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Technical note

# Selecting boundary conditions in physiological strain analysis of the femur: Balanced loads, inertia relief method and follower load

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ABSTRACT

Selection of boundary constraints may influence amount and distribution of loads. The purpose of this study is to analyze the potential of inertia relief and follower load to maintain the effects of musculoskeletal loads even under large deflections in patient specific finite element models of intact or fractured bone compared to empiric boundary constraints which have been shown to lead to physiological displacements and surface strains. The goal is to elucidate the use of boundary conditions in strain analyses of bones.

Finite element models of the intact femur and a model of clinically relevant fracture stabilization by locking plate fixation were analyzed with normal walking loading conditions for different boundary conditions, specifically re-balanced loading, inertia relief and follower load.

Peak principal cortex surface strains for different boundary conditions are consistent (maximum deviation 13.7%) except for inertia relief without force balancing (maximum deviation 108.4%). Influence of follower load on displacements increases with higher deflection in fracture model (from 3% to 7% for force balanced model).

For load balanced models, follower load had only minor influence, though the effect increases strongly with higher deflection. Conventional constraints of fixed nodes in space should be carefully reconsidered because their type and position are challenging to justify and for their potential to introduce relevant non-physiological reaction forces. Inertia relief provides an alternative method which yields physiological strain results.

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## 1. Introduction

Kinematic models and inverse-dynamics methods have been widely introduced into biomechanical research to map external loading to internal forces and moments [1–7]. These methods usually employ rigid body assumptions and approximation of kinematics to estimate *in vivo* loads. Analyzing tissue straining by means of finite element (FE) technologies facilitates extending local load information towards local strain. Such approaches have proven to be powerful to predict implant loading in various clinical situations after joint replacement [8–10], fracture fixations [11,12] or even simulating bone remodeling following total joint replacements [13] or fracture fixations [14,15].

Mandatory to these mechanical analyses is a proper understanding of mechanical boundary conditions (BCs) represented by the set of muscle and joint contact forces and fixation in space to achieve load equilibrium. Force balancing (FB), inertia relief (IR), and follower load (FL) are different options beside the widely used conventional fixation of nodes in space (displacement constraints) to achieve balanced loading.

*Force balancing*: Musculoskeletal models based on reasonable assumptions (e.g. muscle force optimization criterion) and measurements during gait analysis can be used to evaluate joint and muscle loads at specific points assumed a priori. These loads are in equilibrium with estimated inertial, measured external and estimated gravity forces. When additional information, for instance from measurements, is used to refine or alter the model, these forces may need to be re-balanced to achieve physiologic loading: when forces are distributed (i.e. on joint surface), added (e.g. iatrogenic pull during an operation or rehabilitation, or active implants), dropped (e.g. cut muscles or alternative muscle activation) or if there is high

*Abbreviation:* IR, inertia relief; FB, force balancing; FL, follower load; BCs, boundary conditions; BCS, boundary conditions of Speirs et al. (2007).

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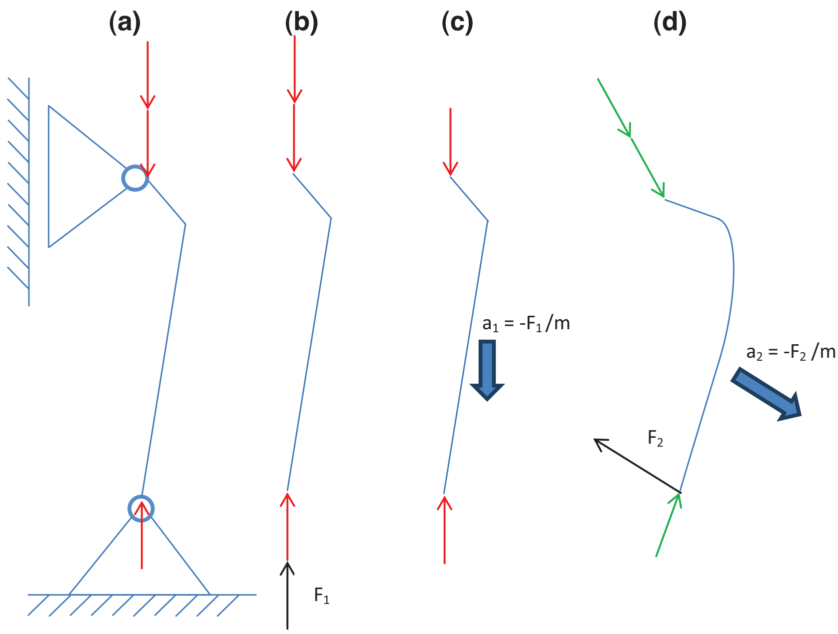


**Fig. 1.** Schematic diagram of the effect of different boundary conditions on a schematic femur depicted as a crooked beam with a strongly unbalanced load (a). Part (b) shows an additional reaction force that appears using a conventionally fixed node at the bottom. Part (c) illustrates how a surplus force $F_1$ is relieved from the structure through inertia relief and an equivalent acceleration is introduced according to the structure mass m. Part (d) demonstrates the change of the direction of the forces according to the deformation (follower load) and the change in the direction of the reaction force ($F_2$) or the acceleration ($a_2$) accordingly.

deformation or deflection, re-balancing of loads (adapting load estimates) within an FE model becomes essential. An example is the medio-lateral load distribution within the knee joint. Distribution of loads is difficult to measure in vivo while total reaction force and moment can be assessed reliably [16,17]. We suggest distributing the total central load using two singular medial and lateral compartment loads to assess physiologic straining of bones while neglecting the exact joint surface pressure distribution [18,19]. This re-distribution may cause non-physiological deflection through an additional moment. The loads have to be re-balanced or widely used constraints for the femur [20] can prevent non-physiological deflection but may result in high reaction forces which do not reflect *in vivo* loads. Reaction forces depend on the selection of the fixed nodes and constraints [20] and thus reflect a somewhat arbitrary concept, and differ among models and studies [21–24].

*Inertia relief*: Standard FE BCs constrain at least six degrees of freedom (DOF). Using the IR method [25,26] enables FE analysis without setting arbitrary constraints, putting the FE model in a dynamic equilibrium simulating static loading conditions without reaction forces [25]. When loads are reasonably well known, one node of relative acceleration in space can be set with optional additional displacement constraints. With IR, only the stress influencing elastic deformation is calculated and forces in excess of force equilibrium are converted to accelerations (Fig. 1), which may serve as a control variable for validation.

*Follower load*: FL has been introduced in spine biomechanics [27] as force which follows surface orientation by rotating with the body. Although follower load may not be a physiological loading configuration itself for all activities, FL has been shown to lead to more realistic results of an in vitro experiment of spine loading [27] and in different computer models [28,29].

BCs defined by Speirs et al. [20] can be used as a simple standard for FE models of the whole femur [53]. The purpose of the present study is to compare the BCs inertia relief method (IR), force balancing (FB) and follower load (FL) in respect to their relevance on bone strains in intact and fractured femoral bones using a patient-specific FE model. Our hypothesis is that adopting a combination of IR, FB and FL leads to a set of BCs with results that are physiologically realistic and comparable to results when using the definitions of Speirs et al. [20] (BCS) as described in the methods section. We aim at concluding with recommendations for the use of these BCs in biomechanical investigations of bone straining in FE analyses.

## 2. Methods

### 2.1. Patient data

A data set including gait, quantitative computed tomography (qCT) and ground reaction force data of one representative patient [13,30] forms the basis of this analysis.

### 2.2. Geometry

Patient-specific FE models were created from the qCT data set using Amira v.5.3 (Visage Imaging, San Diego, USA) for segmentation, Geomagic Studio 10 (Geomagic, Morrisville, USA) for creation of NURBS (Non Uniform Rational B-Splines) and Abaqus/CAE v.6.9 (Dassault Systèmes, Vélizy-Villacoublay, France) to create a solid body mesh of second-order tetrahedral elements [31–33]. Refinement in areas of high local curvature was applied. Initial mesh seed was set to minimum 7 mm distance (characteristic element length 1.67 mm, 240798 DOFs), or 5 mm (1.37 mm, 459825 DOFs) or 3.5 mm (1.10 mm, 894900 DOFs). Characteristic element lengths of meshes show that most elements have lengths well below recommended global edge length of approximately 2 mm for fine material distribution and good convergence [32,33].

In pre-tests, consistent models showed no relevant differences in evaluated results between mesh sizes except for the finest mesh with conventional constraints [20]. Strong local deformation at the single node could be overcome by extension of the constraint to neighboring nodes. The mesh with initial mesh seed of 7 mm led to consistent results with finer meshes regarding evaluated parameters so that further calculations were performed using this mesh.

### 2.3. Material model

Correlation between qCT image density (HU) and mineral density (g/cm$^3$) of a defined phantom was calculated ($R^2 > 0.999$; $p < 0.001$)

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