European Journal of Radiology 88 (2017) 61–70

Contents lists available at ScienceDirect

European Journal of Radiology

journal homepage: www.elsevier.com/locate/ejrad

Quantifying metal artefact reduction using virtual monochromatic dual-layer detector spectral CT imaging in unilateral and bilateral total hip prostheses

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ARTICLE INFO

Article history:
Received 25 March 2016
Received in revised form
24 November 2016
Accepted 3 January 2017

Keywords:
Virtual monochromatic
Dual-layer detector
Spectral CT
Metal artefacts
Hip prostheses

ABSTRACT

Purpose: To quantify the impact of prosthesis material and design on the reduction of metal artefacts in total hip arthroplasties using virtual monochromatic dual-layer detector Spectral CT imaging.
Methods: The water-filled total hip arthroplasty phantom was scanned on a novel 128-slice Philips IQon dual-layer detector Spectral CT scanner at 120-kVp and 140-kVp at a standard computed tomography dose index of 20.0 mGy. Several unilateral and bilateral hip prostheses consisting of different metal alloys were inserted and combined which were surrounded by 18 hydroxyapatite calcium carbonate pellets representing bone. Images were reconstructed with iterative reconstruction and analysed at monochromatic energies ranging from 40 to 200 keV. CT numbers in Hounsfield Units (HU), noise measured as the standard deviation in HU, signal-to-noise-ratios (SNRs) and contrast-to-noise-ratios (CNRs) were analysed within fixed regions-of-interests placed in and around the pellets.
Results: In 70 and 74 keV virtual monochromatic images the CT numbers of the pellets were similar to 120-kVp and 140-kVp polychromatic results, therefore serving as reference. A separation into three categories of metal artefacts was made (no, mild/moderate and severe) where pellets were categorized based on HU deviations. At high keV values overall image contrast was reduced. For mild/moderate artefacts, the highest average CNRs were attained with virtual monochromatic 130 keV images, acquired at 140-kVp. Severe metal artefacts were not reduced. In 130 keV images, only mild/moderate metal artefacts were significantly reduced compared to 70 and 74 keV images. Deviations in CT numbers, noise, SNRs and CNRs due to metal artefacts were decreased with respectively 64%, 57%, 62% and 63% ($p < 0.001$) compared to unaffected pellets. Optimal keVs, based on CNRs, for different unilateral and bilateral metal hip prostheses consisting of different metal alloys varied from 74 to 150 keV. The Titanium alloy resulted in less severe artefacts and were reduced more effectively compared to the Cobalt alloy.
Conclusions: Virtual monochromatic dual-layer Spectral CT imaging results in a significant reduction of streak artefacts produced by beam-hardening in mild and moderate artefacts by improving CT number accuracy, SNRs and CNRs, while decreasing noise values in a total hip arthroplasty phantom. An optimal monochromatic energy of 130 keV was found ranging from 74 keV to 150 keV for different unilateral and bilateral hip prostheses consisting of different metal alloys.

Abbreviations: CT, computed tomography; CTDI, computed tomography dose index; CNR(s), contrast-to-noise-ratio(s); PMMA, polymethyl methacrylate; ROI(s), region(s) of interest; SNR(s), signal-to-noise-ratio(s); UHMWPE, ultra-high molecular weight polyethylene.

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http://dx.doi.org/10.1016/j.ejrad.2017.01.002
0720-048X/

1. Introduction

Metallic implants are known to generate artefacts in computed tomography (CT) images due to beam-hardening, scatter effects and photon-starvation [1]. These artefacts impede the diagnostic accuracy of soft tissue and bone pathology in patients after total hip arthroplasty. The beam-hardening effect is caused by the absorption of the polychromatic X-ray beam [2]. Since low-energy X-ray photons are attenuated more easily than the remaining high-energy photons, polychromatic beam transmission does not follow the exponential decay seen with a monochromatic X-ray. This process particularly generates artefacts when high atomic number materials such as metals are present [1]. The severity of these metal artefacts generally increases with the atomic numbers of the implant material.

Metal artefacts can be reduced using virtual monochromatic images computed from Dual-Energy CT or Spectral CT scans by eliminating the beam-hardening effect [3–8]. Monochromatic images reported in kiloelectron volt (keV), depict objects as if the X-ray source produced X-ray photons at a single energy level only, instead of a range of X-ray photons up to a certain maximum energy or peak kilovoltage (kVp) in conventional CT [7]. These monochromatic images can be computed from the CT reconstructions from projection data with low and high average photon energy. Extrapolating to high energies results in monochromatic images with reduced metal artefacts compared to those at low energies due to reduced beam-hardening effects.

Several Dual-Energy CT studies investigated the value of virtual monochromatic imaging in the reduction of metal artefacts with optimal energies varying between 95 and 190 keV [5,6,9–16]. In these studies, the benefits of high keV virtual monochromatic imaging regarding metal artefacts is well recognized for several Dual-Energy CT implementations. However, a thorough quantitative evaluation of the effect of the type of material used for the implant on the performance of virtual monochromatic imaging as a tool for metal artefact reduction has not been investigated yet. Also, the quantification of metal artefact reduction using a novel dual-layer detector approach has not previously been described.

Hence, the aim of this study was to quantify metal artefact reduction in the CT imaging of different unilateral and bilateral hip prostheses types using dual-layer detector Spectral CT imaging at various monochromatic energies. For the evaluation of the residual metal artefacts we use quantitative CT image quality parameters i.e. CT number accuracy, noise values, signal-to-noise-ratios (SNRs) and contrast-to-noise-ratios (CNRs).

2. Materials and methods

2.1. Spectral CT imaging using dual-layer detector technique

The recently introduced dual-layer detector technique detects two different photon x-ray spectra simultaneously enabling a spectral separation in the projection space without the need of spatial and temporal interpolations [17]. This system consists of a single tube and uses detectors with simultaneous high and low energy discrimination where the low energy photons are captured in the first layer and the high energy photons are captured in the second layer of the detector. With this approach virtual monochromatic images can be extracted and extrapolated ranging from 40 up to 200 keV.

2.2. Hip phantom and prostheses

The custom made hip phantom consists of polymethyl methacrylate (PMMA) or Perspex with dimensions of 320 mm width, 130 mm height, and 290 mm depth (Fig. 1). Additional PMMA shields were placed below and on top of the phantom to increase the sagittal diameter to 190 mm, based on the water-equivalent diameter of 291.5 mm and coronal diameter of 320 mm derived from a body mass index of 25 using a formula of Menke et al. (Fig. 1) [18]. The phantom was filled with water and different prosthetic configurations were inserted. Since we wanted to determine the effect of different prosthetic composites and the use of unilateral and bilateral prostheses we have composed six prosthetic compositions referred to as 'Boxes', shown in Table 1. Scans were obtained with and without the insertion of three different hip prostheses with different stem, head and cup composites. The prostheses were fixated with custom-made PMMA moulds in order to prevent movement and provide correct alignment at the middle of the phantom. The phantom contained 18 cylindrical hydroxyapatite calcium carbonate pellets representing bone with a certified density calibration and documented tolerance of $\pm$0.5% [19] placed at relevant radiological zones in the femur (the Gruen zones, pellets 1–7) and acetabulum (DeLee and Charnley zones, pellets 0 and 8) [20,21]. 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 (Fig. 1).

2.3. Image acquisition and reconstruction

Scans were reconstructed with iterative reconstruction and were analysed with a standardized measurement template mask. The phantom was scanned on a Philips IQon 128-slice dual-layer detector Spectral CT scanner at standard dose with a Computed Tomography Dose Index (CTDI) of 20.0 mGy at 120-kVp and 140-kVp. Static scan parameters were 64×0.625 mm collimation, 0.9 mm slice thickness with 0.45 mm increment, 330 mm field-of-view, 0.392 pitch, 512×512 image matrix, high resolution and a rotation time of 0.75 s. The hard and sharp reconstruction filter D was chosen in order to enhance edges and to optimize the contrast between hard and soft materials in the CT imaging of metallic components. The iterative reconstruction algorithm contains a de-noising step which takes the noise in the spectral decomposition into account to de-noise the photoelectric and Compton images [17]. Without the use of the de-noising algorithm the noise heavily increases at low and high keV, which would result in a suboptimal results at very high keV regarding metal artefact reduction.

2.4. Quantitative analysis of the image quality

The effects of virtual monochromatic Spectral CT imaging, degree of metal artefacts and effectiveness in metal artefact reduction were quantified by analysing CT numbers, noise values, SNRs and CNRs within fixed regions of interest (ROIs) placed in and around the pellets (Fig. 2). The quantitative analysis was executed using the image-processing programs ImageJ (version 1.48 v) and Matlab® (version 2014b). A standardized measurement template mask was developed and used for each scan in order to enhance the reliability of the measurements. The coronal slice aligned at the middle of the pellets was stored at a Philips Intellispace Portal Workstation. A single coronal slice was loaded into ImageJ where a template was manually created with 9 left pellet ROIs (L0–L8) and 9 right pellet ROIs (R0–R8) (Fig. 2a). Matlab was used to perform the actual quantitative measurements. Pellet ROIs had a diameter of 14.7 pixels or 6.6 mm thus mitigating partial volume effects. The number of pixels of background ROIs was matched to the number of pixels of pellet ROIs. Fig. 2b illustrates a single pellet with its inner pellet ROI 1 and surrounding background ROI 2. CT numbers were calculated by measuring mean pixel intensities within local ROIs (Eq. (1)). Noise was measured by calculating the standard deviation of pixels in an ROI of a uniform section of the image (Eq. (2)).