Bone 48 (2011) 1291–1297

Contents lists available at ScienceDirect

Bone

journal homepage: www.elsevier.com/locate/bone

Quantitative characterization of subject motion in HR-pQCT images of the distal radius and tibia

Miki Sode [a,b,*], Andrew J. Burghardt [b], Jean-Baptiste Pialat [b,c,d], Thomas M. Link [b], Sharmila Majumdar [a,b]

[a] Joint Graduate Group in Bioengineering, University of California at San Francisco and Berkeley, San Francisco and Berkeley, CA, USA
[b] Musculoskeletal Quantitative Imaging Research Group, Department of Radiology and Biomedical Imaging, University of California at San Francisco, San Francisco, CA, USA
[c] INSERM U831 and Université de Lyon, Lyon, France
[d] Edouard Herriot Teaching Hospital, Hospices Civils de Lyon, Lyon, France

ARTICLE INFO

Article history:
Received 10 December 2010
Revised 8 March 2011
Accepted 14 March 2011
Available online 21 March 2011

Edited by: Harry Genant

Keywords:
Motion artifacts
HR-pQCT
Sinogram
Similarity measures
Trabecular bone structure

ABSTRACT

Image quality degradation due to subject motion is a common artifact affecting *in vivo* high-resolution peripheral quantitative computed tomography (HR-pQCT) of bones. These artifacts confound the accuracy and reproducibility of bone density, geometry, and cortical and trabecular structure measurements. Observer-based systems for grading image quality and criteria for deciding when to repeat an acquisition and post hoc data quality control remain highly subjective and non-standardized. This study proposes an objective, quantitative technique for measuring subject motion in HR-pQCT acquisitions from raw projection data, using image similarity measures applied to parallelized projections at 0° and 180°.

A total of 88 HR-pQCT exams with repeated acquisitions of the distal radius ($N=54$) or distal tibia ($N=34$) of 49 women (age $=59\pm14$ year) and 3 men (46 ± 2 year) were retrospectively evaluated. All images were graded from 1 (no visible motion artifacts) to 5 (severe motion artifacts) according to the manufacturer-suggested image quality grading system. In addition, to serve as the reference case without motion artifacts, two cadaveric wrist and two ankle specimens were imaged twice with repositioning. The motion-induced error was calculated as the percent difference in each bone parameter for the paired scans with and without visually apparent motion artifacts. Quantitative motion estimates (QMEs) for each motion-degraded scan were calculated using two different image similarity measures: sum of squared differences (SSD) and normalized cross-correlation (NCC).

The mean values of QME_{SSD} and QME_{NCC} increased with the image quality grade for both radius and tibia. Quality grades were differentiated between grades 2 and 3 using QME_{SSD}, but not with QME_{NCC}, in addition to between grades 4 and 5. Both QMEs correlated significantly to the motion-induced errors in the measurements and their empirical relationship was derived. Subject motion had greater impact on the precision of trabecular structure indices than on the densitometric indices.

The results of this study may provide a basis for establishing a threshold for motion artifacts in accordance to the study design as well as a standardized quality control protocol across operators and imaging centers.

Introduction

An increasing number of single and multi-center research and clinical studies of issues related to skeletal health have used high-resolution peripheral quantitative computed tomography (HR-pQCT) for non-invasive, *in vivo* assessment of trabecular bone structure in the peripheral skeleton. The accuracy for estimating density, cortical geometry, trabecular structure, and mechanical parameters (using micro-finite element modeling) has been validated against gold-standard measurements [1–6]. The *in vivo* reproducibility (CV_{rms}) for densitometric measures and trabecular structure indices is less than 1% and approximately 4.5%, respectively [3,7–9]. The finite precision of these measurements can be attributed to intrinsic performance limitations of the scanner hardware and image formation process, operator-related reproducibility of the acquisition and analysis procedures, limitations of the applied image processing routines, and subject motion.

Subject motion has been, and remains, a challenge in obtaining reliable HR-pQCT scans for quantitative analysis. Repeat acquisitions are often necessary to obtain images of adequate quality. Although it takes less than 3 min for image acquisition during the standard *in vivo* protocol, motion artifacts are commonly observed in the reconstructed images (Fig. 1), especially when imaging the forearm [3]. Subject movements during HR-pQCT image acquisition can include tremor, twitch/spasm, and gradual translations or rotations. Unlike periodic motion due to respiration, cardiac motion, blood flow, etc.,

* Corresponding author at: Musculoskeletal Quantitative Imaging Research Group, QB3/Byers Hall, Suite 203, 1700 4th Street, University of California, San Francisco, San Francisco, CA 94158, USA. Fax: +1 415 514 9656.
E-mail address: miki.sode@ucsf.edu (M. Sode).

8756-3282/$ – see front matter © 2011 Elsevier Inc. All rights reserved.
doi:10.1016/j.bone.2011.03.755

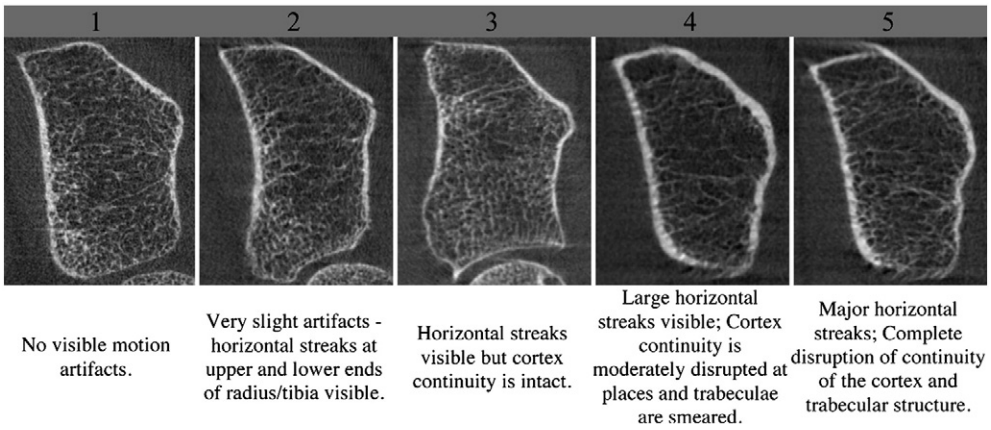

Fig. 1. Image quality grading guideline suggested by the manufacturer and representative reconstructed grayscale image of the distal radius for each grade.

these involuntary, random motions are difficult to predict and monitor.

Subject motion during image acquisition can result in severely degraded HR-pQCT image quality *in vivo*. It introduces substantial error, diminishing the accuracy and reproducibility of measurements obtained from the images. Longitudinal changes in density, cortical geometry, and trabecular structure measurements in postmenopausal women after being on anti-resorptive treatment for 12 months are at the same order of magnitude as the reproducibility [10–12]. However, individual errors can be as high as 12% and 30% with severe motion artifacts (unpublished data). Indices that describe trabecular structure are likely more prone to such errors compared to densitometric indices [3]. Because trabeculae span only 1–3 voxels in width, the depiction of the trabecular structure is subject to significant, variable partial volume averaging — a challenge for threshold-based analyses.

The detection and correction for subject motion in tomographic image data have been the subject of considerable research effort [13–18]. As three-dimensional computed tomography images are reconstructed from a series of projection images collected across at least 180° over a certain integration time, subject motion during acquisition alters each projection according to its magnitude, mode, and timing. Therefore, the set of projections collected during a tomographic acquisition encodes temporal and spatial information of the motion. The majority of approaches focus on an analysis of sinogram shape. The edge of an object in the projection appears sinusoidal in the sinogram space; hence the deviation from the idealized sinusoidal line is assumed to be due to subject motion. Fiducial markers [14] or anatomical landmarks are often traced in the sinogram space.

In an effort to provide a guideline for grading image quality, the manufacturer has provided a qualitative grading system according to the apparent severity of motion artifacts in the image (Fig. 1). The criteria for grading, however, are highly subjective, are not based on a quantitative measurement, and have not been related to error in bone quality outcome parameters. While this image quality grading system can distinguish the worst image quality (grade 4 or 5) from the best quality (grade 1 or 2), the discriminatory power is not linear or reliable (unpublished data).

If subject motion in the image can be quantified, the magnitude of motion-induced error in the measurements can be predicted. Such a procedure is essential for establishing not only a threshold for motion artifacts in order to control image quality to detect the difference in accordance to the study design but also a standardized quality control protocol across operators and imaging centers. It also allows realistic assessment and comparison across study results that use *in vivo* HR-pQCT. Therefore, an objective, standardized procedure for repeating the acquisition based on empirical data that allows immediate decision-making in a clinical setting is necessary.

The objective of this study is two-fold: (1) to develop a metric for quantifying subject motion objectively during an HR-pQCT acquisition (quantitative motion estimate, QME), and (2) to define parameter-specific relationships between the metric and expected precision error.

Methods

Proposed method for an objective detection of subject motion

In this study, we propose an objective technique for measuring subject motion based on the similarity between the parallelized projections acquired at 0° and 180°. Fig. 2 summarizes the workflow of the proposed method for measuring the amount of motion during a single acquisition quantitatively. The proposed method is based on the assumption that if there was absolutely no motion, parallel projection images at 0° and 180° would be mirror images. Any differences between these two parallel projections are, therefore, assumed to be primarily due to subject motion during the acquisition. Therefore, by comparing the differences between parallel projection images at 0° and mirrored parallel projection images at 180° using a similarity measure, the subject motion can be estimated (Fig. 3).

Parallelization

Divergence of the cone-beam configuration of X-ray beam used in the current CT image acquisition introduces magnification that results in dissimilarity in the projections at 0° and 180° due to differences in object location with respect to the source and detector. To eliminate this magnification, a series of cone-beam projections was reformatted to a series of parallel projections (in the azimuthal/fan-beam plane). First, the dark and flat field intensities were corrected in each raw projection image. Parallel rays were collected over an interval of projections equal to the fan angle of the beam (Fig. 2).

To construct a parallel projection image at an angle from the acquired raw cone-beam projection data, the beams parallel to that angle were extracted from 78 sequential projections spanning ±9.32° (Fig. 2). The resulting parallel projection image at each angle, therefore, contained both spatial and temporal information collected over this range. A total of 3 parallel projection images at 0°, 0.24° (the second acquired projection), and 180° were constructed from the raw cone-beam projection using the manufacturer-provided algorithm prior to the calculations of the similarity measures. The resulting images corresponded to the palmar and dorsal projections for the radius, and the medial and lateral projections for the tibia (Fig. 2).

The parallel projection image at 180° was then mirrored with respect to the center of rotation on the detector to match and to be compared to the parallel projection image at 0° (Fig. 2). Finally, a fixed threshold was applied to the parallel projection image at 0°, to identify the region containing bone along the long axis of the detector. On average, this bone region spanned 528 pixels and 598 pixels out of