Post-mortem computed tomography: Technical principles and recommended parameter settings for high-resolution imaging: — Source link ⤢

Dominic Gascho, Michael J. Thali, Tilo Niemann

Institutions: University of Zurich, University Hospital of Lausanne

Related papers:

- Image-guided virtual autopsy findings of gunshot victims performed with multi-slice computed tomography (MSCT) and magnetic resonance imaging (MRI) and subsequent correlation between radiology and autopsy findings

- Radiological investigation of gunshot wounds: a systematic review of published evidence

- Gunshot fatalities: Correlation between post-mortem multi-slice computed tomography and autopsy findings: A 30-months retrospective study

- Standardized medical image registration for radiological identification of decedents based on paranasal sinuses.

- State of the art in post-mortem computed tomography: a review of current literature.

Year: 2018

Post-mortem computed tomography: Technical principles and recommended parameter settings for high-resolution imaging

Gascho, Dominic ; Thali, Michael J ; Niemann, Tilo

Abstract: Post-mortem computed tomography (PMCT) has become a standard procedure in many forensic institutes worldwide. However, the standard scan protocols offered by vendors are optimised for clinical radiology and its main considerations regarding computed tomography (CT), namely, radiation exposure and motion artefacts. Thus, these protocols aim at low-dose imaging and fast imaging techniques. However, these considerations are negligible in post-mortem imaging, which allows for significantly increased image quality. Therefore, the parameters have to be adjusted to achieve the best image quality. Several parameters affect the image quality differently and have to be weighed against each other to achieve the best image quality for different diagnostic interests. There are two main groups of parameters that are adjustable by the user: acquisition parameters and reconstruction parameters. Acquisition parameters have to be selected prior to scanning and affect the raw data composition. In contrast, reconstruction parameters affect the calculation of the slice stacks from the raw data. This article describes the CT principles from acquiring image data to post-processing and provides an overview of the significant parameters for increasing the image quality in PMCT. Based on the CT principles, the effects of these parameters on the contrast, noise, resolution and frequently occurring artefacts are described. This article provides a guide for the performance of PMCT in morgues, clinical facilities or private practices.

Technical report

Post-mortem computed tomography: Technical principles and recommended parameter settings for high-resolution imaging

Dominic Gascho[1], Michael J. Thali[1] and Tilo Niemann[2]

Abstract

Post-mortem computed tomography (PMCT) has become a standard procedure in many forensic institutes worldwide. However, the standard scan protocols offered by vendors are optimised for clinical radiology and its main considerations regarding computed tomography (CT), namely, radiation exposure and motion artefacts. Thus, these protocols aim at low-dose imaging and fast imaging techniques. However, these considerations are negligible in post-mortem imaging, which allows for significantly increased image quality. Therefore, the parameters have to be adjusted to achieve the best image quality. Several parameters affect the image quality differently and have to be weighed against each other to achieve the best image quality for different diagnostic interests. There are two main groups of parameters that are adjustable by the user: acquisition parameters and reconstruction parameters. Acquisition parameters have to be selected prior to scanning and affect the raw data composition. In contrast, reconstruction parameters affect the calculation of the slice stacks from the raw data. This article describes the CT principles from acquiring image data to post-processing and provides an overview of the significant parameters for increasing the image quality in PMCT. Based on the CT principles, the effects of these parameters on the contrast, noise, resolution and frequently occurring artefacts are described. This article provides a guide for the performance of PMCT in morgues, clinical facilities or private practices.

Keywords

Post-mortem CT, forensic radiology, computed tomography, CT guide, virtual autopsy, Virtopsy

Introduction

The use of computed tomography (CT) has been well-established in clinical radiology for a long time, and CT scanners have experienced various stages of development. Nowadays, most scanners use the x-ray fan-beam technique combined with detector arrays opposite the x-ray tube, perform helical (spiral) scanning and utilise reconstructions based on filtered back-projection (FBP) or iterative image reconstructions, which are considered third-generation multi-slice CT (MSCT) scanners.[1,2] MSCT, multi-row CT or multi-detector CT (MDCT) scanners are based on the use of several rows of multiple small detectors, and the scanners are specified by the maximum number of slices that can be acquired simultaneously (e.g. 4-, 8-, 16-, 32-, 64- and 128-slice MSCT).[3] Acquiring slices simultaneously results in a reduced acquisition time and reduced heating of the x-ray tube. Applying a

widened x-ray beam along the z-axis was introduced in 1992 (CT-Twin), but the birth of the MSCT can be dated to 1998.[3] In 2002, the first 16-slice CT scanners were installed, followed by 32- and 40-slice scanners in 2003 and 2004, respectively, and 64-slice scanners in 2005.[3] At the same time, post-mortem CT (PMCT) and PMCT angiography (PMCTA), as a supplement to autopsy using the term 'Virtopsy' or 'virtual autopsy', arose throughout the world.[4–8] As forensic

[1]Department of Forensic Medicine and Imaging, Zurich Institute of Forensic Medicine, University of Zurich, Switzerland
[2]Department of Radiology, Cantonal Hospital Baden, Switzerland

Corresponding author:
Dominic Gascho, Zurich Institute of Forensic Medicine, University of Zurich, Winterthurerstrasse 190/52, Zurich, 8057, Switzerland.
Email: dominic.gascho@irm.uzh.ch

radiology, and especially PMCT, continues to grow, forensic institutes that have recently started performing PMCT depend on the vendor's program settings or the knowledge of clinical users. However, the needs of clinical and forensic radiology differ greatly, and two major advantages in PMCT that lead to an improvement in the image quality are the ability to neglect radiation exposure and the absence of motion artefacts.[9] Therefore, the PMCT and PMCTA parameters have to be adjusted to use these advantages to achieve the best image quality. Apart from the technology used, the image quality itself is dependent on user-selected pre-scan acquisition parameters and user-selected post-scan reconstruction parameters. Below, we explain the CT principles, and based on these, we recommend relevant parameter adjustments for PMCT. The aims of this review are to give an overview of parameter adjustments to enhance the image quality in PMCT and to offer information on current CT technologies.

Contrast, noise and spatial resolution

First, the term 'image quality' must be defined. It can be divided into objective and subjective quality. Subjective image quality mainly depends on the perception, individual preferences and former experiences of the observer, while the objective quality criteria can be measured objectively. The main characteristics of objective image quality are the contrast, noise and spatial resolution. The contrast can be described as the depiction of different adjacent densities and depends on the administered x-ray photon energy and the type of tissue or material that has been penetrated.[10] The number of detector-captured photons is limited, and the detector measurements are randomly fluctuating, which causes quantum mottle (noise), displayed as graininess in the reconstructed images.[1,11] The spatial resolution describes the ability to distinguish between the smallest, most closely spaced elements in an image and is mainly limited by the aperture and detector size.[11] The noise and the spatial resolution are image characteristics that are interlinked. So, trade-offs have to be made according to the anatomical region of interest. Apart from the technology used, the image quality itself is dependent on the acquisition parameter settings and the reconstruction parameter settings.

Artefacts

Further, objective image quality is affected by the presence of artefacts. In CT imaging, a variety of artefacts with different causes may occur. In the following, only the most common artefacts are mentioned, which can be significantly reduced by parameter adjustments.

Beam hardening (BH) may occur as cupping artefacts. Cupping artefacts refers to the x-ray photons penetrating the centre of an object being hardened due to the higher radio-opacity in the centre than at the edges of an object.[12] BH may appear as hypodense dark streaks along the x-ray path through thick regions of bone or contrast media.[11,13] The Hounsfield bar is a known example and is described as the appearance of a dark streak between the petrous bones in the base of the skull.[13] Partial-volume artefacts (PV) may occur if detector elements are partly shaded by the edge of a high-contrast structure.[13] Thus, the attenuation measured by the detector element is a blend of the high-contrast structure and the surrounding low-contrast structure, which is displayed as an incorrect HU value. PV artefacts occur in-plane and along the z-axis.[13] Windmill (spiral) artefacts (WM) are caused during helical data acquisition by high edge differences between two detector rows due to the layout of the anatomical structures.[11,14,15] These artefacts appear as streak-like patterns that seem to rotate when scrolling through the images.[16] Metallic artefacts due to implants or foreign bodies are caused by a mixture of the above-mentioned artefacts but are mainly due to heavy BH. Objects or body parts beyond the in-plane scan field may cause out-of-scan-field artefacts[17] that appear as streaks or shading in the image.[12] Apart from those caused by obese bodies, out-of-scan field artefacts may occur especially by decomposed, charred or frozen bodies due to inadequate positioning.

Generally, the adjustments of the PMCT parameters allow for the reduction of the extent of all these artefacts in post-mortem imaging. But first of all, adequate body positioning is essential.
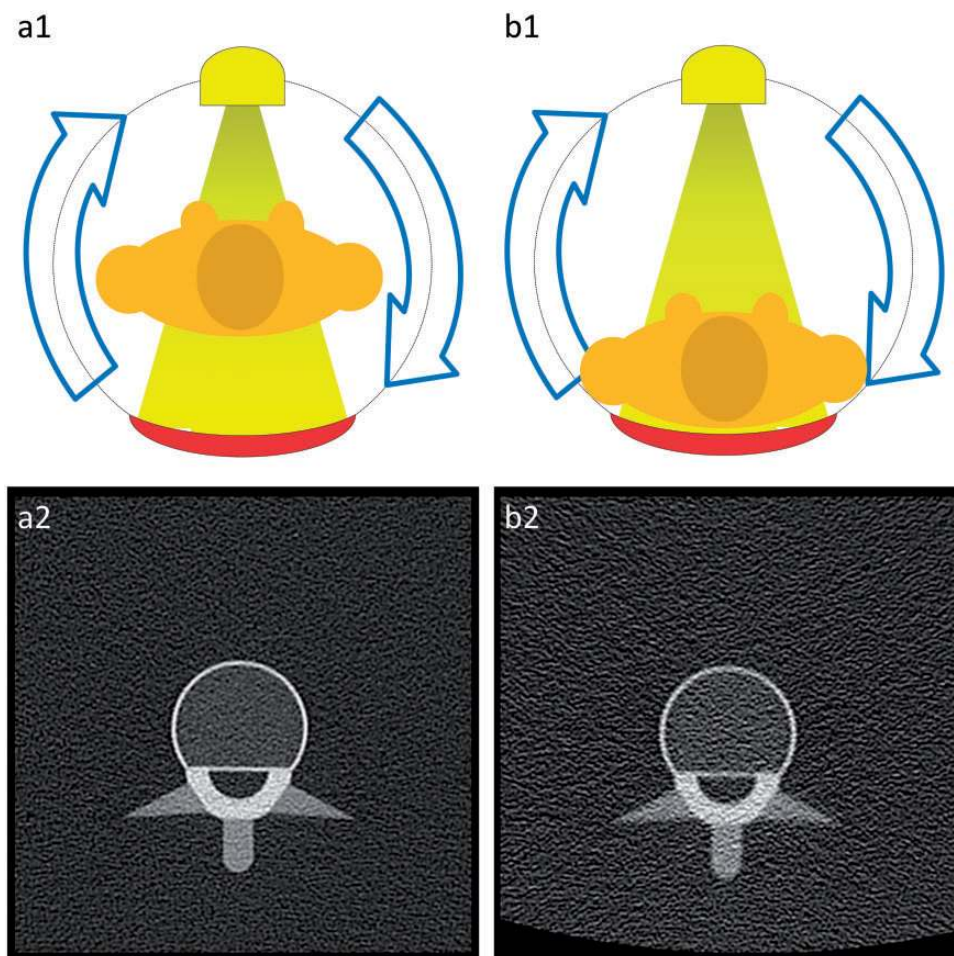
Body positioning

It is advised to put the deceased into a body bag or on a plastic foil to prevent any soiling of the CT table or damage due to the leakage of any type of body fluid. Depending on the condition of the deceased, a plastic foil beneath the body is preferable, as a body bag may have handles and zippers, which should be removed from the scan region as they may cause artefacts in the images. Another significant reason is that the body bag may be visible in post-processing reconstructions (e.g. volume renderings) or even limit the post-processing possibilities regarding body surface, as it covers the body. Usually, the corpse is in a supine position, depending on its condition. Generally, the arms should be pressed slightly against the body by using position aids and restraining straps to avoid out-of-scan-field artefacts and to be able to use an adjusted small reconstruction field later on. For the purpose of including as much of the body as possible in the scan

range along the *z*-axis, the corpse may have to be pulled cranially on the CT table. To ease the process of pulling the body, silicone or glide sprays can be used to impregnate the table's mattress overlay in order to keep the table slippery. From a radiological point of view, the head should be carefully aligned along the body axis with a median position of the nose for a better radiological assessment of the head and neck region. Furthermore, as the number of captured x-ray photons is affected by the size of the penetrated object, the BH artefacts and image noise can be reduced for thoracic and abdominal scans by moving the arms above the head.[14,18] However, the risk of post-mortem alteration and the potentiality of causing post-mortem findings due to body repositioning have to be considered. Therefore, a previous PMCT scan for documentation purposes, without the anatomically adjusted repositioning of the head or the elevation of

the arms, is advisable. Thus, it is recommended to perform a whole-body scan as a basic scan without repositioning the head or elevating the arms in order to detect post-mortem findings due to body repositioning during the PMCT examination. However, a correct symmetric position is always dependent on the extent of rigor mortis and on the condition of the corpse, for instance it may not be possible to place charred bodies in a specific scan position.

The body has to be positioned in the centre of the gantry to obtain the best image quality (Figure 1). Additionally, recent dose-modulation algorithms generally calculate the dose in the centre of the gantry, so careful patient centring is required. First, the body has to be placed in the middle of the table width. Second, the height of the table has to be selected so that the decedent is located in the middle of the gantry to achieve the best image quality. This approach may be

Figure 1. Correct body position in the centre of the gantry is required for the best image quality. In the left column, a spine phantom is positioned almost in the iso-centre of the gantry (a1). In contrast, the right column displays a spine phantom at the edge of the standard scan field (b1). The reconstructions of the spine phantom in the centre of the gantry (a2) exhibit better image quality and significantly less noise compared to the reconstructions of the spine phantom at the edge of the standard scan field (b2). The image quality decreases with the distance to the iso-centre.

difficult for obese or decomposed, bloated bodies, as the skeleton may be in a low position and thus at a greater distance from the iso-centre of the CT gantry due to the extended size of the abdominal girth. The gantry diameter is larger than the scan field-of-view, and this should be taken into account by performing an additional scan with the bloated abdomen beyond the standard scan field but with the skeleton positioned closer to the iso-centre of the CT gantry.

After adequate body preparation and positioning, acquisition parameters (prior to the PMCT scan) and reconstruction parameters (after the PMCT scan) have to be adapted in order to achieve the best quality. Table 1 (acquisition parameters) and Table 2 (reconstruction parameters) display a summary of recommended parameter adjustments for PMCT and their main effects regarding contrast, noise and resolution as well as artefacts.

Acquisition parameters

The acquisition parameter settings have various influences on contrast, noise and spatial resolution. Most of these parameters affect the radiation dose. PMCT is not limited by radiation dose considerations towards the examined person. Therefore, the acquisition parameters can be optimally adjusted to achieve a high image quality.

Tube voltage

The contrast is mainly affected by the x-ray photon energy, which can be adjusted by the x-ray tube voltage settings.[10] An increased tube voltage reduces the contrast, and thus the tube voltage settings should be adjusted according to the radiological interests, which have to be assessed.[19] The user has to select a tube voltage level based on the kilovolt peak.[19,20] Usually,

Table 1. A summary of recommended acquisition parameters for PMCT examinations.

	Parameter settings	Area of application	Main effects regarding contrast, noise, resolution and artefacts
Tube voltage	120 kVp	Standard scans	Compromise between soft tissue and bones regarding the contrast
	140 kVp	Dental scan, metallic objects, skeletal remains	Higher kVp settings decrease beam hardening but decrease the contrast
	100 kVp	PMCTA	Improved visualisation of iodine-contrasted vascular structures but increase beam hardening
Dual energy	80/140 kVp	PMCTA (bone removal), metallic objects, material differentiation, pediatric imaging	Dual-energy bone removal achieves better implementation than single-energy bone removal, mono-energetic reconstructions can be used for the metallic artefact reduction, dual-energy provides attenuation information from different energy levels, dual-energy allows to define an optimised contrast in paediatric imaging
Tube current	300 mAs	Whole body	An increased tube current significantly reduces the noise
	1000 mAs	Head/neck	
	500 mAs	Thorax/abdomen	
	Dose modulation	Whole body (300 reference mAs), thorax/abdomen (500 reference mAs)	Uniform noise occurrence over the whole scan
Detector collimation	0.6 mm	Always recommended	Small collimation size results in a higher spatial resolution and reduces the susceptibility to partial-volume and windmill artefacts
	≤0.4 mm	Dental scan, laryngeal scan	Smallest collimation size is applicable for a small scan length to achieve highest spatial resolution
Pitch	≤0.5	Always recommended	A small dose-independent pitch enables the user to increase the mAs value (which reduces the noise), improves the z-axis resolution and reduces the susceptibility to windmill artefacts
Rotation time	≥1 s	Always recommended	Slow rotation time along with a small pitch increases tube current (thus reduces the noise) and improves the z-axis resolution

The indicated values are guide values, as depending on the CT scanner, the technical capabilities vary.
PCMT: post-mortem computed tomography; PMCTA: PMCT angiography; CT: computed tomography.

Table 2. A summary of recommended reconstruction parameters for PMCT examinations.

	Parameter settings	Area of application	Main effects regarding contrast, noise, resolution and artefacts
Window	Presets	According to the kernel	Windowing enables the variation of the contrast of a slice stack; the window settings can be changed on the reconstructed slice stacks anytime later
Extended CT scale	Selectable via a checkbox	Dental implants, metallic objects	Detailed depictions of tiny surface patterns of metallic foreign objects or dental implants
Reconstruction filter (kernel)	Sharp (hard)	Bones, lung tissue, dental scans, metallic objects	A sharp kernel displays small details or edges for high spatial resolution but increases the noise
	Smooth (soft)	Soft tissue	A smooth kernel decreases the noise and allow for better soft-tissue discrimination
	Medium smooth	Larynx calcification, coronary vessels	Medium-smooth filters are recommended for cartilage structure and calcification
Field of view (FOV)	Downsized to region of diagnostic interest	Always recommended	A small FOV minimises the in-plane voxel size to achieve high in-plane resolution
	Extended FOV	Obese, charred, frozen or bloated bodies	An extended FOV displays body parts beyond the standard maximum FOV, which may otherwise be truncated
Slice thickness	≤2 mm	Whole body	The slice thickness should be adjusted according to the scan length in order to avoid too large data sets and according to the reconstruction FOV to obtain iso-tropic voxels
	≤1 mm	Thorax/abdomen	
	0.6 mm	Head/neck	
Increment	Three-quarters of the slice thickness	Always recommended	A smaller increment smaller than the slice thickness results in overlapping images, which may be more suitable for scrolling through a slice stack

The indicated values are guide values, as depending on the CT scanner, the technical capabilities vary.

120 kVp is used in clinical CT, as it provides the best compromise between imaging the soft tissue and bones,[18] which is also applicable for standard PMCT. Higher kilovolt peak settings reduce BH, with the disadvantage of a lower contrast.[14] Therefore, in PMCT, additional scans with higher kilovolt peak settings (e.g. 140 kVp) should be considered for dental scans or scans with a prosthesis, spondylodesis or osteosynthesis to reduce the metallic artefacts due to dental or metallic implants. Further, high-energy PMCT may also be considered for exhumed bodies or any skeletal remains for best osseous depiction, neglecting soft-tissue contrast. Low kilovolt peak settings (e.g. 100 kVp) in turn improve the visualisation of iodine-contrasted vascular structures in angiographic examinations, as the attenuation of the iodine-containing contrast media will be increased,[21,22] although lower kilovolt peak settings generally increase the susceptibility for BH.[19,22] If available, dual-energy technology offers special advantages for metallic artefact reduction[23,24] or CT angiography, for example regarding bone removal.[25,26] Using dual-energy CT (DECT), the different contrast behaviours of iodine according to the energy spectrum can be used to highlight the contrast media optimally and differentiate it from the bones or soft tissue.[27]

Regarding post-processing, dual-energy bone removal achieves better implementation than single-energy bone removal.[27]

DECT

DECT uses two different polychromatic x-ray spectrums for acquiring measurements of the same slice stack. Several vendors offer different DECT technologies, such as utilising two scanner-integrated x-ray sources (dual-source DECT), performing twin-beam single-source CT using gold (low-energy) and tin (high-energy) pre-filters, applying rapid kilovolt peak switching associated with a gemstone scintillator detector and using a dual-layer detector or alternating kilovolt peak settings with each rotation (dual-scan single source).[28–30] The DECT data of two different energy spectra allow for extrapolating mono-energetic reconstructions at different selectable kiloelectron volt levels.[31] The term 'kiloelectron volt' is usually used in DECT, as it describes the mean energy of the x-ray spectrum.[28] Those mono-energetic reconstructions can be used for the metallic artefact reduction of dental restorations[24] or any orthopaedic implants,[32] which is significant for radiological identification. DECT may

also be considered in post-mortem paediatric imaging to define an optimised contrast by selecting the desired kiloelectron volt value using mono-energetic reconstruction. Further, DECT provides attenuation information from different energy levels, which shows potential in forensic imaging, for example to differentiate foreign bodies,[33] to discriminate drugs in body packs[34] and to discriminate ferromagnetic from non-ferromagnetic projectiles for magnetic resonance imaging.[35]

Tube current

The use of an increased radiation dose to obtain a high image quality in PMCT is frequently mentioned in the literature,[18,36,37] although it is intended to increase the tube current (tube amperage). The tube current is essential for the intensity of the noise (Figure 2). Thus, an increased tube current significantly reduces the noise.[14,38] The tube current level is limited by the scan length. Thus, in addition to a whole-body scan, separate head/neck and thorax/abdomen scans with higher milliampere-seconds settings are recommended in the literature.[18] The tube current is vendor-dependent, indicated together with the slice scan time as the tube current-time product using milliampere-seconds.[11] The radiation dose is directly proportional to the administered tube current-time product based on the milliampere-seconds value.[10,11,20] Image acquisition using increased milliampere-seconds settings is the foremost advantage in PMCT compared to clinical CT.

Dose modulation

The initial purpose of automatic exposure control or dose-modulation techniques is decreasing the radiation dose by the modulation of the tube current for the regional body anatomy,[39–41] which at first glance may be considered unnecessary in PMCT. However, the noise will be adjusted as the tube current is adjusted using dose-modulation techniques,[39,40] which results in more uniform noise occurrence over the whole scan. Several vendors provide automatic exposure control with different technical approaches. A uniform noise distribution is also beneficial for PMCT. By selecting a high initial reference milliampere-seconds value, the overall administered radiation dose can be increased to improve the image quality combined with dose-modulation, resulting in more uniform noise. However, the use of dose modulation for a combined scan of the head and neck regions may result in unwanted dose distribution, as significantly less exposure to the head region compared to that to the lower neck will occur due to the shoulders. Thus, dose
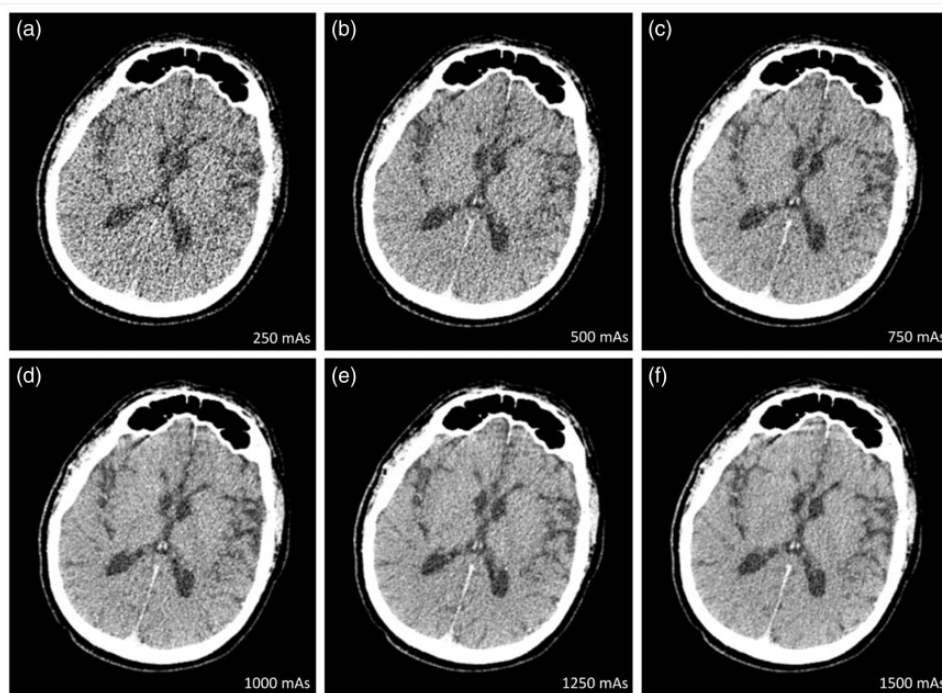
Figure 2. The difference between 250 mAs and 1500 mAs with respect to the presence of noise is impressively illustrated by a head scan. The different stages of the milliampere-seconds settings show the reduction of the image noise from 250 mAs to 1500 mAs. When increasing the tube current, the effect of the noise reduction will approach a limit. As shown, the difference in image quality is less pronounced between values >1000 mAs.

modulation for the head and neck region may not result in the desired image quality for the head region. Furthermore, it must be of concern that paediatric cases as well as heavily decomposed, mummified or charred bodies may initially be low dosed due to the automatic dose modulation, so the reference milliampere-seconds value has to be kept in sight and adjusted if necessary.

Detector collimation

The user-selected detector collimation affects the spatial resolution[42] and defines the reachable, minimal slice thickness (SL).[3] A small collimation size is always recommended in any PMCT examination, as it results in a higher spatial resolution[11] and reduces the susceptibility to PV[13] and WM[3,15] artefacts, with the drawback of increased noise.[43] If necessary, after the PMCT scan, thicker slices can be reconstructed from the same raw data, or the SL of thin-sliced reconstructions can be increased using dedicated post-processing software.[44]

Pitch and rotation time

Helical scanning means that the table moves through the gantry during x-ray rotation with the constant acquisition of data,[3] which is described by the pitch. However, two different definitions of the pitch factor are used by vendors[38]: detector pitch[3,43] and volume pitch.[43] Detector pitch is defined as the table feed per rotation of the x-ray tube divided by the total detector collimation and therefore is the total number of all simultaneously active detector rows.[3,44] In contrast, volume pitch excludes the number of active detector rows, and thus it is defined as the table feed per rotation of the x-ray tube divided by the detector collimation of a single row.[45] The pitch factor along with the rotation time affects the image quality in different ways. In general, a larger pitch may cause gaps between x-ray beam trajectories and thus gaps in the data acquisition,[1,15,43] which results in a lower image resolution. In contrast, a small pitch enables more x-ray photons to contribute to the calculation of an image due to the overlapping beam trajectories.[1,2,15,43] Thus, a small pitch automatically affects the tube current, and a (vendor-dependent) dose-independent pitch enables the user to increase the milliampere-seconds value, as a higher maximum dose is applicable. Additionally, a reduced rotation time allows for increasing the tube current. Further, a small pitch improves the *z*-axis resolution, as fewer anatomical changes are located within the range of interpolation.[2,46] Major changes of the anatomical structures within the interpolation range may cause WM artefacts.[3,15,16] So, apart from the narrowed collimation, the likelihood of these WM artefacts is also decreased by using a small pitch.[3,15]

Reconstruction parameters

After the data acquisition, a raw data set is computed that is the basis for further image reconstructions. These raw data sets are used for the reconstruction of slice stacks in accordance with the radiological interests for the subsequent evaluation and interpretation of the examination. From the same raw data set, several slice stacks with different reconstruction parameters can be computed.[15] However, raw data sets are not stored due to their size, and so they must remain on the CT workstation until the reconstructed slice stacks have been reviewed in the event that further special reconstructions are desired.

Window settings

The measured attenuation values are converted into a calculated integer (CT number) indicated in HU.[1] The attenuation coefficient of water is used as a reference for the calculation of each CT number. Thus, the CT number for water is approximately 0, for air approximately –1000 and for dense cortical bone approximately +1000[1] for a commonly used x-ray beam energy value (120 kVp). The CT numbers are displayed as grey levels.[1] The number of grey levels that could be visually discernible is limited.[1,2] Thus, the range of grey levels has to be adapted to the amount of CT numbers. Otherwise, for example, soft tissue would not differ visually, as the CT numbers of soft tissue have too little variation and will not be visually distinguishable. Windowing allows for allocating grey levels in the image based on the window width and window centre for diagnostic interests (e.g. bony structures, soft tissue or lung structure). Thus, the range of displayed CT numbers can be subsequently adjusted to achieve the desired image contrast for the radiological interests by windowing. However, the CT number (and the HU value) is fixed independently from the windowing. The window width defines the quantity of CT numbers that should be displayed.[1] This means that the window width sets the range of CT numbers in which all grey levels are covered, and any values outside this range are displayed either in black (lower values) or white (higher values). The window level or window centre defines the central point of the window width along the HU scale[1] and thus the central CT number value over which the window width ranges.[2] Thus, windowing enables the variation of the contrast of a slice stack. Usually, CT scanners offer various specific window presets for image review and reconstruction,[2] but the window

settings can be individually and interactively set and changed on the reconstructed slice stacks anytime later.

Extended CT scale

In addition to the discussed reconstruction variations, the extended CT scale (ECTS) technique extends the maximum window range of a reconstructed slice stack tenfold, as the CT numbers will be scaled down by a factor of 10, which in turn expands the CT scale from approximately 4000 HU to approximately 40,000 HU.[47] Thus, this approach does not necessarily reduce the artefacts, but it enables better image contrast for a hyper-dense material by windowing into a higher range. Therefore, a more accurate depiction of materials with high radio-opacity, such as metallic objects, can be achieved. Reconstructions using ECTS enable detailed depictions of tiny surface patterns of metallic foreign objects for a clear identification of the objects.[48] The precise depiction of foreign bodies is a major task of PMCT, for example to distinguish between a common coin and a button battery is of substantial importance.[48,49] Furthermore, the ability to use ECTS yields advantages for dental identification, as dental implants are depicted more accurately[50] on PMCT for better comparison to ante-mortem x-rays. Additionally, high-density objects can be discriminated based on the possibility of measuring higher HU values (Figure 3) at different kilovolt peak settings.[33,35]
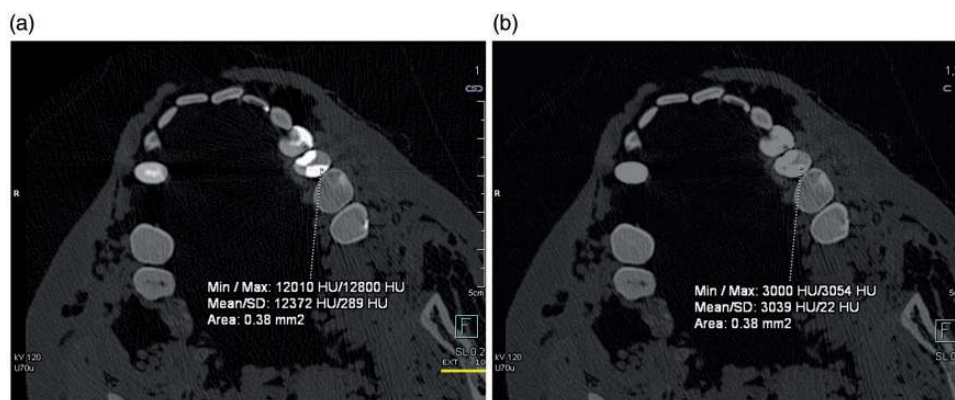
Reconstruction filter

A reconstruction filter (kernel), actually an algorithm, defines a compromise between image noise and spatial resolution by blurring the image, as blurring reduces the image noise.[1] Regarding the radiological interests, soft tissue will usually not need well-defined edges for diagnostics. Thus, reduced image noise is preferred over spatial resolution. Smooth (soft) filters are applied that decrease the noise and allow for better soft tissue discrimination.[1,2,11] In contrast, displaying small details or edges of bones requires a high spatial resolution, and the image noise interferes less with these images, and hence sharp (hard) filters are applied.[1,2,11] Thus, smooth (soft) filters for soft tissue and sharp (hard) filters for bones and lungs have to be used for separate reconstructions. Of particular note is the use of medium-smooth filters for the laryngeal region[18] in cases of strangulation due to the cartilage structure. Further, medium-smooth filters are recommended to display the calcification of the coronary vessels better.[51] Additionally, special filters have been developed for defined anatomical regions, for example for a separate head scan.

Field of view

Reconstructed sectional slices are three-dimensional (3D), and the picture elements are cuboid-shaped. Thus, a reconstructed slice consists of volume elements, so-called voxels. The size of the voxel depends on the matrix size and the size of the reconstructed field of view (FOV; *x*- and *y*-axis), and the slice thickness (*z*-axis; Figure 4). Thin-sliced image stacks with almost isotropic voxels are essential for further high-resolved 3D reconstructions. Voxels are considered isotropic if their extension is equal in all three spatial axes.[3,15] Primarily, the *z*-axis resolution is limited by the detector collimation and spiral interpolation algorithm (*z*-filtering).[15] Therefore, the *z*-axis resolution or the smallest possible slice thickness has to be mainly defined by the setting of the acquisition parameters, whereas the in-plane resolution is mainly defined by the size of the reconstructed FOV. The general term

Figure 3. Apart from depicting high-density objects (e.g. dental implants) more accurately, reconstructions using (a) the extended computed tomography scale (ECTS) enable HU measurements over a higher range compared to (b) standard CT reconstructions. ECTS (in combination with multi-scan, multi-energy post-mortem computed tomography) enables the discrimination of high-density objects to a certain degree.

'field of view' relates to the size of the scanning FOV (sFOV) in axial orientation or to the size of the reconstructed FOV (rFOV) in axial orientation. The extension of the sFOV is described by a diameter, as it is circular according to the rotational movement of the x-ray tube, whereas the rFOV is usually square according to the image matrix. Most types of CT scanners allow for adjusting the rFOV after the CT scan, and the size of the sFOV does not affect the in-plane resolution, while the size of the rFOV significantly does. Usually, the maximum rFOV is 500×500 mm
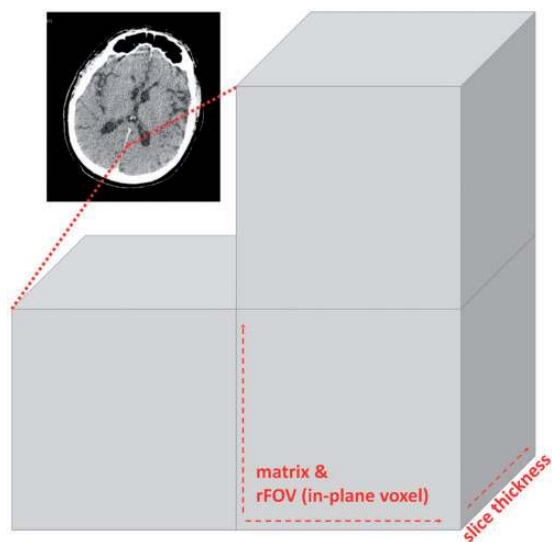
according to the maximum sFOV (500 mm), with the centre of the gantry as its centre point. Within the range, the rFOV can be downsized and adjusted to any region of interest to minimise the in-plane voxel size and to achieve high in-plane resolution. Special reconstruction algorithms also allow the rFOV to be extended beyond its initial maximum size with the restriction of less accuracy in the outer range.[2] Such an enlarged rFOV is described as an extended FOV (eFOV). A separate reconstruction using eFOV is recommended for obese decedents to display (with lower quality) the upper extremities on the body, which may otherwise be outside the sFOV and hence truncated.[52] Using eFOV is also suitable for charred, frozen or bloated, decomposed bodies, which may not fit into the standard range of the rFOV.[18]

Slice thickness and increment

Current CT technology allows for reconstructing several slice stacks with different slice thickness values from the same raw data set.[21] The slice thickness is usually abbreviated as SL. To reconstruct overlapping images, which may be more suitable for scrolling through a slice stack, the increment should be smaller than the SL (three-quarters of SL). To reduce the noise in the slice stacks along the *z*-axis, SL can be increased (Figure 5), which may be suitable for soft-tissue diagnostics. However, reconstructions from the raw data set with thick slices are not necessarily required, as the SL of thin-sliced image stacks can also be increased during viewing directly on a dedicated workstation, according to the diagnostic interests. The main advantages of the thin-sliced image stacks are their high
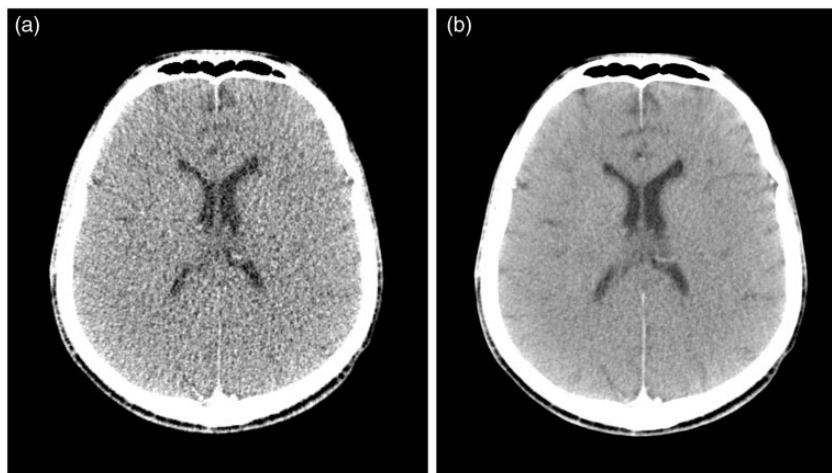

Figure 4. Each CT slice (a) consists of small voxels (b). The size of the voxel is dependent on the matrix, the reconstructed field of view (FOV; in-plane resolution) and the slice thickness (z-axis resolution).

Figure 5. The slice thickness (SL) of thin-sliced reconstructions can be increased by using a dedicated workstation for soft-tissue diagnostics. Thin-sliced (SL: 0.6 mm) reconstructions (a) exhibit increased noise compared to thick-sliced (SL: 3 mm) reconstructions (b), but they provide the basis for further high-resolution three-dimensional reconstructions.

spatial resolution, possibilities for almost iso-voxel 3D reconstructions and reduction of partial volume artefacts.[38] An adequate SL for an appropriate scan length is recommended, dependent on the possibilities of the CT scanner or detector collimation. Along with an adjusted rFOV, thin-sliced image stacks consist of small voxels, which provide the basis for further high-resolution 3D reconstructions.[21,53] Regarding the radiological identification of decedents,[54] it is recommended to perform additional thin-sliced reconstructions of prostheses or anatomical characteristics with an adjusted small rFOV. For the comparison of ante-mortem data sets and PMCT data sets, the PMCT reconstructions can be matched with ante-mortem data sets using multi-planar reformation for a side-by-side comparison or even image fusion, depending on the applied workstation.[54–56]

3D post-processing

The reconstructed slice stacks allow for computing volumetric reconstructions. High-quality 3D post-processing requires thin-sliced image stacks reconstructed in an adjusted small rFOV to achieve the highest resolution in all three spatial planes.[21,53]

Multi-planar reformation/reconstruction (MPR) allows one to compute and display two-dimensional slices with a defined thickness from the whole volumetric data set in any desired plane.[57] Thus, MPRs can be used for reconstructions in any arbitrary plane to display coronal, sagittal or any anatomically aligned oblique plane in high resolution.[43] Currently, MPR allows for the real-time viewing of an already reconstructed slice stack in any arbitrary orientation. However, MPR real-time viewing requires thin-sliced image stacks with almost isotropic voxels for high resolution without step artefacts.[2,53] If desired, the slice thickness can be modified during MPR real-time viewing to reduce the noise. A very thick slice of several centimetres ('ray sum') may resemble a conventional x-ray image, and this method is frequently used for comparison between ante-mortem x-rays and PMCT reconstructions for identification purposes.[54] Even non-linear or curved reconstructions are feasible[2,58] and can be used to reconstruct a panoramic, tomographic image for comparison with an ante-mortem orthopantogram for radiological identification based on dental structures and implants.[54,59] Further curved MPR can be used for the purpose of displaying curved anatomical structures in a planar view, for example contrast-enhanced vessels.[60] Generally, MPR reconstructions are advantageous in PMCT, especially if a body could not be positioned correctly in the supine position, for example charred or frozen bodies.

The maximum intensity projection (MIP) depicts the maximum intensity of all data points from the entire volume (Figure 6(a) and (b)). This technique can be described as volumetric visualisation, as the reconstructed
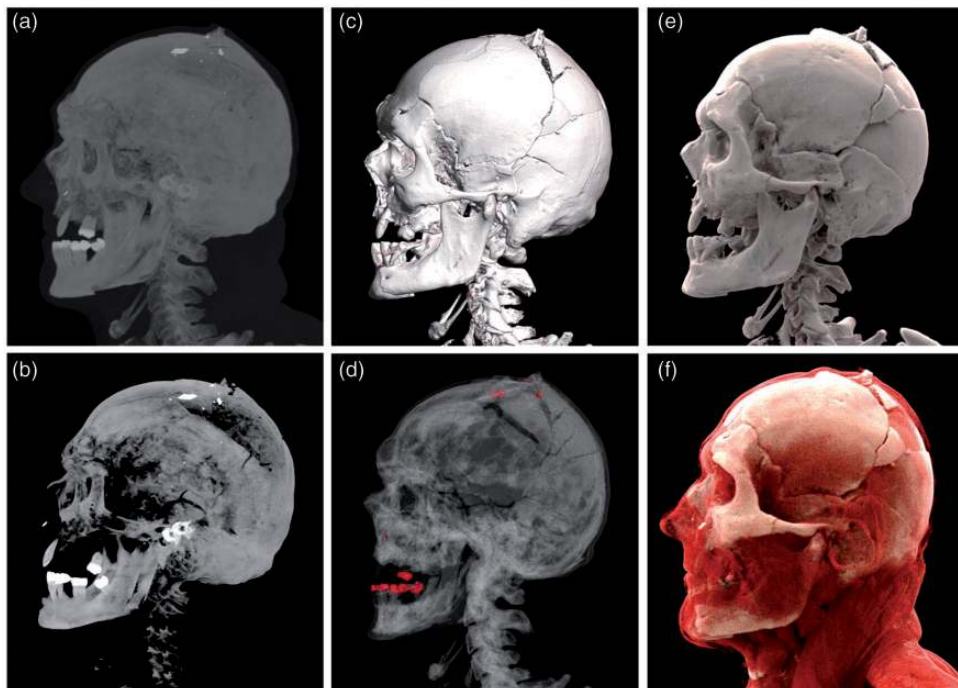
Figure 6. Three-dimensional reconstructions of a headshot displayed as maximum intensity projection ((a) and (b)), volume rendering ((c) and (d)) and cinematic rendering ((e) and (f)) using two different presets for each technique.

MIP can be rotated. Thus, the viewpoint is freely selectable.[58] Usually, MIPs are standard tools in CT-angiography for the better visualisation of enhanced vessels or better differentiation between contrast agents and plaques.[61] Further, this technique can be used to display an overview of the bones, metallic implants or calcifications.[58]

Shaded surface display (SSD) can be described as a surface modelling,[57] as it is based on rendering using a mosaic of connected polygons.[58] Therefore, this visualisation technique allows for the surface of an object to be displayed based on HU threshold segmentation, which can be used to identify surface injuries.

However, SSD has mostly been replaced by volume rendering (VR), as SDD could not be established in clinical radiology.[58] VR means that a specific colour and opacity will be assigned to each voxel based on predefined attenuations related to different types of tissue (Figure 6(c) and (d)). Furthermore, a projection technique based on simulated light rays is used, so that each voxel will alternate the colour of the light rays according to its assigned colour and opacity.[62] As different trapezoids are used for tissue separation, the rendered volume can be adjusted based on the radiological interests by changing the gradient of the trapezoid.[62] Vendors offer several presets for different illustrations. This visualisation technique can be used to display the anatomy and pathology in a 3D image, which can be used for real-time viewing. A more recent visualisation technique, comparable to VR, is cinematic rendering (CR). CR may provide further advantages, as this technique computes photorealistic depictions[62–64] (Figure 6(e) and (f)). VR and CR play a particular role in postmortem radiology, as they may provide a more understandable overview of pathologies and injuries for district attorneys and judges.[65] Compared to autopsy, this visualisation technique may provide the in situ depiction of injuries, for example multiple head fractures,[66] which may fall apart during autopsy due to fragmentation, or it may enhance the illustration of the gas distribution across the entire body. For the illustration of the findings, it must be decided whether to use thin-sliced data sets with a smooth (soft) or a sharp (hard) reconstruction filter for the VR or CR reconstructions.

Summary

This article describes the CT principles from acquiring image data to post-processing and recommends parameter adjustments to obtain the best image quality in PMCT. The summary in Tables 1 and 2 provides a quick introduction to PMCT with high image quality and provides a guide for the performance of PMCT in morgues, clinical facilities or private practices.

References

1. Goldman LW. Principles of CT and CT technology. *J Nucl Med Technol* 2007; 35: 115–128.
2. Adamson KL and Walmsley BR. Basics of CT. In: Fogelman I, Gnanasegaran G and van der Wall H (eds) *Radionuclide and hybrid bone imaging*. Heidelberg: Springer, 2012, pp.135–148.
3. Goldman LW. Principles of CT: multislice CT. *J Nucl Med Technol* 2008; 36: 57–68.
4. Farkash U, Scope A, Lynn M, et al. Preliminary experience with postmortem computed tomography in military penetrating trauma. *J Trauma* 2000; 48: 303–308; discussion 308–309.
5. Thali MJ, Yen K, Schweitzer W, et al. Virtopsy, a new imaging horizon in forensic pathology: virtual autopsy by postmortem multislice computed tomography (MSCT) and magnetic resonance imaging (MRI) – a feasibility study. *J Forensic Sci* 2003; 48: 386–403.
6. Levy AD, Abbott RM, Mallak CT, et al. Virtual autopsy: preliminary experience in high-velocity gunshot wound victims. *Radiology* 2006; 240: 522–528.
7. Ljung P, Winskog C, Persson A, et al. Full body virtual autopsies using a state-of-the-art volume rendering pipeline. *IEEE Trans Vis Comput Graph* 2006; 12: 869–876.
8. Grabherr S, Djonov V, Friess A, et al. Postmortem angiography after vascular perfusion with diesel oil and a lipophilic contrast agent. *Am J Roentgenol* 2006; 187: W515–W523.
9. Christe A, Flach P, Ross S, et al. Clinical radiology and postmortem imaging (Virtopsy) are not the same: specific and unspecific postmortem signs. *Leg Med* 2010; 12: 215–222.
10. Huda W, Scalzetti EM and Levin G. Technique factors and image quality as functions of patient weight at abdominal CT. *Radiology* 2000; 217: 430–435.
11. Goldman LW. Principles of CT: radiation dose and image quality. *J Nucl Med Technol* 2007; 35: 213–225.
12. Barrett JF and Keat N. Artifacts in CT: recognition and avoidance. *Radiographics* 2004; 24: 1679–1691.

13. Raupach R and Flohr T. Artifacts in MSCT. In: Bruening R, Kuettner A, Flohr T (eds) *Protocols for multislice CT*. Heidelberg: Springer, 2006, pp.41–47.

14. Boas FE and Fleischmann D. CT artifacts: causes and reduction techniques. *Imaging Med* 2012; 4: 229–240.

15. Flohr TG, Schaller S, Stierstorfer K, et al. Multi-detector row CT systems and image-reconstruction techniques. *Radiology* 2005; 235: 756–773.

16. Kyriakou Y, Kachelrieß M, et al. Impact of the z-flying focal spot on resolution and artifact behavior for a 64-slice spiral CT scanner. *Eur Radiol* 2006; 16: 1206–1215.

17. Ohnesorge B, Flohr T, Schwarz K, et al. Efficient correction for CT image artifacts caused by objects extending outside the scan field of view. *Med Phys* 2000; 27: 39–46.

18. Flach PM, Gascho D, Schweitzer W, et al. Imaging in forensic radiology: an illustrated guide for postmortem computed tomography technique and protocols. *Forensic Sci Med Pathol* 2014; 10: 583–606.

19. Huda W. Dose and image quality in CT. *Pediatr Radiol* 2002; 32: 709–713.

20. McNitt-Gray MF. AAPM/RSNA physics tutorial for residents: topics in CT. *Radiographics* 2002; 22: 1541–1553.

21. Prokop M. Multislice CT: technical principles and future trends. *Eur Radiol* 2003; 13: 3–13.

22. Marin D, Nelson RC, Schindera ST, et al. Low-tube-voltage, high-tube-current multidetector abdominal CT: improved image quality and decreased radiation dose with adaptive statistical iterative reconstruction algorithm – initial clinical experience. *Radiology* 2009; 254: 145–153.

23. Bamberg F, Dierks A, Nikolaou K, et al. Metal artifact reduction by dual energy computed tomography using monoenergetic extrapolation. *Eur Radiol* 2011; 21: 1424–1429.

24. Stolzmann P, Winklhofer S, Schwendener N, et al. Monoenergetic computed tomography reconstructions reduce beam hardening artifacts from dental restorations. *Forensic Sci Med Pathol* 2013; 9: 327–332.

25. Persson A, Jackowski C, Engström E, et al. Advances of dual source, dual-energy imaging in postmortem CT. *Eur J Radiol* 2008; 68: 446–455.

26. Deng K, Liu C, Ma R, et al. Clinical evaluation of dual-energy bone removal in CT angiography of the head and neck: comparison with conventional bone-subtraction CT angiography. *Clin Radiol* 2009; 64: 534–541.

27. Morhard D, Fink C, Graser A, et al. Cervical and cranial computed tomographieß angiography with automated bone removal: dual energy computed tomography versus standard computed tomography. *Invest Radiol* 2009; 44: 293–297.

28. Johnson TRC. Dual-energy CT: general principles. *Am J Roentgenol* 2012; 199: S3–S8.

29. Coupal TM, Mallinson PI, Gershony SL, et al. Getting the most from your dual-energy scanner: recognizing, reducing, and eliminating artifacts. *Am J Roentgenol* 2015; 206: 119–128.

30. McCollough CH, Leng S, Yu L, et al. Dual- and multi-energy CT: principles, technical approaches, and clinical applications. *Radiology* 2015; 276: 637–653.

31. Apfaltrer P, Sudarski S, Schneider D, et al. Value of monoenergetic low-kV dual energy CT datasets for improved image quality of CT pulmonary angiography. *Eur J Radiol* 2014; 83: 322–328.

32. Filograna L, Magarelli N, Leone A, et al. Value of mono-energetic dual-energy CT (DECT) for artefact reduction from metallic orthopedic implants in post-mortem studies. *Skeletal Radiol* 2015; 44: 1287–1294.

33. Ruder TD, Thali Y, Bolliger SA, et al. Material differentiation in forensic radiology with single-source dual-energy computed tomography. *Forensic Sci Med Pathol* 2013; 9: 163–169.

34. Leschka S, Fornaro J, Laberke P, et al. Differentiation of cocaine from heroine body packs by computed tomography: impact of different tube voltages and the dual-energy index. *J Forensic Radiol Imaging* 2013; 1: 46–50.

35. Winklhofer S, Stolzmann P, Meier A, et al. Added value of dual-energy computed tomography versus single-energy computed tomography in assessing ferromagnetic properties of ballistic projectiles: implications for magnetic resonance imaging of gunshot victims. *Invest Radiol* 2014; 49: 431–437.

36. Saunders SL, Morgan B, Raj V, et al. Post-mortem computed tomography angiography: past, present and future. *Forensic Sci Med Pathol* 2011; 7: 271–277.

37. Palmiere C, Binaghi S, Doenz F, et al. Detection of hemorrhage source: the diagnostic value of post-mortem CT-angiography. *Forensic Sci Int* 2012; 222: 33–39.

38. Rydberg J, Buckwalter KA, Caldemeyer KS, et al. Multisection CT: scanning techniques and clinical applications. *Radiographics* 2000; 20: 1787–1806.

39. Kalra MK, Maher MM, Toth TL, et al. Techniques and applications of automatic tube current modulation for CT. *Radiology* 2004; 233: 649–657.

40. McCollough CH, Bruesewitz MR and Kofler JM. CT dose reduction and dose management tools: overview of available options. *Radiographics* 2006; 26: 503–512.

41. Söderberg M and Gunnarsson M. The effect of different adaptation strengths on image quality and radiation dose using Siemens Care Dose 4D. *Radiat Prot Dosimetry* 2010; 139: 173–179.

42. Hu H. Multi-slice helical CT: scan and reconstruction. *Med Phys* 1999; 26: 5–18.

43. Prokop M. General principles of MDCT. *Eur J Radiol* 2003; 45: S4–S10.

44. Kopp AF, Klingenbeck-Regn K, Heuschmid M, et al. Multislice computed tomography: basic principles and clinical applications. *Electriomedica* 2000; 68: 94–105.

45. Prokop M. General principles of MDCT. *Eur J Radiol* 2003; 45: S4–S10.

46. Flohr TG, Stierstorfer K, Ulzheimer S, et al. Image reconstruction and image quality evaluation for a 64-slice CT scanner with z-flying focal spot. *Med Phys* 2005; 32: 2536–2547.

47. Coolens C and Childs PJ. Calibration of CT Hounsfield units for radiotherapy treatment planning of patients with metallic hip prostheses: the use of the extended CT-scale. *Phys Med Biol* 2003; 48: 1591.

48. Gascho D, Gentile S, Bolliger SA, et al. Charon's coins. *Forensic Sci Med Pathol* 2016; 12: 384–387.

49. Marom T, Goldfarb A, Russo E, et al. Battery ingestion in children. *Int J Pediatr Otorhinolaryngol* 2010; 74: 849–854.

50. Jackowski C, Lussi A, Classens M, et al. Extended CT scale overcomes restoration caused streak artifacts for dental identification in CT-3D color encoded automatic discrimination of dental restorations. *J Comput Assist Tomogr* 2006; 30: 510–513.

51. Renker M, Nance JW, Schoepf UJ, et al. Evaluation of heavily calcified vessels with coronary CT angiography: comparison of iterative and filtered back projection image reconstruction. *Radiology* 2011; 260: 390–399.

52. Hsieh J, Chao E, Thibault J, et al. A novel reconstruction algorithm to extend the CT scan field-of-view. *Med Phys* 2004; 31: 2385–2391.

53. Ford JM and Decker SJ. Computed tomography slice thickness and its effects on three-dimensional reconstruction of anatomical structures. *J Forensic Radiol Imaging* 2016; 4: 43–46.

54. Hatch GM, Dedouit F, Christensen AM, et al. RADid: a pictorial review of radiologic identification using postmortem CT. *J Forensic Radiol Imaging* 2014; 2: 52–59.

55. Ruder TD, Brun C, Christensen AM, et al. Comparative radiologic identification with CT images of paranasal sinuses – development of a standardized approach. *J Forensic Radiol Imaging* 2016; 7: 1–9.

56. Heimer J, Gascho D, Gentile S, et al. Antemortem identification by fusion of MR and CT of the paranasal sinuses. *Forensic Sci Med Pathol*. Epub ahead of print May 20 2017. DOI: 10.1007/s12024-017-9873-6.

57. Zhang Q, Eagleson R and Peters TM. Volume visualization: a technical overview with a focus on medical applications. *J Digit Imaging* 2011; 24: 640–664.

58. Lundström C, Persson A, Ross S, et al. State-of-the-art of visualization in post-mortem imaging. *APMIS* 2012; 120: 316–326.

59. Brough AL, Morgan B and Rutty GN. The basics of disaster victim identification. *J Forensic Radiol Imaging* 2015; 3: 29–37.

60. Achenbach S, Moshage W, Ropers D, et al. Curved multiplanar reconstructions for the evaluation of contrast-enhanced electron beam CT of the coronary arteries. *Am J Roentgenol* 1998; 170: 895–899.

61. Prokop M, Shin HO, Schanz A, et al. Use of maximum intensity projections in CT angiography: a basic review. *Radiographics* 1997; 17: 433–451.

62. Dappa E, Higashigaito K, Fornaro J, et al. Cinematic rendering – an alternative to volume rendering for 3D computed tomography imaging. *Insights Imaging* 2016; 7: 849–856.

63. Ebert LC, Flach P, Schweitzer W, et al. Forensic 3D surface documentation at the Institute of Forensic Medicine in Zurich – workflow and communication pipeline. *J Forensic Radiol Imaging* 2016; 5: 1–7.

64. Fellner FA. Introducing cinematic rendering: a novel technique for post-processing medical imaging data. *J Biomed Sci Eng* 2016; 9: 170–175.

65. Ampanozi G, Zimmermann D, Hatch GM, et al. Format preferences of district attorneys for post-mortem medical imaging reports: understandability, cost effectiveness, and suitability for the courtroom: a questionnaire based study. *Leg Med* 2012; 14: 116–120.

66. Flach PM, Thali MJ and Germerott T. Times have changed! Forensic radiology – a new challenge for radiology and forensic pathology. *Am J Roentgenol* 2014; 202: W325–W334.