Guide to Low Dose

Answers for life.

SIEMENS

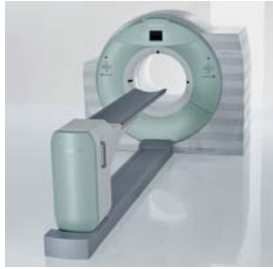# Guide to Low Dose

Foreword

Dear reader,

Radiation and radiation dose reduction are arguably the most controversial topics in medical imaging today, subjecting patients to a great deal of uncertainty and putting medical professionals under increased pressure to justify imaging procedures. Public discussions, predominantly in the U.S., but also in other countries around the world, are adding to the anxiety surrounding the question of dose.

As a leader in healthcare innovation, Siemens has a strong legacy in dose-reducing technologies. As you know, a fundamental focus of our research and development has been to reduce dose without compromising image quality and clinical outcome.

This history and knowledge is now bundled in our second edition of a short yet comprehensive discussion on dose. With our "Guide to Low Dose", we want to help healthcare professionals like you stay up to date on what is new and developing in the field of diagnostic imaging – specifically, dose-reduction opportunities from Siemens and their benefits to patients and operators.
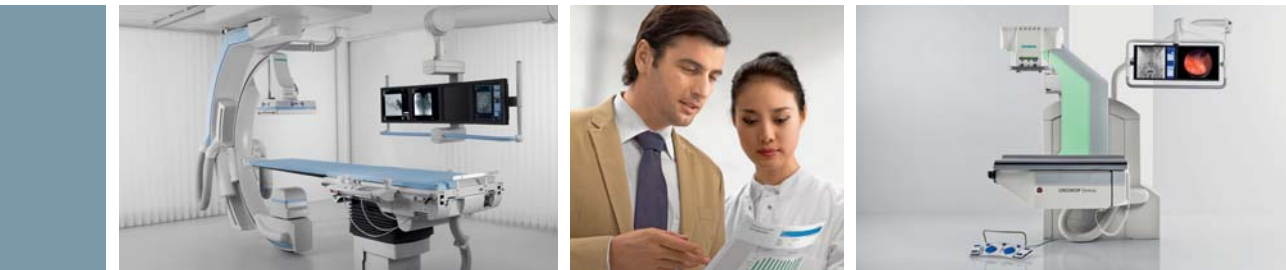
Imaging systems like the SOMATOM® Definition Flash computed tomography scanner, the Biograph mCT PET/CT scanner and the Artis zee® angiography systems are described using concise, understandable language in order to provide the latest information not only for those new to the debate about dose, but also for professionals with a long history in the field.

With this information, we hope not only to help you in your daily decisions about dose application, but also to assist you in better explaining medical imaging procedures to patients who are increasingly knowledgeable and proactive about their healthcare.

Sincerely,

Bernd Montag
CEO Imaging & Therapy Systems
Siemens Healthcare

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I. Radiation Basics

The topic of radiation dose in medical imaging has become the focus of an intense public and technical discussion. There is no doubt that medical imaging saves lives, and thus its utilization has increased immensely over the years.

As an example, in the U.S., the annual per capita radiation dose from medical exposure rose from 0.53 mSv to 3.1 mSv over the past three decades (Figure 1). Computed Tomography (CT) is under special scrutiny because it has become the single largest contributor to man-made radiation exposure. The radiation dose level from medical exposure is today in the same range as the annual natural background radiation of 3.1 mSv. The same patterns of utilization are observed in all industrialized countries, and it can be expected that other countries will follow suit as well.

As an innovation leader in dose reduction, Siemens has long applied a comprehensive approach to all areas of diagnostic and interventional imaging: CARE – Combined Applications to Reduce Exposure. Today, the Siemens CARE standard brings together a wide variety of advanced technologies and applications to meet the needs of patients and physicians for appropriate radiation dose with the best possible outcomes for diagnosis and interventions.

On the other hand, we at Siemens strongly believe that education and training are equally important pillars in this area. Only with sufficient education and training will physicians be able to weigh the potential risks and benefits of using ionizing radiation and take advantage of existing dose reduction technologies to their fullest extent.

Therefore, in this guide we aim to provide you with comprehensive background information on the physics of radiation, risk models for radiation dose and available dose reduction technologies.

1980

All other 0.05

Medical 0.53

Natural background 2.4

2006

Interventional radiology 0.4

All other 0.14

Diagnostic radiography 0.3

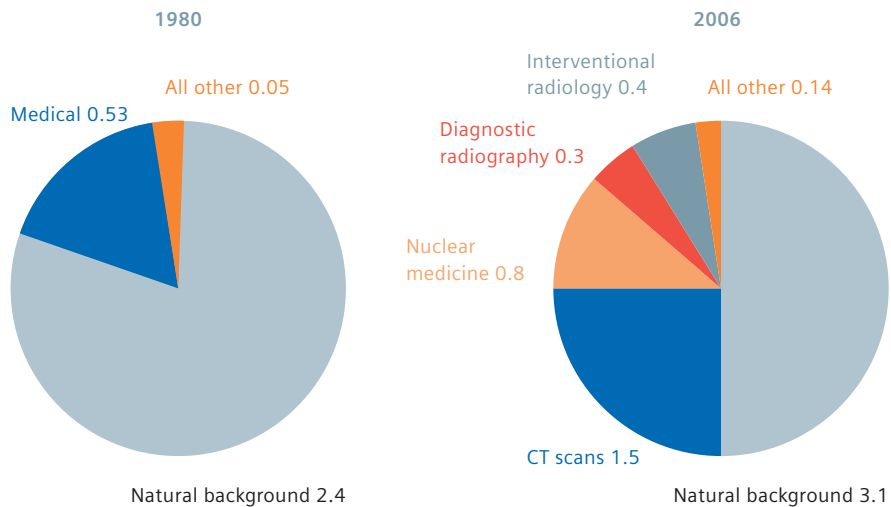Nuclear medicine 0.8

CT scans 1.5

Natural background 3.1

Fig. 1:
Distribution and development of annual per capita dose in mSv to the population from 1980 to 2006 in the USA as an example for the development in industrialized countries.[1]

[1] Sources and Effects of Ionizing Radiation, UNSCEAR 2008 Report. United Nations Scientific Committee on the Effects of Atomic Radiation, New York, 2010.

I.A. Radiation Defined

This section provides a short historical background of X-rays and other ionizing 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, despite 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 for these observations. 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, convinced of his observations, he published his paper, "On a New Kind Of Rays."[2] Today, Röntgen is considered the father of diagnostic radiology, the medical specialty that uses imaging to diagnose disease.

A year later, in 1896, French 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[3]) 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 the Greek "radius," describes the phenomenon of different forms of energy that are emitted outward in all directions from a central source. When we throw a stone in a still pond, the waves (kinetic energy) expand in concentric 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 that transport energy from the sun, expanding in circles in all directions and being absorbed or partially reflected by the objects they encounter in their path.

[2] Röntgen W. Ueber eine neue Art von Strahlen. Sitzungsberichte der Wuerzburger Physik.-medic. Gesellschaft, Wuerzburg, 1895.

[3] Robert R. Marie Curie. New American Library, New York, 1974, p. 184.

Electromagnetic waves 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 wavelength (Figure 2) of the wiggle.

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 within the matter. Depending on the energy and thus the wavelength of the electromagnetic radiation, the atoms may lose electrons, thereby changing their structure and becoming electrically charged (Figure 3). This phenomenon is called ionization. Not all electromagnetic radiation is ionizing – for example, visible sunlight, with a wavelength between 800 nm and 400 nm, is not ionizing. Only 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, is ionizing, and can alter or damage living tissue by changing the DNA.
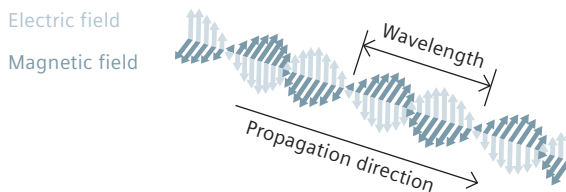
Electric field

Magnetic field

Wavelength

Propagation direction

Fig. 2
Schematic illustration of an electromagnetic wave propagating through space. The smaller the wavelength, the larger the energy that the electromagnetic wave carries.
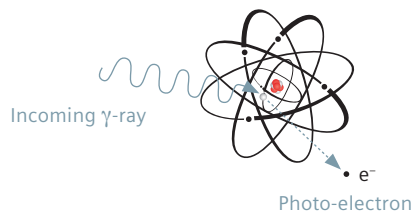
Incoming γ-ray

e⁻

Photo-electron

Fig. 3
Schematic illustration of the ionization of an atom by electromagnetic
radiation. The energy of the incoming electromagnetic wave (incoming
γ-ray) is large enough to kick an electron (the photoelectron) 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 (nuclei of helium), beta
particles, (quick-moving electrons or positrons), or gamma rays (electromagnetic radiation from
the atomic nucleus). Alpha and beta particles can cause damage to organic tissue but they can
be easily blocked – alpha particles by a piece of paper, and beta particles by a sheet of
aluminum.

Remember: **There are different sources and types of radiation that can be ionizing.
The radiation used in Computed Tomography (CT) conventional radiography and
angiography, for example, is electromagnetic radiation (i.e., X-rays). Positron Emission
Tomography (PET), Single Photon Emission Computed Tomography (SPECT) and other
nuclear imaging methods use radiation emitted during the decay of radioactive tracers
(i.e., gamma rays).**

	Radio waves	Micro waves	Visible light	X-rays	γ-rays →

| | AM | FM | Infrared | Ultraviolet | | |

| Wavelength | 1 km | 1 m | 1 mm | 1 μm | 1 nm | 1 pm | 1 fm |

| Frequency | 1 kHz | 1 MHz | 1 GHz | 1 THz | | | |

| Energy | 1 neV | 1 μeV | 1 meV | 1 eV | 1 keV | 1 MeV | 1 GeV |

α

β

γ

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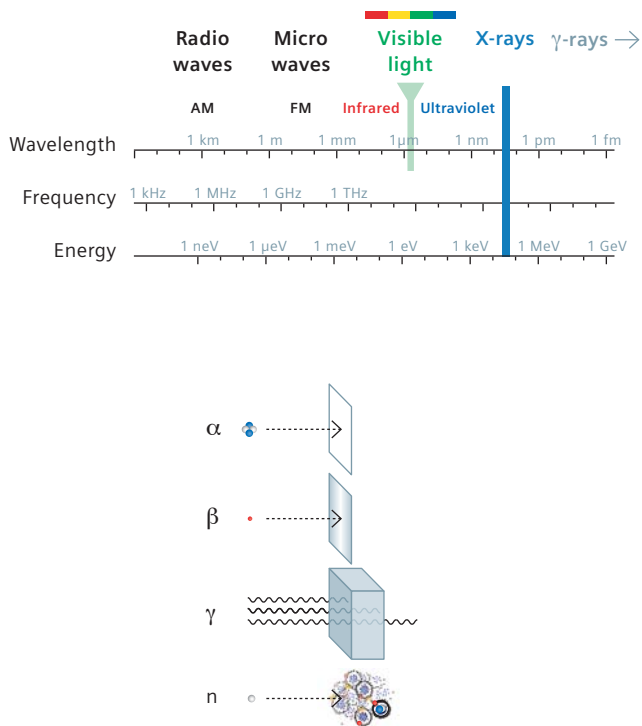
Fig. 4
Alpha (α) radiation consists of fast-moving Helium-4 (⁴He) nuclei and can
be stopped by a sheet of paper. Beta (β) radiation, consisting of electrons,
can be blocked by an aluminum plate. Gamma (γ) radiation and X-rays,
consisting of energetic photons, are eventually absorbed as they penetrate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

3. Natural Radiation Sources

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

Sunlight, warmth and all forms of energy (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, overexposure to the sun's UV radiation, which is ionizing, may cause skin cancer.[4]

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 today.[5]

Radon is responsible for most of the mean public exposure to ionizing radiation. In fact, 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. Its concentration is variable according to location, and no one can avoid exposure to radon even though this may potentially cause damage. Breathing high concentrations of radon can cause lung cancer and, according to the United States Environmental Protection Agency, may even be the second most common cause of lung cancer.[6]

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

[4] Wang SQ et al. Ultraviolet A and melanoma: a review. J Am Acad Dermatol. 2001 May;44(5):837-46.

[5] Toxicological profile for radon. Agency for Toxic Substances and Disease Registry, U.S. Public Health Service, Atlanta, 1990.

[6] A citizen's guide to radon: the guide to protecting yourself and your family from radon. United States Environmental Protection Agency, Washington D.C., 1992.

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

X-rays are electromagnetic waves, similar to visible or UV light. X-rays used in CT, for example, 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 an 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 40–140 kV that is applied between the cathode filament and the anode. The electron beam strikes the anode and part of its kinetic energy is converted into X-ray photons, while the remainder is converted 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 the cathode and anode, as well as 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 speed up, producing more X-rays and X-rays with higher energy 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. Some of the X-rays pass through the patient, while some are absorbed. The resulting radiation pattern is then detected by detectors. In earlier times, silver bromide film was used to detect the X-rays directly. Modern radiology uses mostly digital methods to detect radiation patterns. For example, modern CT scanners employ solid-state detectors in which scintillation crystals convert the X-ray energy into visible light and semiconductor photodiodes measure the light intensity.
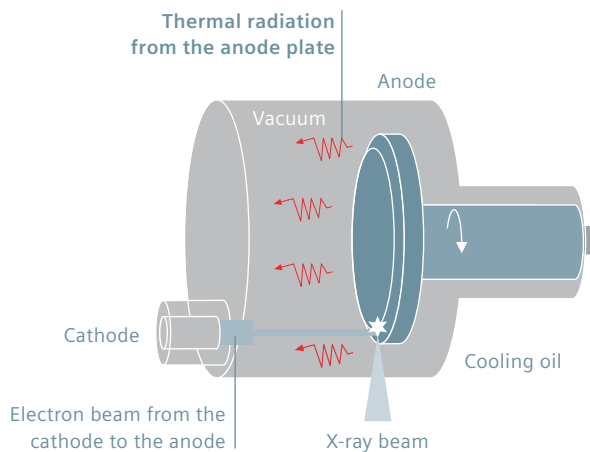
**Thermal radiation
from the anode plate**

Anode

Vacuum

Cathode

Cooling oil

Electron beam from the
cathode to the anode

X-ray beam

Fig. 5
Illustration of an 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.

5. Radionuclides and Radioactive Tracers

Atomic nuclei consist of neutrons and protons. An element is defined by the number of protons its nucleus contains, while isotopes of an element vary in the number of neutrons. Nuclei are stable only when the numeric relationship between neutrons and protons is well balanced.

There are three categories of nuclear radiation, named alpha (α), beta (β) and gamma (γ). Nuclei with a surplus of neutrons frequently exhibit β decay, in which a neutron is converted to a proton, an electron (β radiation) and an antineutrino. On the other hand, nuclei with a surplus of protons frequently exhibit β+ decay, in which a proton is converted to a neutron, a positron (β+ radiation) and a neutrino. Often, additional γ radiation is emitted to lower the energy level of the nucleus. The resulting new isotope has a better balanced number of nucleons (protons and neutrons) than the original one. Finally, alpha radiation, which consists of helium nuclei, occurs only in the radioactive decay of heavy nuclei.

For medical imaging, only isotopes with gamma or positron emission are useful. Gamma rays can penetrate human tissue and can, therefore, be detected outside the body by a medical device. Positrons have a very short range in the tissue, but upon contact with an electron, the resulting positron-electron annihilation produces two 511 keV photons (electromagnetic radiation), which can penetrate the body like gamma or X-rays. For radionuclide therapy, radiation with a short range is preferred if the isotope accumulates in the diseased tissue, in order to protect healthy tissue. This is true for isotopes emitting β radiation, α radiation or Auger electrons.

The most important properties of a radioactive isotope are its half life (the time it takes for 50% of the atoms to decay), type, probability and energy of the emitted radiation.

I.B. Biological Effects of Radiation

In this section, the differences between long- and short-term damage to biological tissue are 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, such as "absorbed," "equivalent" and "effective" dose, are introduced.

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 two-fold: (1) radiation directly destroys the DNA of the cells by ionizing atoms in its molecular structure and, (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 radiation poisoning has definitely occurred; in addition, 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 diagnostic imaging procedures are far below the threshold for deterministic radiation damage. However, deterministic effects are an important consideration in external radiation therapy and radionuclide therapy.

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 in years prior. Cellular self-repair mechanisms may fail, and some cells may experience non-lethal DNA modifications that are passed on through 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, no-threshold model and the threshold model. The linear, no-threshold model assumes that the response is linear (i.e., directly proportional to the amount) at all levels of radiation exposure. The more radiation 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 (Figure 6). Some authors even postulate that low levels of radiation have a bio-positive effect (hormetic model).
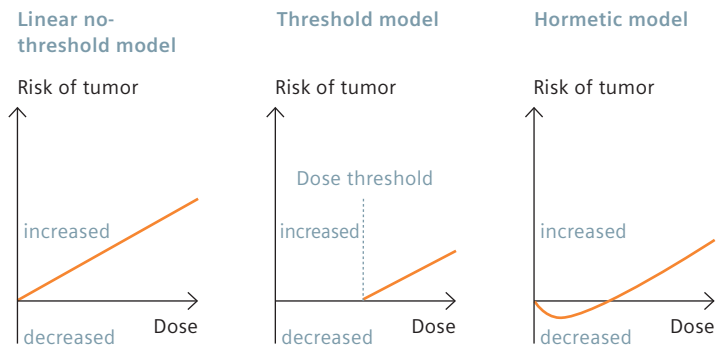
Linear no-threshold model

Risk of tumor

increased

decreased   Dose

Threshold model

Risk of tumor

Dose threshold

increased

decreased   Dose

Hormetic model

Risk of tumor

increased

decreased   Dose

Fig. 6
Linear no-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 differing opinions of internationally recognized scientific institutions, the linear no-threshold model is currently the most accepted risk model for low levels of radiation as well. For example, the Committee to Assess Health Risks from Exposure to Low Levels of Ionizing Radiation of the National Academy of Sciences concluded in its most recent report: [7]

> "The committee concludes that current scientific evidence is consistent
> with the hypothesis that there is a linear, no-threshold dose-response
> relationship between exposure to ionizing radiation and the development
> of cancer in humans."

But they also note:

> "New data and analyses have reduced sampling uncertainty, but
> uncertainties related to estimating risk for exposure at low doses and
> dose rates and to transporting risks from Japanese A-bomb survivors
> to the U.S. population remain large. Uncertainties in estimating risks
> of site-specific cancers are especially large."

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

[7] Health Risks from Exposure to Low Levels of Ionizing Radiation: BEIR VII – Phase 2. Committee to Assess Health Risks from Exposure to Low Levels of Ionizing Radiation, National Research Council, National Academies Press, Washington D.C., 2006, p. 15.

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 1 kg of water is irradiated with 1 Gray,
> 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.

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_f) that weighs the damage caused to biological tissue by a particular type of radiation. In the case of X-rays, γ-rays, β-rays and positrons, the weighting factor is 1; therefore the equivalent dose is the same as the absorbed dose. In the case of α-rays, which occur naturally and are emitted, for example, by some types of uranium isotopes, the absorbed dose must be multiplied by a factor of 20. This indicates that α-rays and other heavy particles such as neutrons and protons 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; for example, 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 must 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:

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

where $w_{thyroid}$ and w_{brain} indicate how sensitive these organs are to radiation and $H_{thyroid}$ and 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 quantification technologies advance, these factors may change.

The Recommendations of the International Commission on Radiological Protection of 2007 (ICRP 103) lists different coefficients compared to the previous recommendations from 1990 (ICRP 60).[8] In particular, the 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.[8]

[8] The 2007 Recommendations of the International Commission on Radiological Protection. ICRP publication 103. Ann ICRP. 2007;37(2-4):1-332.
1990 Recommendations of the International Commission on Radiological Protection. ICRP Publication 60. Ann ICRP. 1991;21(1-3):1-201.

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

In summary:

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 in 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 of different organs to radiation. It is the sum of the equivalent doses in all irradiated organs multiplied by the respective tissue weighting factors w_i.

II. Modality-Specific Challenges and Siemens Solutions

In this chapter, we will discuss radiation in the context of various imaging modalities and the solutions Siemens has developed to reduce radiation dose to a minimum while still obtaining optimal results.

II.A. Computed Tomography

Because of its undoubted clinical benefits, the number of Computed Tomography (CT) examinations has grown over the years, and today is the biggest medical contributor of collective dose to the population. Therefore, in this first section, in-depth information is provided about dose parameters used in CT and dose reduction features available on Siemens CT scanners.

1. Dose and Radiation Risk in Computed Tomography

We will 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.

CT-Specific Dose Parameters: CTDI and DLP

During a CT scan, cross-sections – or slices – 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 this area due to scattering of the radiation (Figure 7).
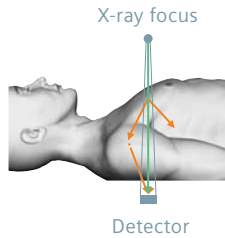
X-ray focus

Detector

Fig. 7
Contribution of direct and scattered radiation to an axial CT slice.

Computed
Tomography

The scattering of the X-rays must be included in calculating 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 Figures 8 and 9), normalized to the nominal slice thickness S.
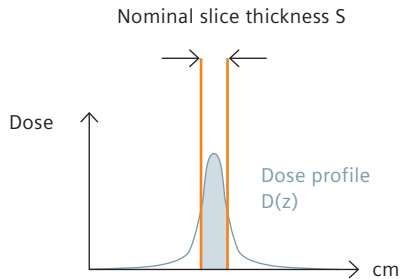

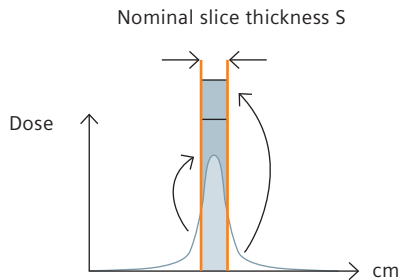**Fig. 8**
Absorbed dose in the slice.

Fig. 9
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 = 1/1000 Gy).

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 Perspex® dummies of 16 cm diameter for the head and 32 cm diameter for the body (Figure 10).
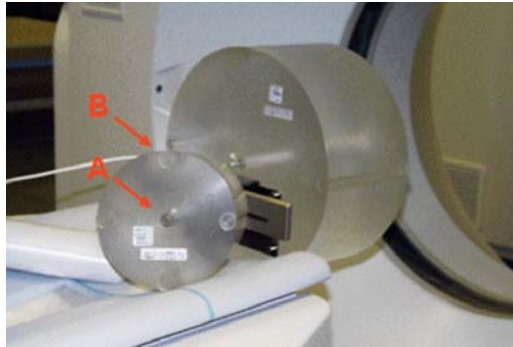
Fig. 10
Perspex® phantoms for measuring the peripheral (B)
and central (A) absorbed dose.

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 center of the patient's body by a weighted sum of the central and peripheral CTDI values.

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

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

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 must be considered: If the table moves slowly, the X-ray beam profiles will overlap (Figure 11).

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.
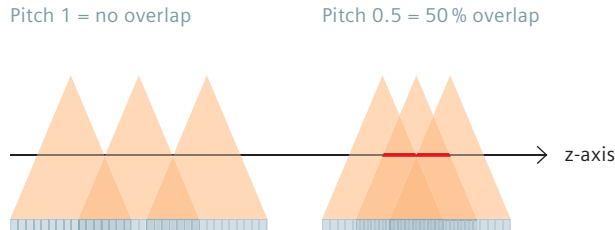
Pitch 1 = no overlap   Pitch 0.5 = 50 % overlap

z-axis

Fig. 11
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:

$$CTDI_{vol} = CTDI_w \cdot 1/pitch$$

If the pitch is smaller than 1, the X-ray beam profiles overlap and the absorbed dose increases. If the pitch is larger than 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 Figure 12).
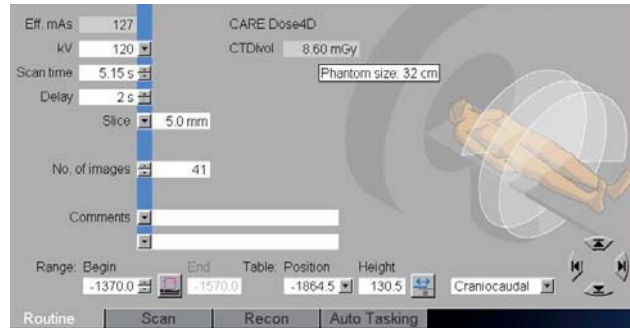

Fig. 12
$CTDI_{vol}$ for the chosen parameters.

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Tomography

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 Figure 13).
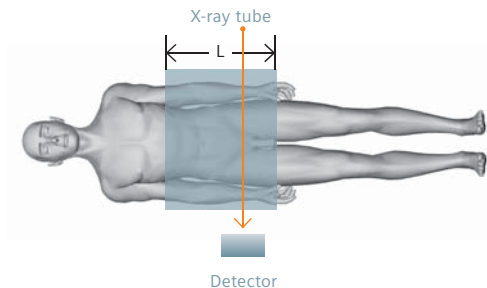

Fig. 13
The X-ray tube and the detector scan the patient along L (examination range) on the z-axis.

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 Perspex® 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.

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

Remember: D = Radiation Energy/kg of matter

Effective Dose in CT

The effective dose in CT takes into account the direct and scattered radiation for all organs in the scan volume. It cannot be calculated exactly for each patient, but a close estimate can be obtained by means of Monte Carlo simulations, assuming an idealized "mean" patient. Figure 14 illustrates a mathematical adult hermaphrodite phantom of the kind used for Monte Carlo simulations of effective doses by the UK National Radiological Protection Board (NRBP) in 1989.
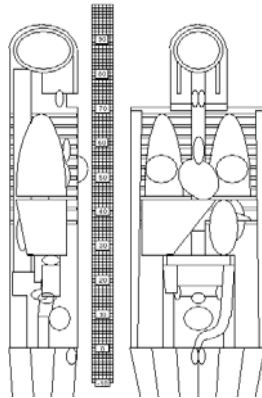

Fig. 14
Typical phantom used to calculate effective doses.

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 Figure 14.

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

Computed Tomography

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[9]:

$$E = DLP \cdot f$$

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

1. **Head: f = 0.0021 mSv/(mGy · cm)**

2. **Neck: f = 0.0059 mSv/(mGy · cm)**

3. **Thorax: f = 0.014 mSv/(mGy · cm)**

4. **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.

[9] Jessen KA et al. EUR 16262: European Guidelines on Quality Criteria for Computed Tomography. Office for Official Publications of the European Communities, Luxembourg, 2000.

Radiation Risk in CT

As listed in Table 2, the effective doses typically used during CT routines are far below the threshold needed for deterministic damage of a part of the body (Table 3).

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?

The answer to this question remains uncertain. There are only a few assumptions and models to quantify this risk.

The most important study that addresses this issue was conducted by Preston et al on 105,000 radiation victims in Hiroshima and Nagasaki, of which 35,000 received radiation doses between 5 and 200 mSv.[10] Unfortunately, this study revealed a high statistical uncertainty in the low-dose range that applies to CT scanning.

There are, according to Muirhead, "... uncertainties about the shape of the dose-response, both for cancer and for non-cancer diseases, below about 100 mSv."[11]

[10] Preston DL et al. Solid cancer incidence in atomic bomb survivors: 1958–1998. Radiat Res. 2007 Jul;168(1):1-64.

[11] Muirhead CR. Studies on the Hiroshima and Nagasaki survivors, and their use in estimating radiation risks. Radiat Prot Dosimetry. 2003;104(4):331-5.

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

In a recent publication, Brenner et al estimated the lifetime risk of death from cancer attributable to a CT scan.[12] Their estimations are shown in Figure 15.
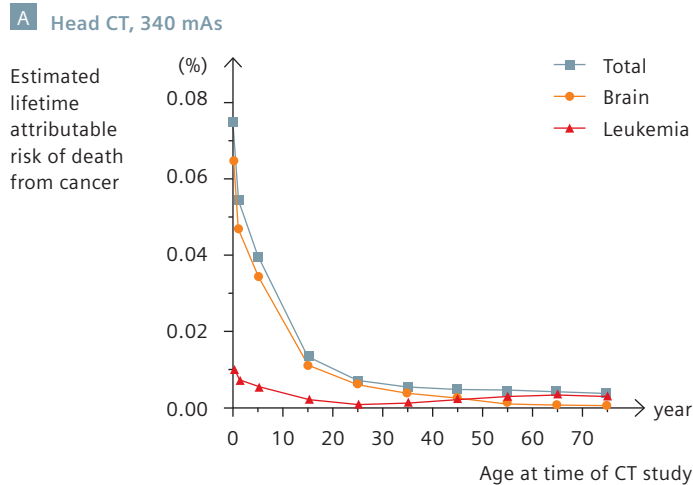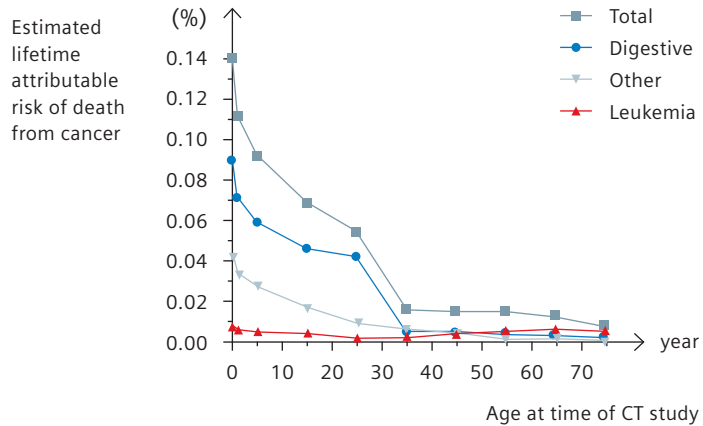
Fig. 15
Estimated risk of death by cancer attributable to a CT scan at different ages.
A: CT of the head at 340 mAs.

[12] Brenner DJ et al. Computed tomography – an increasing source of radiation exposure. N Engl J Med. 2007 Nov 29;357(22):2277-84.

B **Abdominal CT, 240 mAs**

B: Abdominal CT at 240 mAs.

The International Commission on Radiological Protection (ICRP) of 1990 assumed an excess life-time cancer mortality risk of about 5% per Sv. Based on this assumption, a CT examination with 10 mSv increases cancer mortality risk by about 0.05%. This value is in reasonable agreement with Brenner's assumptions (Figure 15).

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.[13]

Therefore, if clinically indicated, the benefit of a CT examination far outweighs 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.

[13] Smith JT. Are passive smoking, air pollution and obesity a greater mortality risk than major radiation incidents? BMC Public Health. 2007 Apr 3;7:49.

Cause of Death	Estimated No. of Deaths per 1,000 Individuals
Cancer[14]	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

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Table 4
Estimated lifetime risk of death from various sources.[15]

[14] Levin B et al. 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 May–Jun;58(3):130-60.

[15] Gerber TC 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 Feb 24;119(7):1056-65.
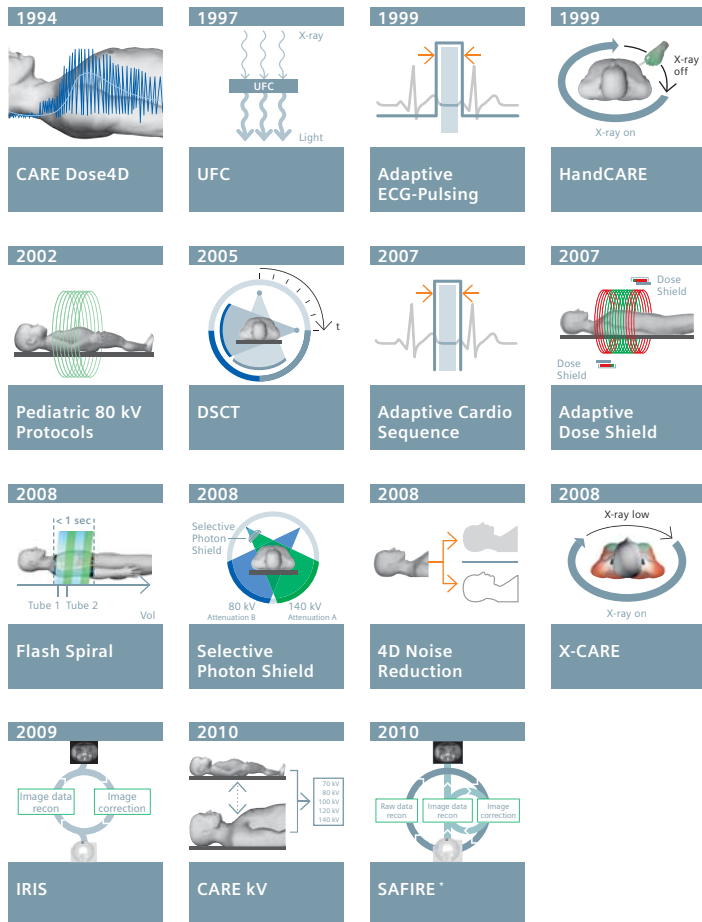
2. Dose Reduction Advances in Computed Tomography

In this section, 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 of the dose reduction methods available in the CT market today. As a leader in the dose reduction field, we also consistently develop our own solutions. As a result, 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.

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 to develop 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 on-going seminars and resources related to dose reduction.

On the following page are brief descriptions of our dose-reduction products and algorithms (more detailed information can be found at **www.siemens.com/low-dose-ct**).

1994

CARE Dose4D

1997

X-ray

UFC

Light

UFC

1999

**Adaptive
ECG-Pulsing**

1999

X-ray
off

X-ray on

HandCARE

2002

**Pediatric 80 kV
Protocols**

2005

t

DSCT

2007

**Adaptive Cardio
Sequence**

2007

Dose
Shield

Dose
Shield

**Adaptive
Dose Shield**

2008

< 1 sec

Tube 1 Tube 2

Vol

Flash Spiral

2008

Selective
Photon
Shield

80 kV
Attenuation B

140 kV
Attenuation A

**Selective
Photon Shield**

2008

**4D Noise
Reduction**

2008

X-ray low

X-ray on

X-CARE

2009

Image data
recon

Image
correction

IRIS

2010

70 kV
80 kV
100 kV
120 kV
140 kV

CARE kV

2010

Raw data
recon

Image data
recon

Image
correction

SAFIRE *

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* The information about this product is being provided for planning purposes.
The product is pending 510(k) review, and is not yet commercially available in the U.S.

Features pioneered by Siemens to reduce dose in CT include:

1. **CARE Dose4D – Real-time anatomic exposure control**
2. **Adaptive ECG-Pulsing – ECG-controlled dose modulation for cardiac spiral CT**
3. **Adaptive Cardio Sequence – Flexible ECG triggered sequential scan**
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 and SAFIRE[*] – Iterative reconstruction techniques**
8. **CARE kV – Automated dose-optimized selection of the X-ray tube voltage**

CARE Dose4D – Real-time Anatomic Exposure Control

The most efficient way to reduce radiation dose in CT is through 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. For example, they may not be aware that, when scanning an area where the patient's diameter has decreased by only 4 cm, the tube output can be reduced by a factor of two while still maintaining adequate image quality. 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.

[*] The information about this product is being provided for planning purposes. The product is pending 510(k) review, and is not yet commercially available in the U.S.

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 for efficiently reducing dose in the shoulder and pelvic region, where the lateral attenuation is much higher than the anterior-posterior attenuation. Figure 16 demonstrates the working principle of CARE Dose4D. Figure 17 is a clinical example obtained with the use of CARE Dose4D.

Computed Tomography
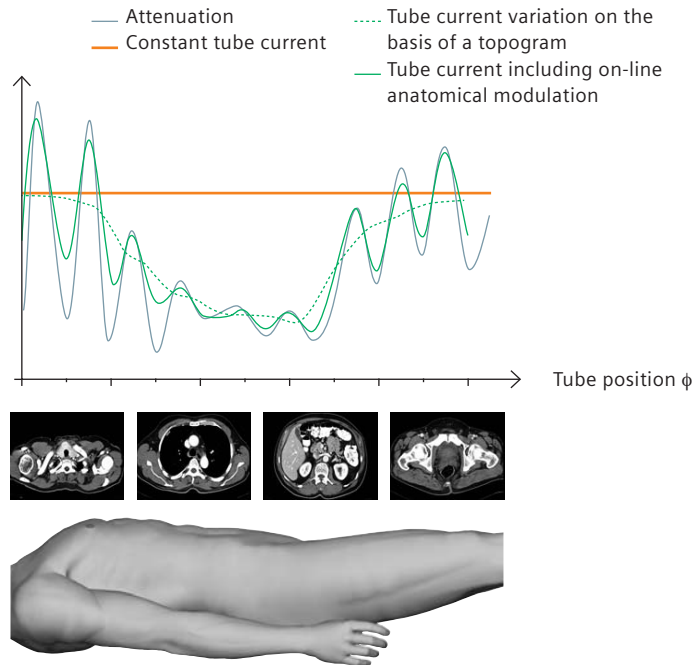
Fig. 16
Illustration of the working principle of CARE Dose4D. With constant tube current, regions in the shoulder and the pelvis would be under-dosed, while the thorax and abdomen would be significantly over-dosed. On-line anatomical dose modulation efficiently adapts the tube current and hence the radiation dose to the patient's attenuation.

140 mAs   55 mAs

110 mAs   130 mAs
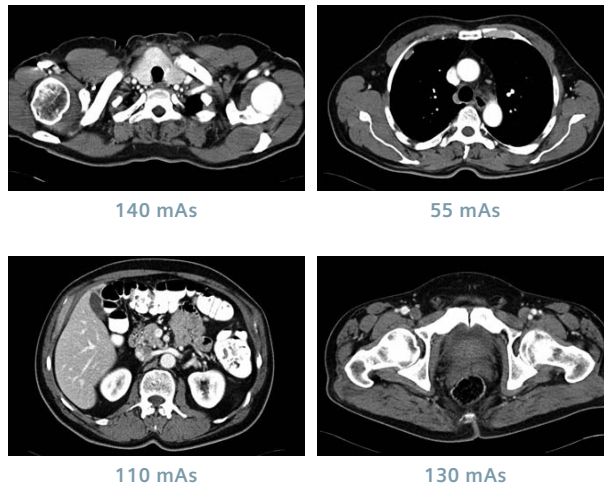
Fig. 17
CARE Dose4D for a scan from the shoulders to the pelvis produces optimized radiation dose for all anatomic regions.

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 (Figure 18).
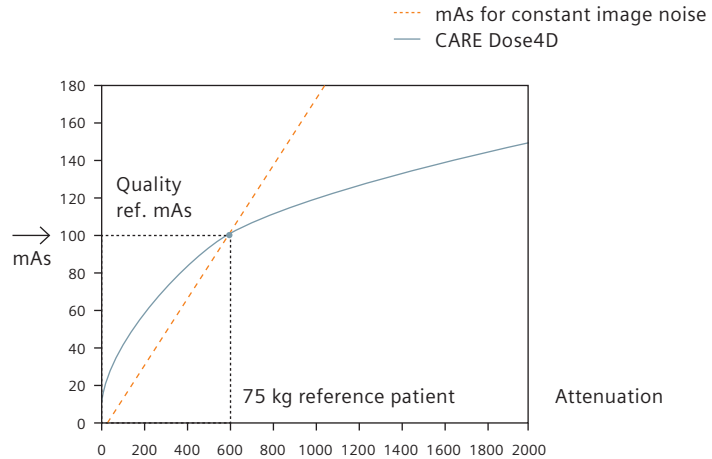
Fig. 18
Dose adaptation with CARE Dose4D.

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% (Figure 19).[16]

[16] Jakobs TF 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 May;12(5):1081-6.

Computed Tomography

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 that react flexibly to arrhythmia and ectopic beats and have the potential to considerably enhance the clinical application spectrum of ECG-controlled dose modulation.
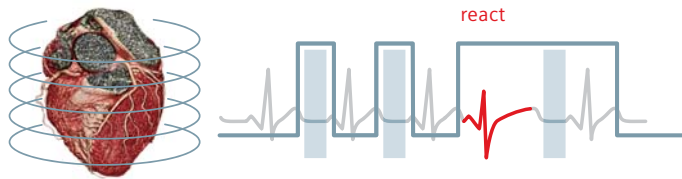
Fig. 19
The CT generates images only during a pre-defined phase of the heartbeat. During this phase, the tube current (gray line) is 100% of the necessary level to achieve adequate image quality, but between these pre-defined phases, the current is reduced to 20% or even 4%. Recently introduced algorithms can react flexibly to arrhythmia.

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 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 either skipped if the ectopic beat happens earlier than the predicted scan start, thus saving unnecessary dose, or repeated at the same position.

Hence, the application spectrum of ECG-triggered sequential scanning is extended to patients with high and irregular heart rates.
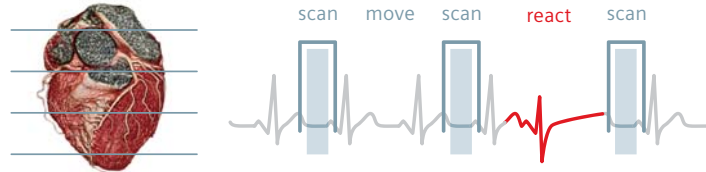

Fig. 20
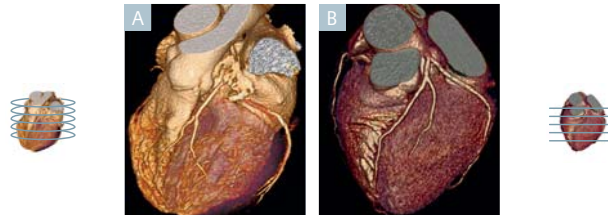Each slice of the heart is scanned during the same ECG phase.

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Tomography

Fig. 21
Using Adaptive ECG-Pulsing, an ECG-gated spiral scan of the heart (A) can be performed at a dose of 4–9 mSv.[17] With the Adaptive Cardio Sequence, an ECG-triggered sequential scan of the heart (B) requires a dose of only 1–3 mSv.[18]

[17] Stolzmann P et al. Radiation dose estimates in dual-source computed tomography coronary angiography. Eur Radiol. 2008 Mar;18(3):592-9. Leschka S et al. Low kilovoltage cardiac dual-source CT: attenuation, noise, and radiation dose. Eur Radiol. 2008 Sep;18(9):1809-17.

[18] Stolzmann P et al. Dual-source CT in step-and-shoot mode: noninvasive coronary angiography with low radiation dose. Radiology. 2008 Oct;249(1):71-80.

Adaptive Dose Shield – Dynamic 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. This problem is typical for spiral CT and commonly referred to as "over-ranging" (Figure 22).
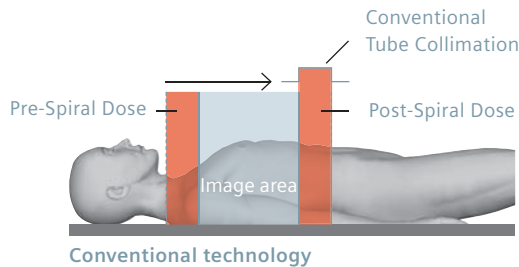

Conventional technology

Fig. 22
Conventional pre-patient collimator. The areas marked in red are out of the necessary scan range but still irradiated with full power.

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 (Figure 23).
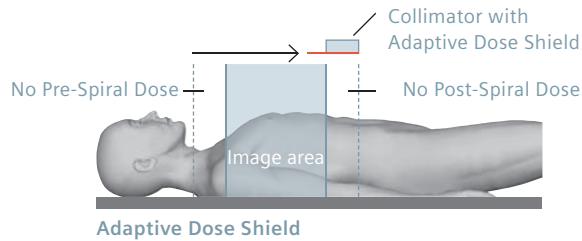
Collimator with
Adaptive Dose Shield

No Pre-Spiral Dose

No Post-Spiral Dose

Image area

Adaptive Dose Shield

Fig. 23
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
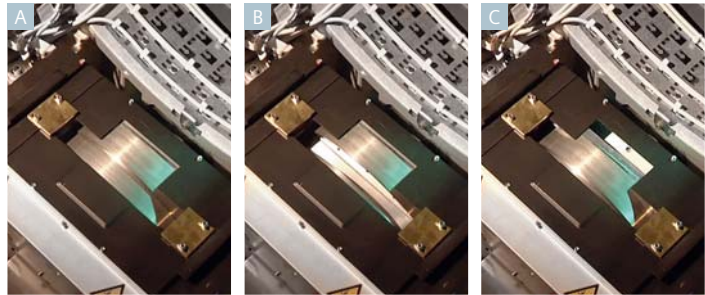the collimator closes asymmetrically.

Fig. 24
The two collimators of the Adaptive Dose Shield.
A: Closed
B: Open left
C: Open right

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Tomography

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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 (Figure 25).
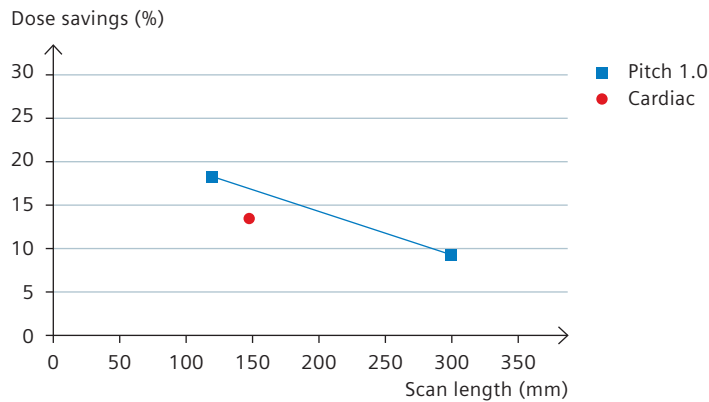

Fig. 25
Dose savings with Adaptive Dose Shield for different CT scan lengths.

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 2.0 to ensure gapless volume coverage along the z-axis. If the pitch is increased, sampling gaps occur (see Figure 26) 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 Figure 26). In this way, the pitch can be increased up to 3.4 in a scan field of view (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.
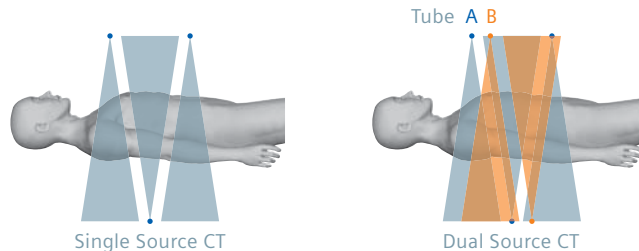

Tube A B

Single Source CT   Dual Source CT

Fig. 26
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.

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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, and the phase of images adjacent in the z-direction is slightly shifted (Figure 27). 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. Initial publications have demonstrated that reliable coronary CT angiography (CTA) is feasible at radiation dose values below 1 mSv.[19, 20]

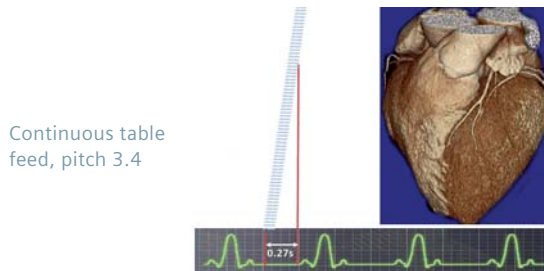Continuous table feed, pitch 3.4

Fig. 27
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 preselected cardiac phase after acceleration to maximum table speed. Data acquisition begins at this pre-selected z-position. Because of the rapid movement of the table, 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 from 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.

Figure 28 shows images reconstructed in this mode with an acquisition time of 250 ms, a temporal resolution of 75 ms, 100 kV and a resulting effective dose of 0.8 mSv.
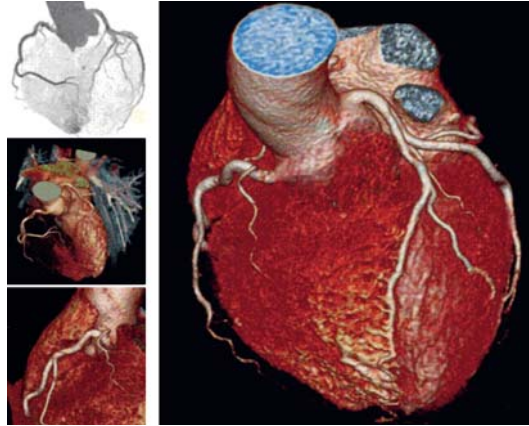

Fig. 28
CT angiography of the coronary arteries acquired with the high pitch DSCT spiral mode (Flash Spiral). Courtesy of Prof. S. Achenbach, Erlangen, Germany.

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

[19] Achenbach S et al. Coronary computed tomography angiography with a consistent dose below 1 mSv using prospectively electrocardiogram-triggered high-pitch spiral acquisition. Eur Heart J. 2010 Feb;31(3):340-6.

[20] Leschka S et al. Diagnostic accuracy of high-pitch dual-source CT for the assessment of coronary stenoses: first experience. Eur Radiol. 2009 Dec;19(12):2896-903.

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Tomography

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 – even without being the object of interest – is irradiated and should therefore be especially protected. Siemens X-CARE, an organ-based dose modulation mode, 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 Figure 29.
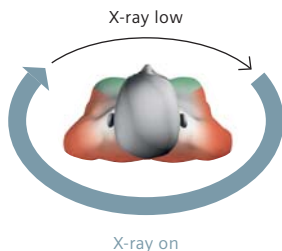

Fig. 29
Illustration of the X-CARE principle.

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 Figure 30.
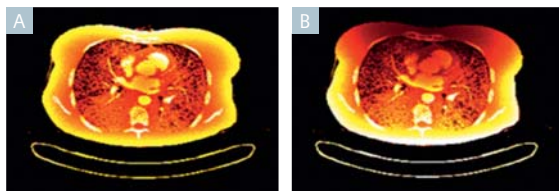

Fig. 30
A: Radiation doses without X-CARE and **B:** with X-CARE. Darker areas indicate lower absorbed dose.

IRIS and SAFIRE[*] – Iterative Reconstruction Techniques

Image quality in CT is mainly determined by spatial resolution and image noise. Conventional reconstruction methods in CT – filtered back-projection (FBP) – are limited by a trade-off: Higher spatial resolution comes with higher image noise. Iterative reconstruction approaches on the other hand, allow decoupling of these two parameters. A correction loop is introduced into the image generation process to reduce image noise in multiple iterative steps – without deteriorating spatial resolution.

IRIS (Iterative Reconstruction in Image Space) and SAFIRE[*] (Sinogram-Affirmed Iterative Reconstruction) are unique methods that reduce image noise without loss of image quality or detail visualization. The significant image noise reduction provided by IRIS and SAFIRE[*] 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 extremely high computer processing power (Figure 31).
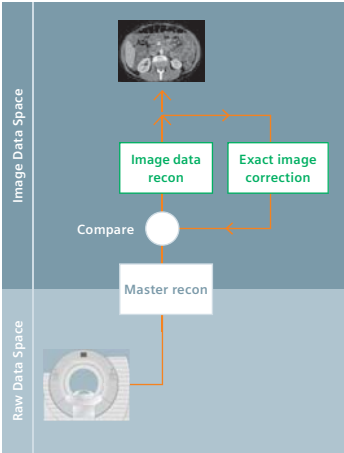
Computed
Tomography
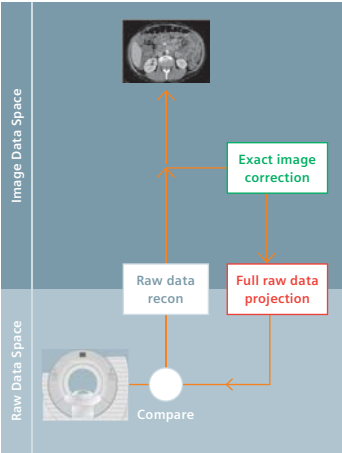
Fig. 31
IRIS – Comparison of the Theoretical Iterative reconstruction with IRIS –
Siemens entrance in the iterative reconstruction world.

Siemens developed IRIS as a unique method to translate the iterative reconstruction loop into the image domain, hence avoiding the time-consuming traditional re-projection. IRIS offers both significant image noise reduction and fast reconstruction for routine clinical use. In addition, the noise texture of the images is similar to standard well-established convolution kernels. The 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 (3 to 5 iterations), enhancing object contrast and reducing image noise with each iteration. IRIS is more advanced than simplified iterative reconstruction attempts because of the special master reconstruction that is used to start IRIS and the special iterative structure of the image enhancement steps (Figure 31 and Figure 32).

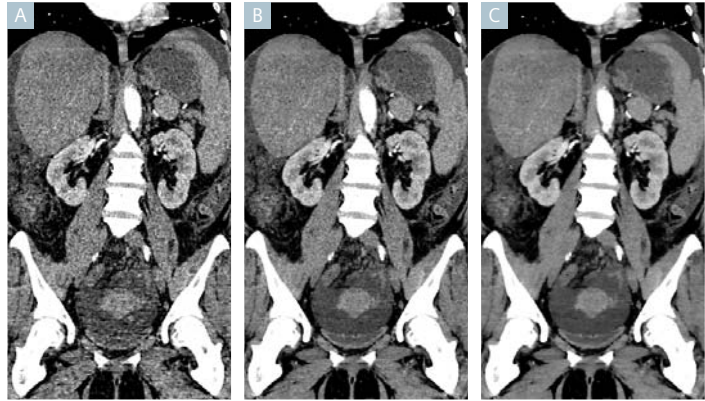**Filtered Back Projection   40% less noise with IRIS**

Fig. 32
Contrast-enhanced CT scan of the abdomen. **A:** Standard Filtered Back Projection Reconstruction, kernel B31. **B:** IRIS reconstruction, **C:** SAFIRE* reconstruction. Note the significantly decreased image noise from each approach without loss of resolution.

SAFIRE* is the first raw-data-based iterative reconstruction. This new generation of image reconstruction hardware allows for the first time the use of raw data during the iterative construction loop. SAFIRE reduces noise and significantly improves image quality – for both robust dose reduction AND high performance.
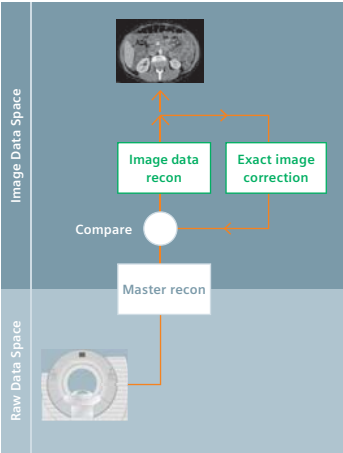
* The information about this product is being provided for planning purposes.
 The product is pending 510(k) review, and is not yet commercially available in the U.S.

Theoretical IR
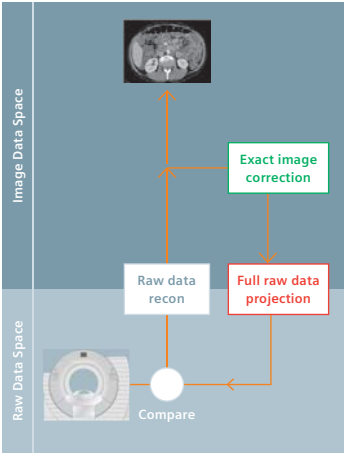
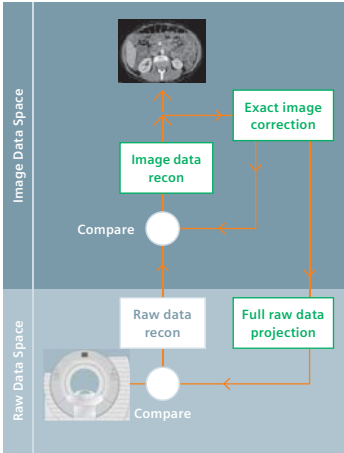
- Dose reduction or image quality improvement
- Well-established image impression
- Very time-consuming reconstruction

IRIS

- Dose reduction or image quality improvement
- Well-established image impression
- Fast reconstruction in image space

SAFIRE*

- More powerful dose reduction than image-based methods
- Well-established image impression
- Superior image quality
- Fast reconstruction in image and raw-data space and improved workflow with variable settings

Fig. 33

SAFIRE* – Siemens approach to iterative reconstruction now adding up raw-data utilization.

With further enhanced algorithms and by better using the now available hardware capacities, also raw-data information (typically visualized in a so-called sinogram) can be used in the iterative image improvement process with SAFIRE*. In addition to the well-established approach of IRIS (also improved and optimized now), the data is now also re-projected in the raw-data space, allowing to additionally validate (or affirm) the images with the measured data. The detected deviations are reconstructed using the Weighted Filter Back Projection, yielding an updated image. Image noise can be further subtracted by correcting geometrical imperfections of the initial reconstruction. This also applies for potential artifacts that occur with any system using the FBP. With this, SAFIRE also allows to reduce dose by up to 60%, but for a much wider range of applications and with superior image quality, even surpassing the already impressive image quality achieved with IRIS.

* The information about this product is being provided for planning purposes. The product is pending 510(k) review, and is not yet commercially available in the U.S.

CARE kV – Automated Dose-Optimized Selection of the X-Ray Tube Voltage

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 great 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 any or all of these parameters will render a better image and enable the reading physician to make a more precise diagnosis. For example, when 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 angiography). 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 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 (Figure 34).

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, 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 therefore necessary.
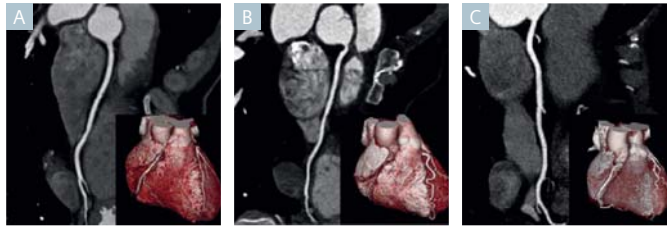
Computed Tomography

Fig. 34
Three CT angiographies with three different current and voltage settings.
Note that the mean contrast in the aorta is higher with 100 kV.

A: 120 kV 330 mAs, $CTDI_{vol}$ = 43.1 mGy, mean contrast aorta: 322 HU

B: 100 kV 330 mAs, $CTDI_{vol}$ = 31.8 mGy, mean contrast aorta: 561 HU

C: 100 kV 230 mAs, $CTDI_{vol}$ = 21.2 mGy, mean contrast aorta: 559 HU

3. 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. 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, for example, offers effective doses below 0.5 mSv in pediatric applications, with full diagnostic image quality. Because of the fast scan speed using very high pitch values (Flash Spiral), even uncooperative children can be examined without sedation, reducing stress for the patient and saving time and money.

Dose Structured Reports

On SOMATOM scanners, the reporting of established dose parameters such as Computed Tomography Dose Index (CTDI) and dose length product (DLP) has been implemented since 1990. For each exam, the information is available in the patient protocol, and can be viewed and archived after the scan.

As the first CT manufacturer, Siemens now provides the new Dose Structured Reports (Dose SR) for nearly all products within our CT portfolio.

Dose SR contains comprehensive data for each irradiation event, including the accumulated dose and basic information about the context of the exposure. One radiation event is defined as one continuous irradiation applied to a patient; for example, a CT topogram and the associated spiral scan are two separate events. The details include patient demographics, study information, imaging technique and geometry and all typical dose metrics (e.g., $CTDI_{vol}$, DLP). The data is provided in electronic format that can be sent to any system which receives, stores or processes dose information, such as conventional PACS or workstations for further analysis.

CARE Analytics

The Dose SR can serve as the central component of an institution-wide dose quality control initiative. To evaluate and analyze the information, Siemens provides a new tool, CARE Analytics. It is a stand-alone tool and can be installed on any workstation or office computer.

With CARE Analytics, the data of the Dose SR can be automatically processed or included in institution-specific databases. It gives a comprehensive overview of the applied radiation dose of different examinations and provides information to optimize scan protocols in a very efficient way. Thus, it enables greater transparency in terms of radiation dose. With the implementation of Dose SR and CARE Analytics, SOMATOM scanners are facilitating the next step in radiation dose management.

Computed Tomography

II.B. Dose Documentation for SOMATOM Definition Systems with Utilization Management

Siemens Utilization Management

Siemens Utilization Management (UM) is a proactive service offering from Siemens UPTIME Services that provides system-specific usage data which have been collected electronically using Siemens Remote Service (SRS). Access to these detailed data enables customers to leverage their systems' full potential. Customers receive extensive equipment utilization analyses and anonymous benchmark information about comparable systems at other facilities operating in similar environments, as well as dose information on a regular basis.

The reports are generated by the Siemens service organization and made available through personalized access to the LifeNet UPTIME Services Portal. This is Siemens' secure web portal that gives customers the information they need to manage the productivity of their Siemens diagnostic equipment – anywhere internet access is available.

A New Feature: The Dose Report

The dose report is a new feature of Utilization Management (UM) that keeps users aware of the data regarding total radiation exposure and dose usage on a monthly or a multi-monthly basis.

With UM, dose information can be accessed in all basic and advanced reports for SOMATOM Definition systems (from SW version *syngo* CT 2008B). It is an easy way to track system and dose utilization details at the point of care. With these reports, users can see if any specific values have been exceeded, indicating that a significant radiation dose has been administered.

With the new dose report, operators obtain an overview of the total number and the percentage split of all selected protocols, as well as the available dose information for the dedicated period. The relevant dose values for a CT examination are the CT dose index volume ($CTDI_{vol}$) and the dose length product (DLP).

Key benefits for the user include:

1. **Visibility of dose usage data in a monthly or multi-monthly period; increased awareness of radiation exposure among clinicians**

2. **Easy way to track system and dose utilization details at the point of care**

3. **Increased patient safety**

4. **Need for specific procedures' dose details is satisfied**

5. **Available in four languages (German, English, French, Spanish)**

Through customized service agreements, Siemens UPTIME Services can offer almost unlimited options, from proven service modules and freely adaptable solutions to shared service options that support in-house staff in performing individual tasks themselves. More information is available at **www.siemens.com/Performanceplans**.

Dose
Documentation

II.C. Molecular Imaging – Nuclear Medicine

1. Dose and Radiation Risk in Molecular Imaging – Nuclear Medicine

Unlike in CT, the radiation burden in nuclear medicine does not directly depend on the imaging device, but on the administered radiopharmaceutical and its properties. This is independent from the number of acquired images. The effective dose is proportional to the administered activity, which indicates the amount of the isotope. Activity is measured in Becquerels (Bq), where 1 Bq equals 1 decay per second, and Curies (Ci), where 1 Ci equals 3.7×10^{10} decays per second.

Additional physical parameters are physical half-life, and type and properties of the emitted radiation. Therefore, isotopes with a physical half-life in the range of the total examination duration should be applied in order to prevent additional unused radiation that would result from using isotopes with longer half-lives. In addition, pure gamma emitters such as ^{99m}Tc or pure positron emitters such as ^{18}F are preferred. Additional radiation components do not contribute to the diagnostic image, but contribute to the overall effective dose.

Biological parameters are the pharmacokinetic properties of the radiopharmaceutical, which influence the biological half-life and distribution pattern, and change dynamically after the application. Effective half-life combines the effects of physical decay (physical half-life) and excretion (biological half-life):

$$1/T_{1/2eff} = 1/T_{1/2phys} + 1/T_{1/2biol}$$

The calculation of effective dose in an individual is extremely cumbersome, because in addition to the above-mentioned parameters, anatomy (organ sizes and form, distances) also influences the result. Dose calculation not only reflects the tissue content of the tracer and its self radiation, but also the dose delivered to the rest of the body by radiation types with a long range (γ radiation and annihilation radiation). Thus, some simplifications – such as standardized anatomy – are made for the assessment of dose for diagnostic purposes, as in Organ Level INternal Dose Assessment (OLINDA).

Package inserts describe the effective doses per administered MBq as well as the organ doses per MBq for, at minimum, critical organs for a standard patient with respect to size, weight, distribution pattern and excretion.

The impact of the system on the dose is only influenced by the system's minimum activity requirements for a given image quality and acquisition duration.

As in planar X-ray diagnostics and CT, there is a wide range of typical effective doses for different nuclear medicine procedures. Some are far below 1 mSv (e.g. the Schilling test), while others may exceed 10 mSv (e.g. gallium scintigraphy). Most procedures, however, result in doses between 1 and 10 mSv.

In hybrid devices, the burden of radiation stems not only from the use of radiopharmaceuticals but also from the CT component as well. Thus, advances in scanner technology allow the reduction of injected activity, and hence, the effective dose. In addition to activity reduction, other considerations include:

1. **The usage of ionizing radiation always requires a justified indication. Can the same clinical result reasonably be achieved using any other examination?**

2. **Can the overall diagnostic dose be reduced by judiciously combining examination types? Good clinical planning may reduce the number of CT examinations if diagnostic CT and CT for hybrid imaging are combined in a single scan.**

3. **Isotopes with a shorter half-life and favorable radiation type can reduce radiation exposure dramatically. For example, the use of ^{123}I-MIBG instead of ^{131}I-MIBG in the diagnosis of neuroendocrine tumors improves image quality and reduces radiation exposure, due to a much shorter half-life (13 hours instead of 8 days) and radiation type (gamma emitter with a main peak of 151 keV versus combined beta and gamma emitter). Another example is the use of FDG-PET instead of ^{67}Ga scintigraphy in lymphoma and inflammation cases. In addition to a much lower radiation exposure, image quality and precision are much better and the overall time from injection to end of scan is substantially reduced (1.5–2 hours versus 3–4 days).**

4. **The choice of the radiopharmaceutical can also reduce the radiation burden. ^{99m}Tc-MAG3 shows a much higher renal uptake and clearance than ^{99m}Tc-DTPA. This allows injected activity to be reduced by a factor of 2 while maintaining the same image quality and precision with respect to urodynamics.**

Molecular Imaging

5. Not every clinical question requires striving to achieve the best possible image quality through the combination of high injected activity and advanced acquisition and reconstruction tools. In some cases, a somewhat reduced image quality, which permits lower injected activity, still fulfills the clinical need.

6. Several countries have issued guidelines for maximum standard activities according to examination type. Pediatric activities can be calculated adherent to special rules. In the past, effective doses in children were rather high due to insufficient adjustments to the guidelines, which have since changed.

2. Dose Reduction Technology for PET•CT

Features pioneered by Siemens to reduce dose in MI include:

1. CARE Dose4D
2. LSO HI-REZ
3. TrueV
4. HD•PET
5. Adaptive Dose Shield
6. Adaptive Cardio Sequence
7. ultraHD•PET
8. IRIS

Biograph

1994	1999	2006	2007
CARE Dose4D	LSO HI-REZ	TrueV	HD•PET

2007	2007	2008	2009
Adaptive Dose Shield	Adaptive Cardio Sequence	ultraHD•PET	IRIS

As concern over radiation dose to patients rises, hybrid modalities are coming under intense scrutiny due to their dual imaging nature – the two components of the PET•CT scan both impose a certain radiation burden. Because of this dual nature, both modalities must maximize the diagnostic information obtained while reducing the injected radiopharmaceutical dose and the applied radiation from the CT scan. Innovations in both areas have allowed the clinician to reduce the radiation dose by 50–60%.

The following sections describe the major innovations contained in the Siemens Biograph® family of PET•CT scanners. Many of the innovations are unique to the Biograph, offering a wide range of flexibility in patient dose reduction and scan acquisition speed.

Biograph mCT offers the ability to reduce the patient dose while simultaneously increasing scan speed. No longer must the choice be made between the two.

Please visit **www.siemens.com/mindosemaxspeed** to find out more.

Molecular Imaging

LSO HI-REZ

In the 1990s Siemens pursued the development of a new scintillator crystal for clinical PET scanning called lutetium oxyorthosilicate (LSO). This new crystal has the properties of emitting more light than previously used materials when struck with a 511 keV photon from a PET radiopharmaceutical. LSO also emits the light more rapidly and has a shorter "afterglow." Both of these qualities enabled the PET detector system to become more efficient, without losing any necessary information or reducing image quality. This, in turn, allowed the injected dose of radiopharmaceutical to be reduced while maintaining image quality and speed of the examination.

Along with dose reduction, the improved characteristics of LSO allowed a second innovation, called HI-REZ, to take place within the detector configuration. The additional light output enabled changes to be made to the crystal configuration while simultaneously increasing the resolution. The detector went from 6 mm square pixels to 4 mm square pixels, enabling an increase in resolution of approximately 250%.
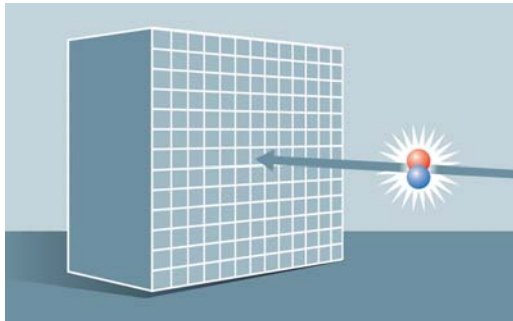

Fig. 35
LSO HI-REZ.

True V

The Biograph family of PET•CT systems possess a unique feature in the market, called TrueV, that is directly aimed at dose reduction. This unique feature is based on extending the field of view (FOV) of the PET detectors in the z-direction of the patient in order to capture more information in each PET bed position. PET technology in the Biograph uses what is known as 3D acquisitions. By employing this method, any extension of the field of view leads to tremendous benefits in data acquisition. Increasing the field of view by 33% actually yields an improvement in scan productivity of about 75%. PET is a modality that uses bed positions or bed stops in order to cover the patient's body from the eyes to mid-thigh for most scans. With TrueV, it takes less bed stops to cover the same patient volume than with other systems. In addition, each bed stop is more sensitive. By combining both of these benefits, the injected dose to the patient can be reduced by 50% while keeping all the other scan parameters the same, including image quality.
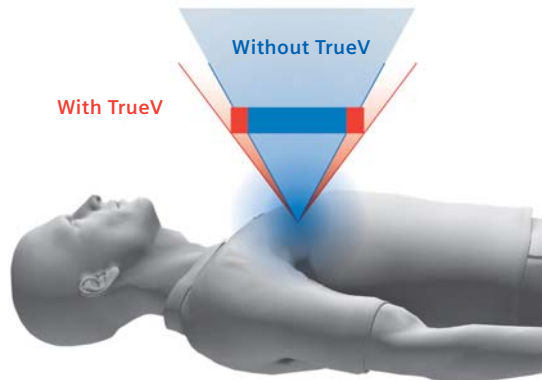

Without TrueV

With TrueV

Fig. 36
True V.

Molecular Imaging

HD•PET

Of utmost importance to the diagnosing physician is the image quality being produced by the PET•CT system. Lowering dose, or scanning for less time is not advantageous if a confident diagnosis cannot be made. Because of this, the Biograph has another unique feature called HD•PET, a resolution recovery technique employing the use of point spread functions during the image reconstruction. By understanding the characteristics of the PET detector system, we are able to improve the resolution on the image, and maintain that resolution across the entire field of view. Without using a spatially variant point spread function in the reconstruction, there is a sharp drop-off of measured resolution further away from the center of the field of view. Another important benefit of the HD•PET innovation is that as it recovers the lost resolution from the detection system, it also reduces image noise as well. This is because the reconstruction process can do a better job at reproducing the reality of the measurement, knowing the precise characteristics of the detection system.
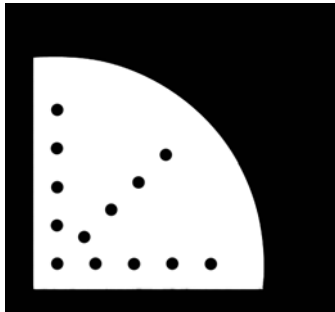

Fig. 37
HD•PET.

ultraHD•PET

ultraHD•PET further reduces the image noise by using the HD•PET algorithm, which improves image resolution and uniformity across the field of view (FOV), and adds time-of-flight (TOF) information in the PET data. By measuring the arrival times of both incoming photons onto the detectors, the sophisticated electronics are able to determine where the measured event took place along the line of response. Because photons travel at the speed of light, the timing resolution of the PET system must be able to measure a difference in arrival times of the photons in the pico-second range. By being able to localize the event in the body along the line of response with this time-of-flight information, the reconstruction process is able to reduce image noise further to a point where the injected dose to the patient can be reduced by half and the image quality will match a full dose scan.

Biograph mCT is the only PET•CT system that can combine the extended FOV of TrueV with ultraHD•PET to offer a system that can reduce the injected dose to the patient by 50% and offer improvement in scan speed at the same time, eliminating the need to choose between a fast scan or a low-dose scan.
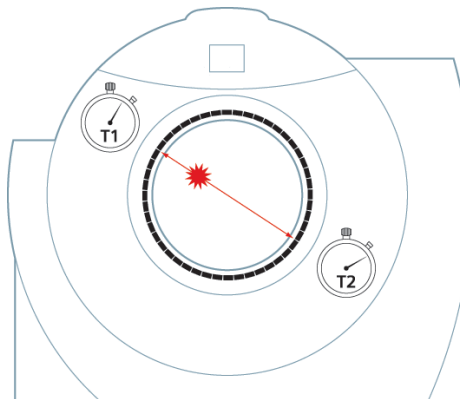
Fig. 38
ultraHD•PET.

Molecular
Imaging

CT Innovations in the Biograph

IRIS, CARE Dose4D, Adaptive Dose Shield, and Adaptive Cardio Sequence have all been explained in the CT chapters of this guide. All of the CT innovations for dose reduction that are available on the Emotion, Sensation and Definition AS line of CT systems are also available on the Biograph mCT.
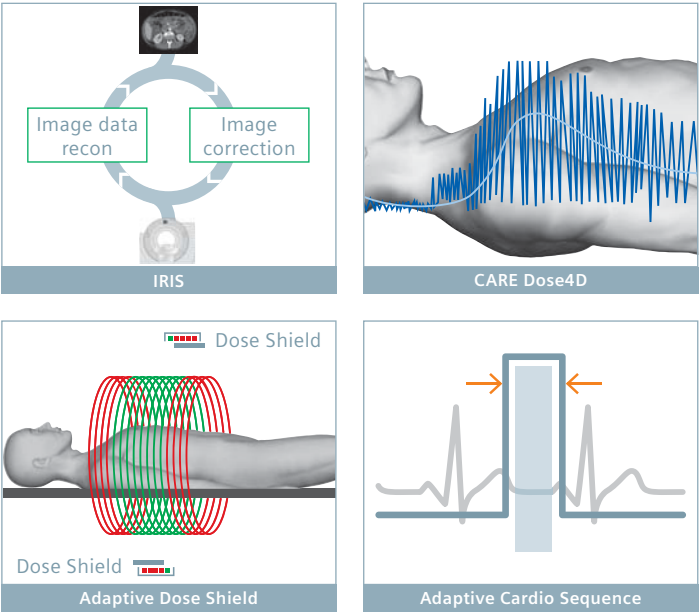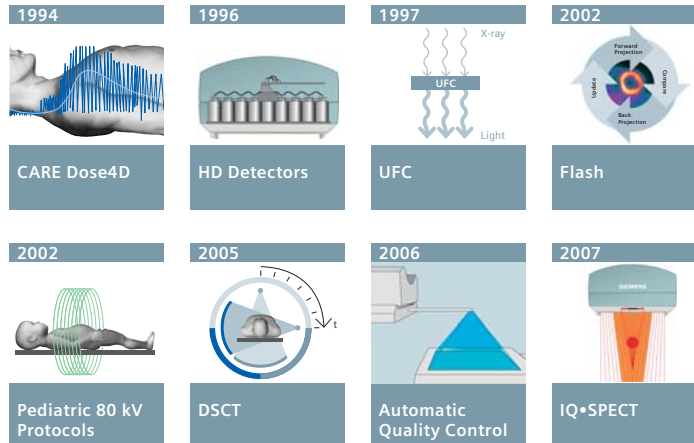

Fig. 39
CT Innovations in the Biograph.

3. Dose Reduction Technology for SPECT

Features pioneered by Siemens to reduce dose in MI include:

1. **CARE Dose4D**
2. **HD Detectors**
3. **UFC Detectors**
4. **Flash Iterative Reconstruction**
5. **Pediatric Protocols**
6. **Diagnostic Spiral CT**
7. **Automatic Quality Control**
8. **IQ•SPECT**

Symbia

1994	1996	1997	2002
CARE Dose4D	HD Detectors	UFC	Flash

2002	2005	2006	2007
Pediatric 80 kV Protocols	DSCT	Automatic Quality Control	IQ•SPECT

Molecular Imaging

SPECT•CT imaging procedures are categorized into four main clinical areas – cardiology, oncology, general purpose imaging and neurology. Minimizing dose is equally important in all clinical segments, but – unlike CT or MRI – hybrid imaging in nuclear medicine (SPECT•CT and PET•CT), permits dose to be optimized in two ways: by reducing the dose of the injected radiopharmaceutical and by minimizing the CT dose emitted from the hybrid scanner. While pressure is increasing to minimize the CT dose, it is also critically important to reduce the injected dose in SPECT imaging to ensure patient safety.

Additionally, but no less importantly, there are only five nuclear reactors worldwide that produce 100% of the most commonly used SPECT tracer – technetium (^{99m}Tc). Some of these reactors have been shut down for temporary maintenance, and the availability of ^{99m}Tc, consequently, has been reduced worldwide. This limited availability has brought a need for more efficient use of technetium. Therefore, injecting less dose per patient is very important so that hospitals can continue to scan their full patient load.

Siemens Symbia® SPECT•CT product line is pioneering dose savings in both SPECT and CT. Not only do we have the most integrated diagnostic SPECT•CT systems, but we also provide features that can save up to 75% of the dose while maintaining image quality. This section provides a short overview of the dose-saving features of Symbia.

Please visit **www.siemens.com/mindosemaxspeed** to find out more.

HD Detectors

Symbia's latest HD Detector technology offers outstanding and consistent image quality. The energy-independent response of the highly integrated detector electronics eliminates isotope-specific floods. Real-time corrections and individual photomultiplier tuning further minimize scheduled system calibrations and user interaction. Together with Siemens' AUTOFORM collimators, Symbia has extremely high sensitivity, enabling more counts and superb image quality.

Symbia HD detectors have up to 26% higher sensitivity, allowing any one of the following benefits:

1. **Lower injected dose**

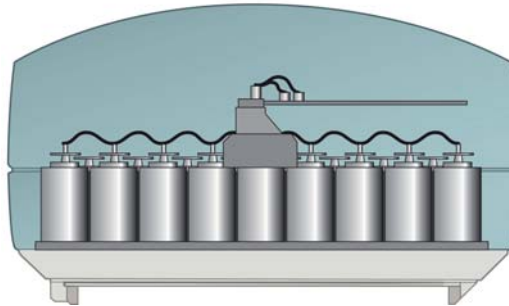2. **Fast image acquisition for more patient comfort**

Fig. 40
HD Detectors.

Flash Iterative Reconstruction

Flash, a leading iterative SPECT reconstruction method, significantly improves image quality. Reconstruction image fidelity depends on the accuracy of the physical models used in image formation. *onco*•Flash and *cardio*•Flash provide higher spatial resolution, reduced distortion and reduced artifacts. As a result, Flash images are more accurate and easier to interpret. Flash technology restores image quality from count-reduced patient scans acquired in less time or with lower injected dose. Planar, whole-body and SPECT studies performed with Flash achieve better image separation, better resolution and better contrast.

Flash enables any one of the following benefits:

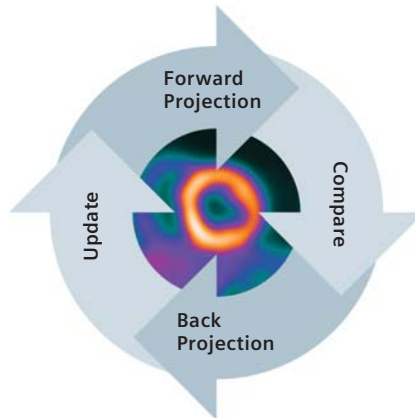1. **Half the injected dose**

2. **Double the scan speed**

Fig. 41
Flash Iterative Reconstruction.

Automatic Quality Control

With automated daily, weekly and monthly tests, quality control (QC) is performed routinely and consistently. Automatic Quality Control (AQC) automatically starts the specified QC tasks so that they are finished before the first scheduled patient arrives. AQC uses built-in shielded sources: a point source for peaking and tuning, a line source for extrinsic flood and a sleeve with five slits for center of rotation and multiple head registration. The process concludes with a report of the completed QC results and integrates with Siemens Remote Service (SRS).

The benefits of AQC include:

1. **Reliable, consistently reproducible QC**

2. **Eliminated risk of spillage with open sources**

3. **Operator dose reduction**

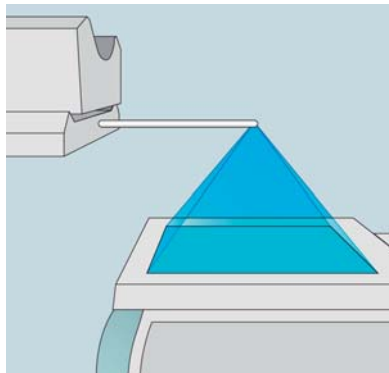4. **Up to 19.5 hours saved per month**

Fig. 42
Automatic Quality Control.

IQ•SPECT

With IQ•SPECT, more information is received from the heart in five minutes than with a conventional SPECT in 20 minutes. Its dual **SMART**ZOOM collimators work in tandem to track the heart and keep it in the "sweet spot" at all times so the acquisition can be quick and accurate, resulting in superb patient comfort and high throughput. Its innovative **SMART**ZOOM collimators, placed in a cardio-centric orbit, collect maximum information from the heart in the least amount of time. IQ•SPECT needs only four minutes for a full-count SPECT scan and just 60 seconds more for CT-based attenuation correction and calcium scoring.

IQ•SPECT provides four times greater sensitivity and any one of the following benefits:

1. Half dose and double scan speed imaging at the same time

2. Up to four times faster scan times

3. Up to 50% decrease in injected dose

Fig. 43
IQ•SPECT.

CT Innovations in the Symbia

CARE Dose4D, pediatric protocols and Ultra Fast Ceramic (UFC) detectors have all been explained in the CT chapters of this guide. All of the dose reduction CT innovations above are available in the Symbia SPECT•CT systems.
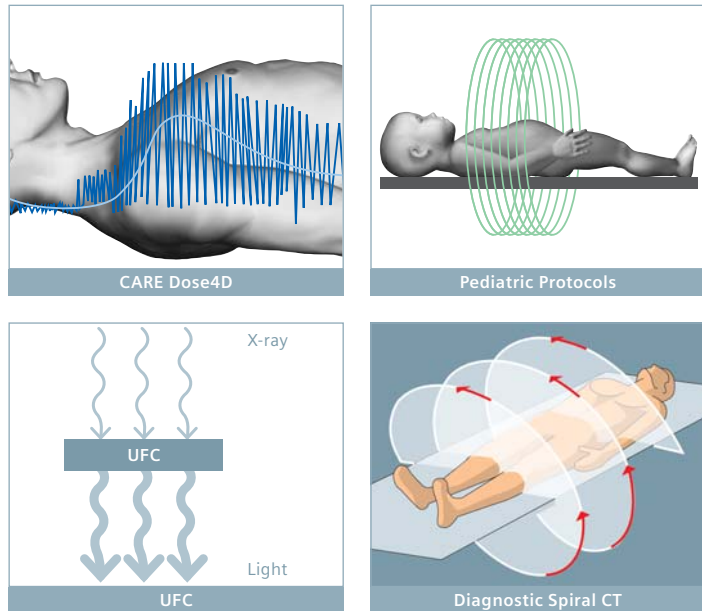

Fig. 44
CT Innovations in the Symbia.

II.D. Angiography

1. Dose and Radiation Risk in Angiography

Angiography-Specific Dose Parameters

Several dose parameters are specific to angiography: the detector dose, the skin dose, the dose rate and the DAP (dose area product), which will be covered in the following sections.

Dose and Image Quality

In the past, when angiography was performed using traditional photographic film technology, the general rule was the higher the dose, the better the image quality.

With today's improvements in imaging technology, is there still a trade-off between improving quality and saving dose?

<div align="center">**Yes**</div>

In general, low dose goes hand in hand with less visibility, while higher image quality requires, among other factors, a higher dose. To obtain a specific image quality, it is necessary to choose the "right" dose for the tissue being penetrated.

Therefore, the solution is to make best use of dose and equipment.

Good image quality can be expressed in terms of detectability. Rose found:[21]

$$\text{Detectability} \sim \text{physical contrast} \cdot \text{object diameter} \cdot \sqrt{\text{dose at detector}}$$

(The physical contrast is the difference in X-ray absorption between rays of the beam running through the object of interest and the rays next to it.)

What does this mean for angiography? Lesion detectability is directly related to detector dose, given a certain physical contrast and diameter of a vessel. In general good contrast, low noise and high spatial resolution are necessary for good image quality. As a result, even fine details are visible.

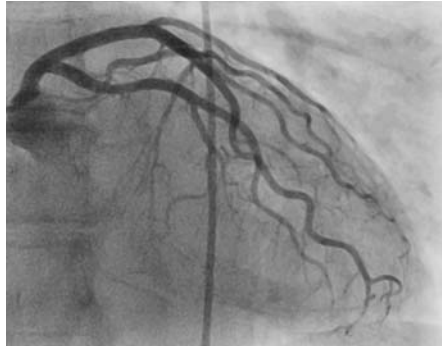[21] Rose A. Vision, human and electronic. Plenum Press, New York, 1973.

Fig. 45
Image example with good contrast, low noise, and high spatial resolution.

Skin Dose and Safety During Fluoroscopy

Modern detector systems make it easy to obtain high quality results simply by selecting the required image quality by choosing an appropriate protocol; the requested dose at the detector entrance will automatically be kept constant (as much as possible) by adjusting the tube output dose. This automatic dose control compensates for patients of different body sizes.

The dose is highest at the point where the beam enters the patient. The absorbed dose at this beam entrance is an important measure: It signifies the accumulated skin dose over the length of the procedure, measured in mGy (1000 mGy = 1 Gy). This accumulated skin dose is relevant for determining the skin burn damage resulting from the intervention.

Detector systems display and report only an estimate of the skin dose at the interventional reference point (IRP). Values at the actual exposure entry location can be different depending on the patient's body shape and other geometric measures, such as table and C-arm position.

Angiography

Figure 46 shows the location of measurement for dose and dose rate.

1. **The red point at the X-ray tube housing indicates the position of the focal spot.**

2. **The source-image distance (SID) is the distance between the focal spot and the image receptor. On the Artis zee, this receptor is the flat detector.**

3. **ISO Center refers to the isocenter of the C-arm; i.e., the central point around which the C-arm rotates.**

4. **The IRP is 15 cm beneath the isocenter, and is assumed to be the skin entrance point. The calculated estimates for the displayed dose values refer to the IRP.**
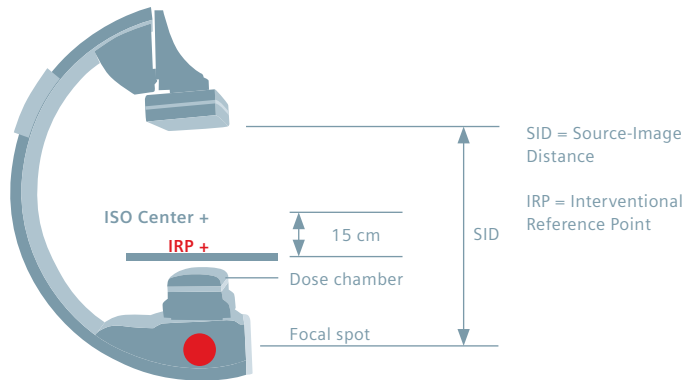

Fig. 46
Schematic C-arm, detector, tube, table; location of dose and dose rate measurement.

The IRP is the measuring point for:

1. **Dose area product (DAP) in µGy m^2**
2. **Dose in mGy**
3. **Dose rate in mGy/min**

> **Note:** **The IRP does not change with table height.**

As a general rule, the closer the beam entrance to the tube, the higher the real skin dose and vice versa.

The Siemens Artis system has two built-in safety regulations for fluoroscopy:

1. **By default, the dose rate at a specific location (30 cm in front of the detector) is limited to a certain level (e.g., 10 R/min = 87 mGy/min for the U.S. and European countries). It is assumed that this point is identical to the patient's skin entry point. It is possible to increase the dose rate by using the "Fluoro +" button (High contrast button). An audible warning occurs.**
2. **After every five minutes of fluoroscopy, a message pops up on the display and a sound is emitted to remind the user of the applied dose. If the operator does not acknowledge this signal, radiation exposure stops after next five minutes of fluoroscopy.**
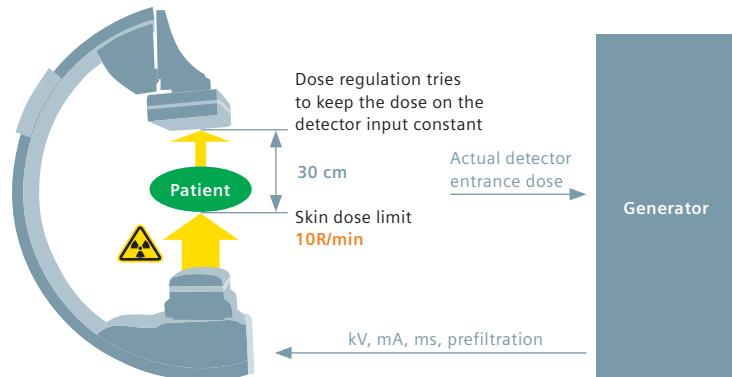

Fig. 47
Principle of dose regulation with constant input dose at the detector and 10 R/min dose rate limit 30 cm in front of the detector.

Angiography

Dose Rate and Dependence of Absorption on Patient's Thickness

Mean skin dose has been reported for interventions in the brain. Although head size varies little among individuals, body size varies greatly.

Figure 48 shows that an increase in patient's thickness of about 3 cm results in twice the entrance dose for a constant detector entrance dose. This rule of thumb is based on the assumption that tissue absorbs radiation in a similar manner as water and that a certain quality of beam is applied.
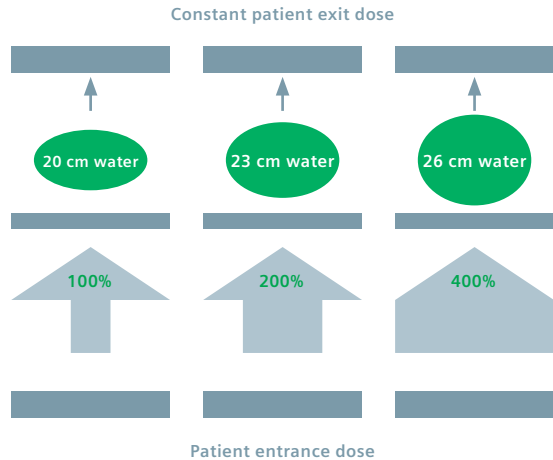
Constant patient exit dose

20 cm water	23 cm water	26 cm water
100%	200%	400%

Patient entrance dose

Fig. 48
Patient entrance dose depends on patient thickness to achieve the same patient exit dose.

A similar effect occurs when the direction of projection is changed to an oblique position (Figure 49). Because the shape of the body is more oval than circular, the length of the X-ray beam is now longer, resulting in a higher entrance dose. True values may differ significantly since the body is not really a homogeneous ellipsoid but consists of bones, organs, etc.[22]
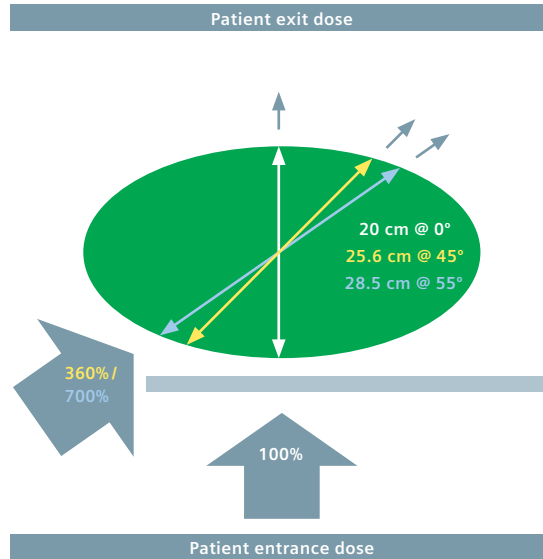

Patient exit dose

20 cm @ 0°
25.6 cm @ 45°
28.5 cm @ 55°

360% /
700%

100%

Patient entrance dose

Fig. 49
Simplified model showing the effect on patient entrance dose when the projection is angulated.

Angiography

[22] Cusma JT et al. Real-time measurement of radiation exposure to patients during diagnostic coronary angiography and percutaneous interventional procedures. J Am Coll Cardiol. 1999 Feb;33(2):427-35.

Dose Area Product and Inverse Square Law

In air X-rays travel in a straight line – thus, their intensity decreases as the distance from the X-ray tube focus increases along with the surface area of the beam. Figure 50 shows the following: the dose D at the distance d from the focal spot F drops to 1/4 D at the distance 2d and to 1/9 of D at three times the distance. This is the "inverse square law for radiation dose."

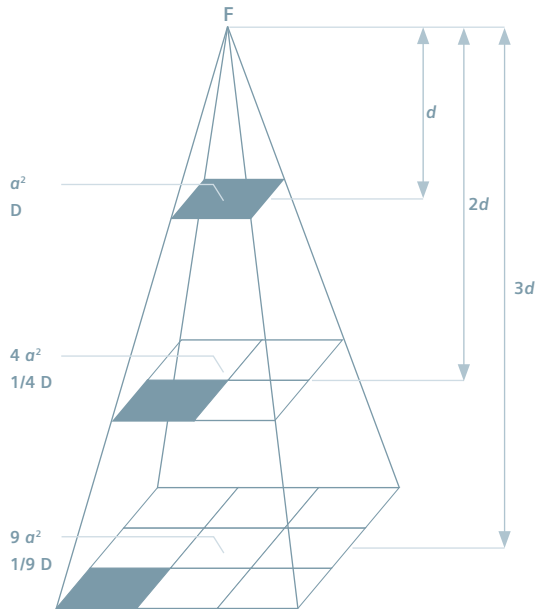**Fig. 50**
Dose area product and inverse square law for radiation dose.

Dose Area Product (DAP)

The DAP of a certain exposed area of constant dose is defined as dose times area and is independent from the distance to the source. An example for distances $d_1 = d$ and $d_2 = 2d$ and for associated doses $D_1 = D$ and $D_2 = 1/4\ D$ and the irradiated areas $a_1 = a$ and $a_2 = a\ d_2^2/d_1^2 = 4a$ proofs:

$$DAP_1 = D_1 \cdot a_1 = D \cdot a$$
$$DAP_2 = D_2 \cdot a_2 = 1/4\ D \cdot 4a = D\ a = DAP_1$$

This means that the Dose Area Product remains constant at different distances.

> **Inverse square law:**
> **The inverse square law for radiation dose shows that at twice the distance from the focus, the dose D is reduced by a factor of four with respect to the quadrupled surface – it is spread across a four-fold area.**

Angiography

Effective Dose in Angiography

Determining the effective dose in angiography depends on several factors, primarily on the variability in organ sensitivity to radiation. Recall that bone marrow is far more sensitive to radiation than the liver (refer back to the section on "Equivalent and Effective Dose"). The degree to which organs are affected by radiation also depends on the angle of the beams. Because dose distribution in angiography is not "homogeneous" as it is for CT, these factors must be considered when estimating the damage caused by irradiation.

Converting skin dose and DAP to effective dose is reliable only if the X-ray parameters and the location of the beam running through the body are known. In modern angiography, the role of the effective dose is not as significant as it is, for instance, in CT.

> **Remember:** The effective dose 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.

Important Parameters that Affect Dose in Angiography

Several parameters affect dose in angiography:

1. **Footswitch on-time: footswitch on-time controls how long the beam is on the body and thus how long the body is irradiated; less time means less radiation.**

2. **Frame rate: high frame rates are used to visualize fast motion without stroboscopic effects. However, the higher the frame rate, the more radiation. Therefore it is best to keep the frame rate as low as possible.**

3. **SID: according to the quadratic law and a constant requested dose at the detector, a greater distance between the source and the imager increases the skin dose. Rising SID from 105 cm (=SID 1) to 120 cm (=SID 2) increases skin dose (i.e. the dose at the IRP) by approximately 30%[*].**

Figure 51 illustrates the setup including the lower (SID = 105 cm) and the upper (SID = 120 cm) position of the detector.

IRP +

SID_2

SID_1

SID = Source-Image Distance
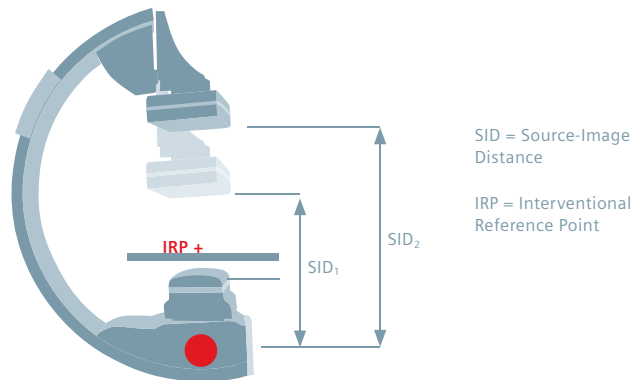
IRP = Interventional Reference Point

Fig. 51
C-arm, two different SIDs, constant table height, location of the IRP.

[*] If C-arm angles, table position, patient, and requested dose at the detector does not change.

Angiography

2. Dose Reduction Advances in Angiography

This section focuses on various technologies that Siemens has implemented or developed to reduce, monitor, and report the radiation dose applied during interventional procedures.

Siemens strives to implement all dose-saving, monitoring, and reduction methods available in the interventional market today. As a leader in the field of dose reduction, we consistently develop our own solutions. As such, we were the first to implement several features that save and monitor the radiation dose in the interventional routine. In addition, we are a leading vendor to offer these cutting-edge solutions for a large number of features.

Our products clearly follow the ALARA principle (As Low As Reasonably Achievable) to reduce radiation dose to the lowest possible level. This desire for as little radiation exposure as possible is at the heart of the CARE (Combined Applications to Reduce Exposure) research and development philosophy.

To maintain our leading position, and to improve health care for patients, we cooperate closely with experts at universities, and at public and private radiology centers all over the world – to convert research developments into practical components of everyday clinical routine.

In addition to implementing the newest technology, dose reduction efforts in angiography require training to become familiar with reduction methods and factors. We therefore attempt to make our dose-saving products as transparent as possible. We also provide a broad range of dose-monitoring products to interventionalists and technologists, and offer an ongoing selection of seminars and resources on dose reduction.

Figure 52 outlines our dose-saving, monitoring, and reporting products and tools for angiography, which are all available with the Artis zee family of products.
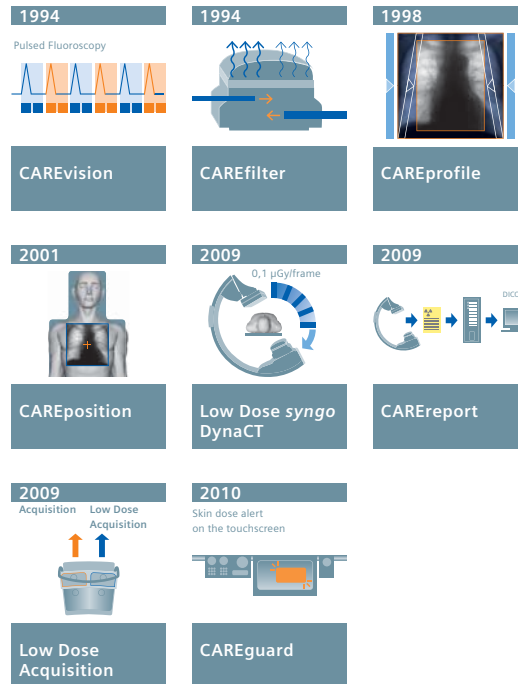
Fig. 52
Timeline of Siemens innovations to reduce, monitor and report dose in angiography.

Dose Saving – CARE

Reducing the dose during interventional procedures is not only critical for the patient, but also for the interventionalist and staff in the examination room. By integrating a broad range of dose-saving features into the Artis zee, Siemens minimizes the dose to both patients and the interventional staff.

CAREvision

CAREvision provides variable fluoro pulse rates. The pulsing frequency of the Artis zee system can be adapted according to the clinical need: from 30 pulses per second (p/s) in various steps, down to 0.5 p/s. This is the easiest way to reduce exposure to the patient. A reduction to half pulse rate reduces the dose by about half. The reduction from 30 p/s to 7.5 p/s results in a dose saving of 75% (Figure 53).
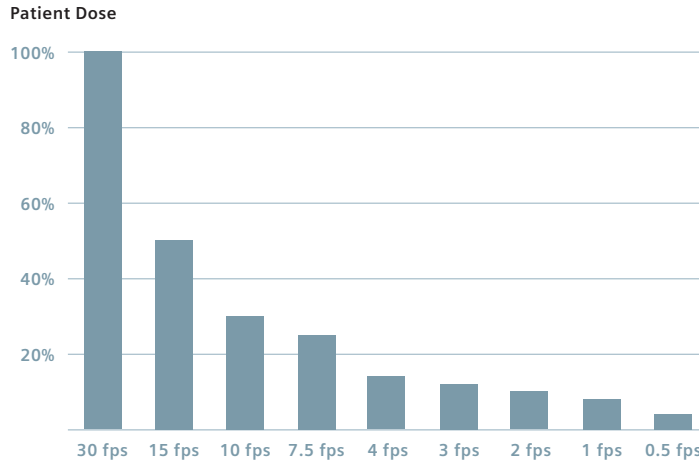
Patient Dose

Fig. 53
Reduction in patient dose by lowering the pulse rate during fluoroscopy.

CAREfilter

CAREfilter achieves skin dose reduction by allowing adjustment of the filter thickness. Additional copper filters reduce the skin dose through beam hardening. The variable filtration, 0.2 to 0.9 mm during fluoroscopy and 0.0 to 0.9 mm during digital acquisition, is adjusted automatically according to the absorption of the patient entrance dose along the path of the X-ray beam through the patient. This automatic filter insertion always maintains the lowest skin dose possible without degrading image quality. The filter selection is shown on the data display part of the monitor. Increasing prefiltering from 0.2 to 0.9 mm at 70 kV results in a dose saving of approximately 50% (Figure 54).
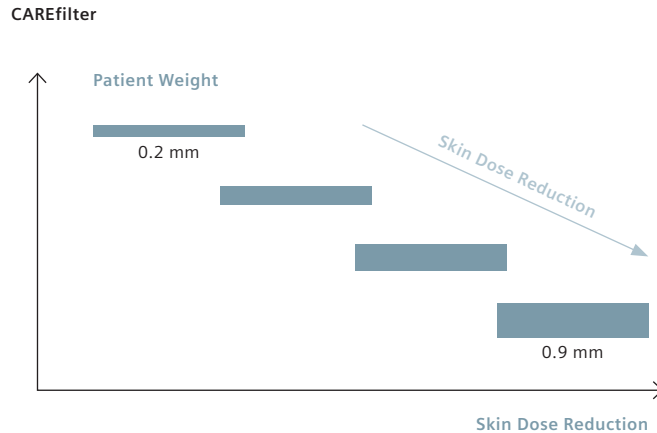
CAREfilter

Fig. 54
Skin dose reduction by automatic copper filtration depending on the absorption of the X-ray beam by the patient.

Angiography

CAREprofile

Using the last image hold (LIH) as a reference, CAREprofile allows radiation-free collimation and semitransparent filter parameter setting to precisely target the region of interest (Figure 55).
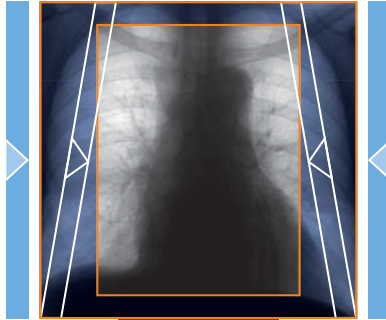

Fig. 55
The collimator position is indicated on the LIH by a white frame.

CAREposition

CAREposition provides radiation-free object positioning. Graphic display of the outline of the upcoming image allows panning the table without fluoroscopic radiation exposure (Figure 56).
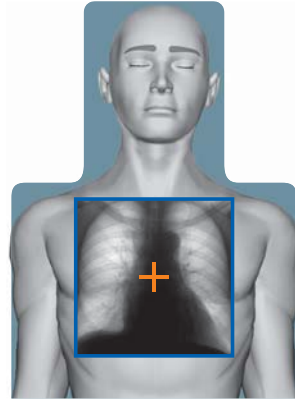

Fig. 56
When panning the table top, the graphical annotation of the field moves to the new position on the LIH.

For specific cardiac interventions, CAREprofile and CAREposition can reduce the overall fluoroscopy time by 0.5 to 3 minutes. Under typical fluoro conditions, this may result in a dose reduction of 20 to 120 mGy.

Angiography

Low-Dose Fluoroscopy

The operator can easily reduce the radiation exposure by changing the fluoroscopy protocol –
i.e., by switching table side ECC/TSC* from "Fluoro Medium" to "Fluoro Low" during fluoroscopy
using the touch screen; or by increasing the exposure to "Fluoro High" because of a thick object
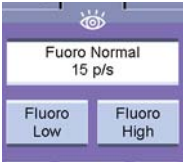or a steep angulation (Figure 57).

Fig. 57
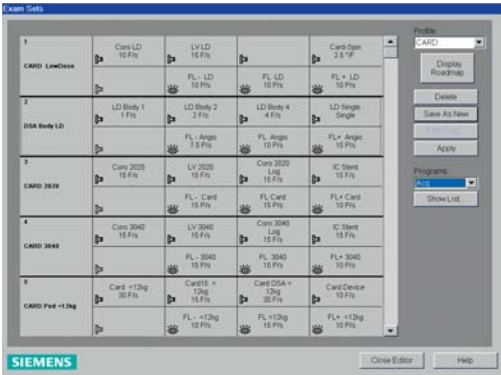Switching between various dose rates at the ECC/TSC.

Fig. 58
Examination Set "Neuro" with "Fluoro + 10P/s" and "DSA 3 F/s".

* ECC = Examination Control Console, TSC = Touch Screen Control

Switching between the three different fluoroscopy modes can be done table side in the Examination tab card (see Figure 57) or in the control room in the Examination Set (see Figure 58). "Fluoro Low" means half dose compared to "Fluoro Medium." Thus, switching from "Fluoro" to "Fluoro Low" results in a dose saving of 50%.

Low-Dose Acquisition

For especially dose sensitive patients, it is possible to generate a special low-dose acquisition protocol. An acquisition pedal of the foot switch can be configured as a low-dose acquisition alternative to the ECC/TSC (Figure 60).

A dose saving of 67% can be achieved by using an acquisition dose of 80 nGy/f instead of 240 nGy/f for interventional cardiology and an acquisition dose of 0.8 µGy/f instead of 2.4 µGy/f for interventional radiology.

Fig. 59
Low Dose Acquisition image of a peripheral ballon dilatation using 67% less dose in comparison to a standard acquisition protocol.
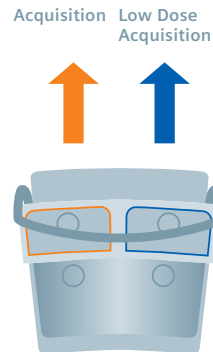

Fig. 60
Pedals for generating standard acquisition and low-dose acquisition.

Angiography

Low-Dose *syngo* DynaCT

The low-dose *syngo* DynaCT protocol achieves acceptable image quality at lowest possible dose values (Figure 61). This protocol is for radiosensitive patients, such as pediatric patients, and provides adequate diagnostic image quality. In clinical practice, the balance between image quality and dose has to be considered. For the prerequisites mentioned above, a five second high contrast DR rotational 3D run applying 0.36 µGy/f can be reduced to 0.1 µGy/f. Switching from 0.36 µGy/f to 0.1 µGy/f results in a dose saving of 72%. Low-dose *syngo* DynaCT in neuro can be achieved with an effective dose of 0.3 mSv.

In combination with *syngo* InSpace3D/3D Fusion, low-dose *syngo* DynaCT results can be fused with diagnostic pre-interventional CT, MR or PET•CT results. These fused data sets provide an excellent basis for planning and guidance during interventional procedures.
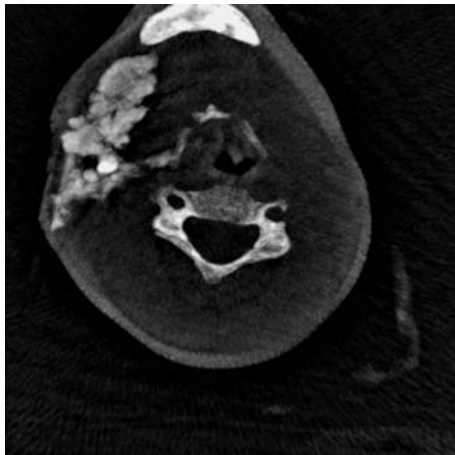
Fig. 61
Check-up AVM treatment using low-dose *syngo* DynaCT.

Slab Reconstruction for *syngo* DynaCT

Slab mode allows you to collimate the image from top to bottom before doing the 3D rotational run (Figure 62). The benefits are lower dose, because of the small exposed area, with better image quality, because of less scattered radiation.

Without collimation

With collimation

Minimum 3.5 cm
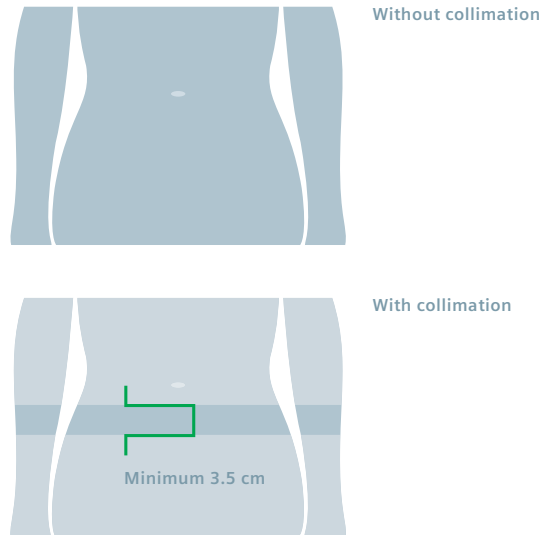
Fig. 62
Slab reconstruction for *syngo* DynaCT.

Angiography

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Fluoro Loop – Storage of Fluoroscopic Scenes

The Artis zee can store the last 513 images from fluoroscopy to hard disk (Figure 63).
This feature can be used for documentation, and can make additional digital acquisitions
unnecessary. The operator only has to press one button at the ECC/TSC console.

Dynamic fluoro scene

HDD Storage
CD-R/Net
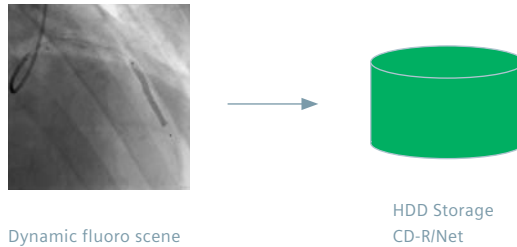
Fig. 63
Fluoro scenes can be stored to disk and CD/DVD, and also sent to PACS.

For specific cardiac protocols, using fluoroscopic recording instead of digital acquisition results
in a dose saving by a factor of 8 to 10.

Removable Grid

A simple way to reduce the dose in pediatric examinations, especially for babies or very thin patients, when scatter radiation can be expected to be negligible, is to remove the scatter grid in the flat detector housing (Figure 64).

Fig. 64
The anti-scattering grid can be removed by simply pressing a button on the flat detector housing.

The grid factor (i.e., the absorption of primary radiation due to the anti-scatter grid compared to free air) is 1.35, which translates into a dose saving of 26% when removing the grid.

Angiography

Dose Monitoring

Monitoring the patient dose is another element to controlling radiation exposure. To keep the burden of this task off the interventionalist, the Artis zee is equipped to monitor patient dose in various ways. This allows for more transparency during and after the procedure as to how much radiation was applied. The following sections discuss how to monitor dose with the Artis zee.

Displayed Dose Values

During the patient examination, the dose values are displayed at the image monitors in the examination room and also in the control room (Figure 65).

1. **When radiation is off, the dose area product and the accumulated dose at the IRP (Figure 65) are displayed.**

2. **When radiation is on, the dose area product and dose rate at the IRP are displayed.**

mGy	A: 123
µGym^2	A: 567.8

Displayed when the footswitch is not pressed:
Dose = 123 mGy
DAP = 567.8 µGy m^2

mGy/min	A: 321
µGym^2	A: 567.8

Displayed when footswitch is pressed:
Dose rate = 321 mGy/min
DAP = 567.8 µGy m^2
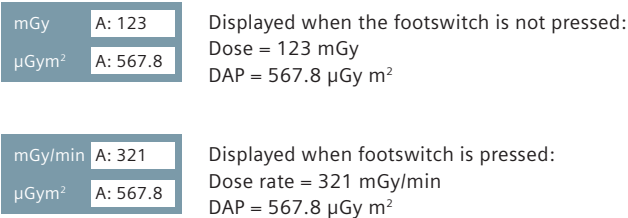
Fig. 65
Displayed dose values at the Artis zee.

CAREguard

CAREguard provides an effective way to control skin dose. Three skin dose threshold values (low, medium and high) can be defined. If the accumulated skin dose exceeds a defined threshold:

1. **An audible warning sound is given**

2. **A skin dose indicator on the live display flashes**

3. **A warning popup is displayed at the ECC/touchscreen (Figure 66)**
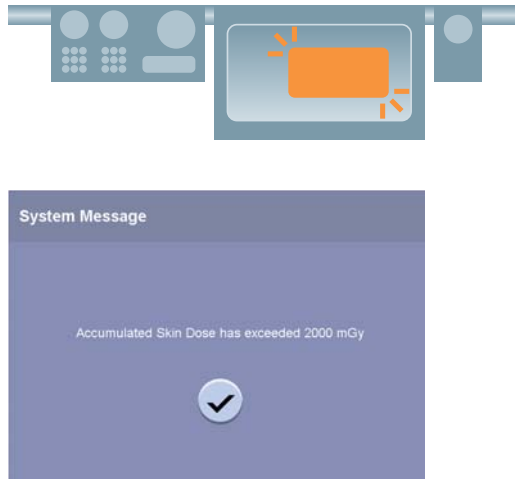
Fig. 66
Warning popup at ECC/TSC when a set dose threshold is exceeded.

Dose Reporting

More and more countries and authorities require the reporting of patient exposure following an intervention. To meet current and future regulations, Artis zee allows effective reporting of dose exposure and thus enables enhanced in-house dose reporting and analysis.

Patient Examination Protocol

After the patient has been examined, an examination or patient protocol is stored together with acquired images. The complete information for each run is stored. At the end of the protocol, the dose information is listed: number of exposures, total fluoro time, total dose area product, and total dose at IRP. These values are separated per plane in the lines below (Figure 67).

The examination protocol can be sent to a PACS system and printed as an image. It can also be stored and sent as a DICOM structured report for further evaluation (see next section for further details).
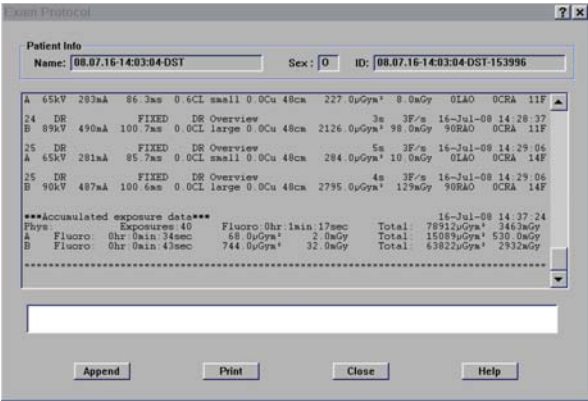
Fig. 67
Examination protocol for a biplane system (for example only, not a real patient examination protocol).

CAREreport – DICOM Structured Report

CAREreport, the DICOM structured dose report contains all patient demographics, procedure and dose information. Using commercially available programs or in-house software, this information can be filtered for further processing, such as dose analysis (Figure 68).

CAREreport provides consistent dose reporting and prepares for future potential legal requirements.

DICOM

Fig. 68
Use of the structured report for hospital statistics.

3. Radiation Protection for the Medical Staff in Angiography

Distance to the X-ray Source

In addition to protecting patients from excessive radiation exposure, physicians, technicians and other medical staff should be protected from unnecessary (i.e., scattered) radiation as well. Scattered radiation does not come directly from the X-ray tube, but rather is scattered by the patient, table or other devices within the path of the X-ray beam (Figure 69). Usually most of the scattered radiation is generated where the X-ray beam hits the patient.
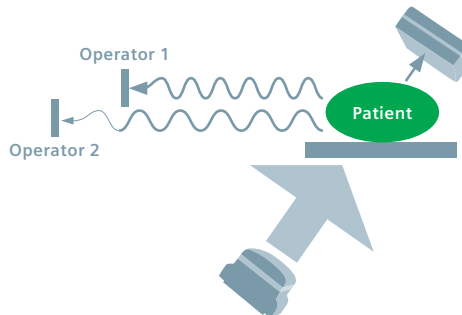
Fig. 69
Scattered radiation decreases with distance from the source (i.e. the beam entrance location of the patient).

Scattered radiation is roughly proportional to the dose area product (DAP) and decreases with distance squared to the location the scatter is generated. That is, twice the distance results in a quarter of the scattered radiation.

Approaches to shielding from scattered radiation include:

1. **Lead apron**
2. **Lead glasses**
3. **Mobile lead walls**
4. **Upper and lower body radiation protection**

C-arm Installation at Interventional Radiology Biplane System

Scattered radiation can be reduced by installing the lateral C-arm with the tube on the left side of the table when the medical staff works on the right side (Figure 70).

Fig. 70
Artis zee biplane system for interventional radiology with the X-ray tube on the left side of the table.

As already mentioned in the last paragraph, scattered radiation is mainly generated at the beam entrance location of the patient, which is left side in this configuration. At the operator's working position (right side), radiation exposure from scatter is much lower.

This setup is possible for biplane systems used for interventional radiologic and neuroradiologic applications only; it is not applicable for cardiac use because in a cardiac setup the detector needs to be close to the heart.

Angiography

Collimation

Scattered radiation is roughly proportional to the dose area product (DAP). If the area of the irradiated field is reduced by half, scattered radiation is likewise reduced by 50% (Figure 71).
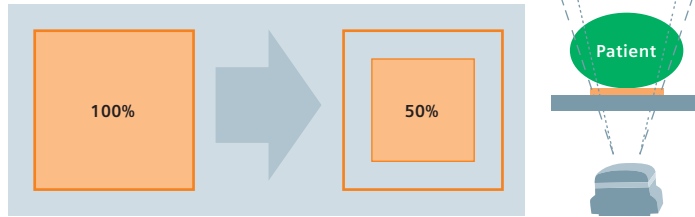
Fig. 71
Collimation should be used if possible.

Upper and Lower Body Radiation Protection

Radiation protection reduces medical staff exposure to scattered radiation by 99% – a highly effective method (Figure 72).

Fig. 72
Lower and upper radiation protection.

Further Reading

A comprehensive overview on dose issues and image quality can be found in the following article:

Stephen Balter et al. ACCF/AHA/HRS/SCAI Clinical Competence Statement on Physician Safety and Image Quality in Fluoroscopically Guided Invasive Cardiovascular Procedures, Journal of the American College of Cardiology, Vol. 44, No. 11, 2004

Angiography

II.E. Special Clinical Fields

There are many other clinical fields in which technology that uses ionizing radiation is used, such as surgery, mammography, and urography. In this section, you will find a short overview of challenges and solutions with regards to dose reduction in these fields.

1. Dose Reduction Advances in Mammography

Breast cancer is one of the most common cancers in women in the Northern Hemisphere. More than 10% of all women can expect to have some manifestation of the disease during their lifetime. Although earlier detection and better treatment may have reduced mortality in recent years, 30% of women with breast cancer will die from the disease.

X-ray mammography is still the gold standard of investigational procedures. Digital mammography has improved diagnostics, especially in younger women and in women with denser breasts. In most countries, screening programs have been established in order to support prevention efforts and early breast cancer detection. The right balance between low dose and high image quality for diagnostic confidence is of utmost importance.

Full-Field Digital Mammography

Several randomized controlled clinical trials have shown that mammography screening can reduce breast cancer mortality. Numerous clinical studies have shown that full-field digital mammography (FFDM) may provide some diagnostic advantages over screen/film mammography for certain population subgroups, such as young women with dense breasts and pre- and perimenopausal women.

It is well known that digital radiation detectors that are part of an FFDM system may have different absorption characteristics than a screen-film receptor in an analog mammography system. Therefore, the question arises whether the X-ray spectrum and the beam quality applied in screen-film mammography are also suitable for FFDM systems. In digital mammography, the radiographic technique can be optimized independently of the exposure to the image receptor, thus exploiting the wide dynamic range and the linear characteristic curve of digital detectors. Image quality and patient dose can be optimized at the same time. There is still some debate regarding the best anode/filter material to be used in digital mammography.

Anode/Filter Combinations in Digital Mammography

Some authors prefer molybdenum/molybdenum (Mo/Mo) for the examination of smaller breasts and rhodium (Rh) as the filter material for thicker breasts. Other users switch from molybdenum to rhodium or tungsten (W) as the anode material for thicker breasts, or recommend the general use of rhodium or tungsten as the anode material, in combination with rhodium as the filter material for all breast thicknesses and tissue composition.

A specific dose-related study was conducted to compare the average glandular dose (AGD) of Mo/Mo, Mo/Rh and W/Rh in full-field digital mammography based on an amorphous selenium detector with clinical data.[23] The main purpose of the study was to determine whether W/Rh can be used to reduce the glandular dose without loss of image quality in all breast thicknesses of the examined population.

[23] Uhlenbrock DF et al. Comparison of anode/filter combinations in digital mammography with respect to the average glandular dose. Rofo. 2009 Mar;181(3):249-54.

The average glandular dose corresponding to the images acquired with Mo/Mo was 2.29 ± 1.15 mGy with a mean compressed breast thickness of 46 ± 10 mm. For the Mo/Rh cases with a mean compressed thickness of 64 ± 9 mm, an AGD of 2.76 ± 1.31 mGy was obtained. The W/Rh cases with a mean compressed thickness of 52 ± 13 mm resulted in an AGD of 1.26 ± 0.44 mGy.
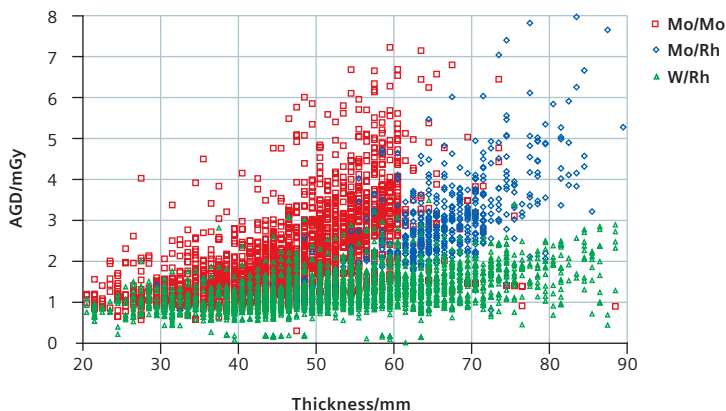
Fig 73
Scatter plot of the average glandular dose (AGD) values of all 4867 exposures as a function of the thickness.

The image quality of all mammograms in the study was evaluated visually by the readers in order to be adequate for the diagnostic task. In a few example cases that were imaged with both X-ray spectra, no degradation of the evaluated lesion's signal-to-noise ratio was found. One such image example is shown in Figure 74. This MLO image of a 67-year-old patient with a compressed breast thickness of 72 mm had been acquired with an average glandular dose of 3.8 mGy for Mo/Mo and 2.4 mGy for W/Rh.
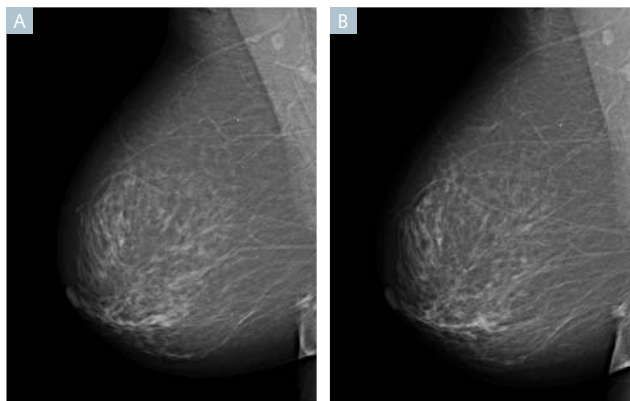
Fig. 74
Mammograms (MLO view) of the same breast acquired with
A: W/Rh and **B:** Mo/Mo.

Analysis of 4867 images revealed that the anode/filter combination W/Rh has advantages over Mo/Mo or Mo/Rh in terms of glandular dose, especially in patients with large breasts. Whereas the dose reduction with W/Rh is relatively small for small breasts, the effect becomes larger for increasing breast thickness, particularly above a compression thickness of 50 mm. For the largest compression thickness, W/Rh reduces the patient dose by more than a factor of 2 compared to Mo/Rh.

2. Dose Reduction Advances in Surgery

In today's clinical environment, surgery would be unimaginable without X-ray. Surgical C-arms help to give insight into the human body before, during and after any surgical intervention. For example, a surgeon is able to check screw placement in a complicated fracture using 2D and even 3D images during the intervention.

This helps to increase safety and accuracy of the intervention and also prevents additional post-operative X-ray follow-up; thus avoiding unnecessary radiation for patients and clinical staff in the OR.

Nonetheless, every radiographic exam, however necessary it might be, means exposure to X-ray for the patient as well as for the staff. This is why Siemens strives to combine the best possible image quality with the lowest possible dose for all C-arm systems.

Direct and Indirect Dose Savings

Dose-saving techniques can be embedded within the system itself and also enabled through easier workflows or new technologies that help to reduce dose emission throughout the entire procedure.

With ARCADIS Orbic 3D, Siemens introduced the first truly isocentric C-arm with the capability to provide intraoperative 3D images. ARCADIS Orbic 3D incorporates all available dose-saving features. Its true isocentricity provides the advantage of always maintaining the central beam in the isocenter, nearly eliminating the need for repositioning. Beyond the time savings, this imparts considerable savings in dose.

In addition, every ARCADIS system comes equipped with EASY (Enhanced Acquisition System), a bundle of automatic image processing features that allow ARCADIS systems to automatically analyze the images during acquisition and optimize brightness, contrast and dose.

Furthermore, ARCADIS Orbic 3D features NaviLink 3D, an integrated navigation interface with automatic image transfer. The open interface allows computer-assisted surgery systems to be connected to our C-arms, permitting radiation-free navigated interventions. Thus, the surgeon is able to reduce indirectly the radiation measured for the entire intervention.

Dedicated CARE features have also been allocated to our other surgical C-arms.
These include for example:

1. Freely positionable collimators for pinpoint X-ray images
2. Laser light localizers for positioning without dose application
3. 3D features to minimize postoperative imaging
4. Interface for remote surgical navigation systems
5. Pulsed fluoroscopy for minimal dose application
6. Detachable grid for further dose reduction
7. Additional copper filters
8. Special pediatric kits

In addition to these individual features, Siemens provides comprehensive dose monitoring and reporting for every intervention. Elements such as visualizing the air kerma rate on the display and providing a defined threshold with warning messages when a certain limit of dose application is reached, assists physicians and staff in limiting X-ray exposure.

3. Dose Reduction Advances in Urology

Key parameters that influence radiation dose and image quality include:

1.   **Dose setting**
2.   **kV setting**
3.   **Prefiltration of the X-ray beam**
4.   **X-ray beam collimation**
5.   **Radiation grid**

Organ programs were developed to simplify clinical workflow and handling of the X-ray machine. They allow application-dependant optimal settings to be achieved, thereby reducing the patient dose while maintaining the necessary image quality. With organ programs, dose reduction can be realized even in cases where the physician and/or staff do not have the comprehensive knowledge of a radiologist.

UROSKOP Omnia

The new flat-detector-based urology system UROSKOP Omnia enables patient dose reduction. In the past, the urologist had to acquire two images in order to cover the entire urinary tract (kidneys, ureters, bladder [KUB]). Now, only one shot is necessary to get a KUB image thanks to the dynamic flat detector with a size of 43 cm x 43 cm (17 inch x 17 inch) and the large field of view.

The patient entrance dose can be reduced by up to 38% in RAD mode by flat panel technology versus the conventional cassette exposure technique.

With DR mode, even greater dose reduction can be realized. This is possible because of the optimal setting of X-ray parameters and the excellent image processing of UROSKOP Omnia.

Patient Skin Dose Measurement at UROSKOP Omnia

- Simulation of patient thickness with PMMA
- Collimator format 30 x 30 cm
- FFD 115 cm; FTD 108 cm
- Dose measurement with Unfors XI (patient skin dose)

Program	Det. Dose (µGy)	Prefilteration (mm Cu)	Grid
Standard cassette exposure	2.5	2.5 mmAl	N15 r80 f115
(400 system) [24]	2.5	+0.1 Cu	N15 r80 f115
Organ program UROSKOP Omnia			
Standard-RAD	2.5	0.1	N15 r80 f115
MCU-RAD	1.25	0.3	N15 r80 f115
Standard-DR[*]	1.39 x 0.37	0.1	N15 r80 f115
MCU-DR	1.39 x 0.37	0.3	N15 r80 f115
Pediatry-DR	0.87 x 0.37	0.2	N15 r80 f115
Pediatry-DR (without grid)	0.87 x 0.37	0.2	no grid [**]

[24] Standard cassette exposure parameters: Kidney/Gallbladder a.p.; 19 cm 66 kV 50 mAs
(see: Hoxter E. Röntgenaufnahmetechnik. Publicis Corporate Publishing, Erlangen, 1991.)
Simulation of cassette exposure with RAD-Mode.

[*] DR-Mode: exposure parameters will automatically be calculated from Fluoro parameters (1 point
technique). To get the real det. dose, the indicated dose has to be multiplied with zoom factor 0.37.

[**] At 20 cm PMMA image quality is not good enough for certain applications; we would recommend
to use a grid for such objekt thickness.

5 cm PMMA		10 cm PMMA		20 cm PMMA				"Relative
µGy	kV	µGy	kV	µGy	kV	mAs	µGym²	Dose in %"
28.78	66.00	128.80	66.00	1817.00	66.00	35.10	97.47	100.00
22.32	66.00	89.31	66.00	1124.00	66.00	48.00	67.20	61.86
18.19	77.00	64.02	77.00	682.90	77.00	19.50	39.68	37.58
6.10	81.00	21.48	81.00	206.30	81.00	10.90	11.91	11.35
5.68	68.00	25.01	71.00	147.10	79.00	4.10	8.58	8.10
3.68	73.00	11.20	75.00	91.31	81.00	4.90	5.33	5.03
3.56	70.00	11.06	73.00	81.33	79.00	3.40	4.76	4.48
1.56	70.00	2.75	73.00	11.66	79.00	0.60	0.76	0.64

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, please see **www.siemens.com**.

III. Glossary

A

Absorbed dose	the amount of energy deposited in matter after being exposed to a certain amount of radiation
Activity	amount of a radioactive isotope measured by number of decays per time
Acute radiation syndrome (ARS)	death of a 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, analyzes whether the heartbeats 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 (α) particles	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; apoptosis 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

Beam hardening	increasing the energy level of the beam by filtering out low-energy photons
Bone marrow	soft highly vascular modified connective tissue that occupies the cavities of most bones and occurs in two forms: a yellowish bone marrow consisting chiefly of fat cells and predominating in the cavities of the long bones (also called yellow marrow) and a reddish bone marrow containing little fat, being the chief site of red blood cell and blood granulocyte formation, and occurring in the normal adult only in cancellous tissue, especially in certain flat bones (also called red marrow)

C

CARE	Combined Applications to Reduce Exposure
CARE Dose4D	a system that not only modulates radiation intensity according to the size of the patient, but also to the anatomic region that is being irradiated as the CT scan progresses
Cathode	the electron-emitting electrode of the X-ray tube
Collimator	a thick lead sheet with thousands of small holes that allows passage of only those gamma rays that travel at right angles to the plane of the crystal
Computed Tomography Dose Index (CTDI) – $CTDI_w$	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	a device for detecting the presence of electromagnetic waves or of radioactivity
Deterministic damage	damage of organic tissue that is certain to occur as a result of exposure to a high amount of ionizing radiation
Dose Length Product (DLP)	the product of $CTDI_{vol}$ and the length of the examination range

E

Ectopic beat	a heartbeat that is spurious and is not synchronized with normal heartbeats
Effective dose E	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)
Electron	an elementary particle that consists of a negative charge and spins around the atomic nucleus
Equivalent dose	for any type of radiation defined as the absorbed dose multiplied by a factor (wf) 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	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 heartbeat
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 wavelength of less than 1 pm

H

Half-life	time span until only half of the initial isotope is left due to radioactive decay (physical half-life), excretion (biological half-life), or a combination of decay and excretion (effective half-life)

I

Ionization	the process by which atoms are converted into ions (electrically charged atoms)
Ionizing radiation	radiation that can ionize matter
IRP	Interventional Reference Point
Irradiation	radiation emitted with a specific purpose, in this case, for medical reasons

K

KERMA	Kinetic Energy Released in Matter; the energy released per unit mass of a small volume when it is irradiated by ionizing radiation. Measured in Gray (Gy)
Kinetic energy	energy stored in a moving object

M

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

N

Neutron	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 the context of imaging, the grainy structure of an image

P

PET	Positron Emission Tomography
Photon	a quantum of electromagnetic radiation
Pitch	table feed during one revolution of the X-ray tube divided by the nominal scan width in mm
Plexiglas phantoms	dummies made of plexiglass used to measure the radiation doses on different parts of the body
Positron	a positively charged particle with the same mass as an electron

R

Radiation	the process of emitting radiant energy in the form of waves or particles
Radioactive substances	substances that emit radiation of different types
Radioactivity	the property possessed by some elements (e.g., uranium) or isotopes (e.g.,carbon-14) of spontaneously emitting energetic particles (such as electrons or alpha particles) by the disintegration of their atomic nuclei
Radionecrosis	destruction of organic tissue by radiation
Radon	a 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 (such 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
SPECT	Single Photon Emission Computed Tomography
Spiral CT	CT scan during which the table and the X-ray tube move continuously
Stochastic damage	damage that may potentially happen; involves chance or probability

T

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 reduced when the patient is irradiated from the front
X-rays	electromagnetic radiation that has an extremely short wavelength of less than 100 angstroms and has the property of penetrating all solids to varying degrees

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