Medical Engineering and Physics 000 (2018) 1–9

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

A patient specific finite element simulation of intramedullary nailing to predict the displacement of the distal locking hole

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ARTICLE INFO

Article history:
Received 22 December 2016
Revised 6 March 2018
Accepted 13 March 2018
Available online xxx

Keywords:
Interlocking nail
Locking screws
Nail deformation
Targeting systems
Femoral fracture fixation

ABSTRACT

Distal locking is a challenging subtask of intramedullary nailing fracture fixation due to the nail deformation that makes the proximally mounted targeting systems ineffective. A patient specific finite element model was developed, based on the QCT data of a cadaveric femur, to predict the position of the distal hole of the nail postoperatively. The mechanical interactions of femur and nail (of two sizes) during nail insertion was simulated using ABAQUS in two steps of dynamic pushing and static equilibrium, for the intact and distally fractured bone. Experiments were also performed on the same specimen to validate the simulation results. A good agreement was found between the model predictions and the experimental observations. There was a three-point contact pattern between the nail and medullary canal, only on the proximal fragment of the fractured bone. The nail deflection was much larger in the sagittal plane and increased for the larger diameter nail, as well as for more distally fractured or intact femur. The altered position of the distal hole was predicted by the model with an acceptable error (mean: 0.95; max: 1.5 mm, in different tests) to be used as the compensatory information for fine tuning of proximally mounted targeting systems.

1. Introduction

Intramedullary nailing is a common operative method for treatment of fractures of femoral and tibial shafts [1]. In this method, a metallic tubular rod or nail is inserted in the medullary canal to hold the bone fragments together. In order to provide a stable fixation, which is critical for the healing process, interlocking screws are used at the proximal and distal ends of the nail, to maintain the length, alignment and rotation of the bone.

The procedure of intramedullary nailing can be technically challenging due to the difficulties encountered in the insertion of the interlocking screws. For proper insertion of screws, it is necessary to drill the bone at the proximal and distal sites such that the resulting holes are exactly coaxial with the interlocking holes on the nail. In general, proximal locking might be easily performed using a mechanical guide which is mounted on the proximal end of the nail. However, such method is not applicable for distal locking due to the deformation of the nail after insertion. Considering the requisite of entering into the canal, the intramedullary nail has usually a different curvature from the medullary canal; the average radius of curvature of the femoral medullary canal is about 722 mm and that of the nail in the range of 1500–3000 mm [2]. This substantial difference results in a relatively large deformation in the distal nail, reaching up to 18 mm in the lateral plane [3]. As a result, the position of the distal hole is changed during the insertion procedure and cannot be located using a pre-designed proximally mounted targeting device that does not compensate for nail deformation [4,5].

The typical method of finding the position of the distal hole is the freehand technique. In this method, a large number of fluoroscopic shots are taken from the inserted nail, at different intensifier adjustments in the coronal and sagittal planes, until perfect circles are obtained for the holes. Obviously, this technique exposes the patient and operating room staff to excessive radiation [6], and

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https://doi.org/10.1016/j.medengphy.2018.03.004
1350-4533/

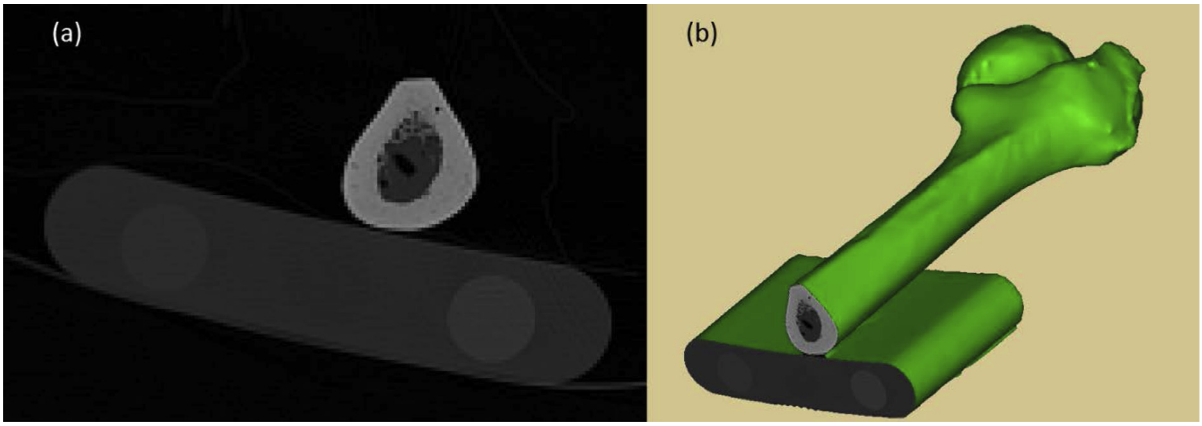

Fig. 1. Bone data acquisition. (a) A CT slice of the cadaver femur specimen and the BMD calibration phantom. (b) The 3D reconstructed geometry of the femur specimen.

prolongs the operation time [4,7,8], with magnitudes depending heavily on the surgeon's experience and skill.

There have been several proposed solutions to facilitate the distal locking procedure [9], based on electromagnetic systems [7,8,10], stereo and virtual fluoroscopy [11,12], mechanical devices mounted on the image intensifier [13,14], and proximally mounted radiation dependent mechanical guides [15–17]. Such solutions, however, are technically demanding and require extra equipment and trained staff that might not be affordable for many operating rooms. There have been also attempts to develop improved image-based techniques in order to recover the axis of a distal hole using two [18,19] or even one fluoroscopic shot [20,21]. Although these techniques are claimed to reduce the dosage of exposure, their efficacy for practical use has not been verified yet.

Patient-specific finite element (PSFE) modeling has been extensively used in recent years to predict the biomechanical behavior of muscelo-skeletal systems during medical interventions [22]. In particular, several studies have used PSFE to predict the mechanical response of long bones, i.e., stress, strain and displacement fields, after nail insertion [23–25]. However, to the best of our knowledge, there is no previous attempt on the estimation of the deformation of an inserted intramedullary nail, using simulation method.

The objective of this study is to propose and validate a PSFE model of intramedullary nailing, based on the preoperative data, and in order to predict the 3D deflection of the nail before surgery. Using such model, it will be possible to estimate the altered position of the distal hole of the nail, preoperatively, and provide the required compensatory information for tuning the proximally mounted targeting systems during surgery. Besides, this model can help nail manufacturers to optimize the design of the nail by analyzing the sensitivity of the nail deformations or stress distribution with respect to its geometrical parameters.

2. Material and method

2.1. Modeling

A fresh frozen human femur from a 63-year-old male cadaver was obtained with the approval of the local ethics commission. The approximate length of the femur specimen was 380 mm. The bone was cleaned from soft tissues and scanned in air using a CT machine (Brilliance 64, Philips, Germany; intensity: 100 mA; voltage: 120 kV) with a voxel size of $0.33 \times 0.33 \times 1\ \text{mm}^3$ and a slice thickness of 0.67 mm. A bone density calibration phantom (QRM-BDC/3 H200, QRM Gmbh, Germany) with three reference materials of zero, 100 and 200 Hounsfield Unit (HU) was placed next to the bone during scanning to obtain quantified computed tomography (QCT) data (Fig. 1).

An in-house segmentation software (Fanavaran Jarahyar Sharif, Tehran, Iran) was used to segment the CT data and reconstruct the 3D geometry of femur, in which the harsh points were removed and boundaries were smoothed out [26]. The HU number of each voxel was mapped onto the bone mineral density (BMD in mgHA/cc), according to the reference HUs of the calibrating phantom, that was then used to calculate the modulus of elasticity at each voxel, using the relations presented by Keyak and Falkinstein [27,28]:

$$\rho_{\mathrm{EQM}} = 10^{-3} \times (a \times \mathrm{HU} - b)\ \left[\frac{g}{\mathrm{cm}^3}\right] \tag{1}$$

$$\rho_{ash} = 1.22\rho_{\mathrm{EQM}} + 0.0523\ \left[\frac{g}{\mathrm{cm}^3}\right] \tag{2}$$

$$E_C = 10,200\rho_{ash}^{2.01}\ [\mathrm{MPa}] \tag{3}$$

where ρ_{EQM} and ρ_{ash} represent the equivalent mineral density and the ash density of bone, respectively. Also, E_C is the Young's modulus in the cortical region, and a and b are calibration parameters determined from the phantom data. The femoral bone was assumed to be an inhomogeneous isotropic linear elastic material [29] with the modulus of elasticity at each voxel determined using Eqs. (1) to (3), and the Poisson ratio kept constant at $\nu = 0.3$.

The femoral bone was meshed using linear tetrahedral elements, as suggested by previous investigations [30]. In order to assign the modulus of elasticity to each element, first, we defined a 3D scalar field which contained the moduli information of all voxels. The modulus of elasticity at each nodal coordinate was then calculated using trilinear interpolation. Finally, the modulus of elasticity of each element was calculated by averaging the values at the surrounding nodes [31].

Two stainless steel tubular nails (Osveh Asia, Mashhad, Iran) were used in this study, with the 3D models provided by the manufacturer. In order to consider the probable permanent deformation of the nail, should the yield stress (i.e. 316 MPa) be exceeded, a linear elastic-isotropic hardening plastic model was assumed for the nail material with $\sigma_{0.01}, \sigma_{0.2}$ and σ_u equal to 190, 316 and 616 MPa, respectively [32].

A Cartesian coordinate system was attached to the proximal end of the nail to describe its geometry and deformation in the 3D space (Fig. 2). The x axis was defined along the longitudinal axis of the nail, the y axis in parallel with the holes' axes, and the z axes perpendicular to x and y. Using these three axes, the spatial deformation of the nail could be projected into three planes: anterior–posterior (z) deflection due to bending in sagittal

Please cite this article as: J. Mortazavi et al., A patient specific finite element simulation of intramedullary nailing to predict the displacement of the distal locking hole, Medical Engineering and Physics (2018), https://doi.org/10.1016/j.medengphy.2018.03.004