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Case Study for Contact Pressure Improvisation with Graded Implant Material in Articular Cartilages of Knee Joint

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Abstract In this study, the effect of graded design in comparison to homogeneous cartilage material is investigated for contact pressure distribution in a human knee joint. The knee implants are assumed as homogeneous material. In reality, the cartilages are not homogeneous, and in order to replicate the heterogeneity of cartilages, a graded design is proposed. The simulation results show an improved contact pressure distribution in the knee joint due to the graded composition of cartilages. The results are helpful in designing a new class of implant materials.

1. Introduction

The articular cartilage serves as the load-bearing component for knee joints [1]. Osteoarthritis (OA) is a significant public health concern due to knee joint pain during walking, climbing, kneeling. Also, the increase in knee replacement operations (TKR) in recent years, among younger patients is direful [2-4]. OA is a chronic progressive musculoskeletal disorder characterised by degradation of articular cartilage in knee joints. Thus an understanding of degradation phenomena requires knowledge of the stress fields within the tissue [5-6]. The numerical method is a better way to understand the behaviour of cartilage [7-8]. Since experimental measures having limitations like accuracy of in-vivo tests, measurements at lower spacial scales, synchronous measurements of joint level kinematics or kinetics, difficulty in measuring variables such as stress, fluid flow, and contact pressure within the cartilage [9-10]. Finite element simulation is a widely adopted technique to investigate the biomechanics of the knee joint at the cell, tissue, and joint levels [11-13]. Further, well-known material models such as linear elastic, isotropic poroelastic, isotropic hyperelastic, transversely isotropic poroelastic, and fibril-reinforced poroelastic models usually proposed to give insight into the constitutive dependency of physical response. Thus, these models help to investigate the mechanical behaviour and damage mechanisms of articular cartilage [15-16].

Articular cartilage can be divided into three zones: superficial, transitional/middle, and deep. The collagen fibers that are oriented to merge with the articular surface are considered a superficial zone. The orientation perpendicular to the boundary of cartilage with bone is a deep zone, and randomly oriented ones are in the middle zone [17]. The zonal thickness of superficial, transitional, and deep zones are roughly around 12%, 32%, and 56%, respectively, of the total width of cartilage [18]. Following this, a graded material design can impart unique advantages that can be used to improve the durability of implants.

Though cartilages are non-isotropic and inhomogeneous, these tissues are modelled as a simple linear elastic material for qualitative knee joint analysis [19]. Researchers have primarily used cartilage tissue as a linear elastic isotropic material, as this helps to investigate instantaneous response [20]. As the mechanical response of cartilage tissues is understood to be nonlinear and involves large deformation, the adoption of a hyperelastic material model is more appropriate [8], [14], [21].

This short article's objective is to highlight and understand the influence of proposed 'graded' material on contact pressure distribution in articular cartilages. A finite element analysis (FEA) on a knee model with proposed graded implant material is implemented at full extension position.

2. Materials and Methods

2.1 Geometry & Material Models

The geometry of the knee joint is taken from an open-source project (open knee) for studying the knee joint and its effect on different loading conditions on underlying tissues [22]. The specimen details along with its coordinate system used in this work as gender-female, age-70 years, weight- 77.1 kg, side-right knee and height-5'6" and the coordinates like x-axis as anterior-posterior, y-axis as proximal-distal and z-axis as medial-lateral directions.

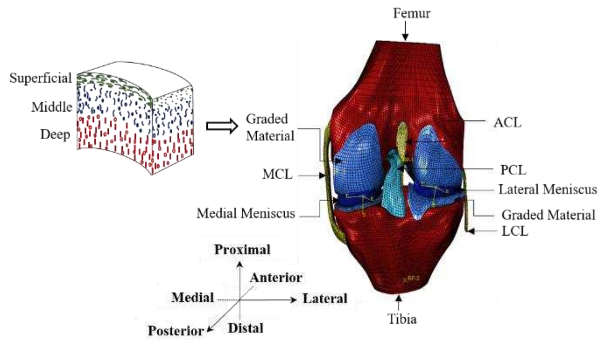

Fig. 1. Finite element model of the knee joint including components such as femur, tibia, femoral cartilage, tibial cartilage, medial meniscus, lateral meniscus, medial collateral ligament, lateral collateral ligament, anterior cruciate ligament and the posterior cruciate ligament. The red lines in graded material show the fibre orientation perpendicular to the articular surface, the blue line shows the cross direction, and the green line indicates fibre orientation parallel to the articular surface. The coordinate system chosen as medial-lateral as x coordinate, proximal-distal as y coordinate and anterior-posterior as z coordinate.

The graded design for articular cartilage is given in Fig. 1 and the model is applied to both femur and tibial cartilages. The graded design replicates the realistic cartilage structure that is the orientation of collagen fibres in cartilage from the articular surface to bone.

A dynamic implicit analysis of knee joint is performed by simulation software- Abaqus 6.14 (Dassault Systems Simulia Corp., Providence, RI, USA). The knee joint model contains four ligaments (anterior cruciate ligament, posterior cruciate ligament, medial collateral ligament and lateral collateral ligament), two cartilages (femur cartilage and tibial cartilage), two menisci (lateral and medial) and two bones (femur and tibia) as indicated in Fig. 1. Patella and patellar tendon are neglected for reducing the complexity of the model as our primary focus is on the tibio-femoral joint.

The displacement of the meniscus is constrained in such a way that it can mimic the horn arrangement of the meniscus as well as retain their position between cartilages. Nodes on the medial face of the lateral meniscus and the lateral facet on the medial meniscus are constrained in the z-direction along with the nodes on the interior edge of these faces are constrained in x and z-direction. The cartilage and bones tied together using rigid body constraint.

The tibia is constrained in all rotational and translational DOF where the femur is free to advance in five degrees of freedom and restricted in knee flexion (rotational DOF) to simulate a gait load at full extension. Then, the femur is given by distal (compressive) displacement of 1mm rather than force (to avoid convergence issues) at the reference point (RP-2) from time 0-1 second results in linear ramping. A dynamic implicit analysis conducted using Abaqus version 6.14 for determining the micro-scale changes in the tissue.

In this work, bone is selected as a rigid body due to its higher stiffness (several orders of magnitude higher than soft tissues), modulus of elasticity and Poisson's ratio shown in Table 1.

Table 1. Material parameters for different material models of the knee components such as bone, ligaments, meniscus and cartilage

Knee components		Material models	Parameters	Values
Bone		Linear elastic [12]	E (MPa) ν (-)	8000 0.3
Ligaments	ACL	Isotropic hyperelastic (Neo Hookean) [13]	C_1 (MPa) D_1 (MPa)$^{-1}$	1.95 0.00683
	PCL		C_1 (MPa) D_1 (MPa)$^{-1}$	3.25 0.0041
	MCL		C_1 (MPa) D_1 (MPa)$^{-1}$	1.44 0.00126
	LCL		C_1 (MPa) D_1 (MPa)$^{-1}$	1.44 0.00126
Meniscus		Transversely isotropic linear elastic [23]	E_θ (MPa) $E_z = E_r$ (MPa) ν_{rz} (MPa) $\nu_{r\theta} = \nu_{z\theta}$ (MPa) G_{rz} (MPa) $G_{r\theta} = G_{z\theta}$ (MPa)	120 20 0.2 0.3 8.33 57.7
Cartilage		Homogeneous material [13]	E (MPa) ν (-)	15 0.46
		Graded material	E_s (MPa) E_t (MPa) E_d (MPa) ν (-)	5 10 15 0.46

Nearly incompressible isotropic hyperelastic material model (Neo Hookean) is used to represent the behavior of all ligaments with strain energy function given by Eq. (1). The Neo Hookean material constants of Eq. (1) are given by Table 1.

$$W = C_1(I_1 - 3) + \left(J - 1\right)^2 / D_1 \tag{1}$$

where I_1 corresponds to the first invariants of the left Cauchy-Green deformation tensor B, J is Jacobian = det(F) and F is the deformation gradient. C_1 and D_1 are material constants defined by the shear modulus (G) and bulk modulus (K) which depends on modulus of elasticity (E) and Poisson's ratio (ν) given in Eq. (2). C_1 represents the deviatoric part and D_1 represents the volumetric part of the strain energy per unit volume while deforming.

$$C_1 = G/2,\ D_1 = 2/K,\ G = E\ /\ 2\left(1+\nu\right),\ K = E\ /\ 3\left(1-2\nu\right) \tag{2}$$

The meniscus modelled as transversely isotropic linear elastic material with material constants circumferential, axial and radial elastic modulus, similarly Poisson's ratio and shear modulus in circumferential, axial and radial directions. For the case of the gradient material model, the elastic constants are chosen in such a way that Young's modulus increased from the superficial zone to the deep zone and the values given in Table 1.

3. Results

Contact pressure on the tibial and femoral cartilage at full extension position of a stance phase (standing/equilibrium position) is given by Fig. 2. By using the values of the material parameters from the literature (Table 1), contact pressure in the knee joint with graded and homogeneous cartilage was analysed. The maximum contact pressure generated on gradient linear elastic tibial cartilage is 4.232 MPa which is less than homogeneous linear elastic tibial cartilage 6.45 MPa, as shown in Fig. 2.

Similarly for femur cartilage, the contact pressure generated is 4.483 MPa on homogeneous material and 3.00 MPa on graded material. Also, by comparing femur and tibial cartilage, a significant contact pressure appeared in the medial part of tibia cartilage and moderate contact pressure observed on the lateral part femur cartilage. The obtained results also show, for both material contact pressure is more evenly distributed on the lateral and medial compartment of femur cartilage while for tibial cartilage pressure is more concentrated on the lateral compartment.

Similar to contact pressure, Von-Mises stress also has a vital role in predicting knee pathologies accurately. The maximum Von Mises stress generated on tibial cartilage for graded material is 1.671 MPa, which is less than homogeneous material 2.274 MPa, as shown in Fig. 3.

Also for femur cartilage, the maximum Von-Mises stress generated for graded material is 1.671 MPa which is less than homogeneous material 1.897 MPa. The mises stress generated on the articulating surface of the cartilage is higher for the homogeneous elastic material model as compared with gradient material. Also, the maximum Von Mises stress for femur cartilage (2.274 MPa) generated is higher than tibial cartilage (1.897 MPa). The maximum contact pressure generated on the knee joint, in the range of 2.3 - 5.08 MPa during standing position (full extension) given by Table 2 [24–29].

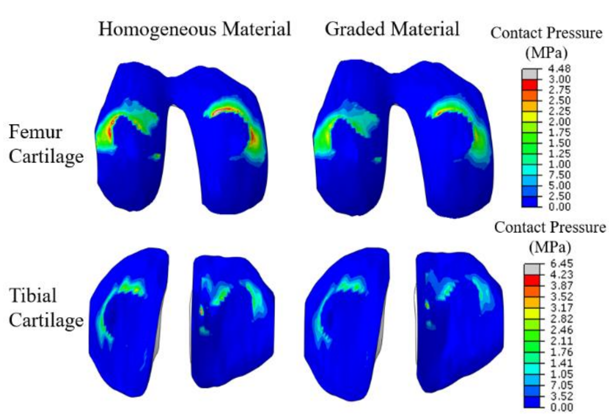

Fig. 2. Comparison between the contact pressure of articular cartilage in the femoral surface of the knee with homogeneous and graded material at full extension (standing/equilibrium) position. The graded material shows a lesser contact pressure generation rather than homogeneous.

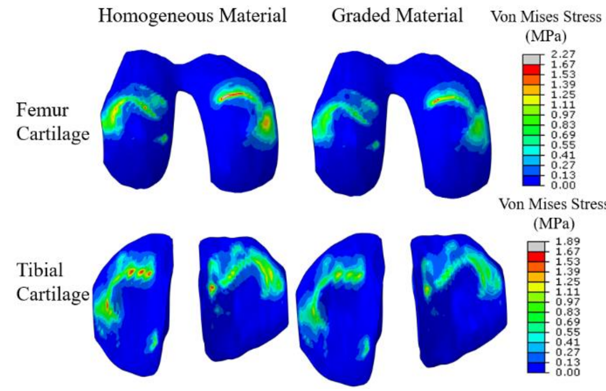

Fig. 3. The maximum cartilage von-mises stress distribution on the tibial and femoral surface for homogeneous and graded material at full extension position. The von-mises stress generated on graded tibial or femoral cartilages is lower than the homogeneous material model.

4. Discussions

The intact knee joint supports the contact pressures, compression stresses and shear stresses over a large area of femoral and tibial cartilage. The present study shows the effects of graded design articular cartilage over homogeneous cartilage in human knee joint, particularly in contact pressure distribution and compression stresses generated.

By comparing femur and tibial cartilage, the contact pressure appeared in tibia cartilage is higher than femur cartilage. The main reason behind this is, the reaction force on tibia cartilage is higher than femur as body force is acting in a downward direction. Hence there is a higher chance of degradation for tibial cartilage than femur cartilage.

Also, in this study, we investigated how material properties of cartilage affects the contact pressure and stresses generated on the articulating surface during loading (simulating standing). However, the results imply gradient material helps in reducing the contact pressure and stress generation on the articular surface, and it also helps to predict results more accurately. Therefore graded material design can be used as an alternative material for homogeneous cartilage component in artificial knee implants.

Table 2. Comparison of contact pressure on the cartilage of the current study with results from the literature.

Experimental/Computational study		Compression load (N)	Peak cartilage contact pressure (MPa)
Experimental	Morimoto et al. [24]	1000 N	3.6 MPa
	Marzo et al. [25]	1800 N	5.08 MPa
	Allaire et al. [26]	1000 N	5 MPa
	Lee et al. [27]	1800 N	4.5 MPa
	Paletta et al. [28]	1800 N	2.3 MPa
Computational	Halonen et al. [29]	1800 N	3 MPa
	Current study	800 N	6.45 MPa

The maximum contact pressure generated on cartilage during loading from literature is given in Table 2. From the current study (graded material), the maximum contact pressure generated is 6.45 MPa for 800N load, while other results show a lesser contact pressure corresponds to a higher load. There is no significant cause for the phenomenon, and this may be due to the limitations of the model such as no standard constraints, different geometry (variation in the contact area), different analysis software packages etc.; however, the maximum contact pressure generated on the cartilage is in the range of 2 to 8 MPa for a knee at full extension position is confirmed.

Other limitations to the situation in Vitro is the nutrition of cartilage as per its deformation and 'weeping lubrication' which is a severe limitation of an artificial design and a challenge for future development. The weeping lubrication in the knee joint is not considering in the simulation. Hence the predicted values obtained through simulation are not accurate when compare with actual values. If such properties are included (weeping lubrication around joint), the results would marginally change; however, they should not change any of the conclusions concerning the gradient material model. These critical findings will be utilised to optimise the required mechanical characteristics of the articular cartilage and eventually accomplish successful cartilage transplantation in a clinical scenario.

5. Conclusions

The proposed gradient material can be used as an implant material for osteoarthritis patients. Also, these discoveries and proposals are applicable to consider in biomechanical models to investigate treatments (surgical or traditionalist) related to knee osteoarthritis. Hence, the outcome of our finite element simulation can be extrapolated into the in-vivo scenarios.

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Nomenclature

ACL	: Anterior Cruciate Ligament
PCL	: Posterior Cruciate Ligament
MCL	: Medial Collateral Ligament
LCL	: Lateral Collateral Ligament
E	: Elastic modulus
ν	: Poisson's ratio
C_1, D_1	: Neo Hookean constants
E_θ	: Circumferential elastic modulus
E_z	: Axial elastic modulus
E_r	: Radial elastic modulus
ν_{rz}	: Circumferential Poisson's ratio
$\nu_{r\theta}$	: Axial Poisson's ratio
$\nu_{z\theta}$	: Radial Poisson's ratio
G_{rz}	: Circumferential shear modulus
$G_{r\theta}$	: Axial shear modulus
$G_{z\theta}$	: Radial shear modulus
E_s	: Superficial zone elastic modulus
E_t	: Transverse zone elastic modulus
E_d	: Deep zone elastic modulus

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