New techniques in imaging reconstruction: SPECT and SPECT/CT

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Single-Photon Emission Computer Tomography (SPECT) is a method that from a set of 2D planar scintillation camera measurements reconstruct 3D tomographic images where each image represents a section in the patients. Thus, the images are still in 2D but the section in the patient for which the image represents is known. The advantage if SPECT is a much better image contrast because of the absence of overlapping activity contribution but also greater possibilities for performing accurate corrections of physical effects such as photon attenuation, scatter contribution and distance dependent collimator resolution (1).

Image reconstruction (2) is required to obtained tomographic images from a set of SPECT projections. The most common method up to now has been the filtered back-projection method (FBP). This method is fast and easy to use. However, due to the, in general, noisy data from scintillation cameras some image processing is needed which is often made by applying a low-pass filter. A commonly used filter is the Butterworth filter for which the magnitude of the filtering can be described by its cut-off frequency and order. FBP can also cause some streak artefacts due to incomplete angular sampling and correction for attenuation, scatter and collimator response needs to be made outside the reconstruction.

A family of reconstruction methods that have gained increasing interest is the iterative methods and especially the MLEM/OSEM methods. These methods are based upon a first estimate of a reconstructed image and a comparative step where an estimated projection calculated from the knowledge of how a SPECT system work is compared to a measured projection. An error projection is calculated bin by bin and the results are backprojected to form an error image. After compared data from all projection angles the initial guess is updated by multiplying the first estimate by the error image. This updated image can then be used to calculated new error images. These iterations continue until the error between estimated and measured are acceptable low. The Maximum Likelihood Expectation Maximization (MLEM) is the most common method for SPECT and PET since its derivation based on the Poisson statistics. The derivation may seem complicated but ends up with a relative simple expression.

MLEM has a slow convergence and this has hampered the clinical implementation of the method. However, in 1994 Hudson and Larkin (3) published the method OSEM (Ordered Subsets Expectation Maximization) that is many ways are similar to MLEM but with the important difference that only a small number of selected angles are used to calculated error image before the update. For example, if 64 angles have been acquired then MLEM use data from all 64 projections to create the error image but OSEM only us a subset (typically four) before updating the image. This means that the initial image will be updated 16 times during each iteration as compare to only one update for the MLEM. The results with therefore converge much faster and the process has thus been accelerated.

It is assumed that the final image is an accurate estimate of the radionuclide distribution since its related projection matches the measured projections. However, this assumption is only valid if the model of the camera and the acquisition process represent the real measurements. If the camera model do not include attenuation effect, scatter contribution and the effect of collimator resolution then the result from reconstruction method will converge toward a wrong image because the way estimated projection is created do not match the way measured projections are created.

It has been generally known that a knowledge of the anatomical information helps very much in establish the location of, for example a tumour. This anatomical information has shown to increase the specificity. Previously, image registration was conducted using CT information obtained from a dedicated CT as a separate study. There was then a need for the images to be transferred and registered to match an acquired SPECT images. The difficulties were here how to make good image registrations and interpolations since the patient in fact had been moved between the two studies. This has now been solved to a high degree in modern SPECT systems in that a CT is combined in the same gantry. Thus, many of the problems with image registration can now been solved by a proper hardware and camera design and the accuracy in the registered images is thereby higher mainly also because the patient do not need to move between the studies but instead remain on the couch.

The available CT information has also lead to a much better and consistent attenuation correction as compared to earlier method with transmission radionuclide sources. Because one usually wanted to do both transmission and emission at the same time transmission measurements with radionuclides lead to problems with down-scatter events from photons emitted from the administered radionuclide that contributes to the transmission data.

This problem affected the accuracy in the determined attenuation coefficients. Furthermore, the noise levels and the spatial resolution of the transmission images with radionuclides are much lower than then corresponding CT information. The first generation SPECT/CT system was equipped with a single slice low-dose CT. For 40 slices the acquisition took about 10 minutes. The spatial resolution was relatively poor and in the order of 3-4 mm which does not make the images diagnostically useful. The second generation SPECT/CT now have diagnostic spiral CT with much better resolution and also the acquisition time is greatly reduced. The argument to include these high-quality CT in SPECT system is also that the system can be used for stand-alone CT studies. It should be remembered that when fusing SPECT images with CT images the spatial resolution is mainly determined by the SPECT camera. Furthermore, the acquisition of the SPECT data is based on a continuous mode that includes patient movements due to respiration. Care must therefore be taken when CT data taken in a short time/slice acquisition because these images may not reflect the average breathing. This could influence the attenuation correction and probably also the interpretation of s fused set of images.

The conversion of Hounsfield numbers from the CT images to attenuation coefficients for a particular radionuclide is also not trivial since the HU number originates from a spectrum of bremsstrahlung photons. The differences in spatial resolution between CT and SPECT may also cause artefacts in the attenuation correction mainly at locations close to boundaries of different densities. Therefore, sometimes the CT data needs to be blurred by a Gaussian function that results in a more comparable spatial resolution between the two modalities.

This lecture will describe and discuss both filtered back-projection and iterative reconstruction and show examples of the pros and cons with these two methods. SPECT/CT systems have been very successful in both diagnostic nuclear medicine studies as well as a very important to for improved and accurate dosimetry for radionuclide therapy planning. In this lecture examples of activity quantitation will also be given for 3D based dosimetry protocol where the CT unit is used for attenuation correction but also as a device to generate 2D transmission images for activity quantitation based on the conjugated-view method.

References

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