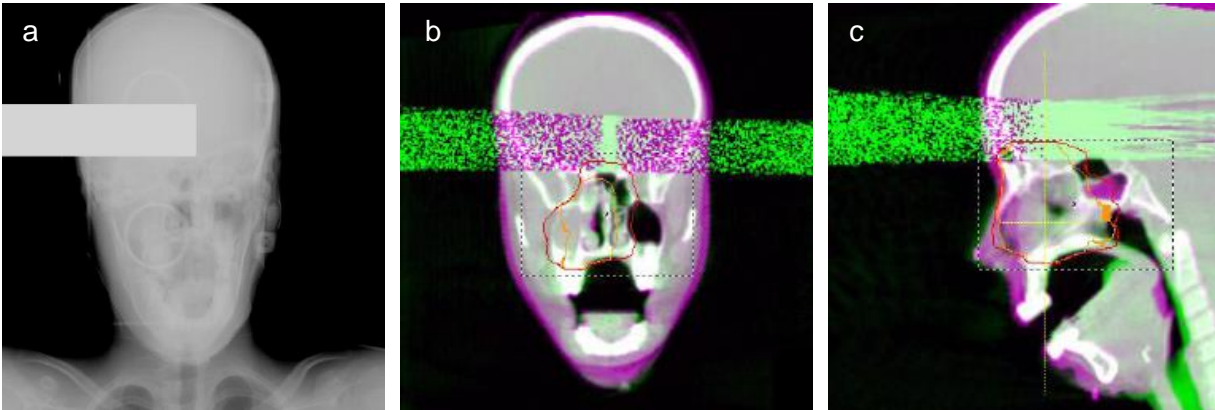


Figure 5.14. (a) A single projection image with the block artefact, (b),(c) different views of a CBCT reconstruction containing the block artefact.

5.4 Summary

All tests suggested in this chapter are summarized in table 5.3.

Table 5.3. Summary of suggested image quality tests (# A: acceptance; U: upgrade; S: service; M: monthly, An: annual).

Test	Tolerance	Frequency # (aSi MV/kV imagers)	Frequency # (camera based MV imagers)	Frequency # (kV CBCT)
Spatial resolution	Baseline**	An and AUS	M and AUS	½An and AUS
Low-contrast detectability	Baseline**	An and AUS	M and AUS	½An and AUS
Uniformity (and noise)	Baseline**	3M (MV) and AUS	M and AUS	½An and AUS
CT-numbers	To be determined locally	-	-	To be determined locally
Scaling	See section 3.3	An and AUS	M and AUS	½An and AUS
Orientation	See section 3.3	An and AUS	M and AUS	½An and AUS
Geometric distortion	Baseline**	-	M and AUS	-
See		§5.2.1.1	§5.2.1.2	§5.2.2

\*\* Initial tolerance as specified by vendor.



## 6 Safety

### 6.1 Mechanical safety

Linear accelerators with a kV CBCT and a MV EPID have, sometimes robotic, arms which can be close to the patient during the imaging procedure. Special care must be taken to avoid machine to treatment couch collisions, and moreover, machine to patient collisions as this may cause serious injuries to the patient. All machines have some kind of detection system to cause the machine to stop when a collision occurs. This can be hardware (touchguards) or some kind of laserguard detection system. These detection systems must be checked on a weekly basis [3,18]. Some machines have software to avoid machine to treatment couch collisions (Varian TrueBeam® platform). With the treatment couch centred the probability of a collision is low, but you have to be careful when a couch-shift, especially a lateral one, and/or when a couch rotation has been applied. Perform a dry run to check that there is no machine to patient collision when the imaging procedure is conducted.

#### Test procedure

- Check if the system detects a collision when activating a touch guard or laser guard.
- This should be done for all touch guards on the imaging systems.

#### Frequency

- Checks should be performed on a weekly basis.

#### Tolerance

- All touch guards must stop the machine movement upon activation.

*Table 6.1. QA recommendations for mechanical safety (# D: daily; W: weekly).*

Report	Frequency #	Tolerance
NCS "CBCT QA"	W	Functional
NCS report 8 [42] & 9 [43]	W	Functional
AAPM TG 104 [18]	D	Functional
AAPM TG 179 [3]	D	Functional

## 6.2 Radiation safety

### 6.2.1 Interlocks and warning lights

The treatment room should be equipped with room door interlocks and associated warning lights that are also functioning for kV imaging. The functionality of these interlocks must be tested.

#### Test procedure

- Open the treatment room door and check if the kV beam stops.
- Check if the radiation warning light is “ON” during the kV imaging procedure.

#### Frequency

- Weekly.

#### Tolerance

- kV beam should be interrupted when opening door.
- Warning light should be “ON” during operation.

*Table 6.2. QA recommendations for room door interlocks and warning lights (# D: daily; W: weekly).*

Report	Frequency	Tolerance
NCS “CBCT QA”	W	Functional
AAPM TG 104 [18]	W	Functional
AAPM TG 179 [3]	D	Functional

### 6.2.2 Radiation leakage

A measurement of the radiation leakage from the kV source should be performed by either the hospital or the manufacturer.

#### Test procedure

- Use a gantry angle such that the kV source is at 0-degrees.
- Select the smallest field size possible with maximum kV and typical mAs settings.
- Put a suitable detector at 1 m SDD in isocentre.
- Measure the output in the isocentre and at 100 cm distance from the isocentre in the GT and AB directions.
- Additionally, wrap the kV head in radiochromic film to detect leakage spots.

### Frequency

- At acceptance of the machine and after replacement of the X-ray tube.

### Tolerance

- At 100 cm distance from the isocentre in the GT and AB directions, the leakage dose should be less than 1‰.

*Table 6.3. QA recommendation for radiation leakage (# A: acceptance; U: upgrade).*

Report	Frequency #	Tolerance
NCS “CBCT QA”	AU	@100cm distance from isocentre in GT or AB direction <1‰

The described radiation leakage test is not mandatory, but if the hospital decides not to perform this test, the hospital has to obtain proof of the leakage measurements from the vendor of the machine.

## 6.3 Recommendations

*Table 6.4. Summary of suggested safety tests (# A: acceptance; U: upgrade; W: weekly).*

Test	Tolerance	Frequency #	See
<i>Mechanical safety</i>			
Verify touch- or laserguards	Functional	W	§6.1
<i>Radiation safety</i>			
Verify radiation interlocks and radiation warning lights	Functional	W	§6.2.1
Measure radiation leakage	<1‰	AU	§6.2.2

## 7 4D CBCT

### 7.1 *Basic principles*

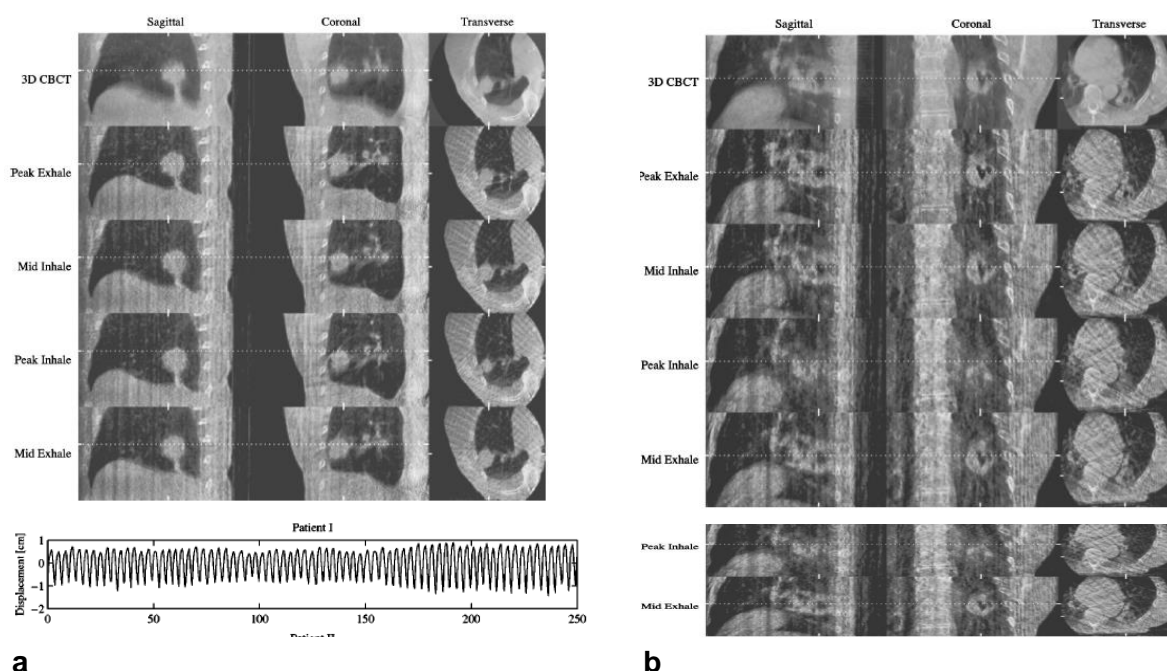
Current implementations of 4D CBCT rely on sorting of the projection data based on the breathing phase. The Varian solution relies on an external recording of the breathing signal and retrospective reconstruction, which is limiting it to offline use. In the case of Elekta's XVI, software extracts the respiratory signal by tracking diaphragmatic and other breathing correlated movement from the actual projection data and does not rely on external monitoring devices. After sorting the projection data into the different breathing phases, 3D images are reconstructed for the different phases using the same algorithm that is utilized for the 3D reconstructions, and subsequently concatenated to form the 4D scan [44].

Because the number of projections per phase is generally much smaller than the number of projections in a typical 3D scan, 4D CBCT scans usually suffer from more noise and undersampling/aliasing artefacts. However, image quality is typically sufficient for image guidance of lesions in the lung.

There are many aspects of the reconstruction, registration and correction processes that are shared between the 3D and 4D methods, hence duplicating the QA procedure for 4D workflows is not required for most steps. In the following sections the major differences and their consequences for QA will be discussed.

### 7.2 *Image quality*

The image quality of the individual phases of 4D CBCT is governed by the same parameters as 3D CBCT, i.e. dose, spatial resolution, reconstruction parameters, etc. Specifically for 4D CBCT, the image quality can be severely deteriorated by irregular breathing of the patient (see figure 7.1), but unfortunately this cannot be prevented by QA measures. Therefore, QA of the image quality in 4D CBCT using phantom measurements will have limited scope.



*Figure 7.1. The effect of breathing irregularities on image quality: (a) a patient that is breathing regularly, (b) an irregularly breathing patient [44].*

Since 4D reconstruction is basically a 3D reconstruction algorithm combined with a sorting method that allocates projections to phases, the recommendation for 4D image quality QA is to validate whether 4D and 3D reconstructions will yield similar results when given the same effective input. Practically, this will entail scanning a 4D phantom in a stationary position with a 3D protocol and with a 4D protocol whilst moving. By customizing the protocols in such a way that the total number of projection images and mAs in the 3D reconstruction is the same as the number of projections and mAs in one phase of the 4D scan, the image quality of the 4D scan can be visually validated against the 3D “golden standard”. This test would reveal inadequacies in both reconstruction and sorting algorithms.

### 7.3 Geometric accuracy and distortions

Since the same reconstruction engine is used for both 3D and 4D CBCT, in principal no specific tests in addition to the 3D QA are required. However, due to breathing motion additional blurring can occur, which may result in degraded geometrical accuracy. To estimate these effects, scans of a 4D phantom with varying amplitudes could be used. One specific test that can be performed for this purpose will be discussed in section 7.5.

#### *7.4 Registration accuracy for 4D CBCT*

For small breathing amplitudes, the registration accuracy between 3D or 4D CBCT, and an anatomically correct 3D planning CT (i.e. breath-hold or mid-position) will typically be similar. For larger amplitudes (above about 8 mm, see reference), the accuracy of the registration with an anatomically correct 3D planning CT will become worse for 3D CBCT due to the poorly defined edges of a moving object in 3D CBCT [45]. In this case, either a registration with a similar “average” reconstruction of the planning CT, or a 4D registration with a multi-phase 4D reconstruction of the CBCT acquisition is advised.

As was noted as well for image quality, the registration accuracy will be patient dependent. Especially for patients with irregular breathing patterns, where image quality will be poor, 4D registration accuracy may suffer. In these cases one option would be to postpone the treatment and coach the patient in obtaining a more regular breathing pattern. That failing, a fall back to 3D CBCT in combination with manual registration, or 3D CBCT in combination with an “average” reconstruction of the planning CT are other options.

Visual validation of registration results should always remain mandatory. The success rate of registration algorithms is strongly dependent on the image quality of the input data, which means that there will be non-negligible failure rate. Since the image quality for 4D CBCT is usually worse than 3D this argument becomes even more important when registering to the individual phases of a 4D CBCT.

#### *7.5 Comparison between 3D and 4D IGRT workflow*

Because most of the processing steps in 4D CBCT are shared with its 3D counterpart, additional QA is not required. However, since this knowledge is solely based on data from literature [44,45] and the manufacturers, it is recommended to perform a comparison test between the 3D and 4D IGRT workflows.

The recommended protocol is limited to validating that the results from a complete IGRT procedure (i.e. scan, register, correct) are identical when using 3D versus (multi-phase) 4D CBCT, or 3D versus average (4D) CBCT. To perform this protocol, a 4D phantom is required (e.g. CIRS InSight, Modus Quasar, or a home built phantom, see figure 7.2). The phantom should be able to move in a “breathing like” pattern, and be able to be positioned stationary along arbitrary points on a breathing pattern chosen by the user.

In this procedure, the method for acquiring an anatomically correct 3D reference CT scan, e.g. through a breath-hold, an extreme phase, average, or mid-ventilation technique, is not included. It is recommended to implement a separate quality assurance programme for this.

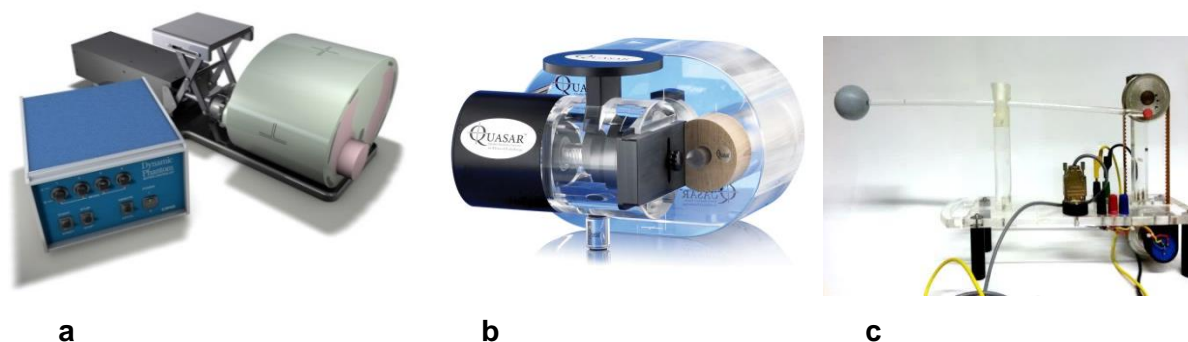


Figure 7.2. (a, b) The CIRS InSight, and Modus Quasar phantoms which can be programmed to follow arbitrary tumour trajectories at different frequencies; (c) A simple home-built phantom.

Clinically, two methods are commonly used to deal with breathing motion: Either an anatomically correct 3D planning CT is registered to one or all phases of a 4D CBCT, or an average 3D CT is registered to an averaged 4D or average 3D CBCT. The actual tests that follow below will cover both strategies.

#### Preparation for 4D registration workflow

- Acquire a 3D CT scan of the phantom with the tumour in a static position, using a protocol that is standard for radiotherapy planning in your hospital. Deviations from the true average position will show up as a setup error during the actual test.
- Transfer the scan-data to the planning system.
- Delineate the tumour in the phantom as GTV (as a structureset may be required to be able to import the plan in the image guidance system), place the isocentre in the centre of the GTV, and construct a simple plan with one or two fields (this will enable import of the plan and scan in the image guidance system).
- Transfer the reference CT, plan and structures to the linac and the IGRT system.

#### At the linac

- Position the phantom on the treatment couch in the same way it was aligned on the CT scanner. Correcting the setup error is not necessary as the test only looks at differences between 3D and 4D workflows.

- Select a constant-speed sinusoidal tumour movement of e.g. 2 cm in the cranial-caudal and 1 cm in the anterior-posterior direction, resulting in an ellipsoidal movement with a repetition cycle of 4-5 seconds.
- Acquire a CBCT scan of the stationary phantom with the tumour at its most cranial, posterior, caudal and anterior position using a 3D protocol.
- Acquire a CBCT scan of the moving phantom using a 4D protocol.
- Register the four 3D CBCT scans and the 4D CBCT scan on the tumour.
- Calculate the average translation from the four 3D CBCT scans.
- The actual values will reflect the deviation of the average of the tumour trajectory from the position in the planning CT scan.
- The difference between the average 3D translation and the result from the 4D registration reflects deviations of the 4D reconstruction/registration algorithm and should be less than or equal to 1 mm in all directions.
- Optionally, the test could be repeated with other amplitudes and/or different speeds and breathing patterns.

As the differences between 3D and 4D are all software based, it is recommended to only perform this test at commissioning/acceptance of the system and after every software update.

#### Preparation for 3D registration on average anatomy workflow

- Position the phantom on the CT scanner and select a constant-speed sinusoidal tumour movement of e.g. 2 cm in the cranial-caudal and 1 cm in the anterior-posterior direction, resulting in an ellipsoidal movement with a repetition cycle of 4-5 seconds.
- Acquire a 3D CT scan of the static phantom using a protocol that is standard for radiotherapy planning in your hospital and that creates an average anatomy. Deviations from the true average position will show up as a setup error during the actual test.
- Select a well-defined fixed position in the motion cycle and acquire a second scan of the stationary phantom.
- Transfer the scan-data to the planning system.
- Delineate the tumour in the phantom as GTV (as a structureset may be required to be able to import the plan in the image guidance system), place the isocentre in the centre of the GTV, and construct a simple plan with one or two fields (this will enable import of the plan and scan in the image guidance system).
- Transfer the reference CT, plan and structures to the linac and the IGRT system.



### At the linac

- Position the phantom on the treatment couch in the same way it was aligned on the CT scanner and select the same motion pattern. Correcting the setup error is not necessary as the test only looks at differences between 3D and 4D workflows.
- Acquire a 3D CBCT scan of the stationary phantom with the tumour at the same position as during the static planning CT acquisition.
- Acquire a 4D averaged CBCT scan of the moving phantom.
- Register the 3D and 4D averaged CBCT scans on the tumour.
- The actual values will reflect the deviation of the average of the tumour trajectory from the position in the planning CT scan (i.e. the setup error).
- The difference between the two registrations reflects additional uncertainty introduced by the 4D workflow (due to e.g. the 4D reconstruction/registration) and should be less than or equal to 1 mm in all directions.
- Optionally, the test could be repeated with other amplitudes.

As the differences between 3D and 4D are all software based, it is recommended to only perform this test at commissioning/acceptance of the system and after every software update.

## 7.6 Recommendations

Table 7.1. Summary of recommended tests (<sup>#</sup> A: acceptance; U: upgrade).

Test	Tolerance	Frequency <sup>#</sup>	See
<i>Image quality</i>			
Visual validation with phantom	none	AU	§7.2
<i>Geometric accuracy</i>			
Verify 3D versus 4D registration accuracy	≤ 1 mm	AU	§7.5

## 8 Software

### 8.1 *Aspects of QA of IGRT software*

Software of IGRT systems provides the possibility to acquire, reconstruct and register images and enables remote-controlled movement of the patient couch according to the registration results. The quality of acquisition and reconstruction of the images has been described in previous chapters ('Geometry', 'Image quality and artefacts' and '4D CBCT') and QA procedures have been proposed. For introducing/maintaining an IGRT system in the clinical process several other software features need to be commissioned and/or frequently tested. This chapter considers the following topics: connectivity (8.2); image registration (8.3); data storage, archival/retrieval and backup (8.4); additional functionality and third party software (8.5). The tests described consider specific IGRT functionality in particular. The performance of the hardware and (system) software and system security are to be tested as customary for software products used in radiotherapy.

As IGRT protocols and IGRT software vary between institutes the tests given below are not compulsory tests, but they can be used to set up a test matrix for the acceptance of IGRT software and IGRT software upgrades. It may be necessary to perform some of these tests at each linac, because the functionality of the software might depend on the local configuration.

### 8.2 *Connectivity*

IGRT software is linked to several other systems or software products: 2D or 3D images exported by virtual simulation software (VSIM) or treatment planning system (TPS) are used as reference data for registration (8.2.1); patient selection in the IGRT software may be performed automatically once a patient is selected in the treatment machine control software (8.2.2); the results of a match may be transferred to the treatment couch system for automatic and online position correction (8.2.3); IGRT images may be exported to external systems such as TPS for further evaluation (8.2.4). All these links should be tested to ensure correct execution of image guidance procedures. Of course, requirements and tests for import/export of data from VSIM or TPS have been addressed in NCS report 11 [46] and 15 [38]. In the sections below, each link in IGRT software is addressed from specific IGRT point of view. Parameters and features which have to be tested, if available in the software and clinically applied by the user, are described.

### **8.2.1 Reference data import**

Reference data imported by the IGRT software should be linked to the specific patient. If patient and treatment specific data are deduced from the imported reference data it should be ensured that the data is linked to the proper patient in the IGRT software. The infrastructure used to provide the IGRT software with the data determines which errors may occur.

#### *Suggested tests:*

- Verify if parameters such as patient name, patient ID, plan name and possibly other parameters are imported correctly.
  - *Action level:* data should be consistent.
- Enter data for a second plan or anatomical site for an existing patient and verify the plans are identified individually. New data sets for the reference image are to be created using 1) the original CT, but with a different isocentre as used in the first plan, and 2) a new CT-set.
  - *Action level:* data should be consistent.

Besides images, the reference data contains parameters that are vital for correct interpretation and use. Completeness of the image data requires that all essential information is transported to and entered into the IGRT software. Examples are isocentre position, field edges and contours of delineated structures. Reference images for IGRT can be either DRR's or complete CT scans. Most tests will be applicable to all reference data, some will be more specific to a certain type of reference.

#### *Suggested tests for all types of reference data:*

- IGRT software may not be able to import reference data of all sources or all sizes, e.g. there may be a limit on the number of slices which is accepted. Inquire or test if there are such limits within the IGRT software.
- The IGRT software should be able to handle images with different scales. In many cases scaling is performed automatically using parameters included in the image sets. If scaling is not performed automatically by the IGRT software, it should be possible for the user to scale the image using tools in the software. To test scaling of reference data sets, images of phantoms with known geometry should be imported in the IGRT software and the displayed size should be verified. Check the scaling in the three main orthogonal directions. Reference images with different scan parameters should be tested.

- *Action level:* the measured geometry should be within 1 voxel size.
- Test whether the isocentre in the imported images is at the correct position using scans of phantoms where this location can be easily verified (e.g. using a marker). At least test a plan with the isocentre moved from the original origin in the scanned image.
  - *Action level:* the isocentre should be within 1 voxel distance to the identified marker.
- Make reference images of a phantom with various clinically applied orientations such as head-first, feet-first, prone and supine. Check the orientation in all imported data sets.
  - *Action level:* the orientation should be identical.
- Check the display of the grey values. The image should display sufficient details to enable adequate evaluation of the registration.
  - *Action level:* the structures visible on the CT should be visible in the IGRT software. For example: for DRR's 16-bit greyscale is common; for CT images grey value range according to the HU range is common.

#### *Additional tests for DRR's:*

- In EPIDs knowledge of the position of the field edges with respect to the isocentre may be used for determination of the isocentre in the portal image. Beam properties and field edges from TPS data are imported into the IGRT software. The interpretation of these parameters can be verified using DRR's with field edges visualized in them. If those are not available one could apply DRR's of phantoms, e.g. used to validate scaling. If the isocentre coincides with a certain object within the phantom one can verify the position of the field edges by measurement of distances within the DRR.
  - *Action level:* the field edges should be within one voxel size distance to its proper location.

#### *Additional tests for CT scans:*

- Delineated structures are useful for the evaluation of the correction that is to be performed based on the registration of a CBCT. To verify that the structures are transferred correctly onto the reference image, a phantom with delineations around clearly visible structures should be imported. Also check whether the names of the structures are imported correctly.
  - *Action level:* the contours should be within one voxel size distance to the identified structure indicating the proper location. Names of structures should be consistent.
- It is recommended to test whether the image guidance system can deal with large numbers of structures, contours per structure, and/or points per structure.

*Frequency:*

The tests described above should be performed for each clinically applied source of reference information. The tests should be performed both at acceptance and after upgrades of either the IGRT software or the TPS or VSIM software.

### **8.2.2 Patient selection and saving acquired images**

Patient selection in the IGRT software may be performed manually or automatically once a patient is selected in the treatment machine control software (R&V). In case of automatic selection it should be verified that the selection is performed correctly. It should also be verified that the acquired images are assigned to the proper patient and plan.

*Suggested tests:*

- Select a field or plan of a patient with multiple plans at the treatment console and verify that the proper patient and plan are selected in the IGRT software.
  - *Action level:* patient/plan selection should be consistent.
- Acquire IGRT images for a patient with multiple plans and verify that the images are stored with the proper patient and plan.
  - *Action level:* data storage should be consistent.

*Frequency:*

Although the tests described above need not be performed on each linac if identical software versions are applied on the different machines, it is advised to verify that the connection is functioning at each linac. The tests should be performed both at acceptance and after upgrades of the software involved in the IGRT process (e.g. linac control or Record & Verify software, IGRT software).

### **8.2.3 Remote couch movement**

Motorized remote-controlled couches with three degrees of freedom (translate the patient along three axes) or six degrees of freedom (three translations plus three rotations) are widely available. IGRT software drives the couches to perform the required displacements such as a-priori displacements or registration results. QA of the couch correction process involves testing communication of the couch displacement data to the couch movement system and the assessment of the accuracy of the corrected movements. Tests for the latter are suggested in section 3.4.

*Suggested tests:*

- Verify that couch displacement data from the IGRT software are transferred correctly to the couch movement software. The transferred data can be the result from an online or offline protocol. The correct processed protocol result should be transferred. There are no requirements for the phantom that is used for this test.
  - *Action level:* data transfer should be consistent to the last visible digit allowing for round off with a 1 mm, 0.1 degree upper limit (rotation limit only for couches with 6 degrees of freedom).

*Frequency:*

Testing of this communication should be performed both at acceptance and after upgrades of either the IGRT software or the treatment couch control software. The suggested tests should be performed at each linac as local settings may influence proper functioning.

## **8.2.4 Data export**

### **8.2.4.1 Image data sets**

The volumetric and radiographic imaging data from the IGRT system can be used for evaluation in software packages of the same or other vendors. In some cases data is exported using for instance a DICOM transfer protocol, in other cases the IGRT data is read directly from the IGRT database. Dependent on the purpose of usage the transferred data needs to be verified. For example, strict requirements for export of a 3D CBCT data set should be imposed when the data set is used for re-planning or plan evaluation (see section 5.1.2).

Note that validation of the exported data using a different software product such as a TPS may introduce errors due to the import settings of that product (see also NCS 15 [38]). The exported data should therefore be validated using the intended software for further processing, thereby checking the whole export-import chain. This also applies if IGRT data is used in modules/packages of the same vendor as the IGRT software.

*Suggested tests for export of 2D or 3D images:*

- Verify consistency of patient name, ID, plan, date and time of scan, etc. in the exported dataset.
  - *Action level:* data should be consistent.

- Verify that the geometry of the exported images is correct. Scaling and the isocentre position can be tested using a phantom with known geometry. Beware that the export module may enable export of the dataset in the position before/after registration thus influencing the isocentre position.
  - *Action level:* the measured geometry should be within 1 voxel. The isocentre should be within 1 voxel distance to the identified marker.
- Verify that the pixel grey values of the exported image are correct.
  - *Action level:* pixel values should be identical.

#### **8.2.4.2 Image registration results**

The position errors found during image registration may be used to adjust the patient position following an offline protocol or for analysis of the entire positioning procedure. As not all IGRT software includes complicated decision protocols, the registration results need to be exported to an external software package. The consistency of the exported/imported data needs to be verified.

*Suggested tests for export of patient position error or correction:*

- Verify consistency of patient name, ID, plan, date and time of scan.
  - *Action level:* the data should be consistent.
- Verify that the exported position error is correct (translations and rotations if applicable) by comparing it to the original data for a selected set of patients.
  - *Action level:* the exported/imported data should be identical.

*Frequency:*

Although the tests described above need not be performed on each linac if identical software versions are applied on the different machines, it is advised to verify that the connection is functioning at each linac. The tests should be performed both at acceptance and after upgrades of either the IGRT software or software of the receiving system (e.g. a treatment planning system).

### **8.3 Image registration**

Most IGRT software packages offer various methods for rigid image registration: manual registration, automatic registration on bony anatomy, automatic registration on grey values,

and automatic registration on fiducial markers. The user can decide to take into account both translational and rotational shifts or to take into account translations only.

Many factors influence the quality of image registration. For instance, the non-rigid changes in the patient's anatomy and the rigid approach of the available registration methods constitute an important limit. Limited visibility of target and/or organs at risk will hamper accurate registration as will artefacts in the reconstructed images (see section 5.3 for examples of artefacts). Due to these limitations the user will often have to make a compromise while looking for the best match for the specific clinical case.

Assessment of the quality of image registration methods is important in modern radiotherapy, however techniques to validate these methods are still being developed. A review of current registration approaches and solutions for image registration in radiotherapy is given in AAPM TG report 132 [47]. In this AAPM report, recommendations for quality assurance and quality control of registration in the clinical process are given as well. Unfortunately, the quantitative measures of image registration accuracy described in this report are not a standard feature in the IGRT software, or in many other commercial registration products. For commissioning, phantoms, either physical or virtual (digital or digitally edited), can be very useful. Virtual phantoms are useful for performance tests, physical phantoms are needed for end-to-end tests. The use of geometric phantoms may oversimplify the clinical situation resulting in accuracy not obtainable in clinical images. The validity of registration in the clinic should be assessed qualitatively using the visualization tools present in the IGRT software (e.g. checker board, complementary colour overlay, etc.).

Design of QA protocols of patient specific image registration methods or a full elaboration of the current status is outside the scope of this report. The tests proposed below are limited to obtain confidence in the basic abilities of the available registration procedures and the selected settings. The results may be influenced by the quality of the acquired image, e.g. by artefacts present in the images.

*Suggested tests (do this test for each clinically applied registration method available in the software):*

- Make a CT scan of a phantom with structures that can be used for image registration, such as plastic rods and fiducial markers. Prepare the reference data set in the IGRT software for the registration method that is being assessed. Position the phantom at the

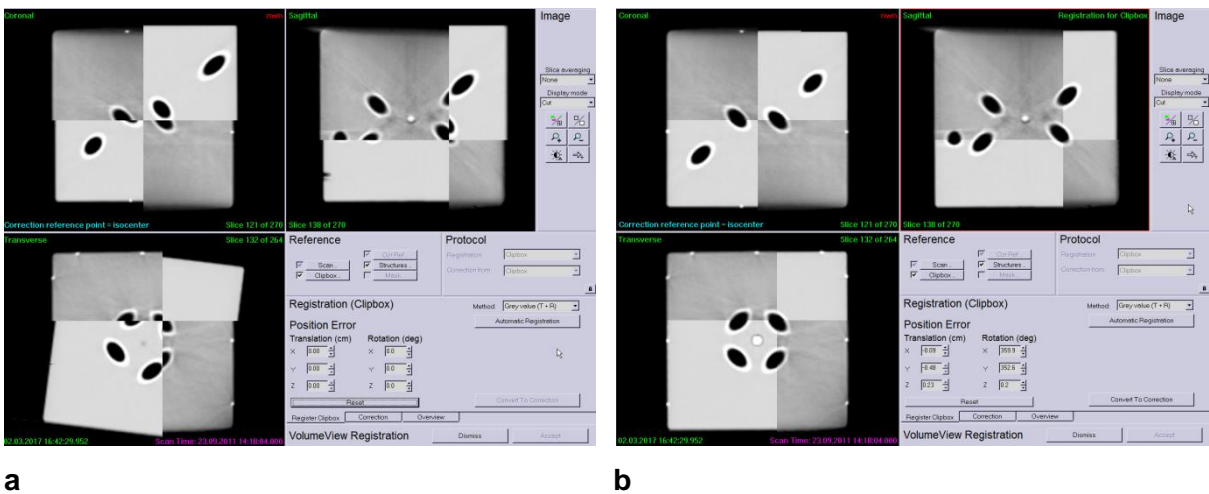


treatment couch as accurately as possible using lasers and marks on the phantom. Verify the accuracy of the initial setup by imaging and registration. Next, apply a well-known displacement of the phantom, e.g. by translating or rotating the treatment couch and make the images for image guidance. Register the images and compare the results to the applied displacement. An alternative approach is to create multiple reference images with known simulated translations/rotations from a single CT scan of the phantom (e.g. by using an image registration software package one can translate/rotate the original reference image and use resampled data as reference image). If CBCT images can be moved from one plan to another, a single CBCT scan of the phantom in the original position can be registered to all reference images. If this is not possible one should make a new scan of the phantom in the original position for each reference image. An example of a test following this approach is demonstrated in figure 8.1, where the phantom in the reference image is digitally rotated around the longitudinal axis. The match result should be in agreement with the simulated translations/rotations.

- *Action level:* the registration results can be assessed visually. For artefact free areas of the image the tolerance is within one voxel size. For a phantom with clear contrasts and a good quality CBCT the position correction errors found after registration should be within 1 mm and 1 degree (per component) of the applied displacement of the phantom or of the simulated translation/rotation of the reference scan.

Frequency:

The tests should be performed at acceptance and after upgrades of the IGRT software.



*Figure 8.1. Checker board representation of CT and CBCT of a phantom before (a) and after (b) automatic registration (using an adapted reference image with known translations and rotations).*

For registration of the CBCT to the reference CT the user is offered the possibility to perform a registration with 6 or 3 degrees of freedom: i.e. using both rotations and translations or translations only. While there are couches available that are able to correct both translations and rotations (the corrections for the rotations with these couches are still limited, e.g. up to rotations of 3 degrees), most couches in clinical use are 3D couches. For facilities with 3D couches one can decide to do 3D registration, however even if the treatment couch is unable to perform rotational corrections, the use of 6D registration can be beneficial. For instance the 6D registration will simplify examination of the registration accuracy and improves perception of anatomical variations. However, in such cases the 6D registration results have to be converted into 3D correction values which can be performed by the treatment couch. The IGRT software may facilitate the derivation of the optimal feasible translational correction, for instance using a user-defined point-of-interest for correcting the present rotations. This point of interest is the only point that will stay in place when the rotations are eliminated (placed for instance at the centre of the target).

*Suggested tests:*

- For instance, scans in the previous test in this section can be used for this purpose. Define a point-of-interest in the reference data set that will be used by the software for converting rotations into translations (for proper testing one should not select the machine isocentre for this point). Register the acquired image and let the IGRT software determine the correction that can be performed using the treatment couch in use.
  - *Action level:* after conversion of the registration results into the feasible correction the defined point-of-interest should still be registered correctly (within 1 voxel).

*Frequency:*

The test should be performed at acceptance and after upgrades of the IGRT software.

Note: In clinical practice the user is strongly advised to visually assess the registration result, both in the target volume and in the surrounding volume, for each acquired image. This verification step should be performed both on the registration result and after conversion of the registration result into a correction which can be applied by the treatment couch in use.

#### 8.4 *Data storage, archival/retrieval and backup*

After image acquisition and registration, the data should be stored as part of the documentation of the treatment, for possible future clinical evaluation or scientific research. The user should verify that the IGRT software stores all relevant information and that the information is correct after retrieval. Requirements with respect to the data storage, for archival or backup, are identical. Upon retrieval all information should be read in correctly by the IGRT software.

##### *Suggested tests:*

- Verify consistency of patient name, ID, treatment, etc.
  - *Action level:* the retrieved data should be equal to the original data.
- Verify consistency of reference data, including contours, isocentre position, and prepared match approach (clipboard/ROI/mask, manual/automatic registration, grey-level/bony-anatomy/marker match).
  - *Action level:* the retrieved data should be equal to the original data.
- Verify consistency of IGRT images, date and time of scan and match results including approval.
  - *Action level:* the retrieved data should be equal to the original data..

##### *Frequency:*

The tests should be performed both at acceptance and after upgrades of the IGRT software.

#### 8.5 *Additional functionality and third party software*

Software for acquiring CBCT images and for registration may provide additional functionality such as couch corrections based on non-isocentric images, decision protocols, gating options or registration analysis. Before clinical use of such functionality it should be validated and QA tests should be included in the QA procedures of the entire IGRT process. It is up to the user to determine action levels and desired frequency of the tests.

#### 8.6 *Summary of tests*

The tests suggested in this chapter are summarized in table 8.1. In general the tests are to be performed at acceptance or after upgrades. For upgrades certain parts of the software may remain unchanged. Based on release notes and experience of the user with previous

upgrades one may decide to limit the amount of tests. Properly designed end-to-end tests mimicking clinical workflow are essential after upgrades to efficiently test proper functioning of the software.

*Table 8.1. Summary of suggested tests.*

<sup>†</sup> Patient/object identification parameters may include (dependent on specific test): patient name, patient ID, case/plan/field name, date/time stamp of images/match.

<sup>#</sup> Frequency: tests need to be performed after acceptance or upgrades of the IGRT software itself or after acceptance or upgrades of software communicating with the IGRT software. Abbreviations: AU-I: acceptance or upgrade of IGRT software; AU-P: acceptance or upgrade of planning or VSIM software; AU-L: acceptance or upgrade of linac control or R&V software; AU-C: acceptance or upgrade of treatment couch control software; AU-R: acceptance or upgrade of receiving software.

Topic	Test	Tolerance	Frequency <sup>#</sup>	See
<i>Item tested</i>				
<b>Connectivity</b>				
<i>Reference data import: general</i>	Verify <sup>†</sup> patient/object identification parameters for a new and an existing patient	Consistent data	AU-I, AU-P	§8.2.1
	Verify scaling of reference image	Within 1 voxel size	AU-I, AU-P	§8.2.1
	Verify isocentre position in reference image	Within 1 voxel distance	AU-I, AU-P	§8.2.1
	Verify orientation of reference image	Correct orientation	AU-I, AU-P	§8.2.1
<i>Reference data import: DRR's</i>	Verify position of field edges	Within 1 voxel distance	AU-I, AU-P	§8.2.1
<i>Reference data import: CT</i>	Verify name and position of contours (delineated structures)	Consistent names. Contours within 1 voxel distance	AU-I, AU-P	§8.2.1
<i>Patient selection and saving data</i>	Verify proper automatic patient/plan selection	Patient/plan selection consistent	AU-I, AU-L	§8.2.2
	Verify proper data storage	Data storage at proper patient/plan	AU-I, AU-L	§8.2.2
<i>Remote couch movement</i>	Verify transfer couch displacement data (for online or offline protocol)	Data consistent to last visible digit	AU-I, AU-C at each linac	§8.2.3
<i>Data export: image data sets</i>	Verify <sup>†</sup> patient/object identification parameters in exported dataset	Consistent data	AU-I, AU-R	§8.2.4.1
	Verify image geometry and isocentre position of exported image	Geometry within 1 voxel/1 mm. Isocentre within 1 voxel	AU-I, AU-R	§8.2.4.1

		distance		
	Verify pixel grey values of exported image	Grey values should be identical	AU-I, AU-R	§8.2.4.1
<i>Data export: image registration results</i>	Verify <sup>†</sup> object identification parameters	Consistent identification parameters.	AU-I, AU-R	§8.2.4.2
	Verify exported error/correction data	Translations/ rotations should be identical.	AU-I, AU-R	§8.2.4.2
<b>Image registration</b>				
<i>Registration quality</i>	Verify each clinically applied registration method using a known displacement	Registration results within 1 mm 1 degree per component	AU-I	§8.3
<i>Conversion of registration into correction</i>	Verify conversion of the registration including rotations into a correction using translations only	The defined point-of-interest for conversion is at the correct position (within 1 voxel)	AU-I	§8.3
<b>Data storage, archival retrieval and backup</b>				
<i>Data consistency</i>	Verify <sup>†</sup> patient/object identification parameters of retrieved data	Retrieved identification data is identical to original identification data	AU-I	§8.4
	Verify retrieved reference data	Retrieved reference data is identical to original reference data	AU-I	§8.4
	Verify retrieved images	Retrieved images are identical to original images	AU-I	§8.4

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## References

- [1] Althof V, de Ost B, Reynaert N, Schubert K, Sterpin E, van de Kamer JB. NCS Report 27: Quality Assurance for Tomotherapy Systems. Delft: 2017.
- [2] Langen KM, Papanikolaou N, Balog J, Crilly R, Followill D, Muty Goddu S, et al. QA for helical tomotherapy: Report of the AAPM Task Group 148. *Med Phys* 2010;37:4817-4853.
- [3] Bissonnette JP, Balter PA, Dong L, Langen KM, Lovelock DM, Miften M, et al. Quality assurance for image-guided radiation therapy utilizing CT-based technologies: Report of AAPM Task Group 179. *Med Phys* 2012;39:1946-1963.
- [4] de las Heras Gala H, Torresin A, Dasu A, Rampado O, Delis H, Hernández Girón I, et al. EFOMP-ESTRO-IAEA Protocol: Quality control in cone-beam computed tomography (CBCT). 2017; [https://www.efomp.org/uploads/2017-06-02-CBCT\\_EFOMP-ESTRO-IAEA\\_protocol.pdf](https://www.efomp.org/uploads/2017-06-02-CBCT_EFOMP-ESTRO-IAEA_protocol.pdf).
- [5] Varian medical systems: On-Board Imager (OBI) Advanced Imaging Maintenance Manual, version 1.6, July 2015.
- [6] Yoo S, Kim GY, Hammoud R, Elder E, Pawlicki T, Guan H, et al. A quality assurance program for the on-board imagers. *Med Phys* 2006;33:4431-4447.
- [7] Bissonnette JP. Quality assurance of image-guidance technologies. *Semin Radiat Oncol* 2007;17:278-286.
- [8] Bissonnette JP, Moseley DJ, Jaffray DA. A quality assurance program for image quality of cone-beam CT guidance in radiation therapy. *Med Phys* 2008;35:1807-1815.
- [9] Elekta: iViewGT Planned Maintenance Manual, 2009.
- [10] Elekta: iViewGT™ R3.4 Instructions for Use, 2010.
- [11] Elekta: XVI R5.0.3 Instructions for Use for: Elekta Synergy<sup>R</sup>, Elekta Axesse™, Elekta Infinity™, Versa HD™, 2015.
- [12] Cho Y, Moseley DJ, Siewerdsen JH, Jaffray DA. Accurate technique for complete geometric calibration of cone-beam computed tomography systems. *Med Phys* 2005;32:968-983.
- [13] Bissonnette JP, Moseley D, White E, Sharpe M, Purdie T, Jaffray DA. Quality assurance for the geometric accuracy of cone-beam CT guidance in radiation therapy. *Int J Radiat Oncol Biol Phys* 2008;71:S57-S61.
- [14] Sykes JR, Lindsay R, Dean CJ, Brettell DS, Magee DR, Thwaites DI. Measurement of cone beam CT coincidence with megavoltage isocentre and image sharpness using the QUASAR Penta-Guide phantom. *Phys Med Biol* 2008;53:5275-5293.
- [15] Klein EE, Hanley J, Bayouth J, Yin FF, Simon W, Dresser S, et al. Quality assurance of

- medical accelerators: Report of AAPM Task Group 142. *Med Phys* 2009;36:4197-4212.
- [16] Gao S, Du W, Balter P, Munro P, Jeung A. Evaluation of IsoCal geometric calibration system for Varian linacs equipped with on-board imager and electronic portal imaging device imaging systems. *J Appl Clin Med Phys* 2014;15:164-181.
  - [17] Lutz W, Winston KR, Maleki N. A system for stereotactic radiosurgery with a linear accelerator. *Int J Radiat Oncol Biol Phys* 1988;14:373-381.
  - [18] Yin FF, Wong J, Balter J, Benedict S, Bissonnette JP, Craig T, et al. The role of in-room kV x-ray imaging for patient setup and target localization: Report of AAPM Task Group 104. 2009.
  - [19] Du W, Gao S, Wang X, Kudchadker RJ. Quantifying the gantry sag on linear accelerators and introducing an MLC-based compensation strategy. *Med Phys* 2012;39:2156-2162.
  - [20] Rowshanfarzad P, Riis HL, Zimmermann SJ, Ebert MA. A comprehensive study of the mechanical performance of gantry, EPID and the MLC assembly in Elekta linacs during gantry rotation. *Br J Radiol* 2015;88:20140581.
  - [21] Das IJ, Cao M, Cheng CW, Misic V, Scheuring K, Schüle E, et al. A quality assurance phantom for electronic portal imaging devices. *J Appl Clin Med Phys* 2011;12:391-403.
  - [22] Varian medical systems: On-Board Imager (OBI) Advanced Imaging Reference Guide, March 2013.
  - [23] Elekta: XVI R5.0 Corrective Maintenance Manual, 2013.
  - [24] Ma CM, Coffey CW, DeWerd LA, Liu C, Nath R, Seltzer SM, et al. AAPM protocol for 40-300 kV x-ray beam dosimetry in radiotherapy and radiobiology: Report of AAPM Task Group 61. *Med Phys* 2001;28:868-893.
  - [25] FDA Title 21, Code of Federal Regulations: Performance Standard for Diagnostic X-ray Systems and their Major Components. 2005; <https://www.fda.gov/downloads/Radiation-EmittingProducts/RadiationEmittingProductsandProcedures/MedicalImaging/MedicalX-Rays/UCM203106.pdf>.
  - [26] Murphy MJ, Balter J, Balter S, BenComo JA, Das IJ, Jiang SB, et al. The management of imaging dose during image-guided radiotherapy: Report of AAPM Task Group 75. *Med Phys* 2007;34:4041-4063.
  - [27] Morin O, Gillis A, Descovich M, Chen J, Aubin M, Aubry JF, et al. Patient dose considerations for routine megavoltage cone-beam CT imaging. *Med Phys* 2007;34:1819-1827.
  - [28] Dixon RL, Anderson JA, Bakalyar DM, Boedeker K, Boone JM, Cody DD, et al. Comprehensive methodology for the evaluation of radiation dose in x-ray computed tomography: Report of AAPM Task Group 111. 2010.



- [29] ICRU Report No. 87: Radiation dose and image-quality assessment in computed tomography. J ICRU 2012;12:1-149.
- [30] International Atomic Energy Agency human health reports No. 5: Status of computed tomography dosimetry for wide cone beam scanners. 2011; [http://www-pub.iaea.org/MTCD/Publications/PDF/P1528\\_web.pdf](http://www-pub.iaea.org/MTCD/Publications/PDF/P1528_web.pdf).
- [31] Dixon RL, Boone JM. Cone beam CT dosimetry: a unified and self-consistent approach including all scan modalities--with or without phantom motion. Med Phys 2010;37:2703-2718.
- [32] Varian medical systems: On-Board Imager (OBI) Advanced Imaging Reference Guide, July 2015.
- [33] Rajapakshe R, Luchka K, Shalev S. A quality control test for electronic portal imaging devices. Med Phys 1996;23:1237-1244.
- [34] Cremers F, Frenzel T, Kausch C, Albers D, Schönborn T, Schmidt R. Performance of electronic portal imaging devices (EPIDs) used in radiotherapy: image quality and dose measurements. Med Phys 2004;31:985-996.
- [35] Low DA, Klein EE, Maag DK, Umfleet WE, Purdy JA. Commissioning and periodic quality assurance of a clinical electronic portal imaging device. Int J Radiat Oncol Biol Phys 1996;34:117-123.
- [36] Herman MG, Balter JM, Jaffray DA, McGee KP, Munro P, Shalev S, et al. Clinical use of electronic portal imaging: Report of AAPM Task Group 58. Med Phys 2001;28:712-737.
- [37] Menon GV, Sloboda RS. Quality assurance measurements of a-Si EPID performance. Med Dosim 2004;29:11-17.
- [38] Bruinvis IAD, Keus RB, Lenglet WJM, Meijer GJ, Mijnheer BJ, van 't Veld AA, et al. NCS Report 15: Quality assurance of 3-D treatment planning systems for external photon and electron beams. Delft: 2005.
- [39] Huijsse S, de Vries A, Greuter M, Schilham A, van Rooijen B, Borns C, et al. NVKF Aanbevelingen kwaliteitscontrole CT scanners, versie 2.0. 2016.
- [40] Barrett JF, Keat N. Artifacts in CT: recognition and avoidance. Radiographics 2004;24:1679-1691.
- [41] Hunter AK, McDavid WD. Characterization and correction of cupping effect artefacts in cone beam CT. Dentomaxillofac Radiol 2012;41:217-223.
- [42] Feenstra SJ, van de Poel JA, van 't Riet A, de Vroome H, Welleweerd J. NCS Report 8: Kwaliteitscontrole van Medische Lineaire Versnellers. Delft: 1995.
- [43] Meijer GJ, van Kleffens HJ, Mijnheer BJ. NCS Report 9: Quality Control of Medical Linear Accelerators. Delft: 1996.

- [44] Sonke JJ, Zijp L, Remeijer P, van Herk M. Respiratory correlated cone beam CT. *Med Phys* 2005;32:1176-1186.
- [45] Rit S, Nijkamp J, van Herk M, Sonke JJ. Comparative study of respiratory motion correction techniques in cone-beam computed tomography. *Radiother Oncol* 2011;100:356-359.
- [46] 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.
- [47] Brock KK, Mutic S, McNutt TR, Li H, Kessler ML. Use of image registration and fusion algorithms and techniques in radiotherapy: Report of AAPM Task Group 132. *Med Phys* 2017;44:e43-e76.