Nuclear Medicine Communications, 2001, **22**, 1181–1184

Nuclear Medicine *Communications*

Scatter modelling and correction strategies in fully 3-D PET

H. ZAIDI*

Division of Nuclear Medicine, Geneva University Hospital, Geneva, Switzerland

Received 10 May 2001 and accepted 15 July 2001

Summary

PET offers the possibility of quantitative measurements of tracer concentration *in vivo*. However, there are several issues that must be considered in order to fully realise this potential. Whilst, a correction for a number of background and physical phenomena need to be performed, the two most significant effects are the photon attenuation in the patient and the contribution in the images of events arising from photons scattered in the patient and the gantry. The non-homogeneous distribution of attenuation within the thoracic cavity complicates the interpretation of PET images and precludes the application of simple scatter correction methods developed for homogeneous media. The development of more sophisticated techniques for quantification of PET images are still required. Recent progress in 3D PET instrumentation and image reconstructions has created a need for a concise review of the relevance and accuracy of scatter correction strategies. Improved quantification of PET images remains an area of considerable research interest and several research groups are concentrating their efforts towards the development of more accurate scatter modelling and correction algorithms. (© 2001 Lippincott Williams & Wilkins)

Keywords: PET, scatter modelling, scatter correction, quantification.

Introduction

The image quality and quantitative accuracy of positron emission tomography (PET) reconstruction are degraded by a number of physical factors including: (1) the finite spatial resolution of the imaging system and the resulting partial volume effect; (2) the attenuation of the photons travelling towards the detector elements; (3) the detection of scattered photons; (4) the limited number of counts one is able to collect when imaging patients; (5) physiological as well as patient motion; and (6) the reconstruction algorithm. While it is well accepted by the nuclear medicine community that the detection of Compton-scattered events degrade image quality, a common question asked by most nuclear medicine physicians is to what extent does scatter affect image interpretation and clinical decision making? Does it reduce diagnostic accuracy? What is the real added value of scatter correction in clinical PET? While not being able to answer the question accurately, physicists are convinced that scatter correction is a vital component in the production of artefact-free, quantitative data.

Modelling the scatter component

Scattered photons arise from the whole attenuating medium, including the imaging table and the PET tomograph itself. In addtion to a decrease in the image contrast, events may also appear in regions of the image where there is no activity (e.g. outside the patient). The issue of scatter detection, modelling and correction in PET is addressed in many publications [1–9]. The ideal research tool (the 'gold standard') for scatter modelling and evaluation of scatter correction techniques is the Monte Carlo method [10]. Nevertheless, the complexity and computing requirements of Monte Carlo simulation led to the development of analytical simulation tools based on simplifying approximations to improve speed of operation. For instance, Beekman *et al*. [11] developed a fast analytical simulator of tomographic projection

*Address all correspondence to Dr Habib Zaidi, Division of Nuclear Medicine, Geneva University Hospital, CH-1211 Geneva, Switzerland. e-mail: habib.zaidi@hcuge.ch

data taking into account attenuation, distance-dependent detector response, and scatter based on an analytical point spread function (PSF) model. Several studies have reported that the scatter fraction defined as the ratio between the number of scattered photons and the total number of photons (scattered and unscattered) detected, represents from 30% (brain scanning) to 50% (whole-body scanning) of the data acquired in the 3-D mode, depending on the scanner geometry, the energy window setting, the region to be explored and patient size [12].

Much research and development has been concentrated on the scatter compensation required for quantitative 3-D PET. Increasingly, sophisticated scatter correction procedures are under investigation, particularly those based on accurate scatter models [6,7], and iterative reconstruction-based scatter compensation approaches [13–16]. Monte Carlo methods give further insight and might in themselves offer a possible correction procedure [5,9]. A number of scatter correction techniques have been proposed and successfully implemented for 3-D PET. The main difference among the correction methods is the way in which the scatter component in the selective energy window is estimated. The most reliable method for determining the actual amount of scatter in the image is physical modelling of the scatter process in order to resolve the observed energy spectrum into its primary (unscattered) and scatter components. By observing how accurately a scatter correction algorithm estimates the amount of scatter under conditions where it can be accurately measured or otherwise independently determined, it is possible to optimize scatter correction techniques.

Historically, once one had obtained the best projection data feasible, one typically applied compensations for these degradations either prior to or after reconstruction with filtered back-projection. Because of expanding diagnostic and therapeutic applications of quantitative PET, scatter correction is evolving from empirically based methods to patient-specific accurate models based on attenuation maps and the physics of interaction and detection of annihilation photons. Although the challenges for the latter algorithms are substantial, rigorous computational models have greater relevance for achieving the accuracy required in quantitative imaging. Currently, the preferred compensation strategy is the incorporation of modelling these degradations into an iterative reconstruction method. The inclusion of scatter estimation in iterative reconstruction is original in PET but was extensively investigated in single photon emission computed tomography (SPECT). Up to now, the feasibility of scatter estimation in PET was hampered by the heavy computational burden. New, faster implementations of scatter models should overcome this limitation. This trend is likely to continue, and these methods become routinely employed clinically [17].

The normal approach for implementation of a scatter model is to incorporate the scatter estimation directly in the transition matrix, although efficiency has been improved by utilizing a dual matrix approach in which scatter is incorporated in the forward projection step only [18]. In this case, the transition matrix is considerably larger than is necessary if only attenuation and geometric factors are included, and computation is therefore slow since scatter is essentially recalculated and added each iteration. In the case of constant precalculated (or measured) scatter (e.g. multi-energy window methods), this scatter estimate can either be subtracted directly from projections prior to reconstruction, or alternatively can be introduced in the denominator of the ML-EM equation. This latter approach results in better noise properties than direct subtraction [13]. The immunity to noise in emission data of statistical reconstruction-based scatter correction methods makes them particularly applicable to low-count emission studies.

Scatter correction schemes

Multiple energy window methods were originally developed for SPECT and have been in use for more than 15 years. The development of the 3-D acquisition mode in PET and improvements in the energy resolution of the detector have allowed the implementation of scatter correction based on the analysis of energy spectra [4]. Several groups investigated the potential of acquiring data in two, three and multiple energy windows to develop corrections for scattering in 3-D PET. The quantitative accuracy is generally improved at the expense of a degradation of signal-to-noise ratio, which can be explained by the scatter subtraction process and the resulting reduction in the statistics of the acquired data sets.

Model-based scatter correction methods use both emission and transmission scans together with the physics of Compton scattering to estimate the scatter distribution [6, 7]. However, as originally proposed, most of these methods do not correct for scatter from outside the field-of-view (FOV). This effect can be directly taken into account by acquiring short, auxiliary scans adjacent to the axial volume being investigated. These algorithms have thus been slightly modified to partially reduce this effect. Ollinger [6] takes the effect of muliple scattering into account by convolving the estimated single scatter component with a Gaussian function and approximates the contribution from out-of-FOV activity. Watson [17] reported on a new numerical implementation of the single-scatter simulation algorithm, which is faster than the previous implementation [7], currently requiring less than 30 s execution time per bed position for an adult thorax. The normalization problem was solved and multiple scatter partially taken into account. The contribution of scatter from outside the FOV remains a challenging issue that needs to be addressed carefully in whole-body imaging especially with large axial FOV 3-D PET scanners.

There is continuing interest in the development of non-stationary convolution–subtraction scatter correction techniques [8], which overcome the inherent limitations of the stationary approach [3] by taking into consideration the dependence of scatter upon source locations, object size, detector angle, etc. Different methods of non-stationary deconvolution have been proposed for SPECT [19] and 2-D PET imaging [8], the extension of such models for 3-D PET should, in principle, be straightforward.

Direct Monte Carlo-based scatter compensation approaches intrinsically handle the effect of multiple and out-of-FOV scatter contribution. It is worth pointing out that this approach may be repeated iteratively to reduce systematic errors introduced by the presence of scatter in the input images and the low statistics in the simulated data [9]. Levin *et al*. [5] proposed using coarse sampling of the input emission and segmented transmission images to reduce computation time. It is, however, recognized that a Monte Carlo-based scatter correction may not be practical for clinical routine applications with common computing facilities available in PET centres. However, powerful multiple-processor parallel processing systems are becoming more accessible to the scientific community, therefore investigation and characterization of such correction techniques and the effect of different approximations on their accuracy is worthwhile. It is hoped, however, that the boost in computing power will render their routine implementation feasible on ordinary desktop computers, obviating the need for sophisticated parallel computing technology.

Accuracy of scatter correction

Evaluation of scatter correction algorithms is inherently difficult and sometimes unconvincing. There is a clear need for guidelines concerning the evaluation of correction techniques and other image processing issues in PET. Most of the algorithms developed so far have been evaluated using either simulated or experimentally measured phantom studies, in addition to qualitative evaluation of clinical data [9]. Modelling and simulation of PET imaging is best done with phantom models that match the gross parameters of an individual patient. Recent three- and four-dimensional computer phantoms seek a compromise between ease of use, flexibility and accurate modelling of populations of patient anatomies, and attenuation and scatter properties and biodistributions of radiopharmaceuticals in the patients. Modelling of the PET imaging process has been improved by more accurate simulation of the physics and instrumentation involved in the process. Monte Carlo software packages, especially those developed specifically for nuclear medicine and with different performance characteristics, have been found useful in the modelling work. The combination of realistic computer phantoms and accurate models of the imaging process allows simulation of PET data that are ever closer to actual patient data. Simulation techniques will find an increasingly important role in the future of nuclear medicine research, especially scatter modelling and correction, in the light of future development of realistic computer phantoms, accurate modelling of projection data and computer hardware. However, precautions must be taken to avoid errors in the simulation process and verification via comparison with experimental and patient data is crucial [10].

In a clinical environment, the evaluation is further hampered by the multiplicity of the medical purposes for which the corrections may be studied. For any specific medical task, the evaluation should ideally be based on the performance of human observers. However, this is costly and complex, since a reasonable number of experienced observers should be used to analyse many images under carefully controlled conditions, etc. One severe limitation of performing psychophysical experiments for evaluation of image reconstruction techniques is that it is time consuming and costly. Furthermore, for optimization of reconstruction algorithms in which possible parameter settings suffer a combinatorial explosion, human psychophysical studies are simply not viable.

There is no single figure of merit that summarizes algorithm performance, since performance ultimately depends on the diagnostic task being performed. Well-established figures of merit known to have a large influence on many types of task performance are generally used to assess image quality [20]. With a few exceptions, most papers dealing with the evaluation of scatter correction techniques compare relative concentrations within different compartments of a given phantom with the background compartment serving as a reference. This approach possibly obscures what is actually going on, does not necessarily reflect the accuracy of the correction procedure and might bias the evaluation procedure. Therefore attempts should be made to put such results into absolute terms.

Conclusion

The major manufacturers of dedicated PET tomographs supply scatter correction software to end-users, whereas the dual-head gamma camera market is still suffering in this respect. However, it is expected that commercial software for accurate PET quantitation will be available shortly. The scatter correction issue in 3-D PET is an area of considerable research interest and many research groups are very active in this field, leading the PET community to forecast promising progress during the next few years.

Acknowledgements

This work was supported by the Swiss National Science Foundation under grant SNSF 3152-062008.

References

1. Links JM. Scattered photons as 'good counts gone bad;' are they reformable or should they be permanently removed from society? *J Nucl Med* 1995; **36:** 130–132.
2. Msaki P, Bentourkia M, Lecomte R. Scatter degradation and correction models for high-resolution PET. *J Nucl Med* 1996; **37:** 2047–2049.
3. Bailey DL, Meikle SR. A convultion–subtraction scatter correction method for 3D PET. *Phys Med Biol* 1994; **39:** 411–424.
4. Grootoonk S, Spinks TJ, Sashin D, Spyrou NM, Jones T. Correction for scatter in 3D brain PET using a dual energy window method. *Phys Med Biol* 1996; **41:** 2757–2774.
5. Levin CS, Dahlbom M, Hoffman EJ. A Monte Carlo correction for the effect of Compton scattering in 3-D PET brain imaging. *IEEE Trans Nucl Sci* 1995; **42:** 1181–1188.
6. Ollinger JM. Model-based scatter correction for fully 3D PET. *Phys Med Biol* 1996; **41:** 153–176.
7. Watson CC, Newport D, Casey ME, deKemp A, Beanlands RS, Schmand M. Evaluation of simulation-based scatter correction for 3-D PET cardiac imaging. *IEEE Trans Nucl Sci* 1997; **44:** 90–97.
8. Bentourkia M, Lecomte R. Energy dependence of nonstationary scatter subtraction–restoration in high resolution PET. *IEEE Trans Med Imag* 1999; **18:** 66–73.
9. Zaidi H. Comparative evaluation of scatter correction techniques in 3D positron emission tomography. *Eur J Nucl Med* 2000; **27:** 1813–1826.
10. Zaidi H. Relevance of accurate Monte Carlo modeling in nuclear medical imaging. *Med Phys* 1999; **26:** 574–608.
11. Beekman FJ, Viergever MA. Fast SPECT simulation including object shape dependent scatter. *IEEE Trans Med Imag* 1995; **14:** 271–282.
12. Zaidi H. Statistical reconstruction-based scatter correction: a new method for 3D PET. *Proc. of the 22nd Annual International Conference of the IEEE* 2000 **1:** 86–89.
13. Beekman FJ, Kamphuis C, Frey EC. Scatter correction methods in 3D iterative SPECT reconstruction: A simulation study. *Phys Med Biol* 1997; **42:** 1619–1631.
14. Kadrmas DJ, Frey EC, Karimi SS, Tsui BMW. Fast implementations of reconstruction-based scatter compensation in fully 3D SPECT image reconstruction. *Phys Med Biol* 1998; **43:** 857–874.
15. Hutton BF, Baccarne V. Efficient scatter modelling for incorporation in maximum likelihood reconstruction. *Eur J Nucl Med* 1998; **25:** 1658–1665.
16. Zaidi H. Reconstruction-based estimation of scatter component in positron emission tomography. *Ann Nucl Med Sci* 2001; **14:** 161–171
17. Watson CC. New, faster, image-based scatter correction for 3D PET. *IEEE Trans Nucl Sci* 2000; **47:** 1587–1594.
18. Kamphuis C, Beekman FJ, van Rijk PP, Viergever MA. Dual matrix ordered subsets reconstruction for accelerated 3D scatter compensation in single-photon emission tomography. *Eur J Nucl Med* 1998; **25:** 8–18.
19. Ljungberg M, Strand S-E. Scatter and attenuation correction in SPECT using density maps and Monte Carlo simulated scatter functions. *J Nucl Med* 1990; **31:** 1560–1567.
20. Furuie SS, Herman GT, Narayan TK, Kinahan PE, Karp JS, Lewitt RM, *et al*. A methodology for testing for statistically significant differences between fully 3D PET reconstruction algorithms. *Phys Med Biol* 1994; **39:** 341–354.