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Implementation of boundary conditions in modeling the femur is critical for the evaluation of distal intramedullary nailing

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ABSTRACT

In previous numerical and experimental studies of the intramedullary nail-implanted human femur several simplifications to model the boundary and loading conditions during pre-clinical testing have been proposed. The distal end of the femur was fixed in the majority of studies dealing with the biomechanics of the lower extremity, be it numerical or experimental, which resulted in obviously non-physiological deflections. Per contra, Speirs et al. (2007) proclaimed physiological deflections as a result of constraining the femur in a novel statically determinate fashion in combination with using a complex set of muscle forces. In tandem with this, we have shown that not only the deflections but also the stress and strain predictions turn out to be much lower in magnitude, as a result of using the latter approach. To illustrate the dramatic change in results, we compared these results with those of two other models employing commonly used boundary and loading conditions in retrograde stabilization of a distal diaphyseal fracture. The model used herewith resulted in more realistic femoral cortical strains, lower stresses on both the nail and the screws, as well as such deflections in the overall structure.

1. Introduction

Although intramedullary nailing has become a standard approach to stabilize fractured bone fragments, failures are still encountered as reported in several studies, emphasizing especially interlocking screw failures due to excessive bending and fatigue [1,2], bone refracture and nail failure [3–6]. With a need for improving these devices, it is imperative that the deflections of the implanted femur, stresses in the nail and screws, and the femoral strains be accurately predicted. The finite element method is the de facto biomechanical modeling standard in orthopedics, and an indispensable tool for predicting *in vivo* biomechanical behavior.

Pre-clinical testing of intramedullary nailing systems by finite element analysis (FEA) is critical to determine stability and performance targets for these devices. While a simplified load case and geometry (e.g. a tube under axial compression) may be justified as a conclusive test method for an intramedullary nail (IM nail), the devices' actual functional performance depends on many factors such as strain distribution in the cortical bone, stress shielding, and shear movement at the fracture site [7,8]. Therefore, it is important to carry out the study on a model reflecting the physiological loading and boundary restraint conditions in order to address and alleviate these issues early in the design stage.

The biomechanical response of a nail implanted femur has been investigated in many FEA studies. In most of these, attention seems to favor the proximal part against the distal. In FEA studies focusing on the proximal part, the distal side was either fixed from the mid-diaphysis [9,10] or from the distal-condylar region [11–16]. Only a limited number of studies used restraint conditions with more care. The restraint conditions at the knee in particular, and the overall femur were considered in Speirs et al. [17]. A free-boundary condition approach based on a consistent self-equilibrating musculoskeletal force system was utilized by various groups [18–20].

In most experimental studies involving mechanical tests of nail-implanted femurs, the distal end was rigidly fixed by some means [7,13,15,21–27]. Over-constraining the distal end and excluding the muscles' forces resulted in a highly over-estimated deflection at the femoral head, as well as very high stresses and strains in the femur and the implant, diverging from *in vivo* conditions. In Speirs et al. [17], it was shown that an intact femur constrained in a statically determinate fashion at both ends and loaded by all muscles of the femur produced much lower deflections in agreement with *in vivo* deflection measurements provided by Taylor et al. [11]. The effects of improved boundary and loading conditions have been further elucidated in [18,19,28]. Based on the lower deflections and bending of the femur, it was argued that employing dominant muscular activity aside from the hip-joint contact force and

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reducing the degree of restraint (mounting, joint restraint, etc.) yield more physiologically relevant results.

Distal femoral fractures are rare, but severe, and mostly reported in young men and elderly women [29]. High incidence rates of septic (13%) and aseptic (14%) non-unions have been reported [30]. This type of fracture is difficult to treat due to a decreased blood supply to the fracture area and a lack of stability in the distal metaphyseal region. Thus, the fracture site stability in this type of fractures is still of special clinical interest. Currently, studies on the influence of boundary conditions on the retrograde stabilization of a distal femoral diaphyseal fracture are lacking. We show that boundary conditions are critical for predicting the mechanical response of bone-implant constructs by comparing the FEA results of three distinct loading and restraint cases. Clinically relevant results such as femoral strains, movement at the fracture site and stresses in the nail and screws were disputed.

2. Materials and methods

To facilitate comparison with other FEA studies and avoid subject-specific variations in geometry, the standardized geometry of the third-generation composite femur model available in the public domain through the Biomedical Research Community (http://www.biomedtown.org) was used in this study. Finite element modeling and analysis were performed in ADINA [31]. The magnitude and orientation of the hip-joint and muscle force components were defined with respect to a right-handed coordinate system centered at the hip-joint center, as described in Bergmann et al. [32]. A new generation of a relatively short retrograde IM nail with four interlocking screws (two at the proximal and two at the distal sides) stabilizing a femur was created. The 5 mm osteotomy was located in the distal one-third of the diaphysis. The IM nail and the bone model used in this study are depicted in Fig. 1. To allow for the placement of the nail in the intramedullary canal, drilling and reaming processes inside the cancellous bone were performed by intersection and subtraction operations in the software. The short tubular IM nail had outer and inner diameters of 9.5 mm and 4.5 mm, respectively, and a length of 250 mm. The maximum gap between the IM nail and the intramedullary canal was 0.25 mm. Solid circular interlocking pins of 4.5 mm in diameter were used to connect the bone and the nail. The complete structure consisting of the bone, the nail and interlocking pins has been referred to as the bone-nail construct, or simply as the construct.

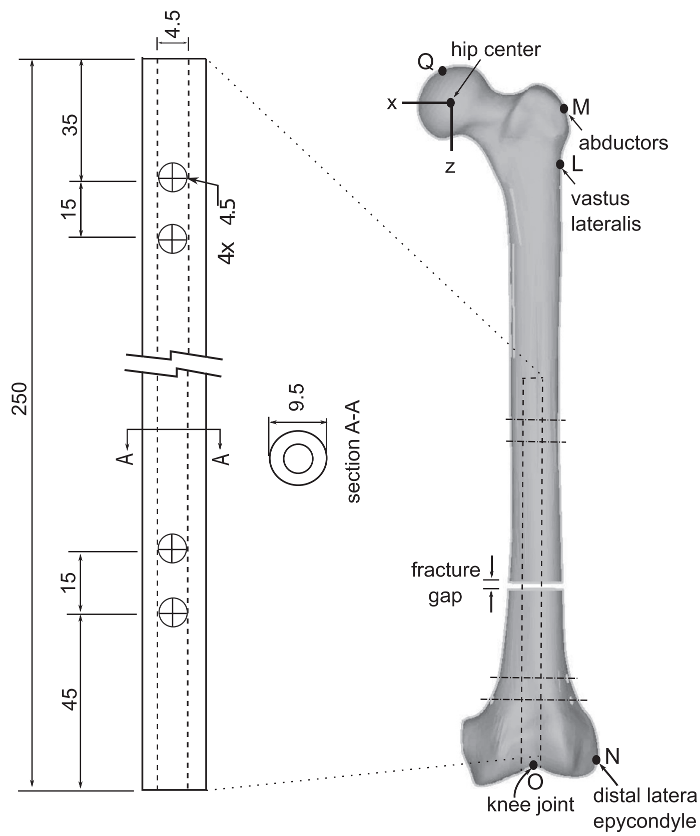

Fig. 1. Simplified model of the IM nail (left) and the third-generation composite femur (right).

2.1. Boundary and loading conditions

We compared three distinct boundary and loading conditions; two of which (cases *a* and *b*) corresponded to models used in the majority of similar studies, where the femur was fully constrained at the distal condyles. Only hip-joint loading was considered in case *a* (cp. [10,14,33]), while the muscles' loads were included in cases *b* and *c* (cp. [12,13,15,16,27]). In our study, the reduced muscle model by Heller et al. [34] has been adopted. As for boundary conditions, a head constraint was utilized in case *c*, as described in Speirs et al. [17] and as shown in Fig. 2. Additionally, the knee center *O* and the most lateral node at the distal femoral epicondyle *N* were also restrained. The head *Q* was only allowed to translate along an axis between the head and the knee center (the local x_l axis from point *Q* to *O*). The knee-joint center *O* was placed at the insertion of the anterior cruciate ligament, as suggested in Horsman et al. [35]. All translational degrees-of-freedom of the knee center were restricted, leaving it essentially as a spherical joint. Point *N* was restricted from translating in the antero-posterior (A-P) direction only. Thus, in case *c*, all the spatial degrees-of-freedom of the construct were constrained, yet it remained statically determinate.

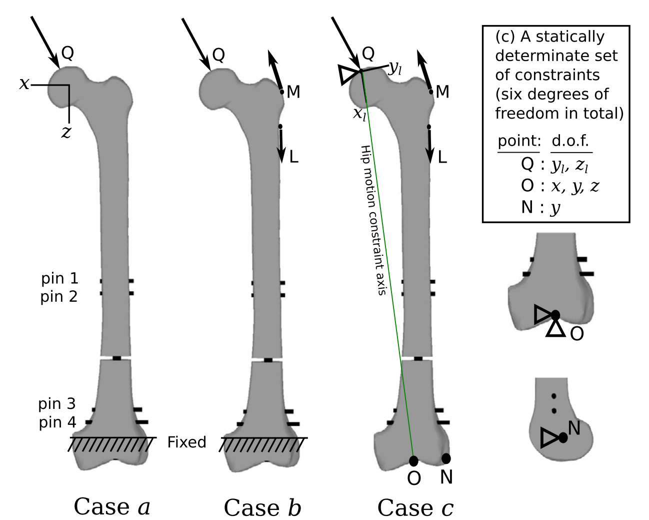

Fig. 2. Boundary and loading conditions considered in this study: case *a* consists of hip-joint contact force with fixed boundary conditions on the distal condyles. Case *b* is the same as *a*, but includes the reduced muscle model (abductor, tensor fascia latae and vastus lateralis) additionally. Case *c* corresponds to the physiological boundary conditions and the reduced muscle model.

As to the question of what activities should be mimicked during an FEA, Bergmann et al. [32] stated that femoral implants should mainly be tested under walking and stair climbing loads, as these activities result in the most critical loading of the femur. This hypothesis was further supported by Morlock et al. [36], where it was pointed out that walking was the most frequent dynamic activity. As far as the peak activity level during walking was concerned, the instant of highest hip-joint contact reaction during walking (first peak at 18% of the