The z-sbDBA, a new concept for a dynamic sheet-based fluence field modulator in x-ray CT

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Purpose: We present a new concept for dynamic fluence field modulation (FFM) in x-ray computed tomography (CT). The so-called z-aligned sheet-based dynamic beam attenuator (z-sbDBA) is developed to dynamically compensate variations in patient attenuation across the fan beam and the projection angle. The goal is to enhance image quality and to reduce patient radiation dose.

Methods: The z-sbDBA consists of an array of attenuation sheets aligned along the z direction. In neutral position, the array is focused toward the focal spot. Tilting the z-sbDBA defocuses the sheets, thus reducing the transmission for larger fan beam angles. The structure of the z-sbDBA significantly differs from the previous sheet-based dynamic beam attenuator (sbDBA) in two features: (a) The sheets of the z-sbDBA are aligned parallel to the detector rows, and (b) the height of the sheets increases from the center toward larger fan beam angles. We built a motor actuated prototype of the z-sbDBA integrated into a clinical CT scanner. In experiments, we investigated its feasibility for FFM. We compared the z-sbDBA to common CT bowtie filters in terms of the spectral dependency of the transmission and possible image variance distribution in reconstructed phantom images. Additionally, the potential radiation dose saving using z-sbDBA for region-of-interest (ROI) imaging was studied.

Results: Our experimental results confirm that the z-sbDBA can realize variable transmission profiles of the radiation fluence by only small tilts. Compared to the sbDBA, the z-sbDBA can mitigate some practical and mechanical issues. In comparison to bowtie filters, the spectral dependency is considerably reduced when using the z-sbDBA. Likewise, more homogeneous image variance distributions can be attained in reconstructed phantom images. The z-sbDBA allows controlling the spatial image variance distribution which makes it suitable for ROI imaging. Our comparison on ROI imaging reveals skin dose reductions of up to 35% at equal ROI image quality by using the z-sbDBA.

Conclusion: Our new concept for FFM in x-ray CT, the z-sbDBA, was experimentally validated on a clinical CT scanner. It facilitates dynamic FFM by realizing variable transmission profiles across the fan beam angle on a projection-wise basis. This key feature allows for substantial improvements in image quality, a reduction in patient radiation dose, and additionally provides a technical solution for ROI imaging. © 2020 The Authors. Medical Physics published by Wiley Periodicals LLC on behalf of American Association of Physicists in Medicine. [https://doi.org/10.1002/mp.14430]

Key words: computed tomography, dynamic beam attenuator, fluence field modulation, image variance, ROI imaging, x-ray fluence filtration

1. INTRODUCTION

Since its early days, computed tomography (CT) has continuously developed toward a major modality in medicine offering three-dimensional (3D) diagnostic information at high spatial resolution and short acquisition times. Nowadays, CT has become a workhorse in medical diagnostics and is widely used in various medical disciplines, such as cardiology, angiography, or neurology. The diagnostic information is acquired, however, using x-ray radiation which always bears...
a health risk.3–5 In accordance with the guiding principle of ALARA (as low as reasonably achievable),4 the trade-off between sufficient image quality for appropriate diagnostic information on the one hand and avoiding unnecessary radiation dose on the other hand has to be balanced carefully for every CT examination. Consequently, the reduction of the radiation dose but also image quality enhancement have always been an active field of research in CT. Various technological inventions have successfully been implemented over the decades to ease the use of CT examinations, for example, beam collimation,6,7 tube current modulation (TCM)8–10 automated exposure control,11,12 or spectral prefiltration.13 These techniques effectively reduce patient radiation dose and homogenize image noise.

Typical patient cross-sections have an elliptical shape, such that the patient attenuation changes (a) across the fan beam angles and (b) the projection angles. Unless properly compensated, the reconstructed image would suffer from horizontal streak artifacts, nonuniform noise, scatter radiation, and unnecessarily high x-ray radiation exposure of the patient.14,15 In addition, inappropriate patient positioning can lead to changes in the CT number and nonuniform image noise which could be addressed by for example adapted x-ray fluence modulation, as shown in recent studies.16–20

Our work deals with a new technical solution for dynamic modulation of the x-ray fluence in order to further improve CT scanners in terms of image noise uniformity and patient radiation dose.

This paper is a significantly expanded version of our previously published conference proceedings paper,21 which introduced the z-sbDBA concept and demonstrated a preliminary proof of concept. Here, we present a more comprehensive characterization of the z-sbDBA, focusing also on noise homogenization, dose reduction and potential region-of-interest (ROI) applications facilitated by FFM. We investigate the filter transmissivity in detail, explicitly taking into account also mechanical and spectral influences. In addition, we demonstrate possible noise distributions on a pixel-wise basis by variance maps. Based on the feasibility of spatially allocating photon fluence, we compare the potential of ROI imaging and associated patient dose reduction among different attenuators.

1.A. X-ray fluence filtration

1.A.1. Static filters

Today, most clinical CT scanners are equipped with static fluence filters — usually referred to as bowtie filters. Bowtie filters typically reduce the emitted x-ray fluence toward larger fan beam angles to compensate for patient attenuation. These filters try to reduce patient dose and achieve a more uniform noise.22,23 Bowtie filters are static in their geometry, and thus constitute a compromise between different body shapes and examination tasks. Obviously, they cannot meet the variations in attenuation across the fan beam angle on a projection-wise basis.

1.A.2. Dynamic beam attenuators concepts

To address this issue, numerous concepts for dynamic beam attenuators (DBAs) have been introduced in the past. DBAs, in general, attempt to dynamically adapt the incident x-ray fluence to the varying patient attenuation as a function of (a) the fan beam and (b) the projection angle — called fluence field modulation (FFM).

Existing concepts can be divided in different groups, based on their basic working principle:

Some DBA concepts adopted the general idea of bowtie filters, extending their functionality by providing variable geometries being penetrated by the x-ray beam.24–26 Likewise, the concepts of the piecewise-linear27,28 (consisting of triangular, opposing wedges) and the piecewise-constant29–31 dynamic attenuator (consisting of flat, opposing wedges) are based on the x-ray attenuation when penetrating solids.

Alternatively, a number of concepts suggests FFM by dynamic fluids attenuating the x-ray beam.32–36 The inverse geometry CT encompasses multiple sources which could constitute a “virtual bowtie filter.”37

Recently, the so-called multiple aperture device (MAD) was introduced. In this DBA approach, overlapping structures of highly absorbing bars allow modulating the fluence by using the Moiré effect.38–40

Numerous studies, including the ones mentioned above, have demonstrated that FFM by DBAs has the potential to reduce patient dose and noise anisotropy. Beyond that, FFM is seen as a promising concept for further CT improvements. For example, it should enable handling also off-centered patients which might reduce prescan preparation time.16,18,41 Furthermore, FFM might be relevant for future photon counting CT (PCCT) technology by avoiding pulse pileup effects due to high photon count rates.15,42–45 Another field of application for FFM is ROI imaging providing high image quality only where it is needed and saving radiation dose (lower image quality) elsewhere. Different attempts have been presented all reporting substantial radiation dose reductions.46–50

1.A.3. The sbDBA

In our previous work,51 we presented the sheet-based dynamic beam attenuator (sbDBA) as an approach for dynamic FFM. In first experiments on a clinical CT scanner equipped with our prototype, we delivered a proof of concept demonstrating its ability to dynamically shape the beam profile. We also identified several additional advantages of this concept w.r.t. regular bowtie filters: The sbDBA

- causes almost no beam hardening to the x-ray beam and the filter transmission is almost independent of the x-ray emission spectrum.
- substantially reduces attenuator-induced scatter radiation toward the patient.
However, we also recognized some challenges regarding the practical usage of the sbDBA: The calibration of the sbDBA, rotatable about two axes, will be a complex issue. Precise reproduction of the angular position is crucial to avoid ring artifacts in the reconstructed images due to miscalibrations. Furthermore, the sbDBA is only capable of producing triangular transmission profiles which are generally not the optimum for typical human body shapes. Moreover, a relatively large tilt is required to achieve a substantial attenuation. To avoid these shortcomings, we developed a new DBA concept — the z-aligned sheet-based dynamic beam attenuator (z-sbDBA) — combining the strengths of the previous sbDBA with a more sophisticated geometry.

2. MATERIALS AND METHODS

2.A. The z-sbDBA concept

Figure 1 depicts the basic principle of the z-sbDBA and describes the coordinate notations. The key feature builds upon an array of highly x-ray attenuating sheets, stacked along the z direction (parallel to the rotation axis of the CT scanner) and perfectly focused toward the focal spot. The periodicity of the sheet array is designed such that the finite size of the focal spot blurs the shadow of the single sheets. As shown in Fig. 1(a), the height of the sheets varies as a function of the fan beam angle $\beta$: At the central rays, the height is very low, symmetrically increasing toward larger fan beam angles. The gaps between the attenuation sheets are (nearly) x-ray transparent.

The z-sbDBA can be tilted about the y axis by the tilt angle $\vartheta$. In focused position, $\vartheta = 0^\circ$, only the width of the attenuation sheets reduces the x-ray intensity, almost independently of the fan beam angle. Thus, the x-ray fluence is hardly affected and given only by the thickness of the sheets in relation to the gaps, as shown in Fig. 1(b).

In Fig. 1(c), the z-sbDBA is defocused by a tilt about the y axis, $\vartheta \neq 0^\circ$, such that the x-ray source faces the sides of the attenuation sheets. Therefore, the area absorbing x-ray quanta can be extended by increasing the tilt angle $\vartheta$. This effect is stronger for larger fan beam angles $\beta$ due to the increasing height of the sheets. Consequently, the x-ray fluence can be modulated across the fan beam width by the tilt angle $\vartheta$, while the transmission is almost unaffected for central rays.

2.B. The z-sbDBA prototype

2.B.1. Manufacturing the z-sbDBA prototype

In order to evaluate the new concept in experiments on a CT scanner, we built a prototype of the z-sbDBA integrated into a standard collimator box. The z-sbDBA is composed of sheets of high density (lead) and sheets of low density (paper), building a stack of alternating highly attenuating and nearly transparent sheets. Independent of their position along the z axis, the attenuation sheets are 21.5 $\mu$m in width and show a smooth, curved transition in height increasing from 0.5 mm in the center to maximum 4.5 mm at the edges. The virtually x-ray transparent inter-space between adjacent lead sheets is approximately 100 $\mu$m. The individual sheets are focused toward the focal spot of the x-ray source (focal length: 105 mm) and the entire stack is protected against physical damage by a cover, see Fig. 2.

2.B.2. Actuation of the z-sbDBA prototype

We directly connected the z-sbDBA prototype to a high-resolution rotational drive (EC-4pole30/16RIO, Maxon Motor GmbH, Munich, Germany) in order to actuate the prototype. Furthermore, we developed a customized firmware package to control the drive and to precisely adjust the
z-sbDBA to the desired tilt angle. The remote control from outside the examination room was accomplished by a Wi-Fi connection.

The bowtie filters of a regular CT scanner (SOMATOM Force, Siemens Healthcare GmbH, Forchheim, Germany) were removed from the collimator box and the z-sbDBA was installed instead. The rotatable suspension, carrying the z-sbDBA, was mounted, such that the z-sbDBA is 105 mm from the focal spot and 490 mm from the iso-center.

2.C. Data acquisition

In our experimental setup, we equipped a clinical CT scanner (SOMATOM Force, Siemens Healthcare GmbH, Forchheim, Germany) with the modified collimator box containing our z-sbDBA prototype. Transmission profiles through the z-sbDBA corresponding to various tilt angles \( \theta \) were acquired, see Section 2.D. for a detailed description. Apart from the tilt angle, also the tube voltage \( U \) was varied in a number of acquisitions to evaluate the spectral dependency of the z-sbDBA on the transmission, see Section 2.E. for details.

For all acquisitions, the following standard scan parameters were used in our investigations: We considered the raw detector signal acquired at 300-mA-tube current, 120-kV-tube voltage, without TCM, at 1 s rotation time and the same focal spot position. In other cases, the modified parameters are denoted. The fan beam covered a detector area of 920 detector channels and 96 detector rows, with a pixel size corresponding to 0.6 mm \( \times \) 0.6 mm at the iso-center. All combinations were acquired twice in two successive scans to enable the calculation of noise-only difference images.

Unless stated differently, in this work, the transmission \( t \) is determined as the raw detector signal divided by a reference measurement without any attenuator (i.e., no bowtie filter, no z-sbDBA) present in the x-ray beam, such that \( t \in [0,1] \). Moreover, the detector signal measured at a central detector row (row no. 42) was averaged over all readings of a full rotation (1400 readings in total).

In addition, we measured the transmission of state-of-the-art bowtie filters, bowtie_1 and bowtie_2, as being part of the un-modified CT scanner. The bowtie_1 represents a standard attenuator for the compensation of patient attenuation in most cases and the bowtie_2 produces more narrow transmission profiles, mainly used for cardiac or pediatric applications. For comparisons between the z-sbDBA and the bowtie filters, the measurements were conducted at the same tube voltages and scan parameters.

2.D. Modulation of the z-sbDBA transmission

By experiments on a clinical CT scanner we performed a verification of our hypothesis stating that the x-ray transmission through the z-sbDBA can be modulated by varying the tilt angle \( \theta \) of the z-sbDBA. For a proof of concept, we evaluated the transmission \( t \) corresponding to a set of different tilt angles \( \theta \in \{0,0.3,0.6\} \) in steps of \( \Delta \theta = 0.3^\circ \).

In addition, we investigated the transmission \( t \) over the entire detector area, covering 920 detector channels and 96 detector rows. In particular, the transmissions along a lower (row 15), a central (row 48) and an upper (row 81) detector row were compared. In this experiment, the z-sbDBA was adjusted to a moderate fluence modulation (\( \theta = 1.5^\circ \)).

2.E. Spectral dependency of the transmission

By experiments, we investigate the influence of the tube emission spectrum on the transmission \( t \) through the z-sbDBA. Deviations in the resulting transmission profiles indicate spectral changes induced when x-ray penetrate through the attenuator. We conducted the experiments at two different tube voltages, \( U = \{80 \text{kV}, 140 \text{kV}\} \) to vary the tube emission spectrum. At both tube voltages, the transmission \( t \) was recorded at two tilt angles \( \theta \in \{1.0^\circ, 2.0^\circ\} \). For a comparison with state-of-the-art technology, the measurements were repeated at the same voltages using the bowtie_1 and the bowtie_2 instead. We evaluate the transmissions at different voltages across the detector width and additionally compare

![Fig. 2. The z-sbDBA prototype (black) is carried by a rotational suspension enabling a rotation about the y axis. A rotational drive (left) is directly connected to the z-sbDBA to perform the actuation. The setup is mounted into a modified collimator box (bronze) of a clinical CT scanner. The photo is taken in beam’s eye view. [Color figure can be viewed at wileyonlinelibrary.com]](image-url)
the difference in transmission depending on the reduction of the transmission.

2.F. Reconstruction and variance map

As the z-sbDBA represents a movable and structured object within the beam path, it bears the potential risk of artifacts in the reconstructed images. For example, small angular deviations due to imperfect realignment or mechanical instability would cause concentric rings in the reconstructed images.

Having a phantom at the iso-center of the CT scanner, we visually evaluated reconstructed images with respect to their image quality. The data were acquired with the z-sbDBA at θ = {1.0°, 2.0°} and reconstructed with filtered back projection. We first acquired the calibration scans (with z-sbDBA, without phantom) at given tilt angles, followed by the set of phantom scans (with z-sbDBA, with phantom) acquired afterwards. As we intentionally moved the z-sbDBA between calibration and phantom scan, we also account for potential mechanical inaccuracies of the motor during the angular realignment. The phantom used is a 32-cm-computed tomography dose index (CTDI) phantom including a 16-cm insert.

The main function of the z-sbDBA is to modulate the x-ray fluence which controls the incident radiation on the object, and thus the image variance. To investigate the influence of different FFM by applying variable attenuator properties, we calculated variance maps showing the effect on the spatial image variance in the reconstructed images.

The image variance was calculated by the variance of 48 noise-only difference images (corresponding to the detector rows 25–72) in each pixel, using all readings. The difference images were based on two successive acquisitions per scan configuration, calculated for 48 slices in z direction. The variance maps were calculated for the z-sbDBA at three tilt angles θ ∈ {1.0°, 1.5°, 2.0°}, the bowtie_1, and the bowtie_2. The attenuators used in this comparison produce individual transmission profiles, consequently, the CTDI phantom was exposed to different incident radiation during the data acquisition. To account for that, we retroactively equalized the radiation exposure among the scenarios by scaling the integral image variance according to the CTDIw measured during the respective acquisition (see next section for dosimetry). Here, we assumed that image variance is approximately inversely proportional to radiation dose which we estimated by measuring the CTDIw. Furthermore, electronic noise was neglected because of a reasonably high tube current (300 mA). Accordingly, we were able to compare the resulting image variance across the phantom cross-section between different attenuator scenarios at equal phantom radiation dose.

2.G. ROI-FFM: spatial separation of phantom exposure

By spatially and temporally adapting the x-ray fluence, we hypothesize that FFM can realize high image quality in diagnostically relevant regions (ROI) and at the same time spare surrounding tissue from unnecessary radiation dose. Thus, we investigated the potential of the z-sbDBA and the bowtie filters for ROI imaging. Therefore, we first measured the dose deposition in the CTDI phantom at a peripheral dose cavity, referred to as skin dose. To guarantee similar conditions in each dose measurement, the peripheral dose was measured at the 12 o’clock position of the CTDI phantom (farthest from the patient table) and the acquisition started at the 6 o’clock position of the CT scanner. Thus, potential tube current variations at the beginning and end of an acquisition would be minimized. We used a calibrated dosimeter (X2 Base Unit, RaySafe AB, Billdal, Sweden) for the dose measurements and then averaged the dose values of the two acquisitions. Second, we determined the image quality in a defined ROI located in the center of the CTDI phantom. The ROI diameter was selected to be 10 cm, as depicted in Fig. 3. As our measure for image quality, we chose the integrated image variance inside the ROI diameter, referred to as ROI variance, based on the difference images described in Section 2.C.4. Third, we normalized the skin dose between the attenuators to equal ROI variance by linear scaling (scaling factor τ). Here, we assumed the image variance to be approximately inversely proportional to radiation exposure. As a result, we can quantitatively compare the skin dose caused to achieve a prescribed level of image variance in the ROI. The bowtie_1 was used as reference scenario.

3. RESULTS

In this section, we present our results based on experiments using the z-sbDBA prototype on a clinical CT scanner. First, we deliver a proof of concept, demonstrating the ability of the z-sbDBA for FFM. Second, we compare the influence of the emission spectrum on the transmission between the z-
sbDBA and two bowtie filters. Third, image reconstructions of a 32-cm-CTDI phantom are computed and the image variance, resulting from the application of either the z-sbDBA or one of the bowtie filters, is compared. In the last section, we compare the potential of ROI imaging by using the z-sbDBA or the bowtie filters.

3.A. Modulation of the z-sbDBA transmission

In Fig. 4, the transmission $t$ through the z-sbDBA is depicted for a set of increasing tilt angles $\vartheta$ over the detector channels, corresponding to the full width of the fan beam. The resulting transmissions are shown in a gray scale plot, Fig. 4(a), visually demonstrating the gradual changes of the transmission profiles, and also in a line plot, Fig. 4(b), with a logarithmic axis of the transmission $t$ to cover the full range of fluence modulation across the fan beam width.

In our experimental setup, incorporating the z-sbDBA mounted on a clinical CT scanner, we can realize bell-shaped transmission profiles as a function of the tilt angle $\vartheta$. As shown in Fig. 4, the transmission profiles narrow for increasing tilt angles and show only low fluctuations of the signal. The x-ray transmission reduces not exactly symmetrically toward the edges of the detector. Upper detector channels tend to higher transmission compared to the corresponding detector rows on the other half of the detector. This deviation might be caused by imperfections of the sheet structuring or the mounting.

The transmission is roughly constant over the detector width in focused alignment ($\vartheta = 0.0^\circ$), reaching a maximum transmission of $t(\vartheta = 0.0^\circ) = 0.81$ for central rays. A slight decrease toward the edges of the detector might be due to small deviations from perfect focusing or the distorted perspective on the focal spot geometry for peripheral parts of the sheets.

By increasing the tilt angle $\vartheta$, the transmission $t$ considerably reduces at peripheral regions. For example, at detector channel 200: $t(\vartheta = 1.2^\circ) = 0.14$ and $t(\vartheta = 1.8^\circ) = 0.05$. In contrast, the transmission is nearly independent from the angular alignment at the center of the fan beam. Only small deviations from the maximum transmission can be noticed which are caused by the residual minimum height of the sheets (about 0.5 mm).

Intersections of the x-ray beam with a tilted attenuation sheet are (statistically) not fully absorbing the x-ray fluence. Hence, the signal toward the edges of the detector does not drop to zero in defocused positions. This is essential, as it avoids issues in image reconstruction due to truncated data, as mentioned in previous DBA investigations.

The distance between attenuation sheets and the focus changes when tilting the z-sbDBA. Consequently, the focusing of the sheets varies according to their position on the $z$ axis. In our setup, sheets covering upper detector rows move toward the focal spot and vice versa. Figure 5 shows the transmission $t$ over the entire detector area for a moderate fluence modulation ($\vartheta = 1.5^\circ$). Apparently, the transmission is affected by the $z$ position of the individual sheets. For lower detector rows (green), the described mechanical characteristic results in an additional defocusing and vice versa for higher detector rows (purple). For example at detector channel 200, the transmission is about $t(\text{row } 15) = 0.05$ for lower detector rows, $t(\text{row } 48) = 0.07$ in the center, and $t(\text{row } 81) = 0.10$ for upper detector rows.

3.B. Spectral dependency of the transmission

Figure 6 depicts the transmission $t$ through the attenuators at different tube voltages or tube emission spectra,
respectively. In Fig. 6(a) a relatively low FFM is shown produced by the z-sbDBA ($\theta = 1.0^\circ$) and the bowtie_1 (standard), whereas Fig. 6(b) shows the z-sbDBA ($\theta = 2.0^\circ$) and the bowtie_2 (narrow) causing a relatively strong FFM.

Generally, the attenuators show a higher transparency of x-ray radiation at higher tube voltages, as expected from the photon interactions with matter. Furthermore, the deviations between both tube voltages are larger at low transmissions. Accordingly, an influence of the emission spectrum on the transmission can be noticed for the z-sbDBA and the bowtie filters. Compared to the z-sbDBA, however, the bowtie filters apparently exhibit larger differences between the considered tube voltages. Especially toward the edges the differences become visibly larger as the penetration length through the bowtie filters increases at larger fan beam angles. These results demonstrate that the z-sbDBA features a substantially lower spectral dependency on the emission spectra compared to bowtie filters.

### 3.C. Reconstruction and variance map

Figure 7 shows reconstructed images of the 32-cm-CTDI phantom, acquired with the z-sbDBA. In case of the low FFM ($\theta = 1.0^\circ$), Fig. 7(a), the image appears without serious ring artifacts or textures. With the strong FFM ($\theta = 2.0^\circ$) instead, Fig. 7(b), the HU values slightly decline toward the outer part of the phantom (somewhat outside the 16-cm insert) compared to the center of the phantom. Apparently, the attenuation measured in the phantom scan was lower than the calibration which could indicate mechanical deviations. As mentioned, we have first conducted all calibration scans at different tilt angles, followed by the actual phantom scans later. Hence, it is very likely that the angular alignment during the calibration could not be exactly reproduced for the phantom scan.

In addition, the image shows stronger textures in the region somewhat outside the 16-cm insert. Higher image noise is to be expected from the low x-ray fluence due to the strong FFM at the periphery, though. This region — roughly matching with detector channels lower than channel 290 and higher than 630 — approximately corresponds to contributions of fan beam angles heavily attenuating the radiation propagation, compare Fig. 4(a).

Figure 8 visualizes the image variance in the reconstructed images of a 32-cm-CTDI phantom acquired with different attenuators. With the bowtie filters, the highest image variance is in the middle of the phantom; likewise, with the z-sbDBA at low FFM ($\theta = 1.0^\circ$). With moderate FFM ($\theta = 1.5^\circ$), the image variance is approximately constant across the phantom diameter — apart from the edges of the phantom. In case of strong FFM ($\theta = 2.0^\circ$) realized by the z-sbDBA, the distribution of the image variance is substantially different from the distributions caused by the state-of-the-art bowtie filters: The image variance is comparably low in the middle of the phantom and increases toward more peripheral regions. Closer to the phantom borders, the image variance decreases again to a level comparable to the other scenarios. Comparing the image variance in the middle of the phantom, this scenario enables the lowest image variance among the scenarios. Small deviations between the upper and the lower half of the variance maps can be noticed. This might be caused by the patient table which is additionally attenuating the x-ray beam.

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**Fig. 5.** (a) Transmission $t$ through the z-sbDBA shown for the entire detector area (horizontal: 920 detector channels, vertical: 96 detector rows) at fixed tilt angle $\theta = 1.5^\circ$. Lower (upper) detector rows are depicted at the top (bottom) of the image. (b) The transmission $t$ corresponding to detector rows 15, 48, and 81, as indicated by colored lines in (a). Attenuation sheets covering upper detector rows are moved toward the focal spot due to the tilt. [Color figure can be viewed at wileyonlinelibrary.com]
3.D. ROI-FFM: Spatial separation of phantom exposure

The potential of ROI imaging by the attenuators considered is reported in Table I. In our setup, the bowtie_2 — particularly dedicated to for example cardiac applications — can reduce the skin dose by 18% compared to the reference bowtie_1. The dose reductions by the z-sbDBA strongly depend on the applied tilt angle: At low FFM ($\theta = 1.0^\circ$), the z-sbDBA reveals a worse dose performance compared to the bowtie filters. At moderate FFM ($\theta = 1.5^\circ$), however, the z-sbDBA can reduce the skin dose by 23% over the reference bowtie_1. In case of strong FFM ($\theta = 2.0^\circ$), the benefits from ROI imaging are substantial: The z-sbDBA is capable of reducing the skin dose by 35% at equal ROI variance compared to the reference scenario.

This investigation demonstrates that the z-sbDBA generally enables ROI-FFM and can provide further dose savings to the patient. According to our methodology, the experiments show that the z-sbDBA facilitates substantial reductions of the skin dose when used appropriately.

4. DISCUSSION

In this work, we introduced a new concept for dynamic FFM applicable to x-ray CT, the so-called z-sbDBA. For experimental validations, we built a first prototype of the z-sbDBA fully integrated into a clinical CT scanner. In experiments we demonstrated its feasibility for FFM by realizing variable transmission profiles.

Furthermore, the new z-sbDBA concept overcomes some limitations of the previously presented sbDBA. Notable improvements due to the revised structuring of the z-sbDBA can be verified: (a) The z-sbDBA reduces the sensitivity to mechanical inaccuracies by avoiding a highly structured object along the fan beam width. As a consequence, fluctuations of the transmitted signal are reduced and the risk of ring...
artifacts in the reconstructed image can be mitigated. (b) The presented z-sbDBA can realize bell-shaped transmission profiles which are generally assumed to be more appropriate for typical human body shapes than the triangular-shaped transmission profiles produced by the sbDBA. (c) The structuring allows to achieve considerable fluence modulations with only a few degrees tilt (compare sbDBA: about 30°–40°), making the technical realization and operation much easier.

Besides these advantages, the main limitation of the z-sbDBA might be its limitation to symmetrical transmission profiles. Unlike other DBA concepts, the z-sbDBA cannot shift the peak transmission along the fan beam width but is centered on the iso-ray. 

We also compared state-of-the-art bowtie filters with the z-sbDBA in experiments on a CT scanner. The comparisons reveal major advantages of the z-sbDBA concept: (a) In particular, the z-sbDBA shows a considerably lower dependency of the transmission on the emission spectra. Similar to the sbDBA, the structure of highly attenuating sheets can reduce scatter radiation such that a higher portion of primary photons transmit the z-sbDBA. (b) Based on our results, we can conclude that the z-sbDBA allows to influence the image variance in the reconstructed image, and thus also the spatial dose deposition. In particular, the z-sbDBA facilitates distributions where the image variance is lower in central than in surrounding regions. This might be essential for applications in ROI imaging. The calibration and the mechanics, however, need to be improved for a more stable usage of the new filter. (c) Moreover, the results demonstrate that the z-sbDBA is generally suitable for ROI imaging. According to our methodology the z-sbDBA reveals substantial reductions of the skin dose when used appropriately.
Our results regarding potential dose reduction and the effect on the image variance distribution seem to be in accordance with the findings of other DBA approaches. It is difficult to compare statements on relative dose saving, of course, because they depend on the point of reference: Was TCM included? Was a standard bowtie included? However, most studies reported dose savings in the range of 30% to 50% w.r.t. conventional bowtie filters. Only the inverse CT geometry yields far higher dose savings of up to more than 80%. Comparable improvements in noise homogeneity are e.g. achieved by the fluid-filled dynamic bowtie filter or the MAD. These two approaches are examples of DBAs that allow the transmission profiles to be shaped very flexibly. Due to the structuring of the MAD, the overall transmissivity reduces to a maximum of 50%, though. Several studies using DBAs for ROI imaging demonstrate comparable spatial variations in image noise as we found in our experiments.

The mechanical stability and reproducibility are generally seen as a crucial issue which should monitored very carefully. We acknowledge that our first prototype of the z-sbDBA shows some inaccuracies w.r.t. the angular alignment which should be improved in future prototype developments. Studies investigating alternative DBA approaches also acknowledged that mechanical implementation is crucial. Moreover, different shapes of the attenuation sheets should be tested to optimize the width of the transmission profiles to common clinical scenarios. The hereby introduced z-sbDBA provides a hardware solution for FFM. In order to use the z-sbDBA efficiently, however, an optimization criterion defining the optimum fluence for a given object should be developed in a subsequent step. It could depend not only on the patient anatomy but also consider task-specific aspects. This topic might include ROI imaging of centered and off-centered ROIs.

5. CONCLUSIONS

In conclusion, our experiments with a prototype integrated into a clinical CT scanner provide a proof of concept showing that the z-sbDBA facilitates FFM. By only small tilts of the z-sbDBA, the intensity of the propagated x-ray radiation is modulated. Hence, the local image variance — and consequently also the radiation dose deposition inside the object — can be adapted according to the needs and tasks. We have demonstrated the substantial improvement in image quality, the reduction in patient radiation dose and the potential benefits of ROI imaging by using the z-sbDBA.

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CONFLICT OF INTEREST

The authors have no conflict to disclose. None of the presented DBA approaches are available in commercial CT scanners.

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