
Future Developments for MSCT

Patrik Rogalla

Contents

1	Speed of Scanning	16
1.1	Clinical Demand.....	16
1.2	Technological Development.....	16
2	Spatial Resolution	17
2.1	Clinical Demand.....	17
2.2	Technological Development.....	17
3	Temporal Resolution	18
3.1	Clinical Demand.....	18
3.2	Technological Development.....	18
4	Image Quality	18
4.1	Clinical Demand.....	18
4.2	Technological Development.....	19
5	Radiation Dose	20
5.1	Clinical Demand.....	20
5.2	Technological Development.....	20
6	And Even More	21
7	Summary	22
	References	23

Abstract

Development and engineering are continually progressing and it would not be the first time in the evolution of computed tomography that some believe we have reached the zenith. Radiologists see technical improvements year after year Chen et al. (Med Phys 38(2):584–588, 2011), sometimes even quantum leaps and breakthroughs in all fields including data acquisition, image processing and post-processing Liu et al. (Radiology 253(1):98–105, 2009). Hounsfield's CT prototype featured a rotation time of about 300 s with a maximum image matrix of 80×80 pixels Hounsfield (Br J Radiol 68(815): H166–H172, 1995). In comparison, today's 320-slice scanner delivers rotation times of 350 ms and an image matrix of 512×512 pixels Rogalla et al. (Radiol Clin North Am 47(1):1–11, 2009). The pace is breathtaking and technological advances appear limitless if there was not radiation exposure that sets clear boundaries to uncontrolled expansion or unrestrained utilisation of CT in humans. However, priorities regarding the direction of development are laid out clearly: more diagnostic information with less radiation in a shorter scanning time and higher resolution. We may look into a crystal ball in an attempt to predict the future, but let us rather take a step-wise approach to illustrate where technological improvements are clinically desired and where short- and long-term advances can be expected.

P. Rogalla (✉)
Department of Medical Imaging, University of Toronto,
Toronto, ON M5G 2N2, Canada
e-mail: patrik.rogalla@uhn.ca

1 Speed of Scanning

1.1 Clinical Demand

Speed of scanning reflects several aspects during the examination procedure. It might refer to the overall duration of diagnostic procedure in its entirety; it might simply reflect body coverage per time, or even refer to temporal resolution both in-plane (i.e., within a given axial slice) and through-plane (i.e., larger body coverage per time).

– Overall duration

Most abdominal imaging protocols in CT are driven by the need to capture specific perfusion phases, thus the overall duration of a CT examination is primarily determined by physiology. Faster scanning would not necessarily result in shorter procedure times.

– Body coverage per rotation time

Clinical necessity to increase body coverage per unit time predominantly arises from the fact that most CT images are acquired during breath-hold in order to reduce motion artefacts. Faster CT scanners achieve greater anatomical coverage during a single breath-hold period. Modern 64-slice scanners need only a few seconds to complete an abdominopelvic CT, however, combined chest, abdomen and pelvic protocols may still exceed the possible breath-hold length for some patients. In addition, new CT applications in perfusion imaging greatly benefit from faster and wider coverage no matter whether helical or axial scanning modes are used.

– In-plane and through-plane (Z-axis) temporal resolution

Dramatic improvements in through-plane resolution have been achieved with the advent of helical scanning technique, but the hunger for further increases in in-plane resolution is mainly driven by cardiac imaging in need of short exposure times to allow motion free images. For abdominal applications, in-plane temporal resolution plays a less important role with the exception of motion artefact reduction in patients who cannot hold their breath or physically move during data acquisition.

1.2 Technological Development

Most modern CT scanners revolve the gantry at least twice per second. It seems appealing to have faster gantry revolving times, however centrifugal forces

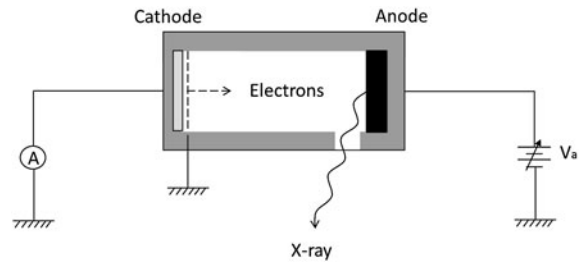


Fig. 1 Simplified principle of carbon nanotubes (CNT). CNTs offer unique advantages for a new generation of X-ray tubes including low power consumption, long lifetime and pulse capability

physically set mechanical limits given current gantry and more importantly, X-ray tube designs. G-forces already exceed 30 g in some current CT scanners. Smaller and lighter X-ray generating tubes are a prerequisite for faster gantry revolving times. It is hard to predict, but minimal gantry revolving times of 150–200 ms may well be a physical limit given today's system components. Such revolving times combined with dual-source technology will lower the exposure window for an axial slice (often inaccurately referred to as in-plane temporal resolution) to less than 40 ms, likely sufficient for virtually all potential cardiac applications.

Another problem becomes more evident with faster scan speeds: the shorter the exposure, the higher the necessary tube output in order to ensure sufficient photon flux on the detector side. Maximum tube power ranges around 100 kW in most CT current scanners. Distributing the energy to two focal spots is current practice and predominantly designed to increase spatial resolution; at the same time, the heat is distributed to a larger area on the anode that helps prevent heat damage to the tube.

Will more tubes, maybe a triple-source CT, solve the problem of limited temporal resolution? Likely not. Unless alternative sources of X-rays such as carbon nanotubes (Fig. 1) will emit enough power sufficient for clinical use, it appears unlikely that more than two conventional tubes (based on a heated tungsten filament) will find space within a given gantry geometry, not to mention subsequent issues with detector position and respective configuration, anti-scatter grids and many more. Inverse geometry CT technology composed of a large distributed X-ray source with an array of discrete electron emitters and focal spots is believed to carry the potential

breakthrough in volumetric coverage, dose efficiency, temporal and spatial resolution (Mazin et al. 2007).

Faster helical scanning—including dual source helical scanning—has been a quantum leap towards rapid longitudinal coverage, albeit a small “overall” enhancement. Apart from the fact that patient acceleration and deceleration may set boundaries to further increase in table speed, faster coverage is beneficial in recently developed scanning techniques for perfusion imaging of the abdomen. The so-called “shuttle mode” enables coverage of larger body areas for perfusion imaging not limited to the effective detector width (Okada et al. 2011).

Sixteen centimetre effective detector coverage represents the current pinnacle in area detector CT and allows for complete organ imaging without table motion in less than 350 ms. Further increases in volumetric coverage inevitably result in increased scattered radiation, cone-beam artefacts, heel effect and over beaming. It remains to be determined where the acceptable boundaries are, however, 20–25 cm coverage appear reasonable and would relevantly reduce scanning time. The entire abdomen, for instance, could be examined using a single gantry rotation.

2 Spatial Resolution

2.1 Clinical Demand

Higher spatial resolution appears to be key to success in CT imaging. In abdominal imaging, both low- and high-contrast resolution are of equal and critical importance. As a classical example: detecting contrast material washout in hepatocellular carcinoma on delayed imaging requires low-contrast resolution, detecting small arterial branches within the liver on arterial perfusion phase images requires high-contrast resolution.

2.2 Technological Development

Many factors influence the maximum achievable spatial resolution. In XY-direction (axial plane), high-contrast spatial resolution is predominantly determined by the number of projections (sampling frequency) and the number and size of detector elements. High-resolution scans should not be acquired with the fastest gantry rotation on any of the currently available CT

scanners, because the sampling rate usually stays constant thereby limiting spatial resolution; conversely, longer gantry rotation times facilitate an increased number of projections. The spatial resolution currently achieved on scanners from all manufacturers varies from 0.33 to 0.47 mm (with a high-frequency reconstruction kernel). The higher the sampling frequency, the more data are to be collected, stored and eventually transmitted to the reconstruction unit. Managing the sheer amount of raw-data signifies a technological challenge, and further advancements in data transmission will facilitate faster sampling rates, provided the detector material allows for respective limiting after-glow effects.

Minimising detector element apertures continues to marshal engineering efforts given extremely high physical demands. As anti-scatter grids and optical barriers (to prevent detector cross-talk) remain necessary, the overall geometrical efficiency of detector arrays may suffer even further using smaller elements. On the other hand, higher resolution aperture should dramatically improve mid- to high- frequency performance allowing for image reconstruction with higher spatial resolution, especially in unification with iterative reconstruction methods to overcome associated increases in image noise (see below). Reconstructible axial slice widths of less than 0.3 mm will become clinical reality in the future.

Image noise represents the limiting factor for low-contrast resolution. Radiation dose concerns in the public and amongst professionals have shifted research and development in CT towards dose reduction techniques. Fewer photons inevitably lead to a lower signal above baseline noise, subsequently resulting in noisier images. A myriad of techniques to confine image noise have been developed, amongst which the recently introduced iterative image reconstruction method attracts the most attention. Currently, model-based iterative reconstruction techniques in both the image and data domain (Fig. 2) are being evaluated (Mieville et al. 2012) and partially implemented into clinical practice. It is foreseeable that further advances in purely raw-data-based iterative reconstruction methods and improved modelling of scanner properties into the algorithm will dramatically improve image quality thus allowing for very low-dose CT imaging in most abdominal indications. Both low- and high-contrast resolution will likely be improved using these new methods.

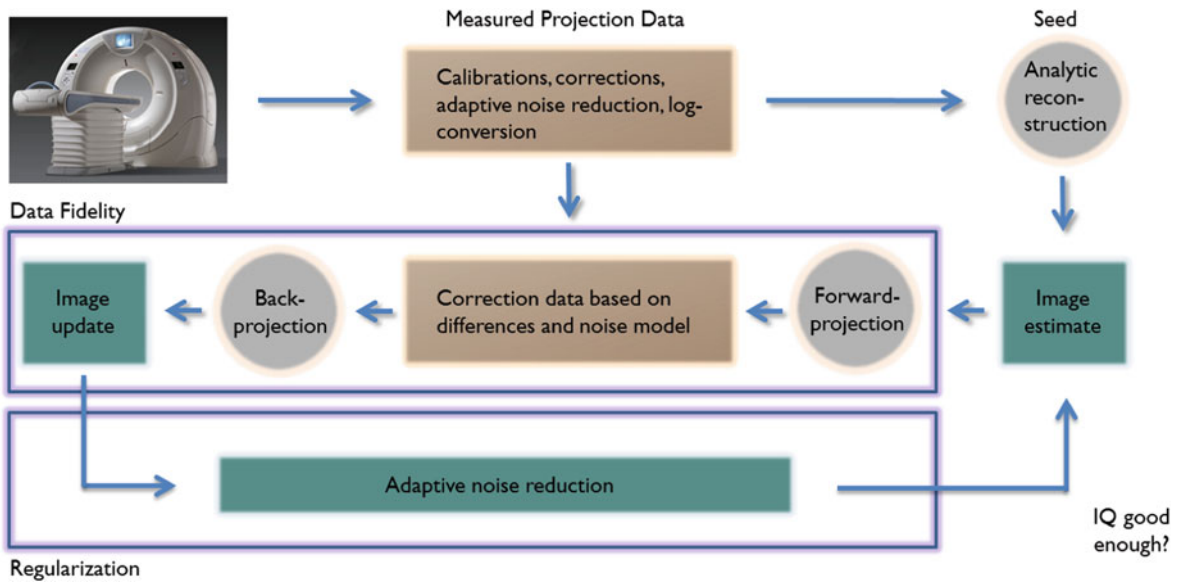


Fig. 2 Schematic drawing of the data flow in raw-data-based iterative reconstruction

3 Temporal Resolution

3.1 Clinical Demand

Improvements in temporal resolution may not lie in the forefront of priorities in abdominal imaging, the demand is mainly driven by cardiac CT. On the other hand, shorter in-plane exposure times will reduce in-plane motion artefacts from breathing, bowel movements, vessel pulsation or patient unrest during scanning.

3.2 Technological Development

Accelerating the gantry is an appealing proposal, however, as described above, basic physical constraints limit faster gantry revolving times. Nevertheless, with the advent of lighter tubes and further miniaturization of all electronic components, revolving times of 150–200 ms are feasible and likely to become reality soon.

Mounting more than one tube within the gantry represents another solution; dual-source CT units have been on the market since 2007 and have significantly contributed to the success story of cardiac CT. Mounting more than two conventional tubes within a gantry might not only be challenging from

the engineering point of view as the gantry becomes even heavier, the diagnostic value might be rather limited in light of subsequent problems including scattered radiation and geometrical constraints (limited fields-of view for each tube-detector system).

Why not remove mechanical constraints as the limiting factor entirely for enhanced temporal resolution? Electron-beam CT with its unrivalled exposure time of 50 ms per sweep has been extensively evaluated for cardiac imaging (Enzweiler et al. 2004). Image quality never reached the standard of conventional gantry designs and output restrictions of the X-ray gun limited its ubiquitous use in abdominal imaging. The scenario might change again as soon as alternative small X-ray sources in fixed position within the gantry (such as CNTs) will produce sufficient power for clinical imaging needs. Prototype scanners, albeit for non-clinical applications, are currently being developed and tested. Preliminary results are promising (inverse geometry CT) (Mazin et al. 2007; Schmidt 2011).

4 Image Quality

4.1 Clinical Demand

Generating artefact-free high-quality images is the essence of research and development in CT technology. Apart from high-contrast, low-contrast, temporal

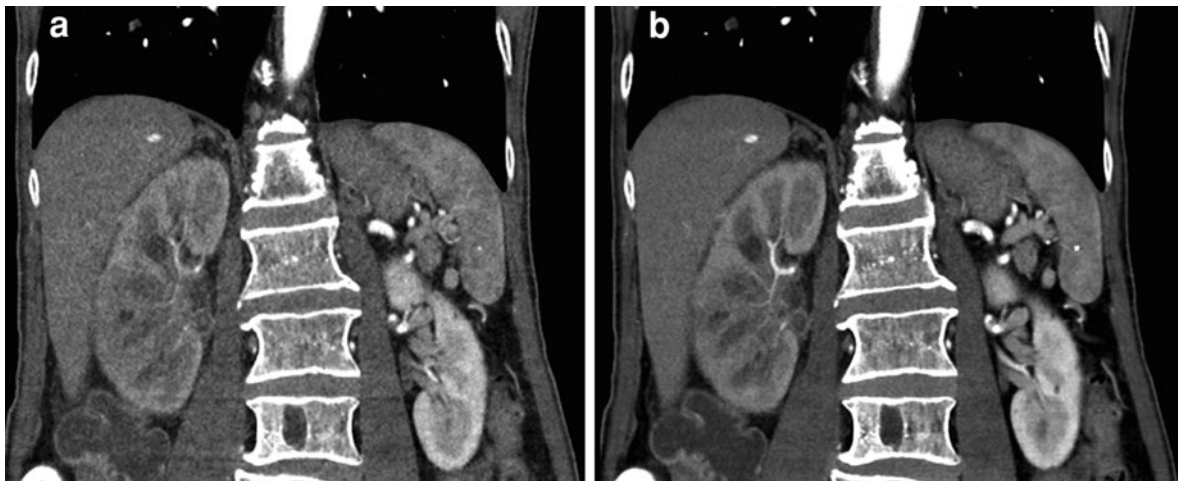


Fig. 3 Example of image quality improvement by means of model-based iterative reconstruction. *Note:* Not only contains image (b) less image noise (translating into $\sim 50\%$ radiation dose reduction), the vascular contrast (best seen within the

kidney) is perceptibly increased at the same time. **a** Conventional filtered back-projection (FBP). **b** Model-based iterative reconstruction (AIDR 3)

resolution and artefact reduction: pixel noise—often referred to as “image noise”—is perceived as the leading parameter that determines overall image quality. With the current focus on radiation dose amongst the public and professionals, dose reduction has attracted a lot of attention and evolved as a determining factor for modality decision making. Unfortunately, radiation dose reduction, unless compensated for by other methods, inevitably leads to noisier images which is why a significant volume of research is devoted towards noise reduction methods. High image noise and severe artefacts may render CT images non-diagnostic (Barrett and Keat 2004).

4.2 Technological Development

Currently, the key to success in noise reduction is iterative image reconstruction techniques (Yu et al. 2011). Be they based on image data, raw-data domain or a combination of both, iterative methods have already demonstrated their effectiveness in noise suppression while maintaining diagnostic image quality. Although available clinical data are still preliminary, it seems that radiation dose reduction of more than 50% is clinically achievable without jeopardising overall diagnostic image quality (Fig. 3). Since image noise predominantly degrades low-contrast resolution,

one might expect relevant improvements in low-contrast resolution using the new reconstruction methods. It is expected that in the near future, raw-data-based iterative reconstruction algorithms with improved modelling will become the standard method in CT imaging.

Reduction of artefacts mainly from radio-dense implants such as metallic prostheses has been a focus of research for decades. Many mathematical approaches for metal artefact reduction (MAR) have been developed and explored of which the most simple, linear interpolation to gap the metal shadow, performed not dissimilar as more complex approaches (Kalender et al. 1987). However, all previously assessed MAR algorithms, besides removing the metal artefacts, introduced new artefacts and cannot completely recover the information from the metal trace. More advanced methods such as empirical beam-hardening correction (EBHC) and normalised MAR (NMAR) are currently under investigation; both have proven to be quite effective in phantom studies and initial clinical applications (Kyriakou et al. 2010; Meyer et al. 2010) (Fig. 4). Without doubt, these or similar mathematical methods will be integrated into future CT reconstruction algorithms and thus will contribute to further image quality improvements in abdominal imaging.

Motion plays a minor role in abdominal CT, although pulsation artefacts may occur in the aorta or

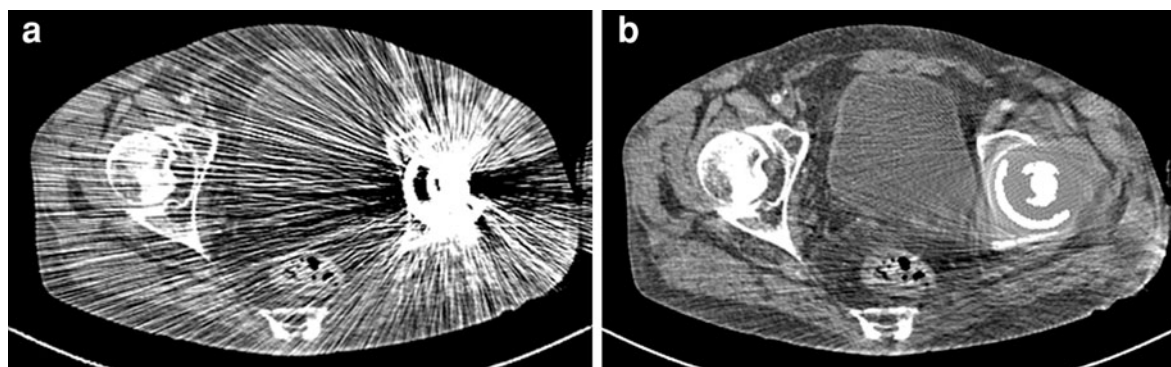


Fig. 4 Axial scan through the hip. The original uncorrected image (a) exhibits severe amounts of streak artefacts caused by the metallic hip implant. Diagnostic image quality may be

restored image (b) using normalised metallic artefact reduction (NMAR) (*courtesy Marc Kachelrieß*)

renal arteries and thus reduce conspicuity of small vascular details. In cardiac imaging, repeated (and to the major part predictable) motion occurs. In principle, feeding preexisting anatomical knowledge into the reconstruction process may be utilised in many ways, contributing to improved temporal resolution and motion correction (Tang et al. 2010). Prior image constrained compressed sensing (PICCS) has been shown to improve temporal resolution especially in cardiac CT. One might expect improvements of temporal resolution by a factor of 2 without modification of scanner hardware, however, evaluation of these methods are necessary in order to assess their applicability and usefulness in general body CT.

5 Radiation Dose

5.1 Clinical Demand

Exposure to radiation has become a major public concern (see following chapter) in particular as hypothetical risks have been falsely perceived as evidenced-based knowledge. In any event, radiation dose concerns have driven research endeavours in CT for the past decade and we are about to see major technological steps towards low-dose and ultra-low-dose imaging strategies. Radiation dose reduction today is all about increased dose efficiency along the chain of detectors, amplifiers (data acquisition system, DAS), data transmission and reconstruction algorithms. The goal however is to utilise the least amount of radiation in order to obtain an image quality

that is believed to be diagnostically sufficient. For that reason, dose reduction may be achieved in many ways including image noise reduction alone or improving low-contrast resolution. Improved temporal resolution may translate into lower radiation doses if shorter exposure times can be applied for motion capture.

5.2 Technological Development

Image noise reduction is the most obvious method for dose reduction as one can afford less radiation dose to achieve similar noise levels in the final image. Quantum noise dominates at higher doses, whereas electronic noise (i.e., noise originating from all electronic components within the detector and amplifier chain) typically dominates at low-dose levels. Increased integration and further miniaturisation of electronic components will contribute to a lower electronic noise floor allowing for improved signal-to-noise ratio in ultra-low-dose applications.

Another development for further noise reduction at the detector side is to reduce the number of steps required for conversion of X-ray photons into an electrical signal. In current solid-state detectors, X-ray photons are converted to light and light is converted to an electrical signal. Photon-counting detectors convert X-ray photons directly into an electrical charge (Fig. 5). The pulses created in the detector by the incident photons are easily detected, since the pulse height is typically an order of magnitude higher than the electronic noise in the detector. Thus the detector counts the number of peak pulses

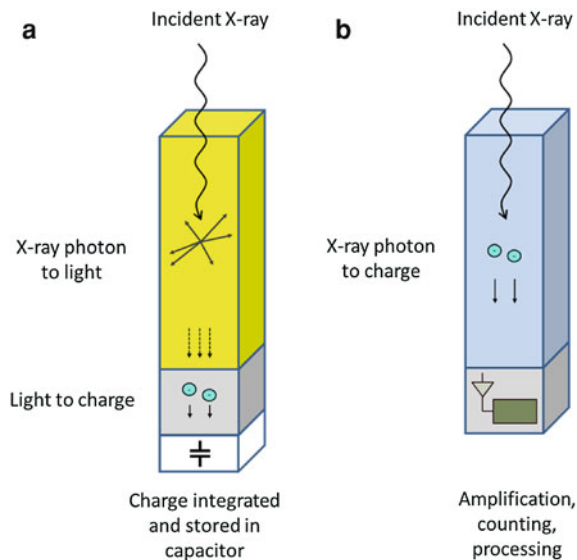


Fig. 5 Conventional detector design. **a** Incident photons are converted to light within the solid-state detector material, and in a second process, light is converted to an electrical charge. The charge is then integrated and stored in a capacitor. Photon-counting detectors (**b**) directly convert X-ray photons to charge that is amplified, counted and processed

and virtually no other signal. Following processing steps including pulse shaping and height analysis, the electric signal may be assigned to more than one energy bin, allowing for spectral analysis of the original X-ray signal(s) (Mieville et al. 2012; Wang et al. 2011). The lack of electronic noise and the ability to perform spectral analysis are paired with other advantages of photon-counting detectors, such as nearly 100% geometric efficiency (in comparison: current CT detector arrays range between 60 and 80% geometric efficiency), potentially higher spatial resolution and contrast-to-noise optimisation. Although the advantages of photon-counting detectors are appealing and obvious, their utilisation in clinical scanners remains far down the road as many challenges are to be overcome, including count-rate capability and count-rate-dependent energy response, not to mention the increased cost of this technology.

The current and near future perspective is that iterative reconstruction techniques, as described above, will continue to be the frontrunner of all dose reduction efforts. The impact of iterative methods is highest when data quality is poor such as with low-dose protocols or in large patients. Dose reduction

without loss of diagnostic image quality has already been proven to be possible by means of statistical reconstruction and simple iterations in the image domain. Further sophistication predominantly in the field of modelling scanner characteristics into the reconstruction process invigorate the assumption that dose reduction of more than 75% will be clinically manageable.

6 And Even More ...

Many current and future developments aim at improving image quality, reducing artefacts and radiation dose at the same time. kV adaptation serves as a good example. Based on the fact that X-ray penetration depends on its energy, kV adaptation to body size, similar to tube current modulation, will reduce artefacts and thereby improve image quality (Pinho et al. 2011). This will likely translate into some radiation dose savings.

Refinements in dual-energy CT regardless of its technical implementation (kV switching, dual-source, back-to-back or dual-energy helical scanning) are imminent and will make this interesting application even more suitable for general clinical use. New scintillator material that features a negligible afterglow in conjunction with a 100-fold faster reaction time allows for recording of 7,000 views per second and may be one of the prerequisites for single-source ultra-fast dual-energy switching, promising almost simultaneous spatial and temporal registration and material decomposition without limiting the scan field-of-view. Fewer beam-hardening artefacts, metal artefact reduction and further improvements in contrast-to-noise ratio are some of the benefits resulting from new scintillator material.

As photon-counting detectors are far from being introduced into clinical scanners, a simple but effective solution for post-patient energy separation lies in a double-layer detector design (Rogalla et al. 2009). The upper layer predominantly receives photons with lower energy and the lower layer receives photons with higher energy. Summing the signal from both detector layers returns the full signal as in a single-layer detector, but separate readouts from both layers allow for simple spectral analysis of the X-ray beam. This technology remains currently under clinical evaluation.

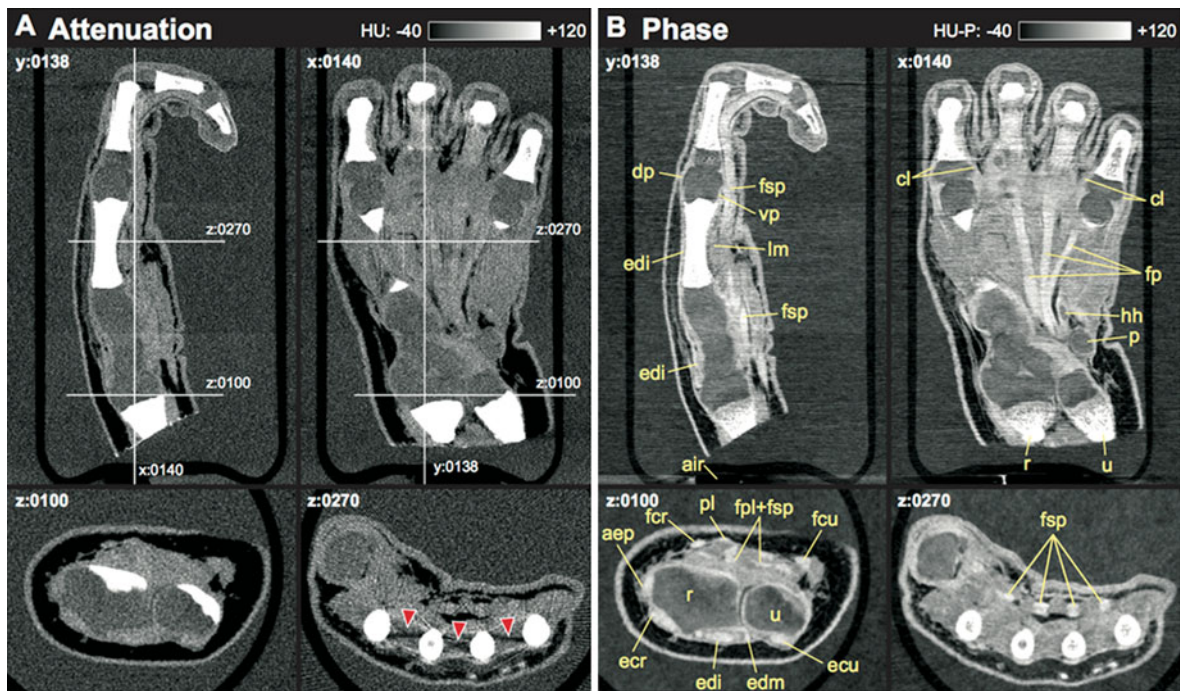


Fig. 6 Initial imaging experience in human tissue using phase-contrast CT (from: Donath et al. 2010)

Although in its earliest stage of development from the clinical perspective, phase-contrast CT carries a lot of clinical potential (Donath et al. 2010). Promising experimental results have recently been obtained in material science, biological applications (Bronnikov 2002) and in smaller body parts (Fig. 6). Moving away from absorption CT might sound revolutionary: various human tissues demonstrate very weak absorption contrast, however, produce significant phase shifts in the X-ray beam. The use of phase information for imaging purposes therefore appears to be a suitable alternative for imaging, likely contributing to further radiation dose reduction in the future.

7 Summary

MSCT has—without doubt—reached a very high degree of maturity, mastering almost all clinical demands. Nevertheless, it would be foolish to believe that technology has reached its true pinnacle. Whilst acknowledging the risks of drifting into speculation, the following technological advances are likely to become clinical reality in the near future:

- faster gantry revolving times, likely around 200 ms,
- improved temporal resolution, probably around 40–50 ms,
- larger area-detectors for extended anatomical coverage without table motion,
- smaller detector elements leading to higher spatial resolution,
- improved image quality by metal and motion artefact reduction,
- drastic radiation dose reduction, predominantly by improved iterative reconstruction methods.

Technological advances currently on the horizon are:

- new detector materials and designs, photon-counting detectors,
- alternative X-ray sources such as carbon nano-tubes,
- inverse geometry CT,
- phase-contrast CT.

In light of most recent quantum leaps in CT technology and all of the above, computed tomography remains one of the most fascinating fields of technological evolution and will undoubtedly

continue to play a pace-maker role for advances in medical imaging on the whole.

References

- Barrett JF, Keat N (2004) Artifacts in CT: recognition and avoidance. *Radiographics* 24(6):1679–1691. doi:[10.1148/rg.246045065](https://doi.org/10.1148/rg.246045065), 24/6/1679[pil]
- Bronnikov AV (2002) Theory of quantitative phase-contrast computed tomography. *J Opt Soc Am A Opt Image Sci Vis* 19(3):472–480
- Chen GH, Zambelli J, Li K, Bevins N, Qi Z (2011) Scaling law for noise variance and spatial resolution in differential phase contrast computed tomography. *Med Phys* 38(2):584–588
- Donath T, Pfeiffer F, Bunk O et al (2010) Toward clinical X-ray phase-contrast CT: demonstration of enhanced soft-tissue contrast in human specimen. *Invest Radiol* 45(7):445–452. doi:[10.1097/RLI.0b013e3181e21866](https://doi.org/10.1097/RLI.0b013e3181e21866)
- Enzweiler CN, Becker CR, Bruning R et al (2004) Value of electron beam tomography (EBT). *Rofo* 176(11):1566–1575. doi:[10.1055/s-2004-813666](https://doi.org/10.1055/s-2004-813666)
- Hounsfield GN (1995) Computerized transverse axial scanning (tomography): part I description of system. 1973. *Br J Radiol* 68(815):H166–H172
- Kalender WA, Hebel R, Ebersberger J (1987) Reduction of CT artifacts caused by metallic implants. *Radiology* 164(2):576–577
- Kyriakou Y, Meyer E, Prell D, Kachelriess M (2010) Empirical beam hardening correction (EBHC) for CT. *Med Phys* 37(10):5179–5187
- Liu X, Primak AN, Krier JD, Yu L, Lerman LO, McCollough CH (2009) Renal perfusion and hemodynamics: accurate in vivo determination at CT with a 10-fold decrease in radiation dose and HYPR noise reduction. *Radiology* 253(1):98–105. doi:[10.1148/radiol.2531081677](https://doi.org/10.1148/radiol.2531081677), 253/1/98[pil]
- Mazin SR, Star-Lack J, Bennett NR, Pelc NJ (2007) Inverse-geometry volumetric CT system with multiple detector arrays for wide field-of-view imaging. *Med Phys* 34(6):2133–2142
- Meyer E, Raupach R, Lell M, Schmidt B, Kachelriess M (2010) Normalized metal artifact reduction (NMAR) in computed tomography. *Med Phys* 37(10):5482–5493
- Mieville FA, Gudinchet F, Brunelle F, Bochud FO, Verdun FR (2012) Iterative reconstruction methods in two different MDCT scanners: physical metrics and 4-alternative forced-choice detectability experiments—a phantom approach. *Phys Med*. doi:[10.1016/j.ejmp.2011.12.004](https://doi.org/10.1016/j.ejmp.2011.12.004), S1120-1797 (11)00160-8[pil]
- Okada M, Kim T, Murakami T (2011) Hepatocellular nodules in liver cirrhosis: state of the art CT evaluation (perfusion CT/ volume helical shuttle scan/dual-energy CT, etc.). *Abdom Imaging* 36(3):273–281. doi:[10.1007/s00261-011-9684-2](https://doi.org/10.1007/s00261-011-9684-2)
- Pinho D, Kulkarni N, Savage C, Sahani D (2011) Initial experience with automated kVp selection for abdomen CT scanning. Annual meeting RSNA Chicago
- Rogalla P, Kloeters C, Hein PA (2009) CT technology overview: 64-slice and beyond. *Radiol Clin North Am* 47(1):1–11. doi:[10.1016/j.rcl.2008.10.004](https://doi.org/10.1016/j.rcl.2008.10.004), S0033-8389(08)00188-7[pil]
- Schmidt TG (2011) What is inverse-geometry CT? *J Cardio-vasc Comput Tomogr* 5(3):145–148. doi:[10.1016/j.jcct.2011.04.003](https://doi.org/10.1016/j.jcct.2011.04.003), S1934-5925(11)00174-2[pil]
- Tang J, Hsieh J, Chen GH (2010) Temporal resolution improvement in cardiac CT using PICCS (TRI-PICCS): performance studies. *Med Phys* 37(8):4377–4388
- Wang AS, Harrison D, Lobastov V, Tkaczyk JE (2011) Pulse pileup statistics for energy discriminating photon counting X-ray detectors. *Med Phys* 38(7):4265–4275
- Yu Z, Thibault JB, Bouman CA, Sauer KD, Hsieh J (2011) Fast model-based X-ray CT reconstruction using spatially non-homogeneous ICD optimization. *IEEE Trans Image Process* 20(1):161–175. doi:[10.1109/TIP.2010.2058811](https://doi.org/10.1109/TIP.2010.2058811)

Multislice-CT of the Abdomen

Zech, C.J.; Bartolozzi, C.; Baron, R.; Reiser, M.F. (Eds.)

2012, XIV, 365 p., Hardcover

ISBN: 978-3-642-17862-7