Virtual Monochromatic Spectral Imaging with Fast Kilovoltage Switching: Reduction of Metal Artifacts at CT[1]

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TEACHING POINTS
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With arthroplasty being increasingly used to relieve joint pain, imaging of patients with metal implants can represent a significant part of the clinical work load in the radiologist's daily practice. Computed tomography (CT) plays an important role in the postoperative evaluation of patients who are suspected of having metal prosthesis–related problems such as aseptic loosening, bone resorption or osteolysis, infection, dislocation, metal hardware failure, or periprosthetic bone fracture. Despite advances in detector technology and computer software, artifacts from metal implants can seriously degrade the quality of CT images, sometimes to the point of making them diagnostically unusable. Several factors may help reduce the number and severity of artifacts at multidetector CT, including decreasing the detector collimation and pitch, increasing the kilovolt peak and tube charge, and using appropriate reconstruction algorithms and section thickness. More recently, dual-energy CT has been proposed as a means of reducing beam-hardening artifacts. The use of dual-energy CT scanners allows the synthesis of virtual monochromatic spectral (VMS) images. Monochromatic images depict how the imaged object would look if the x-ray source produced x-ray photons at only a single energy level. For this reason, VMS imaging is expected to provide improved image quality by reducing beam-hardening artifacts.

Introduction

Conventional computed tomography (CT) of patients with large metal implants is difficult at best. Beam hardening, photon starvation (ie, noise from low photon counts), and scatter artifact normally obscure the immediate surrounding bone, making it impossible to evaluate the metal-to-bone interfaces for loosening, granulomatous particle disease, and fractures. In addition, at CT of patients with hip prostheses, the loss of soft-tissue detail due to streak artifact both near and farther away from the implant can cause concern by making evaluation of intrapelvic anatomy and adenopathy difficult (1). Current CT systems correct this

Abbreviations: MARS = metal artifact reduction software, VMS = virtual monochromatic spectral

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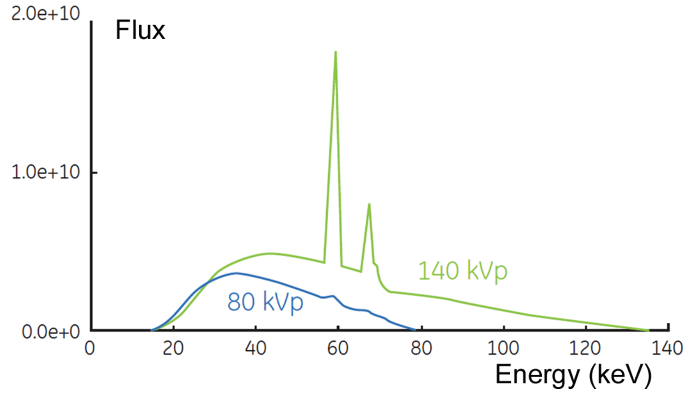


Figure 1. Graph illustrates polychromatic x-ray spectra for 80 and 140 kVp. Conventional CT makes use of a polychromatic x-ray beam composed of photons with a range of energies, with maximum energy expressed as peak voltage (kilovolt peak), which defines the upper-limit x-ray for a polychromatic x-ray beam. *keV* = kiloelectron volts, which is the unit of measure for one x-ray photon and specifies the photon energy for a monochromatic x-ray source. The y-axis represents the number of photons, which is relative; consequently, the y-axis is unitless. (Used with permission of Paul Ayestaran, GE Medical Systems, Buc, France.)

phenomenon, called beam-hardening effective energy shift, with use of calibration data measured in specific phantoms and calculated with the relevant function during the image reconstruction process (2). Other factors may diminish the production of artifacts at multidetector CT. During image acquisition, the use of a high peak voltage (kilovolt peak), high tube charge (in milliampere-seconds), narrow collimation, and thin sections helps reduce metal-related artifacts. During image reconstruction, the use of thick sections, lower kernel values (similar to the standard reconstruction algorithm), and the extended CT scale helps reduce these artifacts (3). However, use of these techniques may not be sufficient. For instance, if the tube charge is increased, the problem of photon starvation is overcome and artifacts are reduced, but the patient receives a higher radiation dose if the beam passes through a region with lower attenuation (2). Despite advances in detector technology and computer software, artifacts from metal implants remain a problem. More recently, the use of virtual monochromatic spectral (VMS) imaging has been proposed as a means of reducing beam-hardening metal artifacts (4,5). VMS images are reconstructed from a pair of accurate material-density images and mass-attenuation coefficients. For a given kiloelectronvoltage, the object is depicted as if it were being imaged with a monochromatic beam at the same voltage.

In this article, we review the theory of VMS imaging and describe our clinical experience with a single-source dual-energy scanner with fast kilovoltage switching (ie, rapid alternation between high- and low-kilovoltage settings) to reduce beam-hardening artifact, using optimized protocols to improve diagnostic performance in patients with metal implants.

Polychromatic X-ray Spectra and Beam-hardening Artifact

At conventional CT, the attenuation of the x-ray beam passing through an object is measured. The x-ray beam quality is commonly defined in terms of kilovolt peak (ie, maximum photon energy [6]), since the x-ray beam consists of a mixture of photon energies (2,5). The polychromatic spectra for 80 and 140 kVp are shown in Figure 1. A polychromatic image is an image generated at conventional single-energy CT due to the full spectrum of photon energies with the kilovolt peak defined by the user (eg, 80, 100, 120, or 140 kVp) (Fig 1). As the beam passes through an object, it becomes “harder”; that is to say, the mean photon energy increases, because the lower-energy photons are absorbed more rapidly than the higher-energy photons (2). The detected x-ray beam contains the higher-energy portion of the spectrum, resulting in dark streaks between metal structures (Fig 2). These streaks occur because the portion of the beam that passes through one of the objects at certain tube positions is hardened less than when it passes through both objects at other tube positions. Because monochromatic dual-energy CT images are generated from projection-space data, they are less affected by beam-hardening artifact and provide more accurate data than do standard single-energy CT images. In addition, because monochromatic dual-energy CT images can be generated for any photon energy level between 40 and 140 keV, a set of images can be created that optimizes contrast differences between two adjacent structures (7). For this reason, VMS imaging has the potential to reduce beam-hardening artifacts.

Teaching Point

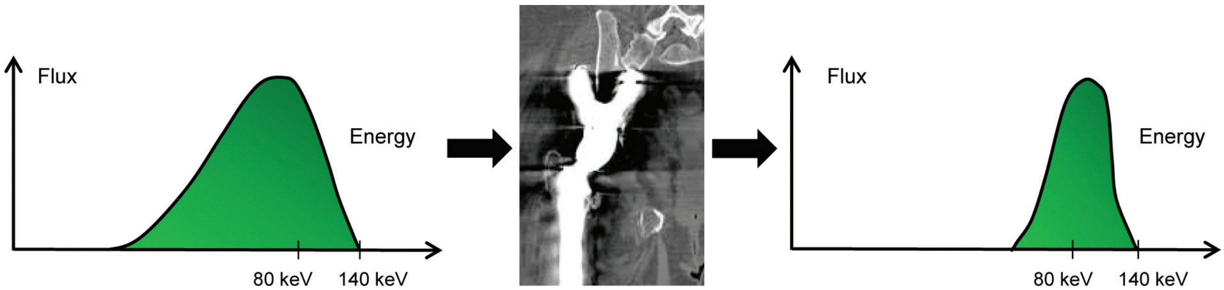


Figure 2. Graphs and image illustrate beam-hardening artifact. As an x-ray beam passes through metal, low-energy x-ray photons are absorbed first, and the remaining high-energy photons are not attenuated as easily. The detected x-ray beam contains the higher-energy portion of the spectrum, resulting in dark streaks between metal structures.

Advantages and Disadvantages of Three Types of Dual-Energy CT Scanners

Type of Scanner	Advantages	Disadvantages
Dual-source	Good spectral separation between high- and low-energy scans, ease of equalizing dose and noise between high and low-energy scans by modulating tube current for each tube, ability to measure attenuation (in Hounsfield units) on virtual unenhanced images	Limited temporal and spatial registration because two separate image datasets are acquired, maximum field of view of 33 cm for dual-energy acquisition, limited flexibility caused by image-domain dual-energy decomposition
Single-source with dual detector layers	Perfect temporal and spatial registration, projection-space dual-energy decomposition can be used	Limited energy separation with substantial spectral overlap
Single-source with fast kilovoltage switching	Good temporal registration between high- and low-energy datasets, which are obtained nearly simultaneously; availability of the full 50-cm field of view for use in image analysis	Limited spectral separation between high- and low-energy scans, higher noise levels on images obtained at a lower peak voltage

Source.—Reference 7.

Principles of Dual-Energy CT and VMS Imaging

Teaching Point

The basic principle of dual-energy CT is the acquisition of two datasets from the same anatomic location with use of different kilovolt peaks (usually 80 and 140 kVp). The two datasets must be acquired together to "freeze" patient and gantry motion. Dual-energy CT allows the analysis of energy-dependent changes in the attenuation of different materials. Three types of dual-energy CT scanners are available: a dual-source dual-energy scanner, a single-source dual-energy scanner with dual detector layers, and a single-source dual-energy scanner with fast kilovoltage switching (7). These scanners differ in the way high- and low-energy CT datasets are acquired, with each scanner having certain advantages and disadvantages (Table).

The single-source CT scanner with a single detector layer (Discovery CT 750HD; GE Healthcare, Milwaukee, Wis) relies on a single x-ray source with fast switching between two kilovoltage settings (80 and 140 kVp) at intervals of 0.5 msec during a single gantry rotation to generate high- and low-energy x-ray spectra. A detector with a fast response and a data acquisition system with a fast sampling capability are used to capture the alternating high- and low-energy data. If the x-ray attenuation of an object is measured at two different (low- and high-kilovolt-peak) spectra, alternating quickly from one view to the next, it is possible to mathematically transform the attenuation measurements into the density (or amount) of two materials that would be needed

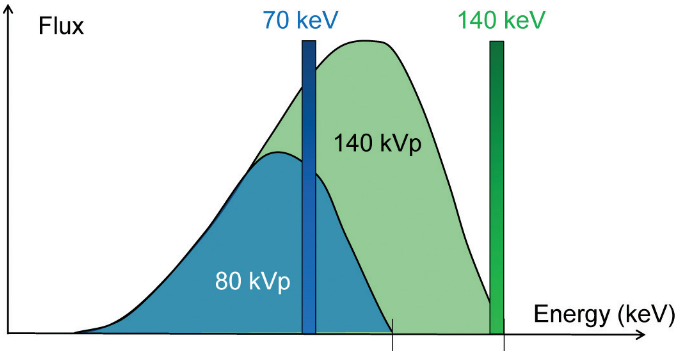


Figure 3. Graph illustrates VMS imaging versus polychromatic spectral imaging. A polychromatic x-ray beam is composed of photons with a range of energies, with maximum energy expressed as the kilovolt peak. A monochromatic x-ray beam is composed of photons with a single, constant energy expressed in kiloelectronvolts, the unit of measure for one x-ray photon, which specifies the photon energy for a monochromatic x-ray source. The VMS image shows how the imaged object would look if the x-ray source produced x-ray photons at only one energy level (eg, 70 or 140 keV).

to produce the measured attenuation. This process is referred to as material decomposition or material separation. It is important to note that material decomposition does not help identify materials. Rather, given two selected basis materials, material decomposition helps determine how much of each material would be needed to produce the observed low- and high-kilovolt-peak measurements (Fig 3). Generally, low- and high-attenuation materials are selected as the basis pair. For medical diagnostic imaging, water and iodine are often used, since they span the range of atomic numbers of materials generally found in medical imaging and thus approximate soft tissue and iodinated contrast material, resulting in material-density images that are intuitive to interpret.

Monochromatic images may be synthesized from the mass-attenuation coefficient values and density images of the two basis materials with a normalization process that makes use of a water-attenuation coefficient for the intended x-ray energy level. The monochromatic image depicts how the imaged object would look if the x-ray source produced x-ray photons at only a single energy level (Fig 3). The unit of measurement for one x-ray photon is kiloelectronvolts and specifies the photon energy for a monochromatic x-ray source (6). Calculation of the monochromatic image is a linear operation performed on the basis material images. Once a spectral acquisition is completed, postprocessing is applied to generate conventional low- and high-kilovolt-peak attenuation, material density, and synthesized monochromatic 40–140-keV images with use of dedicated software (Gemstone Spectral Imaging [GSI] Viewer 2.00 and GE Volume Share 4—AW 4.4, GE Healthcare). This postprocessing usually takes less than 20 seconds. The GSI Viewer allows the radiologist to efficiently obtain the desired clinical information and adjust the energy level (eg, to 70, 80, 90, 110, and 140 keV) to optimally reduce beam-hardening artifacts. Once the optimal energy level is chosen, the reformatted images are sent to the picture archiving and communication system and can be viewed on all commercially available viewers.

Teaching Point

Beam-hardening Correction

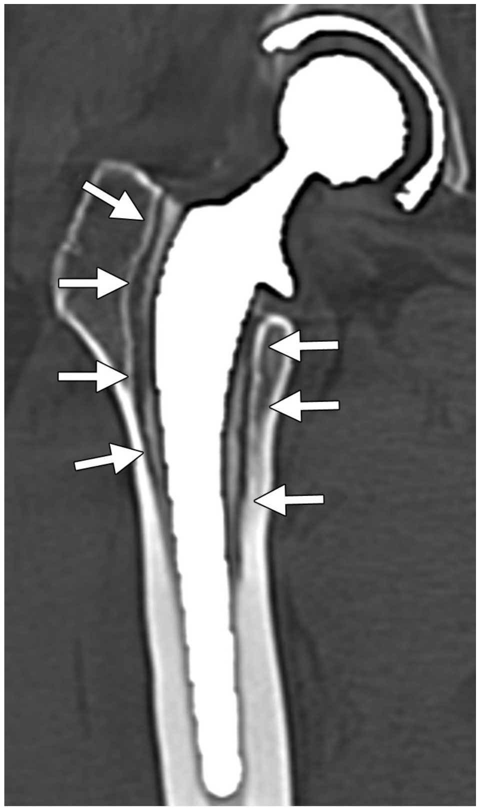
A zone of nonfocal or focal periprosthetic lucency is a common complication of nonseptic hip arthroplasty (8). To improve the visualization of periprosthetic lucency or metal-to-bone or cement-to-bone interfaces, a higher energy (ie, 140 keV) should be used (Figs 4, 5). However, higher energies yield less contrast between materials, and lower energies are needed to visualize soft tissue.

Teaching Point

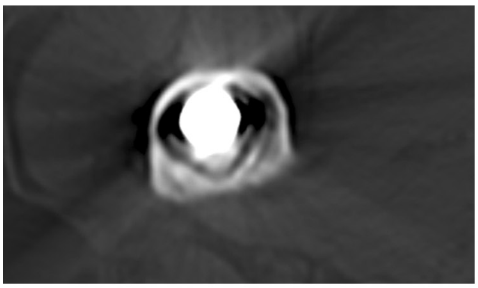
CT is accurate in the detection of painful infection at the site of a hip prosthesis on the basis of soft-tissue findings, rather than bone periprosthetic abnormalities as seen at CT or conventional radiography. Infected hips show evidence of joint distention or fluid collections around the prosthesis (8). To improve the visualization of soft tissue both near and farther away from the implant, it is necessary to improve contrast between materials. Monochromatic images can provide the superior contrast resolution afforded by lower-energy acquisition (ie, 70 or 80 keV), but with less noise. The image noise on VMS images is lowest at approximately 70 keV. Consequently, for a given radiation dose, VMS images that are reconstructed at approximately 70 keV from split 80- and 140-kVp datasets have a lower noise level and a higher contrast-to-noise ratio than do 120-kVp CT images (5). At a lower energy level (eg, 80 keV), VMS imaging is more suitable for soft-tissue detail and easily demonstrates joint distention and fluid collections in infected hips (Fig 6).

Teaching Point

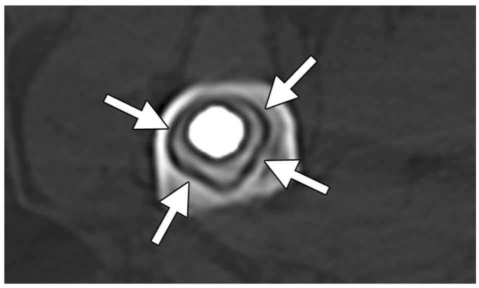
4a.



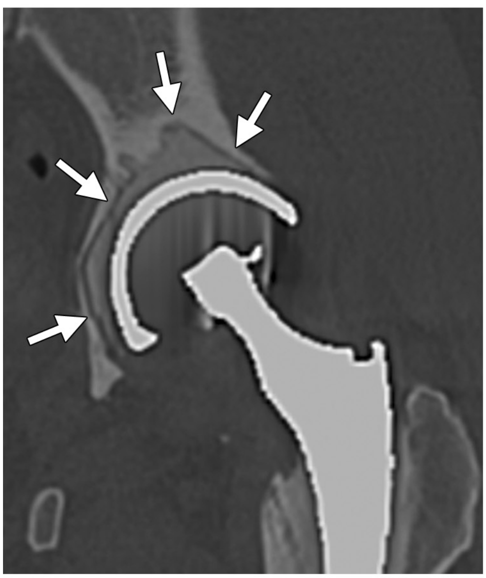
4c.



4b.



4d.



5.

Figures 4, 5. Aseptic prosthetic loosening. **(4)** On coronal reformatted **(a)** and axial **(b)** conventional (polychromatic) CT images, regions of bone around metal implants are obscured by beam-hardening artifacts. **(c, d)** Corresponding VMS CT images (140 keV) clearly depict a nonfocal low-attenuation line (arrows) at the bone-stem interface. Note the undersampling artifact around the metal implant in **c**. **(5)** Coronal reformatted VMS CT image (140 keV) clearly depicts a nonfocal low-attenuation line (arrows) at the acetabular cement-bone interface.

Figure 6. Septic hip prosthesis (*Staphylococcus aureus* infection). **(a, b)** Coronal reformatted **(a)** and axial **(b)** VMS images (140 keV) show large areas of low attenuation around the prosthesis (arrows in **a**), thereby helping visualize the bone-metal interface. **(c, d)** Corresponding VMS images obtained at 80 keV with MARS show joint distention (* in **c**) and allow better visualization of the soft tissues close to the prosthesis.

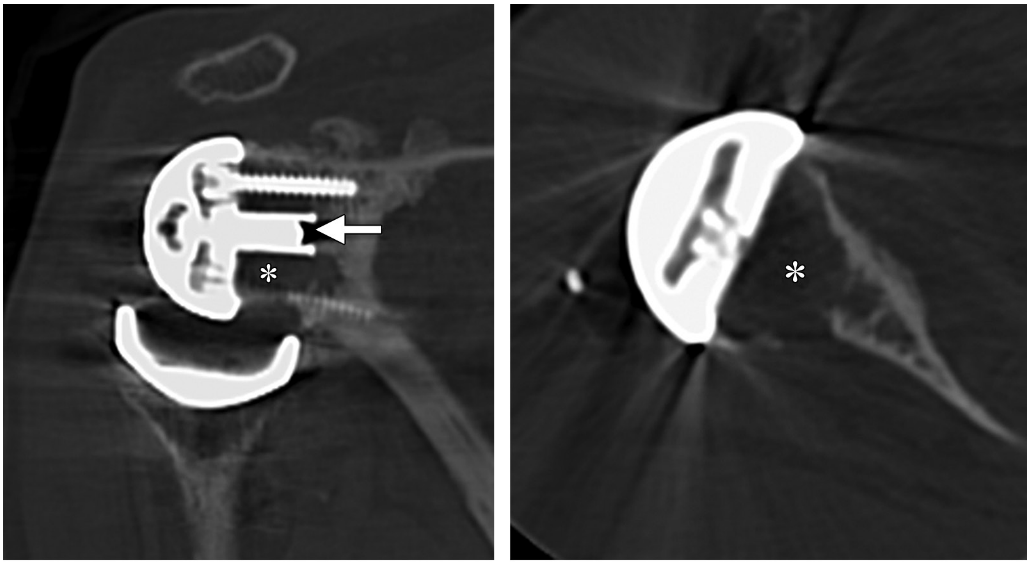
However, low-energy VMS imaging is less efficient in reducing beam-hardening artifact. To improve the visualization of soft tissue near (and not so near) the implant at lower energy, we use a vendor-specific metal artifact reduction software (MARS). With the GSI-MARS method, a metal prosthesis can be segmented on a conventionally reconstructed image based on a CT number threshold. By means of forward projection, the metal artifact–corrected image is then overlaid on the original image. GSI-MARS can also replace the photon-starved regions with information derived from accurate projection measurements by using material decomposition on the corrected projections and monochromatic images (9,10). Therefore, the GSI-MARS technology has the capacity to improve image quality in patients with prostheses (Figs 6, 7). However, the composition of the prosthesis can

Figure 7. Relapse of bone metastasis next to a saddle prosthesis that had been used for reconstruction following excision of metastatic periacetabular tumors from a thyroid neoplasm. **(a, b)** On coronal reformatted **(a)** and axial **(b)** conventional (polychromatic) CT images, the regions of bone around the prosthesis are obscured by beam-hardening artifacts (arrow in **a**). **(c, d)** Corresponding VMS CT images (80 keV) obtained with MARS allow better visualization of the soft tissues near the prosthesis. Arrow = metastasis. **(e, f)** Follow-up coronal reformatted VMS CT images (80 keV) obtained 7 months later with soft-tissue windowing **(e)** and bone windowing **(f)** with MARS show an interval increase in tumor size.

a. c.

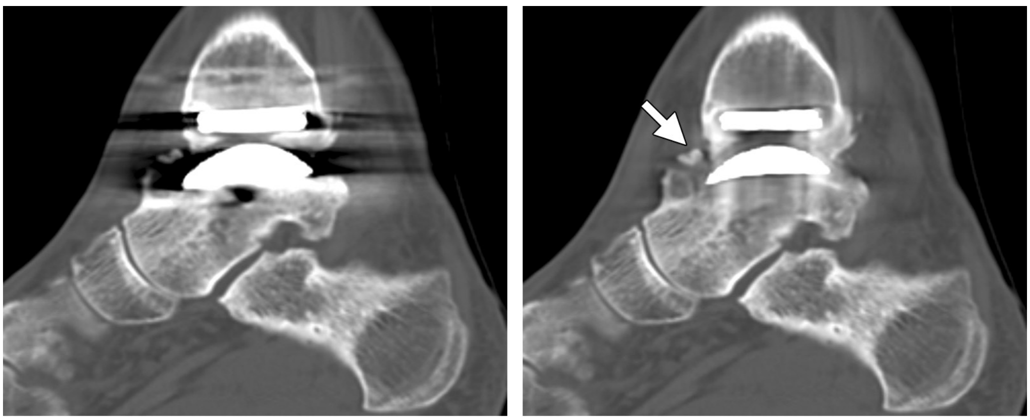
b. d.

e. f.



a. b.

Figure 8. Scapular notch in reverse total shoulder arthroplasty. A scapular notch is a defect of the bone in the inferior part of the glenoid component and corresponds to impingement of the superomedial part of the humeral implant against the pillar of the scapula. Coronal reformatted **(a)** and axial **(b)** VMS CT images (80 keV) obtained with MARS clearly depict the extent of the scapular notch (grade 4 in this case) (*), which obscures the lower screw and extends under the baseplate (arrow in **a**).



a. b.

Figure 9. Beam-hardening artifacts in total ankle arthroplasty. **(a)** On a conventional (polychromatic) CT image, the regions of bone around the metal implant are obscured by beam-hardening artifacts. **(b)** Corresponding VMS CT image (80 keV) obtained with MARS more clearly depicts a loose body in the anterior tibiotalar recess (arrow).

influence image quality. GSI-MARS CT is effective for the visualization of stainless steel prostheses (10). Dense metal alloys such as cobalt-chrome and stainless steel significantly attenuate the x-ray beam and produce more severe artifacts than do less dense metals such as titanium. For this reason, GSI-MARS reconstruction is more effective with dense metal prostheses than with titanium prostheses (10,11).

This technology can be used to reduce metal-related artifacts from a variety of causes: sinuses or facial bones compromised by dental fillings, a pelvic cavity compromised by a hip prosthesis, reverse total shoulder arthroplasty (Fig 8), total ankle arthroplasty (Fig 9), intervertebral fusion devices (Fig 10), lumbar disk arthroplasty (Fig 11), or

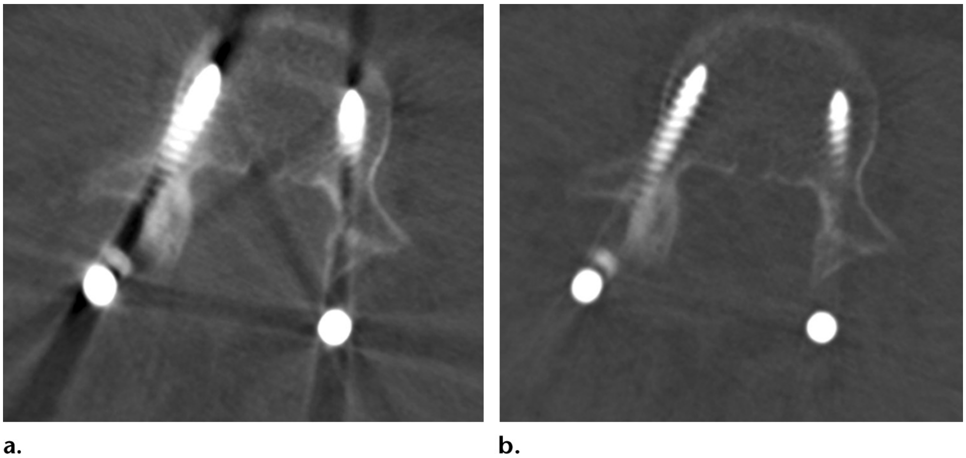


Figure 10. Beam-hardening correction in vertebral fusion. Axial VMS CT images of the lumbar spine obtained at 70 keV **(a)** and 140 keV **(b)** show fixation hardware at L4-L5.

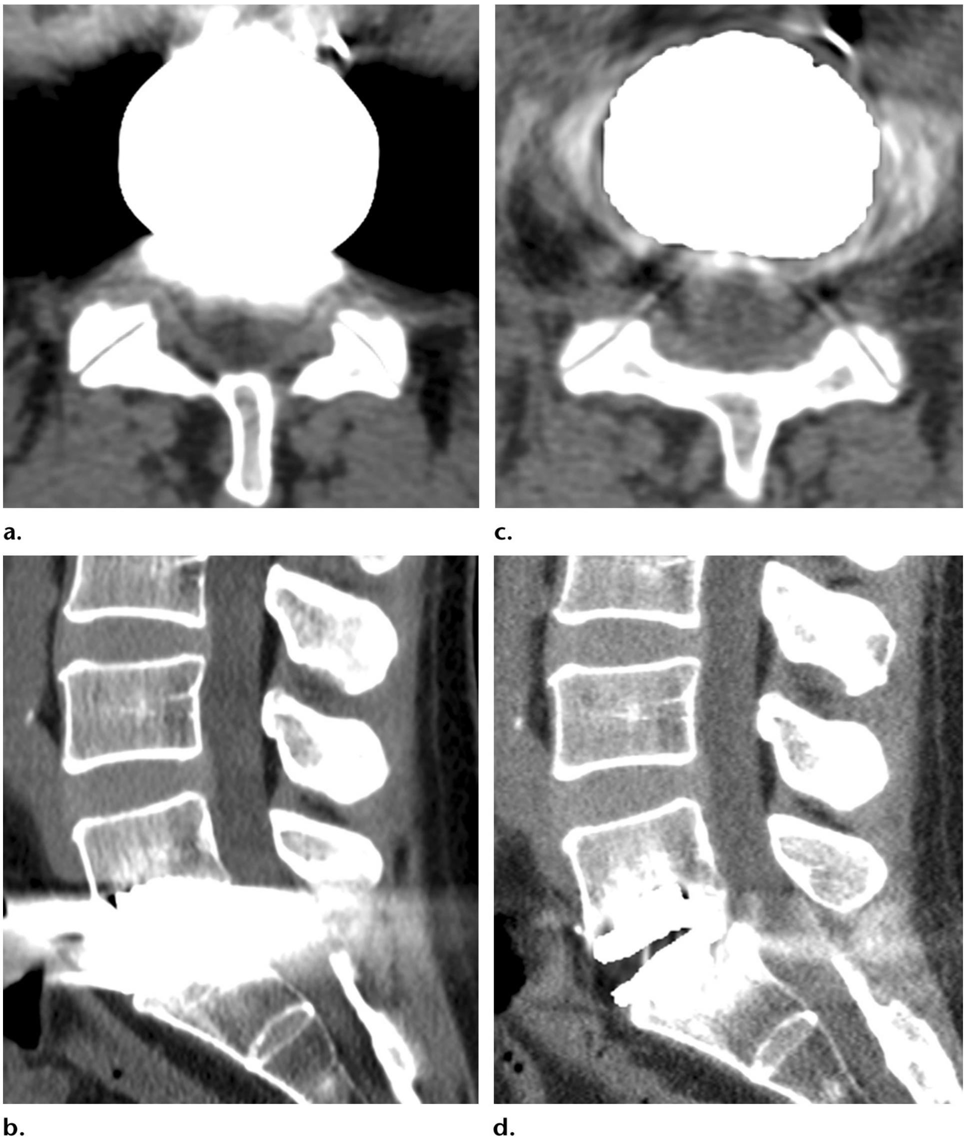
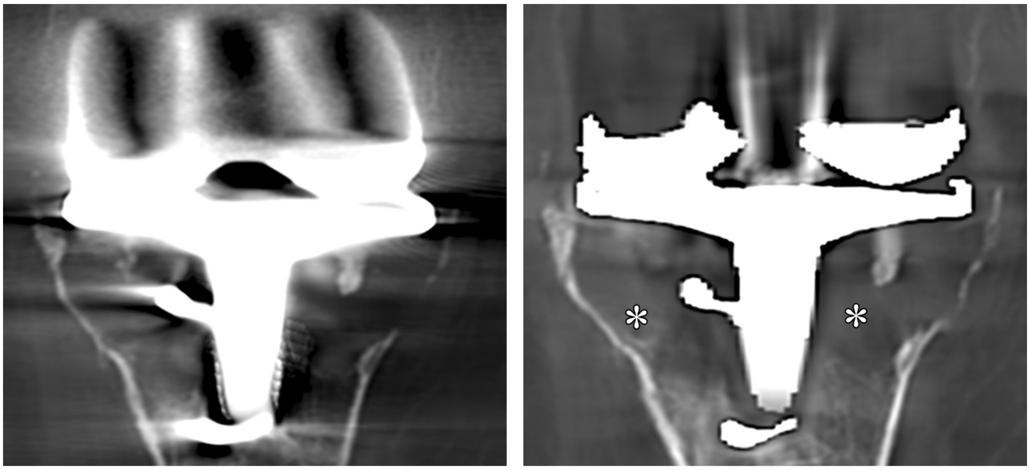


Figure 11. Beam-hardening correction in lumbar disk arthroplasty at the L5-S1 level. **(a, b)** Axial **(a)** and sagittal reformatted **(b)** conventional (polychromatic) CT images of the lumbar spine (soft-tissue windowing) show severe beam-hardening artifacts from orthopedic hardware that obscure the anatomy and limit evaluation of the spine. **(c, d)** Corresponding VMS CT images (80 keV) obtained with MARS show significant but incomplete reduction of beam-hardening artifact around the disk replacement **(c)** and in the spinal canal **(d)**.



a. b.

Figure 12. Mechanical failure with bone resorption and osteolysis after total knee arthroplasty. **(a)** Coronal reformatted conventional (polychromatic) CT scan shows severe beam-hardening artifacts from orthopedic hardware that obscure the anatomy and limit evaluation of the bone. **(b)** Corresponding VMS CT image (110 keV) obtained with MARS shows reduction of the artifacts and allows better visualization of osteolysis (*).

total knee arthroplasty (Fig 12). In most of these cases, the artifacts are not completely eliminated but are substantially reduced (Figs 11, 12), allowing better visualization of the interfaces between the implant, bone, and surrounding tissue. In a study of 31 patients who underwent dual-energy CT of the area around a metal implant, Bamberg et al (4) found that image quality improved by 49% and diagnostic value by about 44%. In clinical practice, the optimal energy level for VMS imaging is variable. To improve image quality, the user needs to select an energy value for VMS imaging that is appropriate given the circumstances surrounding the imaged object.

Radiation Dose Considerations

More recent studies of dual-energy CT performed with fast kilovoltage switching have shown radiation dose levels similar to those with conventional CT. Li et al (12) found that the weighted CT dose index from dual-energy abdominal CT performed with fast kilovoltage switching was 14% higher than that from conventional CT. In a study comparing abdominal examinations performed with dual-energy CT with fast kilovoltage switching and those performed with conventional CT with a similar volume CT dose index (26.27 mGy), Zhang et al (13) found that diagnostic performance based on interpretation of 65-keV monochromatic dual-energy CT images was equivalent to that based on interpretation of conventional images. To the best of our knowledge, however, there are no currently available data on radiation dose levels at dual-energy scanning compared with those at conventional single-energy CT for the evaluation of joints with metal implants (nonabdominal imaging). Further development of dual-energy CT noise-reduction algorithms may allow greater reductions in radiation dose levels (7).

Conclusion

The ability to obtain VMS images gives dual-energy CT potential advantages over conventional CT in reducing metal artifacts and improving image quality and diagnostic value. Evaluation of metal implants and adjacent bone or tissue is enhanced with VMS images reconstructed from dual-energy CT datasets. However, understanding principles of dual-energy CT data processing and image generation is necessary to derive maximum benefit from the dual-energy CT datasets.

References

1. Liu PT, Pavlicek WP, Peter MB, Spangehl MJ, Roberts CC, Paden RG. Metal artifact reduction image reconstruction algorithm for CT of implanted metal orthopedic devices: a work in progress. Skeletal Radiol 2009;38(8):797–802.
2. Barrett JF, Keat N. Artifacts in CT: recognition and avoidance. RadioGraphics 2004;24(6):1679–1691.
3. Lee MJ, Kim S, Lee SA, et al. Overcoming artifacts from metallic orthopedic implants at high-field-strength MR imaging and multidetector CT. RadioGraphics 2007;27(3):791–803.
4. Bamberg F, Dierks A, Nikolaou K, Reiser MF, Becker CR, Johnson TR. Metal artifact reduction by dual energy computed tomography using monoenergetic extrapolation. Eur Radiol 2011;21(7):1424–1429.
5. Matsumoto K, Jinzaki M, Tanami Y, Ueno A, Yamada M, Kuribayashi S. Virtual monochromatic spectral imaging with fast kilovoltage switching: improved image quality as compared with that obtained with conventional 120-kVp CT. Radiology 2011;259(1):257–262.
6. Silva AC, Morse BG, Hara AK, Paden RG, Hongo N, Pavlicek W. Dual-energy (spectral) CT: applications in abdominal imaging. RadioGraphics 2011;31(4):1031–1046; discussion 1047–1050.
7. Kaza RK, Platt JF, Cohan RH, Caoili EM, Al-Hawary MM, Wasnik A. Dual-energy CT with single- and dual-source scanners: current applications in evaluating the genitourinary tract. RadioGraphics 2012;32(2):353–369.
8. Cyteval C, Hamm V, Sarrabère MP, Lopez FM, Maury P, Taourel P. Painful infection at the site of hip prosthesis: CT imaging. Radiology 2002;224(2):477–483.
9. Brook OR, Gourtsoyianni S, Brook A, Mahadevan A, Wilcox C, Raptopoulos V. Spectral CT with metal artifacts reduction software for improvement of tumor visibility in the vicinity of gold fiducial markers. Radiology 2012;263(3):696–705.
10. Lee YH, Park KK, Song HT, Kim S, Suh JS. Metal artefact reduction in gemstone spectral imaging dual-energy CT with and without metal artefact reduction software. Eur Radiol 2012;22(6):1331–1340.
11. Douglas-Akinwande AC, Buckwalter KA, Rydberg J, Rankin JL, Choplin RH. Multichannel CT: evaluating the spine in postoperative patients with orthopedic hardware. RadioGraphics 2006;26(suppl 1):S97–S110.
12. Li B, Yadava G, Hsieh J. Quantification of head and body CTDI(VOL) of dual-energy x-ray CT with fast-kVp switching. Med Phys 2011;38(5):2595–2601.
13. Zhang D, Li X, Liu B. Objective characterization of GE discovery CT750 HD scanner: gemstone spectral imaging mode. Med Phys 2011;38(3):1178–1188.

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