White Paper

DirectDensity™

Technical principles and implications for radiotherapy

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Introduction

CT simulators are ubiquitous in radiotherapy departments across the world. It is an important step in radiotherapy to accurately calculate dose distributions for treatment planning. Within these calculations, the patient anatomy is reflected by images that are obtained with such CT simulators. Typically, treatment planning systems convert the CT value to a certain material coefficient according to a given calibration curve. Therefore, a high proportion of radiotherapy departments use standardized CT simulation protocols with a fixed tube voltage of 120kV. This reduces the number of calibration curves that have to be handled and ensures accurate dose calculation.

However, these CT images are also used by physicians to define the target and organs at risk for treatment prescription. To get optimal contrast and image quality that can help delineate structures, CT images could be acquired using different kV settings based on patient size or on the presence of contrast media[1,2]. Typically, this is not implemented to avoid the complexity of handling several calibration curves but at the expense of potentially improved image quality.

For example, in the case of bariatric patients, who have a higher X-ray attenuation, the output current of the X-ray tube at lower or conventional kV settings may not be sufficient to produce the required contrast-to-noise ratio. For these patients, higher X-ray tube voltages might be necessary. In the case of pediatric patients or younger breast cancer patients, who might be unnecessarily exposed to higher radiation doses with a conventional 120 kV scan protocol, the contrast-to-noise ratio in the images could be maintained by using a low kV scan protocol with potentially lower dose to these patients.

While it may be be beneficial to design a more patient-specific scan protocol, it is often not practical, for several reasons, to optimize the imaging protocol within a busy radiotherapy department. First, it is challenging and time-consuming to optimize CT imaging protocols given the complex relationships between kV, mAs, dose, contrast, and noise. To overcome these challenges, engineers at Siemens Healthcare have implemented functionalities such as CARE Dose 4D and CARE kV[3,4], which can semi-automatically adapt tube current and voltage. However, managing several calibrations for different tube voltages within the treatment planning system, if supported at all, could hinder the workflow and be prone to errors. Thus, it is not always practical to implement changes in kV settings.

In response to these various challenges, Siemens Healthcare introduced the DirectDensity™ algorithm, which directly reconstructs images that can be interpreted as showing relative electron density** at any given kV setting. DirectDensity™ eliminates the need for tube-voltage-dependent system calibration as it is a single linear relationship. This creates scope for tube voltage adaption (e.g., CARE kV) to optimize imaging protocols based on patient anatomy.

** As shown by measurements with a Gammex 467 Tissue Characterization Phantom comparing between standard reconstruction (kernel D30) and DirectDensity™ reconstruction (kernel E30). HU value to relative electron density conversion for the standard reconstruction was based on a two-linear-equations approach with individual calibration for each tube voltage. For DirectDensity™ images a single tube-voltage-independent linear conversion was used.

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The DirectDensity™ algorithm

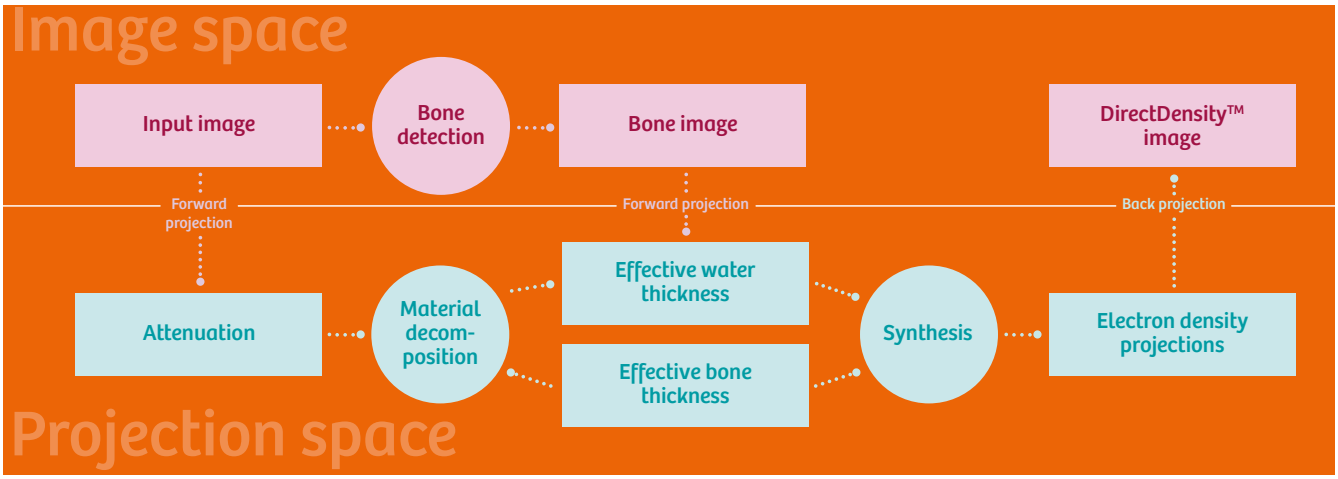


Figure 1: Flow chart of the DirectDensity™ algorithm.

The DirectDensity™ algorithm reconstructs CT images from single-energy CT acquisitions. The resulting CT values can be interpreted as showing relative electron density. This is achieved by combining image-based bone detection with a projection-based material decomposition. Synthetic projections of the relative electron density are obtained using the two-material decomposition of water and bone. DirectDensity™ images are finally reconstructed from the synthetic projections of relative electron density. The DirectDensity™ algorithm can be combined with both standard and iterative reconstruction (SAFIRE* or ADMIRE*). The processing steps are detailed in the following paragraphs, and they are illustrated in the flow chart in Figure 1.

Input image and attenuation

The attenuation in projection space, also called a sinogram, contains the attenuation information for each X-ray beam that travels through the patient from the single energy CT acquisition. The corresponding input images show the distribution of the attenuation coefficient as cross-sectional images. Input images can be obtained from attenuation with a back projection, and attenuation can be obtained from the input images with a forward projection. Both attenuation and input images serve as input for the algorithm.

Bone detection

It is possible to approximate images of the basis material bone by using a threshold on the input images. Voxels in the input images that have a CT value below this threshold are assumed to contain water with variable density depending on the CT value. For CT values above this threshold, it is assumed that a voxel contains a mixture of water and bone. The amount of bone in this voxel increases linearly with CT value. The above is a reliable assumption for natural body materials. Non-natural-body materials, for example metals or contrast agents, are a possible source of errors in this processing step.

* Some features may be optional on SOMATOM systems.

Material decomposition

In general, two-material decomposition from single-energy CT is not possible. Nevertheless, projections of the bone image are used to obtain projections of the effective thickness of the basis material bone. A monotonic relation between effective thicknesses of both basis materials and the attenuation can be established using an exact physical attenuation model. This model incorporates the CT scanner and acquisition-specific parameters, for example tube voltage. This makes it possible to obtain projections of the effective thicknesses of water from the known effective thicknesses of bone and the acquired attenuation. Thus, a complete decomposition into both basis materials is possible.

Synthesis

The sum of the products of the effective thicknesses and the corresponding attenuation coefficients of both basis materials would yield the known attenuation line integrals. Therefore, this is the inversion of the material decomposition in projection space. However, the relative electron density of both basis materials is also known. Adding the products of relative electron density and effective thickness of water and bone for each projection yields line integrals of the relative electron density. This produces synthetic projections, or in other words a sinogram of the distribution of relative electron density.

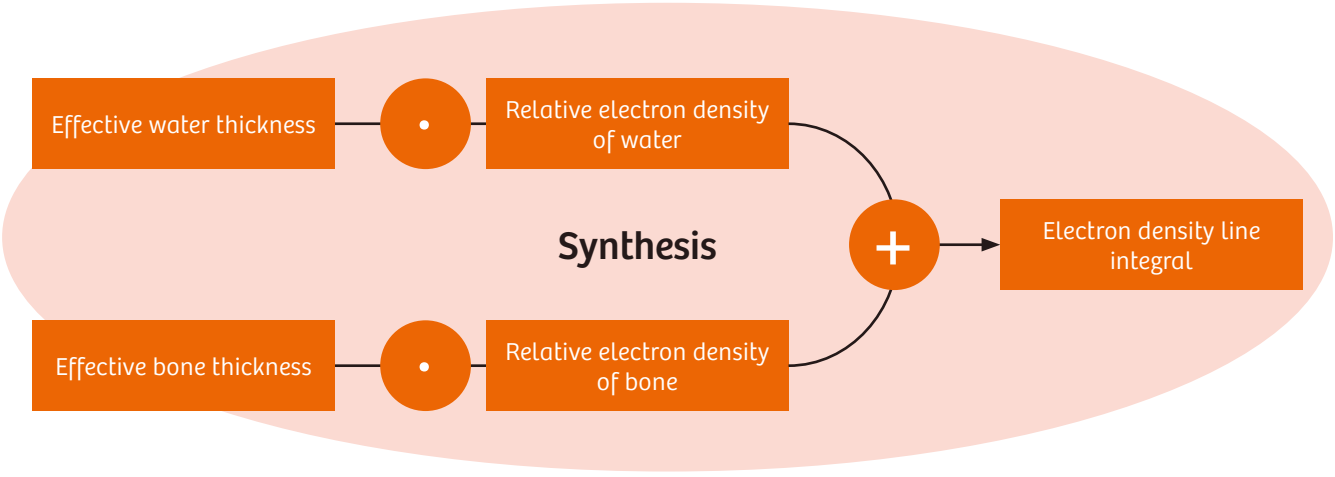


Figure 2: Illustration of the synthesis step. Effective thickness of both materials is multiplied by the corresponding relative electron density. The sum of both products gives the value of a line integral of relative electron density. Calculated for each beam of the CT acquisition, this gives a sinogram of the relative electron density distribution.

How to use DirectDensity™

Enabling DirectDensity™

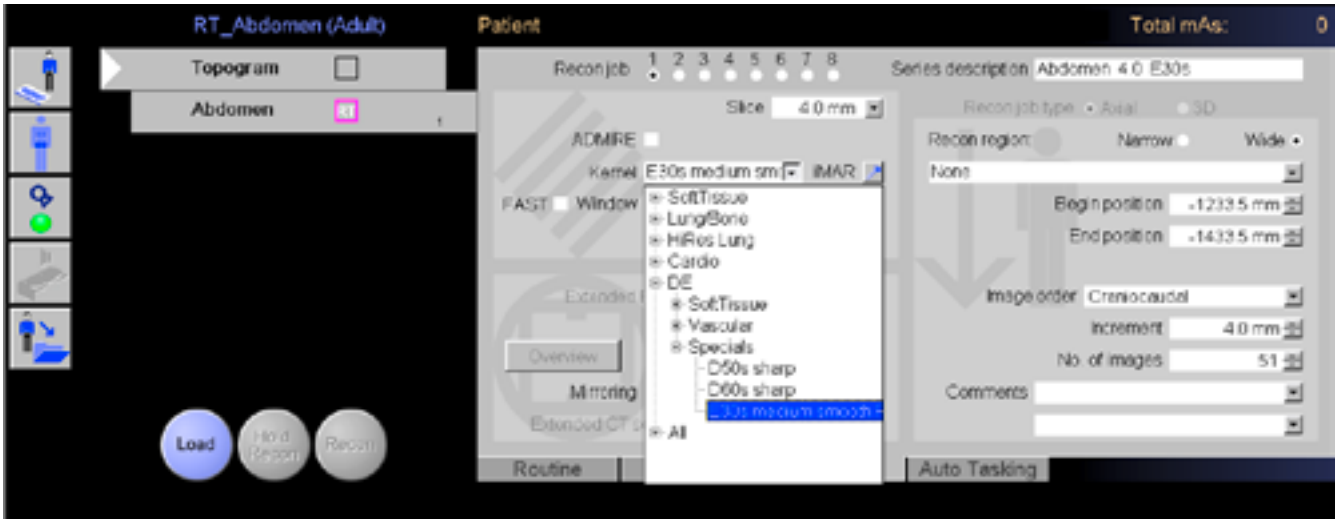


Figure 3: User interface on the CT acquisition workplace, where special E or F kernels can be selected in a recon job to enable the DirectDensity™ algorithm.

The DirectDensity™ algorithm can be enabled during reconstruction by choosing dedicated reconstruction kernels. The special kernels E for a standard reconstruction or F for an iterative reconstruction (e.g., SAFIRE* or ADMIRE*) have to be selected from the list of available kernels. Figure 3 shows the relevant user interface on the Healthcare CT acquisition workplace. The resulting DICOM images of DirectDensity™ reconstructions are provided in the same manner as for standard reconstructions. DirectDensity™ images can be identified by the special kernel names in the relevant DICOM attributes, for example in the series description.

Image values and use in treatment planning systems

DirectDensity™ image values are provided in a HU-like scaling, but are proportional to the relative electron density ρ. This means that image values can be converted to relative electron density using the following equation:

$$\rho = \frac{\text{Image value}}{1000} + 1$$

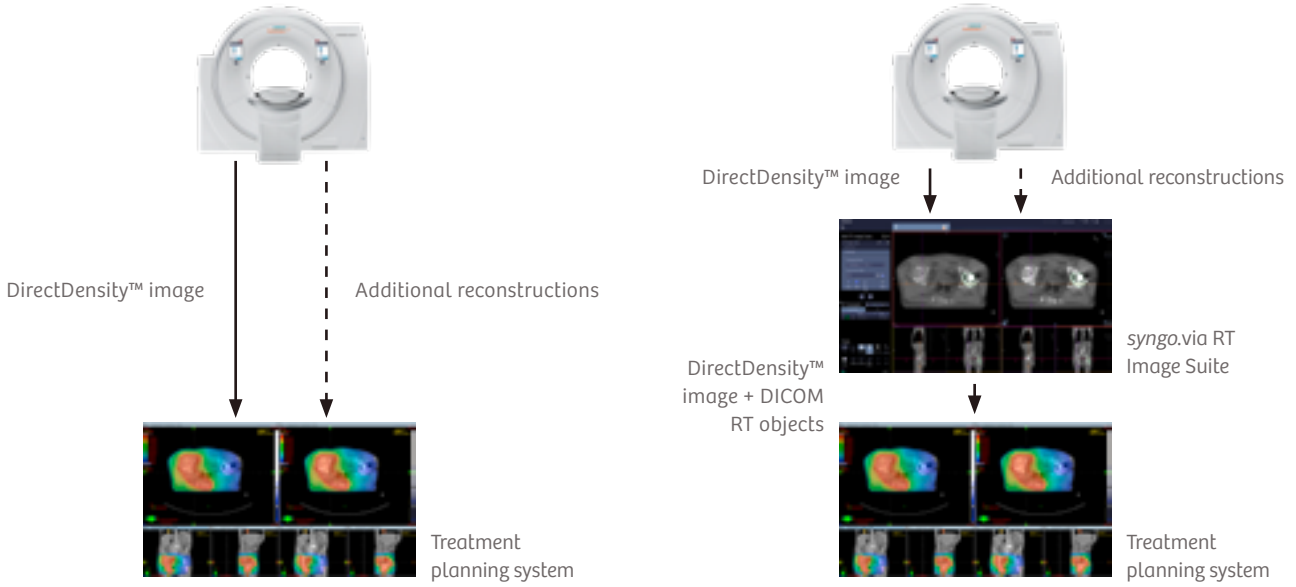
Typically, the treatment planning system will use a calibration curve to convert the CT value of images into relative electron density. This is also true when using DirectDensity™ images. If the treatment planning system expects certain calibration points to be given, then for DirectDensity™ this can be achieved by calculating suitable calibration points, using the linear relationship given above. These calibration points map given CT values to corresponding relative electron densities. An example for the use with DirectDensity™ is shown in Table 1.

Image value	Relative electron density
-1024	0.0
-1000	0.0
+4000	5.0

Table 1: Example of calibration points for use with DirectDensity™. For 12-bit DICOM images, there is usually a range of image values between -1024 and -1000 that would result in negative electron density. In this example, this scenario is prevented by adding an additional calibration point for -1024, which results in the range below -1000 being mapped to zero relative electron density if necessary.

* Some features may be optional on SOMATOM systems.

Unlocking the potential of CT with DirectDensity™



Without *syngo*.via RT Image Suite

- DirectDensity™ reconstructions can be sent directly to the treatment planning system for contouring and dose calculation
- If any additional reconstructions are required for contouring, manual contour propagation is needed

With *syngo*.via RT Image Suite

- The scan protocol can be easily configured with:
  - Primary image: DirectDensity™ image for contouring and dose calculation
  - Up to 7 additional reconstructions for advanced contouring
- Preffered reconstructions can automatically be transfered to *syngo*.via RT Image Suite
- Contours and primary image, i.e., the DirectDensity™ image can be sent to the tratment planning system

Figure 4: Illustration of workflow alternatives for DirectDensity™. Clinical images courtesy of: MAASTRO Clinic, Maastricht, Netherlands.

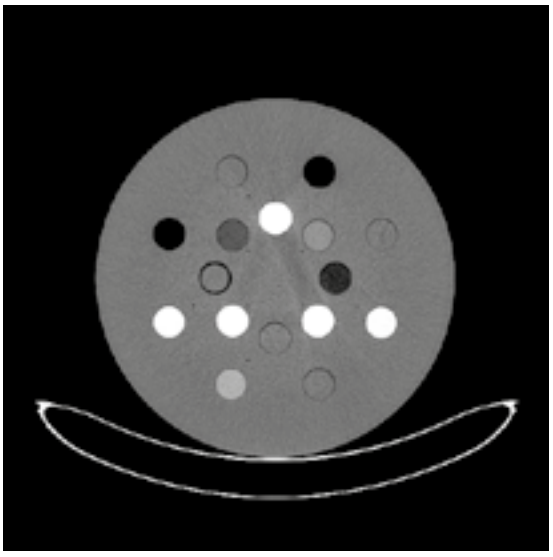
The first and simplest workflow is the following: DirectDensity™ images can be sent directly to the treatment planning system and can be used for contouring and dose calculation.

However, there may be another workflow that unlocks the potential of DirectDensity™ even further. Here, the goal is to have even better images that aid, for example, contouring. In this second workflow, the full potential of patient-specific CT acquisition is used, by adapting tube current and voltage to the patient anatomy, and possibly using automated exposure control mechanisms (e.g., CARE kV).

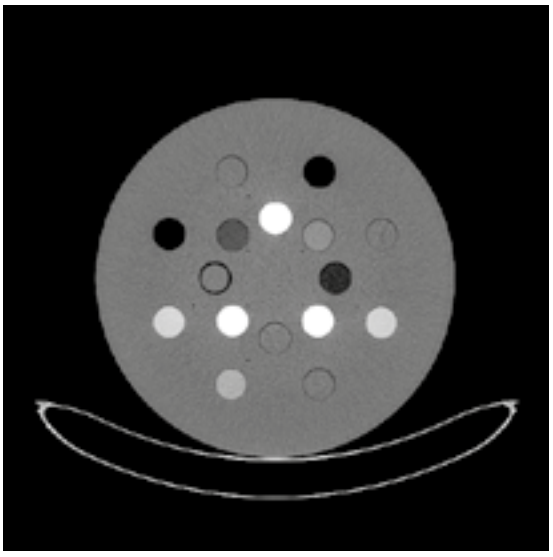
DirectDensity™ has the potential to remove constraints of fixed tube voltage because DirectDensity™ images show reduced variations with regard to tube voltage** and can be used for dose calculation**. In addition, other reconstructions with different settings, such as the kernel, can be performed. These additional reconstructions could be optimized to provide images with potentially better contrast for better visualization that can aid the task of contouring. In summary, DirectDensity™ helps to maintain consistency of dose calculation and could open new possibilities for improved visualization and greater confidence in contouring.

** As shown by measurements with a Gammex 467 Tissue Characterization Phantom comparing between standard reconstruction (kernel D30) and DirectDensity™ reconstruction (kernel E30). HU value to relative electron density conversion for the standard reconstruction was based on a two-linear-equations approach with individual calibration for each tube voltage. For DirectDensity™ images a single tube-voltage-independent linear conversion was used.

Validation of DirectDensity™ on phantoms



Standard reconstruction



DirectDensity™ reconstruction

Figure 5: CT images of a Gammex 467 tissue characterization phantom obtained from the same 120 kV CT acquisition with a standard reconstruction and a DirectDensity™ reconstruction. Window level: C = 40, W = 300.

The Gammex 467 tissue characterization phantom was scanned at different tube voltages to validate the DirectDensity™ algorithm. Figure 5 shows a standard reconstruction and a DirectDensity™ reconstruction from the same acquisition carried out at a tube voltage of 120 kV.

CT value proportional to relative electron density

Figure 6 shows the relationship between measured mean image values for the Gammex 467 tissue substitutes and their true relative electron density. Comparing a standard reconstruction to DirectDensity™, the following observations can be made: With the standard reconstruction, the relationship typically shows a different slope for CT values above and below a certain point between 0 and 100. Additionally, the slope above this point depends on the tube voltage. This implies that a different calibration is needed for each tube voltage. However, when using DirectDensity™, we observe a single linear relationship that is independent of the tube voltage. This shows that one calibration curve in the TPS might be enough even if the tube voltage used for acquisition is varied.

For the standard reconstructions, a calibration relationship using two linear equations was determined for each tube voltage. With this relationship, the mean CT value found for each Gammex 467 tissue substitute was converted to relative electron density $\rho_{standard}$. The single linear relationship given above was used to calculate relative electron density $\rho_{DirectDensity™}$ from DirectDensity™ reconstructions. Using all relative electron densities calculated for the Gammex 467 tissue substitutes, the root mean of the squared relative differences (RMS) of the relative electron densities is calculated by the following equation:

$$RMS = \sqrt{\sum \left(\frac{\rho_{DirectDensity™} - \rho_{standard}}{\rho_{standard}} \right)}$$

This RMS is a measure of relative deviations between relative electron densities obtained from the standard images and the DirectDensity™ images. The RMS values we found are shown in Figure 7 and are below 1.3% for all tube voltages. The deviations are comparable to the magnitude of statistical fluctuations or variations caused by changes in geometry (see Figure 8).

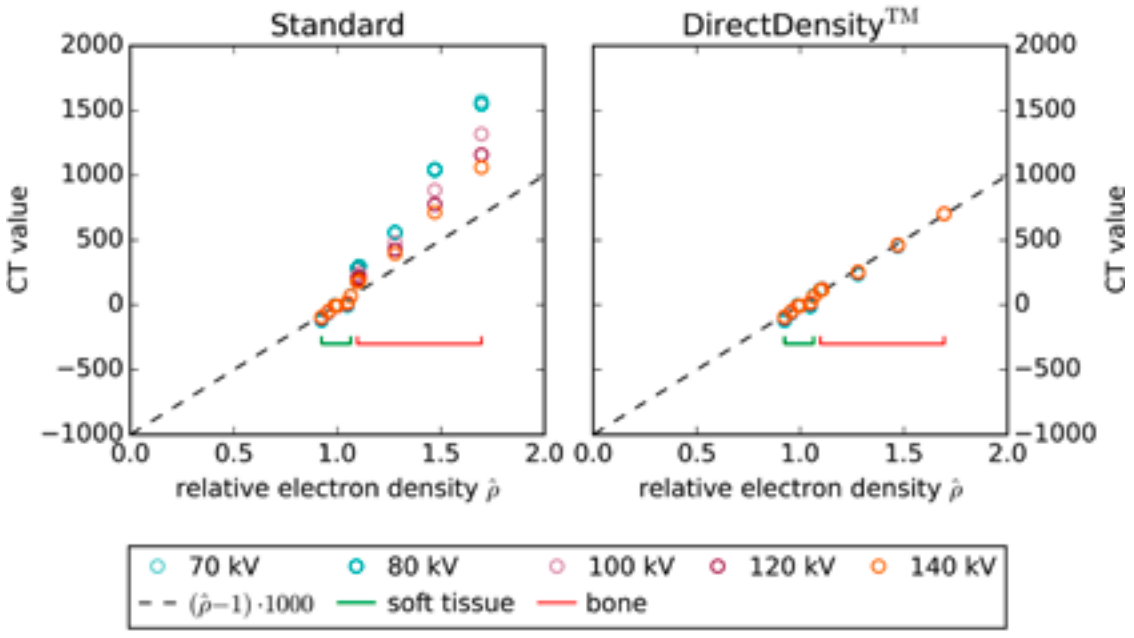


Figure 6: Mean CT values of Gammex 467 tissue substitutes as a function of their true relative electron densities at several tube voltages. Additionally, the dashed line shows the expected relationship between DirectDensity CT values and relative electron density.

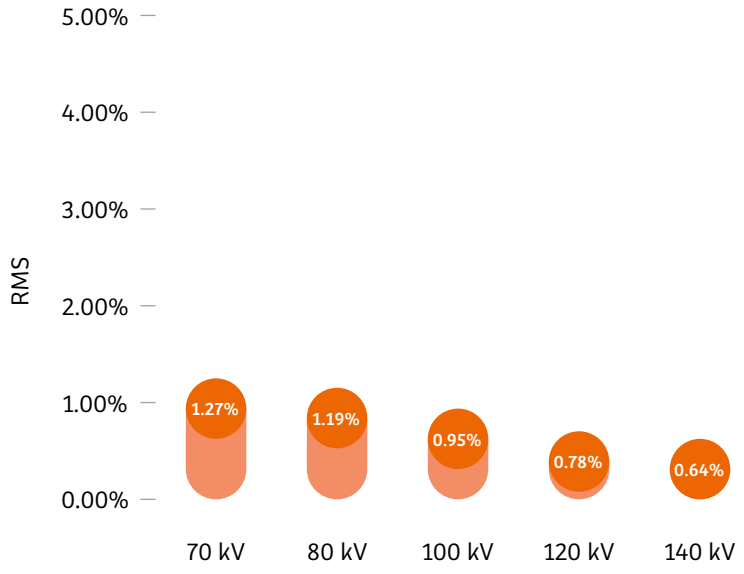


Figure 7: Root mean of the squared relative differences (RMS) between relative electron densities obtained with a standard reconstruction and a DirectDensity™ reconstruction. Relative electron densities of Gammex 467 tissue substitutes were measured. For standard reconstruction, a tube- voltage-dependent two-linear-equation calibration was used. For DirectDensity™ , the direct linear relation was used.

Improved beam hardening correction

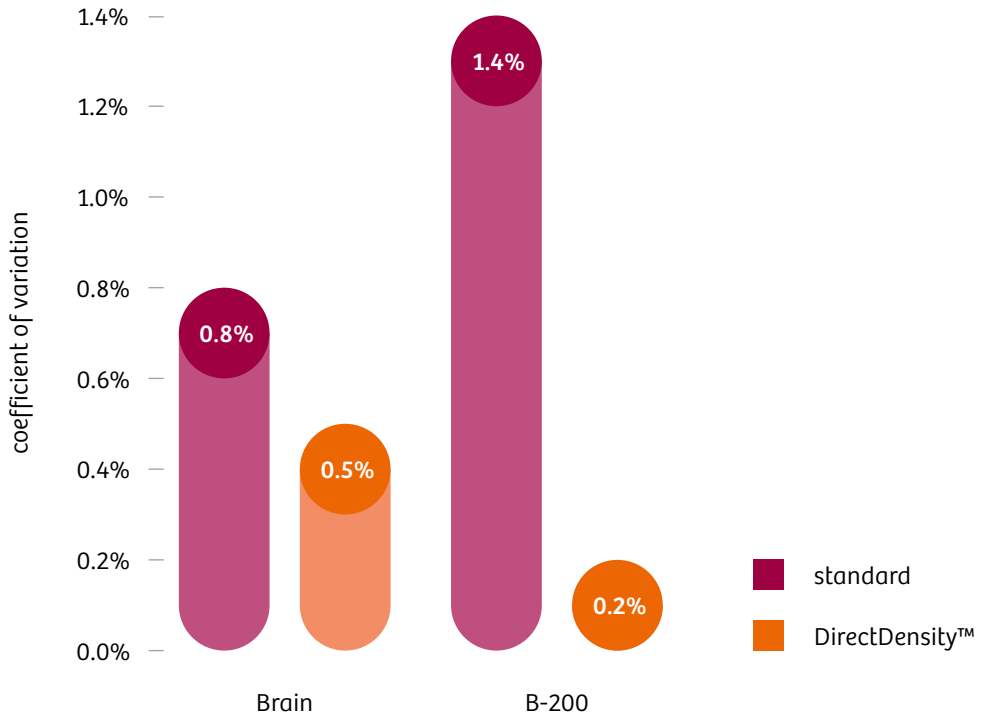


Figure 8: Coefficient of variation measured by varying the position of tissue substitute inserts in the Gammex 467 phantom at 120 kV.

Thanks to the two-material attenuation model, DirectDensity™ supports the reduction of beam hardening artifacts originating from dense, bone-like tissue. This can be observed in the reduction of streak artifacts between bone substitutes in the Gammex 467 phantom, as can be seen in Figure 5. We observed this visible reduction of streak artifacts in the Gammex 467 phantom for all tested tube voltages. Moreover, DirectDensity™ supports increased CT value stability with regard to geometric variations**. This is also an intrinsic effect of the improved two-material attenuation model used in DirectDensity™. This was tested with several repeated scans of the Gammex 467 phantom with varied arrangements of the tissue substitutes.
A coefficient of variation for the CT value of each tissue substitute was calculated to get a measure of the variability with regard to the arrangement.

For example, at 120 kV, the coefficient of variation of the brain substitute with a standard reconstruction was 0.8% This declined to 0.5% with DirectDensity™. With bone-like tissue, an even greater reduction can be observed. For example, for the B-200 low density bone substitute, we observed a reduction in CT value variability from 1.4% to 0.2% with DirectDensity™ at 120 kV. A reduction in the same order of magnitude was observed for all bone-like tissue substitutes. This is a result of the ability of DirectDensityTM and its beam hardening correction to improve CT value stability with regard to geometric variations.

** As shown by measurements with a Gammex 467 Tissue Characterization Phantom comparing between standard reconstruction (kernel D30) and DirectDensity™ reconstruction (kernel E30). HU value to relative electron density conversion for the standard reconstruction was based on a two-linear-equations approach with individual calibration for each tube voltage. For DirectDensity™ images a single tube-voltage-independent linear conversion was used.

Dosimetric evaluation

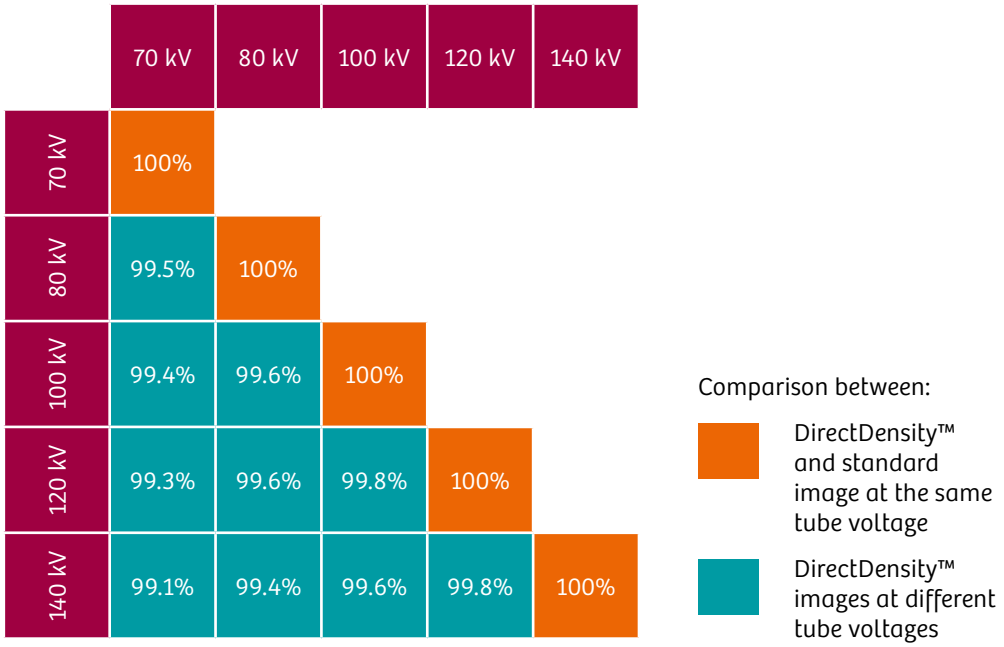


Figure 9: Passing rates resulting from a gamma analysis with criteria of 1 mm and 1%. Dose distributions for a 7-field MV photon treatment plan for an anthropomorphic thorax phantom were calculated. Standard images were converted to relative electron density using a tube-voltage-dependent stoichiometric calibration. DirectDensity™ images were converted using a tube-voltage-independent linear relation.

Recently, Zhao et al.[6] conducted a study in which they reconstructed standard CT images of the Gammex 467 tissue characterization phantom at 70 kV, 80 kV, 100 kV, 120 kV and 140 kV. They also reconstructed DirectDensity™ images for these acquisitions. They used stoichiometric CT calibrations for the standard images and used a single linear conversion as above for DirectDensity™ images. A 9-field MV photon treatment plan was created and the dose distributions were calculated for all standard and DirectDensity™ images.

Dosimetric differences were evaluated using gamma analysis[5] with criteria of 1 mm and 1%. Pass rates equal to or above 99.9% were achieved by comparing standard images and stoichiometric calibration with DirectDensity at the same tube voltage. The same passing rates were obtained in a similar study[7] which calculated a 7-field MV photon treatment plan on an anthropomorphic thorax phantom. Passing rates for the anthropomorphic study are shown in Figure 9.

Clinical examples

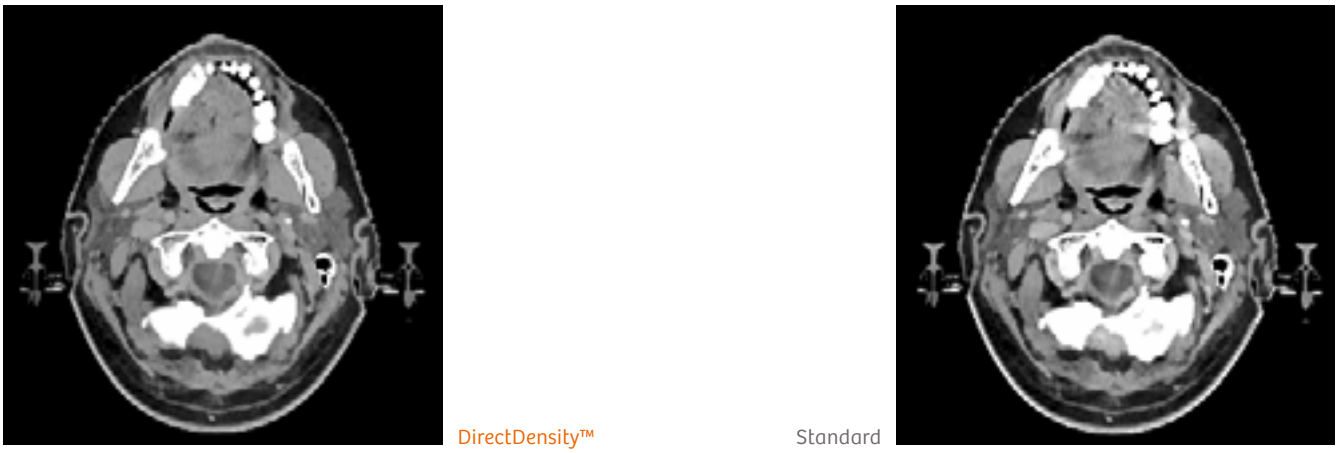


Figure 10: CT images of a head & neck case from the same 120 kV acquisition. Courtesy of MAASTRO Clinic, Maastricht, Netherlands. Window level: C=40, W=300.

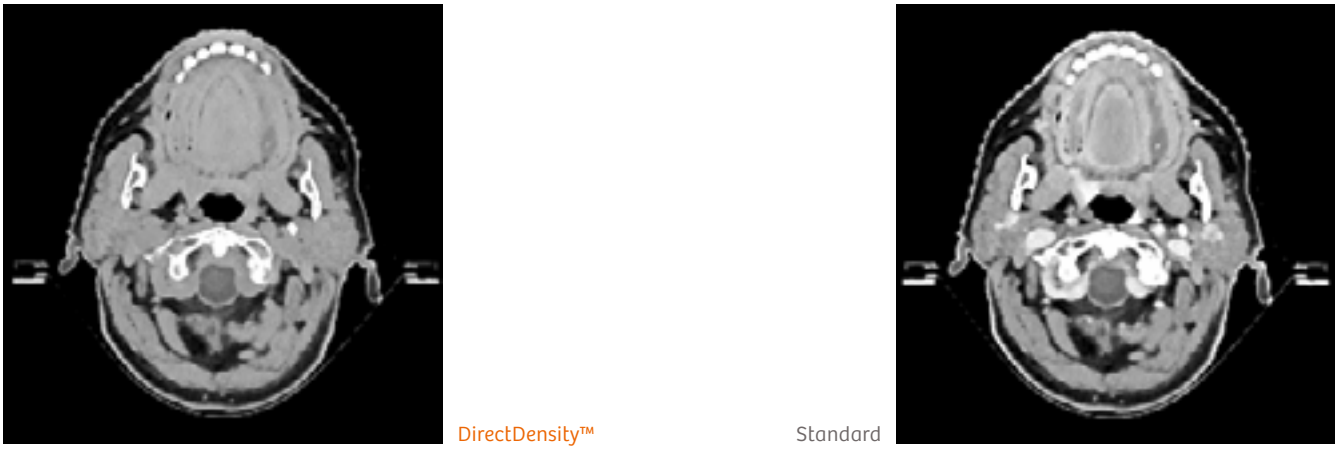


Figure 11: CT images of a head & neck case from the same 100 kV acquisition with contrast media. Courtesy of MAASTRO Clinic, Maastricht, Netherlands. Window level: C=40, W=300.

Figure 10 and Figure 11 show CT images of head and neck scans at 120 kV and 100 kV of different patients, acquired with a SOMATOM Confidence® RT Pro. In both figures, a DirectDensity™ image is compared with the standard image from the same acquisition. In Figure 10, a reduction of beam hardening artifacts with DirectDensity™ can be observed, especially in the area surrounding the teeth. The scan in Figure 11 was performed with contrast media at 100 kV, and shows, that DirectDensity™ images reduce the contrast of the contrast media.

This indicates that when using contrast media, a workflow using DirectDensity™ images for dose calculation and standard images for contouring might be beneficial. In DirectDensity™ images, the influence of contrast media on dose calculation might be reduced to an acceptable level, while in standard images the full benefit of improved contrast media visualization at lower tube voltages can be exploited.

Figure 12 shows a visual comparison of dose distributions for the same beam configuration, calculated on DirectDensity™ images and standard images. Visually, the correspondence between both distributions is high. This finding is supported by the dosimetric evaluation performed on phantoms.

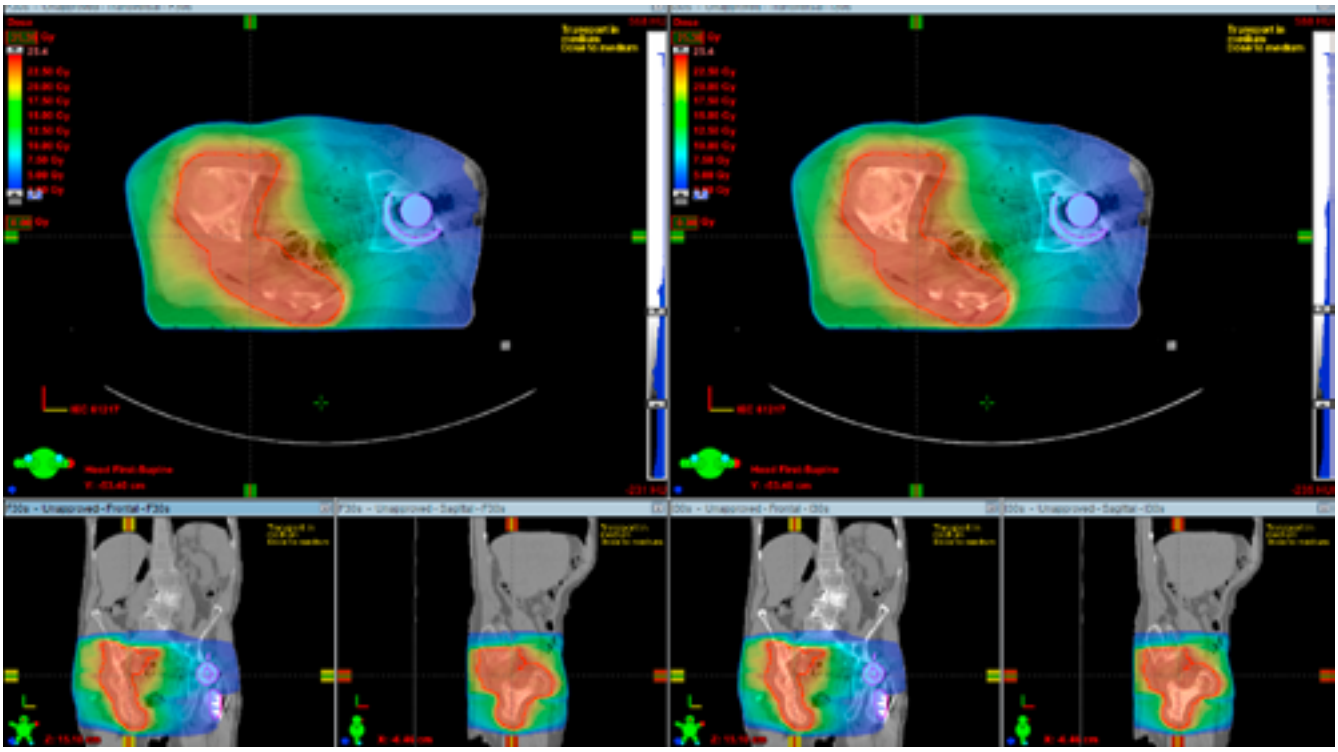


Figure 12: Comparison of dose distributions (color wash) for the same treatment plan calculated on a DirectDensity™ image series (F30, left) and a standard image series (I30, right) within the treatment planning system (Eclipse, Varian Medical Systems). Courtesy of MAASTRO Clinic, Maastricht, Netherlands.

Compatibility and limitations

The current implementation of the DirectDensity™ algorithm available with *syngo*CT VA62 can be used in combination with iterative reconstruction (SAFIRE* or ADMIRE*), iMAR* and respiratory-gated 4D CT*. In this release DirectDensity™ cannot be applied when using extended FoV*, HD FoV*, or extended CT scale*. The DirectDensity™ algorithm is designed based on the assumption that body materials are composed of a mixture of water and bone with variable density. This is a reasonable assumption for natural body materials. Non-natural materials, for example metals and contrast agents like iodine, will decrease accuracy and – as with conventional CT images – can potentially lead to image artifacts. It is technically possible to select DirectDensity™ kernels in reconstructions of Dual Energy scans, but the resulting DirectDensity™ images are not suitable for Dual Energy post processing.

Conclusions

The DirectDensity™ algorithm provides images in which the CT values can be interpreted as showing relative electron density. Currently, radiation therapy treatment planning systems require a kV-specific calibration curve to convert CT image values into relative electron density. A single linear relationship that does not depend on the tube voltage of CT acquisitions can unlock the unused potential of CT imaging in radiotherapy, as it removes the need for scan protocols with fixed tube voltage. In addition, it would reduce the potential sources of errors that may be introduced if the wrong TPS calibration curve is selected. This new capability of varying tube voltage while using one calibration curve offers scope for designing scan protocols that are more personalized to the patient. At the same time, specific reconstructions provide images that may provide better visualization and better contrast for different tasks, especially contouring, in radiotherapy planning.

Overall, we believe that the introduction of DirectDensity™ provides a new opportunity to easily access images that are optimized for both dose calculation and contouring.

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