enerate images including the activity filled rods in background activity sing NU4IQ phantom.

Material and methods: The determination of CRC was measured on ne MiniPET-II and a Siemens Inveon PET scanner utilizing the MHS. his phantom contain four fillable spheres with different radius (1.95 mm, .47 mm, 3.1 mm and 3.9 mm) embedded in a cylindrical chamber diameter: 40 mm, height: 82 mm). The measurements were performed t three different object contrast (OC) ratio (2, 4 and 8). The determination of CRC was performed using the BrainMOD software.

he NU4IQ is inappropriate for the determination of CRC. However, we efined a "t" length list-mode acquisition using the filled NU4IQ, setting wo phantom positions. During the first phantom position scan the uniorm region of NU4IQ is centered in the FOV and scanned for "t0", after ne rods contained phantom region is moved to the scanner center with tr" scan time (t0 + tr = t). Thus, this acquisition method "manually" ums up the uniform and the rod parts of the NU4IQ phantom. After ne measurement the CRC can be easily calculated in same way as in ne human image quality calculation. This measurement was performed n MiniPET-II, Siemens Inveon, GE Explore Vista and Genesis4 small nimal PET systems.

Results: We found that the contrast recovery depends on the OC and ne size of the spheres. Increasing the size of the spheres the CRC mproves. In the case of MiniPET-II the minimum and maximum value f CRC (at OC = 8) are 0.3 and 0.6, respectively. These parameters at iemens Inveon were 0.34 and 0.65, respectively. The CRC parameters were determined width the NU4IQ and the following sequence vas found between the scanners: Siemens Inveon, GE Explore Vista, MiniPET-II and Genesis4.

Conclusion: It can be concluded that the MHS phantom is a practical ool to determine the CRC parameters of small animal PET systems. urthermore, we found that our special measurement protocol with the IU4IQ is a useful method for CRC determination.

039

AUTOMATED REFERENCE REGION DELINEATION OR QUANTITATIVE FDG-PET RAT BRAIN STUDIES

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ackground: The possibility of the quantitative small animal PET study pplications is restricted by the arterial blood-sampling since special ols, reproducible measuring protocol and validated software are eeded. In the absence of these means, methods based on reference gion can be applied, which require more complex data processing id the uncertainty of reference region delineation can influence the lculations. Dynamic rat brain 18F-FDG studies were carried out by e MiniPET-II PET scanner for quantitative analysis in our department. ie "whiskers" areas were used as reference regions.

ur goal was to develop a model-independent automated VOI deeating algorithm, which allows to select reference areas based on namic image series.

aterial and methods: From six rat 18F-FDG studies 60 × 1 min namic images and two integrated static images (first 10 and last minutes) were used for the development of automated VOI delinea-n. The left and right reference region and 3 cerebral regions (left and ht hemisphere and cerebellum) were delineated on static images by nm diameter spherical VOIs.

constructed a model curve from the average of the 12 tissue ves belonging to the reference regions for generating wavelet-coration maps from the dynamic images. Cluster analysis was used to emphasize the highly correlated connected areas on the frontal part of the skull from the correlation map.

Since the identification of the objects in PET images is difficult, the goodness of the automated method was characterized by the changing of noise characteristics, and the trends of the curves in the reference regions. The noise was characterized by the sum of squared differences between the curve and trend curve, the latter was calculated by the wavelet-based multi resolution analysis. The changing in trend curve was defined by the ratio of the average from the first 5 minutes and the average from the last 30 minutes.

Results and conclusion: The automatically delineated reference regions were close to the manually selected regions in each case, as checked by image fusion technique. The noise characteristics of the tissue curves in these regions were better in 5 cases compared to manually selected curves, the relative error was 0.85, the deviation 0.19, and was worse only in one case. There were no relevant differences in the changing of trend curves (average: 1.04). The results prove that the automated method can produce similar curves with better noise characteristics compared to manually delineated curves. One of the possible directions of the development is the combination of the introduced curve characteristics and region growing technique, which can be used in effective "blood curve"-like voxel-set delineation. Additional result is the delineation of the presumably arterial areas, which were basically different from the original reference regions. The application of these areas in kinetic modeling must be investigated in the future.

040

NOISE ANALYSIS OF WHOLE BODY FDG PET IMAGES

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Background: Instead of using continous bed motion during CT examinations, Whole Body PET scans are typically acquired in discrete axial positions. For each bed position the length of the acquisition time conventionally remains the same, however the total acquisition of the scan is usually adjusted to the patients weight or Body-mass-index. Because of the varying amount of attenuation in different sections of the body and the heterogenous activity distributions it is expected that the signal-to-noise values will vary accordingly in each bed position. However, for medical reports it would be more convenient to have relatively constant signal-to-noise through the axial slices, therefore, it would be important to know how could the clinical PET images be described with only one signal-to-noise parameter. The main challange is to estimate the noise (pixel variance), since for this purpose a PET scan should be repeated several times and that is hardly possible to carry out in the clinical routine.

Material and methods: To resolve the problem, we defined a new algorithm that assigns a Standard Deviation/Mean value to every voxel of the reconstructed PET image. These values were obtained from the list mode file of three-minute whole body PET acquisitions, while 12 × 15 second, 6 × 30 second and 3 × 60 second segments of the list mode file were reconstructed into identical image series. This resulted in three separate possible estimation of the image noise. From the identical image series we computed pixel-wise the Mean and Standard Deviation values, resulting in a Mean and a Standard Deviation image volume. As a pixel-by-pixel ratio of these two a Standard Deviation/Mean (SD/Mean) image was created and for each slice a mean value was calculated (as the Noise Parameter) from the pixels inside the body contour.

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.

042

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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