Empirical Multiple Energy Calibration (EMEC) for Material-Selective CT

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Purpose:

Traditional energy integrating (EI) x–ray CT detectors integrate the incident photon energy and provide the result as signal value. Recently, the development of a new detector technology, which counts the number of incident photons showed promising advancements. These photon counting (PC) detectors may additionally allow assigning each incident photon to a distinct energy range. The availability of such photon counting energy–selective (PCES, figure 1) detectors offers new possibilities to the field of material–selective CT, which has been restricted to dual energy CT in most cases, so far. For example, the adoption of contrast agents with k-edges in the range of a typical clinical x–ray spectrum (figure 1) becomes interesting, because the multiple energy measurement allows to automatically calculate images of the exact densities of many contrast agents. This work extends the methods described in reference [1] for single energy CT and in reference [3] for dual energy CT towards multiple energy CT, where the number B of detected spectra and the number M of basis materials is greater than two (figure 3). The advantage of the empirical methods in references [1–3] and in this work is that no knowledge of the spectral properties of the separated materials and of the detected spectra is required.

Materials and Methods:

We assume that the object is a linear combination of M linearly independent materials and that this object is measured with B different detected spectra (figure 3). From the intersection lengths p_m and the mass attenuation coefficient ψ_m of each material m as well as the detected spectrum w_b of each energy bin b, the polychromatic attenuation q_b is calculated. Note that the mass attenuation coefficient and the detected spectra are used here for simulations of the measurement process, but not for the reconstruction. For the reconstruction the non-linear relationship of q_b and p_m must be inverted such that we can calculate p_m when we have measured all q_b. The basic idea of the empirical method proposed in this work, as well as in references [1–3], is to model p_m as an M-dimensional series expansion (figure 5). The coefficient vector c of the expansion is calculated by minimization of the cost function D^2 (figure 6), as proposed in ref. [3]. This is done via one calibration measurement, which is run with the same detected spectra as the subsequent (or preceding) patient measurement. In the case of more detected spectra than linearly independent materials (B>M), there exist several different ways to define an M-dimensional series expansion (figure 7). This redundancy may be used for dose optimizations [4].

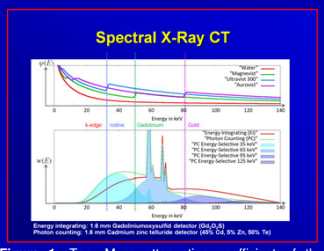

Figure 1: Top: Mass attenuation coefficient of the materials considered throughout this work. Bottom: Detected spectra used for simulations.

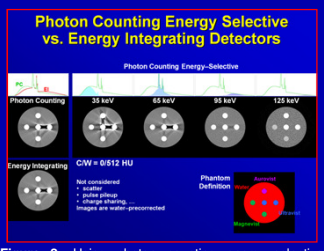

Figure 2: Using photon counting energy–selective detectors, one measurement yields several CT datasets that vary in spectral and noise properties.

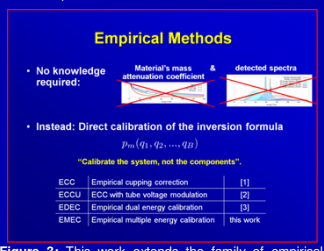

Figure 3: This work extends the family of empirical methods proposed earlier towards multiple spectra and multiple materials. The methods have in common that they do not require knowlege of spectral information.

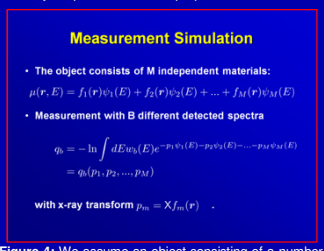

Figure 4: We assume an object consisting of a number of independent materials. Two materials are motivated by photo–effect and Compton effect. The remaining materials need the k-edges to be linearly independent.

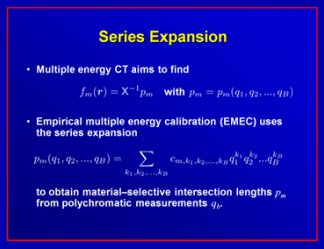

Figure 5: For reconstruction the inverse of the polychromatic measurement must be calculated. Here, the inverse is modeled as series expansion where the coefficients are determined from a calibration scan.

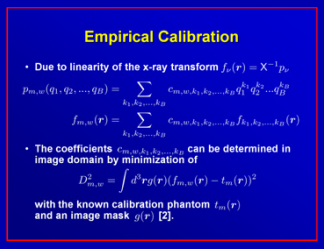

Figure 6: The calibration requires one measurement of a known calibration phantom, which contains the desired materials, with the desired detected spectrum.

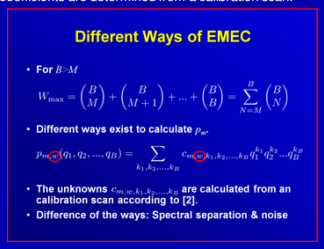

Figure 7: If the number of detected spectra exceeds the number of independent materials, there are different ways possible to perform the calibration. D^2 is a measure of the calibration quality of each way.

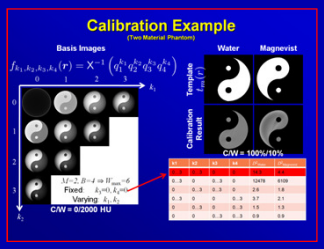

Figure 8: Visualization of the calibration of a two material reconstruction using four detector bins. There are six ways to use two out of the four bins. The (considerably varying) calibration quality D^2 is specified.

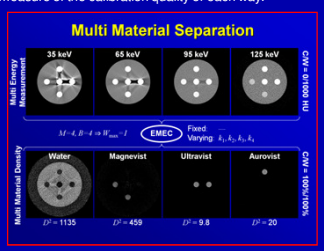

Figure 9: Results of a four material separation using four detected spectra. As usual in material–selective CT, high voxel noise appears in material–selective images. Beside this, the quality of the material separation is very high.

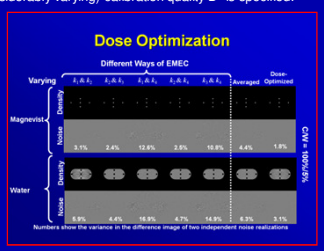

Figure 10: If redundancies exist in the system because more than one way of EMEC is possible, the additional ways can be used to optimize the dose usage. This method is presented in reference [4].

Results:

Figure 8 shows internal information on the calibration procedure of a M=2, B=4 noise-free simulation of the Yin Yang phantom. The ten basis images, which contribute to the minimization when using this particular EMEC way, are shown on the left side. Five other ways to perform EMEC exist in this case. The cost function D^2 (values shown in the table) is a measure of the calibration quality for each way.

In figure 9 we show the reconstruction result of the phantom (M=4) defined in figure 2. The spectral information in figure 1 was used for simulation of the measurement process (B=4), but not for the reconstruction. The reconstruction results show very good separation of the basis materials. An increase in image noise, which mainly depends on the spectral separation of the imaged materials and the detected spectra, is usually observed when performing material–selective CT. Since we have M=B, there are no redundancies in the configuration of figure 9. In figure 10, the CTDI phantom was simulated with M=2 (water and gadolinium), and B=4. This configuration allows six ways to perform EDEC and the result of each material and each way is shown in figure 10. The figure further shows that simply averaging the results of each way results in increased voxel noise because the noise of the way's results is correlated. In order to optimize the dose usage the contribution of each source to the final image's sinogram has to be considered on an object- and detector pixel–dependent method, as discussed in reference [4].

Conclusion:

A straightforward extension of the proven and established ECC and EDEC methods to multiple energy CT and multiple material separation has been proposed and successfully applied in a simulation study.

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