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# Finite element analysis for prediction of adolescent bone strength: micro vs continuum models

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## 1. Introduction

Adolescence is a crucial period for bone growth, and many factors during this period can influence bone quality and the intensity of peak bone mass, e.g., nutritional intakes and physical activity (Chevalley et al. 2011). Carrying out studies in this specific population may help to improve our knowledge of the evolution of bone quantity, quality and biomechanical parameters during puberty, and to better understand the mechanisms of fractures.

HR-pQCT (XtremeCT, Scanco Medical AG, Switzerland) is a scanner used in research to assess *in vivo* densities, bone microarchitecture and biomechanical properties with a High Resolution (isotropic voxel:  $82\mu\text{m}^3$ ). HR-pQCT-based micro Finite Element Analysis (FEA) predicts accurately the mechanical properties and risk of fracture (Laib et al. 1998; Varga et al. 2010; Pistoia et al. 2002), even with a linear homogenized FEA (Hosseini et al. 2017). However, these simulations suffer from discontinuities, are time consuming and device dependent.

The aim of this study is then to validate a continuum FE model by comparison with the micro FEA from HR-pQCT on adolescent cohort.

## 2. Materials and methods

The VITADOS cohort (NCT01832623) is composed of 100 healthy teenagers (50 boys, and 50 girls) from 10 to 18 years of age with five pubertal Tanner stages.

### 2.1. Hr-pQCT based evaluation and $\mu\text{FE}$

Volumetric bone mineral densities and micro-architecture were measured at both distal radius and tibia from a stack of 110 slices corresponding to  $\sim 9\text{mm}$  length of bone.

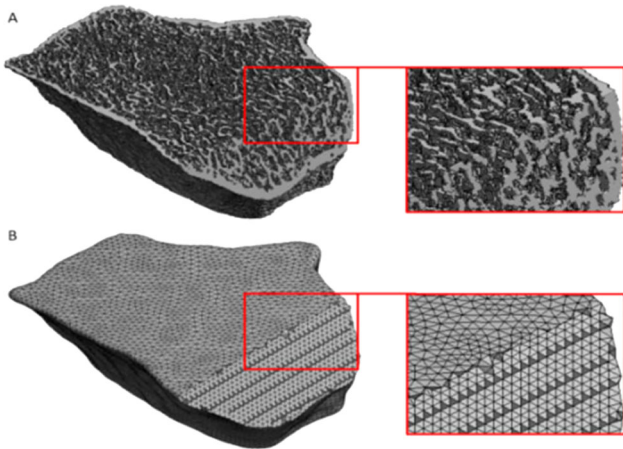
Micro-FE ( $\mu\text{FE}$ ) models of the radius and tibia were created directly from the segmented images using the software dedicated (IPL). Briefly, a voxel-based model converts each voxel of bone tissue into an equally sized hexahedral element (8 nodes 1 integration point) leading to 2 million elements for the radius, 5 million elements for the tibia. A semi-automatic segmentation was used to differentiate cortical from trabecular bone. Material properties were chosen isotropic and elastic. Young's modulus of 20 GPa and 17 GPa were assigned to cortical and trabecular bone elements respectively, with a Poisson's ratio of 0.3 (Vilayphiou et al. 2010). A compression test was simulated with a uniaxial displacement of 1% of bone length applied on the distal face of the bone while the nodes located at the proximal face were fully constrained (Pistoia et al. 2002).

The failure load ( $F_{\text{ult}}$ ) is calculated using Pistoia criterion, adapted by (Vilayphiou et al. 2010; Boutroy et al. 2007) with 2% of the bone tissue exceeding the yield strain of 3500  $\mu$  strain to assess the failure load. The  $\mu\text{FE}$  models were the same for radius and tibia analysis.

### 2.2. cFE modelling

The continuum FE model is built, using the raw dicom, using 3D-Slicer (stl data), then the model is meshed using commercial software (Ansys) with quadratic tetrahedral elements (10 nodes, 4 integration points). After a first parametric study, the size of the elements varied between 0.4 and 0.6 mm and the number of elements is approximately 300000 for radius and 600000 for tibia. Figure 1 represents the difference between continuum FE model (cFE) and micro FE model ( $\mu\text{FE}$ ).

In order to retrieve bone structure a material mapping was performed, i.e., the assignment of Young's modulus to each element using the grey levels of the images. To do so, a calibration phantom with calcium hydroxyapatite equivalent concentrations of 0, 100, 200, 400, and 800  $\text{mg}/\text{cm}^3$  was used to establish the following linear relationship between CT Hounsfield units



**Figure 1.** (A)  $\mu$ FE: only bone was meshed. (B) cFE: all the section was meshed, including bone marrow.

(HU) and calcium hydroxyapatite equivalent density ( $\rho_{\text{QCT}}$ ) in  $\text{g}/\text{cm}^3$ :  $\rho_{\text{QCT}} = 0.041623 + 0.0003555 \text{ HU}$ .

Several density-elasticity laws were tested; however, their results were either overestimated (Keyak et al. 1994; Morgan et al. 2003) or underestimated (Keller 1994; Carter et al. 1977). Then, two previously established density-elasticity relationships (Eqs. (1)–(2)) were combined to assign inhomogeneous linearly isotropic material properties to the finite element models, where Kaneko is assigned to the trabecular bone and Carter affected to the cortical bone.

$$E = 10880 \rho_{\text{ash}}^{1.61}, \text{ Kaneko et al. (2004)} \quad (1)$$

$$E = 2875 \rho_{\text{app}}^3; \text{ Carter and Hayes (1977)} \quad (2)$$

where  $E$  is the elastic modulus (MPa) and  $\rho$  is the density ( $\text{g}/\text{cm}^3$ ).  $\rho_{\text{QCT}}$  was converted to  $\rho_{\text{ash}}$  using

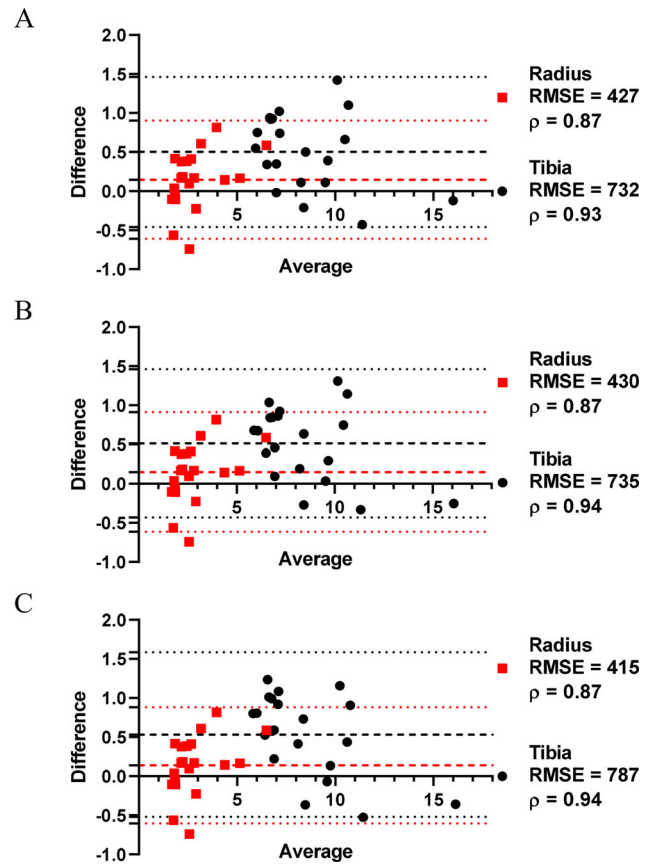
$$\rho_{\text{ash}} = 0.0698 + 0.839 \rho_{\text{QCT}} \text{ (Laib et al. 1998)}$$

The ratio  $\rho_{\text{ash}}/\rho_{\text{app}}$  was considered equal to 0.6 (Schileo et al. 2008).

The passage from one law to another is defined by a density threshold, obtained by the ratio of Cortical Area over Total Area. The percentage of the cortical compartment in the whole bone allows separating the densities of the elements of the cortical from the trabecular densities.

This value was calculated from 4 adolescents in each puberty tanner stage group (3 Tanner groups for boys and 2 Tanner groups for girls). Three categories of threshold density were calculated: (A) mean threshold, (B) threshold-gender and (C) threshold-tanner-gender.

The boundary conditions are the same used for  $\mu$ FE.



**Figure 2.** Bland Altman of  $\mu$ FE and cFE methods using (A) mean threshold, (B) gender threshold, and (C) Tanner-gender threshold of 20 radius and tibia.

### 2.3. Statistical analysis

Non parametric Spearman correlation coefficients ( $\rho$ ) were used to evaluate the correlation. Accuracy between the two methods was expressed by the root mean square error (RMSE).

## 3. Results and discussion

The cFE and  $\mu$ FE were carried out for 20 adolescents whose density thresholds were calculated. The cFE were ten-times faster than  $\mu$ FE.

The results were roughly equivalent for all the three density thresholds (Bland-Altman graphs, Figure 2). The failure load by cFE and  $\mu$ FE was significantly correlated for radius ( $\rho = 0.87$ ) and tibia ( $\rho = [0.93-0.94]$ ), and no statistical difference in RMSE values for all the thresholds. Differences were within the mean of difference (tibia:  $[0.50 - 0.53]$ , radius:  $[0.14 - 0.15]$ )  $\pm 1.96$  SD (tibia:  $[-0.52 - 1.58]$ , radius:  $[-0.61 - 0.91]$ ). The mean threshold is then preferred, as no requirement of differentiation by Tanner stage or gender. To use the combination of two laws (one for cortical, one for

trabecular bone), those results required the microarchitecture analysis (%cortical bone).

#### 4. Conclusion

The cFE models, which do not suffer from discontinuities, showed satisfactory agreement with  $\mu$ FE results. Using the mean threshold obtained on 20 adolescents (radius:  $0.581 \text{ g/cm}^3$ , tibia:  $0.657 \text{ g/cm}^3$ ), the methodology will be validated on 80 remaining subjects of the cohort. HR-pQCT remains useful to assess bone micro-architecture and compartmental densities, but the cFE analysis open to new investigations on adaptive behaviour law.

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**KEYWORDS** FE model; material law; bone Failure load; density threshold; adolescents

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