Reducing metal artefacts and radiation dose in musculoskeletal CT imaging
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CHAPTER 2

COMPUTED TOMOGRAPHY IMAGING OF A HIP PROSTHESIS USING ITERATIVE MODEL-BASED RECONSTRUCTION AND ORTHOPAEDIC METAL ARTEFACT REDUCTION: A QUANTITATIVE ANALYSIS.


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ABSTRACT

Objectives: To quantify the combined use of iterative model-based reconstruction (IMR) and orthopaedic metal artefact reduction (O-MAR) in reducing metal artefacts and improving image quality in a total hip arthroplasty (THA) phantom.

Methods: Scans acquired at several dose levels and kVps were reconstructed with filtered back-projection (FBP), iterative reconstruction (iDose) and IMR, with and without O-MAR. Computed Tomography (CT) numbers, noise levels, signal-to-noise-ratios and contrast-to-noise-ratios were analysed.

Results: Iterative model-based reconstruction results in overall improved image quality compared to iDose and FBP (p<0.001). Orthopaedic metal artefact reduction is most effective in reducing severe metal artefacts improving CT number accuracy by 50%, 60% and 63% (p<0.05) and reducing noise by 1%, 62% and 85% (p<0.001) whereas improving signal-to-noise-ratios by 27%, 47% and 46% (p<0.001) and contrast-to-noise-ratios by 16%, 25% and 19% (p<0.001) with FBP, iDose and IMR respectively.

Conclusions: The combined use of IMR and O-MAR strongly improves overall image quality and strongly reduces metal artefacts in the CT imaging of a total hip arthroplasty phantom.

Keywords: CT; metal artefacts; iterative model-based reconstruction; O-MAR; hip phantom.
INTRODUCTION

Cross-sectional imaging provides important clinical information regarding revision of surgically implanted metal-on-metal (MoM) hip prostheses in large head MoM-related disease of the hip capsule [1,2]. In computed tomography (CT) imaging, metal hip prostheses used for total hip arthroplasties (THA) can cause severe artefacts caused by photon-starvation, beam-hardening and scatter [3]. This strongly deteriorates image quality and significantly reduces the performance of CT in the detection of all sorts of prosthesis-related pathology, such as pseudo-tumours, capsular reactions and other soft tissue and bone pathologies.

Several metal artefact reduction (MAR) techniques have shown to be valuable in the suppression of metal artefacts thereby improving image quality [4–9]. Most MAR approaches are based on replacing the projection data affected by metal with projection data obtained from unaffected regions. Other more advanced iterative metal artefact reduction approaches are based on linear interpolation, algebraic reconstruction or frequency-splitting [5]. Dual-energy CT or virtual monochromatic imaging approaches reduce beam-hardening effects caused by polychromatic X-ray beams. However, their inability to suppress metal artefacts caused by photon-starvation especially in large metallic implants impedes its effectiveness in MAR. The orthopaedic metal artefact reduction algorithm O-MAR is an iterative metal artefact reduction algorithm specially developed for CT-imaging of large metal orthopaedic implants. In O-MAR the metal only images, the tissue classified images and input images are projected forward to generate corresponding sinogram data. In this way, a metal only sinogram is compared to the original sinogram and is used to correct for metal parts. Both streak and darkening effects are mitigated using O-MAR [10]. With O-MAR, Hounsfield Units (HUs) are corrected towards baseline levels and contrast-to-noise-ratios (CNRs) are boosted thereby improving image quality [11–14]. Recently, we showed in a similar phantom study executed on a 64-slice CT-scanner that O-MAR significantly reduces metal artefacts using partial iterative reconstruction technique iDose4 [15]. The iterative model-based reconstruction (IMR) algorithm takes data statistics, image statistics and system models into account in order to produce the image that optimally reflects attenuation distributions. Recent studies show that model-based iterative reconstruction techniques are able to reduce image noise up to 75%-83% and radiation dose up to 75%-80% compared to standard reconstruction techniques [16–21]. Further improvements in reconstruction algorithms, such as IMR, may even further improve the effectiveness in metal artefact reduction.
Therefore, the aim of this study was to quantify the effect of IMR combined with O-MAR in reducing metal artefacts and improving image quality in a total hip arthroplasty phantom on outcome values such as CT-number accuracy, noise values, signal-to-noise-ratios (SNRs) and CNRs.

MATERIALS AND METHODS

Study design, image acquisitions and reconstructions
Phantom scans were acquired with and without the insertion of a unilateral MoM hip prosthesis. All scans were reconstructed with filtered back-projection (FBP), iDose4 and IMR with and without O-MAR and analysed with a standardized measurement template mask. The water-filled phantom was scanned on a Philips Brilliance iCT 256-slice CT-scanner where acquisitions without the insertion of a prosthesis served as a reference. Scans were acquired at half, standard and high-dose with CT dose index (CTDI) of respectively 10.0, 20.0 and 30.0 mGy at 100, 120 and 140-kVp resulting in 9 different acquisitions. Static scan parameters were 64 × 0.625 mm collimation, 0.9 mm slice thickness with 0.45 mm increment, 330 mm field-of-view, 0.398 pitch used in our current clinical practice, 512 × 512 image matrix, high resolution and a rotation time of 1.0s. An IMR prototype reconstruction system (version R11) was used. Noise reduction levels and filter types were matched for standard, iterative and the model-based iterative reconstruction. Philips’ iDose4 can be used in 7 different levels of noise suppression. iDose4 level 4 is used in our current clinical practice and since this is the middle level of noise suppression for iDose4, for fair comparison also the middle level of Philips’ IMR, level 2 was chosen. A hard or sharp filter was chosen in order to increase the contrast and enhance edges between hard and soft materials in the CT imaging of metallic components. Therefore for iDose4 and FBP the hard filter D was used whereas in case of IMR filter Sharp Plus was chosen.
Hip phantom and prosthesis

The custom made hip phantom is made of polymethyl methacrylate (PMMA) with dimensions of 320 mm width, 130 mm height and 290 mm depth. To represent more clinical relevant phantom dimensions, the sagittal diameter of 130 mm was increased to 190 mm. Six PMMA shields in total with a thickness of 10 mm were placed on top and below the phantom. These phantom size calculations were derived from the coronal diameter of 320 mm and a water-equivalent diameter of 29.15 cm, representative for a patient with a body-mass-index (BMI) of 25 using a formula of Menke et al. 2005 [22] (Fig. 1). A large head Metal-on-Metal prosthesis (Biomet Warsaw, Ontario) was inserted with a titanium-aluminium-vanadium (Ti$_6$Al$_4$V) stem. The head and the cement-less ReCap-M2a-Magnum™ cup are made of a cobalt-chrome-molybdenum alloy. The prosthesis was fixated with custom-made PMMA moulds to prevent movement. The phantom contains 18 cylindrical hydroxyapatite/calcium carbonate pellets representing bone with a certified calibration and documented tolerance of ± 0.5% placed at the Gruen zones and DeLee and Charnley zones [23,24]. On each side 9 pellets with a height and diameter of 10 mm were fixated onto PMMA pillars to ensure correct alignment of the pellets at the middle of the phantom.

Figure 1: a) The hip phantom with dimensions. A water-filled phantom was used made of polymethyl methacrylate (PMMA) containing a large head Metal-on-Metal (MoM) prosthesis, shown in Figure 1b, surrounded by 18 hydroxyapatite/calcium carbonate pellets representing bone. Six PMMA shields in total with a thickness of 10mm are placed on top and beneath the original phantom box to represent more clinical relevant phantom dimensions. The cylindrical pellets with a height and diameter of 10 mm were mounted on PMMA pillars.
Quantitative analysis

Image quality, degree of metal artefacts and effectiveness in metal artefact reduction was quantified by analysing CT numbers in HU, noise levels, SNRs and CNRs within fixed regions of interest (ROIs). A standardized measurement template mask was used for each scan in order to enhance the reliability of the measurements. The coronal slice exactly aligned at the middle of the pellets and prosthesis was stored as a Digital Imaging and Communications in Medicine file at a Philips Intellispace Portal Workstation V6.0.3.1220 for each of the 108 datasets. A single coronal slice was loaded into ImageJ version 2.64 where a template was manually created with 9 left pellet ROIs (L0-L8) and 9 right pellet ROIs (R0-R8) (Fig. 2a). Matlab® 2014b was used to perform the actual quantitative measurements. Pellet ROIs had a diameter of 14.7 pixels or 6.6 mm thus minimalizing partial volume effects. Amount of pixels of background ROIs was adapted to the amount of pixels of pellet ROIs (Fig. 2b). Noise, with and without the insertion of a prosthesis, was measured by calculating the standard deviation of pixels in a ROI placed in the pellet and background. Signal-to-noise-ratios were calculated by dividing CT numbers of pellet ROIs by the noise or standard deviation in the pellet ROIs. Contrast-to-noise-ratios were calculated by subtracting the average HU value of the local background from the average HU value of the pellet and subsequently dividing this by the noise of the local background ROI. Metal artefact correction by O-MAR for CT numbers, noise values, SNRs and CNRs were calculated as follows:

\[
\text{Deviation}_{\text{without O-MAR}} = |\text{reference value} - \text{value without O-MAR}| \quad (1)
\]

\[
\text{Deviation}_{\text{with O-MAR}} = |\text{reference value} - \text{value with O-MAR}| \quad (2)
\]

\[
\text{Correction by O-MAR (\%)} = \left( 1 - \frac{\text{Deviation}_{\text{with O-MAR}}}{\text{Deviation}_{\text{without O-MAR}}} \right) \times 100 \quad (3)
\]

Statistical analysis

Statistical analysis was performed by means of repeated measures ANOVA (full factorial, type III), with three within-subject factors notably pellet (‘L0’, ‘L1’, ‘L5’, ‘L7’, ‘L8’) representing different degrees in metal artefacts, reconstruction technique (‘FBP’, ‘iDose^4’, ‘IMR’) and O-MAR (‘off’, ‘on’), generalizing to scan protocol containing the 9 different acquisitions. Additionally a separate analysis was performed for severe metal artefacts in pellet L8 by means of two within-subject factors notably reconstruction technique and O-MAR. Greenhouse-Geisser p-values were interpreted. A 2-sided alpha of 5% was used as significance level.
Figure 2: Coronal CT-slice acquired at 140-kVp and standard dose reconstructed with iterative model-based reconstruction (IMR). Fig. 2a illustrates the measurement template mask including the regions-of-interest (ROIs) of the 18 pellets, 9 left pellets (L0-L8) and 9 right pellets (R0-R8). Fig. 2b shows a single pellet with the inner pellet ROI 1 and outer background ROI 2. The number of pixels of the background ROI 2 is adapted to the inner pellet ROI 1.

Figure 3: A 140-kVp standard dose acquisition reconstructed with iDose\(^4\) level 4 and without the use of orthopaedic metal artefact reduction (O-MAR). Pellet L8, located medial to the head of the metal-on-metal prosthesis, is invisible due to severe metal artefacts.
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RESULTS

Without prosthesis

Iterative model-based reconstruction resulted in lower reference CT numbers compared to FBP and iDose\(^4\) (\(p<0.001\)). Noise values, SNRs and CNRs showed little variation across kVps at similar dose levels and were averaged per CTDI. With IMR, noise levels at CTDIs of 10.0, 20.0 and 30.0 mGy were respectively 46%, 37% and 34% lower compared to iDose\(^4\) and respectively 83%, 77% and 73% lower compared to FBP (\(p<0.001\)). With IMR, at CTDIs of 10.0, 20.0 and 30.0 mGy, SNRs were a factor

![Graphs showing CT numbers, noise values, SNRs, and CNRs for pellets L0, L1, L5, L7, and L8 at 140-kVp and standard dose reconstructed with FBP. Reference values, without the insertion of a prosthesis, are illustrated in red. HUs, SNRs, and CNRs decrease and noise values increase compared to reference values due to metal artefacts. Orthopaedic metal artefact reduction (O-MAR) has no effect on the by metal artefact unaffected pellet L0 and is most effective for the most severe artefacts in case of pellet L8.](image-url)

Figure 4: CT numbers, noise values, signal-to-noise-ratios (SNRs) and contrast-to-noise-ratios (CNRs) for pellets L0, L1, L5, L7 and L8 at 140-kVp and standard dose reconstructed with FBP. Reference values, without the insertion of a prosthesis, are illustrated in red. HUs, SNRs, and CNRs decrease and noise values increase compared to reference values due to metal artefacts. Orthopaedic metal artefact reduction (O-MAR) has no effect on the by metal artefact unaffected pellet L0 and is most effective for the most severe artefacts in case of pellet L8.
3.0, 2.5 and 2.3 higher compared to iDose⁴ and a factor 5.6, 4.0 and 3.4 higher compared to FBP (p<0.001). Subsequently with IMR, at CTDIs of 10.0, 20.0 and 30.0 mGy, CNRs were a factor 3.2, 2.6 and 2.3 higher compared to iDose⁴ and a factor 5.6, 4.0 and 3.4 higher compared to FBP (p<0.001). The SNRs and CNRs at different kVp settings and reconstruction techniques improved proportionally with respect to radiation dose. Reference CT numbers, noise values, SNRs and CNRs without the insertion of a prosthesis are illustrated in Fig. 6.

With prosthesis

The 9 pellets (R0-R8) on the right side of the hip phantom were unaffected by the MoM-prosthesis located on the contralateral side. Therefore we only focused on the left pellets (L0-L8) (Fig. 3). The most affected pellet was obviously pellet L8 located close to the head of the MoM-prosthesis. Pellet L0 (unaffected), pellet L5 –L7 (mild/moderate artefacts), and pellet L8 (severe artefacts) were further analysed and chosen based on visual assessment. Additionally, pellet L1 was investigated since this pellet seems to be affected by a white streak artefact at the lateral side of the large head of the MoM-prosthesis.

![Diagram showing different reconstruction techniques](image)

Figure 5: 140-kVp standard dose images reconstructed with filtered back-projection (FBP), iDose⁴ and iterative model-based reconstruction (IMR) and with and without the use of orthopaedic metal artefact reduction (O-MAR). Pellet L0, L1, L7 and L8 surrounding the head of the prosthesis are shown. Fig. 5 a, b and c show the reconstructed images obtained with respectively FBP, iDose⁴ and IMR. Images in Fig. 5 d, e and f are reconstructed with O-MAR combined with FBP, iDose⁴ and IMR. Pellet L8 is highly influenced by severe metal artefacts due to the large metal head and cannot be visualized without O-MAR for all reconstruction techniques. Pellet L8 is invisible without the use of O-MAR (a, b and c). Using O-MAR, pellet L8 can be visualized but only when combined with iDose⁴ or IMR.
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CT numbers

Fig. 4 illustrates deviated CT numbers, noise values, SNRs and CNRs for pellet L0, L1, L5, L7 and L8 using FBP at 140-kVp and standard dose. Pellet L8 was the only pellet where O-MAR consistently decreases deviations in CT numbers towards baseline values for all reconstruction techniques, kVp-settings and dose levels. CT numbers of pellet L7 were corrected in most of the used acquisition parameters and reconstruction techniques. We further analysed the effects of O-MAR combined with FBP, iDose4 and IMR while mainly focussing on the most severe metal artefacts in pellet L8. The effect of O-MAR combined with the different reconstruction techniques is illustrated in Fig. 5. O-MAR suppresses artefacts and makes pellet L8 visible again when combined with iDose4 or IMR. In case of the most severe artefacts in L8, O-MAR improved CT-number accuracy by 50%, 60% and 63% towards baseline values for respectively FBP, iDose4 and IMR (p<0.05) (Tab. 1, Fig. 6). Pellets L0, R0, L4 and R4 were unaffected by metal artefacts since these pellets were located outside the axial scan field of the prosthesis. CT numbers of these four pellets were similar with and without the use of O-MAR, which confirms the claim that O-MAR does not operate in areas unaffected by metallic artefacts. Deviations in CT numbers compared to reference values increase with the use of O-MAR for pellet L1 and L5.

Noise values

Noise of the pellets was measured by calculating the standard deviation of pixels in the pellet ROIs. Metal artefacts caused an increase in noise values by increasing the standard deviation, especially for pellet L1, L7 and L8 (Fig. 4). Noise values of pellets L0, L4, R0 and R4 were slightly higher compared to reference values without the insertion of a prosthesis. O-MAR is capable of reducing noise values when combined with FBP, iDose4 and IMR. Besides the fact that overall noise values were substantially lower for IMR images, O-MAR was most effective in decreasing noise when combined with IMR with corrections in noise of the most severe artefacts with 1 %, 65 % and 85 % for respectively FBP, iDose4 and IMR (p<0.001) (Tab. 1, Fig. 6).

SNRs and CNRs

The absolute HU correction by O-MAR was clearly greatest for L8, where HU values of other pellets remained relatively unaffected. Despite the fact that O-MAR did not improve CT numbers of all affected pellets to reference values without the insertion of a prosthesis, SNRs and CNRs increased significantly for other pellets also besides pellet L8 (Fig. 4). SNRs of pellet L8 were negative where CNRs of pellet L8 were low and mostly below 1.0, therefore making the distinction between the pellet and
Figure 6: CT numbers, noise values, SNRs, and CNRs in severe metal artefacts regarding pellet L8 at 100, 120 and 140-kVp at CTDIs of 10.0, 20.0 and 30.0mGy using FBP, iDose⁴ and IMR. Reference values, without the insertion of a prosthesis, are illustrated in red.
the surrounding water impossible (Fig. 5(a-c)). Besides the fact that overall CNRs were substantially higher for IMR images, O-MAR was most effective in correcting CNR deviations when combined with IMR with absolute CNR corrections of the most severe artefacts with 0.7 ± 0.3, 2.3 ± 0.5 and 4.8 ± 0.8 for respectively FBP, iDose$^4$ and IMR (p<0.001). Relative improvements in CNRs towards reference CNRs without the insertion of a prosthesis were 13 %, 23 % and 19% for FBP, iDose$^4$ and IMR (Tab. 1, Fig. 6). O-MAR resulted in a CNR improvement in all cases thereby revealing pellet L8, especially when combined with iDose$^4$ or IMR. The MoM-prosthesis heavily decreased CNRs of L8 for all reconstruction techniques. Despite the fact that O-MAR was incapable of bringing SNRs and CNRs back to reference values of unaffected pellets, greatest absolute improvements were seen when combined with IMR (p<0.001). With the use of O-MAR and IMR, SNRs and CNRs of pellet L8 reached values of unaffected pellets in FBP reconstructions (Fig. 6).

O-MAR proved to be most effective in case of the most severe artefacts such as in pellet L8, when combined with IMR and for 140-kVp results (Tab. 1). O-MAR combined with IMR showed an average HU correction for pellet L8 of 60%, 64% and 66% for respectively 100, 120 and 140-kVp results and 64%, 63% and 63% for respectively 10.0, 20.0 and 30.0 mGy. Noise was corrected for with 88%, 79% and 91% for respectively 100, 120 and 140-kVp results and 80%, 87% and 86% for respectively 10.0, 20.0 and 30.0 mGy. SNR corrections were 35%, 49% and 52% for

| Table 1: Corrections in CT-numbers, noise values, signal-to-noise-ratios (SNRs) and contrast-to-noise-ratios (CNRs) of pellet L8 by orthopaedic metal artefact reduction (O-MAR) towards reference values using filtered back-projection (FBP), iDose$^4$ and iterative model-based reconstruction (IMR) for the 9 different acquisitions. Metal artefact correction by O-MAR for CT numbers, noise values, SNRs and CNRs were calculated using equations 1, 2 and 3. |
|---|---|---|---|---|---|---|---|---|---|---|---|
| KVP | CTDI (mGy) | HU (%) | HU (%) | HU (%) | Noise (%) | Noise (%) | Noise (%) | SNR (%) | SNR (%) | CNR (%) | CNR (%) |
| 100 | 10.0 | 42 | 55 | 63 | -13 | 71 | 89 | 31 | 37 | 35 | 23 | 20 | 14 |
| 20.0 | 48 | 54 | 59 | 7 | 41 | 84 | 19 | 30 | 35 | 4 | 20 | 20 |
| 30.0 | 45 | 55 | 60 | 7 | 50 | 91 | 18 | 34 | 41 | 4 | 17 | 19 |
| 120 | 10.0 | 36 | 60 | 64 | -6 | 33 | 69 | 26 | 47 | 49 | 12 | 28 | 19 |
| 20.0 | 55 | 63 | 65 | 3 | 63 | 83 | 29 | 44 | 47 | 18 | 32 | 22 |
| 30.0 | 54 | 62 | 64 | 10 | 58 | 84 | 22 | 47 | 49 | 11 | 24 | 18 |
| 140 | 10.0 | 69 | 64 | 66 | -4 | 81 | 83 | 47 | 54 | 52 | 48 | 35 | 26 |
| 20.0 | 51 | 62 | 65 | 7 | 89 | 95 | 27 | 55 | 55 | 13 | 24 | 17 |
| 30.0 | 51 | 63 | 66 | 2 | 74 | 84 | 24 | 53 | 54 | 10 | 27 | 20 |
| Average (%) | 50 ± 9 | 60 ± 4 | 63 ± 3 | 1 ± 8 | 62 ± 19 | 85 ± 7 | 27 ± 9 | 45 ± 9 | 46 ± 8 | 16 ± 13 | 25 ± 6 | 19 ± 3 |
respectively 100, 120 and 140-kVp results and 45%, 46% and 48% for respectively 10.0, 20.0 and 30.0 mGy. CNR corrections were 17%, 20% and 21% for respectively 100, 120 and 140-kVp results and 19%, 20% and 19% for respectively 10.0, 20.0 and 30.0 mGy.

**DISCUSSION**

Our study shows that iterative model-based reconstruction (IMR) significantly improves image quality compared to standard (FBP) and iterative reconstruction techniques (iDose⁴) at all dose levels and tube voltages with lower noise values and higher SNRs and CNRs (p<0.001). We defined metal artefacts as deviations on (unaffected) mean CT numbers in HU, noise [HU], SNRs and CNRs. O-MAR reduces metal artefacts, is most effective in case of severe artefacts and when combined with IMR based on corrections in deviated HU values (p<0.05), noise values (p<0.001), SNRs (p<0.001) and CNRs (p<0.001). We observed that the combination of O-MAR and (model-based) iterative reconstruction optimizes the image quality. This can be explained by the fact that the O-MAR algorithm post-processes the projection data and provides more regular attenuation profiles before image reconstruction. These more regular attenuation profiles can improve the general performance of iDose⁴ and IMR.

CT numbers were significantly lower for IMR compared to FBP and iDose⁴ where CT numbers of FBP and iDose⁴ were not significantly different. This can be explained by differences in filter type where filter D was used for FBP and iDose⁴ and Sharp Plus was used for IMR. The Sharp Plus filter used in IMR reconstructions uses edge enhancement at the interfaces between structures, which can influence CT numbers in small objects such as pellets. With regard to HU values, O-MAR did not show a positive effect for mild/moderate artefacts with small HU deviations in most cases. This effect was observed for pellets L1, L5 and in most acquisitions for L7. CT numbers, noise values, SNRs and CNRs of pellet L0 and L4 confirmed that O-MAR does not modify the data in regions not affected by metal artefacts. This knowledge subsequently seems to enable more reliable bone mineral density measurements in CT in the acetabular region (L0) that could be of use in THA revision surgery [25]. O-MAR decreased deviations in HUs in severe metal artefacts affecting pellet L8 (p<0.001) in all cases. O-MAR did decrease noise and improved SNRs and CNRs of other affected pellets in most cases. With O-MAR, Pellet L8 was revealed in all cases.
when reconstructed with iDose\(^4\) or IMR, which illustrates differences in effectiveness of O-MAR when combined with (model-based) iterative reconstruction techniques. With FBP, overall noise values were high and remained relatively unchanged when combined with O-MAR due to the fact that FBP itself is incapable of properly handling the noise, especially in severe artefacts. In general, the nature and the position of the artefact were similar when using FBP, iDose\(^4\) and IMR with and without the use of O-MAR. However we did see differences in image quality due to a significant reduction of noise levels using iDose\(^4\) and IMR.

Several metal artefact reduction techniques have shown to be valuable in the suppression of metal artefacts and thereby improving objective and subjective image quality with improved diagnostic confidence as described earlier. O-MAR resulted in an improved CT number accuracy and reduced noise levels [11–14]. However, none of these studies evaluated the value of O-MAR combined with model-based iterative reconstruction. We observed similar positive effects of O-MAR on CT number accuracy and noise compared to other O-MAR studies. When using O-MAR 140-kVp is advised. Using 140-kVp not only decreases the impact of beam-hardening but also decreases statistical noise, which benefits the O-MAR algorithm [10]. We also observed the highest effectiveness in MAR using 140-kVp compared to 100 and 120-kVp. Boomsma et al. [15] quantified the value of O-MAR varying several scan-parameters using iDose\(^4\). They found that iDose\(^4\) did not influence the effects of O-MAR but that it can result in scanning with a lower dose due to improvements in CNRs. They found that O-MAR is capable of reducing severe artefacts by 32% based on CT numbers using iDose\(^4\), where we found a reduction of HU deviations by O-MAR of 60% using iDose\(^4\) and 63% using IMR. They also found no improvement in CNRs by O-MAR in the most affected pellets in pellet L8 unlike our results. A possible explanation for the observed differences can be the use of a different CT-scanner, adaptations in phantom dimensions, reduced amount of water inside the phantom, a different measurement template and the use of a different filter type.

Our study has some limitations. We only performed quantitative analyses using a standardized measurement template. Additional subjective image quality scoring could give more insights in the clinical usefulness and additional value of the CT-techniques. The rectangular shape of the phantom with sharp edges could be improved where an oval shaped phantom would represent more realistic dimensions with regard to the human pelvis. The hydroxyapatite/calcium carbonate pellets with a high density result in high contrast values between the pellets and its background. Adding pellets with different densities or adding soft tissue can give more insights in
the possible additional clinical value in patients. We addressed metal artefacts and the degree of MAR by quantifying CT numbers, noise, SNRs and CNRs. Noise was measured by calculating the standard deviation of pixels in a ROI. We are aware of the fact that both noise and artefacts will influence the standard deviation in a ROI and that it is priori not clear which component is dominant. However, we showed that O-MAR has no influence on CT images without metal artefacts, with similar relatively small standard deviations with and without the use of O-MAR. From those results we conclude that the measured “noise” reduction by O-MAR is mainly due to artefact reduction resulting in a lower standard deviation. Furthermore, despite the fact that CT numbers, noise, SNRs and CNRs are justified and commonly used image quality parameters, additional analyses of the noise-power spectrum (NPS) and modulation transfer function (MTF) can give more insights in the nature of the noise and spatial resolution in metal artefacts using model-based iterative reconstruction and O-MAR.

O-MAR reduces metal artefacts, is most effective in severe artefacts and when combined with IMR compared to iDose⁴ and FBP. Image quality with IMR is superior compared to FBP and iDose⁴. The combined use of IMR and O-MAR improves overall image quality by strongly reducing metal artefacts, decreasing noise and improving CT number accuracy, signal-to-noise-ratios and contrast-to-noise-ratios in the CT-imaging of a MoM hip prosthesis phantom. This will likely improve the diagnostic accuracy in detecting soft tissue and bone pathology in patients after total hip arthroplasty.

REFERENCES


Chapter 2


