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CT density overestimation in enclosed regions and its implications on estimating bone's mechanical properties

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Abstract

We investigated how the density and strength of trabecular bone as measured with computed tomography (CT) could be affected by being enclosed in cortical bone. To this end we measured the opacity of a low density CT phantom enclosed in a sheath of high density contrast fluid. We found a clear correlation between increasing thickness of the radiopaque sheath and increase in measured opacity within the enclosed phantom. This increase was not due to partial volume effects or the scanner's point spread function, but presumably caused by beam hardening and the way the image reconstruction algorithm deals with it. An increase was visible both with and without water submersion, and for all investigated reconstruction kernels. We then estimated what the error in derived mechanical properties for finite element (FEM) models of such bony structures may be. We estimate that the observed effect can lead to inaccuracies exceeding 33% in derived mechanical strength of trabecular bone. Recent skeletal FEM studies seem to insufficiently guard against this issue.

Keywords Computed Tomography, beam hardening, Hounsfield, finite element, density, cortical, trabecular, cancellous, bone

4.1 Introduction

The Finite Element Method (FEM) has proven itself a powerful tool in analysing stress distribution in bone [Taylor et al., 1995, Weinans et al., 2000, Stolk et al., 2002, Schileo et al., 2007]. FEM models need to accurately represent both the geometry and mechanical properties of bone. One way of measuring these properties is by X-ray computed tomography (CT).

CT measures the radiodensity of an object relative to water and air, quantified according to the Hounsfield scale. We can relate the measured CT number of each volume element, in Hounsfield Units (HU), to bone mineral ("ash") density, which in turn can be directly related to the bone's mechanical properties. Several recent studies have used this approach [Viceconti et al., 2004, Taddei et al., 2006, Imai et al., 2008, Bessho et al., 2009].

Beam hardening artefacts, caused by radiodense materials, have been widely described, and result in deviant CT numbers being reconstructed. These artefacts may appear as a lowering of apparent attenuation coefficients, also known as the "cupping" effect [Brooks and Chiro, 1976]. In the human body, radiodense cortical bone often encloses lower density tissues such as trabecular bone. CT opacities measured in these enclosed regions might therefore also be affected by beam hardening.

Beam hardening artefacts can be reduced by enveloping the target in water [Brooks and Chiro, 1976]. Computational beam hardening correction may be applied by scanning a calibration phantom simultaneously with the target in a process known as Quantitative CT (QCT).

While QCT using standard CT hardware is accurate in measuring vertebral bone density [Zink et al., 1997], special dual energy CT is required to fully correct for beam hardening caused by thicker cortical bone [de Casteele, 2004, DeLuca et al., 2009]. However, the use of standard CT to assign material properties to patient specific FEM models is widespread. Several recent studies omit any mention of using QCT or beam hardening correction [Viceconti et al., 2004, Imai et al., 2008, Keyak and Skinner, 1993, Zannoni et al., 1998]. [Taddei et al., 2006] submerged their CT target in water to suppress beam hardening, but made no explicit mention of using QCT. [Bessho et al., 2009] referred to QCT, but did not mention whether single or dual energy was used.

[Tanck et al., 2010] examined the effect which the shape of the surrounding "lower body model" has on volumetric bone mineral density measured within femoral heads. They measured density values from water basin scans that were on average 10.8% lower than in situ. Though recognizing that beam hardening caused by surrounding structures contributes to this discrepancy, they did not quantify the effect caused by the femur's cortical bone itself.

The goal of this investigation was to quantify how cortical bone would affect the CT numbers measured in enclosed trabecular bone, and to provide an indication on what the effect of resulting inaccuracies would be on the bone's derived mechanical properties.

4.2 Materials and Methods

We used a solid cylindrical Perspex rod with a diameter of 20mm and length 150mm to represent low density trabecular bone. Perspex has a known CT number of 122 ± 5 HU [Brown et al., 2008], which is representative for trabecular bone found in elderly patients [Mosekilde et al., 1989]. We performed all scans in a clinical CT scanner (Aquilion One, Toshiba Medical Systems, Japan) using its 64-slice helical mode, operating at 120 kVp and 250 mAs.

The scan target consisted of the Perspex rod in isolation, as well as concentrically enclosed in one of four contrast fluid filled PVC pipes of increasing diameter. The contrast fluid consisted of a mix of water and Iomeprol (Iomeron

400, Bracco Imaging, Konstanz, Germany) with an opacity of 1030 HU, which is in the same range as normal human cortical bone [Mukherjee and Rajagopalan, 2007]. The wall thickness of each pipe was 3mm, with a mean opacity of 380 HU. The four PVC pipes had inner diameters of 25mm, 33mm, 43mm and 68mm respectively. With the Perspex rod inserted this resulted in concentric contrast fluid sheaths with thicknesses of 2.5mm, 6.5mm, 11.5mm and 24mm respectively - see Fig. 4.1.

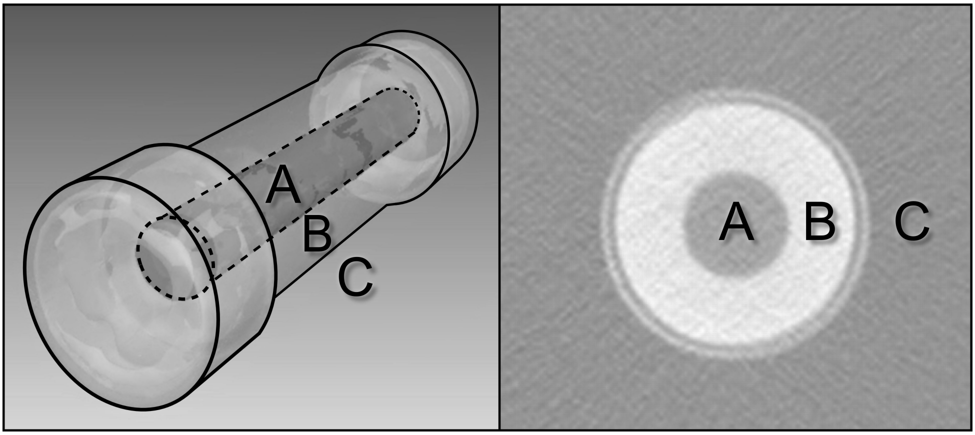

Figure 4.1 – Schematic and axial view of one of the CT phantoms with A) The Perspex rod, B) the contrast filled pipe, C) surrounding water or air. The darkened band between B and C is due to the lower-density construction of the inner pipe wall.

For each parameter set the target was scanned suspended in air, and again when submerged in a rectangular water bath measuring 240mm x 300 mm x 380 mm. To ensure that our results were not determined by the choice of reconstruction kernel, we used three substantially different reconstructions. Both "smooth" and "sharp" kernels were chosen. On our scanner these kernels are labelled FC04, FC14 and FC30. FC30 is a "sharp" bone kernel used as a standard kernel for clinical skeletal imaging. FC04 and FC14 are both "smooth" kernels, the difference being that FC04 incorporates beam hardening correction, and FC14 does not. The Perspex rod was segmented from the reconstructed images using the MITK toolkit [Maleike et al., 2009], and its opacity distribution in each image analysed using the DeVIDE image processing platform [Botha and Post, 2008]. We consistently chose the bar's innermost cylindrical region, 10mm in diameter, as region of interest (ROI), where we recorded the mean CT numbers and their 95% per-voxel confidence intervals.

4.3 Results

Observed opacity in the ROI increased strongly as the surrounding thickness of contrast fluid increased. This trend was observed for all three of the reconstruction kernels - see Fig. 4.2 and Fig. 4.3.

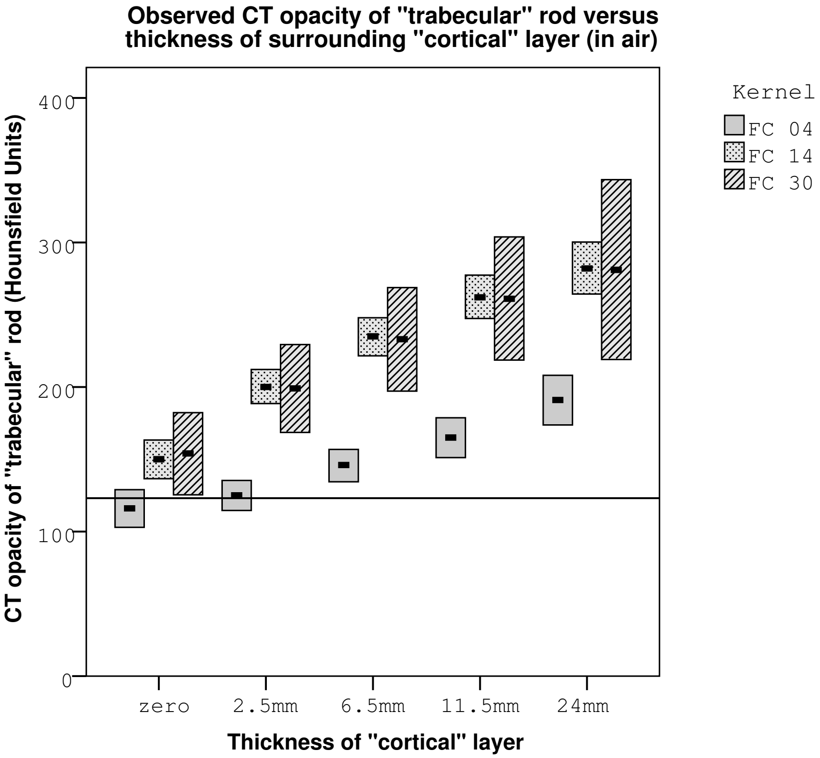

Figure 4.2 – CT values measured in the enclosed Perspex rod (surrounded by air). Average $\pm 2\sigma$ plotted for each region of interest.

While the increase occurred with and without submersion in water, the presence of water somewhat suppressed this trend (Fig. 4.3). This water induced suppression was strongest for the FC14 and FC30 kernels which lack beam hardening correction.

As expected, submersion in water increased image noise, as reflected by the standard deviation of opacity values. This increase exceeded a factor of 10 for the "sharp" kernel (FC30), and a factor of 7 for the "soft" kernels. Noise also increased with increasing diameter of the radiopaque sheath and hence the attenuation – amounting to a factor of 1.3 between an absent layer and the layer of 24mm thickness.

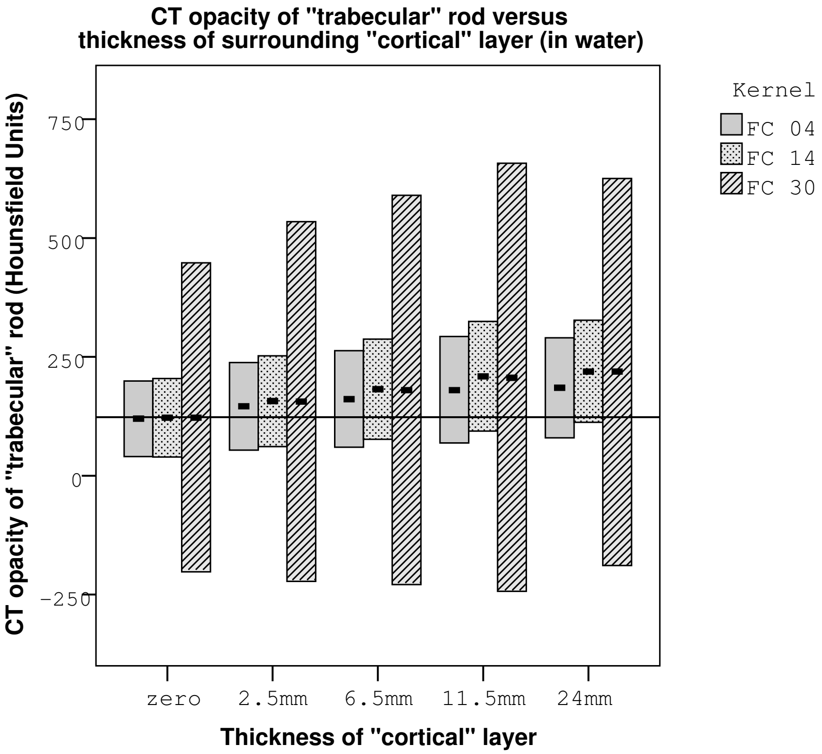

Figure 4.3 – CT values measured in the enclosed Perspex rod (immersed in water). Average $\pm 2\sigma$ plotted for each region of interest.

The only cases in which the ROI's mean opacity matches the known value for Perspex is in the absence of contrast fluid while submerged in water, and in air when a beam hardening correcting kernel (FC04) is used.

4.4 Discussion

Significant overestimation of the phantom's CT opacity was observed whenever it was enclosed by a radiopaque contrast fluid sheath. The overestimation was observed for kernels with and without beam hardening correction, and with and without submersion in water. Beam hardening most likely lies at the root of this problem, but surprisingly has the opposite effect as the "cupping" described by [Brooks and Chiro, 1976], leading to an increase rather than decrease in observed CT numbers. The observed overestimation is not due to partial volume effects since the ROI was chosen in a uniform region of the phantom, separated from its edges by much more than the scanner's point spread radius [Dore and Goussard, 1997].

Femoral cortical bone may reach a thickness exceeding 6.0mm for males, decreasing to 2.5mm for elderly females [Bousson et al., 2000] and has opacity and geometry comparable to the contrast fluid used in this study. The trabecular bone of older patients often has opacities similar to or lower than the Perspex rod used in this study [Ciarelli et al., 1991]. We therefore expect to find comparable behaviour in clinical CT scans.

Perspex has a known CT number of 122 HU ± 5 HU [Brown et al., 2008]. We measured mean CT numbers of 156 HU and 181 HU using the standard FC30 bone kernel to image the rod surrounded by a 2.5mm and 6.5mm sheath respectively while submerged in water. These values increased to 199 HU and 233 HU when the targets were not submerged. When the bone kernel was substituted for the beam hardening correcting FC04 kernel we measured values of 146 HU and 162 HU when submerged but lowering to 125 and 146 HU in air. While the per-voxel noise of the sharp bone kernel had a wide distribution, the standard error of the mean was small and this did not invalidate it as an estimator of mean tissue density.

The observed CT opacity bias, however, will cause errors in the derived density of trabecular bone. The magnitude of this error can be assessed by using the well-established linear relationship between HU and trabecular bone ash density reported by [Mosekilde et al., 1989], namely

$$\rho = 0.0004x + 0.063$$

, where x is the CT number in HU and ρ is the apparent ash density in g/cm^3. Inserting the reference value of 122 HU we obtain an estimated ash density of 0.112 g/cm^3. When instead using 160 HU – the value one could encounter with a sheath of 3mm of cortical bone - we obtain 0.127 g/cm^3 – an overestimation of 13.6%.

We assess the resulting error in estimated elastic properties by using the power relationship described by Keller [Keller, 1994],

$$E = 10.5\rho^{2.29}$$

, where ρ is apparent ash density in g/cm^3 and E is the Young's modulus in GPa. Inserting the values computed above, we obtain Young's moduli of 69.5 MPa and 93.1 MPa respectively – an overestimation of 33.9%. Due to the higher exponent relating cortical ash density to Young's modulus we expect even larger errors if comparable overestimation occurs in cortical bone regions [Schaffler and Burr, 1988].

Such inaccuracies might render derived FEM models inaccurate, and prove disastrous where absolute values of stress and strain are computed. In these cases we suggest using dual energy QCT and/or including a calibration phantom in the same instantaneous field of view as the target. Relative differences between FEM configurations might be less affected. As patient specific FEM modelling becomes ever more popular this issue requires careful model validation.

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