TOPICAL REVIEW

Metal artifacts in computed tomography for radiation therapy planning: dosimetric effects and impact of metal artifact reduction

To cite this article: Drosoula Giantsoudi et al 2017 Phys. Med. Biol. 62 R49

View the article online for updates and enhancements.

Related content
- CT metal artifact reduction method correcting for beam hardening and missing projections
  Joost M Verburg and Joao Seco
- Calibration of CT Hounsfield units for proton therapy treatment planning: use of kilovoltage and megavoltage images and comparison of parameterized methods
  L De Marzi, C Lesven, R Ferrand et al.
- An evaluation of three commercially available metal artifact reduction methods for CT imaging
  Jessie Y Huang, James R Kems, Jessica L Nute et al.

Recent citations
- Dosimetric characterization of carbon fiber stabilization devices for post-operative particle therapy
  E. Mastella et al
Topical Review

Metal artifacts in computed tomography for radiation therapy planning: dosimetric effects and impact of metal artifact reduction

Drosoula Giantsoudi¹, Bruno De Man², Joost Verburg¹, Alexei Trofimov¹, Yannan Jin², Ge Wang³, Lars Gjesteby³ and Harald Paganetti¹

¹ Department of Radiation Oncology, Massachusetts General Hospital and Harvard Medical School, Boston, MA, United States of America
² General Electric Global Research Center, Niskayuna, NY, United States of America
³ Biomedical Imaging Center, Department of Biomedical Engineering, Rensselaer Polytechnic Institute, Troy, NY, United States of America

E-mail: dgiantsoudi@mgh.harvard.edu

Received 3 August 2016, revised 20 October 2016
Accepted for publication 8 December 2016
Published 21 March 2017

Abstract

A significant and increasing number of patients receiving radiation therapy present with metal objects close to, or even within, the treatment area, resulting in artifacts in computed tomography (CT) imaging, which is the most commonly used imaging method for treatment planning in radiation therapy. In the presence of metal implants, such as dental fillings in treatment of head-and-neck tumors, spinal stabilization implants in spinal or paraspinal treatment or hip replacements in prostate cancer treatments, the extreme photon absorption by the metal object leads to prominent image artifacts. Although current CT scanners include a series of correction steps for beam hardening, scattered radiation and noisy measurements, when metal implants exist within or close to the treatment area, these corrections do not suffice. CT metal artifacts affect negatively the treatment planning of radiation therapy either by causing difficulties to delineate the target volume or by reducing the dose calculation accuracy. Various metal artifact reduction (MAR) methods have been explored in terms of improvement of organ delineation and dose calculation in radiation therapy treatment planning, depending on the type of radiation treatment and location of the metal implant and treatment site. Including a brief description of the available CT MAR methods that have been applied in radiation therapy, this article attempts to provide a comprehensive review on the dosimetric effect of the presence of CT metal...
artifacts in treatment planning, as reported in the literature, and the potential improvement suggested by different MAR approaches. The impact of artifacts on the treatment planning and delivery accuracy is discussed in the context of different modalities, such as photon external beam, brachytherapy and particle therapy, as well as by type and location of metal implants.

Keywords: computed tomography, metal artifacts, dose effect, radiation therapy, metal artifact reduction

(Some figures may appear in colour only in the online journal)

1. Introduction

Radiation therapy is one of the primary methods of cancer treatment worldwide with computed tomography (CT) being the most commonly used imaging method for treatment planning. The treatment planning process in radiation therapy typically requires accurate organ delineation, including the tumor volume and sensitive structures expected to be within or close to the treatment area, voxel-based dose calculation and optimization of beam intensities and/or orientations for external beam radiotherapy, or source location for brachytherapy. The accuracy and resolution of the CT image set used for treatment planning is crucial, since the lack of it can greatly affect all of these processes. Errors in the image reconstruction process may lead to erroneous definition of the target volume with adverse effects such as tumor recurrence, in case of underestimation, or increased radiation toxicity, in the case of overestimation of the target volume with unnecessary inclusion of healthy tissues within the treatment area. Inaccuracies in the Hounsfield Unit (HU) values result in error in electron density or stopping power estimate, which may significantly affect the dose calculation and optimization processes.

Advances in orthopedic surgery have resulted in significant increase in the number of patients receiving metal implants. In the United States for example, the number of total hip replacements among patients aged 45 and over doubled in ten years, from 2000 to 2010 (Wolford et al 2015). Considering that, according to statistical data from the American national cancer institute reporting, approximately 14% of men are diagnosed with prostate cancer at some point during their lifetime and that radiation therapy is a primary cancer treatment method worldwide, the number of patients affected by CT metal artifacts due to hip replacement in prostate radiation therapy is expected to increase significantly in the years to follow. Apart from the increasing number of prostate cancer patients with a hip prosthesis and the more common cases of gold or amalgam tooth fillings in head-and-neck patients, rods and screws are commonly present in spinal and para-spinal irradiations, since they are necessary for spinal stabilization after surgical removal of spinal tumors before irradiation. In treatment sites such as head-and-neck tumors more than one type of metal implants may be present, such as dental fillings and metal rods for stabilization of the cervical spine.

In the presence of metal implants in the field of view of a CT scanner, image artifacts are created due to three main effects: beam hardening, scatter and noise (De Man et al 1999, Barrett and Keat 2004). Polychromatic x-ray beams are used by CT scanners and the lower energy photons are more easily absorbed by matter, especially the high atomic number metallic implants. As a result, the average beam energy increases—the beam hardens—and the physics of transmission change. In extreme situations, involving large size, high density or high atomic number metal objects, increased photon absorption may lead to photon starvation, meaning insufficient number of photons, comparable to or lower than the noise level,
reaching the detectors. The signal at these detectors is typically dominated by Compton scattered x-rays. Compton scatter is the dominant type of interaction in CT. Scattering occurs throughout the patient, the path of the corresponding x-rays is altered and they hit the detector off the centerline of the incident beam. The combination of beam hardening and added scattered radiation in the measurements of the metal implant leads to dark streaks in the image along the axis of greatest attenuation, while the metal shows up as white. Another important effect is noise which contributes to artifacts in CT, especially when the photon flux reaching the detectors is significantly reduced due to strong attenuation. Photon detection assumes a Poisson distribution of photon flux, so low photon counts cause higher statistical errors, which present as thin dark and bright streaks in the image.

Current CT scanners include a series of correction steps for beam hardening, scattered radiation and noisy measurements. However, these correction steps are usually based on simplified physics models not accounting for cases of extreme density inhomogeneities. When high-density materials, such as metal implants, exist within or close to the treatment area, these corrections do not suffice, since they are not optimized to efficiently manage the extreme photon absorption by the metal object, leading to prominent artifacts. In the case of common implants, such as dental fillings, spinal stabilization implants or hip replacements, dark and bright starburst-streaking artifacts are generated, which degrade the capabilities of CT images to provide correct information about the HU and electron density of structures. Accurate organ delineation may become extremely challenging and erroneous electron density values are obtained, which can significantly compromise the calculation accuracy of dose distributions. In addition to streaking artifacts, the HU value of metal implants tend to saturate and this effect may become detrimental especially in particle therapy since the HU saturation results in inaccurate estimate of relative stopping power compromising the range and dose calculation accuracy. For large sized implants, such as hip replacements, or materials of high atomic number (high-Z), such as gold, the attenuation of the photon beam is so strong that almost no photons reach the detector (photon starvation) resulting in incomplete or missing data in the projection domain. The dynamic range of the CT scanner then stretches to its lower limit and, when it is back projected into the image, two major effects occur: (a) lower CT numbers are produced for the implants, since a relatively lower attenuation is registered and (b) noise is associated with the artifacts, in the form of bright borders surrounding them (Coolens and Childs 2003).

To compensate for these effects numerous efforts have been made to develop metal artifact reduction (MAR) methods and algorithms. Their effectiveness has been explored in terms of improvement of organ delineation and dose calculation accuracy in radiation therapy treatment planning, for different types of radiation treatment modalities, metal implant locations and treatment sites. This review focuses on the dosimetric effect of CT metal artifacts in treatment planning and the potential improvement suggested by different MAR algorithms.

Section 2 presents a brief description of the main principles of MAR algorithms that have been suggested. Sections 3–5 present the dosimetric effect of metal artifacts on CT-based radiation therapy treatment planning. Since different radiation therapy modalities and treatment sites may require specific treatment planning techniques and algorithms, which can be affected by the presence of CT metal artifacts in different ways, the available literature was reviewed based on the referred treatment modality, type and location of the metal implants and treatment sites.

Section 3 focuses on the dosimetric effect of CT artifacts on treatment planning and delivery of external photon beam radiation therapy. This section is further divided in four sub-sections based on the most common metal implants and the treatment sites where they are encountered: section 3.1 focuses on pelvic implants (hip prostheses) in the treatment of
prostate cancer, section 3.2 on dental implants in the treatment planning of head-and-neck and brain tumors, section 3.3 on spinal implants in the treatment of spinal tumors, and section 3.4 on Onyx embolization for the treatment of arteriovenous malformations (AVM).

Section 4 reports on the dosimetric effect of metal artifacts encountered in brachytherapy treatment of various treatment sites: intracavitary applicator treatment of gynecological (GYN) cancers and the effect of the CT artifacts caused by the metallic applicators, artifacts of permanent implants for low dose rate (LDR) lung and prostate brachytherapy treatments.

Section 5 summarizes the effect of CT artifacts on the dosimetry for particle therapy, including proton therapy, for the treatment sites that are mostly affected by this issue.

Analyzing the findings of numerous publications on this topic, the importance of an improved, automated MAR method becomes apparent. The necessary requirements for a CT MAR method to be effective for the purpose of accurate organ delineation and dose calculation is also discussed in the section 6 of this review, while the conclusions of this review are summarized in the final section 7.

2. CT MAR algorithms in radiation therapy treatment planning


With the sinogram (or projection) completion methods areas of data corrupted by the presence of the metal, are identified in the sinogram space. That data are then treated as missing and is replaced using interpolation routines based on the uncorrupted sinogram regions. CT image reconstruction, typically a filtered back projection (FBP) type algorithm, is then applied. These interpolation-based approaches may achieve significant MAR at the site of metallic implants. However, full removal of projections performed by these approaches is associated with loss of information and inaccuracies in the estimated completion data may lead to additional artifacts in the reconstructed images or loss of spatial resolution. Part of the lost data may include the exact location of edges between the high-density metal and regular-density soft tissues or bone, rendering the contouring of metal structures challenging and estimation of their size highly inaccurate. A first example of a sinogram completion method applied to CT image MAR is the algorithm suggested by Kalender et al (1987), consisting of the following steps: the CT image is initially reconstructed from the original distorted data and the metallic implants are segmented in the image domain. The metal mask is re-projected to determine the metal shadow in the projection data. Linear interpolation is used to fill in the missing data. After optional addition of noise to the interpolated data, the updated image is reconstructed from the modified projection set. A more advanced approach, the normalized metal artifact reduction (NMAR) algorithm, has been used in several studies. Its main principle, as introduced by Meyer et al (2010), is presented schematically in figure 1. Similar to the previous algorithm, the uncorrected image is first reconstructed from the original sinogram. By using appropriate threshold levels the image of the metal is obtained. Along with the metal image, a prior image is also computed by segmentation of soft tissue and bone based on the original uncorrected image. The sinograms of both the metal and prior images are then obtained by forward projection, and the original sinogram is normalized by dividing it by the sinogram of the prior. The missing data in the shadow of the metal object are estimated by
interpolation in the normalized sinogram. Finally, the normalized sinogram is multiplied by
the sinogram of the prior image again to get denormalized and the corrected image is recon-
structed. To maintain the size and the edges of the metal implant, both methods described
above reinsert the metal image from the originally reconstructed image in the final corrected
image. In addition, to improve the edge information of surrounding bone structures the appli-
cation of a frequency split metal artifact reduction (FSMAR) algorithms was introduced by
Meyer et al. (2012).

As an alternative approach to the sinogram completion methods, model-based iterative
algorithms function under the assumption that most artifacts arise because some data are miss-
ing or deviate from the model used for the data acquisition. Utilizing prior knowledge of the
imaging physics, the measurement statistics and the image statistics, iterative reconstruction
is applied to better interpret the measured projection data. Compared to FBP-based recon-
struction algorithms, the more complex models used by the iterative methods can improve the
accuracy of the reconstructed images, at the cost, however, of increased computation time.
Furthermore, the performance of this method is subject to the level of the accuracy of the
physics models utilized and the prior knowledge of the shape and location of the object. A
simplified schematic of the basic principle of an iterative approach is presented in figure 2.
Starting from an initial—often blank—image, the corresponding sinogram is calculated and
compared to the measured sinogram. The sinogram error is calculated, transformed back to
the image domain and subtracted from the current reconstruction. This process is repeated a
number of times until an objective function—a function of the sinogram error—is optimized,
indicating that a good reconstruction is obtained.

Figure 3 presents an example of a MAR algorithm, combining the principles of both the
sinogram completion and iterative algorithm, based on a commercially available MAR algo-

Figure 1. Scheme of NMAR (Reproduced with permission from Meyer et al 2010).
it was evaluated by several studies referenced in this review. However, we should point out that all major CT vendors such as General Electric, Siemens and Toshiba also offer various MAR solutions. O-MAR is mainly focused on reducing the artifacts from orthopedic metal implants and its implementation is an iterative loop where the output correction image is subtracted from the original input image, however, including steps based on a sinogram completion method. Similarly to the NMAR algorithm, the first step is to produce a metal-only image by applying a threshold. In addition to the metal only image, a tissue classified image
is also created by segmenting the input image into tissue and non-tissue pixels. The sinogram data of all three images, the input, the tissue classified and the metal-only, are then produced by forward projection. An error sinogram is then calculated by subtracting the tissue classified sinogram from the original image. The metal sinogram data are utilized as mask to remove the non-metal data points from the error sinogram. Finally this error sinogram data are backprojected to derive the correction image which will be subtracted from the input image and the process can be repeated.

Although most MAR methods result in improved image quality in clinical scans, there are cases where their efficiency is limited. In sinogram completion methods, the metal projection data are discarded, leading to loss of spatial resolution (Boas and Fleischmann 2011). While the main metal artifacts are mostly reduced, new artifacts may also be introduced by both sinogram completion, such as linear interpolation technique, and iterative approaches, such as selective algebraic reconstruction technique (Boas and Fleischmann 2011, Yazdi et al 2011). Sinogram completion methods are more likely to result in additional artifacts or image that are worse than the original. The main distinction between the two types of algorithms is whether the projections through the metal implant(s) are disregarded or attempts are made to use the data by correctly modeling the physics. The level of efficiency for each technique may differ significantly between simulated data and actual clinical scans and for most methods depends greatly on the type of the implant, the metal’s effective atomic number and size. Accounting for \textit{a priori} knowledge of the composition of the implant material a significant improvement can be achieved (Verburg and Seco 2012). In diagnostic imaging, where efforts focus in reducing x-ray radiation dose, increased noise poses an additional challenge potentially making metal artifacts more evident. An elaborate review on MAR algorithms is available in a recent publication by Gjesteby \textit{et al} (2016).

3. \textbf{External photon beam treatment}

3.1. Pelvic implants (hip prostheses)

The dosimetric effect of artifacts in conventional kilovoltage CT (kVCT) imaging caused by metal hip replacements is a significant concern most commonly during CT-imaging for treatment planning of radiation therapy for prostate cancer. When a unilateral prosthesis exists, non-standard beam angles and planning techniques may be employed to overcome the dosimetric inaccuracies caused by beams irradiating through the hip replacement, where considerable artifacts are usually present. Due to the size of these implants, photon starvation is common, resulting in missing data in the projection domain and stretching the dynamic range of the CT scanner to its lower limit. The CT number for the implants is then underestimated and bright borders are observed surrounding them (Coolens and Childs 2003).

In an extensive review on the dosimetric considerations for patients with a hip prosthesis undergoing pelvic irradiation by the Task Group 63 of the American Association of Physicists in Medicine (AAPM), the physics behind the beam attenuation and perturbation due to the prosthesis and resulting dosimetric effects are outlined (Reft \textit{et al} 2003). However the potential effects of the corresponding CT artifact to the dose calculation accuracy is only briefly mentioned. One of the first studies on the effect of metal implants was published by Ding and Yu (2001) and focused mostly on the dose perturbations due to the presence of metal implants. They compared the inhomogeneity correction algorithms of a commercial analytical 3D treatment planning system (TPS) (CADPLAN) to Monte Carlo (MC) in a homogeneous pelvic phantom with hip prostheses simulated as metal bars. The study concluded that the analytical calculation algorithm seriously underestimated the attenuation of hip prostheses due to its
limitation in assigning the electron density of the prostheses. As a result, if no dose correction factors are applied to the analytical algorithm, the dose to the target was overestimated by 14% and 5% for a typical four-field box and an eight-field technique respectively. Smaller discrepancies are expected however with modern TPS that allow assigning the electron density for metals, such as titanium or gold.

In the case of a unilateral hip prosthesis, deviating from the standard beam arrangement and selecting beam directions to avoid irradiating through the prosthesis and/or areas of artifacts may be a solution. However, this is not always an option in the case of bilateral prostheses because the constraints on beam geometry could compromise target volume coverage and increase the dose to organs at risk (OARs). For these cases, in order to minimize any dosimetric effects due to metal artifacts, an effective MAR method is desired. A case study by Su et al. (2005) revealed that if no MAR algorithm is applied, intensity modulated radiation therapy (IMRT), treatments are more robust than non-modulated three-dimensional conformal radiation therapy (3D-CRT). They used the Corvus system (NOMOS Corp., Cranberry Township, PA) for IMRT planning and delivery with 5-field initial plan and 7-field coplanar boost plan, with non-uniformly spaced gantry positions for all beams and inhomogeneity corrections for both plans. It was demonstrated that better sparing of the bladder and the rectum was achieved with the IMRT plan when non-standard beam directions were necessary due to the presence of the metal implant and artifacts. However the IMRT plan was more inhomogeneous, as expected, with a larger part of the target volume receiving more than 105% of the prescription dose.

Before any advanced MAR algorithm was applied in clinical routine, the use of extended CT-scale was suggested as a method to reduce the effect of hip replacement artifacts in structure delineation accuracy (Coolens and Childs 2003, Glide-Hurst et al. 2013). The basic principle of the method, developed by Klotz et al. (1990), dictates that by scaling down the CT numbers by a factor of 10, the range of 12 bit CT numbers can be increased from 4096 HU to 40960 HU, with a CT-scale expanding from −10240 HU to 30710 HU. In a standard CT-scale, metals with CT numbers of about 20000–30000 HU can produce artifacts of several thousand HU and both hip implant and artifact can exceed its limits. By scaling the CT numbers, although some loss in contrast resolution was observed, differentiating between the metal and its surroundings was more plausible, preserving the detailed geometric information and allowing for the metallic objects to be outlined correctly. To investigate whether the extended CT range can be used to determine the composition of artificial hips in radiotherapy patients, Coolens and Childs (2003) evaluated two CT-calibrations methods, one based on material substitution (Henson 1983) and the other a stoichiometric calibration. Although none of these calibration methods was proven accurate enough to predict electron densities of the hip replacement materials, the extended CT range was shown successful in distinguishing between implant densities and determining the physical dimensions to ±2 mm.

In a more recent study Glide-Hurst et al. (2013) investigated the use of extended CT-to-electron-density (CT-ED) curves coupled with an iterative type MAR algorithm. They compared 12- to 16- bit image reconstructions, with and without the use of the commercially available algorithm O-MAR (Philips HealthCare System 2012), of a pelvic phantom with embedded cerrobend rods and of solid water with embedded stainless steel rods. MC dose calculations showed discrepancies of 6.4% downstream of the cerrobend rod between the two images (12- and 16- bit images), and absolute film dosimetry. Comparisons between calculated dose distributions were performed using gamma analysis, which is a criterion combining both concepts of the dose difference between a calculated dose distribution and a reference one, and the distance to agreement (DTA) criterion, which is the distance between a dose point in the reference distribution and the nearest point in the second distribution having the same dose. The gamma analysis downstream of a stainless steel showed that the 16-bit
calculated dose, with and without MAR, agreed better with film results compared to 12-bit reconstructions. The impact of the 16-bit reconstruction improvement was significant in the cases where beams were directly traversing a metal implant, such as in prostate cancer cases with bilateral hip prostheses, but was diminished when the beams were not directly traversing metal (figure 4). Evaluating the effect of the OMAR algorithm, Glide-Hurst et al found negligible differences between CT numbers with and without the correction, demonstrating that applying the OMAR correction did not alter the CT numbers of high-Z material. Gamma analysis between the dose distributions based on the MAR-corrected and uncorrected images showed high passing rates for both 12- and 16-bit reconstructions (93.5% versus 95.3%).

A different approach was chosen by Keall et al (2003) who created a reference image set with a metal implant by virtually adding iron hip prostheses in a non-prosthesis original CT of a prostate patient and calculated the corresponding sinogram based on this reference image (figure 5). A comparison was then performed between two reconstruction methods: filtered back-projection (FBP) and the iterative deblurring method (IDB) (Wang et al 1996, Robertson et al 1997) in terms of image quality and calculated dose. Due to the over-simplified model used for the sinogram calculation, assuming a noise-free detector and a mono-energetic photon beam with no scattered radiation, FBP image reconstruction resulted in minimal streaking artifacts. No significant dose differences were noticed between dose distributions calculated on the differently reconstructed images compared to the image set without the hip prosthesis. A later study by Bazalova et al (2007), on which a MC method was used to simulate a CT scanner and specifically evaluate the effect of scatter and beam hardening on the metal streaking artifacts, showed significant discrepancies between the MC calculated doses (EGSnrc/Phys. Med. Biol. 62 (2017) R49
TOPICAL REVIEW

DOSXYZnrc) for the original and the artifact-corrected CT images. A projection completion MAR algorithm, based on previous work by Roeske et al. (2003) and Yazdia et al. (2005), was used in this study, along with extended CT-scale and assignment of a density of 8.055 g cm\(^{-3}\), corresponding to steel, to all voxels with HU value of 3095. Using a known phantom geometry as reference image, dose calculations errors exceeding 25% were noted for a 6 MV photon beam in the original, uncorrected, CT images without using the extended calibration. These discrepancies decreased to less than 2% in artifact-corrected images when the extended calibration was used. Significant discrepancies in dose calculations were also observed for a five 18 MV photon beam plan between the original and artifact-corrected geometry.

Wei et al. (2006) investigated the dosimetric effect of CT metal artifacts for a variety of treatment modalities, utilizing the smoothing-plus-scaling MAR algorithm for hip replacement, previously developed by the same group (Wei et al. 2004). The principle of this artifact suppression technique, as originally proposed by Chen et al. (2002), relies on scaling down the projection profile peaks caused by the metal objects. The improved version by Wei et al. (2004) introduced a smoothing technique of the bone pixels to avoid bone artifacts introduced by the original one. Dose distributions for 6 MV and 18 MV four-field plans for a prostate patient with bilateral hip-replacement were compared based on artifact-contaminated and artifact-suppressed image sets. Figure 6 (Wei et al. 2006) presents the qualitative comparison between the dose distributions. Based on the significantly more irregular shape of the 170 cGy and 180 cGy isodose lines, the target coverage is compromised when using the artifact-contaminated images for both energies with the 18 MV plan being less affected compared to the 6 MV plan.

Based on dose distributions calculated by the analytical calculation algorithm of the TPS

Figure 5. Schematic diagram of the methodology by Keall et al. (2003, Reproduced with permission from Elsevier).
Topical Review

(Pinnacle³ system), they concluded that in the absence of any CT MAR method, turning off the inhomogeneity correction when performing dose calculations may reduce the dosimetric errors.

The use of megavoltage CT (MVCT) instead of the conventional kVCT imaging, alone or in combination with appropriate MAR algorithms, has been suggested by several studies as an artifact-suppression method. In a case report by Alongi et al. (2011) for a prostate cancer patient with bilateral femoral prostheses MVCT-image-based treatment planning was used for a helical tomotherapy unit (HiArt). Sufficient morphological information of the pelvic anatomy for radical prostate treatment planning was available and the patient completed the planned treatment without severe side effects recorded at 90 d after the end of treatment. While only a case study, this report demonstrated the feasibility of the method. Tomotherapy treatment planning comparison based on kVCT, hybrid kVCT/MVCT and MVCT image sets for prostate cancer patients with bilateral hip prostheses showed that D35 for the bladder was 8% higher in plans based on MVCT and 7% higher in plans based on hybrid images, compared to the kVCT-based plans. Further, D35% for the rectum was 3% higher for both MVCT and hybrid-based plans (Chapman et al. 2014). For a water phantom with a hip prosthesis, kVCT-based plan with O-MAR correction showed better dose agreement with measurement than without the correction, however the best agreement was observed for the hybrid plans (within 1%).

When combining a NMAR algorithm with MVCT images to correct the original kVCT images of phantoms with bilateral steel inserts, Paudel et al. (2013, 2014) showed up to 15.5% discrepancies between the dose distribution on the corrected image compared to the original one (figure 7). The use of MVCT-NMAR algorithm was found superior in terms of image quality to the commercial orthopedic MAR algorithm. Using MVCT-NMAR corrected images in

Figure 6. Isodose curves for four-beam 6 MV ((a) and (b)) and 18 MV ((c) and (d)) treatment plans based on an artifact-contaminated image ((a) and (c)) and an artifact-suppressed image ((b) and (d)) (Reproduced from Wei et al 2006. Institute of Physics and Engineering in Medicine. All rights reserved).
Figure 7. Comparisons of dose distributions for 6 MV $5 \times 10$ cm$^2$ parallel-opposed fields calculated on (a) the reference image, (b) the image corrected using an IMPACT corrected MVCT prior, (c) the image corrected using a FBP reconstructed MCVT prior, and (d) the image corrected using a kVCT prior. The prescription for both the fields is 280 MU. The image window width/level is 1500/0 HU (Reproduced with permission from Paudel et al 2013).

Table 1. For the relative hot or cold dose on OARs, unspecified normal tissue, and TVs when recomputed on the reference data set (SAR-MVC) (Reproduced from Kim et al 2006, with permission from Elsevier).

<table>
<thead>
<tr>
<th>Metal artifact correction method</th>
<th>OARs (Gy)</th>
<th>TVs (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Parotids</td>
<td>Spinal cord</td>
</tr>
<tr>
<td></td>
<td>Right</td>
<td>Left</td>
</tr>
<tr>
<td>UC</td>
<td>Ave.</td>
<td>0.94</td>
</tr>
<tr>
<td></td>
<td>Std.</td>
<td>0.47</td>
</tr>
<tr>
<td></td>
<td>% Vol.</td>
<td>28.72</td>
</tr>
<tr>
<td>HUC</td>
<td>Ave.</td>
<td>0.84</td>
</tr>
<tr>
<td></td>
<td>Std.</td>
<td>0.59</td>
</tr>
<tr>
<td></td>
<td>% Vol.</td>
<td>50.30</td>
</tr>
<tr>
<td>SCC</td>
<td>Ave.</td>
<td>1.17</td>
</tr>
<tr>
<td></td>
<td>Std.</td>
<td>0.36</td>
</tr>
<tr>
<td></td>
<td>% Vol.</td>
<td>33.96</td>
</tr>
<tr>
<td>MVC</td>
<td>Ave.</td>
<td>0.32</td>
</tr>
<tr>
<td></td>
<td>Std.</td>
<td>0.25</td>
</tr>
<tr>
<td></td>
<td>% Vol.</td>
<td>45.02</td>
</tr>
</tbody>
</table>
TomoTherapy treatment plans without directional blocks for a prostate cancer patient resulted in potential for significant dose reduction outside the planning target volume (PTV) (59.2 Gy to 52.5 Gy in pubic bone).

3.2. CT artifacts from dental implants

Compared to the material and the geometry of metal implants for hip replacement, dental metal implants, although of smaller dimensions, are composed of materials of higher atomic number $Z$. Furthermore, clusters of dental implants are very commonly found in head-and-neck patients’ CT images. The combination of high-$Z$ materials and extended clusters of dental implants can then result in prominent streaking artifacts due to missing projection data caused by the close to complete photon absorption by the implant. Although in the case of CT artifacts caused by a hip replacement implant avoidance of the prosthesis might be possible by selected appropriate beam directions, this is not an option for most cases where artifacts are caused by dental fillings. Especially for oral cavity treatment, unacceptably high upstream dose to OARs, due to backscatter from dental filling material (DFM), and target under-coverage due to attenuation from DFM, render the dose calculation accuracy significantly challenging.

The dosimetric impact of CT artifacts due to dental implants in head-and-neck IMRT was studied by Kim et al (2006). They utilized IMRT plans based on five planning CT image data sets, including the uncorrected and corrected images using projection completion methods proposed by Olive et al (2004): (a) uncorrected (UC), (b) homogeneous uncorrected (HUC), where all pixels identified by the tissue class model (Olive et al 2004) as soft tissue, bone or metal were set to the electron density of water, (c) sinogram completion corrected (SCC), where only segments identified as ‘corrupted’ were cut out and replaced by respective model segments before reconstruction, (d) minimum value corrected (MVC), where all pixels identified as soft tissue, bone, or metal were adjusted to the minimum HU of water, and (e) MVC subsequently corrected with a streak artifacts reduction algorithm (SAR-MVC), where the HU of the pixels in the streaks resulting from the presence of metal were replaced as much as possible by the HU of the soft tissue class or bone class. Image set (e) was used as the reference. IMRT plans created for (a)–(d) image sets were projected onto the reference data set (e) and the dose distribution was recalculated and compared to dose distribution from IMRT plan optimized using image set (e). Their results are shown in table 1, indicating that the dose in small structures, such as the parotids, was less affected by the presence of dental metal artifacts than in larger structures, like the clinical target volumes (CTVs) (CTV1 and CTV2). However, small differences were observed on average for all structures.

In a following study by Kim and Tome (2007), the impact of dental metal artifacts on IMRT dose distributions was investigated in terms of tumor control probability (TCP) and normal tissue complication probability (NTCP). With reference to the IMRT plan optimized based on the SAR_MVC image set (e), the larger increase in NTCP for the spared parotid gland and the spinal cord (3.2% and 2.0% respectively) was estimated from the treatment plan optimized on the original CT data-set (UC), accompanied by a 1.3% decrease in TCP. Corresponding deviations for the plans generated based on the MVC CT data were significantly lower. The disadvantage of this study was that no measurements were performed to verify the dose calculation accuracy or the level of MAR methods’ efficiency with reference to a ground truth image such as a known phantom geometry. Later studies by other groups report higher level of dose discrepancies, although it is generally accepted that the dosimetric effect of metal artifacts diminishes with increasing number of IMRT beams.

The presence of the DFM and its effect on dose calculation in RapidArc treatment planning was assessed in a phantom-based study by Mail et al (2012). They reported percent depth dose...
(PDD) data in the site of DFM measured with Gafchomic EBT2 film and a parallel-plate (PP) ion chamber. In order to investigate the effect of the CT artifacts, in addition to the original artifact-contaminated CT image, two CT image sets were created by replacing the streaking artifact regions with a mask of 10 HU and then adding a 2 cm thick 6000 HU virtual filter to compensate for beam attenuation. A following study by the same group (Mail et al. 2013) reported updated data on the phantom study and applied the artifact suppression method in treatment planning for patients. Both studies reported significant backscatter from the DFM for a single, parallel-opposed fields, and RapidArc treatment technique. For a single-field treatment technique, measurements revealed a 22% increase in backscatter dose upstream from DFM compared to that without the DFM, while the downstream dose was decreased by 14%, compromising target coverage. The calculated PTV dose, based on the original

Figure 8. RapidArc plans of a head and neck patient, with dose calculated on (a) an uncorrected CT image, (b) corrected image with an application of 10 HU mask; (c) dose profiles of the treatment plans in (a) and (b) along the red line in (a) (Reproduced with permission from Mail et al. 2013).
uncorrected CT image set, differed from the delivered by 13%. The application of the 10 HU mask and virtual filter reduced the deviations between the calculated and delivered (measured) doses to 2%, showing potential in improving dose homogeneity in the PTV. The use of a virtual filter around the teeth during the planning phase reduces the target underdosage issue in the phantom. The effect of the method applied for a patient is shown in figure 8. Comparison of dose profiles shows the smoother dose distribution predicted by the dose calculation algorithm when the 10 HU mask is applied (figure 8(c)).

A similar phantom-based study was performed by Maerz et al (2015). They created IMRT and volumetric modulated arc therapy (VMAT) plans using both uncorrected and corrected CT data of phantoms consisting of homogeneous water equivalent material surrounding gold or Wirobond® (cobalt-chrome metal-to-ceramic alloy) dental implants. Dose calculations were performed with both pencil beam and collapsed cone algorithms and compared to Gafchromic film measurements. Deviations of ±11% between the measurement and original CT-based calculated data were observed. Corrected CT data, using a simple bulk density correction method, by setting the mass density of uniform water to $\rho = 1 \text{ g cm}^{-3}$, led to higher accuracy of dose calculations. VMAT was found to be preferable over IMRT for patients with metallic implants, since the continuous dose delivery in VMAT leads to a blurring of the inhomogeneity in dose distribution caused by the implants resulting in better dose agreement of the VMAT optimized plans to the measurements.

When using a MAR method based on monoenergetic extrapolation with a dual energy CT (DECT) scanner, Schwahofer et al (2015) observed only minor reductions of artifacts in the images with aluminum and titanium at a scan extrapolated to a mono-energy of 105 keV and practically no artifact reduction for higher material densities. Dose uncertainties in the order of 10–20% were observed for both corrected and uncorrected image sets. In a study by Kwon et al (2015) based on 11 patients and a custom-built phantom with metal bead inserts, the dental implant CT artifact effect on dose calculation was investigated with and without the OMAR (Philips) correction method. When patients were scanned and treated with closed mouth (8/11) little significant difference was recorded in dose distributions. In agreement with the study by Maerz et al (2015), the small dosimetric effect in these cases can be attributed to the use of VMAT as the treatment planning method, in which the continuous dose delivery during wide angle rotational treatment averages out the dose inhomogeneities. However this was not the case for the three patients treated with opened mouth. These patients included cases where targets such as the tongue and the tonsil were close to the air cavity and consequently were irradiated in that cavity region. Since the air is represented by the lower limit of CT HU, the streak artifacts in the air cavities cause increasing HU numbers. These artifacts can make the mouth appear closed, functioning as a dose build-up region during radiation dose calculation and predict a higher dose in the air cavity in the calculated result. As a result, larger discrepancies were observed between OMAR and non-OMAR image based dose calculation, with corresponding gamma passing rates (1%/1 mm) decreased by 3.5%–9% compared to the average passing rate for the other 8 patients. The calculated dose on the OMAR image was found closer to the real delivered dose on a radiochromic film.

3.3. Spinal implants

Although relatively low-Z materials, such as steel, are typically used in manufacturing spinal implants, the CT-artifacts produced by their presence may be detrimental for the image resolution and ability to define structures or accurate electron density of voxels in the slices where the implants are present. These structures are often very complex in geometry. A combination of stainless steel rods and/or a spinal fusion cage is the most common instrumentation used to
stabilize the spinal column after a laminectomy, which is usually part of the treatment course for spinal tumors prior to radiation treatment. The radiation target volume is commonly located within or adjacent to the metal implant, of which the CT-artifacts render the delineation of the target and OARs extremely challenging. Streak artifacts may occupy the regions surrounding the implants and extend along multiple CT slices. Specifically when laminectomy of multiple vertebrae was necessary, large part of information on electron density of tissues of interest is corrupted causing the potential of significant dose calculation errors.

In a patient study for palliative spine treatment, Rong et al (2010) reported that the metal artifacts due to stainless steel spine-stabilizing rods was too severe for treatment planning based on a regular kilovoltage (kV) CT image, despite attempts to correct using density override. A MVCT image of the patient was used instead, acquired by a TomoTherapy Hi-Art unit (TomoTherapy Inc., Madison, WI). The artifacts were not completely eliminated in the MVCT image but were substantially reduced in comparison to kVCT (figure 9). A dose verification measurement of the plan on the tomotherapy ‘cheese’ phantom with high and low density plugs showed a good agreement between ion chamber/film measurements with a 3%/3 mm gamma criterion, suggesting that this can be an option for treatment planning of similar cases, provided that the output stability of the imaging system is verified and the soft tissue resolution allows for tumor and OARs delineation.

In a phantom study by Son et al (2012) the dose calculation accuracy was compared among kVCT extended-scaled kVCT and MVCT image based plans, with reference to corresponding measurements (ion chamber and Gafchromic film). The authors used both a reference phantom without any metal implants and a phantom including titanium implants in realistic positions according to a spinal posterior/posteriolateral fusion. Compared to the reference phantom, the dose calculation accuracy at the center of the phantom, between the titanium implants, was compromised by an average of 2%, with no clear dependence on the image set or radiation treatment system (Siemens ARTISTE, TomoTherapy Hi-Art and Accuray Cyberknife) used for the dose calculation and delivery. A decrease in the dose calculation accuracy was reported while moving closer to the surface of the titanium implants, suggesting that the distance of...
the structures of interest (target or OARs) to the implants should be taken under consideration when assigning dose constraints in treatment planning. These deviations were less significant when MVCT was used for the dose calculation, concluding that the use of MVCT for planning could be advisable for these cases.

The limitations of analytical dose calculation algorithms, when CT artifacts from titanium rods exist, seem to be at least partially waived when density override and an extended CT-to-density look-up table are used to include the titanium density (Wang et al 2013). Using MC calculations (BEAMnrc/DOSXYZnrc (Rogers et al 2001, Walters et al 2005)) as a reference, the accuracy of an analytical (Pinnacle³) dose calculation was evaluated for various 6MV photon small field (2 × 2 to 5 × 5 cm²) irradiations of a 5 mm diameter titanium rod in water. Although at depths more than 1 cm past the rod the analytical algorithm matched MC when titanium density of 4.5 g cm⁻³ was used, the backscattering dose at the water-rod interface was significantly underestimated. In multi-beam IMRT cases of spinal stereotactic radiosurgery treatment, the dosimetric effect of assigning 4.5 instead of 1.82 g cm⁻³ (top density: cortical bone) to titanium implants ranged from minimal effect to 2% dose difference affecting 15% target volume in the study.

In an effort to study the impact of beam hardening and photon starvation effects on dose, Spadea et al (2014) compared MC dose distributions calculated based on original (uncorrected) and artifact-suppressed datasets from actual patient cases. They utilized a MAR method suggested by Verburg and Seco (2012). Although dose discrepancies of 20–25% were observed for the uncorrected images in the regions surrounding high-Z materials, such as gold dental fillings or platinum wire used for an artery embolization, no significant differences between the uncorrected and artifact-suppressed dataset were found for low-Z materials such as spinal titanium implants.

3.4. AVM following embolization with Onyx

AVM of the brain are focal abnormal conglomerations of dilated arteries and veins within brain parenchyma, in which abnormal arteriovenous shunting is present due to loss of normal vascular organization at the subarteriolar level and a lack of a capillary bed. Depending on the size, location in the brain and flow-related characteristics, therapeutic options may include microsurgery, endovascular embolization, stereotactic radiosurgery or a combination of these modalities. If endovascular embolization is followed by stereotactic radiosurgery, depending on the embolic agent used, CT imaging artifacts may occur that can distort the image quality and dose calculation accuracy of the radiosurgery planning process. Onyx (ev3, Inc., Irvine, CA) is a biocompatible liquid embolic agent that is used in the treatment of AVM with advantageous properties. Its composition includes suspended micronized tantalum powder to facilitate the embolization procedure by providing contrast for visualization under fluoroscopy. This high-Z element (Z = 73) is the source of streaking artifacts, due to increased photoelectric absorption, leading to photon starvation and significantly deteriorating the CT image quality after embolization.

Measurements of attenuation and interface effects of two types of Onyx embolization materials (Roberts et al 2012) revealed that effects on dose distributions of a MV therapeutic beam should be dosimetrically negligible for effective thicknesses of less than 8 mm. The measured interface effects are also small, particularly at 6 MV. However large areas of high-density artifacts and low-density artifacts can cause errors in dose calculations and need to be identified and resolved during planning. Figure 10 displays the effect of Onyx artifacts on dose calculation for stereotactic radiosurgery, based on different heterogeneity correction methods (Shtraus et al 2010). Having first measured the difference in attenuation of water
in comparison to the Onyx for beam energy of 6 MV, and approximated to 3%, Shtraus et al suggested a modified heterogeneity corrections method, by assigning individual electron densities to the normal brain, bone and Onyx, based on the actual attenuation measured. Without this manually introduced correction, dose calculation may be unreliable and have dire consequences for patients receiving high doses of radiotherapy.

4. Brachytherapy treatment

There are two main causes of CT artifacts in brachytherapy treatment. In LDR brachytherapy it’s the metallic composition of the permanent implanted radioactive sources that causes beam hardening during CT imaging for the post-implant position and dose verification process.
In intracavitary brachytherapy (ICB) treatment, most commonly for GYN malignancies, the planning CT is performed after the brachytherapy applicators are inserted in the patient. Image artifacts produced by the metal applicators significantly delayed the widespread adoption of CT-based brachytherapy treatment planning. In addition, due to the physics principles in brachytherapy treatment techniques, most commonly the treatment dose is delivered within areas of homogeneous tissue composition. Subsequently, even when CT imaging is used for organ and tumor delineation and source guidance, heterogeneity corrections are not taken into consideration for the vast majority of brachytherapy treatment cases, since the potential dosimetric inaccuracies due to the artifacts would cancel the dosimetric benefit of heterogeneity corrections in tissues.

### 4.1. Intracavitary applicator treatment for GYN

The instrumentation causing the most significant artifact in GYN brachytherapy is the metal components of shielded ovoids, the cylindrical structures that are placed against the entrance of the cervix to facilitate lateral irradiation of the vagina and cervix, while shielding the bladder and other OARs from unnecessary dose. Traditionally, treatment planning for ICB treatment for GYN cases was based on a set of orthogonal x-ray films, which were used to reconstruct the individual source locations and normal tissue dose calculation points, based on the Manchester system (Tod and Meredith 1953), or dose distribution to the target volume, based on the GYN dosimetry system recommended by the ICRU (ICRU 1985). For this treatment site and technique, the dosimetric effect of CT image artifacts is practically equated with the dosimetric differences between orthogonal-film- and CT-image-based treatment planning.

Applying a projection-interpolation MAR algorithm (Roeske et al 2003) or a hybrid approach—combination of projection-based image segmentation and separate reconstruction of metal and non-metal images (Xia et al 2005)—to images containing the ICB applicators, reconstructed images contained virtually no artifacts and allowed for CT-based planning (Roeske et al 2003). Delineating the bladder and rectum boundaries to be used in conjunction with 3D dose calculations showed a factor of two larger maximum dose to the bladder and rectum compared to doses obtained using orthogonal films. The improvement in dose calculation accuracy for these cases may become even more significant, if a global transformation of CT images could be used to estimate the dose from whole pelvic radiation therapy (WPRT) combined with brachytherapy.

As a different approach, Wagner et al (2009) suggested tomotherapy MVCT-based LDR brachytherapy treatment planning for cervical carcinoma to minimize the reconstruction artifacts of kVCT images near high-Z materials, such as the metal brachytherapy applicators. In a phantom study, the ICRU rectal point dose did not differ significantly between three-dimensional (3D) CT-based and two-dimensional (2D) planning. However the ICRU bladder point dose was significantly lower than the DB2cc (minimal dose received by the 2 cm³ of bladder receiving the highest doses), with mean difference being 291 cGy ± 444 cGy. Although MVCT soft tissue image resolution suffered compared to the ideal kVCT image, in terms of discerning the posterior vaginal wall, the use of MVCT imaging for clinical LDR GYN brachytherapy significantly improved the dosimetric accuracy of treatment planning.

To avoid the effects of metal artifacts in CT imaging by the brachytherapy applicators, the use of magnetic resonance imaging (MRI) in image guided brachytherapy (IGBT) is currently recommended by most international guidance groups—including but not limited to the American Brachytherapy Society (ABS), Groupe Européen de Curiethérapie and the European SocieTy for Radiotherapy & Oncology (GEC-ESTRO) and Royal College of Radiologists (RCR)—on IGBT for GYN cancers (Board of the Faculty of Clinical Oncology...
4.2. Lung brachytherapy

One lung brachytherapy technique includes creating an implant during the surgical procedure by weaving strands of $^{125}$I seeds into a vicryl mesh, which is then sutured over the resection staple line. The goal is to deliver 100 Gy at 5–7 mm along the central axis of the resection margin. This process is followed by CT imaging in an effort to identify the seed positions and verify the dose distribution according to TG-43 based calculations (Nath et al 1995, Rivard et al 2004, 2007), assuming seeds in homogeneous water and no interseed attenuation. In low energy brachytherapy, for photon energies of approximately 20–30 keV, the photoelectric effect determines a substantial proportion of interactions, with cross section being proportional to the fifth power of atomic number. Subsequently, MC-based dose calculations for this brachytherapy technique are sensitive to the conversion of CT image pixel values to atomic composition.

To estimate the level of dosimetric effect from CT artifacts on MC dose calculations for lung brachytherapy treatment, Sutherland et al (2014) studied the differences among dose distributions based on five different image sets: (a) uncorrected, artifact-contaminated original CT image, (b) corrected image based on simple threshold replacement technique (STR), where in a $1 \times 1 \times 1$ cm$^3$ cube centered around each seed position any voxel number greater than or equal to 88 HU was replaced with a CT number of $-200$ HU, (c) corrected based on fan beam virtual sinogram (Sutherland et al 2012), (d) corrected by applying a median filter, where each voxel value is compared with a 3D local median value and replaced with an adjacent voxel value if it differs by 0.25 times the standard deviation, and (e) corrected using a combination of fan beam virtual sinogram method and STR (fan + STR). The radionuclide photon spectra were found to be a significant factor affecting dose differences among various MAR techniques, with largest differences observed for $^{103}$Pd seeds and smallest for $^{131}$Cs seeds. When applying lung-constrained tissue assignment schemes to metallic artifact corrected images, improved dose accuracy for permanent implant lung brachytherapy was achieved. Dose metrics, such as D90, increase dramatically with the use of MAR methods, with the fan + STR method showing the largest discrepancies compared to the uncorrected image set: dose calculations based on the uncorrected image set seem to underestimate D90 by up to 69% for the $^{125}$I seeds, 107% for $^{103}$Pd seeds and 46% for $^{131}$Cs seeds. Application of an artifact-suppression method in the presence of permanent brachytherapy seed implant for lung treatment was crucial for the dose calculation accuracy.

4.3. Prostate LDR brachytherapy

Similarly to lung LDR brachytherapy, prostate permanent implant brachytherapy with $^{125}$I seeds requires CT-based seed localization and dose calculation as part of the post-implant dosimetry verification process. However, the permanently implanted seeds can cause significant artifacts on the CT image set to be used for the post-implant dose verification, having potential dosimetric effects if the voxel HU number is to be corresponded to electron density information and used by the dose calculation algorithm. Figure 11 depicts a MAR algorithm suggested by Takahashi et al (2006) to be applied in CT images acquired after permanent implant brachytherapy with $^{125}$I seeds. The first step of their suggested method, utilizing raw projection data as acquired by the CT scanner, was to obtain an image containing only the $^{125}$I seeds by applying a CT number threshold (figure 11(b)). The sinograms of the CT image
with seeds and of the seeds-only image (figures 11(c) and (d)) were then obtained by inverse Radon transform (Lindgren and Rattey 1981). The sinogram without 125I seeds (figure 11(e)) was then provided by subtracting the projection data of the metal image from that of the original one and the corrected image was reconstructed by Radon transform using this sinogram. Significant image improvement was achieved, greatly facilitating the identification accuracy of number and orientation of the seeds, showing potential for decreasing the time required and improving the accuracy of post-implant dosimetry verification process.

5. Proton and heavy ion therapy

In proton and heavy radiation therapy CT numbers are used to determine the stopping power of tissues in the beam path, which are required for calculation of dose distributions and range of proton or heavy ion beams. Artifacts can therefore result in errors in the estimated range, affecting the target coverage and sparing of the OARs. In clinical treatment plans, manual modification of proton stopping power may be applied in an effort to reduce these errors (Staab et al 2011), however the accuracy of such corrections is uncertain, since the artifact-contaminated images may not allow to clearly determine the anatomy and tissue densities. The level of ion range uncertainties based on an artifact-contaminated CT image set also depends on the material composition and geometry of the metal implant creating the artifacts. Phantom studies indicated that the underestimation of ion range due to artifacts alone may amount to 3% for dental fillings and up to 5% and 18% for hip prostheses made of titanium and steel respectively (Jäkel and Reiss 2007). Since the size of the metal inserts cannot be determined correctly in the artifact-contaminated image, a potential range correction in metal also leads to large uncertainties.

In the case of small-diameter metallic objects, such as screws, employing MVCT-based calibration to derive relative proton stopping power in the implant did not affect significantly the values of calculated range (De Marzi et al 2013). However calculations based on MVCT images were more accurate in predicting the dose distribution in the distal part of the dose curve, probably due to the absence of streak artifacts. In the site of spine-stabilizing screws in

Figure 11. Scheme of MAR method suggested by Takahashi et al (2006, © Japan Radiological Society 2006, reproduced with permission of Springer). The white stripes highlighted by arrows represent the projection data of 125I seeds that caused metal artifacts.
the clinical case of proton treatment of a spine chordoma, MVCT-based estimation of proton stopping power resulted in a minor difference of calculated range between −0.4 and 1.5 mm, compared to the kVCT image, depending on the beam direction. However, Ainsley and Yeager (2014) emphasized that when different calibration methods of CT scanners are explored for proton therapy, one should be cautious of using metals in the calibration procedure since they can strongly influence the result of the fitting procedure, especially in the high HU region and they proposed that metallic plugs be avoided when performing the calibration.

5.1. Titanium rod implants in the treatment of spinal bone tumors

Clinical studies of chordoma patients who received proton therapy showed significant association of reduced tumor control (Staab et al. 2011) or increased local recurrence (Delaney et al. 2009) when titanium-based surgical stabilization was involved. Investigating the dosimetric effect of the presence of metal implants to these cases, Verburg and Seco (2013) performed a systematic study to assess the dosimetric impact of CT titanium artifacts when surgical titanium rods implants for spinal tumors are present during proton therapy, based on both phantom and patient data. Dose recalculations on phantom geometries, with accurate knowledge of the detailed geometry, revealed range errors between 1 mm and 10 mm, depending on the proton beam orientation. When the beam was oriented perpendicular to several bright and dark streak artifacts, the dose calculation was almost unaffected, because errors due to several light and dark streak artifacts cancel out. For proton beams passing through the metal implant, and the bright artifacts surrounding it, or beams parallel to the metal artifact, larger range discrepancies of 5–10 mm were observed. Studying the effect on patient treatment plans, metal artifacts resulted in proton field range errors up to 6 mm distal to regions affected by CT artifacts. MC simulations revealed dose differences of more than 10% in the high-dose area (Figure 12), but since these errors are mostly local in nature, the large number of fields limits the impact on target coverage to a small decrease of dose homogeneity. GafChromic film measurements in an anthropomorphic phantom by Dietlicher et al (2014), revealed that when manual artifact correction method of HU assignment was applied, dose calculations agreed well with measurements for both single field uniform dose (SFUD) and intensity modulated...
proton therapy (IMPT) plans. Without artifact corrections up to 18% of measured points failed the gamma criterion for the SFUD plan.

5.2. Hip prostheses in the treatment of prostate cancer

Studying the effect of CT artifacts from bilateral hip prostheses on dose distributions and treatment planning for a variety of treatment modalities, Wei et al (2006) demonstrated that without any artifact-suppression method, a hypothetical treatment target volume, resembling a prostate treatment located in between the hip prostheses and treated by a single anterior–posterior proton beam, would be severely underdosed (figure 13). For the same target volume, proton beam range and modulation calculations based on the artifact-suppressed images...
differed by 13 mm and 9 mm respectively from those generated based on the artifact-contaminated CT images. Similar results were observed by Newhauser et al. (2008), where streak-induced range errors of 5–12 mm were present in uncorrected kVCT-based patient plans for prostate treatment with single or bilateral hip prostheses. Appropriate calibration of TPSs to simultaneously accommodate kV- and MVCT image sets for proton calculation can improve the dose calculation accuracy when the kVCT image is used for soft-tissue contouring and the prostheses are contoured in the MVCT images leading to a co-registered set of kVCT and MVCT images.

Differences in calculated proton beam penetration for the artifact-contaminated and artifact-suppressed image sets of water phantoms with a hip prosthesis were quantified using water equivalent thickness (WET) based on stopping power data by Andersson et al. (2014). Images without metal objects were used as reference data for the analysis of artifact-affected regions outside the metal objects in both the uncorrected and the corrected (O-MAR algorithm) images. Relative to the reference image set, significant differences in WET of up to 2.0 cm were calculated for the uncorrected case, while for the O-MAR corrected images this difference was reduced to up to 0.4 cm.

5.3. Artifacts present in head-and-neck treatments

Due to the complexity of cranial structures and the variety of possible diseases and treatment techniques, different surgical implants might be the source of CT-artifacts affecting the dosimetric accuracy and treatment planning of proton therapy for cranial or head-and-neck treatments. These may range from the most common dental fillings to artifacts caused by titanium mesh cranial implants or metal frame used for stereotactic radiosurgery planning and patient setup. The presence of more than one type of implants, simultaneously affecting the image quality, is not uncommon.

Studying the effect of titanium mesh cranial implants for a mesh thickness of 0.6 mm on passive-scattering proton, Lin et al. (2014) observed a minimal range reduction of 0.5 ± 0.1 mm for a beam perpendicular to the Ti-mesh or when the incident angle was at 60°. The corresponding dosimetric effect was minimal. In a more extended study (Park et al. 2015) based on four clinical cases with a variety of sources of artifacts (dental fillings, surgical clips or head frame used for patient localization), a MRI-based MAR method was used for improvement of proton range calculation accuracy. The corrupted CT data were replaced by mapping CT HU from a nearby artifact-free slice, using a co-registered MRI. Absolute proton range errors of 2.4 and 1.7 cm, estimated for the uncorrected CT-image sets were reduced to less than 2 mm in both cases.

6. Discussion

The increasing number of patients receiving metal implants and the fact that CT is still the preferred imaging method used in radiation therapy, one of the primary cancer treatment method worldwide, renders the issue of CT metal artifacts of significant clinical importance. Depending on the treatment site and delivery method, the dosimetric effect of metal artifacts becomes more significant in 3D conformal radiation compared to IMRT or VMAT delivery techniques and when smaller number of beams is used in the treatment plan. The dosimetric errors decrease when larger number of treatment beams is utilized and especially in VMAT delivery technique where the continuous dose leads to a blurring of the inhomogeneity in dose distribution caused by the implants. In contrary the effect of artifacts on dose calculation
accuracy can be more significant in particle therapy due to the strong dependence of the particle dose calculation algorithm on the HU number.

6.1. External beam radiation therapy

Metal implants used in hip prostheses can pose a significant issue in external radiation therapy of prostate cancer. CT metal artifacts in the form of streaks become significant due to relatively large diameter of the metal implant causing photon starvation under standard scanning techniques. For a 6 MV beam traversing the hip replacement, ignoring the attenuation due to the high-density material can result to a dosimetric error in the order of 15–25% at the target. Assigning the correct density to the metal in the analytical dose calculation process reduces the effect, however significant dosimetric errors can still be observed, especially in the cases of bilateral hip replacement where a dark area is observed in the CT images between the two implants due to photon starvation. The estimated error varies greatly depending on the beam energy and number of beams used in a plan. IMRT technique was shown superior to 3D-CRT treatment, especially for cases of bilateral hip replacement where achieving sufficient dose sparing of the OARs (bladder and rectum) becomes more challenging and non-standard beam directions are necessary. The dosimetric error due to uncorrected metal artifacts decreases for increasing number of beams and beam energy, but it still remains significant in the order of 5% or higher in the target volume. Utilizing MVCT, when available, instead of or in combination with conventional kVCT imaging offers a potential solution, which becomes more efficient when an adequate MAR algorithm can be applied in the images resulting in good dose agreement of the analytical calculation and measurements in phantoms.

Metal artifacts from dental fillings or implants present another important challenge in radiation therapy treatment planning and dose calculation, in the treatment of head-and-neck tumors, especially when extended clusters of dental implants of high-Z materials (gold) are close to the target volume. Phantom-based measurements reveal target dose discrepancies between the delivered and calculated dose distribution on the artifact-affected CT image in the order of 11–13% on average. Simple correction methods, such as manual assignment of appropriate HU in the soft tissue volumes affected by the artifact within or surrounding the target volume, seem to improve the dosimetric agreement. However, these solutions greatly depend on the planner’s ability and experience, are subject to inconsistencies and should be verified by dose measurements to guarantee the necessary patient treatment plan quality. The use of VMAT delivery method was shown advantageous for head-and-neck or brain tumor treatments where significant CT artifacts due to dental implants are present. Since VMAT is delivered continuously during wide-angle rotations of the treatment head, it has a potential averaging effect to the dose inhomogeneity caused by the artifacts. CT artifacts within the oral cavity, due to dental implants, significantly increase the HU number, making the cavities to appear filled with high density material. This can result in significant dosimetric errors, especially if the target volume is adjacent to the oral cavity. To reduce the dosimetric error in these cases proper immobilization device for the head-and-neck, including a bite-block maintaining the mouth’s closed position, is important.

The majority of spinal implants in patients treated with radiation therapy are located within or close to the desired treatment volume, as they are related to the surgical removal of spinal tumors prior to radiation therapy. Although the delineation of the treatment volume becomes very challenging due to the metal artifacts in these cases, their effect to the dosimetric accuracy of external photon beam radiation therapy has been reported as small compared to higher-Z metal implants such as gold dental fillings. MVCT imaging-based treatment planning, when possible, may facilitate organ delineation and improve the dosimetric accuracy,
even though the artifacts are not completely eliminated and soft tissue resolution suffers significantly. MC dose calculations in phantom geometries revealed the potential of increasing dosimetric errors of the analytical calculation algorithm with decreasing distance to titanium rods used as spinal implants. Assigning the appropriate electron density to the titanium rods in the CT images decreases significantly the dosimetric error while inaccuracies in the density assignment of the titanium rods cause only a small error when multi-beam IMRT treatment planning is employed.

Concerning CT artifacts caused by Onyx implants during stereotactic radiosurgery treatment of arteriovenous malformations, due to the originally homogeneous tissue density of the normal brain, manual assignment of individual electron densities to the normal brain, bone and Onyx volumes can be a sufficient solution improving dose calculation accuracy. However, again, this method is subject to planner’s level of experience and expertise, requiring significant effort and subject to the risk of inconsistency errors.

6.2. Brachytherapy treatment

In high dose rate (HDR) brachytherapy artifacts caused by the brachytherapy applicators pose the most significant challenge in CT-based dose calculation accuracy and treatment planning. Due to the metal artifacts and the assumption that homogeneous tissue is surrounding the brachytherapy sources, the use of CT image based 3D dose calculation in brachytherapy treatment was introduced much later compared to external beam radiation therapy. Application of MAR methods to reduce metal artifacts mainly from brachytherapy applicators has allowed for CT-based treatment planning and 3D dose calculation, greatly facilitating organ delineation and significantly improving the dose calculation accuracy. MVCT imaging for clinical GYN brachytherapy has also shown the potential to improve the dosimetric accuracy of treatment planning, despite the decreased soft tissue image resolution of the MVCT image. A combination of both kV- and MVCT imaging could provide an improved solution for these cases.

In LDR brachytherapy, apart from the applicators, the metal encapsulation of the implanted seed imposes an additional source of artifacts. Due to the low energy spectrum emitted by the permanent implanted LDR seeds, the photoelectric effect becomes a significant energy transfer mechanism, increasing the dependence of the dose calculation accuracy to the accuracy of the electronic density of the material as estimated from the HU value and corresponding conversion curves. Application of MAR methods allows for much more accurate CT-based MC dose calculation in permanent seed implant brachytherapy and greatly facilitates the post-implant radiation seed position and dose verification in LDR brachytherapy.

6.3. Proton and heavy ion therapy

The effect of metal artifacts in particle therapy can be more significant compared to conventional therapies due to the stronger dependence of the dose calculation method on the material stopping power determined from CT numbers via appropriate calibration curves. Furthermore, particle therapy plans generally consist of fewer beams compared to photon treatments, increasing the potential contribution of dosimetric errors introduced to the dose calculation in a single beam due to the CT artifacts.

In patients treated for chordoma tumors with proton radiation therapy the presence of titanium-based surgical stabilization was associated with worse outcome in terms of tumor control. Contrary to the photon radiation therapy, if no MAR method is applied, the potential errors in dose distributions calculated for proton beams are substantial. Reported dosimetric
errors of more than 10% in the target volume (Dietlicher et al 2014) and range calculation errors of up to 10 mm or higher (Verburg and Seco 2013) can become clinically significant due to compromised dose coverage of the target volume. Selecting appropriate beam directions, perpendicular to the metal artifact when possible, may lead to increased dosimetric accuracy, however this approach is dependent on each planner’s experience and expertise and subject to inconsistencies.

In proton therapy of prostate patients with bilateral hip replacement, the dark band between the two high-density implants caused by photon starvation, results in significant underestimation of the CT value and the corresponding proton stopping power. If no correction is applied to the CT image, the dose distribution from an anterior–posterior proton beam can be greatly overestimated leading to a range calculation error of more than 10 mm and significantly compromised target coverage. For lateral proton beams, parallel to the artifact, the range calculation error can be even more significant, in the order of 20 mm (Andersson et al 2014). The application of MVCT-assisted treatment planning has the potential to improve the dosimetric accuracy for these cases as well, provided that appropriate calibration of the TPS is performed to simultaneously accommodate kV- and MC-CT image sets for proton dose calculation.

6.4. Need for a global MAR method

Although numerous attempts have been made to address the issue of CT metal artifacts in radiation therapy, their efficiency in terms of improvement of radiation dose calculation accuracy greatly depends on the type of metal implant, the treatment site, and the radiation treatment technique. So far no solution has been proven to provide complete elimination of metal artifacts for a variety of metal implants and treatment sites, since most methods target specific characteristics of metal artifacts. Moreover, common MAR methods on clinical scans have shown mixed results because new artifacts have been introduced in some cases. Sinogram completion methods can lead to a loss of spatial resolution due to the removal of projections, while beam hardening correction methods can be effective only when sufficient photons pass through the implant and model-based iterative approaches require prior knowledge of the implant material which may not be always available in clinical practice.

The physics behind CT metal artifacts is very different depending on the implant material, so the ideal MAR algorithm would need to account for these differences and incorporate appropriate models depending on the physical principles causing each artifact. Mixed approaches combining physics correction algorithm, designed to reduce beam hardening effects, with projection interpolation and iterative projection replacement methods can result in significant improvement of the image quality compared to standard projection interpolation methods for all types of metal implants, without introducing new artifacts (Verburg and Seco 2012). This demonstrates the significance of the a priori knowledge of the implant material in the MAR process. Combining an iterative MAR with frequency-split metal artifact reduction (FSMAR) can also provide improved dosimetric agreement between calculation and measurements in phantoms (Axente et al 2015). However, although this improvement can be sufficient for photon beam plans, in the case of proton beam treatment further improvements are necessary to achieve necessary dosimetric accuracy.

Since no universal solution with the above characteristics currently exists, treatment planners in radiation treatment centers select case-specific solutions based on their experience and expertise. Manual delineation of artifact-affected areas and electron density assignment is a common practice. The accuracy, reproducibility and consistency of such methods however are highly dependent on the planner’s level of experience. Furthermore the level of distortion
of the artifact contaminated image may not allow for accurate delineation of these areas and organs of interest.

In order for standard treatment planning techniques to be used based on CT images from patients with metal implants, correction of the metal artifacts accurately predicting the HU number is necessary. A universal solution would be desirable with the possibility to distinguish and appropriately treat the variety of physics properties and interactions resulting in metal artifacts in a CT image used in radiation therapy treatment planning. A priori knowledge of the composition and geometry of the metal implant has been shown significantly advantageous and all studies point to the fact that a sophisticated segmentation method in the projection space and a combination rather than a single MAR technique will provide such potential. An ideal MAR solution should correct artifact-affected CT images for any geometrical properties of implants causing the artifacts, providing correction steps depending on the size of the affected regions and number of affected projections, distinguishing between the different size and material composition of the implants. Advanced physics correction algorithms correcting for beam hardening should be applied at the presence of low-Z metal or small size seeds or screws of a few millimeters diameter while projection completion methods are more efficient in cases of large diameter or high-Z materials causing significant photon starvation presented with extreme white streaking artifacts. Improved scanning techniques allowing for selective increased photon fluence through areas of interest including metal implants could also be applied as a solution against photon starvation. Also significant specifically to proton therapy is the ability to correctly depict and estimate the dimensions of the metal implant in order to achieve accurate range calculation based on the corrected image when the metal is in the path of the beam. When CT image acquisition is intended for radiation treatment planning, potential increased dose from CT scanning can be considered of secondary priority, given the radiation treatment dose to be received by the patient, since the potential dosimetric accuracy and dose sparing of the OARs based on improved image quality and organ delineation can be orders of magnitude higher.

7. Conclusions

The level of dosimetric effect of CT metal artifacts in radiation therapy varies greatly depending on the implant site, its relative position to the treatment, the level of tissue inhomogeneity and treatment modality. In the presence of high density or large size metal implants, such as hip replacement in the treatment of prostate cancer, dosimetric errors of up to 25% have been reported for 6 MV photon beams. The effect however is reduced for increasing number of beams and higher photon energies. When dental implants exist in the treatment of brain or head-and-neck tumors, the reported effect is lower, in the order of 11–13%. The effect can be more significant in particle therapy, since there is a stronger dependence of the dose calculation method on the material stopping power determined from the CT number. In patients treated for chordoma tumors with proton radiation therapy, dosimetric errors of more than 10% in the high-dose area were reported, leading to range calculation errors of up to 10 mm, due to the presence of CT metal artifacts.

Current MAR algorithms can lead to reduction of the dosimetric error, however further improvement is desirable. Developing a universal MAR solution, with a combination of improved acquisition protocols, physics correction and reconstruction algorithms, requires a collaboration between multiple groups from different disciplines. An additional existing challenge when attempting to evaluate the efficiency of a MAR method is the absence of a standardized approach and tools to be used as benchmark. Phantom geometries utilized
in most studies cannot provide the appropriate level of detail and differ significantly from actual clinical patient cases. When patient cases are used, the lack of a reference ground-truth image-set quality and dose calculation metric only allows for relative studies. Developing a MAR evaluation protocol based on standardized patient-like phantom geometries and actual patient cases with available reference images would be a significantly effective step towards improvement in image quality and dose calculation in CT images affected by the presence of metal implants.

References

Bazalova M, Beaulieu L, Palefsky S and Verhaegen F 2007 Correction of CT artifacts and its influence on Monte Carlo dose calculations Med. Phys. 34 2119
Board of the Faculty of Clinical Oncology R 2009 Implementing image-guided brachytherapy for cervix cancer in the UK R. Coll. Radiologists 1–23
Boas F E and Fleischmann D 2011 Evaluation of two iterative techniques for reducing metal artifacts in computed tomography Radiology 259 894–902

Glover G H 1981 An algorithm for the reduction of metal clip artifacts in CT reconstructions Med. Phys. 8 799

Henson P W 1983 Attenuation coefficient and atomic number calculation involving elements between hydrogen and zinc in the CT scanner energy range of 50 to 100 keV Australas. Phys. Eng. Sci. Med. 6 20–5

ICRU 1985 Dose and Volume Specification for Reporting Intracavitary Therapy in Gynecology (ICRU Report 38) (Bethesda, MD: International Commission on Radiological Units and Measurements)


Kalender W A, Hebel R and Ebersberger J 1987 Reduction of CT artifacts caused by metallic implants Radiology 164 576—7


Lemmens C, Paul D and Nuyts J 2009 Suppression of metal artifacts in CT using a reconstruction procedure that combines MAP and projection completion IEEE Trans. Imaging 28 250–60


Roeske J C, Lund C, Pelizzari C A, Pan X and Mundt A J 2003 Reduction of computed tomography metal artifacts due to the Fletcher-Suit applicator in gynecology patients receiving intracavitary brachytherapy Brachytherapy 2 207–14


Shtraus N, Schifter D, Corn B W, Maimon S, Alani S, Frolov V, Matecnevsky D and Kanner A A 2010 Radiosurgical treatment planning of AVM following embolization with Onyx: possible dosage error in treatment planning can be averted J. Neurooncol. 98 271–6


Wolford M L, Palso K and Bercovitz A 2015 Hospitalization for total hip replacement among inpatients aged 45 and over: United States, 2000–2010 NCHS Data Brief No. 186 (Hyattsville, MD: National Center for Health Statistics)


