



White paper

Shaping the beam

Versatile filtration for unique diagnostic potential within Siemens Healthineers CT

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Contents

Technical implementation	3
Clinical implementation	10
Conclusion	21
Literature	22

Technical implementation

Siemens Healthineers Computed Tomography (CT) scanners, built on over 40 years of innovation and experience, are prime examples of how far technology can evolve.

All CT systems, at a basic level, consist of an X-ray generating source (X-ray tube) and an X-ray detection device (detector).

In this document we will concentrate on the tube side of the equation, and more specifically, the purpose of this white paper is to address the topic of filtration in CT imaging. Firstly by examining the technology, from the initial implementation to unique methods being utilized by Siemens Healthineers, and how each implementation addresses dose topics in CT while maintaining or potentially improving diagnostic image quality. Furthermore, this paper will address the clinical implementation of filtration and how it can be used in a variety of ways in clinical routine to improve diagnostic outcomes.

Beam Generation

The tube consists of a cathode that emits electrons, accelerated by an electrical field, and an anode that the electron beam hits. At the anode (combinations of tungsten, rhenium and/or molybdenum), the electrons are slowed down resulting in an X-ray beam with a certain spectrum (a mixture of Bremsstrahlung and characteristic spectrums).¹

The spectrum includes photons over a wide energy range where the maximum photon energy (keV) of the spectrum is determined by the peak tube voltage (kVp).

Beam Hardening

This theoretical spectrum (Fig. 1, pre-patient spectrum) sent through an object, such as a patient, would be sufficient to be detected at the detector, however, such an X-ray beam may not be ready for diagnostic imaging, due to dose and image quality reasons. Attenuation coefficient curves demonstrate that there is an energy dependency for photon absorption by human tissue (Fig. 2). This leads to two different problems.

With each cm of patient tissue to be penetrated along the beam path, the mean energy of the spectrum is shifted to a higher energy level, since low energy photons are more likely absorbed or scattered by the patient than high energy photons. In other words, the beam hardens, and this effect is greatest for the first cm and then slows down (Fig. 1). With a hardened beam and higher mean energy level, the mean attenuation coefficient changes and affects the stability of Hounsfield units by so-called beam hardening artifacts.

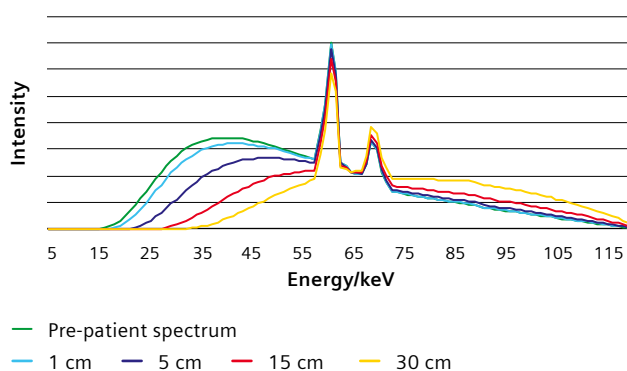


Fig. 1: Spectra before (pre-patient spectrum) and after penetrating water objects of different thicknesses, including the base tube filtration from the anode and tube output window.

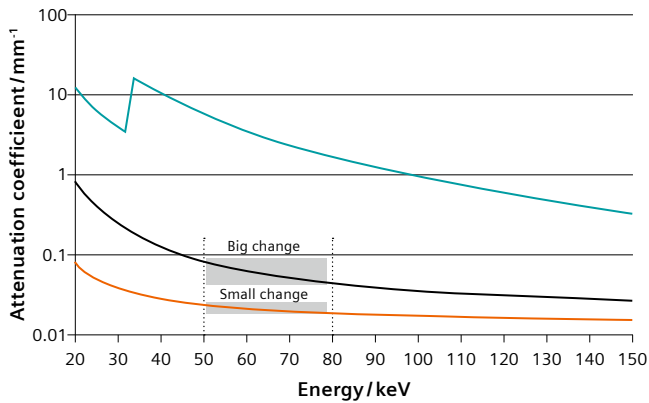


Fig. 2: Energy dependence of the attenuation coefficient of different materials. The different change in attenuation at different energies is illustrated for water (orange) and bone (black) and iodine (petrol).

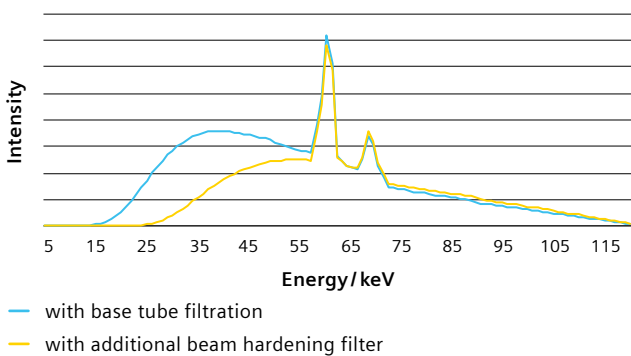


Fig. 3: Comparison pre-patient spectra base tube filtration (anode and tube output window) and additional beam hardening filter. The area between the curves would result in higher dose levels.

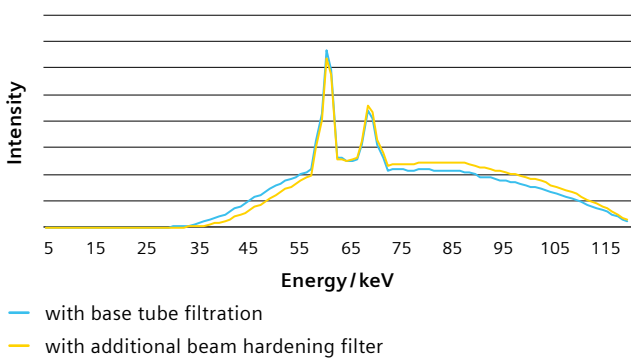


Fig. 4: Comparison post-patient (30 cm water equivalent) spectra with base tube filtration (anode and tube output window) and additional beam hardening filter. This shows that despite a lower dose level, the output spectra are equivalent and therefore image quality is maintained.

Therefore, each SOMATOM CT scanner includes software-based correction methods like water beam hardening correction and iterative beam hardening correction (IBHC) for iodine and bone. Since prevention is better than correction, a reduction of the patient-related beam hardening effect is needed.

The attenuation by the patient is, as such, high for photons within the energy range from 1 to roughly 30 keV such that all these photons of the beam will most likely interact with patient tissue.

Photons in this energy range do not contribute to the measured signal and are therefore useless for imaging but still contribute to patient radiation dose. For diagnostic imaging, the dose level needs to be kept as low as reasonably possible, and therefore photons with energy ranges most likely not contributing to a signal must be eliminated or at least reduced in intensity.

Since the very beginning of diagnostic CT, pre-patient filters of a certain thickness have been permanently placed in the beam path to absorb the very low energy photons and pre-harden the beam (Fig. 3). In this way, a reduction in patient dose level is enabled while maintaining the information level of the detected signal (Fig. 4).

This technique of pre-patient filtration is therefore referred to as beam-hardening filtration.

The X-ray tube assembly already includes a level of filtration that is expressed as the half value layer in an aluminum-equivalent thickness. For example, the Chronon® tube includes a beam hardening filtration equivalent to ≥ 5.5 mm Al at 140 kV as part of the tube housing assembly. As the unit of filtration indicates, materials and combinations other than aluminum can be used for pre-hardening the beam, e.g., brass, copper, lead, titanium, and tin. Within this work, we will stick to aluminum for the sake of simplicity.

In addition to the beam hardening filtration of the X-ray tube assembly, one can add further permanent filters, e.g., as part of the beam-limiting device, such as in the imaging chain of the SOMATOM Force (additional 0.3 mm thick titanium filter).

Beam Shaping

The simplicity of pre-filtration of the X-ray beam varies from the aforementioned filtration based on a certain thickness of aluminum or titanium to more advanced methods such as a bow-tie filtration. The bow-tie filter, as the name suggests, is a separate shaped filter placed in front of the X-ray tube radiation exit window, which is thicker on the periphery of the beam comparative to the center (Fig. 5).

What are the physical and technical reasons for using a filter shaped like a bow-tie?

The intersection length of an X-ray path through the patient differs greatly between central rays and for peripheral rays. This difference in intersection length has several implications that are addressed by the shape of the bow-tie filter.

The beam hardening effect described in the previous section, for example, differs with the intersection length. In order to reduce beam hardening artifacts, the differences in intersection length with the patient can be compensated by a filter with an inverse patient shape like the bow-tie filter has.

The shorter the intersection length, the lower the number of photons being absorbed, and some rays may not intersect the patient at all. The difference in absorption rate results in a large variation of the detector signal, and in a worst-case scenario, in detector saturation with a loss of information.

In addition to this technical reason, the difference in absorption rate also needs to be compensated for further image quality reasons. The image noise level depends on the signal-to-noise ratio (SNR) of the detector signal. High variations in SNR due to different intersection length results in inhomogeneous image noise with lower noise levels at the patient periphery than in the patient center.

In addition, image noise level is closely linked to patient radiation dose. The lower noise level at the periphery indicates a higher local patient dose there.

Photon scattering is another important interaction effect in CT imaging in addition to photon absorption via photo-electric effect. Scatter photons create noise on the primary signal and a high scatter-to-primary ratio results in image artifacts (Fig. 5c). This is the case when a large number of scattered photons from a short intersection length contribute to the signal for a small number of primary photons from a long intersection length. In order to improve the scatter-to-primary ratio, the number of incident photons for peripheral body parts is decreased and thus the number of scattered photons as well.

Fig. 5b shows the different relationship of the shape of the bow-tie filter to the patient size, in relation to the noise (scatter-to-primary ratio). Here we see that the bow-tie 2 would provide the most homogeneous dose distribution of the four options displayed.

The bow-tie filter performs so-called beam shaping in CT imaging and thus accounts for differences in intersection length with the patient in order to avoid detector saturation, to reduce scatter and further beam hardening artifacts, and to enable homogeneous image noise.

By using either of these methods, beam hardening or beam shaping, every CT manufacturer can claim to have implemented a form of pre-filtration of the X-ray beam into their CT scanners.

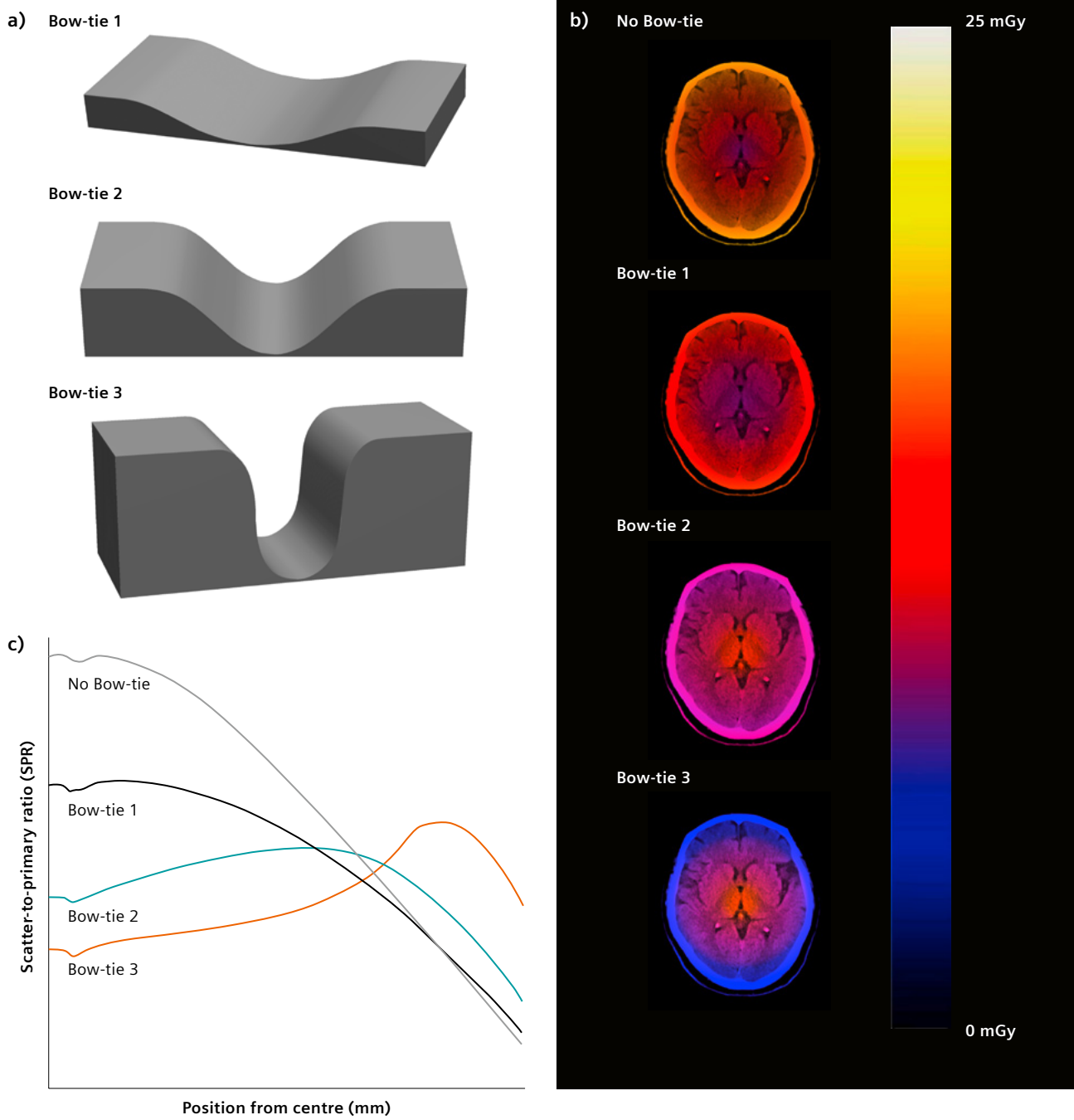


Fig. 5: a) Graphical representation of different bow-tie filters (top left), and b) a representation of the effect on the detector read-out based on the different types (right), and c) the positioning within (bottom left).

Spectral Shaping

So far, we have addressed two aspects of X-ray beam pre-filtration. Removal of low keV photons from the spectrum, utilizing a standard filter (**beam hardening filtration**) and then further removing photons from the periphery of the scan field by utilizing a shaped bow-tie (**beam shaping filtration**).

These methods have been at the basis of CT for many years, and for many providers of CT systems, these have not been extensively adapted or modified despite the advances in high-voltage generator and tube technology and detector technology.

Siemens Healthineers have placed further emphasis on the concept of filtration prior to the patient in an effort to provide a more precise diagnostic tool for a range of pathologies, be it by providing new information with Dual Energy spectral imaging or with ultra-low-dose scanning to enable CT, where previously patient dose was a concern to referring physicians.

One such concept, as outlined by Kelcz et al.⁶, was to introduce pre-filtration to the tube in order to reduce even further the number of low keV photons and modify the average intensities of the X-ray spectrum to an even greater extent.

Tin Filter

Selected from a range of practical nonradioactive, nonvolatile (solid at room temperature), machinable, single elements within the atomic number range of $40 \leq Z \leq 83$, the element Sn (tin) was evaluated as the element with the most potential for additional spectral shaping.

A moveable, selectable filter made of Sn (tin) of a certain thickness, referred to as the Tin Filter, is then positioned in front of the X-ray tube radiation exit window, in addition to the existing standard filtration (permanent beam hardening plus beam shaping filter). This leads to a narrower X-ray tube spectrum with fewer quanta at lower energies and a resulting higher mean energy level (Fig. 6 and Table 1).

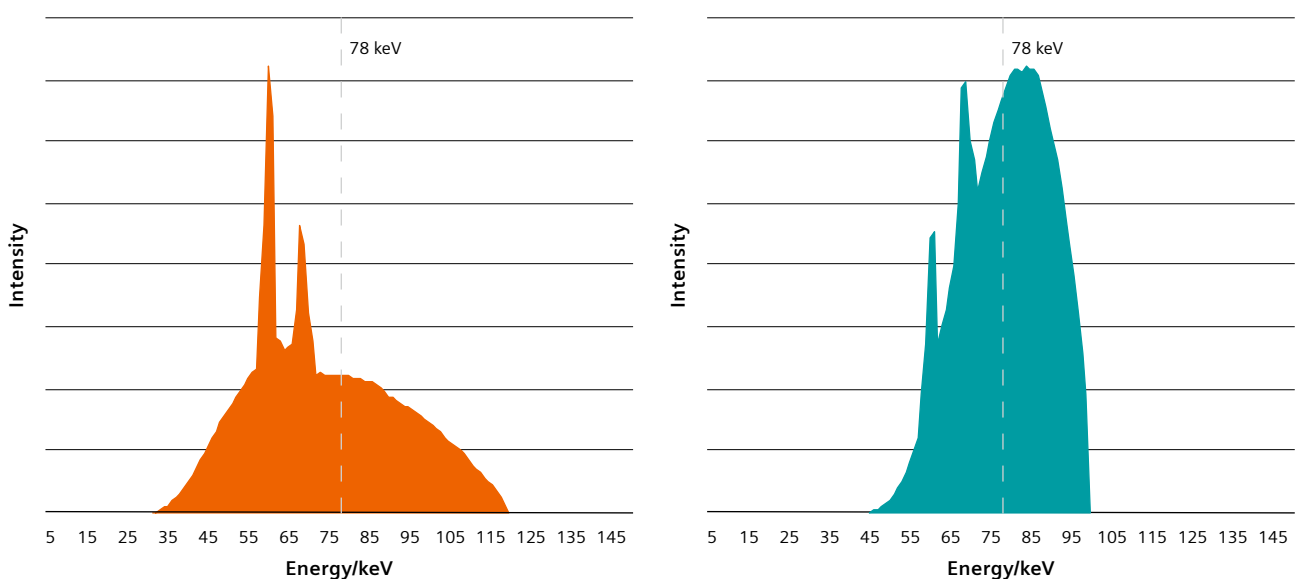


Fig. 6: (left) 120 kV spectrum with standard pre-filtration; (right) 100 kV spectrum with standard pre-filtration plus the additional Tin Filter (Sn100 kV), showing that the average mean intensity is similar even with different X-ray potentials.

Mean energy of X-ray spectrum after passing through object				
X-ray source spectra and filtration	Air [keV]	20 cm [keV]	30 cm [keV]	40 cm [keV]
120 kV	69	78	81	84
Sn100 kV	75	78	79	80

Table 1: The beam hardening effect at 120 kV and Sn100 kV of the additional water object. Mean keV values derived from the different spectra simulations.⁹

Additionally, the Sn100 kV spectrum (100 kV X-ray tube voltage with standard plus tin pre-filtration).

This is highly desired, because it reduces uncertainty in CT imaging, and may lead to a potential improvement in quantitative measurements of the X-ray intensity.

Note also that at Sn100 kV the image noise at the same radiation dose is always lower than at the standard X-ray tube voltage of 120 kV (for all water-equivalent diameters) (Fig. 7). This additional noise reduction would allow for dose reduction potential if a constant image noise is maintained.

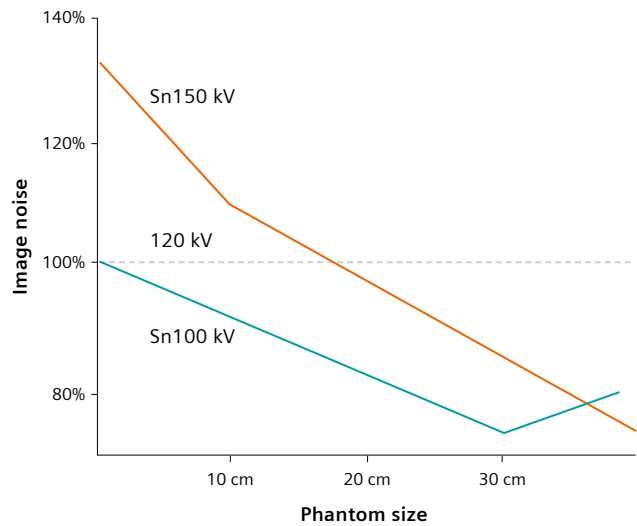


Fig. 7: Relative image noise (normalized to the image noise at 120 kV) as a function of the water-equivalent diameter of various phantoms, for the same radiation dose but different X-ray spectra: 120 kV, 100 kV + Tin Filter (Sn100 kV), 150 kV + Tin Filter (Sn 150 kV). Results from Siemens Healthineers in-house measurements.

In other words, using the Tin Filter at 100 kV would lead to beam hardening effects similar to the standard 120 kV spectra, but at a significantly reduced dose.

In the X-ray attenuation range of typical lung and colon examinations, corresponding to 20–35 cm water-equivalent diameter, the image noise is up to 30% lower compared with the standard 120 kV setting.⁹

Tin-filtered pre-filtration was first introduced with the SOMATOM Definition Flash and has since been implemented to reduce dose with standalone tin-filtered imaging for non-contrast enhanced CT imaging of high-contrast objects.

Gold Filter

Another consideration was to shape the spectrum in the opposite direction, reducing the highly attenuating photons and enhancing the low kV spectrum.

Gold (Au) possesses a K-edge at 81 keV, therefore the X-ray photons with an energy just above this K-edge will be strongly absorbed by pre-filtration with a gold filter as compared to the low-energy photons in the spectrum.¹¹

This filtration behavior differs from the previously mentioned filters like the Tin Filter. Here, the mean energy of the X-ray spectrum is shifted to lower energies while with the filtration materials such as tin (Sn), it is shifted to

higher energies. Because of these different characteristics, a combination of gold filtration and tin filtration across the X-ray beam allows Dual Energy (DE) spectral imaging based on a singular tube voltage.

This technology was introduced as part of the TwinBeam Dual Energy solution where it was configured as a movable split filter with one half Au (gold) and the other half Sn (tin) (see the Dual Energy clinical section for more information).

Each of these additional, switchable spectral shaping methods is used for selected, suitable situations in clinical practice, across the whole Siemens Healthineers CT portfolio.

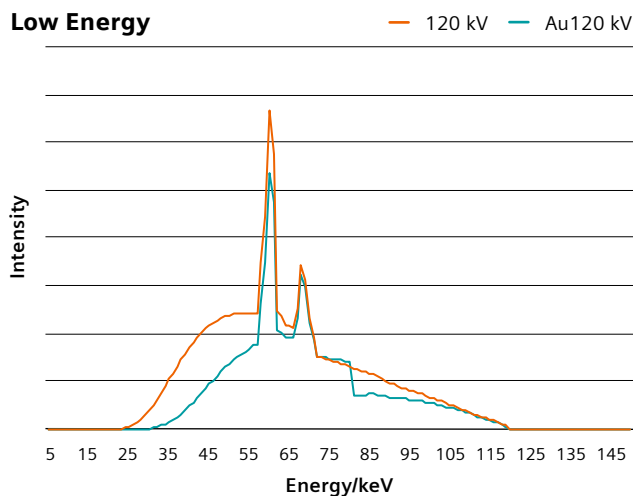


Fig. 8: 120 kV spectrum with additional gold (Au) pre-filtration, with reduced high-energy keV photons.

Clinical implementation

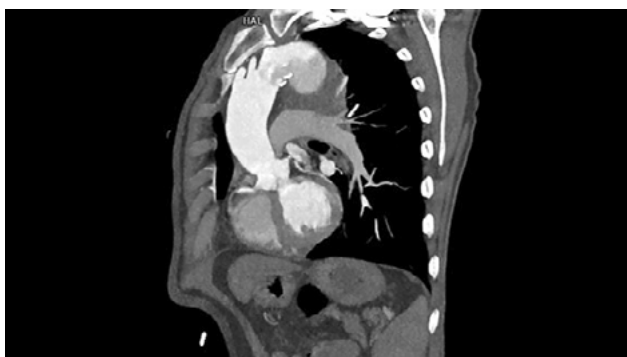
Each method of beam filtration inside our scanners is in routine clinical use as well as being used to investigate new methods to improve treatment outcomes for patients in many clinical fields.

Versatile filtration in general CT imaging

The combination of beam hardening and beam shaping forms a baseline of filtration and is consistent in all CT scanners. Filter thickness and design/shape varies with each X-ray tube type, because they are fitted across all CT scanners in the SOMATOM® portfolio.

The baseline of filtration, beam hardening and beam shaping, is used in every clinical use case by default. This allows CT imaging in routine examination across anatomical regions like the head, chest and abdomen. Routine imaging includes a huge variety of patients, from newborns and small children to adults weighing 300 pounds or more. This spectrum is partially covered by a wide range of tube voltages from 70 kV up to 150 kV.

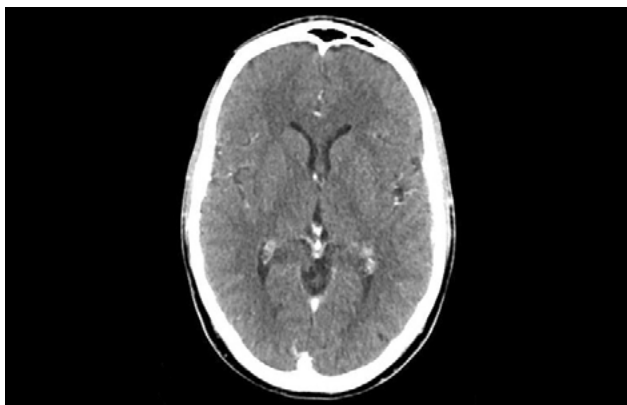
Imaging at low kV settings is a desired CT scanner capability for routine applications (Fig. 9) to save patient dose or enhance contrast, e.g., in scans with contrast media like angiographies.¹⁰ But low kV imaging (70 and 80 kV)



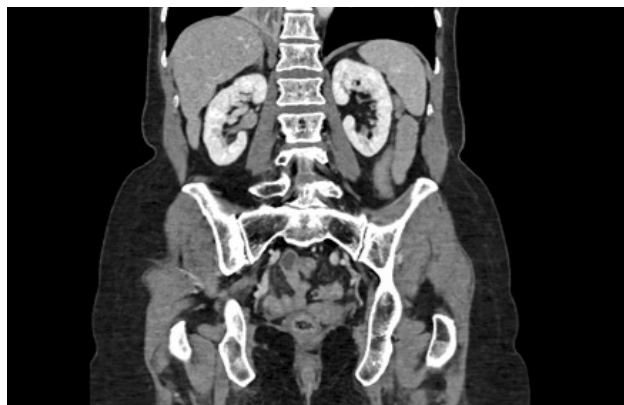
Courtesy of Erasmus MC, Rotterdam, The Netherlands



Courtesy of Amakusa Medical Center, Kumamoto, Japan



Courtesy of Fakultni Nemocnice Polzen/Plzen, Czech Republic



Courtesy of University Hospital of Erlangen, Erlangen, Germany

Fig. 9: High-quality images utilizing the basic filtration, beam hardening and beam shaping. (Top left: SOMATOM Edge Plus) Arterial thoracic section of a whole-body study using 70 kV. (Top right: SOMATOM Drive) Portal venous abdominal study performed at 70 kV. (Bottom left: SOMATOM Drive) Brain studies performed at 90 kV. (Bottom right: SOMATOM go. Top) Abdomino-pelvic case performed at only 70 kV despite large body habitus.

is particularly challenging in technically because it demands a high tube current at low kV. The standard filters are designed to support this essential imaging technique in all recent SOMATOM CT scanners.

This basic but powerful filtration is complemented further by what is termed the Adaptive Dose Shield: additional movable collimators that stop the X-ray beam completely and allow for dynamic collimation. These are used clinically

to stop over-beaming of helical scans so that extra dose is not to the patient when it will not enable image creation.¹²

Depending on the patient size or region of interest, a movable beam-shaping filter that differs in shape may be used instead of or in addition to the default filter. Examples include cardiac and pediatric imaging, mostly combined with the previously highlighted low kV imaging at 70 kV (Fig. 10).



Fig. 10: Left: Cinematic rendering of a cardiac scan performed at 70 kV in one heart beat.
Right: Cinematic rendering of an arterial phase thorax in a small pediatric patient performed at 70 kV in a fraction of a second.

Versatile filtration in non-contrast imaging

Beam hardening and beam shaping filters, the baseline of filtration, cover a wide range of applications. The necessary compromise associated with these filters calls for further optimization for dedicated clinical use cases. In high-contrast CT examinations not based on an iodinated contrast agent. In those cases, spectral shaping can additionally take away many quanta of low energy that only contribute to the patient's radiation burden.¹³

Consequently, scanners without a selectable spectral shaping filter have degraded dose efficiency potential for non-enhanced CT scans and require a higher radiation dose. Furthermore, beam-hardening artifacts are significantly reduced using spectral shaping.¹⁰

Lung with Tin Filter

Gordic et al.¹⁴ evaluated the image quality and sensitivity of ultra-low radiation dose single energy CT with tin (Sn) filtration for spectral shaping, and the iterative reconstruction for the detection of pulmonary nodules in a phantom setting. It was concluded that the sensitivity was higher, and the number of false positives lower with the Tin Filter.

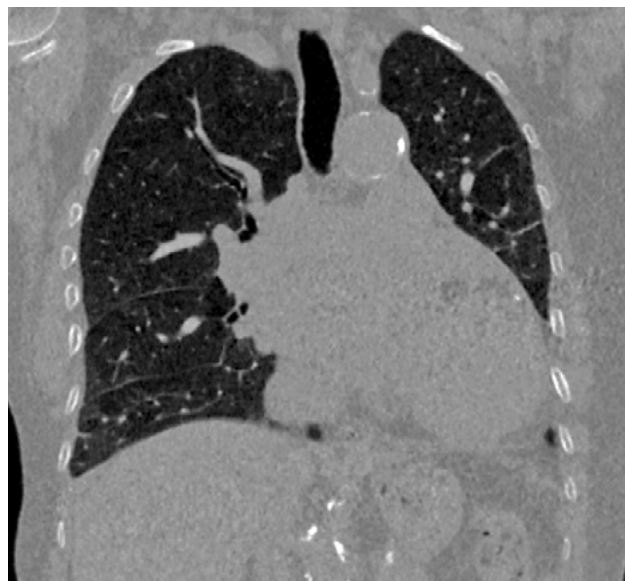
According to Gordic et al., the patient images acquired at Sn100 kV were of diagnostic quality both at 1/10th of the standard dose (corresponding to an effective patient dose of about 0.13 mSv) and at 1/20th of the standard dose (corresponding to an effective patient dose of about 0.06–0.07 mSv).

The authors conclude that “chest CT for the detection of pulmonary nodules can be performed producing high image quality, sensitivity, and diagnostic confidence at a very low effective radiation dose of 0.06 mSv when using a single-energy protocol at 100 kV with spectral shaping and when using advanced iterative reconstruction techniques”. Note that this is the dose level of a chest X-ray examination.

Additionally, a pediatric imaging study focused on evaluating the lung parenchyma compared low-dose CT at 70 kV with spectral beam shaping at 100 kV (Sn 100 kV) using a Tin Filter.¹⁵ Here it was concluded that the Tin Filter method significantly reduced radiation dose compared to low-dose CT.

Additionally, studies have been performed looking into the feasibility of detection of sub-solid nodules with ultra-low dose protocols. The study suggests that with a combination of tin filtration and ADMIRE, the mean volume CT dose index ($CTDI_{vol}$) of chest CT can be lowered considerably, whereas sensitivity for nodule detection remains high. For solid nodules $CTDI_{vol}$ was 0.10 mGy, whereas sub-solid nodules, required a slightly higher $CTDI_{vol}$ of 0.28 mGy.¹⁶

In summary, employing spectral shaping with a Tin Filter offers several advantages and is ideal for high-contrast protocols without contrast and for reducing beam hardening effects and radiation dose.⁹



Courtesy of University Hospital Erlangen, Erlangen, Germany

Fig. 11: With Tin Filter, a non-contrast thorax CT can be performed on patients at X-ray dose levels, with easy positioning and on some systems without breath-hold.

Urinary Stones with Tin Filter

Diagnosis of urinary calcification within the kidneys, ureters, or bladder was originally performed with plain film radiography, before CT began to be utilized for this purpose. However implementing this into routine protocol is sometimes hampered by the question of dose in CT.

Here spectral shaping with Tin Filter can help, with one publication assessing radiation dose during urinary stone detection in unenhanced CT. The authors compared tin-filtered 150 kV (Sn150 kV) and automated kV selection (110–140 kV) based on the scout view on the same CT-device.¹⁷ The authors found more than a 30 percent decrease in the $CTDI_{vol}$ with Tin Filter spectral shaping, while at the same time, they saw an improved image quality.

An additional study also concluded that using spectral shaping with tin filtration can substantially reduce radiation dose compared with routine standard- and low-dose abdominal CT for urinary stone disease.¹⁸



Courtesy of University Hospital Erlangen, Erlangen, Germany

Paranasal Sinuses with Tin Filter

A prospective study on CT imaging of the paranasal sinuses compared a protocol at 150 mAs, 100 kV and tin filtration (Sn100 kV) to a low-dose protocol at 50 mAs and 100 kV.^{19,20}

The results showed a statistically significantly lower $CTDI_{vol}$ in the study group (1.2 mGy vs. 4.4 mGy, $p < 0.001$) and a diagnostic image quality.



Courtesy of Erasmus MC, Rotterdam, The Netherlands

Fig. 12: (Left, top) Plain X-ray of an abdomen may not be sufficient to determine evidence or positioning of a urinary stone (left, bottom). Urinary stone seen in the left renal pelvis.

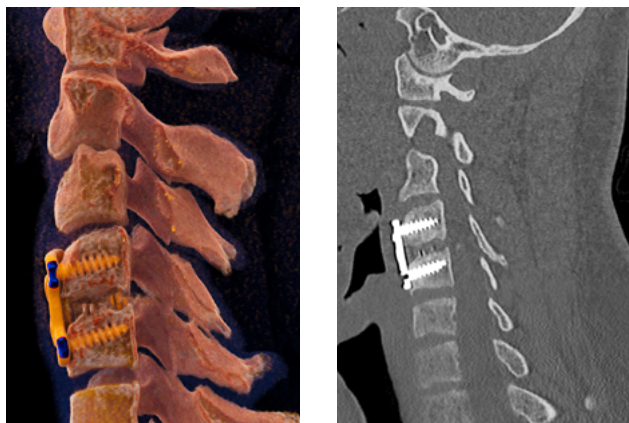
Fig. 13: (Right, top) Plain X-ray images are performed on patients at a similar dose and may require the patient to be placed in multiple positions, with repeats sometimes required. (Above, bottom) With Tin Filter, non-contrast Sinus CT can be performed on patients at X-ray dose levels, in one easy position.

Spine imaging with Tin Filter

Clinical evaluation of cervical, thoracic and lumbro-sacral spine regions is common when searching for underlying acute or chronic fractures or structural anatomical variants, be it in trauma or even oncological settings.

One study to investigate the image quality, radiation dose, and intermodality agreement of cervical spine CT using spectral shaping at 140 kV with a Tin Filter (Sn140 kV) in comparison with conventional CT at 120 kV.²¹

Here it was concluded that cervical spine CT using Sn140 kV improves image quality of the lower cervical region without increasing the radiation dose. Thus, this protocol can be helpful to overcome the artifacts in CT images of the lower cervical spine.

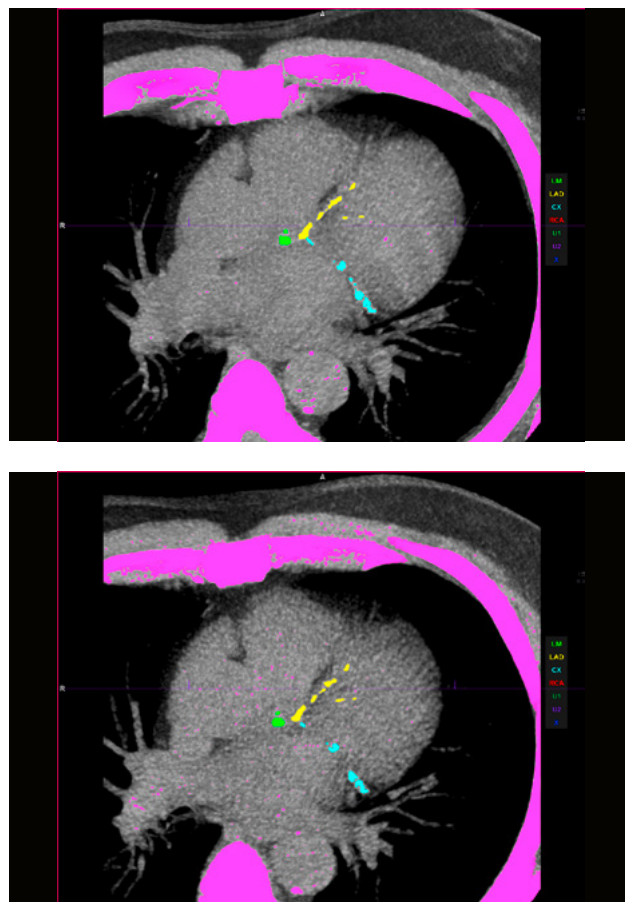


Courtesy of Hôpital privé d'Antony, Antony, France

Fig. 14: (Above) CT of the cervical spine is now the gold standard in emergency imaging. However now, with Tin Filter, the dose is at the level of the old X-ray imaging gold standard.

Calcium Scoring with Tin Filter

Utilizing the Agatston calcium scoring method limits the scan parameters to either 120 or 130 kV. Testing was performed to check if the spectral shaping method using the Tin Filter would correlate to this, because the mean photon intensity is similar. Based on the test results, it was concluded that a scoring method based on thresholds in HU like the Agatston score yields similar scoring results for Sn100 kV* images if compared to the established voltages of 120 kV/130 kV in this phantom setup.^{8, 22}

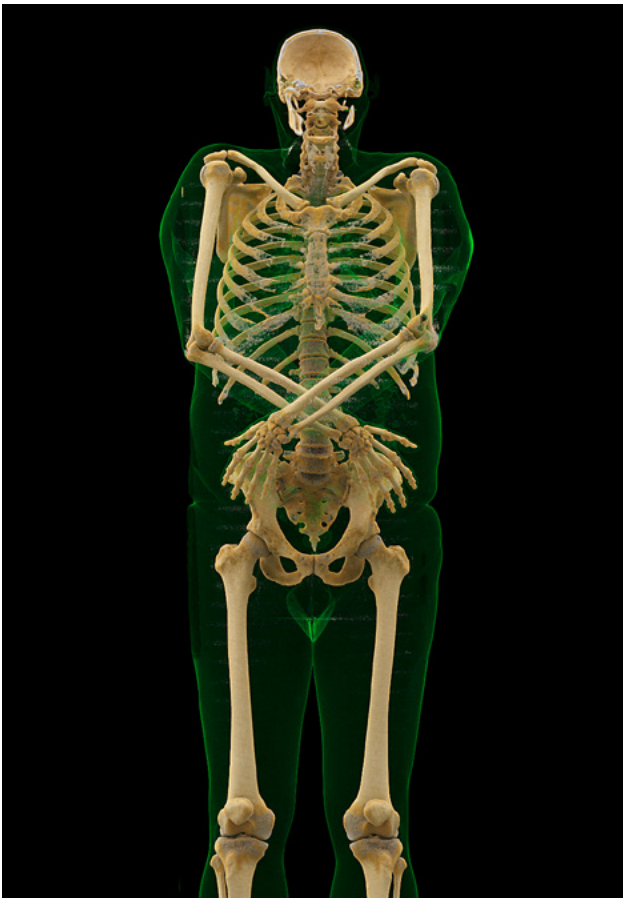


Courtesy of Medscan Barangaroo, Sydney, Australia

Fig. 15: Agatston score results at 120 kV (top) and Sn100 kV (bottom) spectra in the clinical environment showing equivalent scores.

Whole Body CT with Tin Filter

Studies to investigate the radiation dose and image quality of a whole-body low-dose CT using spectral shaping at 100 kV (Sn100 kV) for the assessment of osteolytic lesions in patients with multiple myeloma were performed²³ and concluded that the scans performed with spectral shaping can obtain sufficient image quality to depict of osteolytic lesions while reducing radiation dose by approximately 74%.



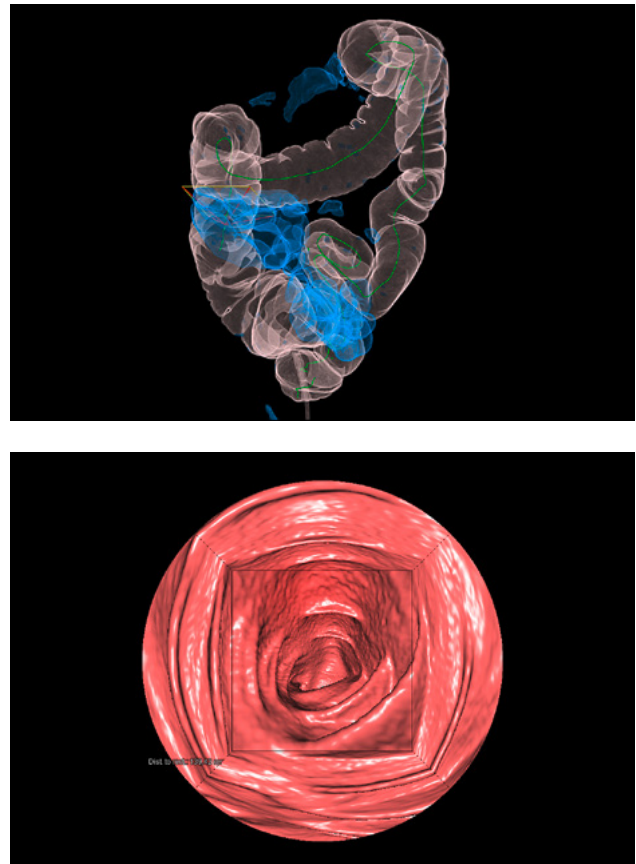
Courtesy of Erlangen University Hospital, Erlangen, Germany

Fig. 16: Full-body, skeletal survey performed in one simple low-dose CT scan.

Colonography with Tin Filter

CT colonography as a non-invasive procedure to detect polyps has been performed for many years, however with the addition of the Tin Filter, it is possible to detect polypoid lesions with very low dose radiation settings.²⁴

Additionally, retrospective studies comparing standard dose protocols versus Tin Filter-based protocols concluded that the Tin Filter enables consistent sub-mSv colonography without substantially impairing image quality.²⁵



Courtesy of ESR EuroSafe Imaging lounge ECR 2018

Fig. 17: (Top) Polyp lesion seen on a CT virtual colonography using Tin Filter (Sn100 kV / 96 mAs at 0.2 mSv) with similar detection potential as a non pre-filtered scan (bottom), despite having roughly 10% of the dose (120 kV / 50 mAs at 2.4 mSv).

Extremities with Tin Filter

Orthopedic examinations have generally been the realm of plain film radiography, however now with spectral shaping and the Tin Filter, it may be possible to move towards a CT scan for these procedures.

This would provide high quality 3D data of CT, with a clear reduction of 2D projection errors (overshadowing or positional), at the same dose as the plain X-ray series.



Courtesy of Medical University of Vienna, General Hospital AKH, Vienna, Austria



Courtesy of Erlangen University Hospital, Erlangen, Germany

Fig. 18: (Top) Hand CT performed with Tin Filter for ultra-low dose and more detailed information. (Bottom) Pelvis examination performed at X-ray dose level.

Topogram with Tin Filter

Given the significantly reduced dose of CT, especially with implementation of low-kV imaging and tin-filtered imaging, the dose of the routine topogram or planning scan may make up a significant portion of the overall CT dose.

Therefore, it is also possible to implement the Tin Filter for the topogram scan²⁶, enabling a significant dose reduction while also maintaining the dose selection parameters of a routine kV topogram.

Intervention with Tin Filter

Invasive interventional procedures in CT are becoming more commonplace. Scanning of the patient to monitor the needle position is important in these cases, however there may be the potential for increased dose when the procedure or patients anatomy is complex. Utilizing the Tin Filter, the overall dose of interventional procedures may be reduced, because each individual scan uses the dose reduction potential.

Versatile Filtration in Dual Energy

As mentioned previously, adding **spectral shaping** with Tin Filter technology to routine beam hardening and beam shaping was first considered for Dual Energy (DE).

Scanning at two different energies provides the required information for material differentiation (Fig. 2) because of the variation in the energy dependence of X-ray absorption for different materials.

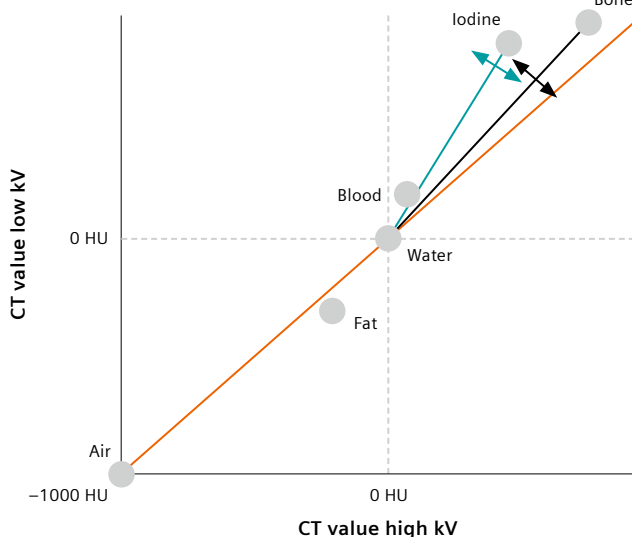
Although the X-ray spectra in CT are not monoenergetic, the average energy of the spectra is different for different tube voltages. Clinical benefits may result from the material characterization, going beyond simple morphology. For example, in conventional single-energy CT, the CT value of a bony structure and a vessel with iodinated contrast agent may be the same; however, Dual Energy CT (DECT) allows differentiating bone from iodine, which can be used for bone removal to provide a clear view of vessels in CTA studies.

Although all aspects of the CT scanner (e.g., spatial and temporal resolution) are of importance in high quality DE imaging, the sections most important are spectral separation and dose efficiency. And here spectral shaping plays a crucial role. References [27]–[29] discuss the different DECT approaches in this regard. Alternatively, also refer to the Dual Energy White Paper³⁰ for a more detailed explanation.

Spectral Separation

For reliable and quantitative material decomposition with DE, spectral separation plays a key role, as illustrated in Fig. 19. Spectral separation can be visualized by the overlap of the detected low- and high-energy X-ray spectrum (Fig. 20), with better spectral separation meaning less overlap.

Bad spectral separation



Good spectral separation

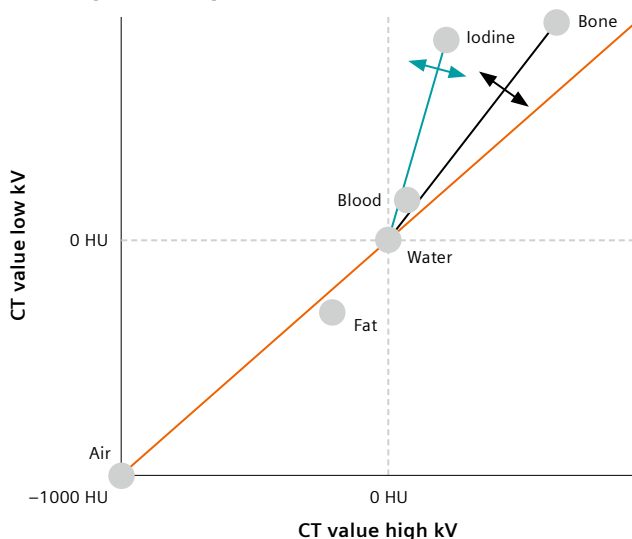


Fig. 19: Spectral separation as defined by the material characteristic slope (e.g., iodine ratio) in the Dual Energy diagram. The arrows orthogonal to the iodine (petrol) and bone vector (black) illustrate a certain noise level defined by the dose used for the image acquisition. At a higher dose, this spread will become smaller. With a bad spectral separation (top), this noise leads to misclassification of certain voxels, i.e., crosstalk between the two materials. This situation can be improved by using a higher dose. With a good spectral separation (bottom) a higher noise level can be tolerated without significant crosstalk.

Dose Efficiency

When performing Dual Energy, the level of overlap of the two energies, which is directly related to the spectral separation previously mentioned, has an effect on dose efficiency. With less separation comes more dose, because there is an inefficiency in the detected photons.³¹

For Dual Energy applications, the essential noise balancing between low and high kV images is enhanced by spectral shaping since less noise allows for less dose on high kV images, which allows for more dose and less noise on the noise-critical low kV images. The higher dose efficiency enables better image quality at the same dose level.

Implementation of Spectral Shaping

One way to separate the spectra and increase dose efficiency as mentioned above is to use spectral shaping on one of the two energies. Because Tin Filter removes the low keV portion of the spectra, it is perfect for this process. Combining a low kV spectrum with a tin-filtered high kV spectrum provides the widest possible spectral separation available in routine Dual Energy (Fig. 20, middle), and as the separation is almost complete, the noise profile of the two spectra provides a highly efficient dose profile.

An additional method to achieve dose efficiency as well as separate out the X-ray spectra is to incorporate the Tin Filter and a gold filter combined, to split, and then shape, the X-ray beam. This allows for improved spectral separation. The introduction of this technology, called TwinBeam, is further explained in the TwinBeam Dual Energy White Paper.³²

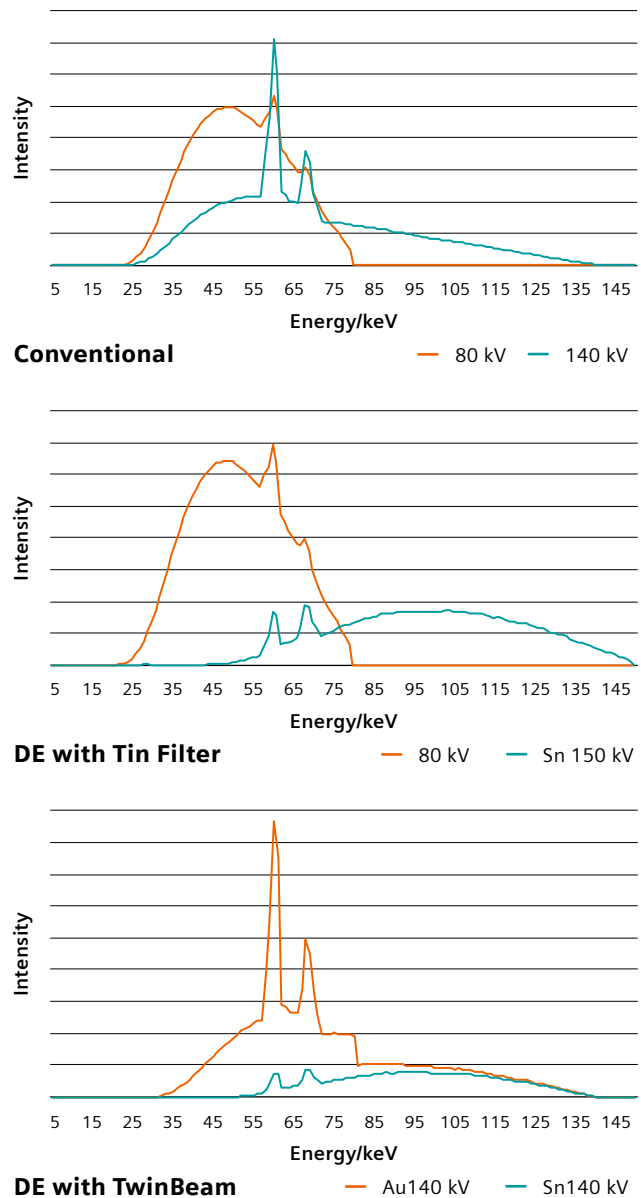


Fig. 20: Conventional Dual Energy without spectral shaping will have varying forms of this conventional spectral overlap, as demonstrated on a Dual Source CT with the Tin Filter turned off (top). The spectra of the high and low energies are highly separated with the introduction of the spectral shaping Tin Filter, helping to achieve a wide spectral separation (middle). With TwinBeam the spectra of the high and low energies are significantly separated with the introduction of the spectral shaping Tin Filter and the gold filter (bottom).

Clinical applications with spectral shaping filters

More and more radiologists are relying on the rich diagnostic possibilities offered by Dual Energy, especially where the spectrum is widely separated, because this can add to quantitative capabilities of the technology dramatically, as discussed, while keeping doses neutral to single-energy scanning.^{32,33,34,35}

Routine angiography or other post-contrast scans

Dual Energy allows areas of contrast enhancement to be accentuated, which can be useful when limited contrast media is available or when the injection has been suboptimal owing to complications, i.e., extravasation.

Furthermore, you can utilize the monoenergetic imaging functionality, which will allow both a shift in calculated single energy levels (keV) to improve iodine enhancement³⁶, to easily compare and quantify lesions and tissues, or to reduce metal artifacts.

Chemical characterization of different materials

The attenuation of X-rays, depends on the electron density and the effective atomic number, and both parameters are characteristic for different materials. With spectral shaping, the electron density and effective Z maps can be accurately calculated, which may provide key insights for radiation therapy planning and is also being researched for other pathologies.

Accurate and non-invasive diagnosis of gout

Gout detection facilitates the visualization of deposited uric acid crystals in peripheral extremities by automatically color-coding these crystals to visualize the deposits.³⁵

Advance the treatment outcome for patients with kidney stones

With the identification and characterization of different kinds of kidney stones³⁷, this approach utilizes the DECT data in order to calculate a visual color-coded material decomposition into uric acid and other stone types, which may provide insights into the best treatment pathway for each patient.

Lesion enhancement quantification in one image

With DE, it is possible to perform a contrast scan and to view a virtual non-contrast image retrospectively.³⁸ Not only does this provide an understanding of contrast enhancement between the two phases, but it also quantifies the iodine uptake [mg/mL] in tissue and lesions. The iodine uptake may correlate with the malignancy of a lesion.

Assessment of perfusion defects and affected vessels to provide treatment outcome potential

For patients with pulmonary emboli or other lung defects, it is important to try to highlight not only the structural pathologies, but also to provide insights into their downward affect and potential treatment planning. With spectrally separated, dose-efficient lung imaging, it is possible to highlight pathological vessels with a significantly lower iodine concentration than nonaffected vessels and to also enable fast evaluation of the related lung perfusion defects without the use of an additional non-contrast scan.³⁹

Further vessel analysis can be performed with other DE functionalities to accurately highlight vessel structures⁴⁰ or to differentiate hard plaques⁴¹ from contrast agent. This may provide greater pre-interventional insights, improve the overall planning for these procedures, and lead to a faster treatment and improved clinical outcome. DE applications may also then provide additional information after an intracranial intervention, by allowing for visualization of iodine concentration and distribution in the brain. Lesions and bleeds may show significant iodine uptake in the image, whereas inactive hemorrhages are not enhanced.⁴²

Some systems may also provide the unique potential to visualize and quantify the iodine concentration in the myocardium to reveal perfusion defects.⁴³

Orthopedic imaging of acute bone injury

Orthopedic imaging of acute bone injury bone bruises after trauma and diffuse tumor infiltrations have been predominantly seen using MRI, however with the benefits of strong spectral separation with versatile filtration, Dual Energy CT imaging can now also aid in diagnosis.⁴⁴ These are some but not all applications covered with Siemens Healthineers Dual Energy solution. For further information on Dual Energy, please refer to the dedicated white paper on Dual Energy.³⁰

Full potential of versatile filtration is still unknown

Beyond the functionalities discussed in this section about clinical implementation, there are other regions of the body that may also benefit from the pre-filtration package available with all recent SOMATOM CT scanners.

In particular, the full potential of spectral shaping is still unknown and awaits stronger external clinical proof beyond the testing performed by Siemens Healthineers. Low-dose brain imaging and low-dose facial bone imaging in trauma are just two examples.

Conclusion

Filtration is no longer a topic in just beam hardening or beam shaping. The advancement in additional spectral shaping techniques, such as Tin Filter or the Tin and Gold split filter, shift the conversation from mere routine imaging to high-quantification Dual Energy or the potential of plain X-ray dose levels for CT scans.

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