On a Novel Approach to Compton Scattered Emission Imaging

Mai K. Nguyen, Tuong T. Truong, Clemence Driol, and Habib Zaidi

***Abstract*—Imaging processes built on the Compton scattering effect are currently under intense investigation. However, despite many innovative contributions, this topic still pose a formidable mathematical and technical challenge. In this work, we argue that, in the framework of single-photon emission imaging, collecting Compton scattered radiation from an emitting object, allows to image the radiotracer distribution in vivo. Data is acquired by a stationary collimated gamma camera under the form of *compounded conical projections* of the activity density function. Mathematically, the image formation process is described by the so-called Compounded Conical Radon Transform (CCRT) and three-dimensional object reconstruction is based on an inversion formula of the CCRT. We perform numerical simulations to show the feasibility of this new imaging modality, which offers the remarkable advantage of operating in stationary mode without the need of bulky and cumbersome spatial rotational mechanism of conventional gamma cameras. This is highly attractive for applications in medical imaging, industrial non-destructive evaluation, nuclear waste storage surveillance and homeland security monitoring. Finally, to improve drastically the sensitivity, we introduce a new feature allowing to acquire data without mechanical collimation and support the findings with some preliminary simulation results.**

***Index Terms*—Compton imaging, Compton scattering, computed tomography imaging, SPECT, SPECT reconstruction.**

I. Introduction

SINCE the early days of the last century, ionizing radiation in particular gamma-rays, thanks to their penetrating property, are used to image the interior of objects. This is done with an external source of radiation, which illuminates an object and projects the shadows of its internal structure on a detecting surface. Later on, it was shown that a true three-dimensional image can be reconstructed if there is a sufficient number of such two-dimensional projections generated by the displacement of the source/detector assembly on a specific space curve. In essence, this reconstruction procedure relies on the inversion of the so-called x-ray transform, which is known since many decades [1], [2].

It should be noted that one can also turn passive objects into radiating ones. For example, in nuclear medicine, this is achieved by injecting a radiotracer such as $^{99m}\mathrm{Tc}$ to produce a nonuniform distribution of the tracer within the patient's body. An image (or projection) can be produced by a parallel-hole collimated gamma camera, set to register 140 keV photons emitted by $^{99m}\mathrm{Tc}$. If the gamma camera is made to rotate around the patient's body, so as to generate a series of images from distinct view angles, then the tracer distribution hidden inside of the body can be reconstructed. This imaging modality is known as single-photon emission computed tomography (SPECT) [3].

Owing to the interaction of radiation with matter, gamma-ray imaging is plagued by Compton scatter, which ruins image quality and degrades spatial resolution. Contribution from scattered photons should at best be eliminated or at least reduced [4]. However, a more astute point of view would be to take advantage of their properties either for improving image quality or for generating new imaging processes. In fact, there are many ways to exploit Compton scattering for imaging purposes. This activity goes back to the 50's [5] and has remained vivid ever since [6]. The idea has many highly desirable features. In the field of diagnostic medical imaging, radiography using scattered radiation could provide a direct and quantitative measurement of the density of the studied object. In non destructive testing, three advantages can be pointed out. It permits to place both the radiation source and the detector on the same side of the object. It has also greater sensitivity to low density materials such as gases. Finally, it allows direct spatial definition with high contrast resolution. With the advent of the x-ray computerized tomography (CT) scanner, interest in Compton scatter imaging has waned for a while. But research in this field has remained very much alive and a large variety of imaging techniques have been developed [3], [7], [8].

Earlier modalities for Compton scatter imaging are classified according to the way measurement of the spatial distribution of scatter radiation is done or the number of simultaneous volume elements being scanned: i.e., point by point, line by line, or plane by plane (see reviews [7], [8]). Most of the devices work at constant scattering angles (generally at 90 degrees). In the mid 90's, the concept of Compton scatter tomography was introduced by Norton [9], and subsequently developed by many other workers ([10], [11]). A prominent example in which Compton scattering acts as imaging agent without mechanical

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collimation, is the co-called Compton camera [12]–[14], as well as gamma-ray tracking imaging or the like. More recently Compton scatter imaging using annihilation pair photons with coincidence measurements has appeared on the scene as a yet unexploited imaging technique [15]. Related concepts allow enhancing the detection efficiency by reconstructing a significant fraction of events which underwent Compton scattering in the crystals [16].

In this work, we describe a different approach in which Compton scattered radiation is used for three-dimensional emission imaging. We first point why scattered photons should be used instead of primary photons in gamma ray imaging. Then we show how they are actually used through a mathematical modeling of image formation by a new invertible Radon transform. Finally, after studying the features of the corresponding point spread function (PSF), we show results of numerical simulations to demonstrate the feasibility of this new imaging principle.

II. Why Use Scattered Photons?

To convey heuristically our idea, we shall first speak of optical analogies, which are closer to everyday life and easier to grasp. Consider a point source emitting a monochromatic red light by a clear day. A human eye, placed at a certain distance from this source, would see only a red spot. But if a fog cloud sets in, then the eye would see a diffuse red cloud much larger than the red spot detected before. The fog cloud has made itself visible because light is scattered by the fog droplets and re-emitted as scattered light by the fog droplets acting as scattering centers. The fog cloud has become a kind of secondary radiating object, more visible to the human eye and indirectly, it reveals the existence of a red light source in it.

In the gamma range, a similar phenomenon occurs. A single emitting point source behaves exactly in the same way with respect to a gamma-ray sensitive "eye". But if the gamma-ray emitting point source is embedded in a medium of finite volume – which plays the role of the fog cloud – then light wave scattering by water droplets is replaced by Compton scattering of emitted photons with electrons of the surrounding medium. If light emerges from scattering without changing its wavelength, the emerging scattered gamma photon looses energy over a continuous spectrum as energy is transferred to electric charges of the traversed medium. The gamma-ray sensitive eye would now "see" a *red-shifted polychromatic radiation* emanating from the scattering medium volume. The wavelengths of the scattered gamma rays are longer and given by the Compton formula [17]:

$$\lambda = \lambda_0 + \frac{h}{m_e c}(1 - \cos\omega),$$

where λ_0 is the originally emitted wavelength of the radiopharmaceutical, m_e the electron mass, h the Planck constant, c the speed of light in vacuum and ω the Compton scattering angle.

Now instead of having a single point source, consider a nonuniform three-dimensional distribution consisting of a set of point sources embedded in a medium of finite volume. The same effect, but much stronger, is reproduced by a gamma-ray sensitive "eye", such as a standard collimated gamma camera.

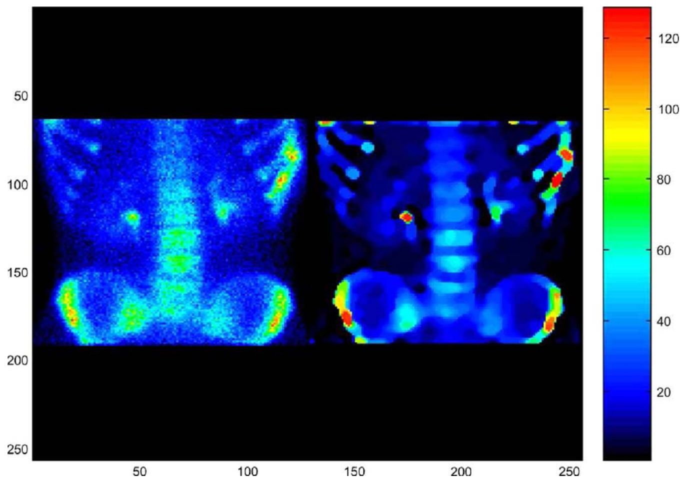

Fig. 1. Image restoration without (left) and with (right) the use of Compton scattered radiation.

If this gamma camera is then set to work at fixed energy (or analogously with a "color filter" in the optical visible range), one would obtain (ignoring the limited energy resolution of the detector) a series of energy labeled images (or analogously "colored" images of the object). Thus one may raise a natural question: is scattered radiation any good for imaging?

As early as 2001, we have observed that scattered radiation images of an object may be sorted out at a given energy (or at a given wavelength) using standard gamma camera data acquired in listmode format (see [18]). Then by a careful treatment, they may be taken into account in the restoration of nuclear medicine images, where small details, unresolved before, emerge clearly separated from each other. In Fig. 1, small hot spots or nodules invisible on the left image become perfectly distinguishable on the right image. This is, in fact, very valuable for medical diagnosis and therapy treatment planning. Thus the newly revealed resolving power brought by scattered radiation has appeared very attractive for further development.

In this work, we go one step further and propose to collect data for three-dimensional Compton scatter imaging. This amounts to instate a novel imaging principle, referred to as *scattered radiation emission imaging*, which differs from the principle used in the Compton camera as well as from gamma-ray tracking imaging modalities. Here *naturally* scattered radiation from the object itself is collected for imaging purpose, *without* the use of a secondary solid-state detector and thus *without* requiring any hardware modification of commercially available gamma cameras.

III. How are Scattered Photons Used?

In conventional SPECT, only primary photons (e.g., of energy 140 keV for $^{99m}\mathrm{Tc}$) are collected by a collimated gamma camera, registering the contribution of the sum of point sources lying on a line as *linear projection* of the object activity density (see Fig. 2).

To get a set of complete data needed for three-dimensional reconstruction, it is necessary to move the detector in space around the object, so as to get all the lines traversing the object in all directions. Hence for a given single linear projection, only voxels

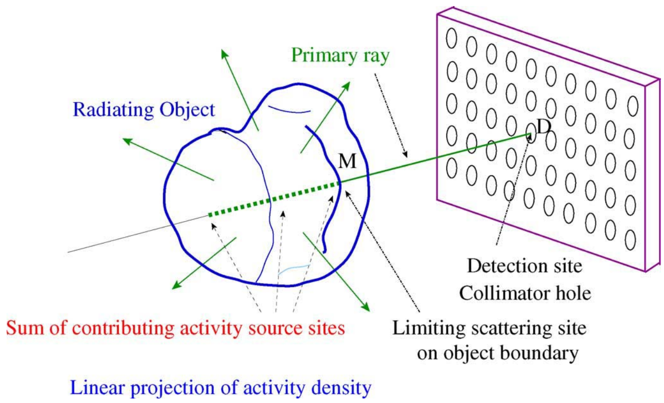

Fig. 2. Principle of data acquisition in conventional emission imaging.

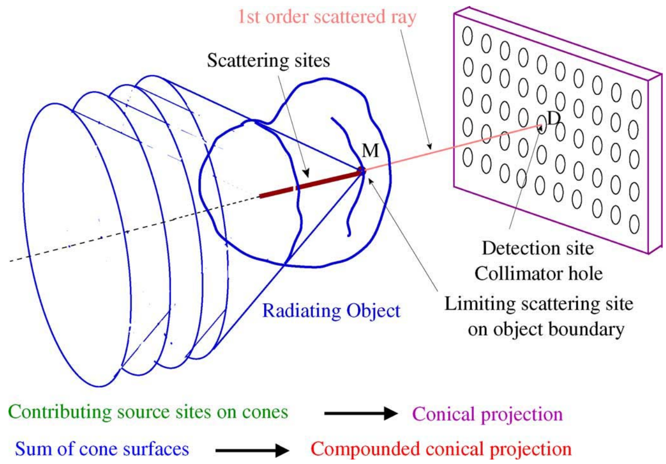

Fig. 3. Emission imaging using Compton scattered radiation via *compounded conical projections*.

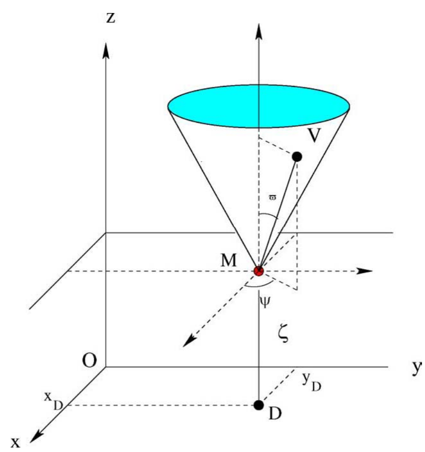

Fig. 4. Coordinate system used in CCRT computation.

lying on a straight line are concerned. Because of photon attenuation and the presence of a physical collimator, few from these primary photons (actually one out of 10^4) would actually reach the detector [19].

A natural alternative to imaging with primary photons is to use the more plentiful scattered photons, as advocated in [20]. Of particular interest are photons generated by scattering with electrons in the medium surrounding the emitting point sources of the radiating object, but not those scattered by an external scattering detector, as in the Compton camera, which is primarily designed to achieve electronic collimation. On the whole, scattered photons are predominantly single-scattered, since higher order scattered photons have much lower probability of being detected, as reported in many Monte Carlo simulation studies [4], [21]. These photons can be collected at a given energy by a collimated detector. We can easily picture them as originating from the part of the object inside a *conical volume*, having for axis the axis of a collimator hole and for vertex the intersection of this axis and the object surface boundary facing the detector. This conical volume contains a vastly superior number of contributing point sources as compared to the number of primary radiation point sources in SPECT. It may be viewed alternatively as a "vertical" sum of a continuous set of "parallel" cones with vertices lying on the common cone axis, which is also the locus of photon scattering sites. Thus each of these cones constitutes a *conical projection* of the activity density. The measured photon flux density at the detector site is a sum of conical projections (Fig. 3) and is called *compounded conical projections* [22], [23]. It should be noted that data acquisition no longer requires to rotate the detector in space (as it is done in a conventional emission imaging). Instead it is sufficient to adjust the energy acquisition window on a motionless detector. Thus it appears that the role of the spatial rotation angle is played now by the scattering angle.

IV. Mathematical Modeling

In this section, we describe the image formation process by a compounded conical projection. Using the coordinate system of Fig. 4 and following the path of a photon from emission to absorption via one scattering at a site $\mathbf{M}$, the expression of the flux density on the detector at a site $\mathbf{D} = (x_D, y_D, 0)$, $g(x_D, y_D, \omega)$ has the form of a linear integral transform of the object activity density $f(x, y, z)$,

$$g(x_D, y_D, \omega) = \int\int\int dx\, dy\, dz \; \mathcal{K}_{PSF}(x_D, y_D, \omega | x, y, z)\, \widehat{f}(x, y, z), \quad (1)$$

where the function $\widehat{f}(x, y, z)$ is defined by

$$\widehat{f}(x, y, z) = \int d\zeta\, \nu(\zeta)\, f(x, y, z + \zeta), \quad (2)$$

the integration kernel is

$$\begin{aligned} &\mathcal{K}_{PSF}(x_D, y_D, \omega | x, y, z) \\ &= K(\omega) \times \nu\left(\sqrt{(x - x_D)^2 + (y - y_D)^2 + (z - \zeta)^2}\right) \\ &\quad \times \delta\left(\cos\omega \sqrt{(x - x_D)^2 + (y - y_D)^2} - (z - \zeta)\sin\omega\right), \end{aligned} \quad (3)$$

$K(\omega)$ is the Compton kinematic factor, and $\nu(d)$ is a function describing a photometric factor for a distance d, e.g., $\nu(d) = 1/d^2$. By definition, $g(x_D, y_D, \omega)$ is called the compounded conical Radon transform (CCRT) of $f(x, y, z)$.

We have not included photon attenuation factors to avoid unnecessary complications which would mask the main idea.

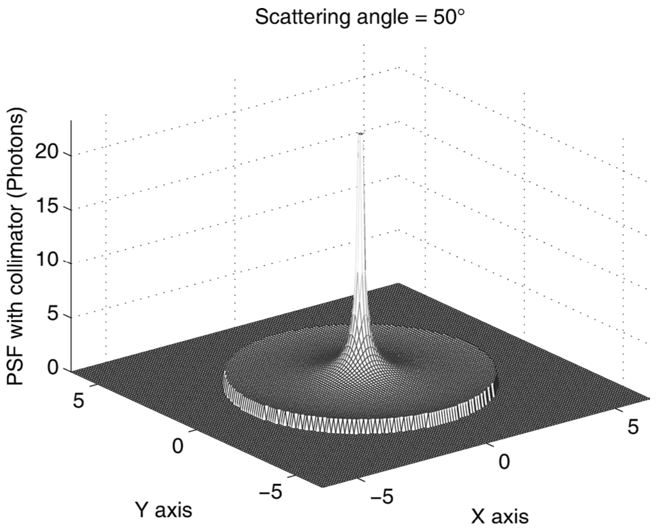

Fig. 5. Two-dimensional plot of the PSF at scattering angle of 50 degrees with collimator.

The inversion of the kernel $\mathcal{K}_{PSF}(x_D, y_D, \omega | x, y, z)$ is then obtained via a kind of central slice theorem in Fourier space of the detector plane for the function f^*, to which one should add a deconvolution to get the Fourier transform $\bar{F}(u,v,w)$ of $f(x,y,z)$, see [22], [24]:

$$\begin{aligned} &\bar{F}(u,v,w) \\ &= \int d\sigma \exp[2i\pi\sigma w] \frac{[-|z|\sqrt{u^2+v^2}]}{\mathcal{J}(w)} \\ &\int_0^\infty t\, dt\, J_1(2\pi|z|t\sqrt{u^2+v^2}) \\ &\left[Y\left(\omega - \frac{\pi}{2}\right) \frac{\partial}{\partial t} \frac{G(u,v,t)}{K(t)} + Y\left(\frac{\pi}{2} - \omega\right) \frac{\partial}{\partial t} \frac{G(u,v,-t)}{K(-t)} \right], \end{aligned} \tag{4}$$

where

— $\mathcal{J}(w)$ is the Fourier transform of $\nu(x)$,
— $J_1(x)$ is the Bessel function of order 1,
— $Y(x)$ is the Heaviside unit step function,
— $t = \tan\omega$,
— $G(u,v,t)$, the Fourier transform of $g(x_D, y_D, t)$,
— $K(t)$ is the Compton kinematic factor as a function of t.

Finally $f(x,y,z)$ is recovered by three-dimensional Fourier transform.

V. Point Spread Function and Results of Simulations

A way to get an idea of what the CCRT could be or do is to construct its Point Spread Function (PSF). To concentrate on the scattered imaging principle, we make some simplifying assumptions to stress only the effects of scattering [22]:

- absence of attenuation for the propagating radiation [19],
- constant density of electrons n_e in the scattering medium,

The PSF turns out to be explicitly computable as an analytic function. It has the shape of a Mexican hat as shown in Fig. 5 [24].

The gamma detector operates now at a *fixed position*. No coincidence detection, as in Compton cameras, is required. Numerical simulations have demonstrated the feasibility of this new imaging principle [24]. Issues related to higher order scattering contribution, nonuniform attenuation, Poisson emission noise, detection sensitivity, collimator efficiency are part of ongoing research with innovative extensions to transmission imaging, leading eventually to an efficient dual-modality imaging procedure [25].

To provide more convincing arguments regarding the viability of this idea, we present numerical simulations which illustrate the reconstruction of a simple cylindrical object using the analytic inversion formula with the following working conditions:

- the used γ-detector is a conventional SPECT camera. It has discretized dimensions N length units $\times N$ length units. We have chosen $N = 16$ to keep the calculations required at a reasonable level.
- the scattering medium is represented by a cube of dimensions $N \times N \times N$,
- the electron density in biological medium is $n_e = 3.5 \times 10^{23}$ electrons/cm^3,
- the radionuclide employed is $^{99m}\mathrm{Tc}$ with an activity concentration corresponding to 4.84 $\times 10^{10}$ counts per minutes per cm^3,
- the acquisition time per projection is set to 0.1 sec,
- the 3D original object (cylinder of height 6 arbitrary units) is placed at the center of the scattering medium (cube),
- the distance from the camera to the upper face of the scattering medium cube is $l = 200$ arbitrary units.

Fig. 6 represents the original object. Fig. 7 shows the series of images of the object at various scattering angles ω ($5° < \omega < 175°$). In Fig. 8, the reconstructed object in the absence of noise is illustrated with a relative mean square error $(\mathrm{RMSE}) = 1.2\%$, which is perfectly reasonable. We observe a good performance of the CCRT for modeling the new imaging process.

Concerning spatial resolution, the intrinsic resolution depends on the camera design (collimator, crystal, photomultiplier tubes and measurement electronics). The reconstructed system resolution is further determined by the reconstruction algorithm used. The inclusion of scattered radiation increases considerably the number of detected photons, which might contribute to improve the signal to noise ratio (SNR) and the resolution of the imaging system. To evaluate accurately the spatial resolution, it is necessary to use real data and to compare it with conventional methods which do not make use of scattered radiation. At the present time, it is too early to use our preliminary simulation results for this purpose. This work is ongoing using realistic experimental conditions.

Since our main objective in this paper is to show how to exploit advantageously Compton scattered radiation to suggest a new imaging principle, we focus on results illustrating the image formation process as well as image reconstruction from scattered photons.

In real situations, of course, one must take into account other factors such as photon attenuation by the medium, Poisson noise and the imperfections of the detector system including the collimator and electronics.

The case of uniform attenuation (often assumed in the literature) was included in our recent work [24]. The exact treatment of inhomogeneous attenuation poses enormous mathemat-

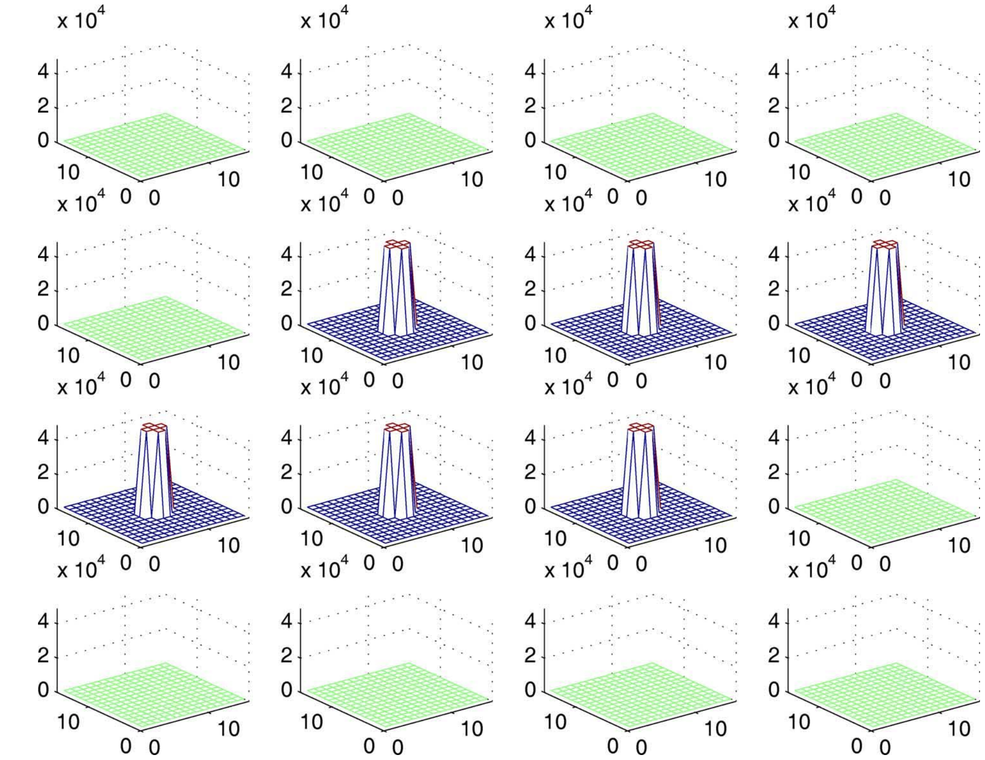

Fig. 6. Original object (cylinder) in a cube consisting of 16 transaxial planes.

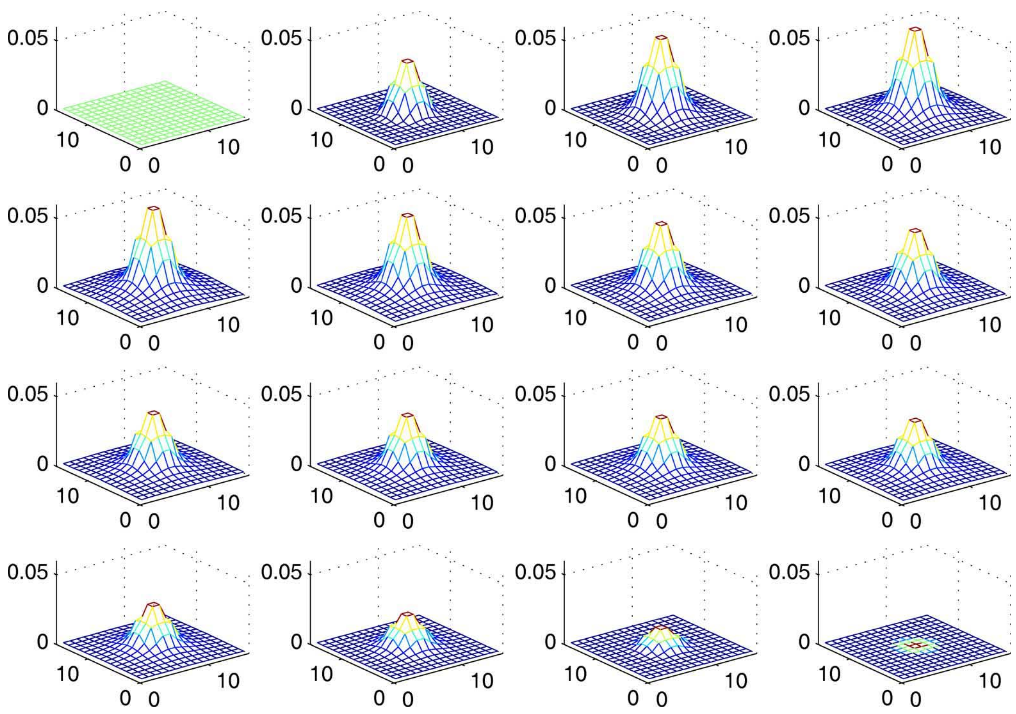

Fig. 7. Series of images parameterized by the angle of Compton scattering ω ($5^\circ < \omega < 175^\circ$).

ical difficulties. Concerning emission noise, several approaches have been suggested to deal with it such as Maximum Likelihood or wavelets method. They may be used for "denoising" the measured data beforehand or jointly with the inversion process.

As for the imperfections of the detector, the standard way for treating this problem is to make use of a response function usually modeled as a Gaussian defined both in spatial and energy coordinates. These issues are discussed in detail in [18].

VI. How to Increase Sensitivity?

As mentioned earlier, the presence of a mechanical collimator restricts severely the sensitivity of the imaging process. We

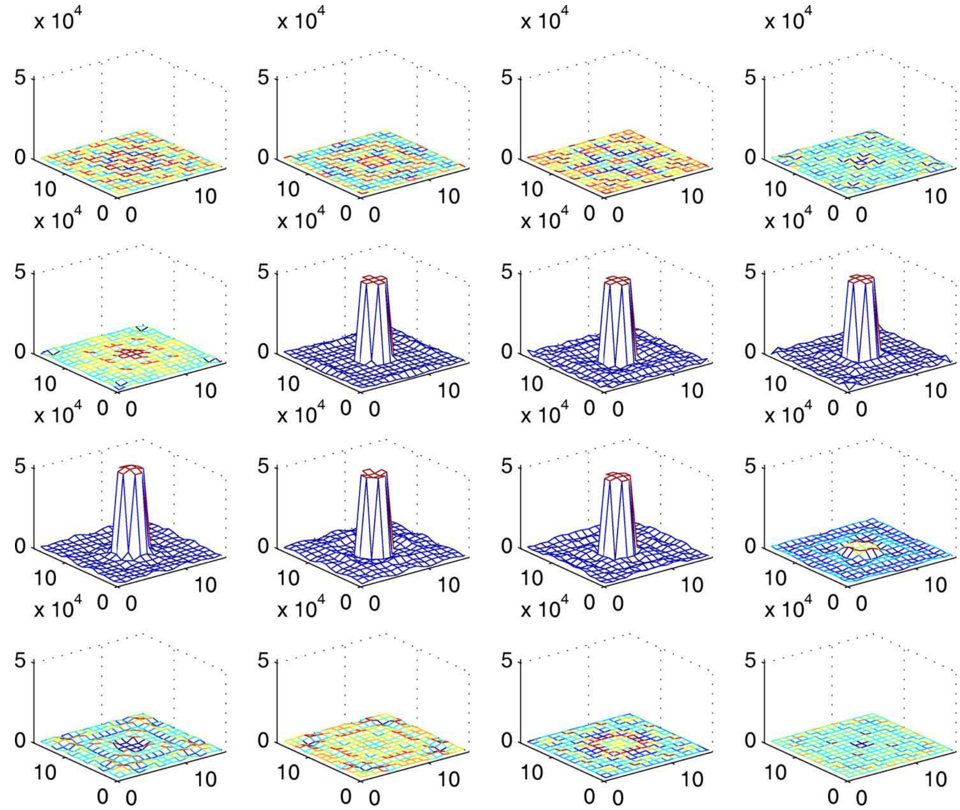

Fig. 8. Reconstructed object in the absence of noise ($\mathrm{RMSE} = 1.2\%$).

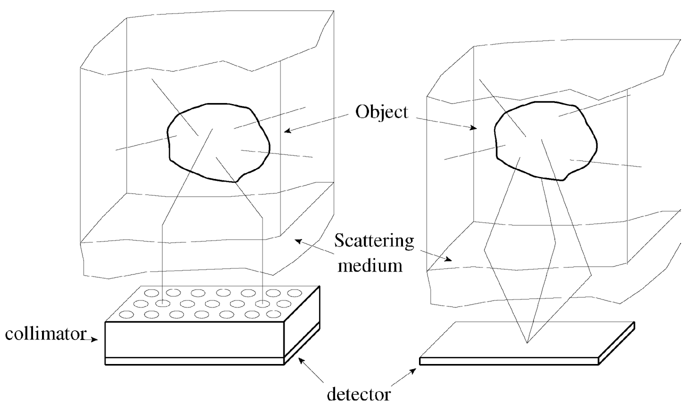

Fig. 9. Two imaging modalities using Compton scattered radiation with and without collimation.

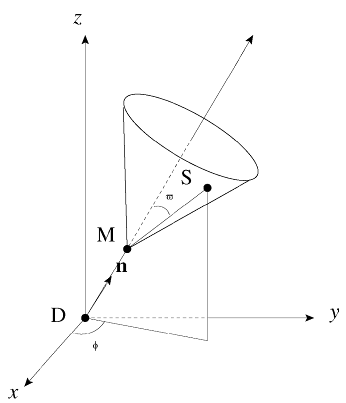

Fig. 10. A conical projection in the generalized CCRT.

have recently advocated a new functional modality following the principle of emission imaging by scattered gamma-rays without mechanical collimation [25], [26]. Removing the collimator from the detector allows gamma rays to reach a detecting pixel from all directions coming from the upper-half space of this site, therefore increasing the strength of the signal (Fig. 9). An introductory study in two dimensions has recently been performed [27] to demonstrate conclusively the feasibility of this idea and to motivate the present work.

The modeling of the image formation process is then done by a generalized compounded conical Radon transform, whereby one sums over conical projections at one detection pixel over all cone vertices in the upper half space. Fig. 10 shows the position of one conical projection in the generalized CCRT.

This transform is obtained by summing over all sites $\mathbf{M}$ for a given site $\mathbf{D}$ on the detector. The mathematical expression of one arbitrary conical projection is quite involved and given in [23]. The summation over all such objects can be still expressed as a linear integral transform of the activity density $f(x,y,z)$

$$g(\mathbf{D},\omega) = \int dx\,dy\,dz\;\mathcal{K}_{PSF*}(\mathbf{D},\omega|x,y,z)\,f(x,y,z). \quad (5)$$

Unfortunately, the explicit form of $\mathcal{K}_{PSF*}(\mathbf{D},\omega|x,y,z)$ is too complicated to yield a simple interpretation, and will not be addressed here. However, this PSF, although no longer a computable function is in fact an integral of the electronic density

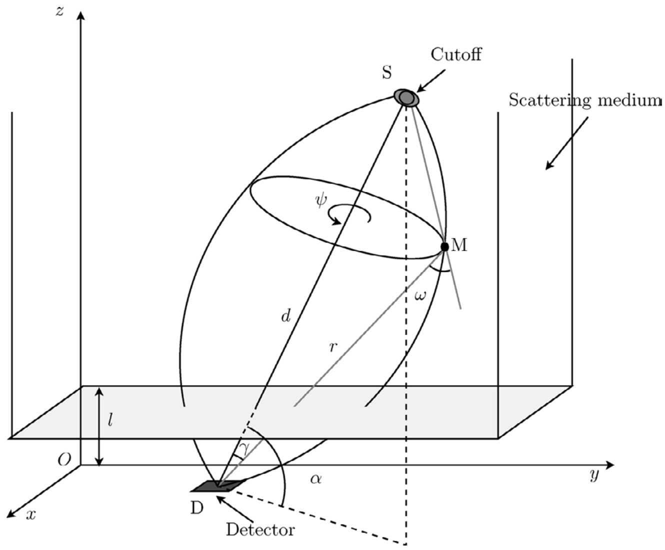

Fig. 11. Torus surface of scattering sites for the case without collimator.

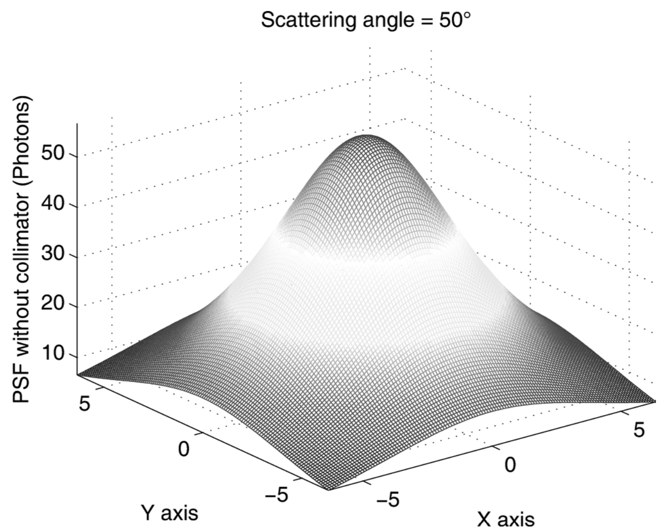

Fig. 12. Plots of the PSF at a scattering angle of 50 degrees without collimator.

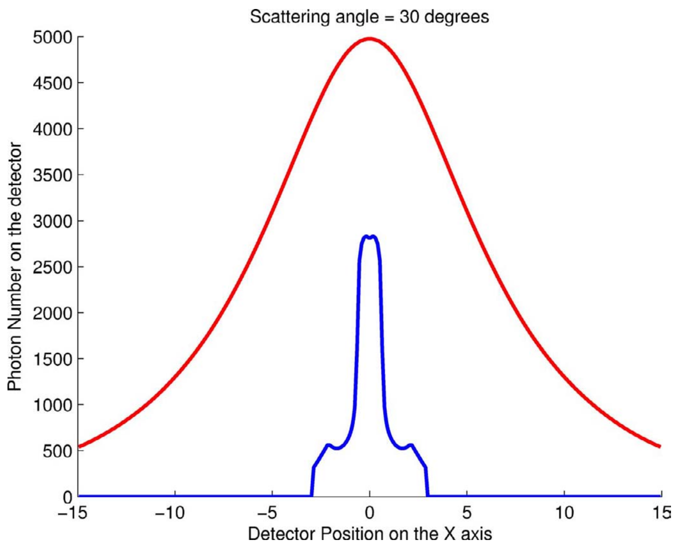

Fig. 13. Comparison of the PSF with (lower blue line) and without collimator (upper red line).

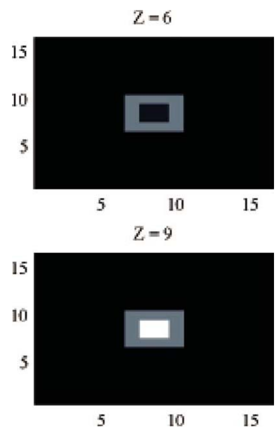

Fig. 14. Two representative slices of the original object illustrating two transaxial slices corresponding to the 6th and 9th planes, respectively.

over the surface of a torus of revolution whose axis is the line connecting the point source to the detection point (Fig. 11).

The shape of the PSF is now completely different compared to the collimated geometry (Fig. 12).

To compare with the collimated detector geometry, Fig. 13 shows the computed PSFs for both cases at a scattering angle of 30 degrees. The PSF without collimator is about 10 times stronger than that with collimator.

To demonstrate the viability of this idea, we have used data generated for a simple object, and applied classical algebraic reconstruction methods. We have taken a simple source immersed in a cubic scattering medium. The source itself consists of two concentric cubes with different activity concentrations (Fig. 14).

A 256×256 pixels detector is placed on the xy plane. The pixel size is 0.4×0.4 mm^2. The scattering medium is a rectangular box of dimensions 30 cm by 30 cm by 15 cm, which is at a distance of 1 cm above the planar detector. The electronic density inside the scattering medium is $n_e = 3.341023$ electrons/cm^3 since most biological tissues have an electronic structure close to that of water. The radionuclide used in this simulation is $^{99m}\mathrm{Tc}$, which emits photons at an energy of 140.1 keV. The scattering medium is discretized with 13 voxels in x and y axis directions and with 9 voxels in z axis direction. The detector is reduced to 13×13 pixels. We construct the weight matrix of the medium by calculating from our previous models, for each point of the mesh, the PSF of the detector at the different scattering angles. The reconstruction is carried out using the conjugated gradient method with positivity constraint, see Fig. 15.

These preliminary results are incentive to pursue our investigation on this new imaging modality.

VII. Conclusion and Perspectives

In this work, we have presented a new approach to Compton scatter imaging. This concept exploits the gamma radiation naturally scattered by the bulk of the emitting object using a stationary gamma camera. Two operating modalities are examined. The first one uses a standard SPECT camera and its working principle is based on the inversion of a new integral

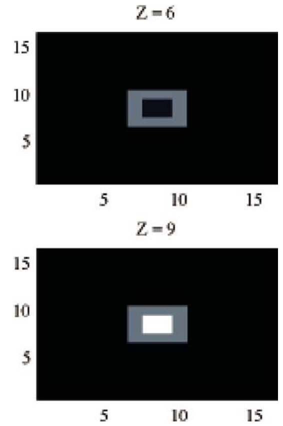

Fig. 15. Two representative slices of the reconstructed images corresponding to the object shown in Fig. 14.

transform, referred to as the compounded conical Radon transform (CCRT). Numerical simulations shows that it is feasible, probably with new generation of solid-state detectors having very high energy resolution. The second one operates without mechanical collimation and coincidence circuitry and is based on a generalized CCRT. Although no analytic inversion formula exists at present, preliminary numerical calculations with data generated using a forward model show promising results.

Future challenging topics to be discussed in this imaging process are:

— Stability and robustness against noise and perturbations,

— Accounting for non-uniform attenuation in the scattering medium,

— Treatment of higher-order scattering events [28],

From a technical standpoint, a new camera conception with a built-in high spatial and energy resolutions without a heavy and cumbersome mechanical apparatus for spatial rotation is to be designed. It would be well-suited for operation under challenging conditions. This should be highly attractive for a number of fields of application such as medical imaging, industrial non-destructive control, surveillance of radioactive hazardous material storage (or transport), localization of contaminations in a nuclear reactor, … etc.

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