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ATTENUATION CORRECTION EFFECTS ON SPECT/CT PROCEDURES: PHANTOMS STUDIES

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Abstract—**Attenuation correction is widely used in SPECT/CT (Single Photon Emission Computed Tomography) procedures, especially for imaging of the thorax region. Different compensationmethods have been developed and introduced into clinical practice. Most of them use attenuation maps obtained using transmission scanning systems. However, this gives extra dose of radiation to the patient. The purpose of this study was to identify when attenuation correction is really important during SPECT/CT procedures.For this purpose, we used Jaszczak phantom and phantom with three line sources, filled with technetium (^{99m}-Tc), with scattering materials, like air, water and acrylic, in different detectors configurations. In all imagesacquired were applied analytic and iterative reconstruction algorithms; the last one with or without attenuation correction. We analyzed parameters such as eccentricity, contrast and spatial resolution in the images.The best reconstruction algorithm on average was iterative, for images with 128x128 and 64x64 matrixes. The analytical algorithm was effective only to improve eccentricity in 64x64 matrix and matrix in contrast 128x128 with low statistics. Turning to the clinical routine examinations, on average, for 128x128 matrix and low statistics counting, the best algorithm was the iterative, without attenuation correction,improving in 150% the three parameters analyzed and, for the same matrix size, but with high statistical counting, iterative algorithm with attenuation correction was 25% better than that without correction. We can conclude that using the iterative algorithm with attenuation correction in the water, anditsextra dosegiven, is not justified for the procedures of low statistic counting, being relevant only if the intention is to prioritize contrastinacquisitions with high statistic counting.**

I. INTRODUCTION

Only recently diagnostic procedures in Nuclear Medicine have access to resources provided by the images generated by CT. Using similar techniques to those practiced in the procedures Positron Emission Tomography, PET / CT, it's possible now to determine more precisely, via CT images fusion, the anatomic location of the region with low or increased uptake revealed by CT Scans for SPECT. This feature, of relevant clinical interest, doesn't exhaust the significance of the images of transmission to the current Nuclear Medicine. These images are also valuable sources of information about the attenuation suffered by gamma rays coming from the decay of the radiotracer in its trajectory toward the camera scintillation detectors. With them it is possible to construct attenuation maps to be used to reduce artifacts resulting from radiation of heterogeneous regions of the patient's body. These maps allowto minimize the effects of attenuation on CT scans per emission, greatly improving the quality of clinical images [1, 2].

In clinical procedures, due to the variation of tissue thickness in different regions of the patient's anatomy, the magnitude of error introduced by the photons attenuation can also varies regionally in radionuclide imaging [3]. Thus, a lesion located deep within the body will produce a signal which is attenuated to a greater degree than that for a superficial lesion. Similarly, a region of tissue, with uniform radionuclide content, which lies below the tissue, with a variable thickness, will produce an image with variable density counting [4]. Reconstructions of CT images without attenuation correction can cause a false high density counts and reduce the image contrast in regions with low attenuation, such as the lungs.

In order to circumvent this problem, the technique of attenuation correction by applying maps correction can be used. These techniques consist to scale, with the best accuracy possible, the effects of attenuation to energies of 70-364 keV order. These energy values are used to obtain emission images in Nuclear Medicine from the data obtained in transmission of images generated by CT, with energies of about 80 keV. Several studies have been performed looking for the best correlation between the CT number and attenuation coefficient, resulting in the use of different linear relationships for estimating the attenuation by tissue and bone [5].

This study is about the real importance of attenuation correction in SPECT/CT procedures. Our data acquisition were based on clinical protocols used in imaging studies of bone and kidney, using the same parameters and algorithms of the clinical routine, in order to compare the structures visualized in phantoms with structures that doctors want to view during exams. Our purpose is not to study the efficiency of reconstructionalgorithms, but the clinical utility of using attenuation correction in conjunction with one of them in real situations rather than ideal conditions.

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A. Analytic reconstruction algorithm

The Filtered Backprojection (FBP) is the classical analytical method for emission (i.e., radionuclides) and transmission (i.e., x-ray) image reconstruction. With FBP, distribution radionuclide f(x, y) is reconstructed from data acquired projection g(s, θ) in two steps: (a) back projection, in which the profiles are designed in an matrix thatwill produce the CT slice image, and (b) the algorithm generates an artifact known as star artifact, which is removedby applying a specific filter (ramp filter) which, in turn, enhances the high frequencies.Therefore, it is necessary to apply a window filtering (Butterworth, Hanning, Parzen, among others). In clinical practice, most images reconstruction is done with analytical methods, because they are quick and simple to implement compared to iterative methods. The equation 1 describesthis algorithm:

$$f(x,y) = \int_k^p g(s_k, \theta_k) d\theta \qquad (1)$$

Where p is the number of projections acquired in π radians, θ_k is the *k-th* angular position, s_k is the location along the detector and $d\theta$ is the angular interval between two successive projections.

B. Iterative reconstruction algorithm

The iterative reconstruction methods include a step of back projection to estimate the concentration of radionuclide from the values contained in the data profiles. They also incorporate an iterative operation, which estimates the profiles and compares them to those originated from the tomography acquisition. This process occurs repeatedly until the difference between the average profile derived from the reconstruction and the actual profile is below the tolerance established by the operator.

Among the most popular iterative methods are the ML-EM (Maximum Likelihood-Expectation Maximization) and OSEM (Ordered Subsets Expectation Maximization) algorithm. The ML-EM goal is to calculate the distribution of the radioactive source that best reproduces the data projection [9]. This process takes into account that the projections can be modeled assuming that the photon count follows a Poisson distribution, taking into account the stochastic nature of the physical process of radiation generation. This algorithm consists of two phases: a Ephase which calculates the expected value of the conditional likelihood function, and the M phase, in which the expected value is maximized with respect to the estimated picture reconstructed previously, providing a new estimate. The ML-EM algorithm proposed by Lange and Carson (1984) [10] can be described by the equation 2:

$$\bar{f}_j^{(k+1)} = \frac{\bar{f}_j^k}{\sum_{i=1}^n a_{ij}} \sum_{i=1}^n \frac{g_i}{\sum_{i=1}^m a_{ij} \bar{f}_j^k} a_{ij} (2)$$

Where a_{ij} is the probability of detecting the photon emitted at pixel j at position I of the detector, f_j is the average number of disintegrations in the pixel j and g_i is the average number of photons detected at position I of the detector.

The algorithm OSEM processes data in subsets, within each iteration, in a way that accelerates the convergence by a factor proportional to the number of the subsets.

C. Attenuation Correction Methods

According to Hasegawa[4], methods of attenuation correction to emission tomography require reliable determination of an attenuation map, which represents the spatial distribution of the linear attenuation coefficients in patient's anatomical region and precisely defines the outline of structures in the body. Once the attenuation map is generated, it can then be incorporated in the reconstruction algorithm, correcting the emission data for errors due to scattering photon attenuation or other physical disturbance. The process of attenuation correction can be applied (a) before reconstruction, (b) after reconstruction (e.g., Chang algorithm) or (c) integrated within the matrix of transition iterative reconstruction algorithm. The iterative reconstruction algorithm uses the body structure information to calculate the attenuation limit which attenuating region for each pixel along the ray between SPECT emission and detection points. For each iteration the projection estimated values are calculated and compared to the projection measured values. The comparison process is used to generate a correction factor that is used to update the estimated concentration in the picture. This process iteratively improves the image precision estimated by modeling the photons attenuation that is present in the measured projection data.

II. MATERIALS AND METHODS

The realization of this work used the resources of the Nuclear Medicine Division of Clinical Hospital and Center for Biomedical Engineering of the University of Campinas. Besides the facilities, provision of radionuclides (^{99m}Tc) and clinic team support, it was used a calibrator dose, the scintillation camera SPECT / CT Siemens Symbia 2T, Jaszczak SPECT tomography simulator and various accessories, like a set of three linear sources to evaluates the tomography spatial resolution, software Image J and a set of cylinders and spheres, cold and hot, to determine the contrast.

To investigate the influence of attenuation correction it was used the following pattern: first, for each type of test, with the guidance of nuclear physicians, it was considered a typical set of patients and it was determined the activity pattern of radiotracer administered, the duration of the examination and the average counts per projection in SPECT acquisitions. Dose and duration of the procedure influence the acquisition of CT scans per emission as they have major influence on the signal/noise ratio of the projections and this, in turn, affects the reconstruction algorithm that used.

With references defined, we began the study with the simulator varying acquisition parameters, such as matrix (64x64 and 128x128), the arc of acquisition (180° and 360°), the statistical scores (high and low), the medium spreader (water, air and acrylic), the activity of the radiation source, the configuration of the detector ('L' or 'H', as in Fig. 1), the number of detectors acquiring images (one or two), their sense rotation (clockwise and counterclockwise), the angular sampling, time per projection (12s for high and 5s for low statistical) and zoom. Each configuration performed was used determine the spatial resolution, the eccentricity of the sources and hot and cold contrasts.

Fig. 1 Settings of detectors: in “H” (picture at the left) and in “L”(picture at the right).

All images were processed using three reconstruction algorithms: FBP, OSEM without attenuation correction and OSEM with attenuation correction (it was chosen protocol tests on bones). All acquisitions were performed using the technique TEW (Triple Energy Window) [11] for scattering correction and the usual protocol for CT image acquisition transmission. The line sources were numbered: 1, the source central; 2, upper; and 3, right (Fig. 2). All were filled with 0.3ml ^{99m}Tc solution, and the activities were registered.

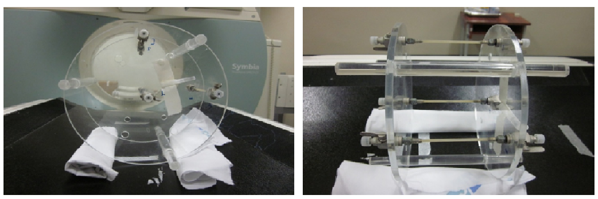

Fig. 2. Insert with 3 linear radiation sources, at the position of air acquisition.

After data acquisition, the images were processed at the Symbia’s processing station and the reconstruction algorithms were applied. The filter used with the FBP algorithm was the Butterworth and in the OSEM algorithm with and without attenuation correction, we used theGaussian filter.

The images were analyzed with the software Image J. By drawing ROI's around each source was possible to obtain the values of maximum and minimum scores relative to the background (BG) and their respective areas.

It was then possible to calculate the contrast given by equation 3. This parameter is important when you want to know whether the patient in this study has some pathological or not.

$$Contrast = \frac{countingsfromradiationsource / area\ fromradiationsource}{countingonBG / area\ onBG} \quad (3)$$

From the ROI's the major and the minor axes of the cross section of sources are obtained, enabling the calculation of the eccentricity by equation4. The closer this value is to unity, the more spherical is the source image. This parameter is important for providing the quantitative image distortion degree, due to its position relative to the center of rotation, the statistics of the acquisition and the sources height relative to the litter.

$$Eccentricity = \frac{minoraxe}{majoraxe} \quad (4)$$

Another value obtained with the ROI is the angle (θ) of the images of the sources relative to the Cartesian plane perpendicular to the system axis rotation. By drawing profiles with these θ angles and their complements (90 ° + θ), in the image of each source, we obtain graphs of these profiles, in which it is possible to calculate FWHM (Full Width at Half Maximum). This parameter is important for characterizing the performance of the equipment; it provides the value of spatial resolution.

III. Results

Given the acquisition matrix and scattering parameters were analyzed eccentricity, spatial resolution and contrast, after application of different reconstruction algorithms, from 60 images acquired in different settings. It was also determined the best configuration of detectors that reproduced the most satisfactory results for these parameters.

Analyzing the acquisition in 64x64 matrix, the best algorithm for obtaining a satisfactory eccentricity of sources was FBP, in 66.7% of the settings, with the detectors positioned at "H" by turning on an arc of 360 °. With the detectors in this same position and same arc, the OSEM algorithm without attenuation correction was more effective for good spatial resolution in 83% of the settings. This same algorithm is effective for obtaining improved contrast in 50% of configurations with detectors positioned at "L", turning arc of 360 ° around the litter and 50% for other positions of detectors other than "L". Percentages are shown in Table I.

Table I. Percentage ofparameters improvedby applyingthe algorithmsin64x64images.

Low statistics counting	**Eccentricity**	**Contrast**	**SpatialResolution**
FBP	66,7%	0%	0%
OSEM without correction	33,3%	100%	100%
High statistics counting			
FBP	66,7%	0%	33,3%
OSEM without correction	33,3%	100%	66,7%

For acquisitionsin 128x128 matrix, the OSEM algorithm with correction for attenuation was better in 83%eccentricity determination of experimental configurations, but the configuration of the detectors was modified according to the middle scattering. A good resolution was achieved in 83% of the settings using the OSEM algorithm without attenuation correction, with the detectors positioned at "H" and turning a 360° arc. The same algorithm without attenuation correction also provided better contrast in 66.7% of the settings for acquisitions with high statistical counting. But for 66.7% of the settings, for low counts acquisitions, the best contrast was obtained using the FBP algorithm. It is noted from Table II that the FBP algorithm was not efficient to improve any parameter in images with high statistical counting.

Table II. Percentage of parameters improved by applying the algorithms in 128 x128 images.

Low statistics counting	Eccentricity	Contrast	SpatialResolution
FBP	0%	66,7%	0%
OSEMwithoutcorrection	33,3%	33,3%	100%
OSEM withcorrection	66,7%	0%	0%
High statistics counting			
FBP	0%	0%	0%
OSEM withoutcorrection	0%	66,7%	66,7%
OSEM withcorrection	100%	33,3%	33,3%

The images in air and in acrylic were acquired only to obtain reference data. The scattering medium that best simulates biological tissue is water. In order to identify the best reconstruction algorithm for certain clinical protocols and whether it is justifiable to use the attenuation correction, the parameters were compared to previously analyzed images with high and low counting statistics (more common place in the clinical routine), matrix 128x128, with the simulator in air (reference) and water. Table III shows the percentages of improvement of the parameters when using OSEM algorithm with attenuation correction compared to OSEM without attenuation correction in the air.

Table III. OSEMwithattenuation correctioncompared toOSEMwithoutattenuation correctionin the air.

AIR	High statistics counting	Low statistics counting
Eccentricity	9,9%	17,7%
Contrast	-85%	-47,7%
SpatialResolution	-34%	-22%

The data in Table IV, with water show that there is a slight eccentricity improvement, a great improvement in contrast to high counting statistics, but spatial resolution worsens using this correction.

Table IV. OSEM with attenuation correction compared to OSEM without attenuation correction on the water.

WATER	High statistics counting	Low statistics counting
Eccentricity	8%	29%
Contrast	161%	-77,6%
SpatialResolution	-35%	-31,2%

IV. Discussion

The bestreconstruction algorithmwasOSEM,on average, for both matrix dimensions.TheFBPalgorithmwas effectiveonly to improveeccentricityin64x64matrixandcontrast in128x128matrix, with lowstatistics. Turning totheclinical routineexaminations, on average, for128x128matricesand lowstatistics,the bestwas theOSEMalgorithmwithoutattenuation correctionin the improvementof the three parametersanalyzed and, for the samearray size, but withhighstatistical,theOSEMwithattenuation correctionwas25%better,on average, thanwithout it.It was verifiedthat the use ofattenuation correctionin a medium,like air,littleattenuator,worsenedparameters, likecontrast andimage resolution.We can also saythat theuse ofOSEMwithattenuation correctionin the wateris not justified forthe acquisition oflowstatistic,beingrelevant only ifthe intentionisto prioritize the contrast in acquisitionswith highstatistic.There was only a slight improvement in the eccentricity due to the attenuation of the photons by litterand the other parameters worsened, which is consistent with the fact that the air doesn't attenuate very much such photons.

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