

Initially, we tried to take into account the estimation of spurious coincidences by adding it to the projection estimate in the denominator of the iteration step, which can be schematically written as

$$A^{n+1} = A^n \times \frac{1}{\tilde{c}I} \times \tilde{c} \frac{S}{cA^n + S_{sc}}, \text{Algorithm 1}$$

where A^n is the activity image estimate at step n , I the identity image, c the projection matrix, S the measured true-coincidence sinogram, and S_{sc} the estimation of the spurious-coincidence sinogram. The operation $+$, $-$ is performed ray-sum by ray-sum, and the operation \times is performed voxel by voxel. Although this method preserves the reconstructed voxel positivity in an elegant, natural way, we observed that Algorithm 1 no longer correctly converges when the estimated term S_{sc} becomes too large (data not published). In ^{86}Y PET imaging, this was especially the case for corpulent patients. This method is currently implemented in the Gemini TF PET system (Philips) for correction of scatter and random coincidences (4,5). Care should thus be taken when imaging low- ^{90}Y specific activity with this lutetium yttrium oxyorthosilicate-based system (6).

Finally, we decided to remove the negative pixels from the subtracted sinogram by transferring to them an appropriate number of counts from neighboring positive pixels (a detailed description of the method has been published (2)). The rationale of this strategy is that Poisson noise is characterized mainly by high-spatial-frequency positive-negative fluctuations. This transfer of counts was performed in a special way that avoids artifact generation in the reconstructed image. Phantom and patient studies showed that this method prevents bias in ^{86}Y PET imaging (2). The method could also be evaluated in ^{90}Y imaging with PET

systems, allowing separated prompt- and random-coincidence acquisitions such as the one used by Tapp et al. (1).

REFERENCES

1. Tapp KN, Lea WB, Johnson MS, Tann M, Fletcher JW, Hutchins GD. The impact of image reconstruction bias on PET/CT ^{90}Y dosimetry after radioembolization. *J Nucl Med.* 2014;55:1452–1458.
2. Walrand S, Jamar F, Mathieu I, et al. Quantitation in PET using isotopes emitting prompt single gammas: application to yttrium-86. *Eur J Nucl Med Mol Imaging.* 2003;30:354–361.
3. <http://atom.kaeri.re.kr/cgi-bin/decay?Y-86> EC. Korea Atomic Energy Research Institute website. Published 2000. Accessed January 12, 2015.
4. van Elmbt L, Vandenberghe S, Walrand S, Pauwels S, Jamar F. Comparison of yttrium-90 quantitative imaging by TOF and non-TOF PET in a phantom of liver selective internal radiotherapy. *Phys Med Biol.* 2011;56:6759–6777.
5. Wang W, Hu Z, Gualtieri EE, et al. Systematic and distributed time-of-flight list mode PET reconstruction. *IEEE Nucl Sci Symp Conf Rec.* 2006;3:1715–1722.
6. Walrand S, Jamar F, van Elmbt L, Lhommel R, Bekonde EB, Pauwels S. 4-step renal dosimetry dependent on cortex geometry applied to ^{90}Y peptide receptor radiotherapy: evaluation using a fillable kidney phantom imaged by ^{90}Y PET. *J Nucl Med.* 2010;51:1969–1973.

Stephan Walrand*

Michel Hesse

Lhommel Renaud

François Jamar

**Université Catholique de Louvain*

Av. Hippocrate 10

1200 Bruxelles, Belgium.

E-mail: stephan.walrand@uclouvain.be

Published online Jan. 29, 2015.
DOI: 10.2967/jnumed.114.152017

Erratum

In the article “In Vivo PET Imaging Demonstrates Diminished Microglial Activation After Fingolimod Treatment in an Animal Model of Multiple Sclerosis” by Airas et al. (*J Nucl Med.* 2015;56:305–310), the author line neglected to mention that Laura Airas and Alex M. Dickens contributed equally to the work. The authors regret the error.