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High-Resolution Extremity Cone-Beam CT with a CMOS Detector: Evaluation of a Clinical Prototype in Quantitative Assessment of Bone Microarchitecture.

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Abstract

Purpose: A prototype high-resolution extremity cone-beam CT (CBCT) system based on a CMOS detector was developed to support quantitative *in vivo* assessment of bone microarchitecture. We compare the performance of CMOS CBCT to an amorphous silicon (a-Si:H) FPD extremity CBCT in imaging of trabecular bone.

Methods: The prototype CMOS-based CBCT involves a DALSA Xineos3030 detector (99 μm pixels) with 400 μm -thick CsI scintillator and a compact 0.3 FS rotating anode x-ray source. We compare the performance of CMOS CBCT to an a-Si:H FPD scanner built on a similar gantry, but using a Varian PaxScan2530 detector with 0.137 mm pixels and a 0.5 FS stationary anode x-ray source. Experimental studies include measurements of Modulation Transfer Function (MTF) for the detectors and in 3D image reconstructions. Image quality in clinical scenarios is evaluated in scans of a cadaver ankle. Metrics of trabecular microarchitecture (BV/TV, Bone Volume/Total Volume, TbSp, Trabecular Spacing, and TbTh, trabecular thickness) are obtained in a human ulna using CMOS CBCT and a-Si:H FPD CBCT and compared to gold standard μCT .

Results: The CMOS detector achieves ~40% increase in the f20 value (frequency at which MTF reduces to 0.20) compared to the a-Si:H FPD. In the reconstruction domain, the FWHM of a 127 μm tungsten wire is also improved by ~40%. Reconstructions of a cadaveric ankle reveal enhanced modulation of trabecular structures with the CMOS detector and soft-tissue visibility that is similar to that of the a-Si:H FPD system. Correlations of the metrics of bone microarchitecture with gold-standard μCT are improved with CMOS CBCT: from 0.93 to 0.98 for BV/TV, from 0.49 to 0.74 for TbTh, and from 0.9 to 0.96 for TbSp.

Conclusion: Adoption of a CMOS detector in extremity CBCT improved spatial resolution and enhanced performance in metrics of bone microarchitecture compared to a conventional a-Si:H FPD. The results support development of clinical applications of CMOS CBCT in quantitative imaging of bone health.

Keywords

cone-beam CT; extremities imaging; CMOS; bone microarchitecture

1. INTRODUCTION

Despite the growing evidence of correlations between bone microarchitecture (e.g. trabecular thickness, spacing and anisotropy) and early onset/progression of osteoporosis (OP) and osteoarthritis (OA), clinical utilization of metrics of bone microstructure is limited because of insufficient spatial resolution of current imaging modalities (typically >0.3 mm) compared to the size of features of interest (e.g. 0.1 mm for the trabeculae). Recently introduced extremity cone-beam CT (CBCT) systems [1]–[3] based on amorphous silicon (a-Si:H) Flat Panel Detectors (FPD) can resolve ~ 250 μm details in high-resolution imaging modes. This improved spatial resolution compared to conventional CT, combined with advantages of logistics, radiation dose and capability for weight-bearing imaging, position extremity CBCT as an attractive platform for comprehensive evaluation of all aspects of bone and joint health. This contribution investigates the benefits of adopting CMOS x-ray detectors to support applications in measurements of bone microarchitecture. CMOS detectors offer improved performance in high resolution imaging compared to a-Si:H FPDs due to their smaller pixel size, lower electronic noise, and faster frame rate. Following from previous work in modelling and test-bench evaluation of CMOS-based extremity CBCT [4], [5], a prototype scanner was developed. We evaluate the imaging performance of the prototype CMOS CBCT and compare to a-Si:H FPD CBCT, in particular with respect to metrics of trabecular bone microarchitecture. Metrics of trabecular architecture on the a-Si FPD and CMOS scanners are evaluated against gold standard micro-CT in a cadaveric sample of human ulna.

2. METHODS**2.1 A prototype CMOS-based extremity CBCT scanner**

The CMOS-based extremity CBCT scanner was built on a gantry that is largely identical to a commercially available a-Si:H FPD extremity scanner (OnSight3D, Carestream Health, Rochester NY). Both systems are compared side-by-side in Fig. 1A. The gantry allows weight-bearing and non-weight-bearing imaging of the extremities and provides a compact geometry with source-detector distance of 560 mm and magnification of 1.3 . The CMOS-based scanner utilizes a compact rotating anode source with 0.3 FS focal spot (IMD RTM 37) and a Dalsa Xineos3030 CMOS detector with 99 μm pixel size and a custom 400 μm thick CsI scintillator, selected in task-based optimization to maximize detectability in high resolution imaging [4], [5]. The imaging protocol involved 90 kVp beam energy, 420 frames over a 210° short scan trajectory, and dose of ~ 15 mGy (central dose in a 16 cm CTDI phantom). Imaging performance of the CMOS-based extremity CBCT was compared to an a-Si:H FPD-based scanner (prototype OnSight3D system) using a Varian PaxScan2530 detector with pixel size of 0.137 mm and CsI:Tl scintillator thickness of 0.7 mm, and a stationary anode x-ray source with 0.5 FS focal spot. The imaging protocol also involved

420 frames and a 90 kVp beam. Central CTDI dose was 12 mGy. The system and acquisition parameters of the CMOS and a-Si scanners are summarized in Fig. 1B.

The CBCT scans for both systems were acquired at the native pixel size of their detectors. This acquisition mode is not commonly used in clinical imaging on the a-Si:H FPD system, where 2×2 binning is typically employed to reduce scan time. As shown in Fig. 1B, the higher frame rate of the CMOS detector results in dramatic reduction in scan time for acquisitions at native pixel binning, from ~60 sec for a-Si:H FPD CBCT to ~17 seconds for CMOS CBCT.

2.2 Performance evaluation

Resolution of the prototype CMOS system was evaluated in comparison to the a-Si:H FPD CBCT in both projection and 3D reconstruction domains. Detector MTF was measured for both systems using a tungsten edge placed at the center of the detector surface at a tilt of ~2° and imaged for 100 frames with the gantry held stationary. An oversampled edge-spread profile was obtained to derive the line-spread functions and MTFs. The detector presampling MTF was then obtained by dividing out a sinc function corresponding to the bin size of the oversampled edge. For characterization of 3D resolution, a ~127 μm tungsten wire tensioned within a hollow plastic cylinder was scanned at the center of FOV of both systems and reconstructed with a ramp filter with Nyquist cutoff frequency at a fine voxel size of 25 μm .

Visual evaluation of image quality for the two systems was performed by an MSK radiologist in scans of a cadaveric ankle.

The high-resolution reconstruction protocols used the Feldkamp algorithm and 25 μm voxel grid. The reconstruction filter was a Hann-apodized ramp. The cutoff frequency was the same for both systems and equal to ~3.3 mm^{-1} .

The high-contrast reconstruction protocol involved binning of the projection data on each system to ~0.3 mm pixel size (2×2 binning on a-Si:H FPD CBCT, identical to the current clinical extremity CBCT protocol, and 3×3 binning on CMOS CBCT). The high-contrast reconstructions also utilized a Hann filter. The cutoffs were ~1.7 mm^{-1} and the voxel size was 200 μm .

The images from the two scanners were registered to facilitate visual comparison. First, a rigid image-domain pre-registration was performed using initial reconstructions. Next, the transform found in the pre-registration was applied to image coordinate system during backprojection to match the position of the two reconstructed volumes with respect to anatomy. This approach avoids interpolation and blurring that would be present if the coordinate transform was applied directly to the reconstructed volume.

Finally, the performance of the two CBCT systems was compared in the following metrics of trabecular microarchitecture: bone volume to total volume ratio (BvTv), mean trabecular thickness (TbTh, mean thickness of a trabecular “ridge”) and mean trabecular spacing (TbSp, the mean size of the cavities surrounded by the trabecular “ridges”). The metrics were computed from a binary image obtained by a segmentation of bone voxels (“ridges”) from the original volume. BvTv was then measured by simply counting the bone voxels. The

trabecular thickness and spacing were then estimated using the sphere-fitting approach of [6]. For TbTh, each voxel in the binary image of trabecular “ridges” was assigned the diameter of the maximum sphere that included that voxel and fit entirely within the ridge. For Tb.Sp, the same procedure was used, but now on the inverted binary image of the bone segmentation. Mean values of TbTh and TbSp were then obtained by averaging the voxel values (sphere diameters) in the two maps. This process is illustrated in Fig. 2.

In this study, the metrics of trabecular microarchitecture were applied to an excised human ulna sample. The sample was scanned on a μ CT system (Skyscan1176, Bruker, BE) to obtain a gold standard characterization of trabecular detail at 28 μ m voxel size. The ulna was then imaged on the a-Si:H FPD extremity CBCT and on the CMOS-based extremity CBCT prototype. High resolution CBCT reconstructions using 28 μ m voxels and Hann filters with cutoffs set at the Nyquist frequency of each system were obtained. The coordinate transform method described above was applied to obtain mutually registered CBCT and μ CT volumes without interpolation.

Trabecular analysis was performed in 16 corresponding ROIs ($4 \times 4 \times 4$ mm³) randomly selected in the CBCT and μ CT images. To segment bone voxels in μ CT ROIs, global thresholding (0.2 of max-min-normalized intensity for the entire image) followed by morphological closing (in a $5 \times 5 \times 5$ neighborhood) was applied.

CBCT ROIs were segmented by sweeping the threshold level for each ROI to find a value that gave maximal Dice similarity index with the μ CT segmentation of the same ROI. This approach uses a fairly simple segmentation algorithm to focus the comparison on the effects of baseline imaging performance of the two systems. Metric values were derived from the ROI segmentations and the correlation between CBCT and μ CT was assessed using Pearson’s coefficient.

3. RESULTS AND BREAKTHROUGH WORK

Figure 3A compares the detector MTF for aSi:H FPD and CMOS CBCT systems. In addition to extended Nyquist frequency, the frequency at 50% modulation is approx. 0.5 lp/mm higher for the CMOS detector compared to a-Si FPD. The frequency at 20% modulation of the CMOS is approx. 1 lp/mm larger than that of the aSi FPD. Fig 3B shows the highest possible resolution reconstruction (ramp filter, Nyquist cutoff) of the 127 μ m tungsten wire. A Gaussian function was fit to line profiles through the wire. The thus estimated FWHM of the CMOS system is approx. 0.1 mm smaller than that of the aSi:H-FPD system.

CMOS and a-Si:H FPD systems are compared in a realistic clinical scenario in Fig. 4A using high-resolution reconstructions of the cadaveric ankle obtained at approx. equal dose. A modest but perceptible improvement in delineation of bony detail with a CMOS detector is apparent, especially in the improved modulation for fine trabecular features in the enlarged view of the calcaneus. Figure 4B shows high-contrast reconstructions of the ankle, where the projection data of both systems was binned to approx. 0.3 mm pixel pitch to reduce noise. Despite the smaller native pixel size and thinner scintillator, the CMOS image

maintains sufficient contrast-to-noise performance to achieve comparable visualization of soft-tissue structures to the a-Si:H FPD system after appropriate post-processing.

As shown in Fig. 5 and 6, the intrinsic resolution advantage of CMOS CBCT leads to improved performance in quantitative evaluation of trabecular microstructure. Visual analysis of an example ROI in Fig. 5 reveals that CMOS-CBCT yields a segmentation that not only better preserves the topology of trabecular microstructure compared to the a-Si:H FPD, but also delivers better modulation of fine detail. This is supported by the observation that CMOS CBCT yielded improvement in the Dice coefficient of the trabecular segmentations (Fig. 6A). For all ROIs, the Dice coefficient for CMOS-based segmentation was 5–15% better than for a-Si FPD. The median Dice value of all ROIs was improved by approx. 10%.

Fig. 6B–C summarizes the performance of both system in measurements of bone microstructure. The improved segmentation achieved by CMOS lead to improved correlation of the trabecular metrics with gold standard μ CT. Both systems achieve correlations better than 0.9 for bone volume over total volume (Fig. 6B) and trabecular spacing (Fig. 6C), where the CMOS detector provides ~10% improvement over aSi:H FPD. The benefit of CMOS is more pronounced for TbTh, where the correlation coefficient is increased by approx. 30%, from 0.49 to 0.74. Trabecular “ridges” are typically thinner (3–4x) than the spaces that separate them. Therefore TbTh is expected to be the most challenging metric to measure in the presence of system blur and stands to benefit the most from improved spatial resolution.

4. CONCLUSIONS

The prototype CMOS CBCT system achieved ~40% improved resolution, better visualization of bony detail at equal scan dose, and improved correlation with gold standard μ CT in metrics of bone microarchitecture compared to the system with conventional a-Si:H FPD. Moreover, the scan time in high resolution mode is ~3x shorter using CMOS than with the a-Si:H FPD operated in 1x1 binning, greatly mitigating the risk of patient motion. Overall, the CMOS-based scanner delivers promising performance in quantitative trabecular imaging. Ongoing studies will extend the current evaluation to assessment of microarchitecture in multiple sites in the extremities using bone core samples, cadavers, and in pilot patient test-retest studies. To further enhance quantitative accuracy both in CMOS CBCT and in a-Si:H FPD CBCT, advanced image processing and trabecular segmentation algorithms are also being pursued.

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Figure 1.
(A) CMOS-based extremity CBCT system along with details of the CMOS CBCT gantry.
(B) Key system and scan parameters for CMOS CBCT and aSi:H FPD CBCT.

Figure 2.

Example of trabecular metric maps derived from a μ CT image ROI of an ulna biopsy sample. From left to right, the reconstructed image, its segmentation, as well as the spacing and thickness maps derived from the segmentation.

Figure 3.

(A) Detector MTF measurements. (B) Line profile through FBP reconstructions of a tungsten wire.

Figure 4. Reconstructions of a cadaveric ankle on a-Si:H FPD and CMOS CBCT systems in **(A)** high-resolution protocol and **(B)** high-contrast protocol.

Figure 5.

Axial slices through the reconstructions of the entire ulna are shown in top row. Sample trabecular ROIs (selected inside the ulna for analysis) are shown in row 2–4 with overlaid trabecular segmentations.

Figure 6.

(A) Distribution of Dice coefficient between the CBCTs and μ CT ROI segmentations and (B)(C)(D) correlation of trabecular metrics obtained with CBCT and μ CT. Dashed opaque lines denote confidence interval corresponding to $\alpha=0.05$. Grey solid line has unit slope and zero intercept (equivalence line representing absolute agreement).