Virtual Monochromatic Imaging in Dual-energy CT: Radiation Dose, Image Quality, and Clinical Applications

MAYO CLINIC

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Purpose

1) Provide an overview on how virtual monochromatic images are synthesized from dual-energy CT using image-domain and projection-domain methods.
2) Compare the quality of virtual monochromatic images with that of polychromatic single-energy images, acquired at different tube potentials and the same radiation dose.
3) Review clinical applications of dual-energy CT-based virtual monochromatic imaging, including beam-hardening correction, contrast and noise optimization, metal artifact reduction, and material differentiation.

Introduction

Since the introduction of dual-source CT scanners and scanners capable of fast tube potential (kV) switching, dual-energy CT has been applied in many clinical areas, including automatic bone removal, stone composition characterization, virtual non-contrast imaging, diagnosis of gout, and assessment of myocardial blood supply. All these clinical applications require dual-energy processing to obtain material-specific information.[1,2] For example, iodine and soft tissue content of each voxel are quantified through a process known as "basis material decomposition" in order to generate iodine map images and virtual non-contrast images.

In addition to the material-specific images, a single set of images is usually generated for routine diagnosis, similar to conventional polychromatic single-energy CT images. These images utilize all the radiation dose in the dual-energy acquisition. A simple approach to generating these images is to perform a linear or non-linear combination of the CT images reconstructed from the low- and high-energy data. Another approach, which may improve the image quality of this single set of images, is to synthesize virtual monochromatic images from the dual-energy scans.[1]

The approach to synthesizing monochromatic images generally relies on basis material decomposition using the knowledge of mass attenuation of the basis materials. Depending on the CT system used to acquire the dual-energy data, the approach of basis material decomposition differs slightly. For dual-energy CT data acquired with a single-source fast kV switching technique, the basis material decomposition is performed in the projection domain. For dual-energy CT data acquired with dual-source CT systems, image domain basis material decomposition is typically used. This is primarily because the projection data from the low- and high-energy scans collected by the dual-source system in a helical mode are not coincident with each other.[3] Iterative methods have been developed to enable projection-domain dual-energy processing of dual-source CT data, but have not yet been implemented in clinical practice.[4]

In this educational exhibit, we summarize how virtual monochromatic images are synthesized from dual-energy CT, using either projection-domain or image-domain methods. We compare the iodine contrast, image noise, and iodine contrast to noise ratio (CNR) between virtual monochromatic images and polychromatic single-energy images acquired at different tube potentials and the same radiation dose. Finally, we will discuss clinical applications of virtual monochromatic imaging, including beam hardening reduction, contrast and noise optimization, metal artifact reduction, and material differentiation.

Synthesizing virtual monochromatic images from dual-energy CT

Projection-based method

Figure 1 illustrates a projection-based method to synthesize virtual monochromatic images. In principle, virtual monochromatic CT images synthesized in the projection domain can fully eliminate beam hardening artifacts. **Figure 2** demonstrates that typical beam-hardening correction methods in single energy CT cannot completely remove the artifacts caused by polychromatic x-ray beams, while the monochromatic image generated from dual energy CT shows no beam-hardening artifacts.

Projection domain basis material decomposition is briefly described below. In the diagnostic x-ray energy range, the linear attenuation coefficient of a voxel at energy can be expressed as a linear combination of the mass attenuation coefficient of two basis materials:[5]

$$\mu(E) = \left(\frac{\mu}{\rho}\right)_1(E)\cdot\rho_1 + \left(\frac{\mu}{\rho}\right)_2(E)\cdot\rho_2 \quad (1)$$

where $\left(\frac{\mu}{\rho}\right)_i(E)$ $i=1,2$ denote the mass attenuation coefficients of the two basis materials, and ρ_i $i=1,2$, the mass densities of the two basis materials. In the original work by Alvarez and Macovski,[1] the attenuation coefficient was decomposed into Compton and photoelectric interactions. The basis material decomposition in Eq. (1) is fundamentally equivalent to the Compton and photoelectric decomposition because two basis materials with distinct atomic numbers can span the same linear space as that by the two types of interactions.[5]
Two non-linear equations are then solved to determine the mass density sinogram of each basis material. In practice, solution of the two non-linear equations is obtained from calibration measurements. The calibration establishes a relationship between the line integral of the mass density of each basis material and the measured data at low- and high-energies. This process inherently takes into account the effect of energy spectra and detector energy response. The mass density map for each basis material can then be reconstructed using typical reconstruction methods.

From the knowledge of the mass density maps of the two basis materials and the mass attenuation coefficients of the two basis materials at each energy, monochromatic images can be synthesized using the linear combination in Eq. (1). This method is theoretically accurate for materials spanning a wide range of atomic numbers.

Image-based method

One can also create virtual monochromatic images based on reconstructed low- and high-energy images. The process is similar to the projection-based method, except that solving for the density maps of the two basis materials is performed using images reconstructed from low- and high-energy scans.

Similar to Eq. (1), the linear attenuation coefficients at low- and high-energy scans after image reconstruction can be expressed as a linear combination of the mass attenuation coefficients of the two basis materials:

$$\mu^k = \left(\frac{\mu}{\rho}\right)_1^k \rho_1 + \left(\frac{\mu}{\rho}\right)_2^k \rho_2 \quad k = L, H \quad (2)$$

Solving the two linear equations, one can obtain the mass density of the two basis materials, which can be used to calculate the monochromatic image at energy E by using Eq. (1).

In practice, image-domain virtual monochromatic images also use a series of calibration measurements of CT numbers for different concentrations of the two basis materials in low- and high-energy scans. To accommodate the effect of different patient sizes, this calibration measurement can be performed at multiple attenuation levels. It has been shown that the monochromatic image generated from image space data are simply a linear combination of the two CT images at low and high-energies, with the weighting factor being a function of monochromatic energy.[3]

Figure 1

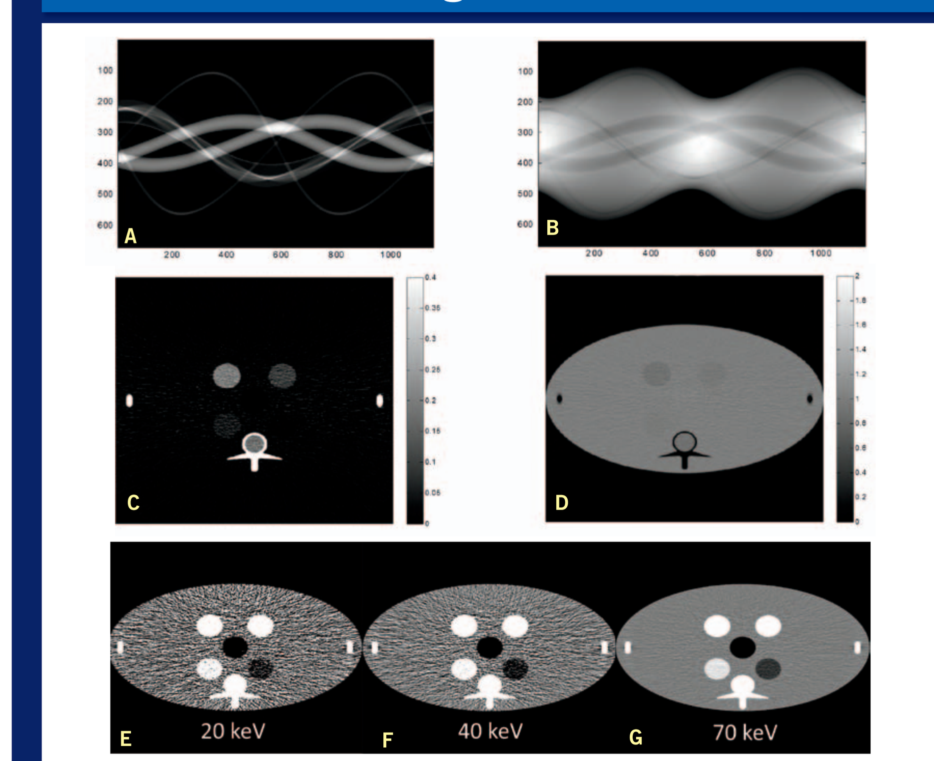

Fig. 1 Computer simulation of projection-based method, synthesizing monochromatic images from dual-energy projection data.
Bone (A) and water **(B)** sinograms after dual-energy basis material decomposition.
Bone **(C)** and water **(D)** density maps after reconstruction.
20 keV **(E)**, 40 keV **(F),** and 70 keV **(G)** monochromatic images synthesized from bone and water maps. (width, 100 HU; level, 0 HU).

Figure 2

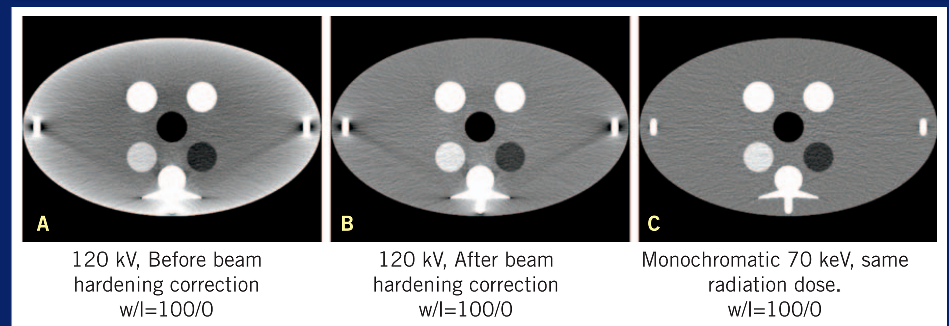

Fig. 2 Computer simulation of beam-hardening effect from polychromatic X-ray beam
120 kV image **(A)** before beam-hardening correction. Note cupping artifact extending inward from phantom edge and dark banding artifacts extending from small dense lateral structures toward spine.
120 kV image **(B)** after beam-hardening correction. Note absence of cupping artifact and retension of phantom banding artifact.
70 keV **(C)**, Complete elimination of beam-hardening artifacts using virtual monochromatic images synthesized from dual-energy scans is theoretically possible.

Image quality of virtual monochromatic images

To evaluate the image quality of virtual monochromatic images synthesized from dual-energy scans, two questions need to be answered. One question is how well the beam hardening artifact can be corrected and how accurate the CT numbers can be. The other question is how the quality of virtual monochromatic images compares with conventional single-energy CT images acquired with polychromatic x-ray beams and the same radiation dose.

Beam hardening artifact correction and quantitative accuracy of CT number

The primary difference between projection-based and image-based methods lies in how the mass density of the two basis materials are determined. The projection-based method solves for mass density integration at each projection first and then reconstructs the mass density images of the two basis materials; the image-based method directly solves for mass density based on images that are already reconstructed. The two approaches are not equivalent due to the non-linear processes involved in solving for density. In principle, projection-based methods should be more effective in terms of beam hardening artifact correction because beam-hardening occurs in each x-ray projection. While beam hardening correction techniques are usually applied prior to image reconstruction, these correction techniques are not always accurate, especially when iodine and other dense structures are present. Even with iterative beam hardening correction methods, residual artifacts may still exist after correction when exact knowledge of the physical model (spectrum, detector, imaging materials) is not available. Therefore, virtual monochromatic images created in the image domain may still contain beam hardening artifacts propagated from the low- and high-energy images, especially in the presence of dense bone or iodine.

Despite the theoretical benefit of projection-based method over image-based method, existing studies have not shown this benefit in practical implementations.

Goodsitt et al[6] evaluated the accuracy of CT number and effective atomic number measured in virtual monochromatic images obtained with a projection-based fast-kV switching dual-energy technique. Their results showed that, although the computed effective atomic numbers were reasonably accurate (within 15%), CT number inaccuracies remained, especially for dense materials at low energies, and there was still a dependency on patient/phantom size. Beam-hardening artifacts still exist, particularly for images at lower monochromatic energies **(Fig. 3)**. Their evaluation demonstrates that the synthesized virtual monochromatic images generated from the current fast-kV switching dual-energy technique are not truly monochromatic, even though they are processed in the projection domain. Further research is required to fully realize the theoretical benefit of the projection-based method. Currently, it remains unclear which implementation, fast-kV switching projection-based or dual-source dual-energy image-based methods, has an advantage over the other regarding beam hardening artifact reduction and quantitative CT number accuracy.

Figure 3

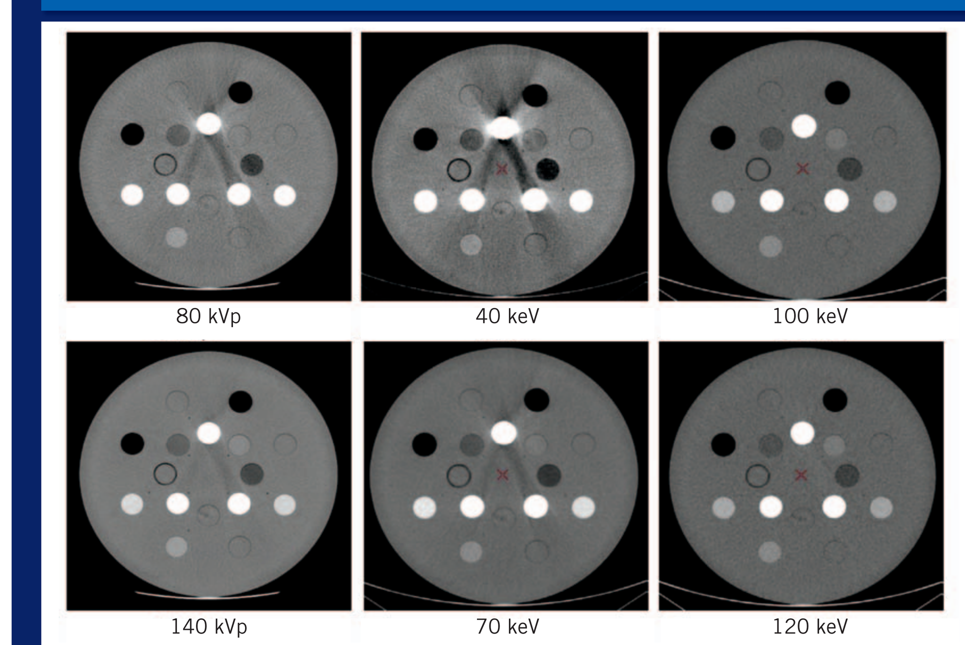

Fig. 3 Polychromatic and virtual monochromatic CT images of electron density phantom (Gammex 467, Gammex) that contains samples of various atomic numbers and densities. Images were acquired using a scanner capable of fast kilovoltage switching (HD750, GE Healthcare). Virtual monochromatic images were reconstructed using projection-based technique. Significant streaking and shadowing can be seen in 80-kVp polychromatic and 40- and 70-keV virtual monochromatic images, but not in 100- and 120-keV virtual monochromatic images. Display window level and width were 50 and 500 HU, respectively. (Reprinted with permission from)[6]

Image quality comparison between monochromatic and polychromatic images at same radiation dose.

In addition to the material-specific information provided by a dual-energy CT exam, interpreting physicians also need a single set of images for routine diagnosis. Virtual monochromatic images appear to be ideal for this purpose. An important question is how the image quality of virtual monochromatic images compares with conventional single-energy CT images acquired with polychromatic x-ray beams at the same radiation dose. If the image quality is the same or better, then dual-energy CT virtual monochromatic images may be used for routine diagnosis without requiring any more radiation dose.

In a recent work, Yu et al[3] evaluated the image quality of virtual monochromatic images generated from dual-source dual-energy scans and compared it with that from single energy CT scans acquired at various tube potentials but using the same radiation dose. The effects of several important technical factors were considered, including patient size, dose partitioning, and image quality metric to be optimized. They found that monochromatic images provided a consistent iodine CT number across phantom sizes, whereas polychromatic single-energy images showed larger differences, primarily because of the beam-hardening effect **(Fig. 4)**. With optimal monochromatic energies, the noise level was similar to that of single-energy scans at 120 kV for all phantom sizes included in this study **(Fig. 5)**. With respect to iodine CNR, monochromatic images were similar to or better than single-energy images at 120 kV for the same radiation dose **(Fig. 6)**. Therefore, if dual-energy CT is performed to obtain material-specific information, monochromatic images synthesized from the dual-energy scan can provide image quality equivalent to or better than at 120 kV with no increase in radiation dose. However, comparing virtual monochromatic images from dual-energy CT to polychromatic single-energy CT at other tube potentials, optimal settings are dependent on patient size. Iodine CNR in virtual monochromatic images from dual-energy CT was lower than that for single energy 80-kV images for small, medium, and large phantom sizes. Therefore, if there is no desire to obtain material-specific information from the dual-energy scan, it is better to simply perform a conventional single-energy scan at the optimal tube potential, which for a given patient size and diagnostic task, can reduce radiation dose or improve image quality. Approaches for automatic selection of the most dose-efficient tube potential have been developed and implemented in practice.

Figure 4

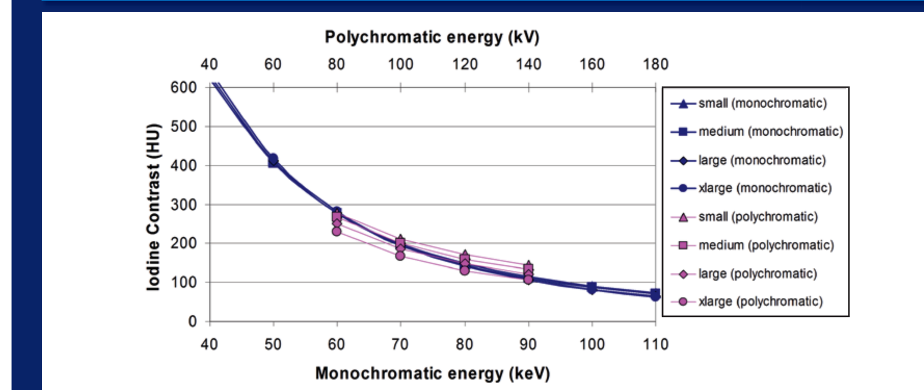

Fig. 4 Iodine contrast as a function of monochromatic energy for four different phantom sizes. Four (blue curves) are very close to each other, indicating that beam hardening in larger phantoms minimally affects CT numbers in virtual monochromatic images. These data were acquired using dual-source scanner (Definition, Siemens Healthcare). Virtual monochromatic images were synthesized using image-based technique. CT numbers of iodine measured from different polychromatic single-energy scans show larger effect of beam hardening due to phantom size (purple curves).

Figure 5

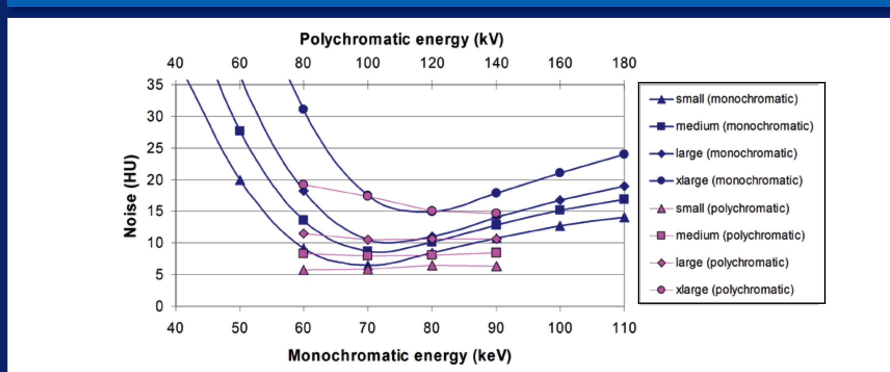

Fig. 5 Noise as function of monochromatic energy for four different phantom sizes (blue curves). For comparison, noise levels at different polychromatic beam energies are also displayed (purple curves). Radiation doses, in terms of volume CT dose index, were matched for each phantom size. Radiation dose allocated to 80 kV in dual-source dual-energy scan was 50%. Data were acquired and reconstructed in same manner as in Figure 4.

Figure 6

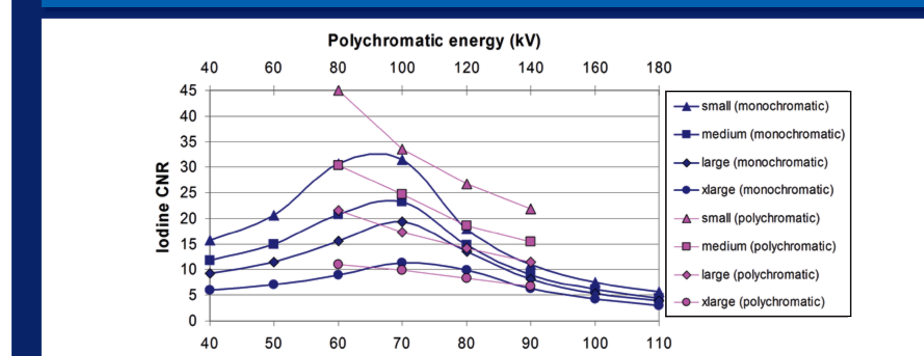

Fig. 6 Iodine contrast-to-noise ratio (CNR) as function of monochromatic energy for four different phantom sizes (blue curves). For comparison, iodine CNR at different polychromatic beam energies are also displayed (purple curves). Note that radiation doses, in terms of volume CT dose index, were matched for each phantom size. Radiation dose allocated to 80 kV in dual-source dual-energy scan was 50%. Data were acquired and reconstructed in same manner as in Figure 4.

Clinical applications of virtual monochromatic imaging

Beam-hardening correction and optimal contrast, noise, and CNR

Virtual monochromatic imaging has the potential to reduce the beam-hardening artifacts and provide more quantitative attenuation information. One clinical application is to reduce the pseudo-enhancement artifact observed in contrast-enhanced scans of renal cysts[7] **(Fig. 7)**. While this is promising, more research is necessary to improve the quantitative accuracy and beam-hardening correction.[6]

Metal artifact reduction

Virtual monochromatic images at high energies have demonstrated the ability to reduce artifacts caused by metal implants on dual-source, dual-energy CT.[8,9] Optimal monochromatic energies vary between 95 keV and 150 keV, depending on the composition and size of the metal implant. **Fig. 8** compares two images of a pedicle screw acquired by single-energy CT at 120 kV and monochromatic images at 127 keV synthesized from a dual-source, dual-energy scan for the same radiation dose. Streaking artifacts caused by the metal implant were almost completely eliminated on the monochromatic image. However, for very dense metal implants, the correction is not effective **(Fig. 9)**. This is mainly because the metal artifacts in the presence of dense metal are caused by factors in addition to beam-hardening, such as photon starvation and non-linear partial volume averaging, which cannot be corrected by synthesizing high-energy monochromatic images.

Using attenuation values from virtual monochromatic images to characterize materials

Using virtual monochromatic images at different energies, a spectral attenuation curve as a function of energy can be plotted. Different chemical compositions may be differentiated based on the characteristics of the spectral attenuation curve. For example, iodine enhancing solid masses may be differentiated from hyper-attenuating renal cysts based on the attenuation curves, since iodine attenuation greatly increases at lower energies while cyst attenuation remains relatively flat at all energies[7] **(Fig. 10)**. This is a qualitative way to differentiate materials. A more quantitative approach to characterizing materials evaluates the effective atomic number, dual-energy index, or the density map of a specific basis material calculated from dual-energy scans.

Figure 7

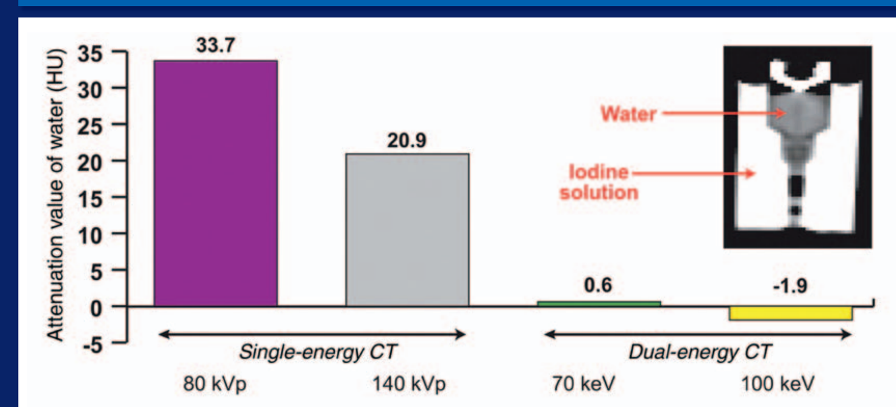

Fig. 7 Reduction of pseudoenhancement using virtual monochromatic images acquired using fast-kilovoltage-switching approach (HD750, GE Healthcare) and created using projection-domain technique. Phantom containing vial of water was placed in iodine solution (20 mg/mL) and scanned using conventional single-energy CT at 80 and 140 kV. Phantom was scanned again with dual-energy CT with fast-kilovoltage-switching technique, and virtual monochromatic images at 70 and 100 keV were synthesized. CT numbers in water were measured on each image. Significant pseudoenhancement was observed on 80- and 140-kV images, with CT numbers much higher than expected CT number of water (0 HU). Virtual monochromatic images at both 70 and 100 keV showed negligible pseudoenhancement. (Reprinted with permission from)[7]

Figure 8

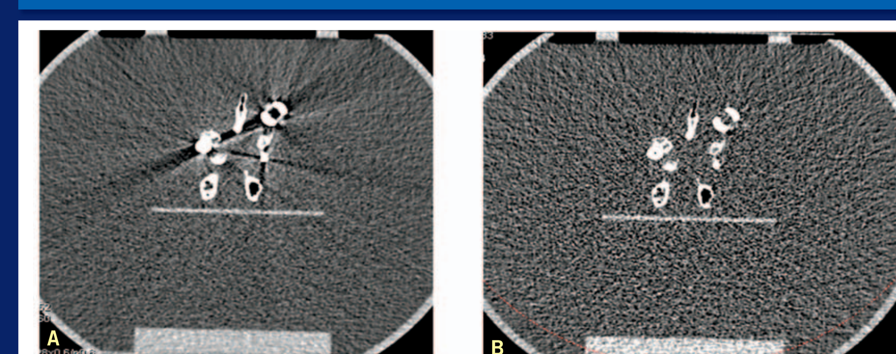

Fig. 8 Metal artifact reduction using virtual monochromatic images synthesized from dual-source dualenergy CT data using image-space techniques. **(A)**, Image shows pedicle screw in water phantom, acquired with single-energy scan at 120 kV. **(B)**, Monochromatic image at 127 keV generated from dual-energy scan with same scanner output (volume CT dose index) as used in single-energy scan. Streaking caused by metal was almost completely eliminated.

Figure 9

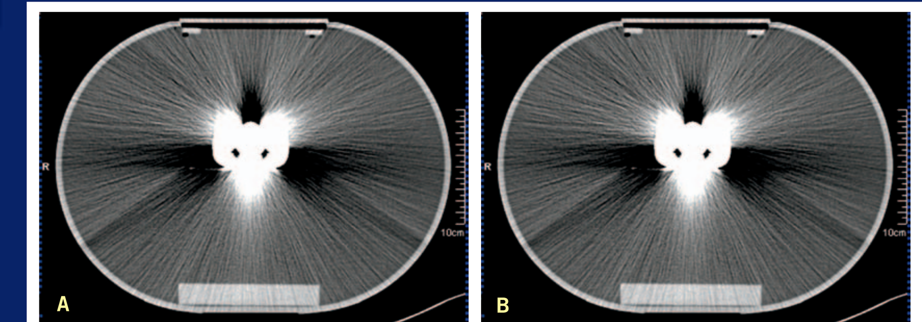

Fig. 9—Metal artifact reduction using virtual monochromatic images synthesized from dual-source dual-energy CT is ineffective for dense metal objects. **(A)**, Image of dense metal implant in water phantom acquired with single-energy scan at 120 kV. **(B)**, Monochromatic image at 127 keV generated from dual-energy scan with same scanner output (volume CT dose index) as used in single-energy scan. Streaking caused by metal is worse on virtual monochromatic image.

Figure 10

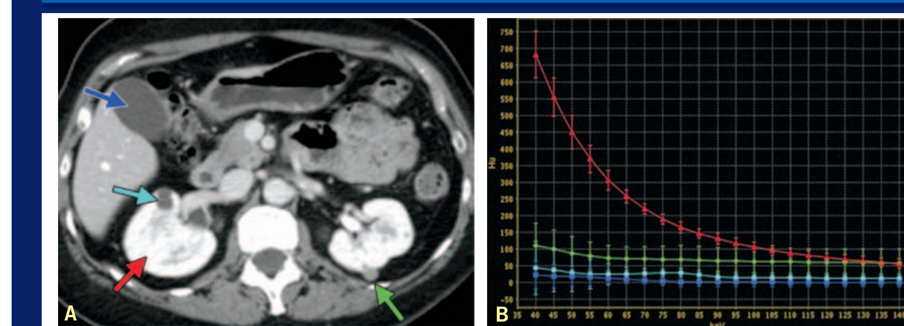

Fig. 10—Use of spectral attenuation curve to determine simple and hemorrhagic renal cyst.
(A), Virtual monochromatic image synthesized from fast-kilovoltage-switching dual-energy CT. Arrows point to hyperattenuating left renal lesion (green), hypoattenuating right renal lesion (light blue), gallbladder fluid (dark blue), and renal parenchymal enhanced by iodinated contrast material (red). Left renal lesion (green) is indeterminate and could be either enhancing mass or hyperdense cyst.
(B), Spectral attenuation curves measured from virtual monochromatic image for left renal lesion (green), right renal lesion (light blue), gallbladder bile (dark blue), and renal parenchyma (red) at various kiloelectron volt values. Attenuation curves of both renal lesions are similar to that of gallbladder fluid, which remains relatively flat, indicating nonenhancing tissues. In contrast, attenuation curve of renal parenchyma shows sharp increase at low energies, which is characteristic of iodine-containing materials. Therefore, spectral attenuation curves obtained from virtual monochromatic images helped to determine that left hyperattenuated renal lesion is hemorrhagic or proteinaceous cyst. (Reprinted with permission from)[7]

Conclusions

1) Virtual monochromatic images can provide similar image quality to conventional 120 kV images for the same total radiation dose.
2) Optimal virtual monochromatic energy depends on patient size, relative dose between low- and high-energy scans, and image quality metric optimized.
3) If there is no desire to obtain material-specific information, it is better to perform a single-energy scan at the optimal tube potential, which can reduce radiation dose and/or improve image quality.
4) Virtual monochromatic imaging has the potential to reduce beam hardening artifacts, and artifacts from light metal implants.

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