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3D patient-specific model of the tibia from CT for orthopedic use

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ABSTRACT

Objectives: 3D patient-specific model of the tibia is used to determine the torque needed to initialize the tibial torsion correction.

Methods: The finite elements method is used in the biomechanical modeling of tibia. The geometric model of the tibia is obtained from CT images. The tibia is modeled as an anisotropic material with non-homogeneous mechanical properties.

Conclusions: The maximum stress is located in the shaft of tibia diaphysis. With both meshes are obtained similar results of stresses and displacements. For this patient-specific model, the torque must be greater than 30 Nm to initialize the correction of tibial torsion deformity.

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1. Introduction

Tibial torsional deformities are one of most important lower limb deformities in children, is defined as any twisting of the tibia on its longitudinal axis which produces a change in alignment of the planes of motion of the proximal and distal articulations. The angle of twist can be measurement as the angle between the posterior axis of the proximal plateau and the transtibial axis of the ankle (Fig. 1). The rear axis of the proximal epiphysis plateau is defined as the line joining the

two most posterior points of the plateau. The transtibial axis of the distal tibia is defined by drawing a line on distal articular surface of tibia connecting the tip of the medial malleolus to the mid-point of lateral border (fibular sulcus).^{1,2}

The necessity for treating tibial torsion is one of the most controversial topics in the orthopedic literature. Many authors confirm that the congenital and developmental deformities usually show spontaneous correction, whereas the post-traumatic variety requires osteotomy if the deformity is severe.² But, others says that it is not possible to predict which deformity will correct spontaneously; therefore, consideration

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Fig. 1 – Measurement of the angle between the posterior axis of the proximal plateau and the transtibial axis of the ankle.

should be given to treatment of the objectionable deformities and to prevention of the development of secondary deformities.¹

Harris³ lists his considerations about when tibial torsion will correct spontaneously and when will not. These factors influence the spontaneous tibial torsion correction: magnitude of the deformities; symmetry of deformity between the two legs; deformities in children approaching age two years is less likely to correct, the success of non-operative management of tibial torsion decreases as the child approaches age two. Consequently, persisting internal tibial torsion past nine years old will either have to be accepted or corrected by tibial osteotomy.⁴ In the non-operative treatment of tibial torsional deformities are used manipulation exercise and orthopedic appliance.⁵

The mechanical contribution to the bone tissue analysis is based in determine the structural response of bone tissue under mechanical loads. Stress, strain and displacement of bone are determined considering its mechanical properties, geometry and loads.⁶

The Finite Element Method (FEM) is the tool used in the modeling of bone behavior under mechanical loads. FEM is capable of performing high-fidelity simulation and accurate analysis of complex structure of human organs, is currently the most widely applied method for biomechanical analysis.⁷

The bone is a complex structure, but its biomechanical modeling can be simplified considering that it not exhibits large deformations and can therefore be suppose a linear behavior. The Young's modulus for cortical tissue is 14.3–21.1 GPa and for trabecular tissue is 1–20 GPa.⁸

Bone geometry is generated with the use of medical images acquired from slices made to bone in a Computed Tomography Scanner (CT or micro-CT) or Magnetic Resonance (MRI or micro-MRI). The 3D reconstruction of bone geometry is a complex process, it can be simplified with the help of software for the manipulation of CT data retrieved in DICOM format (Digital Imaging and Communication in Medicine). The image segmentation and the selection of the region of interest (ROI) make possible to contour the bone's transversal section to generate the 3D bone geometry.^{9,10}

The development of patient-specific finite element (FE) models from computed tomography (CT) data is a powerful tool to non-destructively investigate preparatory surgery,¹¹ Computer Aided Surgery (CAS),¹² locating bone defects, to determine the stresses and strains of bones subjected to loads, bone remodeling,^{13,14} bone-implant integration, and bone healing fractures.^{15,16}

The objective of this paper was to obtain a 3D patient-specific model of the tibia from computed tomography data. This model is used to determine the corrective torque needed to initialize the correction of tibial torsion deformity.

2. Materials and methods

Many orthopedic doctors have not tools in their practices for measurement the tibial torsion deformities. When orthopedic treatment is indicated then the corrective torque is applied empirically and cannot predict the time needed to correct the deformity. In these practices, diagnostic methods of the deformity are done with visual methods.

In Fig. 2 is shown a how include the 3D patient-specific model of the tibia from CT data in the diagnosis and treatment of an orthopedic defect. In this work, to calculate the corrective torque needed to initialize the correction of tibial torsion deformity. The result of the biomechanical analyzes will enable orthopedic doctors select the corrective parameters of tibial torsion treatment, as angle and torque.

2.1. Patient-specific model form CT-data

In the 3D bone reconstruction were used images in DICOM format from the Computed Tomograph Scanner GE Light-Speed VCT (120 kV/89.40 mAs, pixel 0.773 mm, 512 × 512, slice 5 mm) installed on the Instituto Nacional de Rehabilitación y Ortopedia of México.

It was used the software: Mimics 10.01 (Materialise, Leuven, Belgium) for the processing of medical images, 3D geometry reconstruction and in the assignment of the mechanical properties of bone, Abaqus 6.13 (Simulia, Dassault Systems, RI, USA) and Hypermesh 11 (Altair, HyperWorks, MI, USA) in generation and analysis of patient specific models.

The first step in Mimics software is images segmentation. The grayscale is used to select the region of interest like Threshold tool. The software converts the grayscale of images on Hounsfield units (HU). A new mask is defined considering HU limits for bone tissue (156–1799), so that bone tissue is separated from the rest of the image. The second step is select only the tibia from all bone tissue with Region Growing tool, then a new mask is created with tibia bone information. This

Fig. 2 – Use of 3D patient-specific model of the tibia from CT in the diagnosis and treatment of an orthopedic defect.

mask is edited manually to remove imperfections and to prevent future errors in the meshing and in the analysis results of FEM. Third, a 3D model of the tibia is generated from the mask. Fourth, the irregular shape of the geometry is smoothed with smoothed surface. Smoothed surface increases the quality of triangular superficial meshes.

2.2. Meshing and quality of the mesh of patient-specific model

The FEM approximate the bone geometry through the discretization or meshing of continuum. FEM is a numerical method and its solution is not exact. Its accuracy depends on precise definition of model that includes the construction of anatomical structure, and its approximation by meshing (mesh quality).

In biomechanical analysis of bone where the geometry is complex, automatic mesh algorithms are preferred, the continuum is meshed with tetrahedral elements.^{17,18} The meshes with tetrahedral elements can be structured or unstructured; unstructured meshes are used most frequently, irregular size elements are present in these meshes. The Advanced Frontal Technique (AFT) and Delaunay are the most used automatic mesh generation algorithms by finite element software.^{19,20} Tetrahedral volumetric mesh is generated from surface mesh of the outer and inner surfaces of the bone.²¹

The superficial mesh with triangular element was generated in Remesh module of Mimics Software. The element size limits is 2–5 mm.^{22,23} A quality control for triangular elements (shape factor base/height) lower to 0.35 was used. Normal reduction of elements size, Split based method and, finally preserving quality elements Mimics Remesh techniques are used to refine the mesh. Triangles with factor shape lower

that limit are edited manually. A superficial mesh with 8344 triangular elements, an average shape factor of 0.761786, an average aspect ratio of 1.65 and a maximum aspect ratio of 3.29 was generated. This superficial mesh was exported in STL file extension.

In the volumetric meshing of bone was used two variants. In both, the volumetric mesh is generated form STL file, and was obtained meshes with low and high order elements. (4 nodes C3D4 and 10 nodes C3D10 elements).

In the first variant, the volumetric mesh was generated in Abaqus (Simulia, Dassault Systems, RI, USA). The superficial mesh with triangular elements from Remesh module of Mimics was imported into Abaqus. Then an unstructured mesh with 57792 C3D4 tetrahedral elements, non-uniform element size from 2 to 5 mm, average aspect ratio of 1.66, maximum aspect ratio of 4.12 and average shape factor of 0.656129 was obtained. Finally, the mesh is exported in *.inp file extension. The inconvenient of this variant is their impossible to control the mesh density and the quality of volumetric mesh depend of the quality of superficial mesh. This mesh is considered by Abaqus like orphan mesh.

In the second variant, the volumetric mesh was generated in Hypermesh (Altair, HyperWorks, RI, USA). The superficial mesh with triangular elements from Remesh module of Mimics was imported into Hypermesh. Then new superfices form stl file information were created. The superficial mesh with triangular elements was removed and a new volumetric mesh with 50890 C3D4 tetrahedral elements, uniform element size of 2.5 mm, average aspect ratio of 1.57, maximum aspect ratio of 3.96 and average shape factor of 0.718784 was generated from the new superfices. In this mesh was controlled the elements size. Finally, these meshes were exported in *.inp file extension.

Fig. 3 – Stress and displacement results for two loads applied to the surface of distal epiphysis in parallel direction, and opposite sense: (a) stress, (b) displacement.

2.3. Non-homogeneous mechanical properties of bone

Bone is considered as non-homogeneous material, Mimics software is employed to assign mechanical properties. The mechanical properties are function bone density, so that the bone has a non-uniform mechanical properties.^{22,24}

The equations that relate the Hounsfield Units (HU) with density,^{25,26} and density with Young's modulus are defined. A constant Poisson coefficient of 0.3 is applied to all levels of mechanical properties. The density limits for cortical tissue and trabecular tissue (0.05–1.92 g/cm³) and for the HU (156, 1799) were established. With these values are obtained the coefficients of linear correlation of Eq. (1).

$$\rho = 0.00108 \cdot HU + 0.02901 \quad (1)$$

$$E = 6.950 \rho_{app}^{1.49} \quad (2)$$

3. Results y discussion

Bone is a material capable to adapt to the load conditions and modify its structure (Wolff's Law).²⁷ Bone cells detect mechanical signals and integrate these signals into appropriate changes in the bone architecture. Bone's stresses, strain and deformation energy density influenced the bone adaptation. Under this supposition is expected that a strained bone initialized the adaptation. The reference strain value used is 0.02 ϵ .²⁸

In the analyzes of patient-specific model with variant 1 mesh (mesh with tetrahedral element generated in Abaqus) and non-homogeneous mechanical properties, are defined two forces applied in two nodes, each force are in the surface of distal epiphysis in parallel direction, and opposite sense. One node is located near to medial malleolus and the other in the mid-point of lateral border (fibular sulcus). These forces reproduce the twist of the tibia. In proximal epiphysis surface is applied a fastening restriction, removing movement in all direction. Under this force condition it is obtained that maximum stresses are located at the nodes where forces are applied (Fig. 3).

This load definition is incorrect because at the nodes where forces are applied appear stress concentration. Furthermore,

in real life condition this load configuration does not represent the reality.

The patient-specific model with variant 1 mesh (mesh with tetrahedral element generated in Abaqus) and non-homogeneous mechanical properties is used again but a new load condition is defined. In the surface of distal epiphysis are defined a nodes set and later are coupled to a reference point. The reference point is used to apply a moment in axial direction that represent the bone twist.

With couple restriction all nodes on nodes' set will rotate together to the reference point. With this loads configuration the results obtained show that the maximum stresses are now in the shaft of tibia diaphysis (see Fig. 4a) and not in the distal epiphysis as in the previous loads configuration. Maximum strains are in the diaphysis too, nearest to the distal epiphysis of bone (Fig. 4b). This result indicates that tibial torsion correction is performed in the shaft of tibia diaphysis. The maximum displacement is located in the distal section of the tibia. This loads configuration corresponds with the expected behavior of the tibia under torque.

The specific model with variant 2 mesh (mesh with tetrahedral elements generated in Hypermesh) and non-uniform mechanical properties is also subjected to a torque on the distal epiphysis surface as explained above. Again the results obtained show that the maximum stresses and strains are placed in the shaft of the tibia diaphysis too, nearest to the distal epiphysis of bone (Fig. 5).

When comparing the stresses and strains of both analysis results show that there are no significant differences between

Fig. 4 – Results of stress and strain for variant 1 mesh: (a) stress (b) strain.

Fig. 5 – Results of stress and strain for variant 2 mesh: (a) stress (b) strain.

the two models. So it is feasible to use any of the two methods of meshing proposed in this work.

In Fig. 6 is shown the graph of relation between the torques applied to the surface of distal epiphysis and the strain results, for variant 1 and variant 2 mesh generation method. When the torque is bigger than 30 Nm then strain is greater than 0.02. For this patient-specific model, the torque must be bigger than 30 Nm to initialize the correction of tibial torsion deformity.

On the other hand, studies of tibia cross-section variation during growth have been carried out and in the mid-shaft diaphysis the tibia is quasi-circular in children under five years.²⁹

Taking into account that the mid-shaft cross-section of the tibia is quasi-circular and that the maximum stresses are located in this section of the bone, can simplify the model considering a hollow circular cross-section bar subjected to torque. The length of the bar will depend on the length of the bone. You can use both equations of classical mechanics to determine the stress and displacement or the MEF.

Fig. 6 – Relation between the torques applied to the surface of distal epiphysis and the strain results for meshes of variant 1 and 2.

With this type of geometry is simplifies the patient-specific model and serves as a first approximation to the stresses calculation to produce the bone remodeling. But if an approximation to real bone geometry is requires then the most recommended option to use is bone geometry from the CT data.

4. Conclusions

The biomechanical analysis of patient-specific model from CT data can be used as one more step inside the diagnosis and treatment of orthopedic practices. In the correct definition of patient-specific model is necessary take into account mechanical properties and bone geometry, loads and interaction of bone with joints. From CT can be generated a 3D model of the tibia as exact as possible and it enhance the accuracy analysis results. Is recommended to use the CT data to set non-homogeneous mechanical properties. Mechanical properties can be calculated from bone density, and the Eq. (1) of this paper can be used to define it. The loads and boundary conditions influence the results of the analysis. For the 3D patient-specific model of the tibia analyzed here, the torque must be greater than 30 Nm to initialize the correction of tibial torsion deformity. This torque varies from one patient to other, but the calculation process is simple one time that methodology is established.

A torque applied to the reference point coupled to the nodes set in the surface of distal diaphysis is the load condition that reproduces real life behavior of bone under twist. The maximum stress is located in the mid-shaft of tibia diaphysis. How in this section is where bone is strained, then cells activate the adaptation process and the rotation of deformity occur in this region of bone.

In the volumetric meshing of bone can be use either the variant one: unstructured mesh generated in Abaqus with 57792 C3D4 tetrahedral elements, non-uniform element size from 2 to 5 mm, average aspect ratio of 1.66, maximum aspect ratio of 4.12 and average shape factor of 0.656129; or the variant two: mesh generated in Hypermesh with 50890 C3D4 tetrahedral elements, uniform element size of 2.5 mm, average aspect ratio of 1.57, maximum aspect ratio of 3.96 and average shape factor of 0.718784. With both meshes are obtained similar results of stresses and displacements. No significant differences on results are shown, the results difference between two of them are less than 5%.

The patient-specific model can simplify the model considering a hollow circular cross-section bar subjected to torque because the mid-shaft cross-section of the tibia is quasi-circular and the maximum stresses are located in this bone's section. This type of geometry serves as a first approximation to the stresses calculation required to produce the bone remodeling. But if an approximation to real bone geometry is requires then the most recommended option to use is bone geometry from the CT data.

Conflicts of interest

All authors have none to declare.

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