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X-ray scatter correcting methods for digital radiographic imaging

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ABSTRACT

X-ray scattering correction method has been the primary means of enhancing radiographic images for quite some time. X-ray scattering is major deterioration factor that decreases image contrast and increases the image granularity in a radiographic image. However, this can be eliminated by using scatter reduction techniques like air gaps and anti-scatter grids, but the two techniques are cumbersome, and increases patient's radiation dosages. Moreover, it can also cause artifacts whenever anti-scatter grids are used. Recently, commercial software packages have been developed from various x-ray equipment manufacturers that have eliminated the need for anti-scatter grid usage. These recent advancements also allow lower patient dosages. Objective of this review is to summarize and review x-ray scattering and image processing algorithms used for enhancing the performance of the digital image in general radiography. Articles on digital image processing and commercial software for x-ray scatter correcting were thoroughly reviewed to complete this summary. These articles indicate that scatter correcting methods are based on principles of physics which involve of mathematical models of radiographic formation and x-ray scattering estimation methods. One simple model has the total energy absorbed at an image detector forming a primary x-ray plus a scattered x-ray whereby the point spread function of the scattered x-ray is used. Almost all estimations of x-ray scatter are computer simulations. The digital image post-processing algorithms are important factors in the x-ray scattering correction process. Their algorithms are related to mathematical models and the amount of scattering x-rays in an image, and are selected for use based on these considerations. These algorithms include subtraction, de-convolution, and anti-scatter grid simulation techniques. Therefore, x-ray scatter correcting methods for a digital radiographic imaging may be used in general radiography since their image quality is comparable to the images that have used anti-scatter grids, but are also beneficial since radiation dosage can be reduced using this process.

Introduction

X-ray scatter radiation is a major deterioration factor that decreases image contrast and increases granularity in the radiographic imaging. The scatter removing processes like air gaps and anti-scatter grids have been used in the past to improve the image quality, but both of them are cumbersome and increases the patient's radiation doses. They can also cause radiographic image artifacts. Fortunately, commercial software has been developed from the various x-ray imaging system manufacturers that aim to enhance the correcting processes involved in x-ray scattering without the use of anti-scatter grids in general radiographic imaging examinations. Examples include such products as Virtual Grid, SkyFlow, Intelligent Grid, and Scatter Correction for CXDI Series, SimGrid, and SBSC. They come equipped with digital imaging systems like optional software, and have been approved by the FDA. The image qualities are comparable to the images using anti-scatter grid, but radiation dosages

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can be reduced using these products. These techniques are particularly good for pediatric radiography, bedside radiography of patient who cannot be moved, and dual-energy mammographic imaging.Without a grid, the technologists can work more efficiently. These techniques eliminate unwanted image artifacts due to grid misalignments that may affect images enough to require retakes. Consequently, these products make for better patient care. [1-5,15]

When a radiographic image is forming, an x-ray passes through the patient or object being examined, and there are three possible directions that each x-ray photon can take. First, it can penetrate the patient without interacting by what is usually called the primary x-ray (the remnant beam). Second, it can interact with the patient and be completely absorbed by depositing its energy. This is usually called the patient absorption dose. Third, it can interact and be scattered from its original direction with or without depositing a part of its energy by what is usually called a scattered x-ray.[6] In diagnostic radiology, x-ray photons have energy ranges between 17 to 150 keV. This range would also apply to mammography. Three dominant X-ray photon interactions include the photoelectric effect, coherent scatter, and incoherent scatter. To understand radiographic image formation and how to determine the appropriate radiation dose given to patient for enhancing radiographic image quality, we need to understand the interactions of x-ray photons with matter as this relates to photon carry energy, tissue type, and composition of subject to be investigated, especially in the energy ranges used in diagnostic radiology.

Radiographic image formation is one of the radiation transport problems which depends on photon interaction. Generally, photon interaction can be predicted by their probabilities. They depend on energy of photon and the material component, which are themselves random situations. The photon interaction can't easily be solved directly by calculation, but can be estimated by computer application using The Monte Carlo simulation method. Monte Carlo simulation determines the number of x-rays that have penetrated the patient and those that have been absorbed by image detector, as well as the energy deposited in the subject lead to patient dose. The details of scatter x-ray can be demonstrated clearly by Monte Carlo simulations that take into account conditions, various parameters and geometry of user's interesting in their radiation transport problems.

Estimations of the primary x-ray and the amount of scatter x-ray photons are very important parameters for calculating the scatter degradation results, establishing x-ray scatter correction methods, and evaluating which correction method that is the most appropriate. For example, the experimental methods for the Blocker-based techniques and the beam stop techniques use high radiation attenuation materials and a bulk of tissue that is equivalent material for simulating x-ray photons that pass through subjects. The x-ray photons occurring as primary and scatter x-rays are measured by image detector.[8] The Scatter to Primary Ratio (SPR) and Scatter fraction (SF) are the relationships of primary, scatter, and the total x-ray photons. These ratios can be calculated from the data found by this experimentation. SPR and SF quantitatively show the degree of scatter generating related to the exposure technique (kV, mAs, filtration, and field size) and the subject thickness. For assessment the quality of clinical images by quantify the scatter degradation in an imaging system. The differential signal to noise ratio (SNR) or contrast to noise ratio (CNR) are used. The other quantitative assessments use physical indicators like contrast improvement, granularity improvement and sharpness, and IQFinv. All of these assessments can be applied to clinical images of various kind of radiographic imaging systems. [1]

Primary x-ray makes for useful information to visually separate the different tissue types in an image, but the scatter makes for an unusable exposure because it spreads across the image receptor and adds to the amount of exposure to every tissue in the area within the image. This effect reduces subject contrast and increases the granularity over the entry imaging area.[7] The combination of primary and scatter x-ray forming in a radiography image described here is a general model. Many authors explain this model by equations, functions, or mathematical models currently used in engineering applications. These models are very beneficial as they introduce important variables and their relationships which are of particular concerns for use in x-ray scatter estimation and correction methods.

For a long time, the image processing algorithm for x-ray scattering correction remained the main method for enhancing a radiographic image. This process calculates the primary x-ray and the scattering radiation from an image captured by an imaging detector. Separating and correcting the scattering part with image processing algorithms leaves only the primary part forming the image. The physics of radiographic image creation and the development of digital image processing methods plays an important role in the x-ray scattering correction process without the use of air gaps or grids. Such a principle demonstrates the potential for use in general radiographic imaging as more commercial software is being developed. These advances reduce radiation dose, improve workflow and efficiency, and thereby make for better patient care. The objective of this review is to briefly describe the scattering x-ray calculation methods, image processing techniques, and the algorithms used for enhancing the quality of the digital image in general radiography.

1. X-ray photon interaction

Within the x-ray diagnostic radiographic energy range of 17-150 keV, three x-ray photon interactions with matter are observed: The photoelectric effect, coherent, and incoherent scattering. These interactions have an influence on the amount of the primary and scatter x-rays which are produced in an x-ray image.

1.1 Photoelectric effect

Photoelectric effect is the dominant process occurring at lower energy portions of the diagnostic x-ray range. It arises when an x-ray photon collides with an electron bound within an atom shell of medium energy level and transfers all its kinetic energy to the electron. If the photon energy is smaller than the binding energy of the electron in that shell, the x-ray photon can then interact with an electron in the outer shells where binding energy is lower. In the case of x-ray photons, kinetic energy is greater than the binding energy of that shell. Consequently, the x-ray photon can reject the electron making for ionization. The free electron is called a photoelectron and its energy will be the difference between the x-ray photon energy and the binding energy of the atom shell. The free electron can travel in the medium and transfer energy along the path length before becoming absorbed. This process contributes to be the patient's absorption of the radiation dosage. The vacancy left in the atom is filled by electrons in the outer shells producing the characteristic x-ray. This type of x-ray is called a fluorescent x-ray and it can interact with electrons in the outer shells producing new photoelectrons called Auger electrons. Soft tissue of human body consists of the low atomic number materials, so the characteristic x-ray energy suitable for humans is also low, as well. It cannot travel as far and will be re-absorbed into the body.

Probability of a photoelectric event occurrence is called the cross section of photoelectric interaction. To predict the probability that an interaction event will occur, the relationship of the energy of photon and the atomic number of an element can be applied to calculate the probability of the event. Photoelectric effect event is proportional to the atomic number of an element, but is inversely proportional to the energy of the x-ray photon.[6]

1.2 Coherent scatter (Rayleigh scattering, Elastic scattering)

When x-ray photons have less energy than the binding energy of an electron in an atom shell, it collides with that electron. Two interactions may occur in this scattering interaction. First, the Thompson effect can occur whereby the x-ray photon energy is absorbed by an orbit electron. Excitation then takes place to where the photon reemits with original energy. Second, Rayleigh scattering can occur in ways similar to the Thompson effect, but instead photon energy is absorbed by the electron cloud around the atom. The direction of the reemission is at a random angle compared to the original direction of the incoming x-ray photon. To predict this deflection angle.[7] J.J. Thompson equation is used. It describes the differential cross section of x-ray photons scattered by a free electron, but it does not explain the coherent scattering angle because the tissue being investigated in medical imaging have electrons which are bound to an atom. Not being a free electron, it has been modified by multiplying with an atomic form factor term. This term includes the energy of the incident photon and the atomic number of the material in question. This modified equation can explain scattering angular distributions for different energies of x-ray photons and the element being investigated.[6]

1.3 Incoherent scatter (Compton effect, Inelastic scattering)

In incoherent scattering, the x-ray photon has a higher energy than the binding energy of electron within an atomic shell. When it strikes an electron, it imparts some of its energy to it. The electron is rejected from its shell while carrying some energy at a departing angle while the x-ray photon is scattered in a direction at an angle to the original direction. Typically, the electrons found at the outer shells of an atom are more likely to be affected by incoherent scattering. The energy of the x-ray photon occurring at an angle after interaction can be approximated using the Principle of Conservation of Energy and Momentum. The energy of electron can be calculated by subtracting the energy of scattering photon from its incident photon energy. The departing angle of electron can be calculated by using energy and the scattering angle of the x-ray photon. The cross section of the Compton Effect event can be calculated using the Klein-Nishina equation.[6]

1.4 Photon interaction and image formation

The distribution of primary x-ray photons over the entry area of the image detector contains useful information to visually separate the different tissue types in a radiographic image through what is called contrast. It is based on the differing attenuation of these tissues. All human tissues are composed of different kinds and proportions of elements, so x-ray attenuation in any type of tissue can be different even in healthy or bad conditions. X-ray attenuation occurs when removal of x-ray photons takes place from an incident beam by absorption or through scattering interaction. This occurs when photons travel along the tissue before being detected by an image detector.

Linear attenuation coefficient (μ), depends on x-ray energy. It is used to describe the amount of attenuated and un-attenuation photons. The linear attenuation coefficient is determined by multiplying the number of the matter's atoms per unit volume and the total cross section of all photon interaction. For x-ray energy ranges used in diagnostic imaging, the total cross section is the three photon interaction cross section. Therefore, the primary x-ray is accounted for using the un-attenuation x-ray and the un-scattered x-rays which have reached the image detector. The intensity of primary x-ray can be calculated by the Beer-Lambert Law using the energy beam, the linear attenuation coefficient of tissue component, their geometry and the distance required for a photon to pass through a medium of matter.[7] The choice of energy will be a compromise between the patient dosage and high contrast image levels. If the energy is very low, then very few photons will reach the image receptor and the radiation dose to the tissue will be very high. If the energy is too high, then there will be very little difference in transmission through different types of tissue and the contrast in the image will be poor.[9] When x-ray photons are absorbed by image detector, the photo electron and K-shell characteristic radiation are produced and are also absorbed into the image detector due to the photoelectric effect. Locally, electrons will absorb, but characteristic radiations do not. The mean free path of characteristic radiation which arise from an element by the image detector like Ag, Cs, or Gd are long enough to spread widely. This can lead to loss of efficiency and to image blurring.

The scatter x-ray acts as a background. The factors that increase the amount of scatter radiation are the higher level of photon energy, large field sizes of x-ray beam, and a large body part and/or thicknesses of soft tissue.

The large field size of an x-ray beam and large body parts produce significant amount of scatter radiation, but the higher kV does not produce significant scatter radiation in small body parts. Even in thicknesses found in parts of the body such as the abdomen, the visible fogging of images caused by scatter are comparable when radiographed with 80 kVp and 92 kVp. Scattering interactions do not result in absorption of the photon. The coefficient for the energy-absorptive component of the scatter is due to the transfer of energy to recoil electrons. In general radiography, the rejecting or reducing of x-ray scatter have been implemented using techniques such as grid or air gap, but they are difficult to transport and the patient's radiation dosage may increase up to double when using a grid.[7]

1.5 Monte Carlo simulation

The Monte Carlo method is an effective technique used for simulation of radiation transport in matter. This method employs random sampling from probability distribution of an event for construction of the solution to a physical or mathematical problem. The Monte Carlo simulation for x-ray photons used in the diagnostic energy ranges involves three interaction processes that may occur when a photon passes through any medium. When an incident x-ray photon strikes the surface of a medium, the free path length is determined from an exponential distribution and the interaction site is calculated using the free path length and the direction of the incident photon. Afterward, the type of interaction that will occur is determined from the interaction cross section data. If it is photoelectric effect, the total energy of the x-ray photon is absorbed. If a repeated simulation for the next photon predicts an incoherent scattering, the scattering angle and energy of the scattered photon are determined d using the Klien-Nishina equation for the Compton Effect. If it is likely that coherent scattering will occur, the scattering angle is determine using the Thompson equation.[8]

The Monte Carlo simulation uses a large number of x-ray photons in the calculation in order to make a good estimation. It uses the same sophisticated steps in calculation as that of geometric modeling for experimentation and error reduction. It can even speed up the calculation itself. The results of these simulations are energy absorption, scattering, and energy transformation. They are used to describe the primary and scatter behavior in forming the radiographic image formation. Several particle physics Monte Carlo simulation packages currently exists that are widely used in medical physics studies. The most common are EGSx/EGSnr, MCNP/MCNPX, Penelope, GEANT4, and FLUKA. Many researchers use the Monte Carlo method while doing studies on scatter x-ray correction methods. There are x-ray products being sold through venders that use the Monte Carlo method for calculating the amount of scatter x-ray in each take of image and use it for scatter correction processes designed for reader console software.[1-5,14]

2. Radiographic image formation, primary, and scatter x-ray models

2.1 The simple model

The radiographic image formation can be described by a simple mathematical model. X-ray energies absorbed at the image detector, I(x,y), are primary and secondary photons as shown by equation number 1. An x-ray source that emits the photons of energies in the diagnostic range allows the ray of the x-ray beam to safely pass through the subject. It falls perpendicular to the image detector and is locally absorbed at the x, y position of the plane of the image detector. The primary photon energies absorbed by the image detector form the image, but the secondary photons create a background signal. as shown by Figure 1.[9]

$$I(x,y) = primary+secondary \quad (1)$$

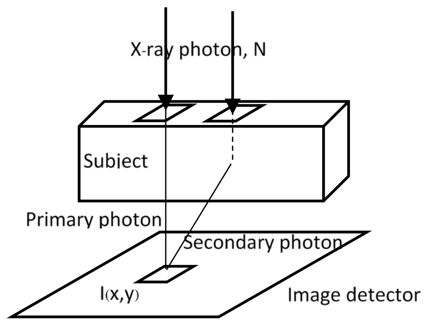

Figure 1 Simple model of the radiographic image formation shows the combination of primary and a secondary x-ray photons striking an image detector.

The x-ray energies absorbed at the image detector can be defined like mathematical integration as show by equation number 2. The first term is the line integral along the path of primary photons arriving and absorbing at point x, y of image detector plane. The second term is the scatter distribution function of energy absorption efficiency (ε), energy range (Es), and solid angle (Ω). The scatter function S has a complicated dependence on the position and the distribution of tissue within the patient. For many applications, it is sufficient to treat it as a slowly varying function, and involves replacing the very general integral with the value at the center of the image.[9]

$$I(x,y) = N\varepsilon(E,0)E\exp(-\int \mu(x,y,z)dz) + \int \varepsilon(E_s,\theta)E_s S(x,y,E_s,\Omega)d\Omega dE_s \quad (2)$$

The usefulness of this simple model is clearly demonstrated in all parameters that are concerned with studying scatter x-rays. This is true not only in a real experiment setup, but also in computer simulations because this equation combines the primary and secondary x-ray photon as the total energy absorbed by image detector. Ryohei ITO *et al*. used this assumption to do the scatter subtracted from the total radiation arriving at the image detector using an algorithm found in some scatter correcting methods.[3] Elena Marimón assumes that the image formed by a digital mammography detector is the linear combination of the primary and scatter x-ray photons and uses the convolution based method to estimate the scatter photons.[11]

2.2 Scatter to Primary Ratio (SPR, STPR) and Scatter fraction (SF)

SPR and SF are the ratio of the intensity of scatter radiation to that of primary radiation. It is useful in determining the degree to which image contrast can be degraded. Evaluation of the characteristics of scattered radiation in diagnostic radiology as a rule will be determined with suitable phantoms, but not with patients.[7] In many studies, they conduct the experiment with water or PMMA of human size phantoms. Sometimes these studies use a Monte Carlo simulation. Not only is it possible to evaluate the characteristics of scatter, but it is also possible to take measures to check the scatter correction technique. Scatter to Primary Ratio is the magnitude of the scattered to primary x-ray at image detector after having penetrated subject. Scatter fraction is the measured quantity of scatter found in the total image record as shown by the following equations, where S and P are scatter and primary x-ray, respectively.[11]

$$SPR=S/P \tag{3}$$

$$SF=S/(S+P) \tag{4}$$

SPR and SF can be estimated by the conventional technique set up for scatter in diagnostic radiography. The Blocker based techniques or the beam stop technique are shown in Figure 2.[11] These techniques consist of imaging subjects made of high radiation attenuation material such as lead (Pb). Lead stops absorption of almost all of the primary x-rays which are incident on the surface of a phantom. Underneath where the primary x-ray beam stops, a block of material is positioned which has the radiation characteristics equivalent to that of human tissue. It is used to demonstrate the scatter x-ray produced when the beam of x-ray passes through the material and is absorbed by image detector. The beam of the x-ray that cannot pass through Pb block only results in scatter x-rays. The primary and scatter x-ray are found elsewhere in the radiation field. As a result of S and S+P outcomes, they are then being combined to calculate the primary (P), SPR, and SF for this particular experimental setup. By varying the x-ray beam energy and scatter material thicknesses, the primary, SPR and SF of each experiment and the setup geometry can be observed.[6,7]

Both the SPR and SF increase as the subject thickness increases; thus, for thick body parts, such as the abdomen, scatter radiation is a serious problem. In chest radiography, the equivalent tissue thickness of the lung is approximately one half that of abdomen. However, tissue thicknesses in the chest may vary from 10 to 30 cm in the mediastinum. Cross scatter from one region of chest to another results in SPR for some areas of the chest where thicknesses may exceed those found in abdominal radiography.

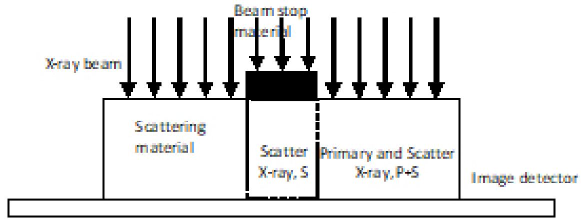

Figure 2 Diagram of the blocker based techniques or the beam stop technique for measuring SPR and SF.

2.3 Scatter point spread function (SPSF)

Point spread function (PDF) is the 2D impulse response function produced from a point source passing through any system. For radiographic imaging, the point spread function represents the spatial distribution of radiation when a narrow beam of x-ray passes through matter as shown in Figure 3(a). PDF is described in polar coordinates PSF(r,θ. PDF demonstrates how x-ray energies are recorded over imaging area by absorption of the energies of primary and scatter x-rays using image detectors. Therefore, the PDF of image detector PDFdetector (r,θ) can be described by the contribution of primary x-ray PDFp (r,θ) and scatter x-ray PDFs (r,θ).[10,11] The later can be called a scatter point spread function (SPSF). Generally, the PDFs(r,θ) is normalized to the total energy deposited by the primary x-ray and integration of the normalized PDF would correspond to the SPR value (Scatter to primary ratio) as seen in equations 5 and 6.

$$PDF_s'(r,\theta) = \frac{PDF_s(r,\theta)}{\int_{\theta=0}^{2\pi}\int_{r=0}^{\infty} PSF_p(r,\theta)d\theta dr} \tag{5}$$

$$SPR = \int_{\theta=0}^{2\pi}\int_{r=0}^{r} PDF_s'(r,\theta)d\theta dr \tag{6}$$

To estimate the pencil beam SPSF by the conventional technique, an assortment of small to large diameter size of x-ray beam stop subjects are used to measure the SPR for each diameter size of beam stop subject. The relationship between diameter size and SPRs are obtained by using curve fitting method. Therefore, SPR at a zero diameter represents a size of point source and can be extrapolated from this relationship with a unit of area-1. The ratio of scattered x-ray to primary x-ray can be found after penetrating the subject with the x-ray irradiation field that is set to the same size as the x-ray detector. As the STPR increases, the image deteriorates as a subject's habitus become thicker. Therefore, the amount of scattered x-ray does not contribute to the depiction as anatomical structures increase.[1]

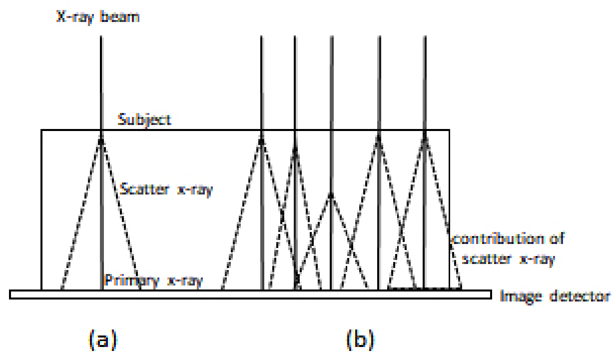

Figure 3 Scatter point spread function (a) single x-ray show SPSF and (b) cumulative SPSS from scatter x-ray distribution over on image.

PDF can be used to estimate scatter forming at image detector as shown by Detlef Mentrup *et al*.[2] They state that the amount of scatter x-ray depends upon thickness and composition of an object. The total scatter present in the image can be thought of as a superposition of scatter contribution generated by a thin pencil-x-ray beam passing through an object. These are called scatter kernels as shown in Figure 3(b). The superposition of all scatter kernels in the imaging area are used to estimate scatters over an entire image.

2.4 Convolution based scatter estimation

Convolution is a mathematical operation on two functions to produce a third function. In convolution based scatter estimation, the primary x-ray as one function are modified or filtered by the other function typically called a convolution kernel to estimate a result of scatter x-ray distribution at image creation location by the primary x-ray as shown by equation 7.

$$I_s = I_p * h_s \quad (7)$$

Is, Ip, and hs are scatter, primary x-ray, and the convolution function, respectively. There are some characteristics of convolution functions that have been revealed in many studies. Detlef Mentrup and Kotre CJ show that scattered radiation is a slowly varying background to image phenomenon. This behavior of scatter x-ray distribution on imaging areas appears similar to that of low spatial frequency. Convolution kernels can be used to explain this characteristic.[2,14] Love, L. A. and Kruger use a convolution-filtering method to estimate the scatter distribution in images by investigating many convolution kernels and applying them to images of anthropomorphic phantoms. They found that two-dimensional exponential kernels with a full width at half a maximum of 50-150 pixels best reproduced the scatter fields within these images.[12] Elena Marimón approximated the scatter by a two-dimensional low-pass convolution filter of primary image.[11] As mentioned before, the radiographic images created from image detectors originate from primary and scatter x-rays. The scatter can be represented as a convolution of primary image with the scattering kernel (hs).

2.5 Other mathematical models

K. Kim *et al*. defined scatter x-rays as an image degradation function like those used in other fields such as engineering.[17] The desired function could have characteristics like that of scatter deterioration. In this study, dark-channel prior is a more suitable, so this method can be used to restore deterioration to the original image. In other study, some types of noise sources have potential for studying scatter x-rays. Pierre Gravel *et al*. study the statistical properties of noise found in medical images. The types of uncorrelated noises are Gaussian, Poisson, and Rician noise are selected to determine the relationship between an image intensity and noise variation, and to evaluate the corresponding parameters for each type of noise. [19]

3. Brief review of scatter x-ray photon estimation techniques

Scatter x-ray has a complicated dependence on position and the distribution of tissue within the patient. For many applications, it is sufficient to treat it as a slowly varying function and to replace the very general integral with the value at the center of the image.[9] Similarly, some researchers use the low spatial frequency convolution function like a low pass convolution or exponential function to represent the scattered x-ray in an image.[2,14] The accuracy of the scatter x-ray estimation is very important in order to correct its effect on any image. In this section, there are examples provided of estimation techniques that use scatter correcting algorithms proposed by particular researchers. The estimation techniques can be divided into two groups: There are those techniques which are found in commercial software, and there are the various techniques used by research groups.

Takahiro KAWAMURA *et al*. estimated the scatter x-rays from the thickness of subject which is under radiographically examination. They use the pixel values from image detector area with a subject and without a subject to estimate the thickness. The subject thicknesses and x-ray photon energy are used to estimate amount of scatter x-ray from pre-calculated values of scatter to primary ratio. Each pixel of radiographic image will be used to estimate for the scatter x-ray pattern over entire imaging area.[1] Ryohei ITO *et al*. developed the algorithm that estimates the thickness of subject for each image pixel from a histogram of the original image and made a body thickness image. They use the body thickness image to estimate scatter x-rays from a table or data base of scatter x-ray ratio to object thicknesses. Their algorithm concerns not only a pixel of interest, but also provides insight regarding various surrounding structures.[3] Detlef Mentrup *et al*. used the model of scatter point spread function to estimate the scatter for the image under radiographically examination. The scatter kernels used by this model are pre-calculated using a data base in a Monte Carlo simulation which can be adapted to patients with different constitutions. The selection is based on the local pixel value and its spatial gradient. The super position of all scatter kernels over entry image area is the total scatter image.[2]

IBEX Innovations Ltd. has developed a scatter correcting method called gridless scatter reassignment. This algorithm measures material type and thickness of subjects in radiographically examinations. The Monte Carlo simulation aids in estimating the scatter x-ray maps from material type and thickness of subject and can be refined until accurate predictions of each subject are obtained.[12] Andreas Fieselmann *et al*. models the scatter radiation in mammography that can be divided into a low-frequency and a high-frequency component. A low-frequency component is a smoothly varying value which added to pixel value of an image. The high-frequency is noise. Fieselmann estimated only a low-frequency component having a scatter point spread function that is simulated by the Monte Carlo method based on subject properties, the geometry, and acquisition parameters.[14]

For commercial software package, Takahiro KAWAMURA and Ryohei ITO estimate the scatter value by subject thickness and pre-calculate scatter ratio data base for each pixel over entry image area. Other authors use thickness coupled with aid from Monte Carlo simulations. Among these, the point spread function or convolution kernel scatter models are used. In all estimation techniques, the scatter x-ray patterns are adjusted to specific parameters for each situation of radiographic examination. A particularly interesting one was based on a subject thickness estimation technique. These researchers used the pixel value in image to estimate the thickness of all anatomical parts in order to estimate scatter x-rays on an image. For some research groups, the Monte Carlo simulation is frequently used in conjunction with convolution scatter x-ray models. There is work being done over several modalities in radiological examinations, not only for general radiography, but also in computed tomography and digital breast tomosynthesis. All these applications have benefited from scatter correcting techniques. There are other the examples provided in these articles, as well

Atila Ersahim *et al.* used a two dimensional convolution of the scatter-glare point spread function to estimate scatter x-rays of image. There are functions involving thickness of subject, beam energy, field size, and air gap of imaging geometry. Theses estimations used Lucite as a tissue equivalent material and measured the primary and scatter x-rays with a beam stop technique.[13] Elena Marimón *et al*. approximates scatter x-ray by a two dimensional low pass convolution filter of the primary image. This approximation was first introduced by a Love and Krunger study. The scatter point spread function was obtained from the Monte Carlo simulation using a GEANT4 toolkit. The experimental geometry was defined in order to simulate the geometry of a mammography system where the phantom image was acquired.[8] Danyk A. Y. *et al*. determined the scattering kernel functions using realistic Monte Carlo simulations and robust scattering estimation algorithms. The set of resulting data is subjected to statistically processes in order to identify a set of the most relevant scattering kernels by clustering analysis. The kernel is used to train the classifier for identifying which area of an image that scatter kernels should be applied.[15] O Diaz *et al*. developed a fast kernel-based methodology for scatter field estimation which is more rapid than the Monte Carlo simulation. This method is thickness-dependent scatter kernel and accounts the full geometry in real digital breast tomography. The computational time is more than five time faster than Monte Carlo simulation and the estimation of error is equal to or less than 10%.[16] K. Kim *et al*. proposed a new model-based radiography restoration method based on simple scatter-degradation. The two models included in their work are the Scatter-degradation and Radiography restoration models. The image restoration function was investigated to determine its relevancy as a well suited function. They found that the dark-channel prior is more suitable for any proposed restoration.[17]

4. Image processing algorithms

First, a simple model has been proposed in which the primary and scatter x-ray is combined to form the radiographic image. The scatter x-ray is directly removed from the original image. Some authors add image processing steps like noise reduction to remove any deteriorating effects of which they are concerned. Ryohei ITO uses this model for any scatter correcting processes. After scatter estimation, he removes the signal amount corresponding to scattered x-rays in the image. Next, the noise caused by the scattered X-ray component and the graininess is reduced in order to improve the noise reduction processing as indicated in step 4 of diagram in Figure 4.[13] Andreas Fieselmann subtracted the low-frequency component of scatter from the original image. No more details about image processing steps taken after correction of scatter x-rays were indicated in this work.[14]

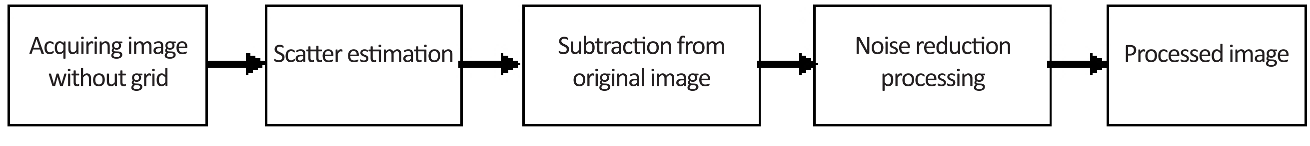

Figure 4 Simple model; scatter is subtracted from original image.

The following is a summary of Grid effect simulation. When a real anti-scatter grid is used, the x-ray photons from patient pass through a grid and are reduced in accordance with the primary and scatter transmission factors of the grid. The scatter x-ray is reduced more than the primary due to grid structure, making the image contrast better. The reduction factors of primary and scatter x-rays are grid ratio, grid frequency, and lead content. This image enhancement method uses the same image processing algorithm that is designed to simulate anti-scatter grid effect. The scatter x-ray estimation technique can calculate the x-ray scatter parts separately from the total signal detected by the image detector and allows the primary x-ray to be calculated immediately. Grid effect simulation algorithms does the contrast improvement work by calculating the reduction of the primary and scatter x-ray using the same factors found on a real grid. After taking the examination, the resulting image is obtained and is similar to the one that was acquired with an anti-scatter grid.[1]

Granularity or noise is the one deterioration factor that arises from scatter x-rays. This component is called "structure less" noise. When a scatter x-ray is reduced, image contrast is improved, but noise effect still remains in an image. A filtering technique can improve the granularity in an image [1]. Grid effect and granularity improvement processes are shown in Figure 5.

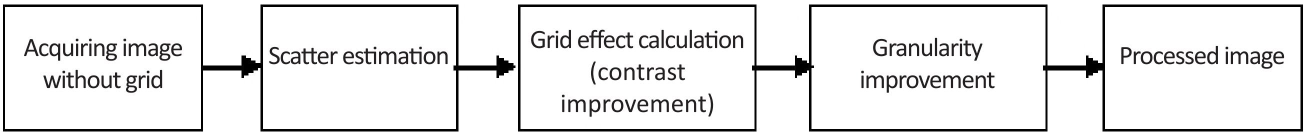

Figure 5 Grid effect simulation and granularity improvement process.

Detlef Mentrup adjusts the scatter x-ray image with a parameter in order to remove parts of them from the original. The resulting image consists of a primary that contains small bits of scatter. The single parameter used to control the algorithm is the grid selectivity which is a quality that is defined in the IEC standard 60627. Scatter x-ray is removed by subtraction from the original image. Afterward, the image needs to be adjusted to obtain a balance between over and underexposed areas in order to bring out the details in very dark and very bright regions necessary for better image quality[2] as shown in Figure 6.

Figure 6 Grid simulation using selectivity parameter for contrast enhancement.

Another technique called the deconvolution model is based on the principle that scatter x-rays are a mathematical convolutions resulting from the primary or the measured image with scatter kernels. Deconvolution or Inverse Fourier Transform can be used to calculate the primary from an original image and the estimated scatter x-rays. J L Ducote advocated using the image deconvolution model to decouple the convolution of scatter from the primary image. The convolution kernel in this work is specified to be used with both primary and scatter. In order to separate the primary x-ray, J L Ducote proposed that the measured image is equal to the convolution of the primary with this specific kernel. The advantage of Inverse Fourier Transform is that it can calculate the primary x-ray which is present in a measured image.[20] Danyk, A.Y. developed a scatter correction technique that improves image contrast to near the possible maximum. This work is based on the idea that scatter can be present as the linear combination of the primary and point spread function or scatter kernel. This work was a simulation study done on a few homogeneous subjects. The measured image is the combination of the primary and the convolution resulting from the primary and scatter kernel. Therefore, the Inverse Fourier Transform can approximate the primary x-ray which is present in measured images in ways similar to the work of J L Ducote. In order to maximize image contrast, clustering analysis and classifiers for identification calculation technique are used.[16]

5. Image quality and patient dose assessment

When comparing the scatter correcting image with non-grid and grid image by visual perception method it is found that the scatter correcting image is clearly an improvement over the ono-grid images. This is especially true in chest and abdomen radiography.[3] Contrast to noise ratio (CNR) is another method currently being used. This method used a Burger and CDRAD phantom and an IQF

(Image quality figure) value to evaluate contrast improvement. Ryohei ITO shows that CNR and IQFinv of image using scatter correcting method is equal to an image using 6:1 grid.[3]

Signal to noise ratio (SNR) used by IBEX Innovations Ltd to compare SNR of scatter correcting and grid image by imaging a phantom that was made up from a PMMA. The results showed that the mean SNR of scatter correcting images had an improvement by a factor of 1.16.[12]

Contrast improvement factor (CIF) proposed by Detlef Mentrup used CIF to compare the image quality of a water phantom and a Lungman phantom. It was found that there was strong agreement that existed between the ICF obtained with the scatter correcting method and the ICF obtained with the grid. For clinical applications, scatter correcting techniques demonstrated a significantly higher image quality for grid-less bedside chest radiography when patient dosages had decrease by 0.6.[2]

V. Tóth showed that the scatter correcting software significantly improved image quality of grid-less bedside chest x-ray, but did not reach the level of quality found in a chest x-ray equipped with an anti-scatter grid. User benefits from the significant reduction in patient dose.[21]

Conclusion

The image processing algorithms used for x-ray scattering correction still are the main concepts used for enhancing a radiographic image. Currently, commercial software packages have been developed by various x-ray equipment manufacturers that have largely replaced anti-scatter grid usage. The enhancement algorithms based on scattering estimation and scatter correcting method. Monte Carlo scattering x-ray pre-calculated database and patient thickness estimation technique used to adjust the algorithm for each specific situation of examination. Users will have the benefits of improved image quality, it will allow reduced dosages for patients, and there is a general improvement in efficiency resulting from using these algorithms necessary for improved workflow that will add to the overall improvement in patient care.

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