Easy Guide to Low Dose

Answers for life.

SIEMENS

Foreword

Dear reader,

You have in your hands what is probably the industry's only short but comprehensive discussion of the hottest topic in CT today – radiation and radiation dose reduction in computed tomography. The interesting history of radiation, from Röntgen to Siemens newest CT Flash scanner, is covered in crisp, understandable language suitable for both medical facility employees, administration and financial department personnel, as well as for radiologists and technologists who are well educated in radiation technology.

In summary, we bring you up-to-date on what is new and developing in the field of diagnostic imaging – specifically, dose-reduction opportunities from Siemens Computed Tomograhy and the benefits to patients from these developments.

And, as an extra benefit, we hope the information contained here will assist you to better explain computed tomography to increasingly knowledgeable and demanding patients.

Good reading!

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I.A.
Radiation Defined

This chapter provides a short historical background of X-rays and other electromagnetic radiation, as well as the working principles of an X-ray tube for medical purposes.

1. Historical Development

Wilhelm Röntgen, born 27 March 1845, was a German physicist who, on 8 November 1895, discovered the radiation known today as X-rays or Röntgen rays. During 1895, Röntgen investigated the effects of radiation outside of various types of vacuum tubes (predecessors of those used in conventional TVs) when an electrical current passed through them. He repeated the experiments using a tube with a thin aluminum "window" that allowed light to exit the tube but maintained the necessary vacuum. At one point in his efforts, he covered the window with cardboard to prevent light from escaping. Yet Röntgen observed that, in spite of the cardboard covering, something caused fluorescence on a small screen outside the tube.

Röntgen speculated correctly that some unknown kind of ray might be responsible. During the following weeks, he ate and slept in his laboratory while he investigated the various properties of the new rays. He named them X-rays. At one point, while he was investigating the ability of various materials to block the rays, Röntgen saw the world's first radiographic image, his own

flickering, ghostly skeleton on a special screen. At that moment, he decided to continue his experiments in secrecy because he feared for his professional reputation if his observations were wrong.

Finally, in December 1895,[1] convinced of his observations, he published his paper, "On a New Kind Of Rays." Today, Röntgen is considered the father of diagnostic radiology, the medical specialty which uses imaging to diagnose disease.

A year later, 1896, physicist Henri Becquerel discovered that uranium salts emitted rays that resembled X-rays in their penetrating power. He demonstrated that this radiation did not depend on an external source of energy but seemed to be emitted spontaneously by uranium itself. Becquerel had in fact discovered radioactivity. Later, Marie Curie, a young Polish physicist working with Becquerel, discovered other radioactive elements (polonium and radium) and postulated the theory of radioactivity (a term coined by her[2]) which explains why some elements lose energy in form of radiation, transforming themselves spontaneously and "decaying" throughout the years. She also conducted the first studies on the treatment of cancer using radioactive substances.

2. Physical Background

Radiation, from Greek "radius", describes the phenomenon of different forms of energy that are emitted by a source and expand in circles. When we throw a stone in a still pond, the waves (kinetic energy) expand in circles to the shore of the pond where some are reflected and others absorbed. Similarly, when we sunbathe, the light we perceive and the warmth we feel are due to electromagnetic waves which transport energy from the sun expanding in circles in all directions and being absorbed or partially reflected by the objects in their way.

[1] W. C. Röntgen "Über eine neue Art von Strahlen", Sitzungsberichte der Würzburger Physik.-medic. Gesellschaft 1895.
[2] Robert Reid, Marie Curie, p. 184.

What are electromagnetic waves? They are among the most interesting and challenging phenomena of physics. To make it simple, we can imagine them as particles (photons) “wiggling” their way through space and matter. They carry a certain amount of energy which is inversely related to the wave length (Fig.1) with which they are wiggling.

Today we use electromagnetic waves for everyday purposes. Life without a TV, a radio or a mobile phone is almost inconceivable. The technology of these appliances (and many others) is based on electromagnetic radiation.

When electromagnetic waves travel through matter, part of their energy is absorbed by the atoms. Depending on the energy and thus the wavelength of the electromagnetic radiation, the atoms may lose electrons, thus changing their structure and becoming electrically charged (Fig. 2). This phenomenon is called ionization. Not all electromagnetic radiation is ionizing, e.g. visible sunlight with a wavelength between 800 nm and 400 nm is not ionizing. Only “ionizing radiation” with wavelengths shorter than 248 nm, which corresponds to an energy level of 5 eV (electron volts), such as UV light and X-rays can alter or damage organic (human or animal) tissue by changing the DNA (modifying the cell nucleus).

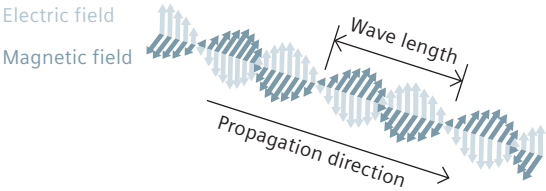


Fig. 1
Schematic illustration of an electromagnetic wave propagating through the space. The smaller the wavelength, the larger the energy that the electromagnetic wave carries.

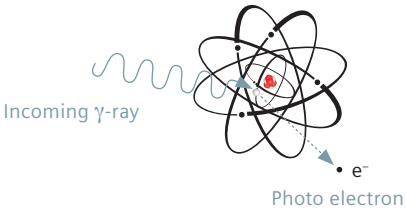
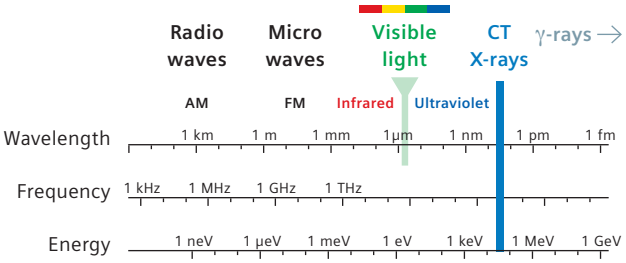


Fig. 2
Schematic illustration of the ionization of an atom by electromagnetic radiation. The energy of the incoming electromagnetic wave (incoming y-ray) is large enough to kick an electron (the photo electron) out of the atom, which is thereby "ionized".

There are other types of ionizing radiation that have not yet been mentioned. Charged particles, such as electrons, positrons, and alpha particles, also interact strongly with electrons of atoms or molecules. Radioactive materials usually release alpha particles, which are the nuclei of helium, or beta particles, which are quickly moving electrons or positrons, or gamma rays. Alpha and beta particles can cause damage to organic tissue but they can be blocked by a piece of paper (alpha particles) or a sheet of aluminum (beta particles).

Remember: There are different sources and types of radiation that can be ionizing. The radiation used in computed tomography (CT) is electromagnetic radiation (i.e. X-rays).



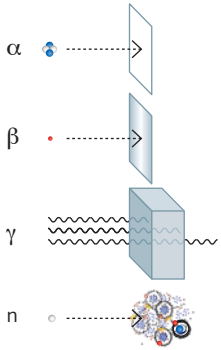


Fig. 3
Alpha (α) radiation consists of fast moving Helium-4 (^{4}He) nuclei and can be stopped by a sheet of paper. Beta (β) radiation, consisting of electrons, can be blocked by an aluminum plate. Gamma (γ) radiation and X-rays, consisting of energetic photons, are eventually absorbed as they penetrate a dense material. Neutron (n) radiation consists of free neutrons which can be blocked using light elements.

3. Natural Radiation Sources

Without radiation from the sun, life on earth would not be possible.

Sunlight, warmth and all energy forms (oil, gas, etc.) that we consume daily are the result of thousands of years of electromagnetic radiation from the sun, generated by atomic transformations at very high pressures and temperatures. Unfortunately for those who love sun bathing, over exposure to the sun's UV radiation, which is ionizing, may cause skin cancer.[3]

There are other sources of natural radiation, such as radon, which is a colorless, odorless, tasteless, naturally occurring, radioactive gas. Radon results from the normal radioactive decay of uranium. Uranium has been present since the earth was formed and has a very long half-life (4.5 billion years). The half-life of a radioactive element is the time required for the radiation to be reduced by half. Thus, radon will continue to exist indefinitely at about the same levels as it does now.[4]

Radon is responsible for most of the mean public exposure to ionizing radiation. Its concentration is variable according to location and it is often the single biggest contributor to the amount of background radiation an individual receives. Radon gas from natural sources can accumulate in buildings, especially in confined areas such as basements. No one can avoid the exposure to radon even though this may potentially cause damage. Breathing high concentrations of radon can cause lung cancer and could even be the second most frequent cause of lung cancer according to the United States Environmental Protection Agency.[5]

[3] Wang S, Setlow R, Berwick M, Polsky D, Marghoob A, Kopf A, Bart R (2001). "Ultraviolet A and melanoma: a review". J Am Acad Dermatol 44 (5): 837–46.
[4] Toxological profile for radon, Agency for Toxic Substances and Disease Registry, U.S. Public Health Service, In collaboration with U.S. Environmental Protection Agency, December 1990.
[5] A Citizen's Guide to Radon". U.S. Environmental Protection Agency. 2007-11-26.

These examples show that everyone is exposed to different sources of natural radiation in daily life, with positive and negative aspects. The additional exposure with X-rays in medicine has to be seen in this context. Without the use of X-rays many diseases could not be diagnosed early enough for efficient treatment. If properly indicated the use of radiation for medical imaging outweighs the additional radiation risk by far.

4. X-rays and Working Principles of an X-ray Tube for Computed Tomography

X-rays are electromagnetic waves, similar to visible or UV light. X-rays used in computed tomography (CT) have a mean energy of 50–70 keV (kilo-electron volts) and a wavelength of 0.018–0.025 nm. This type of radiation is ionizing and can therefore pose a danger to organic tissue, depending on the dose.

In a CT X-ray tube, X-rays are produced by an electron beam striking an anode "target". The electrons that make up the beam are emitted by a heated cathode filament. The electrons are then focused and accelerated towards the focal spot by a high voltage of 70–140 kV that is applied between cathode filament and anode. The electron beam strikes the anode and part of its kinetic energy is converted into X-ray photons, the other part into thermal radiation that heats up the anode. X-rays are emitted in all directions from the anode surface, the highest intensity being around 60° to 90° from the electron beam due to the angle of the anode. There is a small "window" that allows the X-rays to exit the tube with little attenuation while maintaining the vacuum seal required for X-ray tube operation. A generator is used to supply the X-ray tube with a controlled high voltage between cathode and anode, and a controlled current to the cathode. If the current increases, more electrons will be beamed to the anode producing more X-rays. If the voltage between cathode and anode is increased, the electrons will be faster and more X-rays and X-rays with higher energy will be produced in the anode. Hence, changing both the current (mA setting) and the high voltage (kV setting) will alter the output of the X-ray tube.

The X-ray beam is then projected onto the patient to be examined. Some of the X-rays will pass through the patient, while some are absorbed. The resulting radiation pattern is then detected by solid-state detectors, that consist of scintillation crystals which convert the X-ray energy into visible light and semiconductor photo-diodes that measure the light intensity.

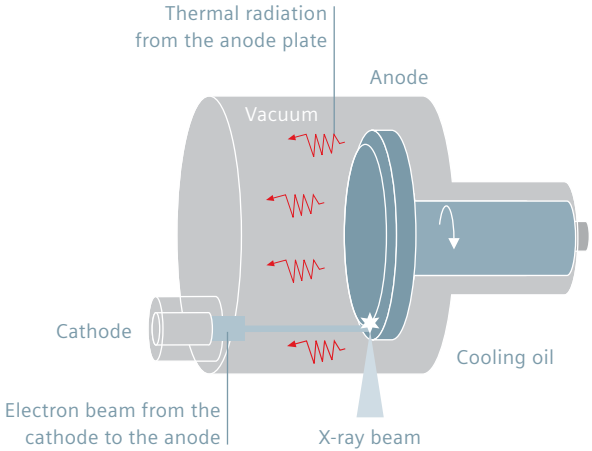


Fig. 4
Illustration of a CT X-ray tube. The rotating anode enables faster heat dissipation. The blue line from cathode to anode represents the electron beam, the light blue cone the X-rays that are emitted and leave the tube through the tube window. Only part of the energy of the incoming electron beam is converted into X-rays, the rest is converted into heat. The thin red arrows represent thermal radiation due to the heating of the anode plate.

I.B.
Biological Effects of Radiation

In this chapter, the differences between long and short-term damage to biological tissue is discussed as well as how to estimate damage in relation to the amount and type of radiation received. Different definitions of radiation dose commonly used in radiology will be introduced, such as “absorbed”, “equivalent” and “effective” dose.

1. Short and Long-Term Biological Damage

As mentioned before, ionizing radiation may, depending on the dose, cause damage to organic tissue. The mechanisms by which radiation damages the human body are twofold: (1) radiation directly destroys the DNA of the cells by ionizing atoms in its molecular structure or, (2) radiation creates free radicals which are atoms, molecules, or ions with unpaired electrons. These unpaired electrons are usually highly reactive, so radicals are likely to take part in chemical reactions that eventually change or harm the DNA of the cells.

The human body can repair damaged cells to a certain extent, but if exposed to a high amount of radiation beyond a given threshold in a short period of time, "deterministic" damage will occur. This term implies that the radiation poisoning definitely occurs and that the damage is dependent on the amount of radiation received. Deterministic radiation damage includes changes of the blood count, hair loss, tissue necrosis or cataract. Exposure levels of typical medical CT scans are far below the threshold for deterministic radiation damage.

Lower levels of radiation may cause long-term or "stochastic" damage. In this context "stochastic" means that the probability of suffering a disease caused by radiation is proportional to the amount of radiation received years before. The self-repair mechanism of the cells fail, and some cells may experience non-lethal DNA modifications that are passed to subsequent cell divisions. Years after exposure, diseases such as cancer or leukemia may occur.

In fact, the effect of the very low amounts of radiation encountered under normal circumstances (from both natural and artificial sources, such as cosmic rays or medical X-rays) is subject to constant debate. There are two main models used to predict the effects of low amounts of radiation: the linear, non-threshold model and the threshold model. The linear, non-threshold model assumes that the response is linear (i.e., directly proportional to the amount) at all levels of radiation exposure. The more radiation was received, the more likely a disease caused by radiation will occur.

The threshold model proposes that anything below a certain level of radiation is safe, and only if this level is exceeded does the probability of radiation damage increase proportionally to the received radiation (Fig. 5). Some authors even postulate that low levels of radiation have a bio-positive effect (hormetic model).

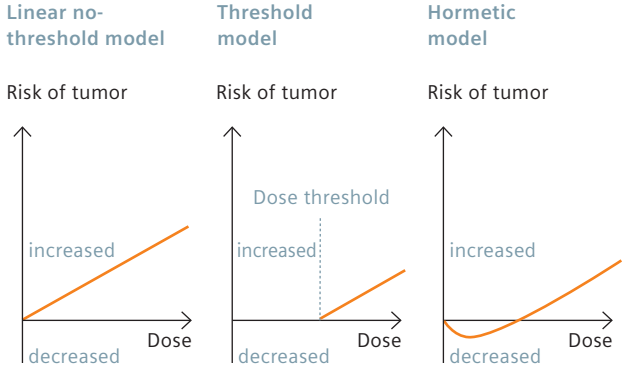


Fig. 5
Linear, non-threshold model, threshold model, and hormetic model. The x-axis represents the radiation amount and the y-axis the likelihood of a potential later damage, such as cancer.

Despite the strong controversy and different opinions of internationally recognized scientific institutions, the linear non-threshold model is the currently most accepted risk model also for low levels of radiation. The United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR) wrote in its most recent report:[6]

"Until the [...] uncertainties on low-dose response are resolved, the Committee believes that an increase in the risk of tumor induction proportionate to the radiation dose is consistent with developing knowledge and that it remains, accordingly, the most scientifically defensible approximation of low-dose response. However, a strictly linear-dose response should not be expected in all circumstances." [6]

We all have an intuitive understanding of what "dose" is, but a radiation dose that reflects the potential damage to organic tissue can't be defined simply as a certain amount of radiation energy per kg or cm^2 of body surface. That is why three different definitions are used: absorbed, equivalent and effective dose. In the following sections, we will define precisely why we need to differentiate between these and how they are defined.

2. Absorbed Radiation Dose

The energy dose or absorbed dose characterizes the amount of energy deposited in matter after being exposed to a certain amount of radiation. The unit used to measure it is the Gray (Gy) and it is defined as the amount of radiation required to deposit 1 Joule (J) of energy in 1 kilogram of any kind of matter. Therefore:

$$1\ \text{Gy} = 1\ \text{J/kg}$$

and the absorbed dose D is

$$D = \text{Absorbed Radiation Energy/kg of matter}$$

When we irradiate 1 kg of water with 1 Gy,
the water stores 1 Joule
and its temperature increases by only 0.00024 °C

Unfortunately this rather simple definition is a physical quantity and does not reflect the biological effects of radiation, since it does not take into account the type of radiation or the damage it might cause in different tissues.

[6] UNSCEAR 2000 REPORT Vol. II: Sources and Effects of Ionizing Radiation: Annex G: Biological effects at low radiation doses. page 160, paragraph 541.

3. Equivalent and Effective Radiation Dose

Equivalent dose:

The biological damages caused by different types of radiation are not the same; therefore even if an absorbed dose of X-rays or α-rays is similar, the damage can be dramatically different.

The equivalent dose for any type of radiation is defined as the absorbed dose (D) multiplied by a factor (w_r) that weighs the damage caused to biological tissue by a particular type of radiation. In the case of X-rays used in CT, the weighting factor is 1; therefore the equivalent dose is the same as the absorbed dose. In the case of α-rays, that occur naturally and are emitted, e.g. by some types of uranium isotopes, the absorbed dose has to be multiplied by a factor of 20. This indicates that α-rays cause much more damage to biological tissue than X-rays.

The unit used to measure the equivalent dose is the Sievert (Sv) and the equivalent dose H is

$$H = D \cdot w_r$$

where w_r is an estimate of the amount of biological damage caused by 1 Gy of the corresponding type of radiation.

Effective dose:

The damage that radiation causes in different types of organic tissue is not identical; e.g. red bone marrow is very sensitive to radiation, whereas the liver is much less sensitive.

When estimating the stochastic damage caused by irradiation of the human body, these differences have to be considered. The effective dose reflects this, because it is a weighted average of the equivalent dose received by the organs:

$$E = \sum w_i \cdot H_{org,i}$$

where w_i is a coefficient that quantifies the sensitivity of the particular organic tissue to the radiation received. Assuming that the brain and the thyroid gland are irradiated, the effective dose would be calculated as follows:

$$E = w_{thyroid} \cdot H_{thyroid} + w_{brain} \cdot H_{brain}$$

where $w_{thyroid}$ and w_{brain} indicate how sensitive these organs are to radiation and $H_{thyroid}$, H_{brain} are the equivalent doses received by these organs.

The weighting factors w_i are estimated and published by the International Commission on Radiological Protection. As research and measuring technologies advance, these factors may undergo changes.

The Recommendations of the International Commission on Radiological Protection of 2007 (ICRP 103) has different coefficients than that of 1990 (ICRP 60). In particular, gonads are less radiosensitive and the breast is more radiosensitive than previously assumed, as shown in Table 1:

Tissue or organ	w_i according to the ICRP 60	w_i according to the ICRP 103
Gonads	0.20	0.08
Red bone marrow	0.12	0.12
Colon	0.12	0.12
Lungs	0.12	0.12
Stomach	0.12	0.12
Breast	0.05	0.12
Liver	0.05	0.04
Esophagus	0.05	0.04
Thyroid	0.05	0.04
Skin	0.01	0.01
Bone surface	0.01	0.01
Salivary glands	–	0.01
Brain	–	0.01
.....	.....	.....
$\sum w_i$	**1.00**	**1.00**

Table 1
Weighting coefficients "w_i" according to the International Commission of Radiological Protection.[7]

Please note that the effective dose E is an approximate measure that was introduced to compare the stochastic risk of a non-uniform exposure of ionizing radiation with the risk caused by a uniform exposure of the whole body. E depends on model assumptions which may not be valid for an individual. Hence, E is not useful to determine the specific risk of an individual after receiving a certain amount of radiation.

[7] **ICRP 103:** Annals of the ICRP. ICRP Publication 103. 2007 Recommendations of the International Commission on Radiological Protection. Oxford, New York, Frankfurt: Pergamon Press; 2008. **ICRP 60:** Annals of the ICRP. ICRP Publication 60. 1990 Recommendations of the International Commission on Radiological Protection. Oxford, New York, Frankfurt: Pergamon Press; 1991.

Summarizing:

Absorbed dose D (also called "energy dose") measured in Gray (Gy) units, characterizes the amount of energy deposited in tissue. It is defined as the amount of radiation required to deposit 1 Joule (J) of energy in 1 kilogram of any kind of matter.

Equivalent dose H, measured in Sievert (Sv) units, takes into account the damage caused by different types of radiation. It is the absorbed dose multiplied by a weighting factor w_r, characteristic for the particular type of radiation. For X-rays $H = D$.

Effective dose E, measured in Sievert (Sv) units, includes the sensitivity to radiation of the different organs. It is the sum of the equivalent doses in all irradiated organs multiplied by the respective tissue weighting factors w_i.

I.C.
Dose and Radiation Risk in Computed Tomography (CT)

We now discuss how radiation dose during a CT scan is estimated and the factors that affect it, including the difficulties in analyzing the real risk that can be attributed to CT scans, and some interesting comparisons with environmental influences.

1. CT Specific Dose Parameters: CTDI and DLP

During a CT scan, cross sections of the body are irradiated. Nevertheless, the X-ray dose delivered to the body is not exactly confined to the user-defined slices, but extends outside due to scattered radiation (Fig. 6).

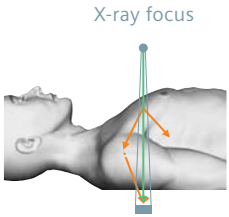


Fig. 6
Contribution of direct and scattered radiation to an axial CT slice.

The scattering of the X-rays has to be included in the calculation of the absorbed dose D. The Computed Tomography Dose Index (CTDI) is the sum of the absorbed dose in the slice and the contributions outside (the tails in Figs. 7 and 8), normalized to the nominal slice thickness S.

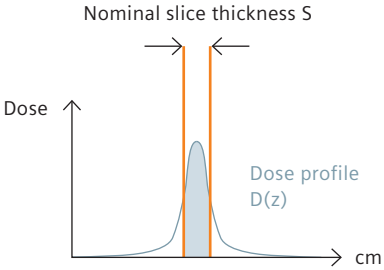


Fig. 7
Absorbed dose in the slice.

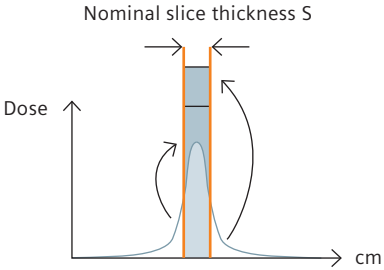


Fig. 8
Absorbed dose including scatter contributions from outside the slice (CTDI).

Mathematically, the CTDI is calculated as the integral of the absorbed dose along the z axis, divided by the nominal slice thickness S.

CTDI is the measure of the dose deposited in a single axial slice of the patient. The unit used to measure it is the mGy (1 mGy = 1 Gy/1000).

In practice, the integration limits cannot be extended to infinity. CTDI as defined by the FDA requires an integration length of 7 nominal slice thicknesses "S" on either side of the irradiated slice. The more common definition today, $CTDI_{100}$, requires an integration range of 50 mm on either side of the irradiated slice. This is more practical, since most ionization chambers used to measure CTDI are 100 mm long. The ionization chambers are placed in the center and the periphery of Plexiglas dummies of 16 cm diameter for the head and 32 cm diameter for the body (Fig. 9).

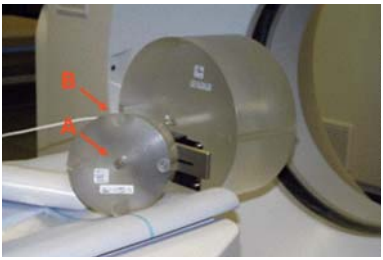


Fig. 9
Plexiglas phantoms for measurement of the peripheral (B) and central (A) absorbed dose.

There are different ways to calculate the CTDI. One of them is to consider the differences between the absorbed dose in the periphery and in the centre of the patient's body by a weighted sum of the central and peripheral CTDI values.

The resulting formula for the weighted $CTDI_w$ that takes into account this difference is:

$$CTDI_w = \frac{1}{3} CTDI_{100}^{A} + \frac{2}{3} CTDI_{100}^{B}$$

2. Important Parameters that Affect the Absorbed Dose in CT

Volume CT scans include many sequential slices during a spiral scan. For this reason, the velocity with which the table moves has to be considered: if the table moves slowly, the X-ray beam profiles will overlap (Fig. 10).

For a spiral scan, pitch is defined as the longitudinal distance in mm that the table travels during one revolution of the X-ray tube divided by the nominal irradiated width of the detector projected to the isocenter of the scanner.

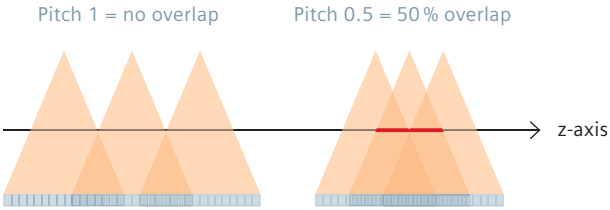


Fig. 10
If the table moves fast (pitch = 1) the X-ray beam profiles do not overlap, if the table moves slowly (pitch = 0.5) the X-ray beam profiles overlap. Please note that the overlap is measured at the isocenter of the scanner (along the z-axis).

For a spiral examination, the $CTDI_{vol}$ is:

$$CTDI_{vol} = CTDI_{w} \cdot 1/\text{pitch}$$

If the Pitch < 1, the X-ray beam profiles overlap and the absorbed dose increases. If the Pitch > 1, the X-ray beam profiles do not overlap, there are gaps in the acquisition and the absorbed dose decreases. This is valid for both single-detector and multi-detector row CT.

The expected $CTDI_{vol}$ is displayed on the user interface of the CT scanner prior to each scan. The operator can therefore easily observe on the screen the absorbed dose according to the parameters chosen for the scan. See Fig. 11:

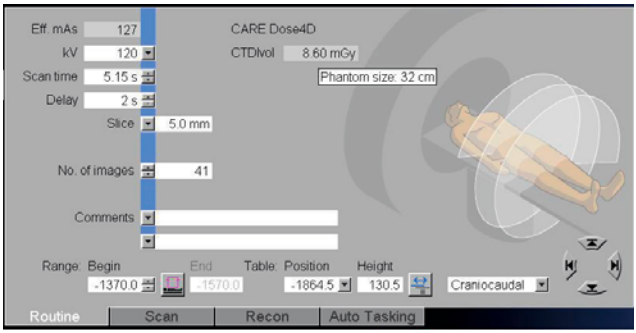


Fig. 11
$CTDI_{vol}$ for the chosen parameters.

In order to calculate the total absorbed dose for a complete CT examination, the range that is being examined must be taken into account. See Fig. 12:

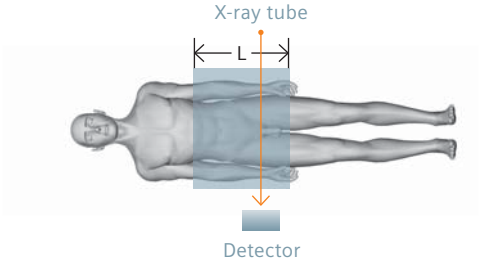


Fig. 12
The X-ray tube and the detector scan the patient along L (examination range) on the z-axis.

The Dose Length Product (DLP) is the product of $CTDI_{vol}$ and the examination range:

$$DLP = CTDI_{vol} \cdot L$$

It is measured in mGy · cm. Both $CTDI_{vol}$ and DLP for each CT examination are stored with the patient protocol and are therefore readily available.

Another aspect to be considered is that the absorbed dose is also related to the size of the patient. If a patient is smaller than the 32 cm plexiglass phantom used to determine the body CTDI, the actual absorbed dose will be higher. If the patient is bigger, the actual absorbed dose will be lower.

Is the patient shape/cross section similar to that one of the CTDI phantom, the $CTDI_{vol}$ can be used as an estimate for absorbed patient dose.

Remember: D = Radiation Energy/kg of matter

3. Effective Dose in CT

The effective dose in CT considers the direct and scattered radiation for all organs in the scan volume. It cannot be calculated exactly for each patient, but it can be estimated closely by means of Monte Carlo simulations, assuming an idealized "mean" patient. In Fig. 13, an adult, hermaphrodite, mathematical phantom is shown that was used for Monte Carlo simulations of effective doses by the UK National Radiological Protection Board (NRBP) in 1989.

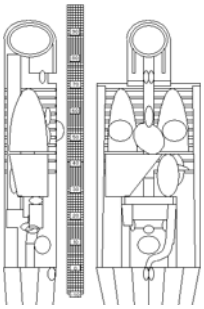


Fig. 13
Typical phantom used to calculate effective doses.

The effective dose in CT is therefore a measure of the mean radiation burden based on a patient group, **not** a measure of the radiation burden of an individual patient, who normally deviates from the idealized "mean" patient as shown in Fig. 14.

Remember: Effective dose = $\sum D_{org} \cdot w_{org}$

The effective dose is the sum of the doses for all organs, multiplied by the respective tissue weighting factors.

For different scan ranges, the effective dose E can be calculated approximately from the DLP.

Remember: DLP is the **total** absorbed dose for a complete CT examination and the range that is being examined is taken into account → DLP = $CTDI_{vol} \cdot L$

$$E = DLP \cdot f$$

f is a mean weighting factor (average between male and female models) for different regions of the human body. i.e.:

Head: f = 0.0021 mSv / (mGy · cm)

Neck: f = 0.0059 mSv / (mGy · cm)

Thorax: f = 0.014 mSv / (mGy · cm)

Abdomen and Pelvis: f = 0.015 mSv / (mGy · cm)

Table 2 shows typical examples of the effective dose for different CT routines:

Protocol	$CTDI_{vol}$	Effective dose
Head Routine 120 kV, 340 mAs, 12 cm	59.7 mGy	1.5 mSv
Thorax Routine 120 kV, 120 mAs, 30 cm	9.2 mGy	3.9 mSv
Abdomen Routine 120 kV, 180 mAs, 30 cm	13.8 mGy	6.2 mSv

Table 2
Effective dose in mSv for head, thorax, and abdomen routines.

4. Radiation Risk in CT

As shown in Table 2, the effective dose during any of the CT routines used are far below the threshold needed for deterministic damage of a part of the body shown in Table 3[8]:

Radiation dose	Damage
> 1 Sv	Bone marrow damage with changes of the DNA
2–10 Sv	Headache, fever, infections, hair loss, vomiting, nausea, cataract
10–15 Sv	Severe bowel damage

Table 3
Equivalent radiation dose for the onset of deterministic radiation damage.

So what is the risk of stochastic damage after one CT scan?

This aspect still remains uncertain. There exist only some assumptions and models that quantify this risk.

The most important study that addresses this, is the one conducted by Preston et.al[9] on 105,000 radiation victims in Hiroshima and Nagasaki, 35,000 having received radiation doses between 5 and 200 mSv. Unfortunately, this study shows a high statistical uncertainty in the low-dose range that applies to CT scanning.

There are, according to Muirhead[10]: "... uncertainties about the shape of the dose-response, both for cancer and for non-cancer diseases, below about 100 mSv."

As already discussed, the assumption today is a linear relationship between the radiation dose and the additional cancer risk with no dose threshold (linear non-threshold model – LNT) and that risk depends strongly on the age at the time of irradiation (the younger a child, the higher the risk).

[8] Source: H. G. Vogt, H. Schultz, "Grundzüge des praktischen Strahlenschutzes".
[9] Preston D L et al, Solid Cancer Incidence in Atomic Bomb Survivors: 1958–1998, RADIATION RESEARCH 168, 1–64 (2007).
[10] Muirhead C R et al, STUDIES ON THE HIROSHIMA AND NAGASAKI SURVIVORS, AND THEIR USE IN ESTIMATING RADIATION RISKS, Radiation Protection Dosimetry Vol. 104, No. 4, pp. 331–335 (2003).

In a recent publication, Brenner et al estimated the lifetime risk of death from cancer attributable to a CT scan[11]. Their estimations are shown in Fig. 14.

A **Head CT, 340 mAs**

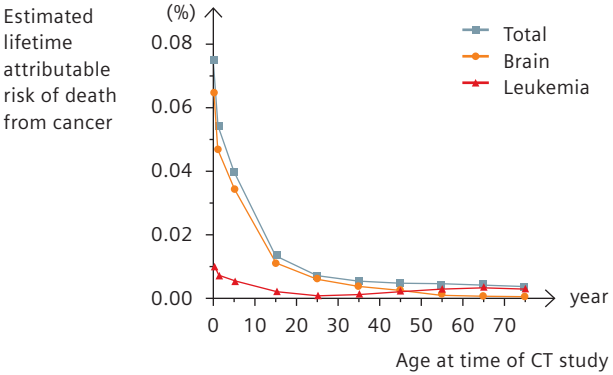
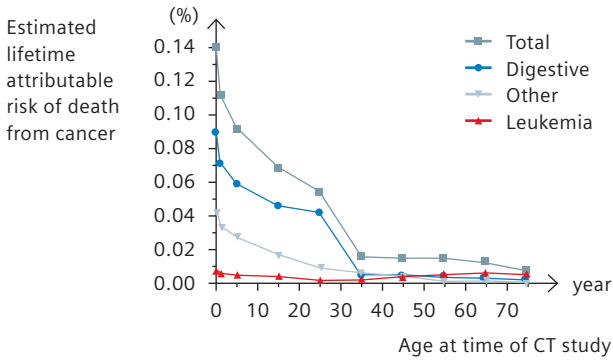


Fig. 14
Estimated risk of death by cancer attributable to a CT scan at different ages.
A: CT of the head at 340 mAs and **B:** Abdominal CT at 240 mAs.

B **Abdominal CT, 240 mAs**



The International Commission on Radiological Protection (ICRP) of 1990 assumed an excess life-time cancer mortality risk of about 5 % per Sv. As a consequence, a CT-examination with 10 mSv increases cancer mortality risk by about 0.05 %. This value is in reasonable agreement with Brenner's assumptions (Fig. 14).

[11] Brenner et al, New England J Medicine, 2007, pp 2277 ff.

However, this risk has to be framed appropriately:

The average cancer mortality risk in a Western society is about 25 %. After a CT-examination with 10 mSv, it is increased only by 0.05 % (25.05 %). This is the same increase of mortality risk as living in Central London for 450 days (death caused by air pollution) or living in the same apartment with a smoker for 540 days.[12]

Therefore, if clinically indicated, the benefit of a CT examination outweighs by far the additional radiation risk for the patient. Nevertheless, Siemens' ultimate goal is to adhere to the ALARA (As Low As Reasonably Achievable) principle, i.e. to use the lowest possible dose to obtain the required diagnostic quality images.

[12] J. T. Smith: Are passive smoking, air pollution and obesity a greater mortality risk than major radiation incidents? BMC Public Health 2007, 7:49.

Cause of Death	Estimated No. of Deaths per 1,000 Individuals
Cancer[13]	228
Motor vehicle accident	11.9
Radon in home	
Average U.S. exposure	3
High exposure (1–3%)	21
Arsenic in drinking water	
2.5 µg/L (U.S. estimated average)	1
50 µg/L (acceptable limit before 2006)	13
Radiation-induced fatal cancer	
Routine abdominopelvic CT	0.5
Single phase, ~ 10-mSv effective dose	
Annual dose limit for a radiation worker	
10 mSv (recommended yearly average)	0.5
50 mSv (limit in a single year)	2.5
Pedestrian accident	1.6
Drowning	0.9
Bicycling	0.2
Lightning strike	0.013

Table 4[14]
Estimated lifetime risk of death from various sources

[13] Levin B, Lieberman DA, McFarland B, et al.; American Cancer Society Colorectal Cancer Advisory Group, US Multi-Society Task Force, and American College of Radiology Colon Cancer Committee. Screening and surveillance for the early detection of colorectal cancer and adenomatous polyps, 2008: a joint guideline from the American Cancer Society, the US Multi-Society Task Force on Colorectal Cancer, and the American College of Radiology. CA Cancer J Clin 2008; 58:130–160

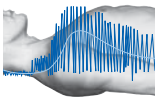
[14] Adapted with permission form: Gerber TC, Carr JJ, Arai AE, et al. Ionizing radiation in cardiac imaging: a science advisory from the American Heart Association Committee on Cardiac Imaging of the Council on Clinical Cardiology and Committee on Cardiovascular Imaging and Intervention of the Council on Cardiovascular Radiology and Intervention. Circulation 2009; 119:1056–1065

II.A.
Dose Reduction Advances in Computed Tomography

In this chapter we discuss different technologies and algorithms that Siemens has implemented or developed to reduce the absorbed dose to a minimum.

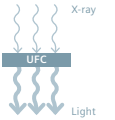
Siemens strives to implement all dose reduction methods available in the CT market today. As a leader in the dose reduction field, we also consistently develop our own solutions. Therefore, we were the first to implement many dose-saving features into clinical routine and for many critical features, we are still the only vendor offering these leading edge solutions.

1994



CARE Dose4D

1997

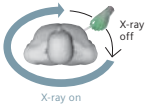


UFC

1999

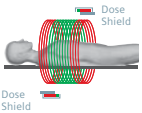
Adaptive ECG-Pulsing

1999



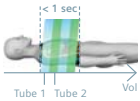
HandCARE

Adaptive Dose Shield



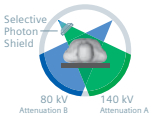
2007

Flash Spiral



2008

Selective Photon Shield



2008

4D Noise Reduction

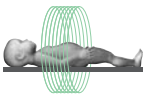
2008

To maintain our leading position and to improve health care for patients, we cooperate closely with experts from around the globe in universities, public clinics and private radiology centers to bring research developments into practical, everyday clinical routine.

In addition to the newest technology, dose reduction in CT requires training i.e. familiarity with dose reduction methods and factors. We therefore attempt to make our dose savings products as transparent as possible to reading physicians and technologists and also offer an on-going choice of seminars and resources relative to dose reduction.

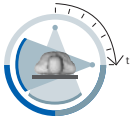
Below, you will find brief exposés of our dose-reduction products and algorithms (More detailed information can be found at www.siemens.com/low-dose):

2002



Pediatric 80 kV Protocols

2005

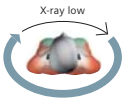


DSCT

2007

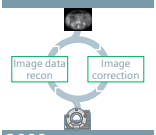
Adaptive Cardio Sequence

X-CARE



2008

IRIS



2009

1. "CARE Dose4D" – Real-time Anatomic Exposure Control
2. "Adaptive ECG-Pulsing" – ECG-Controlled Dose Modulation for Cardiac Spiral CT
3. "Adaptive Cardio Sequence" – ECG-triggered Sequential CT
4. "Adaptive Dose Shield" – Asymmetric Collimator Control
5. "Flash Spiral" – ECG-Triggered Dual Source Spiral CT Using High Pitch Values
6. "X-CARE" – Organ Based Dose Modulation
7. "IRIS" – Iterative Reconstruction in Image Space
8. "CARE Dose kV" – Automated Dose-optimized Selection of the X-ray Tube Voltage (in development)

1. "CARE Dose4D" – Real-time Anatomic Exposure Control

The most efficient way to reduce radiation dose in CT is an adaptation of the scan parameters to the anatomy of the patient. Centering the patient correctly, using the right protocols and adjusting the X-ray tube output to the patient's size and shape help to minimize radiation exposure. Many users, however, may not fully know how parameters should be modified to adjust radiation dose levels for different patients. As an example, they may not be aware that the tube output can be reduced by a factor of two while still maintaining adequate image quality if the patient's diameter decreases by only 4 cm. Hence, in all modern Siemens CT scanners, control mechanisms are available that automatically adjust the radiation dose level to the patient's anatomy – similar to a highly sophisticated camera's automatic exposure mode.

Siemens CARE Dose4D automatically adapts radiation dose to the size and shape of the patient, achieving optimal tube current modulation in two ways. First, tube current is varied on the basis of a topogram, by comparing the actual patient to a "standard-sized" patient. As might be expected, tube current is increased for larger patients and reduced for smaller patients. Differences in attenuation in distinct body regions are taken into account. For example, in an adult patient, 140 mAs might be needed in the shoulder region, whereas 55 mAs would be sufficient in the thorax, 110 mAs in the abdomen, and 130 mAs in the pelvis.

In addition, real-time angular dose modulation measures the actual attenuation in the patient during the scan and adjusts tube current accordingly – not only for different body regions, but also for different angles during rotation. This is particularly important in efficiently reducing dose in the shoulder and pelvic region, where the lateral attenuation is much higher than the anterior-posterior attenuation. Fig. 15 demonstrates the working principle of CARE Dose4D. Fig. 16 is a clinical example obtained with the use of CARE Dose4D.

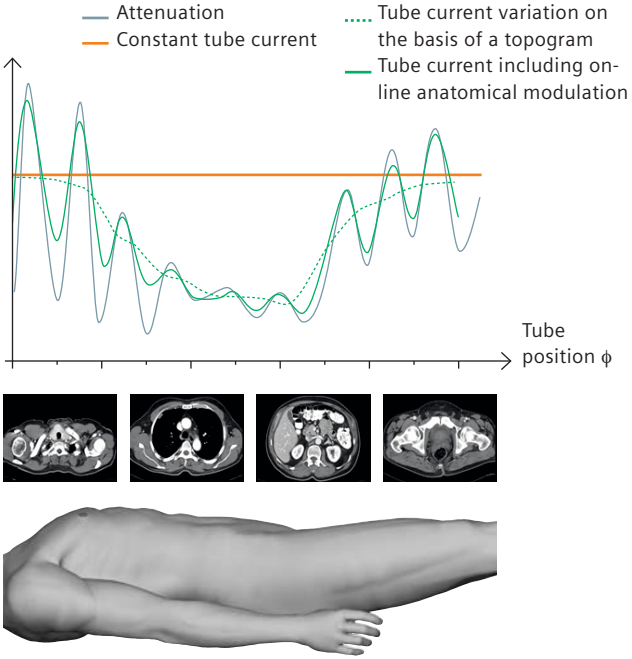


Fig. 15
Illustration of the working principle of CARE Dose4D. With constant tube current, regions in the shoulder and the pelvis would be under-dosed, while thorax and abdomen would be significantly over-dosed. On-line anatomical dose modulation efficiently adapts the tube current and hence the radiation dose to the patient's attenuation.

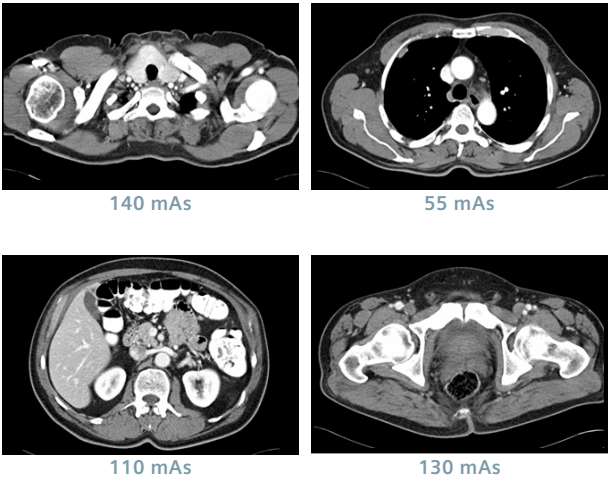


Fig. 16
CARE Dose4D for a scan from the shoulders to the pelvis produces optimized radiation dose for all anatomic regions.

Clinical experience has shown that the relationship between optimal tube current and patient attenuation is not linear. Larger patients clearly need a higher dose than average-sized patients, but they also have more body fat, which increases tissue contrast. Smaller patients need a lower dose than average-sized patients, but they have less fat and less tissue contrast, which would result in noisy images if the dose were too low. Therefore, during real-time dose modulation, CARE Dose4D reduces radiation dose less than might be expected for smaller patients, while increasing the dose less than might be expected for larger patients. This maintains good diagnostic image quality while achieving an optimal radiation dose (Fig. 17).

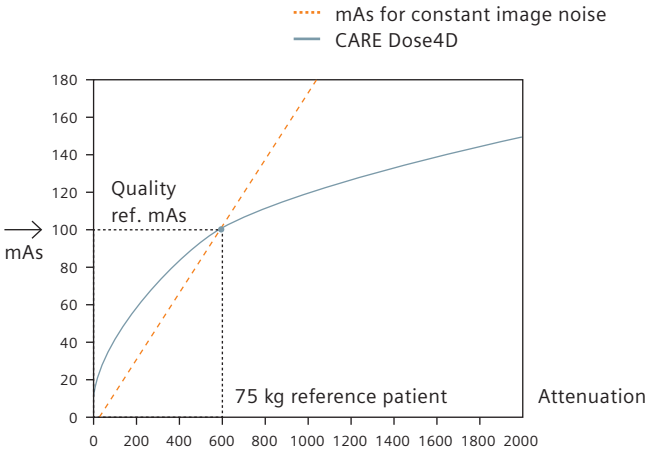


Fig. 17
Dose adaptation with CARE Dose 4D.

2. "Adaptive ECG-Pulsing" – ECG-Controlled Dose Modulation for Cardiac Spiral CT

With this method, the radiation dose is modulated during the complete spiral CT scan by using information from the patient's ECG. The tube current is maintained at 100 % of the desired level only during a predefined "phase of interest" of the patient's cardiac cycle. During the rest of the time the current can be reduced to as low as 4 %, thus reducing the mean radiation dose by up to 30–50 %[15] (Fig. 18).

ECG-controlled dose modulation is based on the continuous monitoring of the ECG and an algorithm that predicts when the desired ECG phase will start by calculating the mean durations of the preceding cardiac cycles. Older ECG-pulsing approaches reach their limitations with arrhythmia patient scans that cannot be predicted by simple averaging. Recently, more versatile ECG-pulsing algorithms have been introduced which react flexibly to arrhythmia and ectopic beats and have the potential to considerably enhance the clinical application spectrum of ECG-controlled dose modulation.

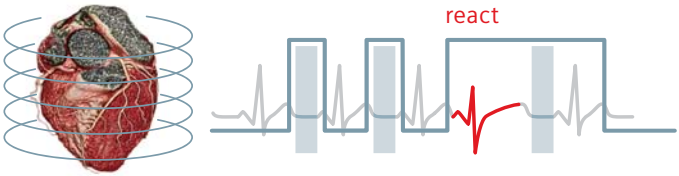


Fig. 18
The CT generates images only during a pre-defined phase of the heartbeat. During this phase, the tube current (blue line) is 100 % of the necessary level to achieve adequate image quality, but between these pre-defined phases, the current is reduced to 20 % or even 4 %. Recently introduced algorithms can react flexibly to arrhythmia.

[15] Jakobs T F et al. Multislice helical CT of the heart with retrospective ECG gating: reduction of radiation exposure by ECG-controlled tube current modulation, Eur. Radiol. 2002, 12: 1081-1086.

3. "Adaptive Cardio Sequence" – ECG-Triggered Sequential CT

Prospective ECG-triggering combined with "step-and-shoot" acquisition of axial slices is a very dose-efficient way of ECG-synchronized scanning because only the very minimum of scan data needed for image reconstruction is acquired during the previously selected heart phase. The patient's ECG-signal is monitored during examination, and axial scans are started with a pre-defined temporal offset relative to the R-waves. With conventional approaches, the method reaches its limitations with patients with severe arrhythmia, since ECG-triggered axial scanning depends on a reliable prediction of the patient's next cardiac cycle by using the mean length of the preceding cardiac cycles. With the adaptive cardio sequence, a more refined analysis of the patient's ECG is performed. Irregularities are reliably detected, and in case of an ectopic beat, the scan can be repeated at the same position. Hence, the application spectrum of ECG-triggered sequential scanning is extended to patients with high and irregular heart rates.

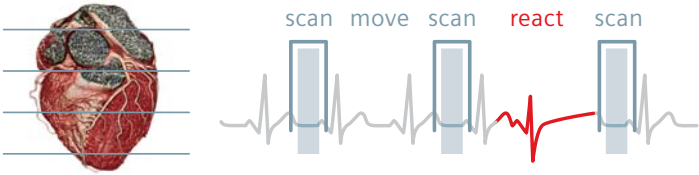


Fig. 19
Each slice of the heart is scanned during the same ECG phase.

Fig. 20
Using Adaptive ECG-Pulsing, an ECG-gated spiral scan of the heart (A) can be performed at 4–9 mSv dose[16]. With the Adaptive Cardio Sequence, an ECG-triggered sequential scan of the heart (B) requires only 1–3 mSv dose[17].

[16] Stolzmann P, Scheffel H, Schertler T, Frauenfelder T, Leschka S, Husmann L, Flohr TG, Marincek B, Kaufmann PA, Alkadhi H. Radiation dose estimates in dual-source computed tomography coronary angiography. Eur Radiol. 2008; 18(3):592-9.
Leschka S, Stolzmann P, Schmid FT, Scheffel H, Stinn B, Marincek B, Alkadhi H, Wildermuth S. Low kilovoltage cardiac dual-source CT: attenuation, noise, and radiation dose. Eur Radiol. 2008; 18(9): 1809-17.

[17] Stolzmann P, Leschka S, Scheffel H, Krauss T, Desbiolles L, Plass A, Genoni M, Flohr TG, Wildermuth S, Marincek B, Alkadhi H. Dual-source CT in step-and-shoot mode: noninvasive coronary angiography with low radiation dose. Radiology. 2008; 249(1):71-80.

4. "Adaptive Dose Shield" – Asymmetric Collimator Control

In spiral CT, it is routine to do an extra half-rotation of the gantry before and after each scan, fully irradiating the detector throughout, even though only part of the acquired data is necessary for image reconstruction. As a result, the wide cone beam exposes tissue that will never be part of reconstructed images (Fig. 21).

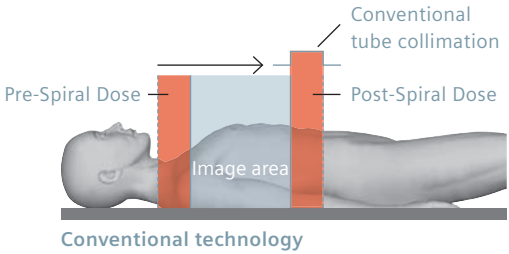


Fig. 21
Conventional pre-patient collimator. The areas marked in red are out of the necessary scan range but still irradiated with full power. This problem is typical for spiral CT and commonly referred to as "over-ranging".

The Adaptive Dose Shield, a technology based on precise, fast, and independent movement of both collimator blades, limits this over-ranging. The pre-patient collimator asymmetrically opens and closes at the beginning and end of each spiral scan, temporarily blocking those parts of the X-ray beam that are not used for image reconstruction. As a result, only the targeted tissue is irradiated (Fig. 22).

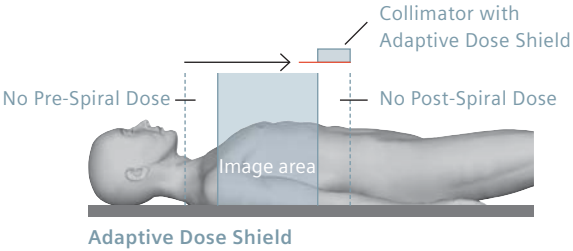


Fig. 22
Adaptive Dose Shield. When the CT scan starts, the collimator opens asymmetrically. In the center of the scan range, the collimator is fully open according to the selected beam width. At the end of the scan range the collimator closes asymmetrically.

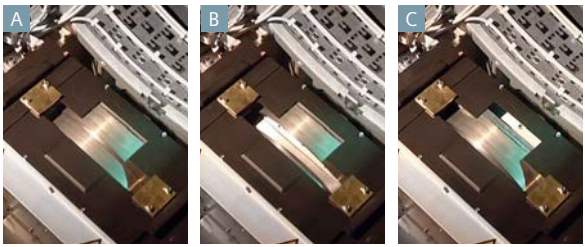


Fig. 23
The two collimators of the Adaptive Dose Shield.
A: Closed
B: Open left
C: Open right

Measurements at the Institute of Medical Physics, University Erlangen-Nürnberg, Germany, and at the Clinical Innovation Center, Mayo Clinic, Rochester, Minnesota, USA, have demonstrated significant dose reductions, depending on the scanned range, without affecting image quality (Fig. 24).

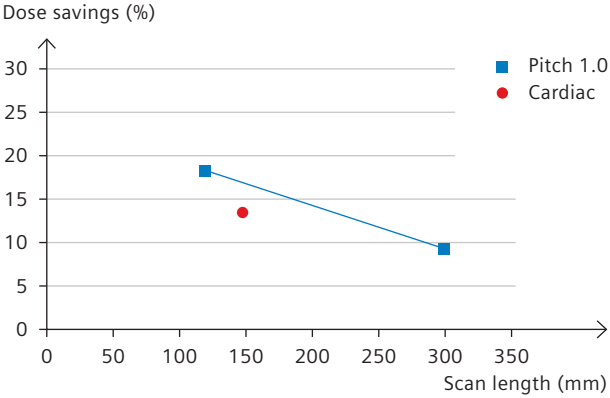


Fig. 24
Dose savings with Adaptive Dose Shield for different CT scan lengths.

5. "Flash Spiral" – ECG-Triggered Dual Source Spiral CT Using High Pitch Values

Dual Source CT (DSCT) offers a way to scan the heart within one heartbeat without using an area detector that covers the entire heart volume. With a single source CT, the spiral pitch is limited to values below 1.5 to ensure gapless volume coverage along the z-axis. If the pitch is increased, sampling gaps occur (see Fig. 25) that hamper the reconstruction of images with well-defined narrow slice sensitivity profiles and without excessive image artifacts.

With DSCT systems, however, data acquired with the second measurement system a quarter rotation later can be used to fill these gaps (see Fig. 25). In this way, the pitch can be increased up to 3.4 in a SFOV that is covered by both detectors. Since no redundant data are acquired due to the high pitch, a quarter rotation of data per measurement system is used for image reconstruction, and each of the individual axial images has a temporal resolution of a quarter of the rotation time.

The SOMATOM Definition Flash offers 38.4 mm detector z-coverage and 0.28 s gantry rotation time. At a pitch of 3.4, the table feed is 450 mm/s, which is sufficient to cover the heart (12 cm) in about 0.27 s. The scan is triggered and starts at a user-selectable phase of the patient's cardiac cycle. Each of the images has a temporal resolution of 75 ms, the phase of images adjacent in the z-direction is slightly shifted (Fig. 26). Since no overlapping data are acquired, the radiation dose of this new mode is very low and even below the dose values of ECG-triggered sequential scanning. First publications have demonstrated that reliable coronary CTA is feasible at radiation dose values below 1 mSv.[19/20]

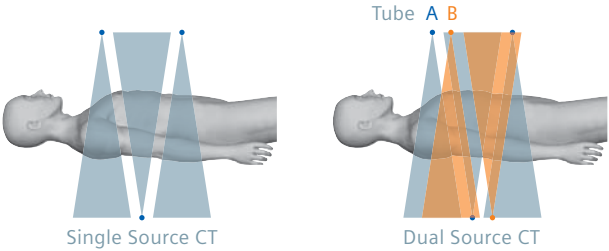


Fig. 25
Sampling scheme along the z-axis for a single source CT operating above the pitch limit of 1.5 (left), and for a dual source CT (right). Here, the sampling gaps are filled with data acquired by the second measurement system, such that considerably increased pitch values are feasible.

Continuous table feed, pitch 3.4

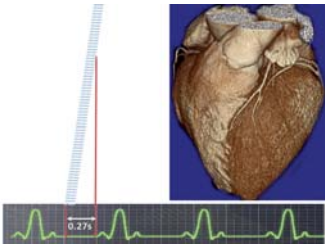


Fig. 26
Principle of ECG-triggered DSCT spiral scan data acquisition and image reconstruction at very high pitch. The patient table reaches a pre-selected z-position (e.g. the apex of the heart) at a pre-selected cardiac phase after acceleration to maximum table speed. At this pre-selected z-position data acquisition is started. Due to the fast table movement, the entire heart can be scanned in a fraction of a heartbeat. The total scan time is typically 0.25–0.27 s. The scan data for images at adjacent z-positions (indicated by short horizontal lines) are acquired at slightly different phases of the cardiac cycle. Each of the images is reconstructed using data of a quarter rotation per X-ray tube, resulting in a temporal resolution of 75 ms per image.

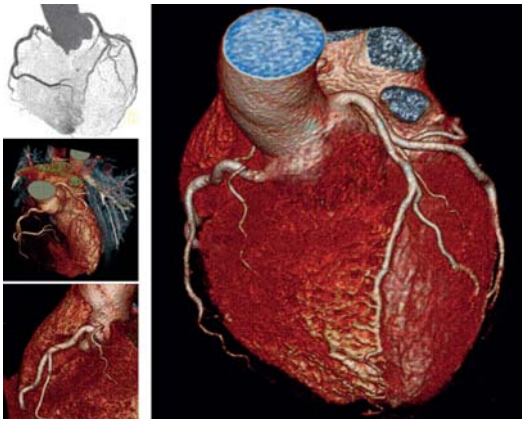


Fig. 27
CT angiography of the coronary arteries acquired with the high pitch DSCT spiral mode ("Flash Spiral").[18]

Fig. 27 shows images reconstructed in this modus with an acquisition time of 250 ms, a temporal resolution of 75 ms, 100 kV and 0.8 mSv.

First scientific papers[19, 20] on the SOMATOM Definition Flash, confirm effective radiation doses of 0.88–0.9 mSv for routine coronary CTA. Please feel free to visit the worldwide, low-dose counter at **www.siemens.com/low-dose** that displays real-time average dose values of Flash Spiral Cardio scanning though out our installed base.

[18] Courtesy of Prof. S. Achenbach, Erlangen, Germany.

[19] Achenbach S, Marwan M, Ropers D, Schepis T, Pflederer T, Anders K, Kuettner A, Daniel WG, Uder M, Lell MM. Coronary computed tomography angiography with a consistent dose below 1 mSv using prospectively electrocardiogram-triggered high-pitch spiral acquisition. Eur Heart J. 2010;31(3):340-6.

[20] Leschka S, Stolzmann P, Desbiolles L, Baumueller S, Goetti R, Schertler T, Scheffel H, Plass A, Falk V, Feuchtner G, Marincek B, Alkadhi H. Diagnostic accuracy of high-pitch dual-source CT for the assessment of coronary stenoses: first experience. Eur Radiol. 2009;19(12):2896-903.

6. "X-CARE" – Organ Based Dose Modulation

According to recently modified tissue weighting factors (recommendations of the International Commission on Radiological Protection of 2007, ICRP103), the female breast is more radiosensitive than previously assumed. In any CT examination of the thorax, the breast, without being the object of interest, is irradiated and should therefore be especially protected. Siemens X-CARE, the organ-based dose modulation, can selectively limit the radiation exposure of sensitive organs. When using this mode, radiation intensity is reduced when the patient is irradiated from the front as shown in Fig. 28.

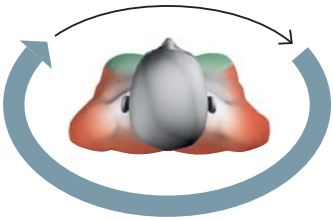


Fig. 28
Illustration of the X-CARE principle.

With this method, the radiation exposure of the breast or the eyes is reduced by 30–40 %, while image noise and detail visualization remain unaffected, as shown in Fig. 29:

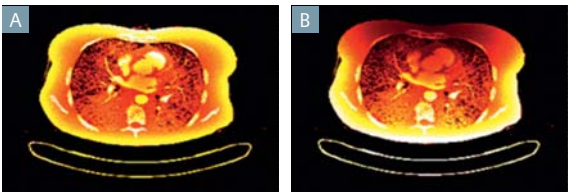


Fig. 29
A: Radiation doses without X-CARE and **B:** with X-CARE. Darker areas indicate lower absorbed dose.

7. "IRIS" – Iterative Reconstruction in Image Space

IRIS is a unique, Siemens-only method that reduces image noise without loss of image quality or detail visualization. The significant image noise reduction provided by IRIS allows for up to 60 % radiation dose reduction in routine clinical use.

Today, CT scanners use standard filtered back-projection methods in which improved spatial resolution can only be achieved at the cost of increased image noise. In contrast to filtered back-projection, iterative reconstruction enables a decoupling of spatial resolution and image noise. It enhances spatial resolution in areas with higher contrast and reduces image noise in low contrast areas, enabling the user to perform CT scans with lower radiation dose.

In an iterative reconstruction, a correction loop is introduced into the image reconstruction process. Once an image has been reconstructed from the measured projections, a ray-tracing in the image is performed to calculate new projections that exactly represent the reconstructed image. This step, called re-projection, simulates the CT measurement process, but with the image as the "measured object." If the original image reconstruction were perfect, measured and calculated projections would be identical. In reality they are not, and the deviation is used to reconstruct a corrected image and to update the original image.

Then the loop starts again. The images are improved step by step, and a significant noise reduction can be obtained by carefully modeling the data acquisition system of the CT scanner and its physical properties in the re-projection algorithm. This method is called "theoretical iterative reconstruction". The drawback of this approach is that the exact modeling of the scanner during re-projection requires high computer processing power, therefore significant hardware capacity is needed to avoid long image reconstruction times that is not available in the near future (Fig. 30).

Simplified approaches with less computer complexity and faster reconstruction are possible, but with significantly less accurate re-projection and calculation of the correction image. This may result in strange, unfamiliar noise textures and a plastic-like look of the images.

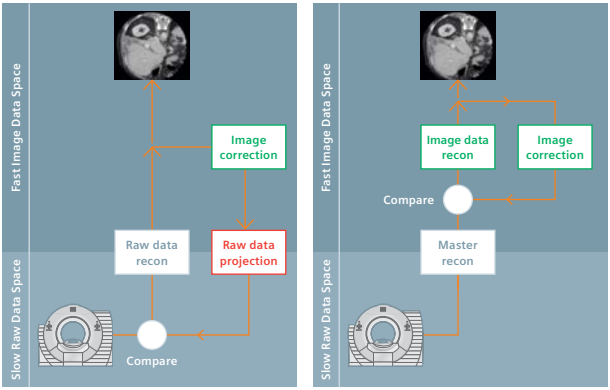


Fig. 30
Comparison of different approaches to iterative reconstruction.

Siemens has developed IRIS (Iterative Reconstruction in Image Space), a unique method that translates the iterative reconstruction loop into the image domain, hence avoiding the time consuming traditional re-projection. IRIS offers both a significant image noise reduction and a fast reconstruction for routine clinical use. In addition, the noise texture of the images is similar to standard well-established convolution kernels. Starting point of the IRIS method is a master volume reconstruction that optimally utilizes all measured data and provides all available detail information but at the expense of significantly increased image noise.

This master volume is then “cleaned up” step by step in an iterative loop using 3 to 5 iterations, enhancing object contrasts and reducing image noise with each iteration. IRIS is more advanced than simplified iterative reconstruction attempts, due to the special master reconstruction that is used to start IRIS and the special iterative structure of the image enhancement steps (Fig. 30 and Fig. 31).

Filtered Back Projection

40 % less noise with IRIS

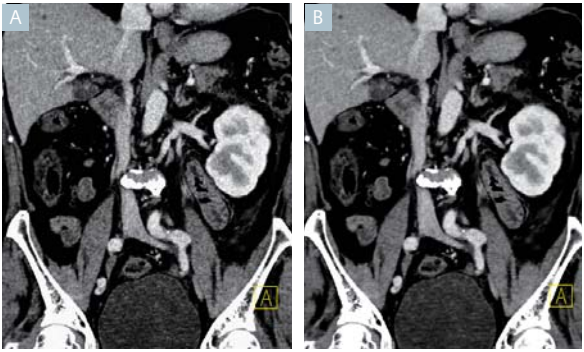


Fig. 31
Contrast-enhanced CT scan of the abdomen.
A: Standard Filtered Back Projection Reconstruction, kernel B31.
B: Iterative Reconstruction. Note the significantly decreased image noise without loss of resolution.

8. "CARE Dose kV" – Automated Dose-optimized Selection of the X-Ray Tube Voltage (in development)

Conventional dose modulation approaches control only the X-ray tube current, while the X-ray tube voltage (the kV-setting) is left untouched. Yet, there is a big potential for dose reduction by adapting the kV setting and thus the radiation energy to the diagnostic task, such that an optimized contrast to noise ratio is achieved.

The quality of CT images is characterized by three parameters: contrast, noise, and sharpness (spatial resolution).

Improving all or any of these parameters will render a better image and enable the reading physician to make a more precise diagnosis. For example, if the contrast is high and the noise is low, the image quality improves.

Additionally, an iodine contrast agent is often administered to improve contrast and thus the visibility of organ structures in CT images (particularly in CT angiographies). The contrast is best with lowered X-ray tube voltage, since the low energy X-rays are better absorbed by the dense iodine than by the surrounding tissue. However, in order to maintain low noise levels, the tube current usually requires an adjustment. Nevertheless, for a constant contrast-to-noise ratio in CT angiographic studies, the radiation dose can be significantly reduced by choosing 80 kV or 100 kV tube voltages instead of 120 kV (Fig. 32).

For larger patients, though, who have a higher X-ray attenuation, the output of the X-ray tube at lower kV settings may not be sufficient to produce the required contrast to noise ratios. For these patients, higher X-ray tube voltages will have to be selected, despite reduced iodine contrasts.

In a busy environment, the technicians and reading physicians often have insufficient time to measure the attenuation of each patient. Automatic tools that define the optimal combination of voltage and current for each patient according to the patient's topogram and the selected examination protocol are necessary and will be implemented in the near future.

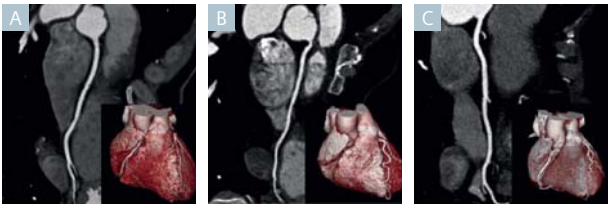


Fig. 32
Three CT angiographies with 3 different current and voltage settings. Note that the mean contrast in the aorta is higher with 100 kV.

A: 120 kV 330 mAs, CTDIvol=43.1 mGy,
Mean contrast aorta: 322 HU

B: 100 kV 330 mAs, CTDIvol=31.8 mGy,
Mean contrast aorta: 561 HU

C: 100 kV 230 mAs, CTDIvol=21.2 mGy,
Mean contrast aorta: 559 HU

II.B.
Pediatric Computed Tomography

Radiographic examinations are used much less frequently for children than for adults, because their organism is still developing and because children seldom understand the cooperation (such as breath-holds, etc.) required of them. Furthermore, the smaller the cross section of the patient, the larger the actually absorbed radiation dose. Nevertheless, computed tomography is of great importance for the treatment of pediatric patients, especially for complex lung imaging, for the treatment of congenital malformations, and in intensive care. As a consequence, the ALARA principle (As Low As Reasonably Achievable) is of particular importance in pediatrics. It calls for always selecting the dose that is as low as possible, yet sufficient for a reliable diagnosis.

The Siemens Dual Source CT SOMATOM Definition Flash system offers effective doses below 0.5 mSv in pediatric applications, with full diagnostic image quality. Due to the fast scan speed using very high pitch values (“Flash Spiral”), even uncooperative children can be examined without sedation, saving time and money and reducing stress for the patient.

A Final Word

We want to thank you very much for reading and familiarizing yourself with Siemens low-dose efforts and progress. Should you have any questions or need support or information of any kind, rest assured that the entire Siemens organization stands ready to serve you.

For further information: www.siemens.com/low-dose

The Siemens Healthcare Sector is one of the world's largest suppliers to the healthcare industry and a trendsetter in medical imaging, laboratory diagnostics, medical information technology and hearing aids. Siemens is presently the only company to offer products and solutions for the entire range of patient care from a single source – from prevention and early detection to diagnosis and on to treatment and aftercare. By optimizing clinical workflows for the most common diseases, Siemens also makes healthcare faster, better and more cost-effective. Siemens Healthcare employs some 49,000 employees worldwide and operates in over 130 countries. In fiscal year 2008 (to September 30), the Sector posted revenue of 11.2 billion Euros and profit of 1.2 billion Euros.

III.
Glossary

A

Absorbed dose	The absorbed dose is the amount of energy deposited in matter after being exposed to a certain amount of radiation
Acute radiation syndrome (ARS)	The acute radiation syndrome (ARS) is the death of large number of cells in the organs impairing their function after exposure to radiation
Adaptive Cardio Sequence	Modus in which the CT scanner registers in real time the ECG, analyses if the heart beats are normal and triggers the scan during a predefined phase of the ECG
Adaptive Dose Shield	Pre-patient collimator in which both collimator blades move asynchronously, thus reducing the radiation dose at the beginning and end of the scan range
Adaptive ECG-Pulsing	CT scan modus in which the current intensity is modulated such that the radiation is maximal during the prescribed phase of the ECG and the radiation is reduced to a minimum during the rest of the ECG
Alpha particles	Alpha (α) particles are fast moving Helium-4 (^{4}He) nuclei
Angiography	Radiographic visualization of the blood vessels after injection of a radio opaque substance (contrast agent). In CT, iodine is frequently used as a contrast agent.
Anode	The electron-collecting electrode of the X-ray tube
Apoptosis	A genetically determined process of cell self-destruction that is marked by the fragmentation of nuclear DNA, it can be activated by radiation

Attenuation	Reduction of the amount force or value of a parameter, in this context, reduction of the intensity of the radiation beam

B

Bone marrow	Soft highly vascular modified connective tissue that occupies the cavities of most bones and occurs in two forms: a: a yellowish bone marrow consisting chiefly of fat cells and predominating in the cavities of the long bones – called also yellow marrow b: a reddish bone marrow containing little fat, being the chief seat of red blood cell and blood granulocyte formation, and occurring in the normal adult only in cancellous tissue especially in certain flat bones – called also red marrow

C

CARE Dose4D	System that not only modulates the radiation intensity according to the size of the patient but also to the anatomic region which is being irradiated as the CT scan progresses
Cathode	The electron-emitting electrode of the X-ray tube
Collimator	Device (shield) for obtaining a beam of radiation (as X-rays) of limited cross section
Computed Tomography Dose Index – $CTDI_w$	Computed Tomography Dose Index (CTDI) is the sum of the absorbed dose in the slice and outside the slice (due to scattering outside the slice)
CT slice	Each transversal image generated by a CT scan
$CTDI_{vol}$	$CTDI_w$ divided by the pitch

D

Detector	Device for detecting the presence of electromagnetic waves or of radioactivity
Deterministic damage	Damage of organic tissue that will occur for sure due to the exposition to a high amount of ionizing radiation
Dose Length Product (DLP)	Dose Length Product or DLP is the product of $CTDI_{vol}$ and the length of the examination range

E

Ectopic beat	A heartbeat that is spurious and is not synchronized as normal heart beats
Effective dose E	The effective dose reflects the sensitivity of each organ and is a weighted average of the equivalent dose received by the organs
Electromagnetic radiation	Radiation that has the properties of particles and waves (photons)
Electrons	An elementary particle consisting of a charge of negative electricity and spinning around the atom nuclei
Equivalent dose	The equivalent dose for any type of radiation is defined as the absorbed dose multiplied by a factor (w_t) that weights the radiation-specific damage caused to biological tissue. In the case of X-rays used in CTs the weighting factor is 1, therefore the equivalent dose is the same as the absorbed dose
Examination range	Part of the body to be scanned along the longitudinal axis (z-axis)

F

Flash Spiral	This is a new ECG-triggered technique that is based on a dual source spiral scan at very high pitch. Enables scanning the heart in only one heart beat

Fluorescence	Luminescence that is caused by the absorption of radiation at one wavelength followed by nearly immediate re-radiation usually at a different wavelength and that ceases almost at once when the incident radiation stops
Free radicals	Atoms, molecules, or ions with unpaired electrons. These unpaired electrons are usually highly reactive, so radicals are likely to take part in chemical reactions that eventually change or harm the DNA of the cells

G

Gamma rays	(γ-rays) Radiation consisting of photons with a wave length of less than 1 pm

I

Ionization	The process by which atoms are converted into ions (electrically charged atoms)
Ionizing radiation	Radiation that can ionize matter
Irradiation	Radiation emitted with a specific, in this case, medical purpose

K

Kinetic energy	Energy stored in a moving object

M

Modulation	Adjustment of a parameter to keep it in the desired range

N

Neutrons	Uncharged elementary particle that has a mass nearly equal to that of the proton and is present in all known atomic nuclei except the hydrogen nucleus
Noise	In this context the grainy structure of a CT image.

P

Photon	A quantum of electromagnetic radiation
Pitch	Longitudinal distance in mm that the table Table feed during one revolution of the X-ray tube divided by the nominal scan width in mm
Plexiglas phantoms	Dummies of plexiglass used to measure the radiation doses on different parts of the body
Positron	Positively charged particle with the same mass as the electron

R

Radiation	The process of emitting radiant energy in form of waves or particles
Radioactive substances	Substances that emit radiation of different types
Radioactivity	The property possessed by some elements (as uranium) or isotopes (as carbon 14) of spontaneously emitting energetic particles (as electrons or alpha particles) by the disintegration of their atomic nuclei
Radionecrosis	Destruction of the organic tissue by radiation
Radon	Heavy radioactive gaseous element formed by the decay of radium
Resolution	A measure of the sharpness of an image or of the fineness with which a device (as a video display, printer, or scanner) can produce or record such an image

S

Sharpness	Clearness in outline or detail of an image, ability to resolve small details
Spiral CT	CT scan during which the table and the X-ray tube move continuously
Stochastic damage	Damage that might happen; involves a chance or probability

T

to trigger	To initiate, actuate, or set off by a certain event or signal
Topogram	Contour of the human body
Tube current	Current applied to the cathode of the X-ray tube
Tube voltage	Voltage between the anode and the cathode of the tube

X

X-CARE	Organ based dose modulation. In this modus the radiation intensity is reduce when the patient is irradiated from the front
X-rays	Electromagnetic radiations that has an extremely short wavelength of less than 100 angstroms and has the property of penetrating various thicknesses of all solids

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