From the 60 second images 4 subreconstructions were performed with shorter times (10, 12, 15, 20 seconds) from which estimated 60 second noise image was created pixel-by-pixel via linear regression.
Results: The average of the SD/Mean image pixels varied between 0.1–0.25, moreover, it was visible on the PET image volume that this parameters correlates with the more noisy sections. The SD/Mean values calculated from the three series reconstructed with different times were consistently decresing with the acquisition time, and this tendency was correlating with the estimation from Poisson statistics. We examined the correlation of the Attenuation Correction Factors and the SD/Mean parameters for 12 patients, and the results slightly but clearly differed from those experienced via former phantom measurements. The image noise measured from the three 60 seconds image sets proposes correlation with the estimated noise image created via linear regression from the sub60 second image noise series. Therefore, the estimation of image noise from subrecostructions could be a possible method to resolve the problem.
Conclusions: This new method could be easily used for the determination of pixel noise for PET scans, therefore, we propose to use it for the optimization of PET acquisition protocols.

041

ACCREDITATION QUALITY CONTROL PERFORMANCES OF DIFFERENT PET SCANNERS

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Background: It is a critical point to assure similar image qualities for PET scanners in multicenter diagnostical studies including several PET-CT centers. For this purpose, the multicenter studies usually define the minimal criteria in terms of image quality for the PET-CT systems that can be included in the study. In the last few years accreditation protocols of two major international associations are also available. However, these two associations, the European Association of Nuclear Medicine (EANM) and the American College of Radiology (ACR) propose to use different phantom scans and image capability criteria for the PET scanners. The ACR recommends a specially designed ACR phantom, and the specific parameters for the quantitative measurement are the SUV Mean, SUV maximum and standard deviation for the region of interests. The EANM accreditation uses the NEMA 2007 IQ phantom, and the protocol includes the calculation of the Activity Recovery Coefficients (ARC) obtained from the intensities in the six spheres of the phantom. Both of the two accreditations limits for the calculated parameters and delivers the PET scanner acceptable or not. The aim of this work was to investigate different PET scanners from several vendors, and their acceptance by the two accreditation protocols mentioned above.
Material and methods: Our investigation included the following PET-CT scanners: Siemens TruePoint HD (Budapest), Siemens Biograph 16 (Nagyvárad), Philips Gemini 64 TF (Debrecen) and Siemens mCT (Doha/Qatar). The EANM protocol was performed on all of the four scanners and the ACR protocol was performed on the Philips Gemini and the Siemens mCT scanners. The EANM PET accreditation protocol gives minimum and maximum criteria for the Activity Recovery Coefficients, while the ACR protocol gives them to the calculated SUV max values. These parameters have been investigated by the "eanm_qc_tools_v15082011" program and with algorithms implemented in Matlab. It has been also investigated in this work, how the different acquisition and reconstruction parameters affect the calculated values of interest mentioned above.
Results: Only the Siemens Biograph 16 system fulfils the EANM PET accreditation requirement using the default clinical acquisition and reconstruction protocol. The other three PET cameras "overfulfil", that is, most of ARC values were higher than the prescribed maximums. In the case of the Siemens TruePoint HD and Philips 64 TF systems the ARC values shifted into the required interval if we changed the reconstruction or acquisition settings from the clinical defaults. For example at the Siemens system we had to turn off the point spread function modelling (TrueX) option during the reconstruction and to apply a Gaussian postfilter (with 4 mm window) on the images. Nevertheless, these modifications could be large impact on the resulted image quality. The all ARC data of the mCT PET camera were definitely higher that the required maximal values independently on the acquisition and reconstruction settings. Considering the ACR phantom measurement, the calculated SUV maximum values were in the required range for the Philips 64 TF camera but could not be fulfilled the ACR accreditation criteria in the case of the mCT system beside any reconstruction and/or acquisition settings.
Conclusions: Though the EANM or ACR accreditation based phantom measurements can be carry out with easily and clearly, the predefined criterions and the related ARC and SUV parameter intervals can be only satisfied if the users impair the default reconstruction and/or acquisition protocol settings used in the clinical routine. The latest (the state of the art) system was the Siemens mCT scanner in this study, nevertheless the results obtained with this camera could not fulfil at all the requirements of the accreditations.

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OPTIMAL PARAMETER SETTINGS OF 3D PARALLEL PROJECTION BASED SPECT RECONSTRUCTION PROCEDURE FOR CLINICAL APPLICATIONS

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Background: Parallel projection based Single Photon Emission Computer Tomography (SPECT) imaging is the most widely used procedure till nowadays, having several limitations in image quality. The following components have important rules in image quality such as: the contradiction between the resolution and sensitivity, the non-linear distant dependent spatial resolution (DDSR) and most of the case the non-uniform attenuation media around the imaged objects. Essentially, we have worked out multi-modality (SPECT/CT) based 3D iterative (MLEM method) reconstruction procedure for parallel projection based human and small animal imaging. Main goal of our current research/developing work is to find out those parameterization ways (including acquisition parameterization too) for the clinical/biological applications and various imaging systems (general purpose and/or dedicated systems) where the acquisition time/processing time/obtained image quality jointly will be significantly better than the traditional 2D method depending on the clinical conditions.
Material and methods: DDSR describing by point spread function (PSF) has determinant effect on the model of parallel projection. PSF is determined by a dedicated calibration procedure. It is necessary to acquire point spread functions at predefined distances from the detector surfaces for all the collimators and isotopes combinations to be used in the imagings. Then will be derived the inherent forward projection operator describing the distant dependent compensation

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