Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2019 February ; 10951: . doi:10.1117/12.2512956.

Prototype system for interventional dual-energy subtraction angiography

Michael A. Speidel^{a,b}, Christiane S. Burton^c, Ethan P. Nikolau^a, Sebastian Schafer^d, Paul F. Laeseke^c

^aDept. of Medical Physics, Univ. of Wisconsin – Madison, Madison, WI, USA

^bDept. of Medicine, Univ. of Wisconsin – Madison, Madison, WI, USA

^cDept. of Radiology, Univ. of Wisconsin – Madison, Madison, WI, USA

^dSiemens Healthineers, Hoffman Estates, IL, USA

Abstract

Dual-energy subtraction angiography (DESA) using fast kV switching has received attention for its potential to reduce misregistration artifacts in thoracic and abdominal imaging where patient motion is difficult to control; however, commercial interventional solutions are not currently available. The purpose of this work was to adapt an x-ray angiography system for 2D and 3D DESA. The platform for the dual-energy prototype was a commercially available x-ray angiography system with a flat panel detector and an 80 kW x-ray tube. Fast kV switching was implemented using custom x-ray tube control software that follows a user-defined switching program during a rotational acquisition. Measurements made with a high temporal resolution kV meter were used to calibrate the relationship between the requested and achieved kV and pulse width. To enable practical 2D and 3D imaging experiments, an automatic exposure control algorithm was developed to estimate patient thickness and select a dual-energy switching technique (kV and ms switching) that delivers a user-specified task CNR at the minimum air kerma to the interventional reference point. An XCAT-based simulation study conducted to evaluate low and high energy image registration for the scenario of 30–60 frame/s pulmonary angiography with respiratory motion found normalized RMSE values ranging from 0.16% to 1.06% in tissue-subtracted DESA images, depending on respiratory phase and frame rate. Initial imaging in a porcine model with a 60 kV, 10 ms, 325 mA / 120 kV, 3.2 ms, 325 mA switching technique demonstrated an ability to form tissue-subtracted images from a single contrast-enhanced acquisition.

Keywords

dual-energy; digital subtraction angiography; C-arm; interventional procedures

1. INTRODUCTION

Digital subtraction angiography (DSA) with a C-arm based interventional x-ray system is a proven tool for the diagnosis and treatment of cerebral and peripheral vascular diseases, providing 2D and 3D contrast-enhanced images of vessel structure and flow abnormalities

without background anatomic clutter. Still, there are classes of interventional procedures in the thorax and abdomen for which DSA is challenging and often not performed, even though vasculature of interest may be obscured by anatomic background. In these anatomic regions, motion of background tissue (e.g. due to respiratory motion) leads to misregistration of the early mask images and later contrast-enhanced images. The resulting artifacts in 2D DSA can be as large as the vessel signal itself. In 3D DSA, the problem is exacerbated by the long time required to perform two rotational C-arm acquisitions.

We are investigating 2D and 3D dual-energy subtraction angiography (DESA) techniques based on fast kV switching for interventional applications which traditionally suffer from DSA subtraction artifacts. Figure 1 compares DESA and conventional time-subtracted DSA data acquisition strategies in 2D and 3D. DESA using kV-switching and weighted subtraction of high and low energy images received early attention for its potential to cancel moving tissue,¹⁻⁴ although commercial interventional solutions are not currently available. Potential advantages are i) the two images used to form the subtraction are always acquired very close in time, reducing misregistration artifacts in 2D mode, ii) the mask phase is eliminated, shortening the overall imaging time in 3D mode, and iii) material identification/decomposition techniques are possible.⁵⁻⁷

This work reports initial steps towards adapting an interventional x-ray angiography system for 2D and 3D DESA. We report measurements of kV-switching x-ray tube control and develop an automatic exposure control algorithm suitable for interventional imaging in 2D and 3D modes. The potential impact of residual motion between low and high energy image pairs in the presence of respiratory motion is evaluated through simulations. Initial dual-energy imaging is demonstrated in a porcine model.

2. METHODS

2.1 X-ray angiography platform

The dual-energy prototype is based on an interventional x-ray angiography system (Artis Zee, Siemens Healthineers, Forchheim, Germany). Investigations were performed on two dedicated research systems at the UW-Madison (nonclinical). Each system was equipped with an x-ray tube (Megalix Cat+) capable of operation at up to 125 kV, 80 kW (nominal) power with the 1.0 mm focal spot, and up to 60 frame/s pulse rates. The detector on these systems is a 30 cm x 40 cm flat panel detector with 0.154 mm element pitch.

2.2 X-ray tube control

Fast kV switching was implemented without modification to system hardware using a strategy that was recently employed by Müller et al. on the Siemens Artis Zeego platform.⁵ The Artis Zee system was equipped with a prototype rotational acquisition mode that follows the x-ray tube technique specified in a user-modifiable configuration file. The configuration file defines the accelerating potential (kV), pulse width (ms), and tube current (mA) on a frame-by-frame basis throughout the acquisition.

The dual-energy prototype allows the user to obtain image data for arbitrarily defined tube switching techniques at up to 30 energy pairs/sec. Frame-to-frame switching over wide

ranges (e.g. 60 kV to 120 kV) is a mode of operation that does not resemble conventional clinical imaging. To determine the accuracy of x-ray tube control in the dual-energy mode, time-resolved kV waveforms were measured using a RadCal AccuGold meter with an AGMS-D+ sensor (0.1024 ms sample period). To enable measurements during a 3D rotational acquisition, the sensor was affixed to the x-ray detector and the x-ray table was removed from the beam. Measurements were made with the following types of techniques programmed: i) combined kV & ms switching ii) kV stepping at fixed pulse width, iii) pulse width stepping at fixed kV. The response of the RadCal AGMS-D+ was also compared to that of a RadCal 4083 Accu-kV meter at fixed kV values.

2.3 Design of automatic exposure control algorithm

For dual-energy imaging to be practical in the interventional setting, it is necessary to automatically select an x-ray tube technique for the imaging task at hand. The technique should deliver a desired task performance at the lowest possible radiation dose, while also staying within the operational constraints of the x-ray tube and detector. It should also account for whether imaging is being performed in 2D mode, in which case dual-energy technique may be adjusted dynamically based on patient radiographic thickness in the chosen C-arm gantry angle, or in 3D mode, in which case a fixed technique may be desired throughout a C-arm rotational acquisition. Finally, the framework should be general enough to account for different tasks and image reconstruction methods.

The framework adopted for this work is outlined in Eq. 1. First, the patient water-equivalent thickness p is estimated based on i) the raw detector image brightness for the current tube technique and projection angle, and ii) prior tabulation of detector brightness versus water thickness and tube technique. Then, given a system parameter vector s that captures imaging geometry, focal spot size, and total anticipated imaging time, the AEC algorithm selects a technique vector $\hat{x} = (kV_{low}, ms_{low}, kV_{high}, ms_{high}, mA)$ by solving the following constrained optimization problem:

$$\begin{aligned} \hat{x} = \underset{x}{\operatorname{argmin}} \quad & Dose(x; p, s) \\ \text{such that} \quad & task(x; p, s) \geq goal, \\ \text{and} \quad & x \leq tubelimit(s), \quad b(x; p, s) \leq detectorlimit(s), \end{aligned} \quad (1)$$

where $Dose(x; p, s)$ is the air kerma at the interventional reference point summed for the dual-energy pair, $task(x; p, s)$ is a task performance metric (e.g. Rose model CNR) for the technique x , $goal$ is the requested task value, $tubelimit(s)$ returns a technique vector x_{lim} describing the limits of the x-ray tube, $b(x; p, s)$ is the detector brightness vs. technique and patient thickness, and $detectorlimit(s)$ is the saturation level of the detector. The x-ray tube operating limits were obtained from the system manufacturer. The kV and pulse width are switched frame-by-frame, whereas mA is fixed for a frame pair. However, the mA is allowed to vary gradually over time to compensate for overall changes in patient attenuation which commonly occur due to changes in the C-arm angle during 2D imaging. For 3D cone-beam CT acquisitions, an additional constraint can be added to lock in the kV switching technique during C-arm rotation.

Solving Eq. (1) for given p and s amounts to finding an iso-task surface in a 5D technique space, excluding regions of the surface according to constraints, and locating the point (i.e. technique) of minimum dose on the remaining surface. This process was numerically implemented by discretizing kV and ms values while leaving mA as a continuous variable. An inverse function relating the task performance metric to mA was stored for each possible discrete combination of $(kV_{low}, ms_{low}, kV_{high}, ms_{high})$. This enabled tabulation of a finite set of 5D vectors delivering an exact task value (*goal*). Thus, the initial search space was reduced to the set of pre-calculated technique vectors

$$\mathbf{x}_{ijkl}(\text{goal}; \mathbf{p}, \mathbf{s}) = (kV_{low} = kV(i), ms_{low} = ms(j), kV_{high} = kV(k), ms_{high} = ms(l), mA = f_{ijkl}(\text{goal})) \quad (2)$$

for all possible combinations of (i, j, k, l) , where i and k are indices into a vector of discrete kV values, j and l are indices into a vector of discrete ms values, and f_{ijkl} is the inverse function mapping a *goal* value to an mA value for the specific kV pair and ms pair (and for the given $\mathbf{p}, \mathbf{s}$). The techniques were culled based on x-ray tube and detector limits, and of the surviving techniques, the technique with lowest calculated air kerma per frame pair was selected.

The solution to Eq. (1) depends on experimental data and also factors outside the scope of the present work, such as scatter correction and image denoising methods which can influence the task metric. To facilitate a basic investigation of the AEC algorithm behavior, simulated images of an iodine object (1 mm thick, 350 mg/mL, 1 mm² area) with a water background (5 – 30 cm) were generated. The simulations used polyenergetic TASMIP x-ray spectra with 2 mm Al added filtration,⁸ NIST-tabulated x-ray attenuation coefficients,⁹ and a detector converter with 0.153 g/cm² CsI. These calculations assumed no filter switching at the source. The task metric was the Rose model contrast-to-noise ratio.

2.4 Study of inter-frame respiratory motion

Although 2D DESA imaging offers a reduction in motion-induced misregistration artifacts compared to time-subtraction DSA, a dual-energy solution based on kV switching at conventional frame rates (e.g. 15–60 Hz) should account for the potential for small but finite patient movements between each image frame. A study of inter-frame respiratory motion was performed to evaluate the magnitude of this effect. Dual-energy pulmonary angiography was considered, since traditional pulmonary DSA is susceptible to misregistration artifacts under free breathing conditions or when breath holding is poor (e.g. in pulmonary embolism patients). To facilitate comparisons with the ground truth DESA images that would be obtained with truly simultaneous low/high energy imaging, the study was based on simulated imaging of the dynamic XCAT phantom.¹⁰

Multi-frame image sequences were simulated centered on three points in a 5 second respiratory cycle, corresponding to maximum diaphragm speed during inspiration (at $t = 0.75$ s), maximum expiratory speed ($t = 3.25$ s), and the minimum speed at the pre-inspiratory pause ($t = 4.75$ s). Image frame rates of 30 and 60 frame/s corresponding to dual-energy pair rates of 15 and 30 Hz were considered. The dual-energy technique was 60/120 kV with a 10 ms pulse width at both energies. To produce an image frame corresponding to

a single x-ray pulse, labeled XCAT volumes centered on the chest were generated for 5 time points within the pulse period. The XCAT voxel size was $(0.3 \text{ mm})^3$. Each voxel was labeled as either pulmonary artery, airway, lung, soft tissue (muscle), bone, or air. An in-house polyenergetic ray tracer written in MATLAB (R2018b, MathWorks Inc., Natick, MA) implementing Siddon's method¹¹ was then used to create forward projections. Energy-dependent mass attenuation coefficients and material densities maintained by NIST were assigned to each material label, with the exception of pulmonary artery, which was assigned iodine mass attenuation coefficients and a density of 0.15 g/cm^3 to mimic injection of contrast agent, and lung, which was assigned a modified density of 0.2 g/cm^3 .¹² The ray tracer calculated the integral of linear attenuation coefficient for each source-detector ray and each energy bin (10–130 keV, 1 keV bin width). The results were then weighted by the x-ray spectrum of interest (e.g. 60 kV TASMIP) to obtain the transmitted polyenergetic fluence. Projections were performed in the Artis Zee geometry for a 120 cm source-to-detector distance. The detector was modeled as an energy integrator with CsI converter, and small amount of Gaussian blur ($\sigma = 1.5$ pixel) was added to approximate the effects of focal spot and detector blur. Finally, the projections for the 5 time points within the single x-ray pulse were summed to simulate motion blur. Noise was not simulated since the goal was to evaluate frame-to-frame motion.

After generation of the low and high energy image data, 2D DESA images were formed according to

$$D_{i,j} = \ln\left(\frac{L_i}{L_{air}}\right) - w \ln\left(\frac{H_j}{H_{air}}\right) \quad (3)$$

where L_i is the low-energy image from the i^{th} frame period, H_j is the high-energy image from the j^{th} frame period, L_{air} and H_{air} are low and high energy air scan images obtained in the same projection geometry with no objects in the beam, and w is a user-selected weighting factor used to cancel a material in the DESA image. Each DESA image was compared to the ground truth $D_{i,i}$ that would be obtained if it were possible to acquire both energies simultaneously.

Normalized root mean square error in the pixel values was calculated as

$$nRMSE = \sqrt{\frac{1}{n} \sum_{k=1}^n (D_{i,j}(k) - D_{i,i}(k))^2 / (\max(D_{i,i}) - \min(D_{i,i}))} \quad (4)$$

where k is pixel index and the summation is over n pixels in the image, excluding a 20-pixel border.

2.5 Demonstration in a porcine model

Imaging of a 54 kg female pig was performed as an initial demonstration. Rotational acquisition was performed with a 60 kV, 10 ms, 325 mA / 120 kV, 3.2 ms, 325 mA switching technique during injection of an iodinated contrast agent into the carotid artery (300 mg/mL Omnipaque). A total of 240 projections were acquired at 30 frame/s with a 0.8 deg angular increment. For 3D reconstruction, software on a separate workstation was used to parse low and high energy projection sets, pre-process projections, and then reconstruct

high & low energy image volumes. A weighted subtraction of low and high energy volumes was formed using a weighting factor adjusted to isolate iodine signal.

3. RESULTS

3.1 X-ray tube control

Figure 2 shows an example of a measured kV waveform for a dual-energy switching technique of 70 kV, 10 ms, 100 mA / 120 kV, 3.2 ms, 100 mA. The frame rate was 30 fps nominal (26.3 fps measured). Figure 3 shows kV and pulse width test waveforms used to check the accuracy and stability of tube control. The kV values were obtained after cross-calibration of the RadCal AccuGold readings with the RadCal 4083 meter. The ms values represent the full-width-half-max of the measured dose traces for the x-ray pulses.

The measured kV and ms values demonstrated small biases relative to requested values (+1% to +5% for kV; +4% to +12% for ms). The magnitude of the kV overshoot tended to increase with requested kV, as demonstrated in the top left panel of Figure 3. The discrepancy in ms was characteristic of a constant offset (see Fig. 3, lower left), likely due to the finite rise and fall time of each x-ray pulse. The measured vs. requested kV and ms data were each fit to a linear model of the form $y = mx + b$, as shown in the right hand panels of Fig. 3. Calibration curves for programming kV and ms were obtained by inverting the linear models, yielding: $kV_{prog} = 0.941 \text{ kV} + 2.25$, $ms_{prog} = 0.989 \text{ ms} + 0.49$ ($R^2 > 0.99$). kV calibration on a second x-ray system resulted in similar coefficients ($kV_{prog} = 0.945 \text{ kV} + 1.91$).

3.2 Automatic exposure control algorithm

Figure 4 demonstrates the calculated kV and ms switching technique as a function of background water thickness for requested Rose model CNRs ranging from 3 to 10. For this demonstration, the behavior of the AEC algorithm was investigated using polyenergetic simulations of images of an iodine object (1 mm thick, 350 mg/mL, 1 mm² area) with a water background ranging from 5 cm to 30 cm thick, TASMIP x-ray spectra with 2 mm Al added filtration, and assuming a 0.153 g/cm² CsI detector. X-ray tube technique constraints for 30 frame/s acquisition, 5 seconds of imaging, and a large focal spot were assumed. The CNR was calculated for 2D DESA images with the high energy image weighting set to cancel background water. The kV choices were discretized in 2 kV increments, and the pulse width choices were discretized in 0.4 ms increments.

For the low energy image, the selected kV was relatively constant (52–58 kV) for all water thicknesses and goal CNRs evaluated, and the selected pulse width was 16 ms for 15 cm and greater water thickness. For less than 15 cm, the selected pulse width for the low energy image declined, depending on goal CNR. For the high energy image, the selected pulse width was the minimum 3.2 ms across all thicknesses and CNRs evaluated, and the selected tube voltage was 124 kV for 15 cm and greater water thickness. For less than 15 cm and goal CNRs of less than 8, the AEC algorithm called for a reduced kV in the high energy image. The mA value selected for both energies increased with goal CNR, as expected. The full

range of goal CNRs (3–10) was achieved for water thicknesses up to 20 cm; for greater thicknesses, there was a drop off in maximum achieved CNR due to tube load limitations.

Since AEC technique versus thickness is relatively well behaved, the results can be stored in tables for rapid lookup and interpolation as estimated patient thickness changes. Figure 5 shows a control flowchart demonstrating how these tables can be used in a real-time 2D DESA imaging scenario.

3.3 Evaluation of inter-frame respiratory motion

Figure 6 shows the XCAT phantom-based 2D DESA pulmonary angiograms for 30 frame/s imaging (15 energy pairs/s) with a 60/120 kV switching technique in the cases of the greatest and least severe respiratory motion ($t = 0.75$ s and $t = 4.75$ s). A weighting factor $w = 1.3$ was selected to cancel tissue in this example. The corresponding ground truth DESA images with simultaneous high/low energy acquisition and the difference between ground truth and DESA are also shown. While the DESA images were free of severe respiratory motion-related artifact, close inspection of the $t = 0.75$ s case revealed bands of increased brightness near the smaller moving vessels (see inset). The nRMSE values were 1.06% and 0.33% for $t = 0.75$ s and $t = 4.75$ s, respectively, in the 30 frame/s images. As expected, the residual motion effect was reduced for a 60 frame/s acquisition (30 energy pairs/s). In the 60 frame/s images, the nRMSE values were 0.54% and 0.16%, respectively, for $t = 0.75$ s and $t = 4.75$ s.

Based on these findings, potential methods for correcting residual respiratory motion were investigated. The chosen correction scheme is outlined in Figure 7. To process the energy pair corresponding to frame i and frame $(i-1)$, deformable registration is applied between frame i and frame $(i-2)$, the most recent frame acquired at the same energy as frame i . The resulting vector field describing the mapping from frame $(i-2)$ to frame i is then divided by 2 to obtain the vector field mapping from frame $(i-1)$ to frame i . This approach avoids the complication of performing feature matching between frame i and frame $(i-1)$ where the features have different contrasts at low and high energies, although the temporal interpolation step assumes that local velocities are constant over the time scale of two frames. Figure 7 (right) demonstrates the performance of the correction algorithm for the scenario of 30 frame/s (15 pair/s) and $t = 0.75$ s (lowest frame rate, maximum velocity). The nRMSE was reduced from 1.06% to 0.21%. The MATLAB implementation of a non-rigid demons registration algorithm was used for this test (*imregdemons*, MATLAB R2018b).

3.4 In vivo images

Figure 8 shows cone-beam CT reconstructions of the low and high energy projections from a single contrast-enhanced rotational scan of the porcine cranium, and a weighted subtraction of the low and high energy image volumes. Volumetric images are presented with a Volume Render Technique (VRT) and a Maximum Intensity Projection (MIP) technique. A subtraction of the high energy image from the low energy image naturally provided a degree of soft tissue cancellation, due to the definition of CT number (water = 0). Application of a weighting factor to the high-energy image prior to subtraction further facilitated the removal of bone content when performing thresholding and window/level operations during VRT and

MIP creation. The 3D DESA images shown here used a weighting to emphasize iodine and de-emphasize bone, analogous to conventional 3D digital-subtraction angiography (3D-DSA). Some loss of small scale vasculature is visible in the dual-energy images, although the main cranial anatomy is well visualized. Further investigations are needed to optimize the 3D approach.

4. DISCUSSION

This study reports on the initial development of a prototype kV-switching dual-energy subtraction angiography capability for an interventional x-ray angiography platform. In diagnostic CT, dual-energy imaging has been applied for automated bone removal, perfused blood volume imaging, virtual non-contrast imaging, and material identification, among other applications.^{13–15} In the interventional setting, DESA may offer reduced motion artifact compared to time-subtracted 2D DSA, and shorter acquisition times in 3D rotational mode due to the elimination of the non-contrast mask rotation. However, implementation of kV switching on an interventional platform requires the analysis of several practical problems, including but not limited to the characterization of x-ray tube behavior, development of an automatic exposure control algorithm, and handling of residual frame-to-frame motion.

For this study, kV and pulse width switching were implemented in a prototype rotational acquisition mode which allowed for acquisition of 2D images (with rotating C-arm) or 3D images. Measurements of kV and pulse width yielded results generally consistent with those previously reported by Müller et al.⁵ Small deviations between requested and achieved values can be corrected via calibration.

The automatic exposure control algorithm described here is a general task-driven framework in which a switching technique delivering the requested task performance at the minimum dose is selected, given system constraints. A calculation for imaging of iodine on a water background with a CNR-based task metric was used to demonstrate the algorithm; however, the framework can be extended to a variety of scenarios. For example, the AEC algorithm can be applied when a filter is used with the high energy beam,¹⁶ or when more sophisticated spatial-frequency dependent task metrics are used.¹⁷ In practice, the AEC algorithm should be informed by experimental data. Specifically, the relationship between tube current and task metric should be determined for different combinations of kV and ms. Although this is a potentially laborious task, theoretical calculations of the expected switching technique can be used to guide experimental work. After the AEC calculations have been completed versus phantom thickness, the results can be stored in lookup tables.

While DESA is designed to be motion-insensitive, there is potential for minor organ shifts between low and high energy images when using kV switching. For the example scenario of pulmonary angiography with respiratory motion, a subtle effect can be expected during the period of maximum diaphragm velocity in the inspiratory phase, when employing 30 frame/s imaging (15 pair/s). In this scenario, the normalized RMSE was 1%. Simulations indicate that this deviation could be reduced, if desired, either by operation at 60 frame/s or

through application of a simple non-rigid registration scheme. Cardiac motion (e.g. coronary angiography) was not investigated and may require different strategies.

Several issues remain to be investigated. It is known that the simple weighted subtraction scheme can amplify noise, depending on the weighting factor employed, and various dual-energy denoising methods have been described.^{1,18} Similarly, the presence of scatter in the x-ray images can alter the quality of subtractions. The solutions to these issues will influence achievable image quality and dose efficiency. Additionally, further optimization of the 3D DESA acquisition and reconstruction method is needed. Future work will focus on experimental investigation of these issues.

5. CONCLUSIONS

Dual-energy imaging via kV switching was investigated on an interventional C-arm system without hardware modification. With calibration, kV and pulse width control was achieved for switching ranges from 60 to 120 kV and 3.2 to 10 ms. A task driven automatic exposure control algorithm was developed for selecting a minimum dose switching technique given an imaging task goal and the operating constraints of the x-ray system. Initial work supports the feasibility of interventional dual-energy subtraction angiography in applications where conventional DSA is influenced by respiratory motion. Further evaluation in phantoms and animal studies should be performed.

ACKNOWLEDGMENTS

Partial financial support was provided by NIH Grant No. R21 EB023008 and funding received from Siemens Healthineers. The concepts presented in this paper are based on research and are not commercially available.

REFERENCES

- [1]. McCollough CH, Van Lysel MS, Peppler WW, and Mistretta CA, "A correlated noise reduction algorithm for dual-energy digital subtraction angiography," *Med. Phys.* 16(6), 873–80 (1989). [PubMed: 2586373]
- [2]. Molloy S, Detrano R, Ersahin A, Roeck W, and Morcos C, "Quantification of coronary arterial calcium by dual energy digital subtraction fluoroscopy," *Med. Phys.* 18(2), 295–298 (1991). [PubMed: 2046617]
- [3]. Van Lysel MS, Miller WP, Senior DG, Gupta VK, Ende DJ, and Albright DJ, "Left ventricular dual-energy digital subtraction angiography: a motion immune digital subtraction technique," *Int. J. Card. Imaging* 7(1), 55–65 (1991). [PubMed: 1753160]
- [4]. Ducote JL, Xu T, and Molloy S, "Dual-energy cardiac imaging: an image quality and dose comparison for a flat-panel detector and x-ray image intensifier," *Phys. Med. Biol.* 52(1), 183–96 (2007). [PubMed: 17183135]
- [5]. Müller K, Datta S, Ahmad M, Choi JH, Moore T, Pung L, Niebler C, Gold GE, Maier A, and Fahrig R, "Interventional dual-energy imaging-Feasibility of rapid kV-switching on a C-arm CT system," *Med. Phys.* 43(10), 5537 (2016). [PubMed: 27782692]
- [6]. Zbijewski W, Gang GJ, Xu J, Wang AS, Stayman JW, Taguchi K, Carrino JA, and Siewerdsen JH, "Dual-energy cone-beam CT with a flat-panel detector: effect of reconstruction algorithm on material classification," *Med. Phys.* 41(2), 021908 (2014). [PubMed: 24506629]
- [7]. Gang GJ, Zbijewski W, Stayman JW, and Siewerdsen JH, "Cascaded systems analysis of noise and detectability in dual-energy cone-beam CT," *Med. Phys.* 39(8), 5145–5156 (2012). [PubMed: 22894440]

- [8]. Boone JM and Seibert JA, "An accurate method for computer-generating tungsten anode x-ray spectra from 30 to 140 kV," *Med. Phys.* 24(11), 1661–1670 (1997). [PubMed: 9394272]
- [9]. Hubbell JH and Seltzer SM, "X-ray mass attenuation coefficients," [NIST Standard Reference Database 126], National Institute of Standards and Technology, Gaithersburg MD, 20899.
- [10]. Segars WP, Sturgeon G, Mendonca S, Grimes J, and Tsui BMW, "4D XCAT phantom for multimodality imaging research," *Med. Phys.* 37(9), 4902–4915 (2010). [PubMed: 20964209]
- [11]. Siddon RL, "Fast calculation of the exact radiological path for a three-dimensional CT array," *Med. Phys.* 12(2), 252–255 (1985). [PubMed: 4000088]
- [12]. Van Dyk J, Keane TJ, and Rider WD, "Lung density as measured by computerized tomography: implications for radiotherapy," *Int. J. Radiat. Oncol. Biol. Phys.* 8(8), 1363–1372 (1982). [PubMed: 7141916]
- [13]. McCollough CH, Leng S, Yu L, and Fletcher JG, "Dual- and multi-energy CT: principles, technical approaches, and clinical applications," *Radiology* 276(3), 637–653 (2015). [PubMed: 26302388]
- [14]. Bai A, Sun Y, Qi L, Yang Y, and Hua Y, "Improving the image quality in computed tomographic pulmonary angiography with dual-energy subtraction: a new application of spectral computed tomography," *J. Comput. Assist. Tomogr.* 37(5), 718–724 (2013). [PubMed: 24045247]
- [15]. Meysman M, Everaert H, Bult N, Nieboer K, and de Mey J, "Comparison of ventilation-perfusion single-photon emission computed tomography (V/Q SPECT) versus dual-energy CT perfusion and angiography (DECT) after 6 months of pulmonary embolism (PE) treatment," *Eur. J. Radiol.* 84(9), 1816–1819 (2015). [PubMed: 26074385]
- [16]. Primak AN, Ramirez Giraldo JC, Liu X, Yu L, and McCollough CH, "Improved dual-energy material discrimination for dual-source CT by means of additional spectral filtration," *Med. Phys.* 36(4), 1359–1369 (2009). [PubMed: 19472643]
- [17]. Richard S, and Siewerdsen JH, "Optimization of dual-energy imaging systems using generalized NEQ and imaging task," *Med Phys.* 34(1), 127–139 (2007). [PubMed: 17278498]
- [18]. Richard S, and Siewerdsen JH, "Cascaded systems analysis of noise reduction algorithms in dual-energy imaging," *Med Phys.* 35(2), 586–601 (2008). [PubMed: 18383680]

Figure 1.

Data acquisition for conventional time-subtracted DSA (left) versus dual-energy subtraction angiography (DESA) using kV switching (right). In conventional 2D mode, an early mask is subtracted from later contrast-enhanced frames. In conventional 3D mode, a mask rotation is followed by a contrast-enhanced rotation and images are subtracted angle-wise before volume reconstruction. In 2D DESA, mask data is not acquired and subtraction images are formed from low/high energy frame pairs acquired during contrast enhancement. In 3D DESA, the mask rotation is eliminated, and a subtraction volume can be generated either in image space after reconstructing high and low energy volumes, or by reconstructing a set of 2D DESA images formed in projection space.

Figure 2.
Excerpt of kV waveform for a switching technique of 70 kV, 10 ms / 120 kV, 3.2 ms.

Figure 3. Top row: measured kV values for a kV stepping program (left), and linear fit to the measured vs. requested kV data (right). Bottom row: measured ms for pulse width stepping program, and linear fit.

Figure 4. Calculated kV and ms switching techniques for constant iodine CNR in 2D DESA versus background water thickness. The same tube current (right) is used for both frames in a low/high energy pair.

Figure 5.
Automatic exposure control flowchart.

Figure 6.

Comparison of 30 frame/s pulmonary DESA (15 energy pair/s) to ground truth for two points in the respiratory cycle, based on XCAT simulation. Images are shown for maximum motion in the inspiratory phase (top row, $t = 0.75$ s) and minimum motion during the pre-inspiratory pause (bottom row, $t = 4.75$ s). The difference (right) between DESA and the ground truth image is displayed with a tight window to exaggerate differences. 60/120 kV switching was simulated and a high energy weighting of 1.3 was used to cancel soft tissue and leave iodine/bone.

Figure 7. Registration method to account for residual respiratory motion. The motion vector field derived from the current (L_i) and the most recent image of the same energy (L_{i-2}) is interpolated to the most recent image of the opposite energy (H_{i-1}). Images at right show simulated pulmonary angiography during maximum inspiratory motion, without and with the registration algorithm applied.

Figure 8. Left: volume rendering (VRT) of the low and high energy image reconstructions of the head. Cranial bone, teeth, and contrast-enhanced vasculature are visible. Right: VRT and MIP rendering of a weighted subtraction of high and low energy image volumes. The weighting was adjusted to emphasize iodine signal.