Intraoperative and dynamic 3D rotational X-ray imaging

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Chapter 2

*High contrast image quality of a mobile 3D Rotational X-Ray device*

Bart Carelsen, Henk W. Venema, Marcel van Herk, Patricia van Kemenade, Wilko Grolman, Niels J. Noordhoek, Geert J. Streekstra

*Submitted*
2.1 Introduction

3D Rotational X-ray (3D-RX) with a mobile C-arm, also termed cone beam CT (CBCT), provides intra-operative 3D images to support the surgeon. In an emerging number of surgical fields this technique has become a standard procedure to achieve an optimal operation result.[1-3] Intra-operative 3D-RX replaces post-operative and occasional pre-operative Computed Tomography (CT) examinations. Examples of application areas of 3D-RX imaging are trauma, hand, orbital, paranasal sinus and cochlear implantation surgery. These applications require a high spatial resolution and a proper display of the high contrast transitions between bone and (soft) tissue and tissue and air, as well as (to a lesser extent) a proper depiction of metal implants.

In clinical practice, the choice of image quality related device settings like tube voltage, tube charge and magnification mode are based on evaluation of images of cadaver specimens and are also based on acceptability of reconstruction times. Although the 3D-RX image quality as observed in clinical practice gives some directions for choosing device settings, these settings may be sub-optimal due to subjective user preferences. A better starting point for determining image quality for a certain application is by comparison of 3D-RX image quality with that of CT, since CT is the clinical reference standard for therapy planning and verification.

In literature, comparisons of the image quality of 3D-RX and CT have been made by considering the detectability of anatomical landmarks of bone within tissue and/or tissue within air of a certain size.[4; 5] The drawback of such an approach is that the anatomical landmarks are not well defined and therefore only a qualitative and subjective comparison is possible.

We propose a quantitative estimation of the image quality and detectability of small high contrast details based on phantom experiments. This approach provides for an objective measure of the ability to detect bone details, which may help to choose the appropriate system settings for each surgical application.

Recently, two studies were published in which the image quality of 3D-RX systems was investigated quantitatively.[8; 9] The study of Fahrig et al. investigated spatial resolution, dose and perception of details of low contrast of a ceiling mounted C-arm based CBCT system.[8] The study of Daly et al. concentrated on the assessment of dose and the dose dependency of the visibility of low and high contrast details, and optimal imaging geometry for head and neck imaging with a prototype mobile C-arm.[9] Both studies did not make a comparison of image quality of 3D-RX with CT and were based on perception of contrast details.

The purpose of this study is to investigate the image quality of a mobile 3D-RX device in comparison to the clinical standard CT. To this end we measure CNR and spatial resolution, as well as radiation dose. From the CNR and resolution measurements we estimate the detectability of small bone details using a slightly modified Rose model.[6; 7]
2.2 Method & Materials

3D-RX system

The BV Pulsera with 3D-RX (Philips Medical Systems, Best, The Netherlands) is a mobile C-arm (Fig. 1) with an additional servo driven propeller movement and 3D reconstruction software. During 30 s propeller rotation of the C-arm over an angle of 200° a number of pulsed fluoroscopy projections is acquired. By adjusting the acquisition speed 450, 225 or 90 projection images can be acquired. More projection images result in better quality and higher dose as the average dose per projection image is independent of the number of projections for this particular system.

The 12” X-ray image intensifier (XRII) has three magnification modes which allows for 3D reconstructions of different fields of view (FOV) and spatial resolution. The three different XRII magnification settings result in 179, 124 or 91 mm diameter spherically shaped FOVs each inscribed in a cube of 256x256x256 voxels. During the rotation, projection images are acquired with dynamic dose adjustment, which consists of adjusting the tube voltage. In this way the variation of the average intensity at the entrance window of the XRII is reduced, which is necessary to optimally use the dynamic range of the XRII. Each image is made with a 10 ms X-ray pulse and 60 mA tube current. In Tables 1 and 2 an overview of the settings used for 3D-RX is provided.

![Diagram](image.png)

*Fig. 1: Left: Side view Line drawing of the BV Pulsera with 3D-RX C-arm stand and the placement of the RX phantom. Right: front view at the C-arm. The C-arm rotates around the Z axis.*
Table 1: Settings of the 3D-RX regarding magnification.

<table>
<thead>
<tr>
<th>XRII Magnification</th>
<th>12 inch (31 cm)</th>
<th>9 inch (23 cm)</th>
<th>6 inch (17 cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volume diameter (FOV)</td>
<td>179 mm</td>
<td>124 mm</td>
<td>91 mm</td>
</tr>
<tr>
<td>Voxel size</td>
<td>0.70 mm</td>
<td>0.54 mm</td>
<td>0.40 mm</td>
</tr>
<tr>
<td>Tube voltage (RX phantom)</td>
<td>52 kV</td>
<td>53 kV</td>
<td>55 kV</td>
</tr>
<tr>
<td>Tube voltage (head examination)</td>
<td>64-67 kV</td>
<td>65-73 kV</td>
<td>68-79 kV</td>
</tr>
</tbody>
</table>

Table 2: Settings of the 3D-RX regarding image quality.

<table>
<thead>
<tr>
<th>Quality</th>
<th>High</th>
<th>Medium</th>
<th>Low</th>
</tr>
</thead>
<tbody>
<tr>
<td>Projection images</td>
<td>450</td>
<td>225</td>
<td>90</td>
</tr>
<tr>
<td>Total tube charge</td>
<td>270 mAs</td>
<td>135 mAs</td>
<td>54 mAs</td>
</tr>
<tr>
<td>Reconstruction time</td>
<td>6 min.</td>
<td>3 min.</td>
<td>1 min.</td>
</tr>
</tbody>
</table>

Table 3: Acquisition and reconstruction settings of the MSCT scanner. Bottom: the 3D-RX settings that are used for the particular examination.

<table>
<thead>
<tr>
<th>MSCT Protocol</th>
<th>Orbit</th>
<th>Skeleton</th>
<th>Ear</th>
</tr>
</thead>
<tbody>
<tr>
<td>Protocol code</td>
<td>CT-O</td>
<td>CT-S</td>
<td>CT-E</td>
</tr>
<tr>
<td>Number of rows x collimation</td>
<td>4 x 1.0 mm</td>
<td>2 x 0.5 mm</td>
<td>2 x 0.5 mm</td>
</tr>
<tr>
<td>Reconstruction kernel</td>
<td>H60s</td>
<td>B60f</td>
<td>U90u</td>
</tr>
<tr>
<td>Slice thickness/increment</td>
<td>1.25/0.6 mm</td>
<td>0.5/0.3 mm</td>
<td>0.5/0.2 mm</td>
</tr>
<tr>
<td>FOV (512² matrix)</td>
<td>200 x 200 mm</td>
<td>211 x 211 mm</td>
<td>150 x 150 mm</td>
</tr>
<tr>
<td>Axial length scan</td>
<td>70 mm</td>
<td>100 mm</td>
<td>60 mm</td>
</tr>
<tr>
<td>Tube Voltage</td>
<td>120 kV</td>
<td>120 kV</td>
<td>120 kV</td>
</tr>
<tr>
<td>Effective total tube charge</td>
<td>100 mAs</td>
<td>105 mAs</td>
<td>165 mAs</td>
</tr>
</tbody>
</table>

3D-RX equivalent | Medium Quality 12” | Medium Quality 9” | High Quality 6” |
Multi Slice CT scanner

To compare the measurements on a 3D-RX system with those on a clinical standard CT we used a 4 slice CT scanner (Somatom Sensation 4, Siemens, Erlangen, Germany). We used three protocols that are also used for patients who are operated upon with the aid of intra operative 3D-RX, i.e., the imaging of the orbit, the skeleton and the inner-ear or mastoid.[1; 4] In this last setting the spatial resolution of the CT scanner is improved by inserting additional pins between the detectors in order to reduce detector aperture.[10; 11] The settings for these protocols are shown in Table 3.

RX Phantom

Standard CT image quality phantoms imaged with 3D-RX yield incomplete phantom reconstructions and truncation artifacts, especially when the smallest field of view is used. Therefore, we used a custom-made RX phantom to measure both the spatial resolution and the CNR (Fig. 1). The RX phantom is a 12 cm long, 9 cm diameter polypropylene cylinder, with a 3 cm cylindrical hole in the centre (Fig. 1 and 2). Polypropylene (approximate CT value of -70 HU at 120 kV) represents (soft) tissue and its attenuation is approximately constant in the range 50-120 kV. A Polyvinyl chloride (PVC) insert (CT value approximately 950 HU at 120 kV) represents bone. PVC has an approximately 2 times higher CT value at 52-55 kV than at 120 kV, similar to bone.[12] The RX phantom was scanned in the 3D-RX system according to Fig. 1. In the CT scanner the cylindrical axis of the RX phantom was aligned with the axis of gantry rotation.

Fig. 2. 3D-RX High Quality 12” images of the RX phantom. The top left is a cut plane rendering and the remainder images are orthogonal planar reformattings. The definition of the axes as used in Figure 1 are drawn in the figure. The dashed rectangles indicate the regions which the ESF is determined in all three directions. The filled rectangles and circle indicate the size and position of the CNR measurements, from top to bottom representing air, (soft) tissue and bone. In the lower right part the axial image is shown, the apparent streak from left to right around the middle is due to the absence of Parker Weighting.[19]


2.3 Experiments

A. Spatial resolution

Generally 3D-RX systems have an anisotropic point spread function (PSF) due to limited angle information.[13] Consequently, we obtained the spatial resolution in all three directions. To this end the edge spread function (ESF) was measured in each direction.[14] Next, the ESF data was fitted with the convolution of a step edge and a Gaussian PSF, with the full width at half maximum (FWHM) of the PSF as a free parameter. The PSF of CT systems can be reasonably approximated by a Gaussian.[15]

For the 3D-RX system the ESF measurements were performed in ROIs in 6 adjacent Z-Y and Z-X slices of the RX phantom perpendicular to the edge, as indicated in Fig. 2. In our study the edges in the phantom images are cylindrical, whereas for measurement of the ESF, edges should ideally be straight.[14] However, the radius of curvature of the cylindrical edge is relatively large compared to the size of the PSF and the edge can therefore be considered locally straight. For the CT scanner used in this study it appeared that accurate measurements of the ESF in this way were not feasible because the CT values at the lower part of the edge are clipped. Therefore, in the CT scanner edge images were acquired with a PVC cube in water. Estimation of the FWHM of the PSF was done in the same way as described above.

B. Contrast and noise

In CT images the CT values of the voxels are standardized attenuation values relative to those of water. They are customarily expressed in Hounsfield Units (HU). In 3D-RX images the RX values of the voxels are not standardized.

In both 3D-RX and CT the CNR values of tissue in air and bone in tissue were determined by measuring the mean and standard deviation (SD) of the CT and RX values in a region of interest (ROI) of approximately 1 cm² (see Fig. 2) for 6 different slices. The CNR is defined by:

\[
CNR = \frac{|\bar{X}_{\text{soft tissue}} - \bar{X}_{\text{material}}|}{SD_{\text{soft tissue}}},
\]

where \(\bar{X}\) is the average CT- or RX-value and \(SD_{\text{soft tissue}}\) is the standard deviation in the ROI in polypropylene.

C. Effective Dose

For comparison of the dose of 3D-RX and CT examinations, the effective dose was estimated for a typical head examination for both modalities. For 3D-RX dose calculations, PCXMC software was used.[16] The effective dose for one rotational run was estimated as the average dose of 2 lateral, 1 AP and 2 oblique projection images of the head times the number of projection images, using the parameters of Tables 1 and 2. Dose calculations for the head examination with CT were done with the CT dosimetry spreadsheet of ImpactScan.[17] The dose of the CT skeleton protocol could not be calculated because the software does not provide data for the extremities.
D. Prediction of bone detail detectability

3D-RX images are relatively noisy in comparison with CT images, which reduces visibility of details in the image. This could imply that a small object is imaged with sufficient spatial resolution, but nevertheless remains invisible because of the high noise level.

The Rose model has been used extensively in the literature as a first approximation of the relation between the visibility of a detail and its signal to noise ratio (SNR). [6; 7] The SNR is defined as

\[ SNR = CNR \cdot \sqrt{n} \]  

where \( n \) is the number of voxels of the detail. Hereby the assumption is made of uncorrelated noise and a detail with sharp edges. When the SNR of a detail exceeds a certain threshold value \( k \) it should be visible. Conventionally the threshold \( k \) is assumed to be approximately 5 to 7.[6, 7]

We used a slightly modified version of the Rose model to estimate the minimal diameter of a bone detail in tissue that should be detectable at a given CNR and FWHM of the PSF for each 3D-RX and CT setting. Starting point is the image of a circular detail representing the cross-section of a small bony cylinder (mimicking a small ossicle) which is aligned along the z-axis of the imaging system.

Firstly, we determine the apparent diameter and the apparent contrast of the circular detail after convolving the cylinder with a 2D Gaussian representing the in-plane PSF of the 3D-RX or CT system. In our slightly modified Rose model we approximate the apparent diameter (\( d_{\text{apparent}} \)) of the detail by the FWHM of the blurred cylindrical cross-section and the apparent contrast of the detail (\( C_{\text{apparent}} \)) by the maximum intensity within the cross-section relative to its surroundings. For the FWHM of the 2D-Gaussian representing the in-plane PSF we take the measured average FWHM of the PSF in X and Y direction.

Secondly, the noise in 3D-RX and CT imaging is spatially correlated and for that reason the \( 1/\sqrt{n} \) dependency of the SD of the intensity of an object on \( n \) no longer holds.[22] Therefore, the SD was determined separately for the area of each circular detail (\( SD_{\text{detail}} \)) with diameter \( d_{\text{apparent}} \). Because the area corresponding to \( d_{\text{apparent}} \) appeared generally to be between 1 and 2 voxels, we determined the SD in ROIs of 1 voxel and 2x2 voxels and used linear interpolation to estimate the \( SD_{\text{detail}} \) for details with intermediate areas. In case the area corresponding to \( d_{\text{apparent}} \) was slightly smaller then the voxel size the noise corresponding to a ROI of 1 voxel was used.

Next, we combine the apparent contrast \( C_{\text{apparent}} \) with the \( SD_{\text{ROI}} \) of each detail to estimate the SNR of the object,

\[ SNR = \frac{C_{\text{apparent}}}{SD_{\text{ROI}}} \]  

(3)
As a last step, we determine the minimally detectable bone detail diameter as the diameter for which the SNR equals the threshold value $k$. We take $k=10$ as the detectability threshold to be on the safe side in view of all approximations involved.

The ability to image small bone details is illustrated by a demanding task of intraoperative 3D-RX regarding bone detail: the imaging of the middle ear.[1] In the middle ear there are three ossicles: the stapes, the malleus and the incus which all have sub millimetre dimensions. Images of the latter two are used to visually compare the image quality of 3D-RX and CT and relate these to the estimated minimal detectable bone diameter. For the visual evaluation the images are automatically registered in order to facilitate the comparison.[18]

2.4 Results

A. Spatial resolution

![Diagram](image)

Fig. 3. The full width at half maximum of the Point spread function of 3D-RX and MSCT. The markers are the average of 6 measurements while the error bars represent the mean plus and minus one standard deviation.
In Fig. 3 the FWHM of the PSF of the 3D-RX system in X, Y and Z direction is shown as function of the magnification of the XRII and the number of projections. The FWHM of the PSF decreases for a smaller FOV and increases for a smaller number of projections. The latter effect is due to courser angular sampling, which introduces some blur and artifacts. For a smaller FOV the FWHM of the PSF decreases due to the higher spatial resolution of the XRII images in combination with a sharper reconstruction kernel. The difference between the FWHM of the PSF for 3D-RX in the X and Y direction is due to the absence of Parker weighting (see also Fig. 2).[19] At the highest magnification of the image intensifier and the maximal number of projection images, the FWHM of the in-plane PSF of the 3D-RX is comparable to the FWHM of the CT scanner with the CT-E protocol and considerably better than the FWHM of the other CT protocols (Fig. 3).

B. Contrast and noise

![Contrast to Noise ratio](image)

**Fig 4.** Contrast to Noise ratio's of bone-tissue and air-tissue contrasts, measured for the 3D-RX and MSCT with different settings. The margin error of the CNR measurements is negligible and was therefore omitted.

In Fig. 4 the results of the CNR measurements are shown. The CNR of 3D-RX decreases with a decreasing FOV and a decreasing number of projection images, due to the increase in noise in these situations. For 3D-RX the CNR of bone relative to tissue is larger than that of tissue relative to air which is due to the higher attenuation of bone relative to tissue at low tube voltages. The CNR of air-tissue contrast and tissue-bone contrast for CT is approximately equal. The best CNRs of 3D-RX can contend with the CNRs obtained with the CT-E and CT-S protocols.
C. Effective Dose

For a typical head examination the dose for CT is higher than for 3D-RX (Table 4) mainly because of the higher kV settings used in CT (120 kV versus 60-80 kV). The dose of the CT-E protocol is larger than the dose of the CT-O protocol. This is because of the higher mAs value of the CT-E protocol that was chosen to compensate for the inefficient use of radiation when a very narrow collimation (2*0.5 mm) is used. Without this higher mAs value, noise would be substantial when a narrow slice width, additional collimation pins and a sharp reconstruction kernel are used.[20]

Table 4: Estimated effective dose of each 3D-RX and CT setting for a head scan.

<table>
<thead>
<tr>
<th></th>
<th>3D-RX</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>12 inch (31 cm)</td>
<td>9 inch (23 cm)</td>
<td>6 inch (17 cm)</td>
<td></td>
</tr>
<tr>
<td>High quality</td>
<td>0.19 mSv</td>
<td>0.11 mSv</td>
<td>0.07 mSv</td>
<td></td>
</tr>
<tr>
<td>Medium quality</td>
<td>0.09 mSv</td>
<td>0.05 mSv</td>
<td>0.04 mSv</td>
<td></td>
</tr>
<tr>
<td>Low quality</td>
<td>0.04 mSv</td>
<td>0.02 mSv</td>
<td>0.01 mSv</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>MSCT</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT ear</td>
<td>0.99 mSv</td>
</tr>
<tr>
<td>CT orbit</td>
<td>0.40 mSv</td>
</tr>
</tbody>
</table>

D. Prediction of bone detail detectability

Fig. 5. Relation between the actual object diameter and the SNR calculated for the three clinically used 3D-RX settings and their MSCT equivalents: 6" High Quality-CT Ear, 9" Medium Quality-CT Skeleton, and 12" Medium Quality-CT orbit.
Fig. 6. Minimal detectable bone detail diameter, average FWHM of the PSF in the X-Y plane and voxel size.

Fig. 5 shows the SNR of a bone detail in tissue as a function of its diameter for three clinically used 3D-RX and CT settings. Fig. 6 shows the size of the minimally detectable bone detail at a SNR of 10 as well as the voxel size and the average FWHM of the PSF in the X-Y plane.

For 3D-RX the use of an increasing number of projection images makes smaller objects visible, as expected, due to a better CNR of the reconstructed images. The use of a larger magnification hardly increases the detectability of small objects, as the higher resolution obtained with the larger magnification is annulled by the higher noise level. The size of the minimally detectable bone detail in CT-E and CT-S is nearly equal to the 3D-RX equivalent protocols, which are High Quality 6” and Medium Quality 9”, respectively. The minimally detectable bone detail for the CT-O is smaller than its 3D-RX equivalent. This is due to the relatively large slice thickness for this protocol that results in a higher SNR for the detail and the fact that the cylindrical detail is perpendicular to the scan plane in this example.
Fig. 7. Pre-operative Coronal slices of CT ear (upper row) and intra-operative 3D-RX (6" High Quality, lower row). The slices above each other are approximately equally positioned by automatic registration, [16] the arrows indicate the malleus M, the incus (I), and the cochlear implant (Cl). The 3D-RX images are acquired during cochlear implantation which explains the liquid in the middle ear cavity and auditory canal.

In Fig. 7 a pre-operative CT and an intra-operative 3D-RX scan of the same patient during cochlear implantation are shown. The objective measures for image quality are comparable (i.e. CNR and FWHM of the PSF). Consequently, the ossicles, the incus and the malleus, all of sub millimetre size, are equally well visible in both scans even though the metal cochlear implant electrode array causes some distortions in the 3D-RX images. The effective dose of the CT examination is an order of magnitude larger than for 3D-RX, while the image quality is comparable. The dose difference can be explained by the differences in tube voltage, the smaller size of the irradiated volume in the case of 3D-RX and the inefficient use of CT detectors when the CT-E protocol is used.
2.5 Discussion

In this study we determined the spatial resolution, the CNR and the dose of a mobile 3D-RX system in comparison to that of a CT scanner. Additionally, we predicted detectability of small bone details in tissue from contrast to noise ratio and resolution measurements for protocols that are of relevance in surgical applications of the system.

The mean in-plane spatial resolution of 3D-RX is under optimal circumstances (HQ 6”) better than that obtained with two of the three CT protocols. This is to be expected because the 2D detector resolution of 3D-RX is better than that of CT and thus a higher 3D resolution is expected.[21] The in-plane spatial resolution of 3D-RX using the HQ 6” protocol is comparable to that of the CT-E protocol. However, additional collimation pins are inserted at the expense of reduced dose efficiency for this CT protocol. [10; 11]

3D-RX reconstructions with a smaller voxel size and a larger reconstruction matrix than the currently applied 256\(^3\) matrix, in combination with a sharper kernel, would result in an even better spatial resolution. We did not evaluate this option in this study because it is rarely used in clinical practice due to the higher noise levels introduced by the sharper kernel and increased reconstruction time.

The CNRs for 3D-RX bone imaging are under optimal circumstances (HQ, 6”) comparable to the CNRs obtained for the CT-E protocol. Higher CNRs are obtained with CT for the CT-S and CT-O protocol. In the latter case this is at the expense of a lower resolution in the Z-direction. Calculations showed that in the CT imaging protocols the dose is 2 to more than 50 times higher than those of the 3D-RX protocols. The higher dose of CT is due to the larger volume scanned, the higher kV and the reduced dose efficiency when a narrow collimation and/or additional collimation pins are used.

With CT-O the estimated size of the bone details that can be detected are smaller than those with the corresponding 3D-RX protocol. This is mainly due to the large slice thickness causing the noise levels to be reduced. The CT-E and CT-S protocols match their clinical 3D-RX equivalents with respect to minimally detectable bone detail.

The incus and malleus, which are objects with sub millimetre dimensions, can be visualized equally well with both systems (Fig. 7). This can be explained by the fact that both CNR and FWHM of the PSF are comparable for both systems. Moreover, the order of magnitude of the size of the incus and malleus corresponds to our theoretically predicted minimally detectable bone detail size of approximately 0.3 mm.
If one requires the visibility of bone details with a diameter larger than approximately 0.5 mm, all magnifications and qualities can be used. This argues for less projection images and thus faster reconstruction and lower dose, than the currently used 3D-RX protocols (Table 3). However, the measurements in this study were performed under optimal circumstances and with one particular system. In practice, (partly) scanned objects that are in the scan region but outside the field of view and metal artifacts, affect the image quality negatively. Therefore a somewhat higher image quality may be needed in practice.

Finally, the comparisons in this study are based on preset and clinically used protocols for both 3D-RX and CT. For 3D-RX there are opportunities to improve imaging characteristics such as application of flat panel detectors and correction of reconstruction artifacts introduced by the half circular trajectory of the image intensifier.[8; 9; 11; 20]

## 2.6 Conclusion

With the phantom-based approach presented in this paper the most relevant imaging parameters of 3D-RX and CT protocols were estimated and compared. The image quality of 3D-RX with the settings used in clinical practice approximates that of CT for high contrast imaging. 3D-RX provides sufficiently high spatial detail and sufficient contrast to distinguish air-tissue-bone transitions for traumatology, hand surgery, maxillofacial, sinus, ear and orbital surgery.

References


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