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Medical X-Ray Imaging with Scattered Photons

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ABSTRACT

All medical x-ray imaging today is done using the transmitted photons, i.e., those x-ray quanta which do not suffer any interaction within the patient. An alternative is to use the more plentiful scattered photons. Backscatter is almost entirely Compton (incoherent) scatter, which is principally sensitive to the number of electrons per unit volume. Forward scatter is dominated by coherent scatter, which is the basis of x-ray diffraction. Its cross section varies with angle and photon energy in a material-specific manner, even for amorphous materials. The dependence on Z and chemical structure allows it to be very useful in distinguishing tissues within the patient. Many workers have demonstrated utilization of both types of scatter in the lab, but it has been difficult to compare the performance of these systems with conventional transmission imaging. Therefore, we devised a semi-analytic model of scatter imaging. Our calculations predict that for some imaging tasks the contrast and signal-to-noise ratio achieved by collecting a portion of the scatter (in an annular cone) will be superior to that achieved by conventional transmission imaging, for the same number of photons incident on the patient. Our analysis is reliant on the limited published data for coherent scattering for biological materials. To confirm and extend these data, we are measuring the coherent scatter form factors for biological materials, and also plastics, using diffractometers made available to us by the National Research Council of Canada in Ottawa.

1. INTRODUCTION

The basis of diagnostic radiology conventionally has been to form projection images using x-ray photons. The spatial distribution of energy absorbed by the receptor forms the image. The radiation from a diagnostic x-ray tube spans a range of energy per photon from $h\nu \approx 16$ keV up to that corresponding to the potential, usually ≤ 150 kV.

Attenuation in the patient is via three processes: photoelectric absorption, inelastic or incoherent (often called Compton) scattering, and elastic or coherent scattering. For $h\nu > 25$ keV, over 50% of the interactions in tissues are scatterings.[1] Figure 1 shows for the two scattering processes, for H_2O, the differential cross sections per unit angle versus scattering angle θ. The incoherent cross section is reduced for $\theta \approx 0$ because of the atomic binding of electrons. The coherent cross section is strongly peaked at a small θ and at 35 keVexceeds that for incoherent for $\theta < 17.5^o$. For lower $h\nu$ this angle is larger. Hence, radiation which reaches the image receptor after one scattering has a large coherent-scattered component.[4-6]

In conventional projection imaging, most of the photons approaching the image receptor have been scattered in the patient; the scatter-to-primary ratio can be as high as[7] 10. Scatter can significantly degrade projection image quality [contrast (*C*) and signal-to-noise ratio (*SNR*)]. The usual fix is geometric rejection using an antiscatter grid of miniature Pb septa arranged like a venetian blind.

An alternate approach is to use the scattered x rays. Previous investigators have demonstrated imaging for medicine using the forward scatter[8-10] and backscatter,[11,12] and use of forward scatter for non-destructive testing.[13,14] In principle, simultaneous measurements of forward scatter, backscatter, and transmitted primary x rays could be made, as in Fig. 2.

2. MODEL OF X-RAY SCATTER IMAGING

To date, it has been difficult to compare quantitatively the performance of scatter imaging to primary imaging, and to compare different scatter imaging approaches to each other. For primary imaging the standard analysis tool is the model of Motz and Danos.[15] We have formulated analogous models for scatter imaging[16] and used them to quantify the ultimate performance of scatter imaging based on the fundamental input parameters, namely the differential scattering cross sections, independent of the engineering of a particular apparatus. The models are semi-analytic and intentionally simple. For a given photon fluence entering the patient, the models calculate C and *SNR*, where the "signal" in the *SNR* calculation is the difference in measurement between two objects that are to be distinguished. We analysed both forward scatter (2^o- 12^o) and

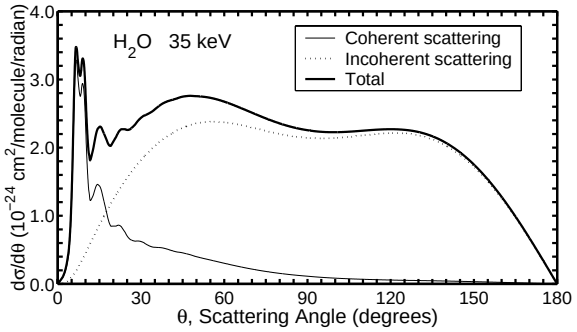

Figure 1. Cross sections at 35 keV for x-ray scattering in H_2O at angle θ into a ring of infinitesimal width $d\theta$. The integrals of these curves are the total scatter cross sections. (From data of Refs. 2, 3).

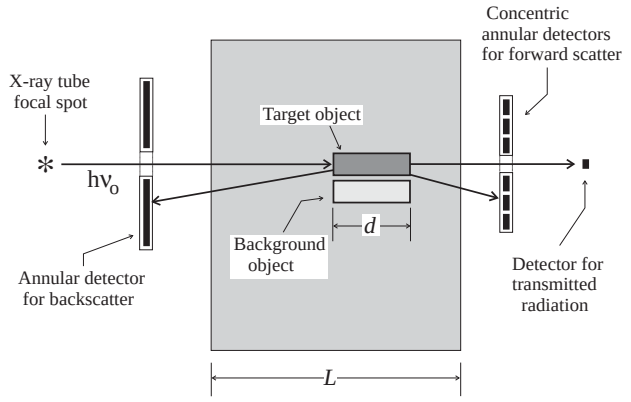

Figure 2. Information can be acquired using the transmitted primary, backscattered, and forward-scattered x rays.

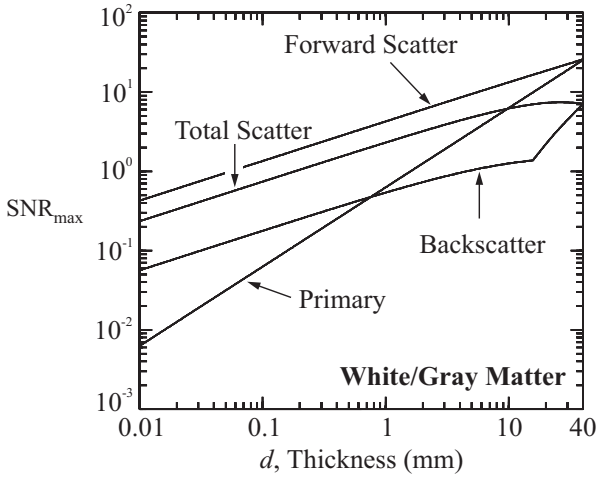

Figure 3. The maximum SNR over all energies for distinguishing gray matter from white matter, of thickness *d*, cross sectional area 1.0 mm^2, inside a 15 cm diameter spherical H_2O phantom. Incident energy fluence fixed at 1.88 x 10^{11} keV cm^{-2}. (From Ref. 18).

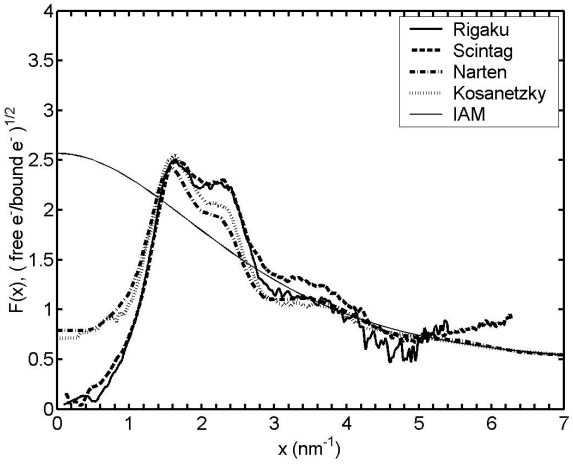

Figure 4. Comparison of coherent scatter form factor for H_2O measured by us (Rigaku and Scintag machines), Narten,[2] and Kosanetzky *et al*.[20] The IAM curve is theoretical, and assumes no inter-atomic interference.

backscatter (158°-178°) imaging. We also calculated *C* and *SNR* assuming hypothetical capture of all scatter over 4π steradians. Our forward-scatter model was verified experimentally using polyenergetic beams and plastic targets.[17]

Figure 3 shows numerical predictions for distinguishing gray from white brain matter in neuroradiology. Conventional CT scanners can just barely do this, as the primary contrast is only ≈ 0.5 %. Two small targets of white and gray matter are modelled to be inside a sphere of water, radius 7.5 cm. Shown are maximum achievable values of *SNR*, where the maximum is obtained by optimizing hν. Using forward scatter gives a better *SNR* than using primary for all target sizes $d < 40$ mm. Also, using only the forward scatter gives a better *SNR* than using all the scatter. Although there are less photons available in the former case, so that the fractional Poisson counting noise is greater, the difference in *total* scatter cross section is larger, and therefore so is the *SNR* of the gray/white matter difference. These results are for monoenergetic beams. Calculations using typical medical polyenergetic spectra show that the *SNR* reduction due to spectral blurring is modest.[19]

3. MEASUREMENT OF SCATTER CROSS SECTIONS

The coherent scattering cross section per electron per steradian is $\frac{d_e\sigma}{d\Omega} = \frac{r_e^2}{2}(1+\cos^2\theta)\frac{F^2(x)}{Z}$, where r_e is the classical electron radius, Z is the number of electrons in the stoichiometric unit considered, and *F* is the coherent scatter form factor. The parameter *x* arises from considerations similar to Bragg's Law for crystalline specimens, and is $x = \lambda^{-1}\sin(\theta/2)$. While a number of groups have reported values of $d_e\sigma/d\Omega$ or of *F* for tissues,[20-25] there remains uncertainty about the values, and

this induced us to commence our own measurements.[26,27] Measurements can be made at any hν, and then translated to the energy of interest through *x*. Using low hν facilitates the experiment, since the data shift to higher θ. We have used two diffractometers made available to us by the National Research Council of Canada: Rigaku (6.94 keV), and Scintag (8.06 keV). Using two machines, at different energies, allows methodology developed on one to be checked on the other.

We have found that accurately measuring diffraction from amorphous materials on such machines is difficult. The machines are designed for accurate peak location but continuous background interferes with amorphous sample data. Figure 4 shows a comparison of our results for H_2O with that of others. The variation is significant and must be resolved.

4. CONCLUSIONS

By treating scattered radiation as an additional information source rather than as a nuisance to be suppressed, a new dimension is added to x-ray imaging. Our calculations predict that in some cases, such as neuroradiology, scatter images will have better *C* and *SNR* than projection images, for the same radiation dose. Alternatively, the *SNR* of conventional imaging could be matched by scatter imaging at lower dose. Better data for tissue scattering cross sections are needed.

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