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Validation of density–elasticity relationships for finite element modeling of human pelvic bone by modal analysis

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ABSTRACT

In total hip arthroplasty and particularly in revision surgery, computer assisted pre-operative prediction of the best possible anchorage strategy for implant fixation would be a great help to the surgeon. Computer simulation relies on validated numerical models. In the current study, three density–elasticity relationships (No. 1–3) from the literature for inhomogeneous material parameter assignment from CT data in automated finite element (FE) modeling of long bones were evaluated for their suitability for FE modeling of human pelvic bone. Numerical modal analysis was conducted on 10 FE models of hemipelvic bone specimens and compared to the gold standard provided by experimental modal analysis results from a previous *in-vitro* study on the same specimens. Overall, calculated resonance frequencies came out lower than measured values. Magnitude of mean relative deviation of numerical resonance frequencies with regard to measured values is lowest for the density–elasticity relationship No. 3 (−15.9%) and considerably higher for both density–elasticity relationships No. 1 (−41.1%) and No. 2 (−45.0%). Mean *MAC* values over all specimens amount to 77.8% (No. 1), 78.5% (No. 2), and 83.0% (No. 3). *MAC* results show, that mode shapes are only slightly influenced by material distribution. Calculated resonance frequencies are generally lower than measured values, which indicates, that numerical models lack stiffness. Even when using the best suited (No. 3) out of three investigated density–elasticity relationships, in FE modeling of pelvic bone a considerable underestimation of model stiffness has to be taken into account.

1. Introduction

In total hip arthroplasty and particularly in cases of acetabular revision surgery, general guidelines for pre-operative planning are missing. However, prediction of the best possible anchorage strategy for implant fixation is desirable for surgeons. Computer simulation such as Finite Element Analysis (FEA) can be a valuable tool in decision making even when complex structures and non-linear material and contact behavior are concerned. However, simulations of real effects, such as stress distributions, implant-bone micromotions or periprosthetic bone remodeling demand detailed models of bony structures.

The state of the art in FE modeling of bone includes automatic material parameter assignment based on CT data (Taddei et al., 2004), but, functional relationships between CT values and material parameters were developed and validated only for long bones (Helgason et al., 2008a; Taddei et al., 2007; Schileo et al., 2007; Eberle et al., 2012). It is known, that the cortical shell of pelvic bone plays a central role in the load transfer across the human hip (Dalstra and Huiskes, 1995). Whatsoever, to date no experimental data is available on material properties of pelvic cortical bone (Helgason et al., 2008a). Thus, material parameter assignment based on other than pelvic bone samples was used in the past (Anderson et al., 2005; Dalstra et al., 1995; Kluess et al., 2009).

Principally, experimental validation is required, before clinical conclusions can be drawn from computer simulations (Viceconti et al., 2005). Schileo et al. (2007) investigated validity of density–elasticity relationships from the literature (Carter and Hayes, 1977; Keller, 1994; Morgan et al., 2003) for FE modeling of the femur and identified Morgan et al. (2003) as the best choice. This finding was later confirmed by Sitzer et al. (2011) and Eberle et al. (2012).

Since cortical shell is known to be much thinner at the pelvis as opposed to the femur, it is questionable, whether the method of FE modeling established for long bones can as well be adopted for modeling of pelvic bone. Thus, in the current study it was questioned,

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Nomenclature

E	Young's modulus
ν	Poisson's ratio
ρ	density
mass	mass
V	volume
CT	CT values in Hounsfield units (HU)
R^2	coefficient of determination
RMSE	root-mean-square error
MAC	modal assurance criterion
Φ	normalized eigenvector
RD	relative deviation
ω	resonance frequency
i, k, n	number of items

if one of three density–elasticity relationships (Carter and Hayes, 1977; Keller, 1994; Morgan et al., 2003), that were investigated by Schileo et al. (2007) on the femur, can be confidently adopted for automatic FE modeling of pelvic bone. Experimental modal analysis results of 10 hemipelvic bone specimens from a previous study serve as a gold standard (Neugebauer et al., 2011).

2. Materials and methods

2.1. Experimental modal analysis

In Hobatho et al. (1991), Couteau et al. (1998) and Taylor et al. (2002) modal analysis is described as an effective method for validation of FE models of femur and tibia. Experimental modal analysis (EMA) is particularly suitable as a measurement method on fresh cadaveric specimen due to the simple experimental setup and the short period of time required for measurements. Taylor et al. (2002) validated an FE-model of the femur by experimental modal analysis using an impact hammer and an unidirectional piezoelectric accelerometer at 16 measurement points. However, for methodological reasons Conza and Rixen (2006) recommended laser vibrometry as method of choice for experimental modal analysis on fresh specimen of cadaveric hemipelvic bone. In a previous work by Neugebauer et al. (2011) 10 fresh-frozen human hemipelvic bones were subjected to experimental modal analysis using 3D laser vibrometry. The applied test setup and measuring procedure are described in detail elsewhere (Neugebauer et al., 2011).

2.2. CT scanning

The same 10 fresh-frozen human hemipelvic bones (Neugebauer et al., 2011; Table 1) were μCT scanned (Cone beam method: 200 kV, 150 mA; plane detector: 2024 × 2024 pixels; output data: image stacks, 16bit TIFF; voxel resolution: 105–130 μm) in fresh-frozen state prior to EMA measurements. A simple self-made phantom containing compartments of air (−1000 HU) (Whitehouse, 2005), H_2O (0 HU) (Whitehouse, 2005), and PMMA (130 HU) (Parodi et al., 2007) was placed alongside each specimen in field of view of the μCT scanner, respectively. Thus, CT grayscale values were subsequently calibrated according to the scale of Hounsfield units (HU).

2.3. Subject-specific FE modeling

FE models were created from μCT image stacks of 10 hemipelvic bone specimens. Automatic segmentation of bony contours was carried out using MIMICS©v12, MATERIALISE NV, Leuven, Belgium. In ANSYS ICEM CFD©v12.0.1, ANSYS Inc., Canonsburg, PA, USA 10-node tetrahedral volume meshes were generated.

2.4. Material parameter assignment

The open source software BoneMat is readily available for automatic assignment of inhomogeneous material parameters of elasticity, also known as the Young's modulus E, and density ρ on FE meshes of bony structures (Taddei et al., 2007). Thereby, an average Young's modulus E is individually calculated for each finite element of the mesh based on local CT values. Thus, just by applying inhomogeneous material parameters to an unstructured finite element mesh a good approximation of cortical shell is achieved by the stiffening effect of higher values at the surface layer of the mesh. The BoneMat algorithm requires the user to select the applicable functional relationships from the literature and type them into the program. Two of the density–elasticity relationships investigated in the current study originate from experimental measurements on pooled samples (Carter and Hayes, 1977; Keller, 1994) and one on samples from the femoral neck (Morgan et al., 2003; Helgason et al., 2008a).

The investigated equations are listed below (Carter and Hayes, 1977; Eq. (1); Keller, 1994; Eq. (2); Morgan et al., 2003; Eq. (3)):

$$E_1 = 3.79\rho_{app}^{3} \tag{1}$$

$$E_2 = 10.50\rho_{ash}^{2.29} \tag{2}$$

$$E_3 = 6.85\rho_{app}^{1.49} \tag{3}$$

with Young's modulus E in GPa and density ρ (g/cm^3). The formula (1) by Carter and Hayes (1977) was used, as modified by Schileo et al. (2007). According to Schileo et al. (2007) the influence of strain rate was considered negligible in the current study since vibrational testing induces only very low strains into the material.

For the functional relationships according to Carter and Hayes (1977) and Morgan et al. (2003) (Eqs. (1) and (3)) it is necessary to pre-calculate apparent density ρ_{app} for which the following functional relationship according to Taylor et al. (2002) was applied:

$$\rho_{app} = b \cdot CT \tag{4}$$

with apparent density ρ_{app} and coefficient b in kg/m^3 and CT values in HU.

For calculation of coefficient b in Eq. (4) Taylor et al. (2002) used two support values, one representing water ($CT_{H_2O} = 0$ HU, ρ_{app,H_2O} 0 kg/m^3) and the other one representing cortical femoral bone of maximal effective density (CT_{max}=2025 HU, $\rho_{eff,max}$=1940 kg/m³) as the upper limit. In the current study, for determination of the upper limit, hemipelvic bone specimen SP10 (Neugebauer et al., 2011) was used, since, according to visual examination of μCT image data, it presented with the strongest cortical shell structure ($CT_{max,SP10}$=1628 HU, $\rho_{eff,max,SP10}$=2033 kg/m^3) out of all specimens. Consequently, coefficient b was evaluated to 1.249 kg/m^3 (compared to Taylor et al. (2002): b=0.958 kg/m^3).

Application of the Eq. (2), according to Keller (1994), requires conversion of apparent density to ash density ρ_{ash} using the following functional relationship according to Schileo et al. (2007):

$$\rho_{ash} = 0.60\rho_{app} \tag{5}$$

For calculation of the material parameter density ρ in the FE model, the following relationship between CT values and effective density ρ_{eff} according to Rho et al. (1995) was used:

$$\rho_{eff} = a \cdot CT + 1000\ \text{kg/m}^3 \tag{6}$$

with effective density ρ_{eff} and coefficient a in kg/m^3 and CT values in HU. This formula requires the coefficient a to be specified. In contrast to the value a=0.523 kg/m^3 proposed by Rho et al. (1995), Taylor et al. (2002) had evaluated the coefficient to a=0.464 kg/m^3 for femoral bone. According to Taylor et al. (2002), the coefficient a had to be identified for the currently investigated group of

Table 1
Pre-test specimen weights and calculated weights of numerical models after calibration.

Specimen	SP01	SP02	SP03	SP04	SP05	SP06	SP07	SP08	SP09	SP10
$mass_{specimen}$ (g) (Neugebauer et al., 2011)	326	340	326	313	495	514	263	476	283	527
$mass_{model,calibrated}$ (g)	368	330	335	315	471	494	277	477	285	519