CHAPTER 3

Patient specific bone and joints modeling and tissue characterization derived from medical images

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Abstract

The objective of the paper is to address the methodology developed to model bone and joints with individualized geometric and material properties from medical image data. An atlas of mechanical properties of human bone has been investigated demonstrating individual differences. From these data, predictive relationships have been established between mechanical properties and quantitative data derived from measurements on medical images. Subsequently, geometric and numerical models of bones with individualized geometrical and mechanical properties have been developed from the same source of image data. The advantages of this modeling technique are its ability to study the ‘patient’ specificity. This should be of importance for quantifying bone and joint deformities and performing individualized preoperative planning surgery or orthopaedic treatment. In the same way, the efficiency of orthopaedic treatment with customised orthese or mechanical behavior of implant in bone could be evaluated. Results would suggest improvement or development of new design.

1 Introduction

Bone and joints are complex structures in their geometry and material properties. In order to assess bone deformities or understand the mechanical behavior of the implant in bone, it is necessary to model the physiological conditions of the bone and joints. The only way to obtain patient specific geometry will be to assess medical images such as magnetic resonance imaging (MRI), computed tomography (CT) and X-ray. This step has been almost achieved by most of the researchers. The other
important factor to be considered is the mechanical properties of the patient. This last step is still poorly investigated as it needs basic research on the assessment of bone and joints material properties using mechanical testing and tissue characterization derived from medical images. Mechanical properties have been investigated for three decades but the data is not always of help for the modeling purpose.

This chapter will address the methodology we have developed in order to model bone and joints with appropriate geometric and mechanical properties derived from medical imaging. Medical imaging systems such as MRI, CT, X-rays are commonly used to evaluate musculo-skeletal disease. The main advantages of MRI and CT techniques are the 3D geometry assessment and the tissue characterization derived from pixels grey level density. That is why these two techniques are mainly in use.

Numerical methods are used for solving physical and mechanical engineering problems. These numerical methods are appropriate for modeling such complex systems as human bone and joints. Literature review demonstrated the extensive use of finite element modeling in biomechanics [1–3]. According to our knowledge, the first two dimensional finite element model derived from medical image were obtained by digitizing radiography [4], and a three dimensional finite element model derived from digitized CT scans [5]. Then, extensive finite element models are obtained from CT data with most often a lack of description of the method allowing to model the geometry. One should note that little attention is given to the acquisition parameters and their consequences to the accuracy of bone modeling. Some authors proposed specific protocol to optimize the reconstruction of a specific long bone (femur) from CT data [6] meanwhile others focus on the hardware and software parameters acquisition influence on the CT image assessment qualitatively and quantitatively [7]. Automatic FE generation have been developed using CT scans voxels [8, 9]. The voxels are converted directly in elements of equal size. Problems occurring with this technique are (1) their limitation in the accuracy of the model at the geometric and material boundaries and (2) the number of elements generated which would require specific algorithms of resolution. Besides these extensive numerical models, most of models are derived from CT data and a few from MRI. Few consider appropriate material properties as they are mostly issued from literature (data or relationships). When experimental data do not confirm numerical simulation, material properties data are assessed experimentally by mechanical testing. This can only be performed on cadaveric specimens. As shown by the previous review, human bone has a non uniform geometry and a heterogeneous structure. One may expect differences from bone to bone and among individuals. In order to consider these intrinsic differences, it was necessary to develop a modeling technique describing appropriately and simultaneously the geometric and material properties. The methodology we have developed is based on a semi-automatic generation of a three dimensional geometric model of bone and joints anatomy derived from medical imaging CT or MRI data. Predictive relationships obtained from previous work demonstrated a significant correlation between the material properties and quantitative measurements derived from imaging techniques [10, 11]. Then, from the same source of medical imaging data, numerical models with individualized geometric and mechanical properties were developed.
2 Methods

The different steps of the methodology would be first to be able to decode the native medical images in order to assess the geometry and the density grey level. It is important to assess native grey level density in order to assess biological material tissue characterization.

The second step would be to determine the mechanical properties of bone or soft tissue and correlate these mechanical properties with bone and joints tissue characterization derived from medical images.

2.1 Assessment of geometry via medical imaging techniques

There are different modalities of medical imaging techniques to explore bone and joints: X-ray radiographs, CT scanner, MRI. The first two techniques are based on absorption of X-rays, and are often used to diagnose bone disease. Meanwhile, the third is based on proton resonance and is more dedicated to soft tissue such as ligaments, cartilage besides bone structures are also visualised.

In order to assess the anatomical data, we have developed a pre-processing medical image CT and MRI [12].

The different steps of the image processing were (1) to decode the stack of medical image data representing the three dimensional of bone and joints structure, (2) to perform an edge detection after a threshold process and (3) to build geometric entities of the bones with creation of an output data file in a neutral format or an IGES format.

2.1.1 Medical image processing

The standard exchange format file of the medical image derived from American College of Radiology and National Electric Manufacturers Association (ACR-NEMA) specifications is Digital Imaging COmmunication in Medicine (DICOM). Once the medical images are decoded, image processing can be performed. A pixel is coded on 12 bits i.e. pixel values varies from 0 (white) to 4096 (black). The next step is to perform the threshold process. It consists in giving threshold values based on the histogram representing the distribution of pixel values. As a result only anatomical contours of interest are visualized on the binary image (only two levels 0 and 1, white and black). Then segmentation by region is used, the edge detection allowed the outlines of the anatomical structure to be obtained. The edge detection consists in Hermite parametric cubic curves interpolation. Their formulation is obtained from a parametric cubic equation (eqn (1)) and geometric constraints on two ending points.

\[
P(u) = [x(u), y(u), z(u)],
\]

\[
P(u) = \sum_{k=0}^{3} a_k u^k, \quad u \in [0, 1].
\]
Their final expression form with blending functions are expressed in eqn (2).

\[ P(u) = P_0(1-3u^2+2u^3) + P_1(3u^2-2u^3) + P'_0(u^2-u^3) + P'_1(-u^2+u^3) \]  \( (2) \)

The geometric continuity of two curves is obtained by defining same tangent at the ending points of the first curve and the first point of the second curve as illustrated in fig. 1.

Surfaces connecting these curves were Hermite bicubic polynomials defined by eqn (3).

\[ P(u) = \sum_{i=0}^{3} \sum_{j=0}^{3} a_{ij}u^i v^j, \quad u, v \in [0, 1]. \]  \( (3) \)

The different steps of a 3D geometrical reconstruction in vivo of a human hip from CT data is illustrated in fig. 2. The pre- and post-processor Patran (MSC.Software) is used for the visualization of the geometric entities written in a neutral file.

Once the geometrical data was obtained, finite element meshing was performed via any commercial software.

In order to model realistic bone anatomy, it is necessary to associate the appropriate mechanical properties of the bone.

2.2 Mechanical properties of human bone

Mechanical properties of bone have been studied for over three decades in order to understand the mechanical behavior of bone in the process of fracture risk, repair
Computational Modeling of Tissue Surgery

and bone related disease. Besides, few data are available or insufficient for human bone modeling. Human bone is highly heterogeneous and anisotropic material. It can be compared to composite materials; it is made of two different tissue, spongious bone (high porosity) and cortical bone (compact bone), depending on the anatomical location. Bone was assumed to have an orthotropic behavior which stiffness matrix containing elastic constants were defined by eqn (4). By reversing the stiffness matrix, the compliance matrix allowed the elastic properties in the three axes of symmetry of the crystal to be determined eqn (5).

\[
[C_{ij}] = \begin{bmatrix}
C_{11} & C_{12} & C_{13} & 0 & 0 & 0 \\
C_{21} & C_{22} & C_{23} & 0 & 0 & 0 \\
C_{31} & C_{32} & C_{33} & 0 & 0 & 0 \\
0 & 0 & 0 & C_{44} & 0 & 0 \\
0 & 0 & 0 & 0 & C_{55} & 0 \\
0 & 0 & 0 & 0 & 0 & C_{66}
\end{bmatrix}, \tag{4}
\]

\[
[S_{ij}] = \begin{bmatrix}
\frac{1}{E_1} & -\frac{v_{12}}{E_1} & -\frac{v_{13}}{E_3} & 0 & 0 & 0 \\
-\frac{v_{12}}{E_1} & \frac{1}{E_2} & -\frac{v_{23}}{E_3} & 0 & 0 & 0 \\
-\frac{v_{13}}{E_1} & -\frac{v_{23}}{E_2} & \frac{1}{E_3} & 0 & 0 & 0 \\
0 & 0 & 0 & \frac{1}{\alpha_{31}} & 0 & 0 \\
0 & 0 & 0 & 0 & \frac{1}{\alpha_{12}} & 0 \\
0 & 0 & 0 & 0 & 0 & \frac{1}{\alpha_{12}}
\end{bmatrix}, \tag{5}
\]

where $E_i$ is the Young’s moduli in the direction $i$, $G_{ij}$ is the Shear moduli in plane $i-j$ and $v_{ij}$ is the Poisson’s ratio stress and strain respectively in directions $i, j$.

Experimentally, an ultrasonic transmission technique was used to assess bone material properties. Different wave propagation techniques were used for velocities measurements of bone tissue. Bulk velocities were measured for cortical bone, and then elastic constants were obtained. Cancellous bone is porous and highly heterogeneous compared to cortical bone. Homogeneous volume needed to be assumed and bar waves were used, allowing to assess directly the elastic properties Ashman et al. [13, 14]. An atlas of mechanical properties was performed in order to obtain a database of material properties of different types of bone (femur, tibia, mandible, patella, lumbar spine, scapula) Ho Ba Tho et al. [11, 15], Mansat et al. [16]. The range of values (minimum-maximum, median, average and standard deviation) of the mechanical properties of the different type of cancellous bone are summarized in table 1.

The range of values (minimum-maximum, median, average and standard deviation) of the mechanical properties of the different type of cancellous bone are summarized in table 2.

Differences of properties between bones and between subjects suggested the use of appropriate characteristic value of properties. When differences are not found to be significant, an average value of properties can be considered as the characteristic
Table 1: Minimum and maximum values, the mean and the average with standard deviation of the mechanical properties of cancellous bone.

<table>
<thead>
<tr>
<th>Cancellous bone</th>
<th>$E_1$ (MPa)</th>
<th>$E_2$ (MPa)</th>
<th>$E_3$ (MPa)</th>
<th>$\rho$ (kg/m$^3$)</th>
<th>$\sigma$ (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distal femur</td>
<td>34–2412</td>
<td>39–2037</td>
<td>213–4419</td>
<td>81–883</td>
<td>0.18–19</td>
</tr>
<tr>
<td></td>
<td>760 ± 589</td>
<td>707 ± 555</td>
<td>1698 ± 993</td>
<td>395 ± 180</td>
<td>5.94 ± 4.13</td>
</tr>
<tr>
<td>Proximal humerus</td>
<td>82–861</td>
<td>102–975</td>
<td>249–1719</td>
<td>117–488</td>
<td>0.30–7.12</td>
</tr>
<tr>
<td></td>
<td>397 ± 215</td>
<td>438 ± 245</td>
<td>813 ± 401</td>
<td>255 ± 91</td>
<td>2.71 ± 2.0</td>
</tr>
<tr>
<td>Lumbar spine</td>
<td>75–659</td>
<td>73–730</td>
<td>375–1939</td>
<td>132–368</td>
<td>0.66–6.23</td>
</tr>
<tr>
<td></td>
<td>292 ± 192</td>
<td>321 ± 218</td>
<td>1057 ± 571</td>
<td>242 ± 96</td>
<td>3.14 ± 1.57</td>
</tr>
<tr>
<td>Patella</td>
<td>286–3553</td>
<td>221–3183</td>
<td>569–4925</td>
<td>391–1210</td>
<td>1.07–17.45</td>
</tr>
<tr>
<td></td>
<td>1650</td>
<td>873</td>
<td>2764</td>
<td>703</td>
<td>6.47</td>
</tr>
<tr>
<td></td>
<td>1484 ± 794</td>
<td>1197 ± 837</td>
<td>2801 ± 1363</td>
<td>731 ± 222</td>
<td>7.71 ± 4.56</td>
</tr>
<tr>
<td>Posimal femur</td>
<td>24–2492</td>
<td>23–2398</td>
<td>105–3669</td>
<td>73–857</td>
<td>0.25–16.82</td>
</tr>
<tr>
<td></td>
<td>713 ± 783</td>
<td>769 ± 734</td>
<td>1267 ± 888</td>
<td>350 ± 190</td>
<td>4.83 ± 3.87</td>
</tr>
<tr>
<td></td>
<td>171</td>
<td>186</td>
<td>604</td>
<td>181</td>
<td>1.88</td>
</tr>
<tr>
<td></td>
<td>202 ± 154</td>
<td>232 ± 180</td>
<td>769 ± 534</td>
<td>198 ± 94</td>
<td>2.58 ± 2.22</td>
</tr>
</tbody>
</table>

value. Conversely, when differences are found to be significant, a range (minimum and maximum) would be more accurate to characterize the properties. The range of values of mechanical properties for cortical bone and cancellous bone from eight subjects are summarized in table 3.

Ultimate strength of cortical specimen was not measured, literature review gave typical values around 150 MPa. Relationships between mechanical properties of cortical and cancellous human bone is illustrated in fig. 3.

The following statements were found according to the atlas of mechanical properties measured experimentally:

- the mechanical properties of cortical bone vary around the periphery with a small variation around 10% and do not vary along the length
- the mechanical properties of cancellous bone vary around the periphery (around a factor of 2) and along the length (around a factor of 3 to 5)
- cortical and cancellous bone are orthotropic materials, however, the degree of anisotropy of cancellous bones is higher than that of cortical bones
- the mechanical properties are different from bone to bone for cortical and cancellous bones
Table 2: Minimum and maximum values, and the average with standard deviation of the mechanical properties of cortical bone.

<table>
<thead>
<tr>
<th>Cortical bone</th>
<th>$E_1$ (GPa)</th>
<th>$E_2$ (GPa)</th>
<th>$E_3$ (GPa)</th>
<th>$\rho$ (kg/m$^3$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femur</td>
<td>7.4–16.2</td>
<td>11.8</td>
<td>11.7 ± 1.9</td>
<td>1581–1996</td>
</tr>
<tr>
<td></td>
<td>8–17</td>
<td>12.2</td>
<td>12.3 ± 2.0</td>
<td>1791</td>
</tr>
<tr>
<td></td>
<td>16.7–24.3</td>
<td>19.8</td>
<td>19.9 ± 2.7</td>
<td>1821 ± 183</td>
</tr>
<tr>
<td>Humerus</td>
<td>6.9–14.4</td>
<td>10.8</td>
<td>10.7 ± 2.0</td>
<td>1552–2075</td>
</tr>
<tr>
<td></td>
<td>7.3–16.5</td>
<td>11.8</td>
<td>11.6 ± 2.1</td>
<td>1790</td>
</tr>
<tr>
<td></td>
<td>13.8–25.5</td>
<td>20.5</td>
<td>20.0 ± 2.7</td>
<td>1779 ± 153</td>
</tr>
<tr>
<td>Mandible</td>
<td>8.6–14.6</td>
<td>11.7</td>
<td>11.6 ± 1.3</td>
<td>1731–1976</td>
</tr>
<tr>
<td></td>
<td>8.6–15.5</td>
<td>12.1</td>
<td>12.4 ± 2.1</td>
<td>1863</td>
</tr>
<tr>
<td></td>
<td>13.6–27</td>
<td>20.5</td>
<td>20.4 ± 2.9</td>
<td>1858 ± 74</td>
</tr>
<tr>
<td>Tibia</td>
<td>8.4–15.3</td>
<td>11.7</td>
<td>11.7 ± 1.3</td>
<td>1616–2063</td>
</tr>
<tr>
<td></td>
<td>8.8–15.4</td>
<td>12.2</td>
<td>12.2 ± 1.4</td>
<td>1859</td>
</tr>
<tr>
<td></td>
<td>15.9–24.6</td>
<td>20.8</td>
<td>20.7 ± 1.9</td>
<td>1840 ± 11</td>
</tr>
</tbody>
</table>

Table 3: Range of values of mechanical properties obtained from eight subjects. Minimum, maximum and median values are given.

<table>
<thead>
<tr>
<th>Bone</th>
<th>Cortical</th>
<th>Cancellous</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_{\text{axial}}$ (GPa)</td>
<td>14–27</td>
<td>0.011–3.12</td>
</tr>
<tr>
<td></td>
<td>20</td>
<td>0.961</td>
</tr>
<tr>
<td>$E_{\text{tangential}}$ (GPa)</td>
<td>7–17</td>
<td>0.023–1.5</td>
</tr>
<tr>
<td></td>
<td>12</td>
<td>0.341</td>
</tr>
<tr>
<td>$E_{\text{radial}}$ (GPa)</td>
<td>7–16</td>
<td>0.024–1.5</td>
</tr>
<tr>
<td></td>
<td>11</td>
<td>0.301</td>
</tr>
<tr>
<td>$\rho$ (kg/m$^3$)</td>
<td>1545–2118</td>
<td>55–774</td>
</tr>
<tr>
<td></td>
<td>1840</td>
<td>257</td>
</tr>
<tr>
<td>CT (HU)</td>
<td>1270–1835</td>
<td>72–512</td>
</tr>
<tr>
<td></td>
<td>1560</td>
<td>143</td>
</tr>
<tr>
<td>$\sigma_{\text{ult}}$ (MPa)</td>
<td>–</td>
<td>0.11–11</td>
</tr>
<tr>
<td></td>
<td></td>
<td>3</td>
</tr>
</tbody>
</table>

- the mechanical properties of cortical and cancellous bone are not all equal between subjects but not all necessarily different
- the relationship between axial modulus and density is linear for cortical bone
- for cancellous bones, linear or power fits were found approximately equal
- powers vary from 1.3 to 1.7 for axial modulus versus density and 1.3 to 2.3 for strength versus density.
These results suggest (1) the use of appropriate characteristic values (average or range) of properties for parameterization study or finite element analysis and (2) the use of appropriate relationships of axial modulus, strength and density upon the type of bone.

Finally, these results suggest the consideration of the heterogeneity of bone material properties and the subject specificity when dealing with bone pathology. One way to consider a patient specificity is to develop numerical models with their geometric and mechanical properties.

2.3 Models with individualized geometric and material properties from CT data

In order to associate geometric and mechanical properties, we assume that measurements derived from medical imaging could predict material properties. We have investigated the relationships between CT numbers derived from CT imaging techniques and mechanical properties of bone. The CT number characterize a linear coefficient of attenuation of X-ray within the tissue. For the CT scan, the pixels values are represented by an empirical number called CT number expressed in Hounsfield units (HU) and is defined by the following empirical equation:

\[
CT (HU) = 1000 \frac{CT - CT_{\text{water}}}{CT_{\text{water}} - CT_{\text{air}}},
\]

\[
CT (HU) = 1000 \frac{\mu - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}},
\]
where \( \mu \) is the linear coefficient of attenuation of X-ray within the tissue (cm\(^{-1}\)).

CT number value is dependent on acquisitions parameters, typical values are 0 and -1000 for water and air, respectively.

Table 4 summarizes some predictive relationships between elastic properties and density and CT numbers, Rho et al. [10], Ho Ba Tho et al. [11, 15] of the proximal tibia (upper extremity of the tibia).

Until now, few investigations were performed concerning the cortical bone, as previous relationships [15] were not significant and could not be used to predict cortical bone material properties derived from CT. That is mainly due to the low range of cortical bone material properties investigated and the limitation of the CT technique at that time.

Correlation was found between mechanical properties and CT number of cortical human bone. The data are obtained from multiscale mechanical and morphological properties characterization of cortical bone, Ho Ba Tho et al. [17] (fig. 4). A new ultrasonic technique has been investigated to provide an acoustic image of

Table 4: Predictive relationships for the proximal tibia. Young’s modulus is expressed in MPa and density in kg/m\(^3\) and CT number in HU.

<table>
<thead>
<tr>
<th>Relationships</th>
<th>Coefficient of determination</th>
</tr>
</thead>
<tbody>
<tr>
<td>( E_{\text{axial}} = 0.51\rho^{1.37} )</td>
<td>( R^2 = 0.96 )</td>
</tr>
<tr>
<td>( E_{\text{tangential}} = 0.06\rho^{1.55} )</td>
<td>( R^2 = 0.90 )</td>
</tr>
<tr>
<td>( E_{\text{radial}} = 0.06\rho^{1.51} )</td>
<td>( R^2 = 0.89 )</td>
</tr>
<tr>
<td>( E_{\text{axial}} = 296 + 5.2\text{CT} )</td>
<td>( R^2 = 0.79 )</td>
</tr>
<tr>
<td>( \rho = 144 + 0.916\text{CT} )</td>
<td>( R^2 = 0.80 )</td>
</tr>
</tbody>
</table>

Figure 4: Cartography of CT numbers (HU) and bulk velocities (m/s) of human femur.
the cortical bone reflecting bone heterogeneity and the relation with microstructure (microporosity) has been investigated with an immersion ultrasonic technique Bensamoun et al. [18, 19].

2.4 Mechanical properties of soft tissue

Modeling joints are mainly obtained by MRI data. In these problems, all soft tissues have to be reconstructed such as cartilage, bone, menisci, ACL ligaments for the knee as an example. When non linear static problems are performed, the mechanical properties of the soft tissue are obtained from the literature.

Our methodology in current investigations will consist of assessing mechanical properties of soft tissue during MRI acquisition in order to correlate MRI ROI which depends mainly on different relaxation time T1, T2 with mechanical properties. The major problem would be to define appropriate MRI acquisitions to visualize and assess tissue characterization.

Mechanical properties in vivo can be defined as shown previously, boundary conditions can also be provided by the images (exact location of ligaments and other biological tissue). But it is still a major problem to know the in vivo load and muscles action of the patient. Geometry or morphological data of soft tissue can be assessed from medical image. Then, one would expect that by quantifying the morphology, the mechanical properties could be assessed. Relationships between morphological and mechanical properties of the muscle were investigated in order to provide the forces developed in relation with the muscle and fiber morphology, Bensamoun [20]. These data are of help as one may expect that forces generated are different from one patient to another.

From the same source of images data, a geometric reconstruction was performed as described previously and a finite element model was then performed via a commercial software. A custom made program matching mesh properties has been developed allowing the matching of the material properties distribution measured on the stack of CT image data and their assignment to the elements properties. The program consists in matching the measured mean CT number of a region of interest (ROI) with the characteristic geometric properties of the elements of the mesh. It should be noted that for the 3D mesh, it is necessary to match the technical acquisition parameters of the CT data with that of the protocol used to get the predictive relationships and the 3D mesh. The same methodology may be used for MRI data.

3 Applications

3.1 Modeling of bone and joints of children

The clinical application of the developed methods is to evaluate bone and joint disease in children (congenital dislocation of the hip (hip deformity) [21], clubfoot (foot deformity) [22], rotational abnormalities of the lower limbs [23, 24], scoliosis (spine deformity) [25]. Geometric model of the knee from MRI, allowed to perform
kinematic analysis in vivo and quantification of contact areas using a non linear analysis [26] (figs 5 and 6).

Geometric modeling of child foot bone in vivo derived from MRI [27] allowed to distinguish the osseous nucleus and cartilage anlage (fig. 7). The length of the foot is around 6 cm. The models allowed to quantify the bone deformity and perform preoperative planning surgery.

Finite element models with individualized geometry and material properties in vivo of a vertebral body of two scoliotic patients (same age and sex). They both had a scoliotic deformity at the same level of the lumbar spine (L1). Figure 8 demonstrates clearly the individual difference in the geometry and material properties range and distribution Périé et al. [28].

3.2 Examples of arthroplasty

Arthroplasty consisted in changing the damaged joint (hip, knee, shoulder) by artificial joints. Figure 9 illustrates a numerical model of an acetabular implant of a patient before and after surgery Hinrichs et al. [29]. In this study, significant influence of the assumptions on material properties have been demonstrated by assuming
the material to be isotropic or anisotropic homogeneous, isotropic or anisotropic inhomogeneous.

It should be noted that Von Mises stress is not appropriate to predict risk of failure for anisotropic elastic properties such as bone. But it is often used for these
materials to simplify the interpretation and to represent the six stress components in a generalized ‘stress intensity’ factor Huiskes and Verdonschot [30].

The methodology has been applied to design and evaluate a customized hip implant from medical images. The protocol consists in designing a customized implant by considering the patient specific geometrical and mechanical properties. In parallel, a 3D finite element model is developed to evaluate the design and predict its mechanical behavior in the patient’s femur Barré [31].

4 Discussion

4.1 Accuracy of the geometric modeling derived from medical images

Accuracy of the geometric model depends on the technical protocol acquisition of the medical images and methodology used for processing them. The geometrical errors could be introduced by the spatial resolution of the image (pixel size is equal to the field of view divided by the matrix of pixels) and algorithm of edge detection and interpolation function. A good spatial resolution allowed a better accuracy for the edge detection after the threshold process. Figure 10(a)–(b) illustrates two different spatial resolutions performed on the same image. The other source of error would be the choice of the values for the threshold process (fig. 10(b)–(c)).

For the same image with different values of thresholding may result different geometric results derived from curve interpolation.

Increasing the spatial resolution will increase the patient irradiation dose for CT image, and time examination for MRI. In general, compromise has to be taken between the quality of the images and patient welfare. An acceptable range of spatial resolution would be less than 0.5 mm (pixel value). It is clear that for each anatomical model sensitivity analysis has to be performed in order to define appropriate technical protocol of acquisition.

4.2 Models with individualized geometric and mechanical properties derived from medical images

Quantitative measurements performed on the CT images required a good ‘density’ resolution. In order to achieve that, the technical protocol of acquisition of the images should be optimised. Influence of these parameters on direct measurements

Figure 10: Geometric errors introduced by the spatial resolution (a) compared to (b) and the threshold process performed (b, c).
of CT number have been quantified and showed variation of 10% for the range of 0 to 1200 HU, Ho Ba Tho and Treutenaere [7]. This should be considered when predictive relationships are used to predict elastic properties from literature.

The methodology of bone modeling with individualized geometric and mechanical properties needed to be improved for MRI data, as tissue characterization is not reflecting significantly the material properties. In fact, the grey level of the region of interest is related to the intensity of proton density, which varies significantly with the acquisition parameters. When predictive relationships are not appropriate, the atlas database is used. When needed, this database is increased with other bones using the transmission ultrasonic technique.

### 4.3 Experimental validation

Before using our method in an *in vivo* case (patient case) our methodology has been validated in an *in vitro* case (cadaveric specimen) for customized hip implant Couteau *et al.* [32, 33], tibial implant Estivalèzes *et al.* [34], glenoid implants Baréa *et al.* [35]. Then *in vivo* modeling could be assessed for testing and predicting the short and long term implant behavior. The results can only be validated by clinical follow up and evaluation on the patient.

### 5 Conclusions and perspectives

Relevance of such techniques is their direct clinical application for modeling patient specific bone and joints. This provides an individual diagnosis and preoperative planning surgery or orthopaedic treatment.

Technical protocol of image acquisitions is of importance in the methodology; spatial and density resolution have to be achieved to reduce geometric and material properties errors.

The methodology needs to be improved by:

- the investigation of assessment of material properties of soft tissue derived from MRI
- the investigation of automatic meshing techniques dedicated to bone and joints anatomy; in order to reduce the time consumed in performing a finite element model of bone and joints.

These are challenging investigations for the next decade. From the results of *in vivo* modeling of bone and joints obtained with these techniques, one would expect objectives criterias for planning surgery or orthopaedic treatment and standardization for development of implants or orthoses.

### Acknowledgments

The author would like to acknowledge Texas Scottish Rite Hospital for Children, Fondation pour la Recherche Médicale, INSERM (Institut National de la Santé et de
la Recherche Médicale), CNRS (Centre National de la Recherche Scientifique) for their support. Clinicians from Orthopaedic Surgery and Radiology Departments of TSHR, CHU Purpan Toulouse and Polyclinique St Côme Compiègne are acknowledged for their scientific contribution.

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