Modeling channel-wise response of a multi-bin detector for photon-counting spectral CT and its applications

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Aims and objectives

Multibin photon-counting x-ray detectors have shown a great potential to be applied in next generation computed tomography (CT). Investigations have shown their substantial advantages in solving the current limitations of conventional charge-integrating detectors. The availability of photon-counting detector makes it possible to generate images in multiple energy bins with a single kVp acquisition[1]. Moreover, multibin detectors open up the possibility for decomposition techniques such as K-edge imaging, which can identify the contrast agents, e.g. gadolinium and iodine, and their densities[2]. Other commonly known advantages of photon-counting detectors are: the improvement in contrast-to-noise ratio by eliminating electronic noise and using optimal energy weighting, and the reduction of the radiation dose while maintaining the image quality comparable to the energy-integrating system in various imaging tasks[3].

However, variations among detector channels in CT very sensitively lead to ring artefacts in the reconstructed images[4]. For material decomposition in the projection domain, the variations can result in intolerable biases in the material line integral estimates[5]. A channel-wise model of the detector is developed and the prediction results are used as inputs for compensation methods to mitigate the effect of pixel-to-pixel variations.

The model is achieved by plugging individual parameters into a well-defined forward model for detector channels[6]. Parameters involved in the forward model include the incident x-ray spectrum, photon flux, absorption efficiency of the detector and the bin sensitivity function where the energy bin thresholds play a role. Since the location of the energy bin thresholds tends to vary among detector channels, an energy calibration method has to be applied to determine the value of energy threshold for each detector channel and each energy bin. A method employing a fitting procedure between the measured and simulated spectra under radiation of a broad x-ray spectrum is used, which has been described in detail in our previous work and shown adequate accuracy[7]. Predictions of the channel-wise model will be applied in two cases: first, to eliminate ring artefacts in CT images with the compensation method described in reference 4; second, to achieve projection-based material decomposition using the maximum likelihood approach described in reference 2.

Methods and materials

Detector description

A segmented silicon-strip detector for photon-counting spectral CT has been developed by our group[8][9]. An image of the detector module is shown in Fig.1, which comprises in
total 450 p-type electrodes that are implanted on n-type silicon substrate with a thickness
of 0.5 mm. The 450 electrodes are arranged as 50 strips each subdivided into 9 depth
segments. Aligned in edge-on geometry, each strip forms a detector pixel with a pixel size
of 0.4 mm × 0.5 mm and a detection depth of 30 mm. The 9 depth segments are read out
individually to cope with the high x-ray fluxes present in clinical CT, with segment lengths
varied such that approximately equal count rates are expected along the detection depth.
Three ASICs provide individual readout channels for each of the detector elements.
Each channel includes a charge-sensitive amplifier, a pulse-shaping unit and a set of 8
comparators to perform pulse height discrimination on the shaped pulse. The amplitude
of the resulting shaped pulse is made to be proportional to the energy deposition
of the inducing photon interaction. By distributing the thresholds across the dynamic range
of pulse amplitudes, pulse-height intervals are formed, each connected to an individual
counter. When a pulse exceeds the lowest threshold a pulse detection period is triggered.
During this period, the highest threshold exceeded by the pulse leads to an increment
of the corresponding counter.

Energy calibration

Since the comparator thresholds have to be set voltage in units of mV, we need to find
out a relationship which can be used to translate the user-set comparator thresholds into
energy thresholds in units of keV that can be used in the forward model. Thus an energy
calibration method, which makes use of the broad x-ray spectrum provided by commercial
x-ray tubes, is developed. Energy calibration parameters for each detector channel
are obtained by a regression analysis that adjusts a simulated spectrum of deposited
energies to a measured pulse-height spectrum. The measured spectrum is obtained by
scanning a given comparator threshold across the range of detectable pulse amplitudes.
At each scanned threshold position, expressed in voltage value, the total number of
counts above threshold is accumulated during a fixed measurement time, resulting in an
integral spectrum in units of mV. The simulated spectrum is generated with the aid of a
detailed Monte Carlo simulation that takes into account the basic characteristics of the
detector modules. Fig. 2 illustrates the fitting results for segments in the central strip of the
detector. By plugging the calibration parameters in the forward model of each detector
channel, count response of the detector can be predicted for any incident spectrum and
a compensation method based on the prediction results can be developed to eliminate
the non-uniform response of the detector.

Data acquisition

A table-top setup with one detector module was used for data acquisition. Fig. 3 shows
a schematic sketch of the measurement setup. An image of the phantom, as shown in
Fig. 4, is took with the table-top setup by mounting a cylindric phantom on the rotation
and translation stage. The phantom was made of PMMA, 90 mm in diameter, hosting
5 cylindrical inserts where polystyrene containers filled with different substances can be
placed in. In this study, the five containers were filled with water, rapeseed oil, packed gypsum powder (containing calcium), iodine contrast agent (53 mg I/ml, or 0.42 mmol l/ml diluted Visipaque) and gadolinium contrast agent (79 mg/ml, 0.5 mmol/ml, Dotarem), as labeled in Fig. 4.

Images for this section:

Fig. 1: Detector module fro photon-counting spectral CT. Four hundred and fifty detector elements are arranged in 50 strips, where each strip id subdivided into 9 depth segments. X-rays enter from the top in the image. Three ASICs provide individual readout channels for each detector element.
**Fig. 2:** Measured integral spectra and the corresponding fitted models for depth segments in the central strip of the detector module.

**Fig. 3:** Schematic sketch of the measurement setup. The detector was mounted in a light-tight box with an x-ray transparent opening. The filtered x-ray beam was collimated and hits the detector strips edge-on.
**Fig. 4:** Photography of the phantom consisting of a PMMA cylinder and five cylindrical inserts where polystyrene containers filled with different substances can be placed in. The diameter of the phantom is 90 mm, the polystyrene containers have an outer diameter of 22 mm and an inner diameters of 18 mm. Substances in the containers are water, rapeseed oil, packed gypsum powder (contains calcium), iodine contrast agent (53mg/ml) and gadolinium contrast agent (79mg/ml).
Results

Eliminating ring artefacts

In Fig. 5 the reconstruction results of the projection data are shown, where 5(a) are the images reconstructed from the projection data compensated by flat-fielding using an air scan, 5(b) are the images reconstructed from the projection data compensated by the compensation method with predictions of the channel-wise model as inputs. Images in the right column are close-up views of the central parts of the images in the left column. Eight reconstructed images of eight energy bins are evenly summed together to generate the final flat-weighted images. All images are normalised to Hounsfield units measured from the circular ROI placed in the water insert. From the images, we can see that ring artefacts are quite strong in the images compensated by flat-fielding and appear negligible in the images corrected by the compensation method.

Material decomposition

The results of basis material decomposition obtained by applying maximum likelihood approach with the aid of forward model are shown in Figure 6. As it can be seen in the images, the decomposition into the four basis functions separated the individual materials. In the PMMA image, the PMMA phantom structure is visible. The calcium insert is most prominent in the Al image and two contrast agents are separately shown in individual images with quite identical features. The iodine and gadolinium concentrations were measured by averaging over ROIs within the corresponding inserts using the images shown in Figure 6. Error bars were derived from the pixel standard deviations within these ROIs. The quantification results are summarised in Table 1, showing good agreement between the measured values and the nominal concentrations.

Table 1. Nominal and measured contrast agent concentrations of the iodine and gadolinium inserts in the phantom shown in Fig. 4.

<table>
<thead>
<tr>
<th></th>
<th>Iodine (mg/ml)</th>
<th>Gadolinium</th>
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<tbody>
<tr>
<td>Nominal</td>
<td>53</td>
<td>79</td>
</tr>
<tr>
<td>Measured</td>
<td>50±12</td>
<td>80±16</td>
</tr>
</tbody>
</table>

Images for this section:
**Fig. 5:** Flat-weighted images of the cylindrical PMMA phantom with 5 different inserts, which are labeled in the images as: 1. water, 2 calcium, 3 gadolinium, 4 iodine, 5 oil. The images in the right column are close-up views of the central parts of the images in the left column. (a) is the flat-weighted image reconstructed from projection data compensated by flat-fielding using air scan. (b) is the flat-weighted image reconstructed from projection data corrected by compensation method with predictions from the channel-wise model as inputs.

**Fig. 6:** Basis material images for PMMA, Al, iodine and gadolinium corresponding to the image in Fig.4. (a) PMMA image. (b) Al image. (c) Iodine image. (d) Gadolinium image.
Conclusion

In an experimental validation, image data reconstructed by projection data corrected by compensation method with predictions from the channel-wise model as inputs show significant improvement with respect to ring artefacts compared to images calibrated with flat-fielding data. Projection based material decomposition gives basis material images showing good separation among individual materials and good quantification of iodine and gadolinium contrast agents. The work indicates that the channel-wise model can be used for quantitative CT with this detector.

Personal information


References


