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Importance of trabecular anisotropy in finite element predictions of patellar strain after Total Knee Arthroplasty

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ABSTRACT

Patellar fracture and anterior knee pain remain major complications after Total Knee Arthroplasty (TKA). Patient-specific finite element (FE) models should help improve understanding of these complications through estimation of joint and bone mechanics. However, sensitivity of predictions on modeling techniques and approaches is not fully investigated. In particular, the importance of patellar bone anisotropy, usually omitted in FE models, on strain prediction is still unknown. The objective of this study was thus to estimate the influence of modeling patellar trabecular anisotropy on prediction of patellar strain in TKA models.

We compared FE-derived strain predictions with isotopic and anisotropic material properties using 17 validated FE models of the patella after TKA. We considered both non-resurfaced and resurfaced patellae, in a load-bearing TKA joint. We evaluated and compared the bone volume above a strain threshold and, in addition, estimated if the difference in isotopic and anisotropic predictions was consistent between patellae of different average bone volume fraction.

Compared to the anisotropic reference, the isotropic prediction of strained volume was 3.7 ± 1.8 times higher for non-resurfaced patellae and 1.5 ± 0.4 times for resurfaced patellae. This difference was higher for patellae with lower average bone volume fraction.

This study indicates that strain predictions acquired via isotropic patellar FE models should be interpreted with caution, especially when patellae of different average bone volume fraction are compared.

1. Introduction

Patellar fracture and anterior knee pain (AKP) remain one of the major complications after Total Knee Arthroplasty (TKA) [1,2]. It is believed that patellar resurfacing can decrease the risk of AKP, while non-resurfacing can help to avoid fracture and other complications associated with resurfacing [2]. These two surgical procedures are widely used in clinical practice and often compared, but no clear advantage of one over the other is reported [2].

Numerical studies suggest that estimation of patellar strain after TKA could help to better understand its pathologies and to choose an appropriate surgical technique [3–5]. However, literature is lacking a validated patellar material law for accurate strain predictions. Currently, the existing patellar numerical homogenized models (hFE) rely on isotropic material laws obtained from other anatomical sites, such as femora or vertebra [3,5,6]

In a previous study, we identified and validated a patellar material law based on morphology–elasticity relationship by means of micro-finite element (μFE) modeling of 20 fresh-frozen cadaveric patellae [7]. We considered two alternative models: isotropic and anisotropic. It was shown that the anisotropic model better replicates the μFE reference. The isotropic model underestimated the stiffness of the patella, and thus tended to overestimate bone strain. However, the validation in that study was conducted on the cuboid patellar section by means of tension and shear load testing applied on the sides of the cuboid. To estimate if the isotropy simplification indeed increases strain prediction in clinical applications, the isotropic model should be compared to the anisotropic model during physiological loading conditions. Although there is a potential to measure anisotropy with standard preoperative computed tomography (CT) scans [8], this modeling approach has not yet been validated. Thus, the isotropic model has higher potential to be used in clinical applications. The estimation of influence of the isotropic simplification on strain prediction is therefore of great importance.

Hypothesizing that anisotropy plays a crucial role in patellar strain prediction, the aim of this study was to compare calculated patellar strains during a loaded knee flexion after TKA using isotropic and anisotropic validated models. We considered both non-resurfaced and resurfaced patella, since these two cases are

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often compared. In addition, we evaluated if the difference in isotopic and anisotropic strain predictions is consistent between patellae of different average bone volume fraction, since it has been suggested that predictions of isotropic models of bones with low bone volume fraction will be higher deviated from anisotropic models predictions [9].

2. Materials and methods

Seventeen fresh-frozen cadaveric patellae (10 males, 7 females; age range 34–93, mean age 70 ± 18) were used for the study. The strains of each patella were evaluated by an isotropic and an anisotropic validated hFE model [7], in a non-resurfaced and resurfaced patella option of TKA. The applied boundary conditions were provided by a validated musculoskeletal knee model [10]. The effect of the isotropic simplification was estimated by comparing the strain predictions with the anisotropic reference.

The model included the patella, the cartilage (non-resurfaced), the patellar component (resurfaced), the surface of the femoral component, the patellar ligament and the four quadriceps muscles: vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF) and vastus intermedius (VI). The geometry of each patella was extracted from segmented µCT scans, obtained during a previous study [7], and imported to Geomagic (Geomagic, Inc., Morrisville, North Carolina, USA) to create non-uniform rational B-spline (NURBS) surfaces. To simulate non-resurfaced cases, we added a cartilage layer for each patella by a uniform extrusion (3 mm) of the posterior articular surface of the patellar bone [6]. To simulate the resurfaced cases, we replicated recommendations of the manufacturer (Symbios, Yverdon-les-Bain, Switzerland). The posterior part of the patella was cut, three cylindrical holes were removed, and the three-peg modified dome patellar component was inserted [2]. The cut depth (5–9 mm), as well as the prosthetic component size (thickness/diameter: 8 mm/31.2 mm, 8.5 mm/34.2 mm, 9 mm/37.2 mm), depended on the patellar size and were aimed to preserve the original thickness of the patella without exceeding critical remaining bone thickness of 12 mm. Only the articular surface of the femoral component was included into the model. The geometry of the femoral and patellar components was obtained from the manufacturer. For simplicity, the cement layer was not modeled. We used CAD software Solidworks (Dassault Systèmes, Vélizy, France) to create the cartilage layer, and to cut the bone and position the patellar component.

We replicated a loaded squat at 60 degrees of knee flexion. The position of the femur and tibia was fixed and imposed by the squat movement. The muscle forces were estimated by a validated musculoskeletal TKA model, assuming a constant body weight (800 N) [10]. The muscle forces (RF: 544 N, VI: 706 N, VL: 1216 N, VM: 778 N) were distributed according to muscles physiological cross-sectional areas [11]. The cartilage (non-resurfaced) and patellar component (resurfaced) were in contact with the surface of the femoral component. The position of the patella was thus only constrained by its contact with the femoral component, the applied muscle forces, and the patellar ligament reaction (Fig. 1).

The Zysset–Curnier morphology–elasticity relationship was considered for the patellar bone [12,13]:

$$E_i = E_0\rho^k\left(m_i^2\right)^l, \quad \frac{E_i}{\nu_{ij}} = \frac{E_0}{\nu_0}\rho^k\left(m_i m_j\right)^l,$$

$$G_{ij} = G_0\rho^k\left(m_i m_j\right)^l, \quad \forall i \neq j = 1, 2, 3$$

where E_i, ν_{ij}, and G_{ij} are engineering constants, E_0, ν_0, G_0, k, l are model parameters, ρ is the bone volume fraction, and m_i are the normalized eigenvalues of the fabric tensor $\boldsymbol{M}$ [14]. In the isotropic case the fabric tensor $\boldsymbol{M}$ was equal to the identity tensor $\boldsymbol{I}$. Model

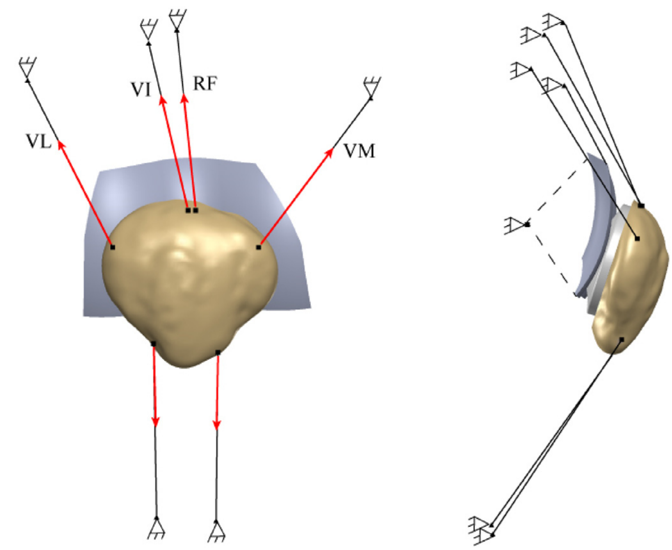

Fig. 1. Patellofemoral TKA model at 60 degrees of knee flexion.

Table 1
Model parameters for isotropic and anisotropic laws.

Law	E_0 (MPa)	ν_0	G_0 (MPa)	k	l
Isotropic	11035.98	0.26	4395.05	2.13	–
Anisotropic	12723.05	0.24	4224.62	2.1	1.02

parameters were identified and validated using micro finite element (µFE) modeling on 20 cadaveric patellae (Table 1) [7] .

In the homogenized isotropic and anisotropic models, the material properties of each bone element were assigned from µCT images using Medtool software (www.dr-pahr.at). A background grid with cubic hex elements (2.0 mm side length) was defined over µCT data set. A spherical volume with 5.3 mm diameter was centered at each node of the grid. Bone volume fraction (bone volume over tissue volume) and mean intercept length (MIL) based fabric tensor were computed for each volume and assigned to the node of the grid. Bone volume fraction and fabric tensor were interpolated to the elements of the patellar bone mesh, providing engineering constants and material orientations for all elements. The cortical bone was not modeled explicitly. Cartilage was assumed Neo-Hookean hyperelastic (C_{10} = 2 MPa, k = 40 MPa, derived from E = 12 MPa, ν = 0.45) [15], polyethylene was assumed linear elastic (E = 572 MPa, ν = 0.4) [3]. The femoral component was rigid. The patellar ligament was modeled by two rigid bars.

The model was implemented in Abaqus v6.13 (Simulia, Providence, RI, USA). Patellar bone and patellar component were meshed with linear tetrahedral elements (2 mm and 1.6 mm element size respectively), while cartilage was meshed with linear hexahedral elements (1.8 mm element size). Bone mesh type and size was done based on previous experience [16]. The muscle and ligament forces were distributed along all nodes of the entire anterior patellar surface with cubic weight function [17]. The system contained around 10^4 degrees of freedom. The implicit solver was used.

We evaluated octahedral shear strain of all patellae [18], for the isotropic and anisotropic models, and for the non-resurfaced and resurfaced cases. To compare isotropy to anisotropy, we calculated for the two cases the volume of bone with a strain above a threshold value. The anisotropic case was used as a reference. Three bone volumes of 2%, 5% and 10% with highest strains in anisotropic case were associated to three stain thresholds. The strain thresholds