GE Healthcare

Technology White Paper

GSI Xtream on Revolution™ CT

Volume. Spectral. Simplified.

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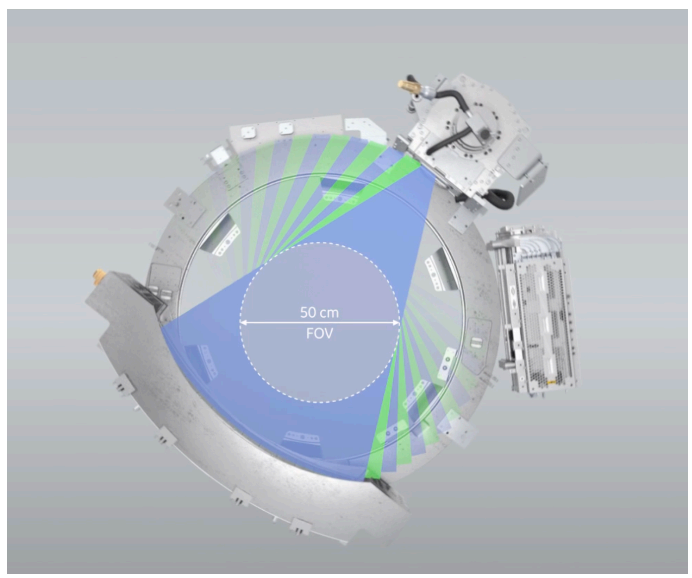

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Introduction

Advent of dual energy CT

Conventional single-energy CT (SECT) uses a polychromatic X-ray source with a fixed peak energy (defined by kVp) to generate images based on linear X-ray attenuation quantified in Hounsfield units. SECT provides useful structural information, but very limited material-specific information. For example, materials like calcium and iodine-water mixture can exhibit very similar or equivalent Hounsfield units in the reconstructed images and cannot be reliably distinguished in SECT. [1-3]

Dual energy CT (DECT) acquires projection data at two different energy spectra, which allows for more accurate material characterization based on energy-dependent attenuation profiles of specific materials. (Fig 1)

DECT was first conceived in the late 1970s, but did not achieve widespread clinical use due to inherent technology challenges such as motion-related mis-registration [4] (a consequence of acquiring two sequential images with long acquisition times), marked image noise at low energy settings [5] and excessive radiation exposure [6].

Figure 1

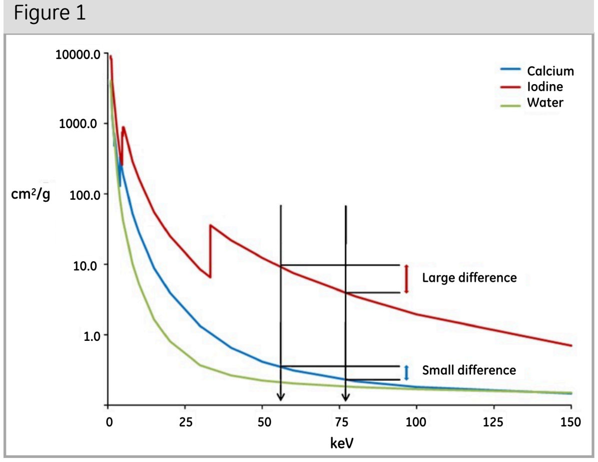


Figure 1: Attenuation curves of calcium, iodine and water. Attenuation difference at two X-ray energies allows for accurate differentiation of calcium and iodine.

Gemstone Spectral Imaging

To address these technical challenges, Gemstone Spectral Imaging (GSI) was introduced by GE on Discovery™ CT750 HD in 2010.

GSI employs fast kV switching during dual energy data acquisition: the x-ray source switches rapidly between 80 kVp and 140 kVp at sub-millisecond speed, producing nearly simultaneous dual energy projections at the same orientation to minimize the impacts of patient and organ motion.

Through innovative projection domain material decomposition processing, GSI overcomes the inherent physics related artifacts such as beam hardening and provides more accurate material-specific attenuation characteristics.

Since 2010, academic research and clinical exploration of GSI have grown rapidly and resulted in rich scientific publications and nice clinical adoption. GSI has shown benefits in:

- Better lesion characterization by providing information about the chemical composition and material characteristics
- Improving lesion detection with enhanced contrast-to-noise ratio
- Reducing beam hardening and metal artifacts
- Optimizing iodine load in contrast enhanced CT studies.

Design of GSI Xtream

To further integrate GSI into daily routine CT practice, GSI Xtream has been developed on Revolution CT.

This white paper shares in-depth insights on how GSI Xtream is designed and engineered to be a volume spectral CT technology with a simplified workflow.

Volume

Hardware Design

Revolution CT is a breakthrough platform that delivers uncompromised image quality and clinical capabilities through the convergence of coverage, spatial resolution, temporal resolution, CT spectral imaging, and dose performance – all in one.

Every component in the imaging chain — from tube and generator, to detector, to iterative reconstruction — has been designed to work seamlessly together to deliver GSI Xtream. (Fig 2)

As a result, Revolution CT with GSI Xtream is the industry leading volume spectral CT scanner designed to scan patients of different sizes, and produce high quality and dose neutral GSI exams with high throughput.

GSI Xtream's hardware innovations include:

- **Ultra-fast kV Switching** with near simultaneous temporal and spatial registration, better energy separation, and 50 cm spectral field of view.
- **Gemstone™ Clarity Detector** with 80 mm GSI collimation as well as CT number and material quantification consistency across the entire volume.
- **GSI HyperDrive** to deliver maximum GSI volume scan speed of 245 mm/sec without compromising the full 50 cm FOV.
- **Dose neutrality** to achieve radiation dose comparable to single energy CT across small, medium and large patients*.

** Demonstrated in phantom testing using small, medium, and large objects. Noise is defined as the standard deviation of the measured signal.*

Figure 2

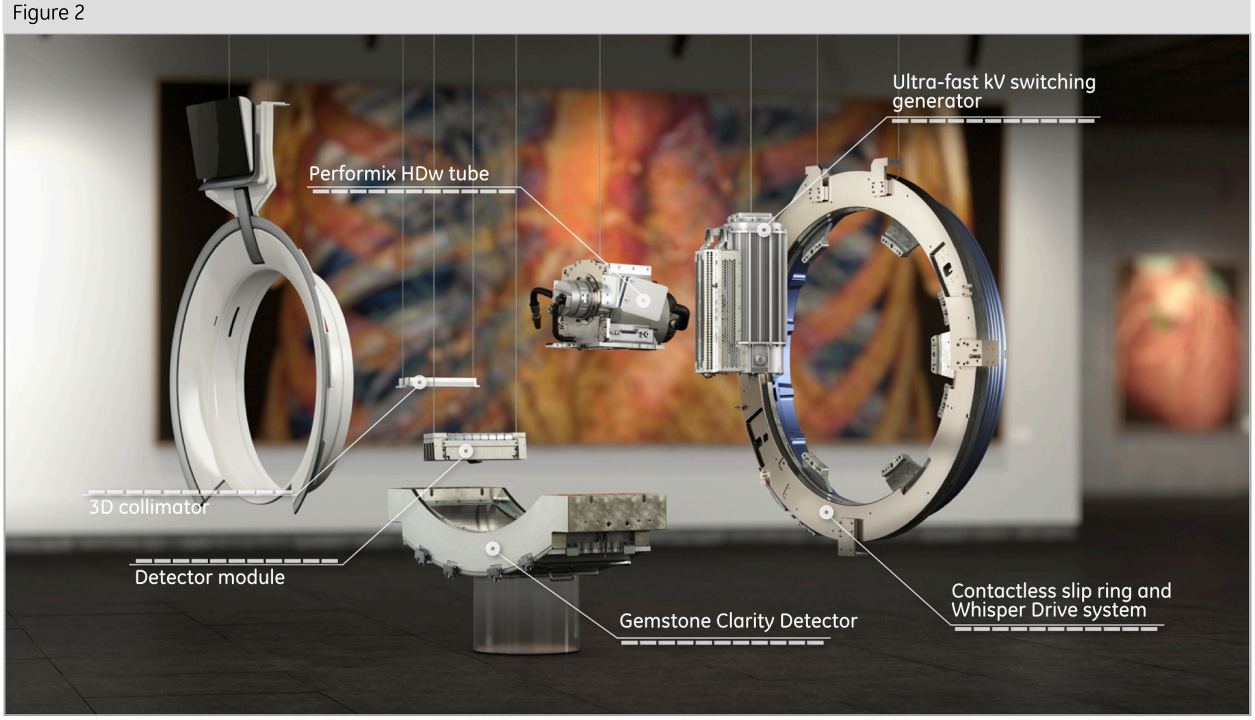


Figure 2: Illustration of hardware components of GSI Xtream on Revolution CT.

Ultra-fast kV Switching

GSI Xtream can switch the beam energy between the low setting (80 kVp) and the high setting (140 kVp) within microseconds to achieve the 0.25ms cycle time (Fig 3). This capability has been achieved by a patented high frequency generator and Performix™ HDw tube with a novel generator control hardware and firmware, a low capacitance interface, and microsecond level tube electrostatic focal spot size and position control.

Simultaneous Temporal and Spatial Registration

Ultra-fast kV switching can register two energy data sets at least 125 times faster than technologies which use multiple tubes and detectors on the same gantry at 0.25 sec rotation speed.

The nearly simultaneous temporal and spatial registration of the two energy datasets can:

- Reduce mis-registration caused by motion in dual-energy CT
- Improve characterization of contrast enhanced studies via low keV imaging
- Enable material decomposition in the projection domain, leading to more efficient beam-hardening correction and more accurate material density measurements [7].

Better Energy Separation

Compared to Discovery CT750 HD, Revolution CT has improved generator hardware enabling faster kV rise and fall times, which results in 20% higher energy separation between the low and high energies. Increased spectral separation results in more accurate material discrimination [8].

50 cm Spectral Field of View

Since ultra-fast kV switching uses a single tube and detector, it has the capability to acquire dual energy data and generate GSI images over the entire 50 cm field of view.

Compared with other dual energy CT technologies which have FOV constraints, GSI Xtream's 50cm FOV has benefits of:

- Scanning larger patients and ensures all relevant anatomy fit within the FOV
- Scanning patients with difficulties bending or moving limbs to fit within the center of the FOV

Figure 4 shows an example of scanning a large patient with GSI Xtream. A liver lesion (arrow), could not be fully included by the limited 33 cm or 35 cm FOV in other DECT technologies, was completely covered by the 50 cm FOV of GSI Xtream.

Therefore, the 50 cm SFOV advantage makes it easier to apply GSI Xtream for larger patients and challenging patients in ER trauma settings.

Figure 3

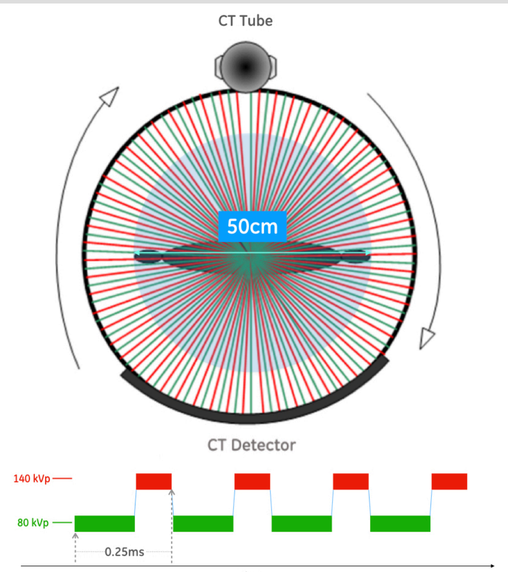


Figure 3: Illustration of ultra-fast kV switching

Figure 4

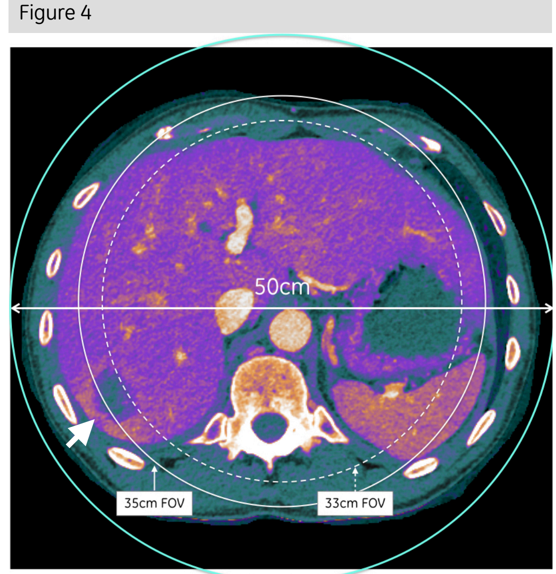


Figure 4: 70 keV with iodine color overlay of liver metastasis in a large patient. Only 50 cm FOV can cover all pathology and anatomy.

Gemstone Clarity Detector

Gemstone Clarity detector is another fundamental hardware component of GSI Xtream (Fig 5). It enables the industry first volume spectral imaging with 245 mm/s GSI volume scan speed.

Gemstone Scintillator

The Gemstone scintillator material is an isotropic ceramic with a highly uniform and translucent cubic structure. (Fig 6)

This unique structure enables faster primary speed (light emission) and shorter after-glow, compared to conventional Gadolinium Oxysulfide–based scintillators (GOS):

- Primary speed is only 0.03 μs, 100 times faster than GOS and is the fastest scintillator in CT industry.
- At 40ms, the afterglow is 0.001%, only 25% of GOS.

As a result, Gemstone scintillator can capture and convert ultra-fast kV switching dual energy signals.

Clarity Data Acquisition System

Equipped with ultra-low capacitance photo diodes, Clarity Data Acquisition System (Clarity DAS) enables 25% reduction in electronic noise and better 80 kVp data quality. This is crucial to GSI image quality and material density measurement accuracy.

3D Collimator

One of the primary challenges of the wide-cone detector CT is the increased scatter. The scatter-to-primary ratio (SPR) for a 160 mm wide-cone CT scanner with a conventional one-dimensional (1D) post-patient collimator, is around 20% for a 35 cm water phantom.

Dual energy quantification accuracy and image quality can be degraded by scatter. Water density measurements using a 1D collimator can shift from 1,000.2 mg/ml for 5 mm collimation to 1,005.82 mg/ml for 160 mm collimation, relative to ground truth of 1,000 mg/ml. (Blue bars in Fig 7)

Figure 7

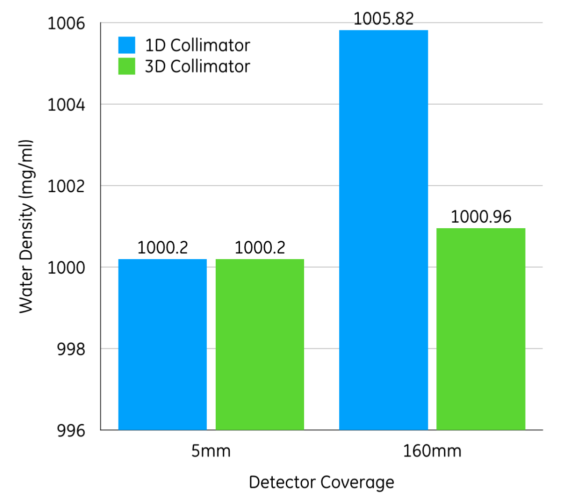


Figure 7: Measured Water Density Quantification. 3D collimator can improve quantification accuracy by reducing scatter.

Figure 5

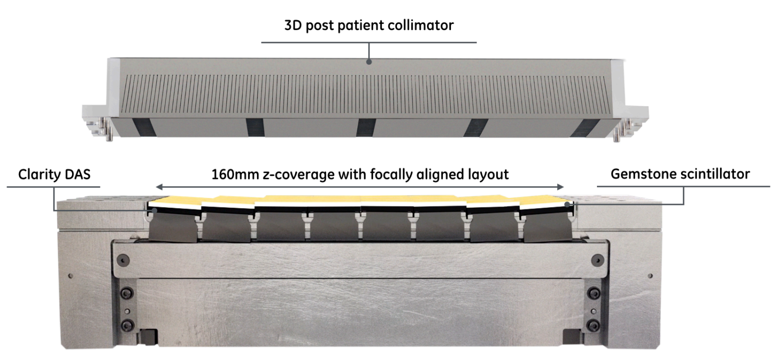


Figure 5: Illustration of Gemstone Clarity Detector Components.

Figure 6

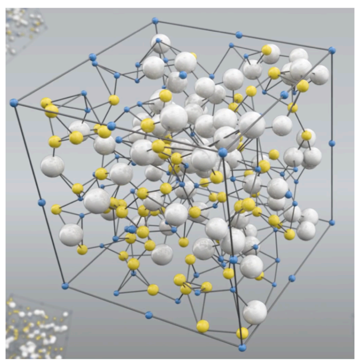


Figure 6: Illustration of material structure of Gemstone scintillator.

The Gemstone Clarity detector has a focally aligned detector layout and 3D collimator to use second-dimensional grids to reject scatter in the z-direction and angular plates in the third dimension to ensure focal alignment to the x-ray source. (Fig 8)

As shown in figure 9, compared to a 1D collimator, at every kV station, the 3D collimator achieves more than 50% decrease in the scatter-to-primary ratio (SPR) at iso-center, which is usually the most vulnerable region to scatter contamination.

Such a low level of SPR at 80 kVp and 140 kVp, along with the intelligent correction algorithm, helps ensure GSI quantification accuracy (Green bars in Fig 7).

Figure 9

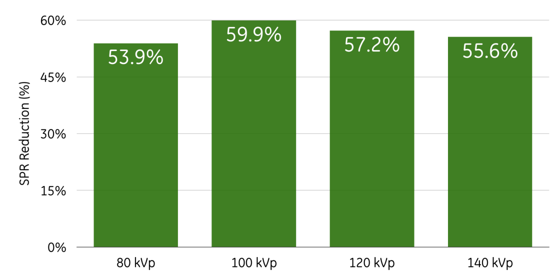


Figure 9: Measured SPR Reduction (3D collimator vs. 1D collimator) using 35 cm Water Phantom.

GSI HyperDrive

GSI Xtream features GSI HyperDrive: a maximum GSI volume scan speed of 245 mm/sec without the compromise of full 50 cm FOV.

This feature is achieved by 80mm GSI helical scan with 1.531 helical pitch and 0.5s/rot.

Because of the unique detector design, GSI CT number uniformity and material quantification consistency are maintained across the entire volume.

This unique capability expands the use of GSI to challenging patients who struggle to hold their breath or stay still in ER settings (Fig 10).

Figure 10

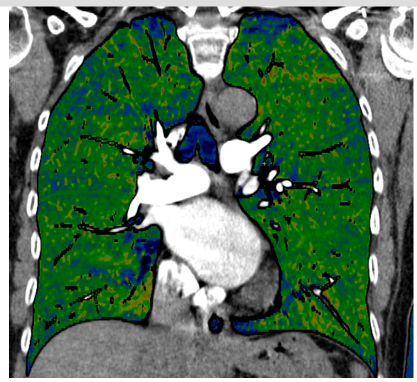


Figure 10: Coronal 70 keV with iodine color overlay in a patient with suspected pulmonary embolism. This patient had difficulty in holding breath. GSI HyperDrive scanned 311 mm length within 1.29 sec at 2 mSv.

Figure 8

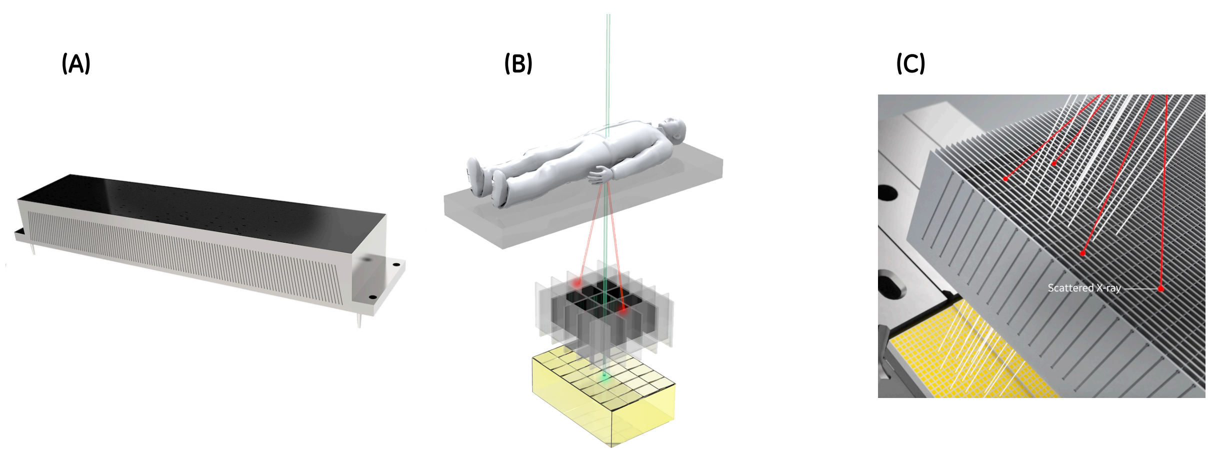


Figure 8: Illustration of internal structure of 3D Collimator. (A): 3D Collimator; (B): two dimensional grids to reject scatter in the z-direction; (C): angular plates in the third dimension to reduce scatter and ensure focal alignment to the x-ray source.

Dose Neutrality

Although two different energy spectra are obtained, GSI's radiation dose is still comparable to single energy CT due to more time assigned to 80 kVp and the use of iterative reconstruction (IR) technology.

Studies performed between 2012 and 2014 demonstrated abdominal GSI doses were between 12.8 and 21.8 mGy. These values are below the ACR's 25 mGy reference dose level for a single phase abdominal CT. The noise levels with GSI were comparable to those with SECT. [9-11]

In order to achieve dose neutrality across patients of different sizes, GSI Xtream integrates several hardware and software technologies, including

- Clarity DAS to reduce electronic noise and produce better 80 kVp data quality
- GSI HyperDrive to achieve faster scans at 245mm/s
- GSI Assist to personalize scan protocols based on anatomy and clinical indication
- ASiR-V*, next generation iterative reconstruction technology

Dose neutrality in different patient sizes was demonstrated by scanning three phantoms, representative of small, medium and large patient sizes, and generating images with comparable image quality between conventional and GSI scanning. (Fig 11)

The adult liver anthropomorphic phantoms (Quality Assurance in Radiology and Medicine, QRM), with and without the body fat ring, approximated large and medium patient sizes, respectively. The scans were acquired at 80 mm collimation, helical pitch of 0.992 and 1.0 sec/rot.

The QRM pediatric thorax phantom represented pediatric and small patients and the scans were acquired at 40 mm collimation, helical pitch of 0.984 and 0.5 sec/rot.

For conventional SECT imaging, 120 kVp was used for medium and large patient sizes, and 100 kVp was used for small patient size.

The CTDIvol for the scans were ~19 mGy (large), ~11 mGy (medium), and ~7.8mGy (small) with ≤ 5% difference between GSI and SECT scans. The SECT images were compared to GSI monochromatic images at 70 keV.

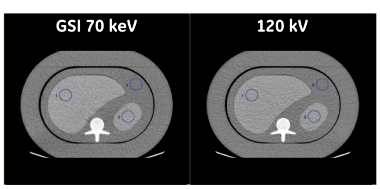
The test demonstrated that noise (defined as the standard deviation of the measured signal) in SECT and GSI 70 keV images was comparable across the three patient sizes.

Both phantom tests and early clinical evidence indicate that radiation doses from GSI Xtream are reliably comparable to those of single energy CT.

** In clinical practice, the use of ASiR-V may reduce CT patient dose depending on the clinical task, patient size, anatomical location and clinical practice. A consultation with a radiologist and a physicist should be made to determine the appropriate dose to obtain diagnostic image quality for the particular clinical task.*

Figure 11

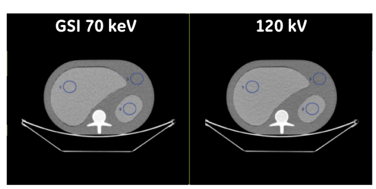
Large Object
(QRM Liver Phantom+body ring, 36cm in diameter)



80mm collimation, pitch 1, 1.0s/rot, ASiR-V 30%, CTDIvol: 19 mGy

Average 70 keV standard deviation	Average 120 kVp standard deviation
23.9 ± 0.3	26.5 ± 0.5

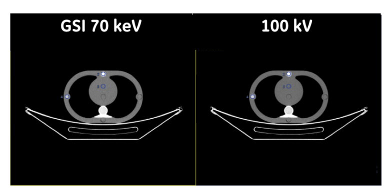
Medium Object
(QRM Liver Phantom, 26cm in diameter)



80mm collimation, pitch 1, 1.0s/rot, ASiR-V 30%, CTDI vol: 11 mGy

Average 70 keV standard deviation	Average 120 kVp standard deviation
15.2 ± 0.4	16.7 ± 0.4

Small Object
(QRM Pediatric Phantom, 16cm in diameter)



40mm collimation, pitch 1, 0.5s/rot, ASiR-V 30%, CTDI vol: 7.8 mGy

Average 70 keV standard deviation	Average 100 kVp standard deviation
9.4 ± 0.6	9.3 ± 1.7

Figure 11: Phantom tests on dose comparability between GSI Xtream scan and single energy scan. Three phantoms simulate small, medium and large patient sizes. Noise is defined as the standard deviation of the measured signal.

SPECTRAL

GSI Image Types

Because of ultra fast kV switching and the Gemstone Clarity detector, GSI Xtream is capable of generating all GSI images types in the projection or raw data domain.

The benefits of projection domain spectral image generation are:

- Better quantification accuracy due to inherent beam hardening reduction in projection domain
- Native reconstruction of all GSI images that can be directly transferred to PACS.

GSI Xtream can generate: monochromatic images, material density images (MD), virtual unenhanced images (VUE), and GSI Metal Artifact Reduction images (GSI MAR).

The GSI images generation flow and their clinical values are explained in this section.

GSI Xtream Image Generation Flow

Figure 12 illustrates the GSI Xtream data acquisition and image generation chain. Data is acquired by simultaneously interleaving prescribed low and high kVp projections, which are then split into low and high kVp projection sinograms.

Low and high kVp data sets are aligned in projection space, and further processed to reduce CT number shifts due to beam-hardening. The high kVp sinogram is reconstructed to produce a 140 kVp image for quality check (QC) purposes.

The interleaved low and high kVp projections are processed using advanced spectral modeling and fully physics-based projection space material decomposition (MD) approach. The heel effect of the x-ray tube and non-uniformity of spectral response on the x-ray detector are taken into account in the spectral and MD modeling.

Using known attenuation curves of the basis materials, water and iodine, the algorithm transforms the low and high kVp data into material density (MD) projection maps of the basis materials. The MD projections are reconstructed to generate the iodine and water images. Noise reduction is then applied to the iodine and water images. A linear combination of these basis material images is finally used to synthesize monochromatic images at any energy from 40 to 140 keV, material density images (i.e. Calcium, Hydroxyapatite (HAP), Fat, Uric Acid) and virtual unenhanced (VUE) images. Metal Artifact Reduction (MAR) algorithms can be applied to all GSI images to reduce metal artifacts.

Figure 12

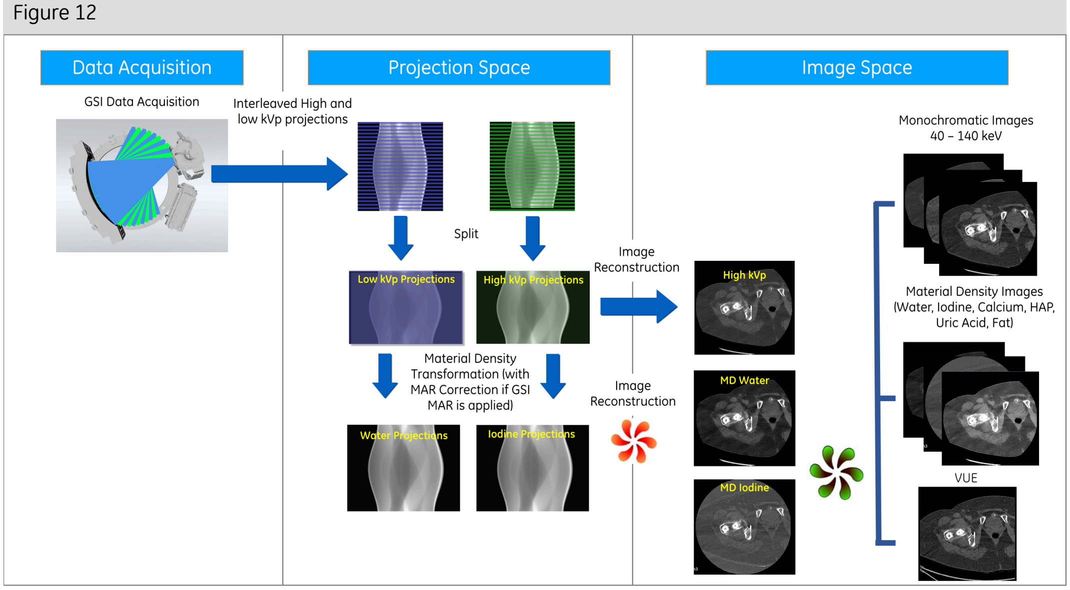


Figure 12: GSI Xtream's Data Acquisition and Image Generation on Revolution CT

Monochromatic images

Monochromatic images, ranging from 40 to 140 keV, depict objects as if they were imaged with a theoretical monochromatic beam whose x-ray energy is measured in kiloelectron volts (keV) instead of peak kilovoltage (kVp).

Monochromatic images with lower energy levels to improve contrast noise ratio (CNR)

Monochromatic images at lower energy levels can even achieve higher CNR than single energy imaging at the same dose. This is because monochromatic images with lower energy levels (40-70 keV) are closer to the k-edge of iodine (33.2 keV) where iodine exhibits much higher attenuation as compared to conventional single- energy CT at 120 kVp.

Due to advancements in hardware and algorithms on Revolution CT, GSI Monochromatic images at lower energy levels have lower noise and better image texture. These improvements have led to a monotonic increase in CNR as keV decreases across patient sizes and without any compromise to MD accuracy (Fig 13).

Clinical studies confirm that low energy monochromatic images (40-70 keV) can depict more subtle contrast enhancement by improving the CNR between a lesion and background parenchyma [13]. Monochromatic images provide more reliable attenuation values than conventional polychromatic CT images.

Monochromatic images can aid in detecting small lesions with insubstantial iodine uptake (Fig 14), evaluating small blood vessels (Fig 15) and depicting slow flows (e.g. subtle vascular endoleaks).

Monochromatic images also have the potential to reduce the amount of contrast material, which may benefit patients at risk for contrast induced nephropathy (CIN) [14]. Studies have shown 25% reduction of iodine contrast dose in abdominal CT [15], 50% iodine load reduction in CT pulmonary angiography [16] and 70% iodine volume reduction in CT aortography [17].

Monochromatic images with higher energy levels to reduce beam hardening

High energy monochromatic images (80-140 keV) can leverage additional information from measurements of the same object at two spectra to reduce beam hardening effects, especially streaking and shading artifacts caused by dense objects.

Figure 14

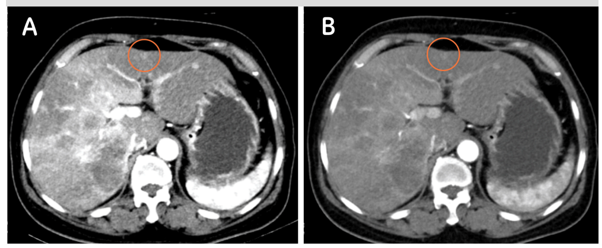


Figure 14: A patient with breast cancer metastatic to liver. Small liver lesion is better visualized in (A) 45 keV image than (B) 70 keV image.

Figure 15

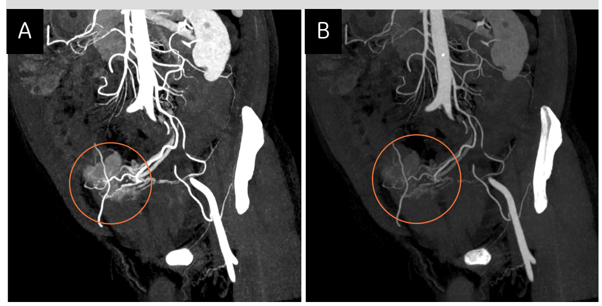


Figure 15: A patient with colon carcinoma. (A) 50 keV image to better visualize small details of feeding arteries of colon carcinoma mass, compared to (B) 70 keV image.

Figure 13

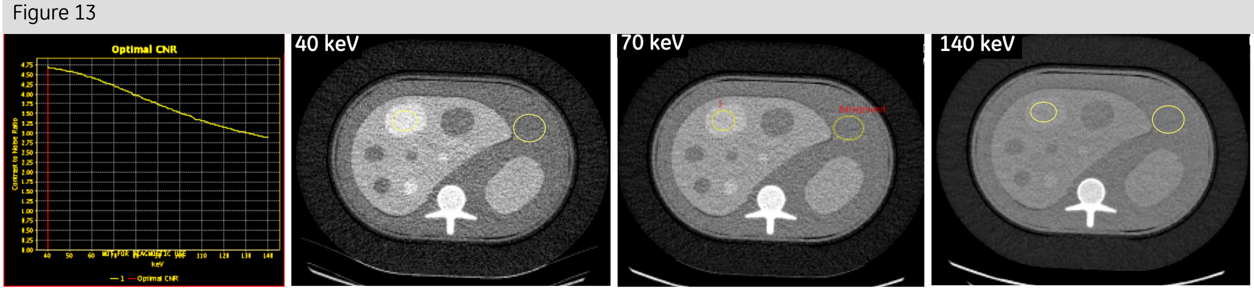


Figure 13: A 30 cm X 40 cm QRM phantom, simulating a large patient, was scanned with 80mm GSI helical at CTDIvol of 24 mGy. The contrast-to-noise ratio (CNR) of the liver ROI against the background was automatically computed using the GSI Viewer Optimal CNR tool on AW. The optimal CNR was achieved at 40 keV.

Material Density images (MD)

GSI Xtream can natively reconstruct Material Density images (MD) of iodine, water, calcium, hydroxyapatite (HAP), uric acid, and fat.

MD images provide qualitative and quantitative information regarding tissue composition and contrast media distribution [18].

Material basis pair and quantification

In GSI Xtream, MD images are generated as a material basis pair (e.g. water/iodine, uric acid/ calcium, etc). For example, when the water and iodine pair is generated, the iodine images illustrate the density of the object with suppressed water information (written as "iodine(water)") and water images illustrate the density of the object with suppressed iodine information (written as "water(iodine)"). Note that the measured pixel intensity values in these images are proportional to material density and are expressed in mg/ml.

GSI Xtream can detect iodine contrast in concentrations as low as 0.5 mg/ml in density at a dose as low as 8 mGy*.

Current and emerging clinical applications

As Sahani et al [18] summarized, material specific images generated by spectral CT expand the current role of CT and overcome several limitations of single energy CT.

Selecting the optimal material pairs for reconstruction is based on the clinical desire and investigation.

The most common material pair used clinically is iodine and water, because the iodine (water) image can be used to assess iodine distribution, to increase tissue contrast and to amplify subtle differences in attenuation between normal and abnormal tissues. This capability facilitates improved lesion detection and characterization, tumor viability quantification, and treatment response monitoring [18, 19]. (Fig 16)

In chest imaging, iodine (water) images, representing the iodine distribution in the lung parenchyma, can also help to identify pulmonary embolism–associated perfusion defects, especially in patients with underlying perfusion abnormalities [20–22] (Fig 17).

Fat (iodine) or fat (water) images can be used to characterize fat content in abdominal lesions, such as cholesterol stone in gallbladder (Fig 18) and a fat-containing renal lesion. Morgan et al [23] demonstrated that fat (iodine) images permit differentiation between low- and high-fat-containing adrenal lesions with high specificity of 94%.

Figure 16

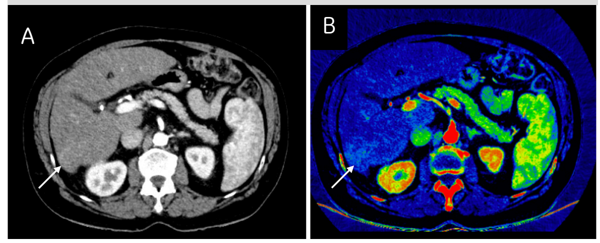


Figure 16: A patient with recurrent liver carcinoma. (B) Iodine color overlay on 55keV can better visualize recurrent cancer lesions (arrows) than (A) 55 keV image.

Figure 17

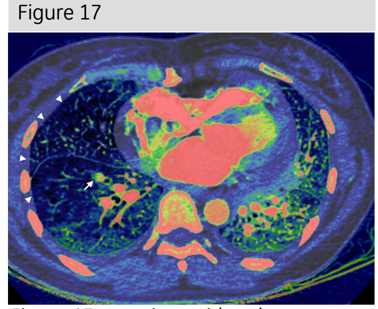


Figure 17: A patient with pulmonary embolus. Iodine color overlay on 70 keV aids in identification of embolus (arrow) and the area of lung perfusion abnormality (arrow heads).

Figure 18

Figure 18: A patient with gallbladder cholesterol stone. Fat (water) image characterizes the fat contents of the stone (arrows).

** Detection of 0.5 mg/mL at 8 mGy was demonstrated in head phantom testing.*

Virtual Unenhanced images (VUE)

GSI Xtream generates virtual unenhanced images (VUE) by subtracting iodine from images. The VUE algorithm is based on multi-material decomposition (MMD), which replaces the volume fraction of contrast by the same volume fraction of blood, producing iodine-suppressed images. The VUE images provide attenuation information in Hounsfield units.

Figure 19 shows examples of a non-contrast scan acquired at 120 kVp, and GSI contrast enhanced 70 keV and VUE images derived from a GSI contrast enhanced scan. The iodine filled vessels are removed in the VUE images. Regions of interest were drawn in the liver and aorta on the non-contrast, contrast, and VUE images on five slices and the average HU values were compared across the images. The HU values in the VUE images were similar to the HU values in the non-contrast images (Table 1).

VUE images can also provide reliable information for characterizing diverse lesions, including those with complex features. As shown in figure 20, in the 70keV image of a contrast enhanced scan, an enhancing region (arrow) is visible in the right ureter, and it is difficult to distinguish between contrast medium and a stone. VUE image shows suppression of the contrast medium and a persistent hyper-attenuating lesion (arrow) indicating a ureter stone.

Figure 19

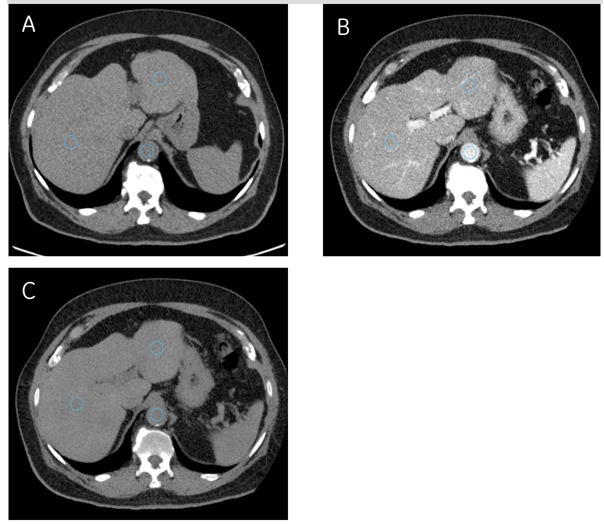


Figure 19: Examples of (A) a non-contrast image acquired at 120 kVp, and (B) GSI contrast enhanced 70 keV image, (C) VUE image derived from GSI contrast enhanced scan.

Table 1: Comparison of CT HU values.

	120 kVp Unenhanced	70keV Contrast enhanced	VUE
Liver 1	50.9 ± 5.8	112.1 ± 6.5	55.4 ± 2.6
Liver 2	59.3 ± 4.3	109.5 ± 11.8	60.7 ± 2.4
Aorta	38.1 ± 3.7	209.2 ± 7.2	37.3 ± 4.7

Figure 20

Figure 20: A patient with ureter stone (arrow). It is difficult to distinguish between contrast medium and a stone in (A) contrast enhanced 70keV image. (B) VUE image shows suppression of the contrast medium and a persistent hyper-attenuating lesion (arrow) which indicates a ureter stone.

GSI Metal Artifact Reduction (GSI MAR)

GSI Metal Artifact Reduction (GSI MAR) is a dual energy metal artifact reduction algorithm designed to reveal anatomic details obscured by metal artifacts. Patients imaged with CT routinely have metal implants which can cause artifacts due to photon starvation, beam hardening and scatter. GSI MAR can reduce metal artifacts using a three-stage correction to address all three factors. This approach generates metal corrected images, while preserving spatial resolution and data integrity near the metal [24-27]. The MAR framework is essentially called twice: once for high energy and once for lower energy, with some communication such as a shared metal mask from the high kVp image. The output sinograms are used as input to the improved spectral imaging chain to generate the final GSI MAR images. (Fig 21)

In the first stage of the MAR correction, the corrupted samples in the projection are identified. These samples correspond to the reading from the detectors that are impacted by the metallic object. Synthesized projections are generated using higher order interpolation and then back-projected to generate first stage of the MAR image.

In the second stage, a prior image is generated. This is done using an innovative signal processing technique to process the first stage MAR image before segmenting it to generate the tissue classified prior images. The tissue-classified image is then forward projected to generate the synthetic data, which is used to replace the corrupted projection samples using an advanced technique. The replacement step is critical and must be done carefully in order to avoid the introduction of other artifacts.

In the third stage, the final corrected projection data is generated using a combination of the original projection data and the in-painted sinogram generated in the second stage. The resultant projection data has the desired characteristics of both the real data (better low contrast resolution) and in-painted projection data (reduced artifacts).

This technique is effective in revealing anatomic details that are hidden beneath the artifacts near the metal (Fig 22).

Figure 22

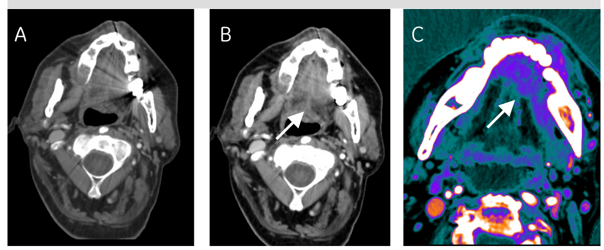


Figure 22: A patient with metal denture and mouth floor carcinoma. (A) 60 keV image shows the artifacts caused by metal denture. (B) 60 keV with GSI MAR and (C) iodine color overlay with GSI MAR can reduce metal artifacts and reveal enhanced mouth floor carcinoma lesions (arrows).

Figure 21

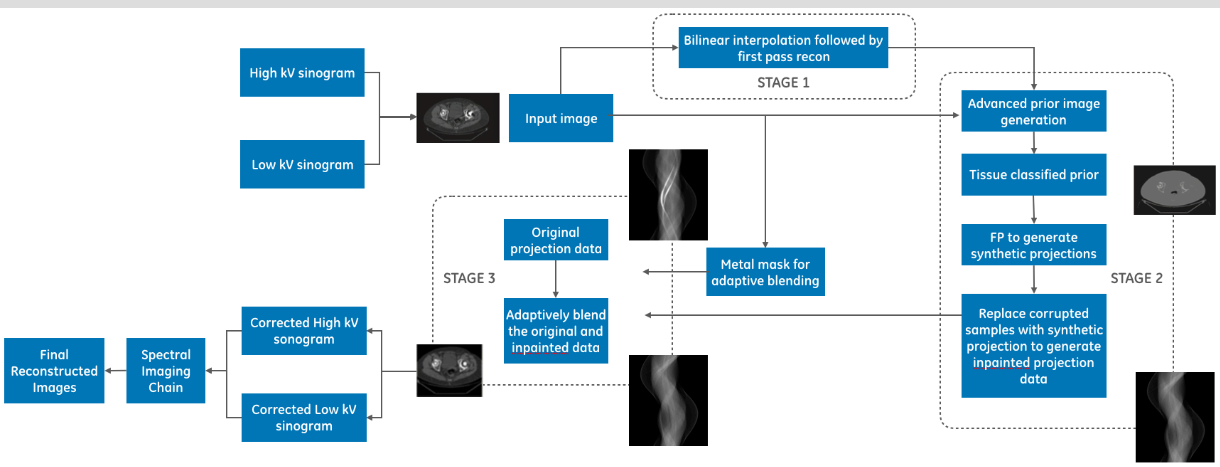


Figure 21: GSI MAR image generation flow.

Simplified

Workflow Design

With a simplified workflow at its core, GSI Xtream is designed to be a main-stream CT tool for daily routine use.

We drew from years of GSI experience to build an entirely new workflow architecture. We developed and incorporated innovative features that fully consider every aspect of workflow (Fig 23) :

- **GSI Profile:** available at a higher level in protocol management to standardize, automate and personalize GSI scan and recon parameters based on clinical needs, to achieve more with less run time decision making.
- **GSI Assist:** personalizes GSI scan parameters to clinical indications and the anatomy of the patient.
- **Xtream Recon:** reconstructs GSI specific images natively on console with real time reconstruction speed.
- **Direct transfer to PACS**: offers the direct transfer of all natively reconstructed GSI images to PACS.

Many of these features are enabled by default, so chief technologists or protocol managers do not need to test extensive configurations.

As a result, GSI Xtream is a major leap forward in workflow for dual energy CT.

GSI Profile

Due to patient-to-patient variation and different clinical applications, scan and reconstruction requirements for every CT scan will be different. Dual energy capability on CT scanners have introduced an additional layer of complexity and variability. As a result, imaging providers have had to create multiple scan protocols and rely upon highly skilled individual technologists to make vital scan time adjustments for each patient.

To address this challenge, GSI Profile was designed and introduced in GSI Xtream on Revolution CT to standardize and streamline scan/recon/data transfer.

Technologists can create GSI Profile, which consists of scan and reconstruction parameters, in the GSI Profile Editor. These parameters can be chosen to meet the image quality considerations and dose targets for a desired clinical task.

Using a GSI Profile within a protocol will automatically pre-populate all the desired parameters to ensure consistency across multiple similar protocols and reduce cognitive load at scan time.

The power of GSI Profile can be illustrated in the following example of a Chest Abdomen Pelvis exam. GE Reference protocol GSI CHST ABD PEL contains two groups: one covering Chest and the other covering Abdomen and Pelvis. GSI Profile GE ROUTINE ABDOMEN CONTRAST is mapped to this protocol.

Figure 23

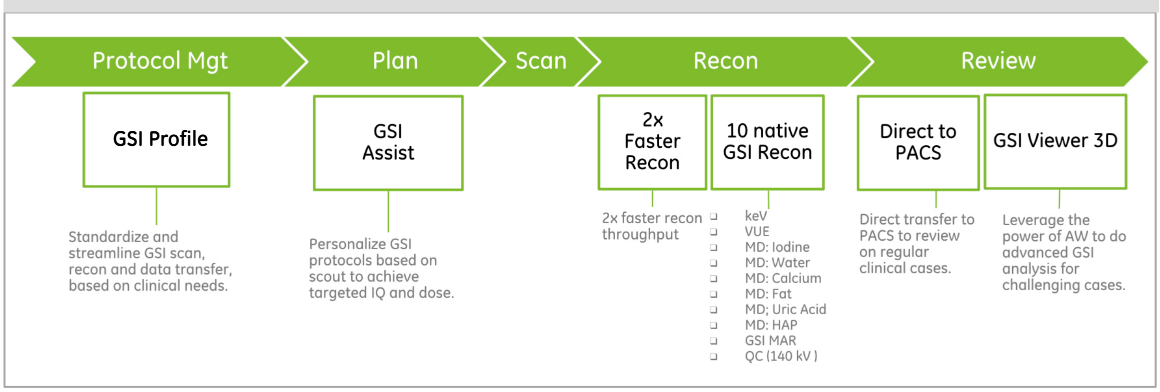


Figure 23: Overview of GSI Xtream workflow design.

For this clinical scenario, it was determined that the primary image quality considerations include general consistent image quality and ensure appropriate dose. Secondary image quality considerations include spatial resolution and MD accuracy. To achieve these image quality requirements, scan settings of all four rotation times (0.5, 0.6, 0.8, 1.0 sec/rot), 0.5 or 0.9 pitch factor, and 80 mm collimation were selected. Combining these scan options support a dose range of ~5.5 – ~45.5mGy. Since spatial resolution is a concern, no scan settings using XLARGE focal spot were included in this GSI Profile. If MD accuracy is a consideration for this clinical scenario, the clinician can easily change the scan parameters such that 0.8 or 1.0 sec/rot time are used. Recommended GSI Images to review for GE ROUTINE ABDOMEN CONTRAST include: monochromatic 70 keV (similar to 120kVp), monochromatic 50 keV (for better contrast and contrast-to-noise ratio of lesions to liver tissue), iodine(water) (lesion quantification and characterization), water(iodine) (differentiation between cyst and mass), and virtual unenhanced images (VUE) (to simulate a true non-contrast).

Thus, having a protocol mapped to a GSI Profile enables the clinicians to quickly tailor the scan acquisition protocol to meet the desired clinical intent without having to make last minute adjustments for each patient at scan time.

GSI Assist

The goal of GSI Assist is to enable the clinician efficiently select a patient-centered GSI scan technique for the clinical task at hand. By incorporating key workflow features of Automatic Exposure Control (AEC), GSI Assist provides the clinician with an intuitive, familiar and powerfully versatile path to achieve the desired GSI image quality at CTDIvol levels appropriate across the patient population.

The routine GSI Assist workflow is outlined in Figure 24. As an initial step, the user selects a GSI Profile to match the desired clinical task. Like the familiar AEC workflow, the GSI Assist user then specifies a desired level of image quality via the Noise Index. GSI Assist, after sizing the patient anatomy via the localizer scout and assuming a reference equivalent AEC technique, then suggests an optimal GSI scan technique chosen from the primary GSI Preset family in the GSI Profile. This primary GSI solution is derived by matching the required CTDIvol of the assumed reference AEC technique to the list of available CTDIvol levels contained in the primary GSI Preset family.

Similar to AEC, GSI Assist provides crucial feedback via the IQ Widget about the expected image quality for the scan. Illustrated in Figure 24, the Average Projected Noise Index (PNI) shown within the IQ Widget is an estimate of the average noise within

Figure 24

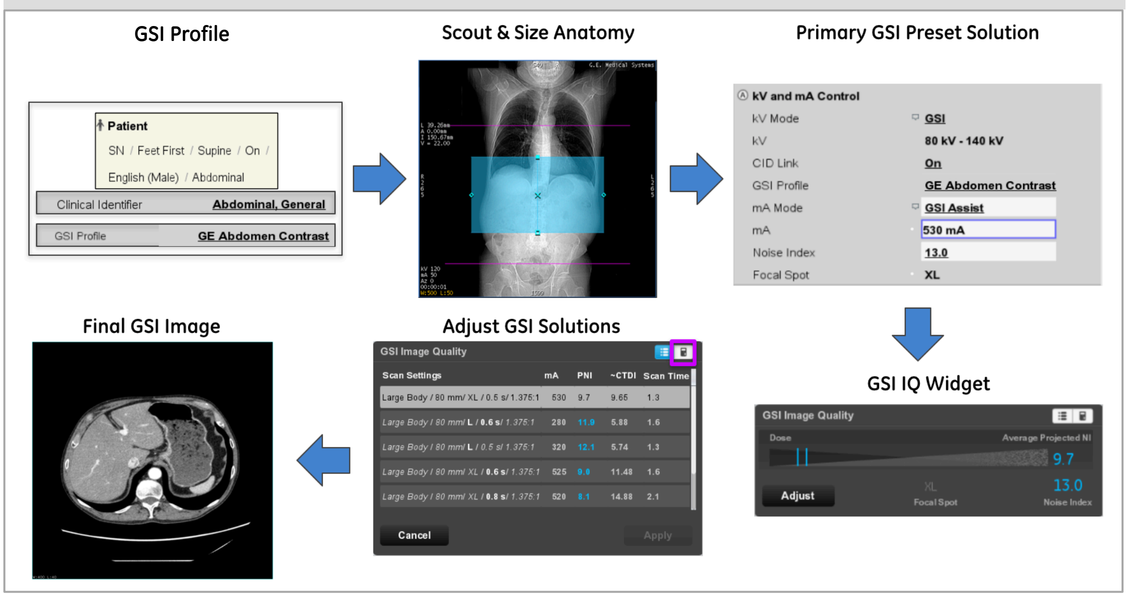


Figure 24: Outline of GSI Assist workflow.

the GSI image volume reconstructed at 70keV. In keeping with AEC, the PNI is colored orange if the average projected noise in the prescribed image volume is estimated to exceed the prescribed Noise Index.

Via the versatile Adjust menu, the user is provided with powerful options to achieve the desired clinical outcome in cases where the primary GSI solution does not achieve the desired tradeoff between dose and image quality. Specially enhanced for GSI Assist, the Adjust menu lists both the currently selected primary GSI preset solution and the most optimal GSI solution for each of the secondary GSI Preset families listed in the GSI Profile. This functionality menu can be valuable in many scenarios, for instance larger patients may benefit from secondary GSI solutions with higher available CTDIvol, while smaller patients may benefit from GSI solutions with shorter rotation times. For all options within the Adjust menu, changes in alternative GSI scan technique settings compared to the currently selected GSI technique are highlighted by bold text. As indicated on the Adjust menu, a change in scan technique elicits a change in the applied CTDIvol, Projected Noise Index and other scan technique factors.

Another key enhancement to the Adjust menu comes in the form of the Calculator option (highlighted on the Adjust Solutions in Figure 24). In cases where certain GSI solutions may induce tube cooling delays, accessing the Calculator gives the user the ability to select alternative solutions from the same GSI Preset family, without incurring tube cooling delays. Thus, the GSI Adjust solutions menu contains several flexible options that can be used to further tailor the suggested GSI preset solution for each individual patient until the desired tradeoff is achieved between dose, image quality and scan timing.

Xtream Recon

Xtream Recon technologies are designed to improve reconstruction speed and streamline patient throughput:

- **Xtream Recon Server**: Enables recon parallelism with two sets of powerful GPUs and CPUs and achieves two times faster baseline recon performance, as compared to previous generation GSI.
- **Smart Recon technology**: Streamlines GSI image generation process by creating an intermediate stage, from which multiple GSI DICOM images can be natively reconstructed (Fig 25). The reconstruction speed can be as fast as 45 images per second.

Early adopters' experiences confirm that GSI Xtream is capable of scanning over 50 oncology patients with multi-phase, contrast-enhanced GSI protocols, on a daily basis, without work flow delays.

Direct Transfer to PACS

The wide-spread adoption of GSI into clinical practice also needs the transfer of GSI DICOM images from console direct to PACS.

GSI Xtream workflow architecture can manage images types to be natively reconstructed and directly transferred to PACS based on the clinical needs at the protocol level.

Figure 25

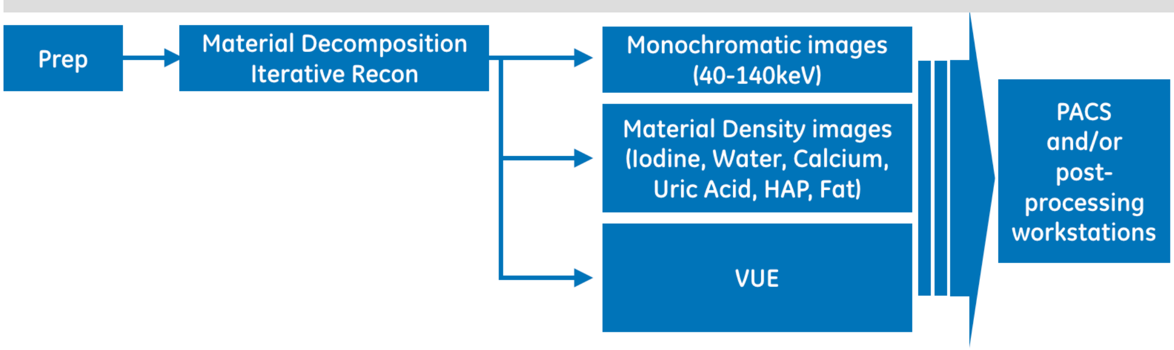


Figure 25: Illustration of Smart Recon technology and direct transfer to PACS. An intermediate stage is created after projection data preparation and material decomposition iterative recon process. From this juncture multiple GSI DICOM images can be natively reconstructed and directly transferred to PACS.

Conclusion

Advanced hardware technology and image processing improvements on GSI Xtream on Revolution CT has made it the first volume spectral CT scanner with a simplified workflow, designed to advance spectral CT into routine practice.

GSI HyperDrive, enabled by advanced 160mm detector design and high helical pitch, achieves 245mm/s GSI volume scan speed without compromising the full 50 cm FOV. This advancement expands the use of GSI to challenging patients who struggle to hold their breath or stay still in ER settings. Ultra-fast kV switching, enabled by generator and tube improvements, achieves 3x faster switching performance between low and high kVp, which helps to improve material quantification accuracy.

Along with these hardware innovations, advancements in spectral modeling, projection-based material decomposition, and noise modeling have led to increased uniformity across Z, consistent quantification accuracy in Material Density images across all dose levels, and lower noise with better noise texture for low keV and Material Density images. These enhancements will help in small lesion detection and tissue characterization across all GSI image types and clinical applications. Further, the three-stage Metal Artifact Reduction algorithm added to GSI images will reduce beam hardening artifacts from metal objects to reveal anatomic details that may have normally been hidden without the correction.

Simplified workflow at scan time (via GSI Profile and GSI Assist), real time reconstruction (via Xtream Recon) and capability to directly transfer GSI images from console to PACS, make the breakthrough of integrating GSI Xtream with routine radiology workflow.

With all these innovations, GSI Xtream on Revolution CT is creating the future of CT spectral imaging, today.

Clinical Images Courtesy of:

- Duke University Medical Center, USA
- Jena University Hospital, Germany
- The First Affiliated Hospital of Dalian Medical University, China

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Reference

1. Alvarez, R. E., & Macovski, A. Energy-selective reconstructions in X-ray computerized tomography. Physics in Medicine and Biology, 1976; 21(5), 733-744.
2. Avrin DE, Macovski A, Zatz LE (1978) Clinical application of Compton and photo-electric reconstruction in computed tomography: preliminary results. Invest Radiol 13:217-222
3. Chiro GD, Brooks RA, Kessler RM, et al.Tissue signatures with dual-energy computed tomography. Radiology 1979;131(2):521-523.
4. Johnson TR, Krauss B, Sedlmair M, et al.Material differentiation by dual energy CT: initial experience. Eur Radiol 2007;17(6):1510-517.
5. Kelcz F, Joseph PM, Hilal SK. Noise considerations in dual energy CT scanning. Med Phys 1979;6(5): 418-425.
6. Graser A, Johnson TR, Chandarana H,Macari M. Dual energy CT: preliminary observations and potential clinical applicationsin the abdomen. Eur Radiol 2009;19(1):13-23.
7. Kaza RK, Caoili EM, Cohan RH, Platt JF. Distinguishing enhancing from nonenhancingrenal lesions with fast kilovoltage-switching dual-energy CT. AJR Am J Roentgenol 2011;197(6):1375-1381
8. McCollough, et al. Dual- and Multi-Energy CT: Principles, Technical Approaches, and Clinical Applications, Radiology. 276(3): 637-653, (2015).
9. Dubourg B, Caudron J, Lestrat JP, et al. Single-source dual-energy CT angiography with reduced iodine load in patients referred for aortoiliofemoral evaluation before transcatheter aortic valve implantation: impact on image quality and radiation dose. Eur Radiol. 2014;24:2659-2668.
10. Lin XZ, Wu ZY, Tao R, et al. Dual energy spectral CT imaging of insulinoma value in preoperative diagnosis compared with conventional multi-detector CT. Eur J Radiol. 2012;81:2487-2494.
11. Shuman WP, Green DE, Busey JM, et al. Dual-energy liver CT: effect of monochromatic imaging on lesion detection, conspicuity, and contrast-to-noise ratio of hypervascular lesions on late arterial phase.AJR Am J Roentgenol. 2014;203:601-606.
12. Wu X, Langan DA, Xu D, et al. Monochromatic CT image representation via fast switching dual kV. Proc SPIE 2009; 7258: 725845.
13. Matsumoto K, Jinzaki M, Tanami Y, Ueno A, Yamada M, Kuribayashi S. Virtual mono- chromatic spectral imaging with fast kilovolt- age switching: improved image quality as compared with that obtained with conventional 120-kVp CT. Radiology 2011;259(1): 257-262.
14. Pinho DF, Kulkarni NM, Krishnaraj A, Kalva SP, Sahani DV. Initial experience with single-source dual-energy CT abdominal angiography and comparison with single-energy CT angiography: image quality, enhancement, diagnosis and radiation dose. Eur Radiol 2013;23(2):351-359.)
15. Clark, Z. E., Bolus, D. N., Little, M. D., & Morgan, D. E. (2015). Abdominal rapid-kV-switching dual-energy MDCT with reduced IV contrast compared to conventional MDCT with standard weight-based IV contrast: an intra-patient comparison. Abdominal Imaging, 40(4), 852-858.
16. Yuan, R., Shuman, W. P., Earls, J. P., Hague, C. J., Mumtaz, H. A., Scott-Moncrieff, A., et al. (2012). Reduced iodine Load at CT Pulmonary Angiography with Dual-Energy Monochromatic Imaging: Comparison with Standard CT Pulmonary Angiography: A Prospective Randomized Trial. Radiology, 262(1), 290-297.
17. Shuman, WP et al. Department of Radiology, University of Washington, 1959 NE Paci c Street, Box 357115, Seattle, WA, 98195, USA, Abdom Radiol (NY). 2017 Mar;42(3):759-765, PMID: 28084544

18. Patino, M., Prochowski, A., Agrawal, M. D., Simeone, F. J., Gupta, R., Hahn, P. F., & Sahani, D. V. (2016). Material Separation Using Dual-Energy CT: Current and Emerging Applications. RadioGraphics, 36(4), 1087-1105.

19. Agrawal, M. D., Pinho, D. F., Kulkarni, N. M., Hahn, P. F., Guimaraes, A. R., & Sahani, D. V. (2014). Oncologic Applications of Dual-Energy CT in the Abdomen. RadioGraphics, 34(3), 589 - 612

20. Bauer RW, Frellesen C, Renker M, et al. Dual energy CT pulmonary blood volume assessment in acute pulmonary embolism: correlation with D-dimer level, right heart strain and clinical outcome. Eur Radiol 2011;21(9):1914-1921.

21. Pontana F, Faivre JB, Remy-Jardin M, et al. Lung perfusion with dual-energy multidetector-row CT (MDCT): feasibil- ity for the evaluation of acute pulmonary embolism in 117 consecutive patients. Acad Radiol 2008;15(12):1494-1504.

22. Chae EJ, Seo JB, Jang YM, et al. Dual-energy CT for assess- ment of the severity of acute pulmonary embolism: pulmonary perfusion defect score compared with CT angiographic ob- struction score and right ventricular/left ventricular diameter ratio. AJR Am J Roentgenol 2010;194(3):604-610.

23. Morgan DE, Weber AC, Lockhart ME, Weber TM, Fineberg NS, Berland LL. Differentiation of high lipid content from low lipid content adrenal lesions using single-source rapid kilovolt (peak)-switching dual-energy multidetector CT. J Comput Assist Tomography 2013;37(6):937-943.

24. Pal D, Sen Sarma K, Hsieh J. Metal artifact correction algorithm for CT. In: Nuclear Science Symposium and Medical Imaging Conference (NSS/MIC). Washington, DC: IEEE, 2013:1–4

25. Girijesh K. Yadava, Debashish Pal, Jiang Hsieh, "Reduction of metal artifacts: beam hardening and photon starvation effects", Proc. SPIE 9033, Medical Imaging 2014: Physics of Medical Imaging, 90332V (19 March 2014);

26. Pal D, Dong S, Genitsarios I, Hsieh J. Smart Metal Artifact Reduction [White Paper]. Technical Report, General Electric Healthcare Company, 2013.

27. Jiang Hsieh, Brian Nett, Zhou Yu, Ken Sauer, Jean-Baptiste Thibault, Charles A. Bouman, Recent Advances in CT Image Reconstruction, 2013.

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