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The Effects of Surface Curvature on Cartilage Behaviour in Indentation Test: A Finite Element Study

Mohd Juzaila Abd Latif[a,b,*], Nur Hikmah Hashim[a], Roszaidi Ramlan[a,b], Jamaluddin Mahmud[c], Aidah Jumahat[c], Mohammed Rafiq Abdul Kadir[d]

[a]*Faculty of Mechanical Engineering, Universiti Teknikal Malaysia Melaka, Hang Tuah Jaya 76100 Durian Tunggal, Melaka, Malaysia*
[b]*Rehabilitation Engineering and Assistive Technology Research Cluster (REAT), Universiti Teknikal Malaysia Melaka, Hang Tuah Jaya 76100 Durian Tunggal, Melaka, Malaysia*
[c]*Faculty of Mechanical Engineering, Universiti Teknologi MARA, 40450 Shah Alam, Selangor, Malaysia*
[d]*Faculty of Biosciences and Medical Engineering, Universiti Teknologi Malaysia, 81310 JohorBahru, Johor, Malaysia*

Abstract

Computational modeling of the behavior of articular cartilage is important in order to improve the understanding of disease processes such as arthritis, and the suitability of biomaterials in surgical treatment. In previous computational studies, the cartilage surface of axisymmetric models was assumed to be flat in order to evaluate the cartilage behavior. This assumption was inappropriate since the synovial joint possessed curvature geometrical shape and may contribute to inaccurate results. Therefore, this study aims to examine the effects of the cartilage surface curvature to the cartilage behavior in indentation test using finite element analysis. Axisymmetric biphasic poroelastic finite element models of flat and various cartilage surface radii, including both concave and convex shapes of the curve, were generated to simulate creep indentation test in order to investigate possible effect to the contact stress and pore pressure of the cartilage. Based on the results, the smaller cartilage surface of 10 mm radius produced higher difference of the cartilage behavior where it generated 39% difference in pore pressure and 6% difference in contact stress, compared to the flat cartilage. This could indicate that the cartilage curvature does affect the cartilage behavior in indentation test particularly the pore pressure of cartilage.

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Keywords: Articular cartilage; finite element; pore pressure; contact pressure.

Nomenclature

E	Young's modulus
Greek symbols	
κ	permeability
v	Poisson's ratio
e	void ratio

1. Introduction

Articular cartilage is a smooth and glistening bluish-white tissue which covers the opposing articular margins

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of the synovial joints in human body. The main functions of articular cartilage in synovial joints are to transmit loads between the opposing joint surfaces, to distribute the stresses over the subchondral bones, and to provide a low-friction articulation [1]. These functions are achieved from the unique material properties possessed by the cartilage.

Articular cartilage consists of two distinct phases which are the fluid and solid phases. The fluid phase is composed of 60-85% water while the solid phase is composed of 15-22% collagens, and 4-7% aggrecan by wet weight [1]. The tissue contains four different zones with respect to depth, which from the surface to the subchondral bone are the superficial, middle, deep and calcified zones [1]. This composition makes the articular cartilage structure inhomogeneous, and possesses anisotropic and nonlinear properties both in compression and tension.

The biphasic properties of the cartilage are commonly characterized using either the experiment method or analytical solutions to determine the elastic modulus (E), Poisson's ratio (v), and permeability (κ) [2,3,4]. However, the indentation test is preferred because of the test set-up can allow the cartilage to be submerged in the fluid during the test and the ease of sample preparation. Recent advancement in computational methods have simulated the indentation test and derived the cartilage properties incorporated with experiment data [5].

Various constitutive material computational models have been used to describe cartilage from single-phase to multiphasic models. However, the biphasic theories developed by Mow and co-workers [2] have been widely accepted to represent the biphasic nature of the cartilage compressive behavior in solid and fluid phases. Both of the phases were individually immiscible and incompressible. The solid phase was porous and permeable to fluid flow, which is subsequently responsible for the compressive behavior of the cartilage.

Although there have been extensive computational studies of cartilage behavior, the cartilage surface in the biphasic axisymmetric finite element (FE) model was assumed to be flat since this appears to have been assumed in most of the previous studies in order to evaluate the cartilage behavior [6,7,8]. However, this assumption was not appropriate because the nature of the cartilage in synovial joint to be curved.

Therefore, the aim of this study was to examine the effect of the cartilage surface curvature to the contact stress and pore pressure of the cartilage during indentation test using axisymmetric FE model. These findings could be used to study the behavior of the cartilage across the articular surface of synovial joint.

2. Methodology

Computational models were developed in order to simulate the creep-deformation phenomenon performed in indentation test experiment. All the FE models were processed using Abaqus 6.9-EF (DS Simulia Corp., Providence, RI, USA). An idealised axisymmetric biphasic poroelastic model was used to examine the contact stress and pore pressure of the cartilage. Subsequently, various cartilage surface radii from 10 mm to 50 mm were employed including both concave and convex shapes of the curve to examine possible effect of this curvature.

2.1. Implementation of Contact Dependent Flow

A contact dependent flow algorithm was implemented to account for the change in the flow conditions between the free surface and the region where the indenter was in contact with the cartilage [9]. In this algorithm, the contact stresses at the cartilage surface nodes were recorded and evaluated. For the stress which was above the set threshold (0.0 MPa), indicating contact with the indenter, a sealed (no-flow) condition was applied at the node, otherwise a free flow condition was applied. The effectiveness of the algorithm can be observed in Fig.1 where the fluid flow in the contact nodes were completely stopped as impermeable indenter was used.

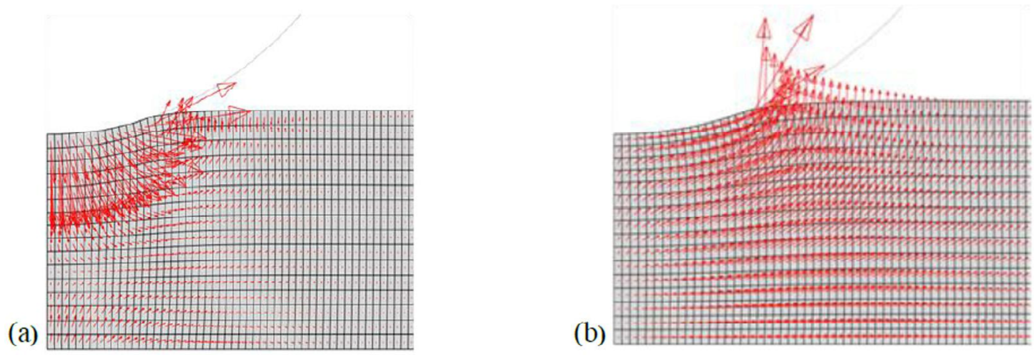

Fig.1. Direction of fluid velocity vector at (a) 2 seconds and (b) 1000 seconds.

2.2. Development of Finite Element Model

An axisymmetric biphasic poroelastic finite element model was generated to simulate the indentation test experiment on cartilage cylindrical pin. The cartilage pin was modeled using 0.5 mm cartilage thickness, whilst the bone was modelled at a constant 1.5 mm height [10]. Four-node bilinear displacement and pore pressure elements (CAX4P) were employed to model the cartilage whilst four-node bilinear elastic elements (CAX4) were used to represent the underlying bone [10]. The 1 mm radius rigid spherical indenter was modelled as an analytical rigid body. Subsequently, cartilage surface radii from 10 mm to 50 mm were modelled including both concave and convex shapes of the curve in order to examine possible effect of this curvature. The finite element models for the cartilage and bone are shown in Fig. 2.

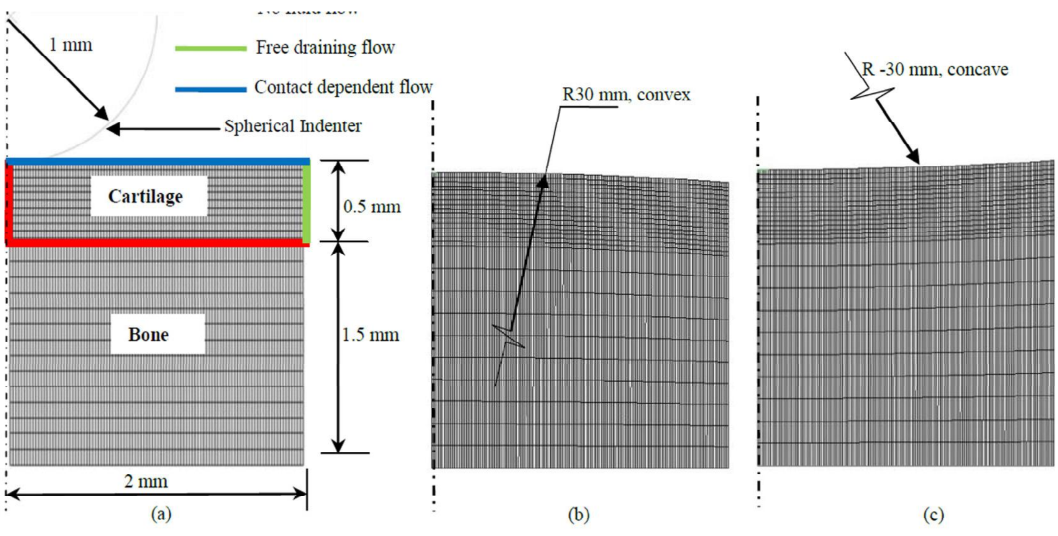

Fig. 2. Axisymmetric FE model of cartilage pin (a) flow boundary condition of cartilage, (b) convex cartilage curvature and (c) concave cartilage curvature.

Boundary and interface conditions were applied on the cartilage and indenter, to imitate the experimental creep indentation test set-up [10]. The bottom nodes of the bone were constrained in both horizontal and vertical directions, whilst the nodes on the axis were constrained in the horizontal direction. The spherical indenter was only permitted to move in the vertical direction, as the horizontal direction and rotational movements were constrained. For the cartilage fluid flow, as illustrated in Fig. 2a, flow was prevented at the bottom and vertical symmetry axis of the cartilage surfaces whilst the outer edge nodes of the cartilage were maintained at zero pore pressure so as to allow unrestricted fluid flow. For the upper cartilage surface, the contact dependent flow algorithm was imposed where the flow conditions at the nodes were changed depending on the contact stress. Table 1 shows the material properties applied in the finite element models.

Table 1: Material properties for FE model

Parameter	Value	Reference
Young's modulus, $E_{cartilage}$	0.76 MPa	[10]
Poisson's ratio, $\nu_{cartilage}$	0.08	[9]
Permeability, $\kappa_{cartilage}$	1.61 x 10^{-15} m^4/Ns	[10]
Void ratio, $e_{cartilage}$	3.75 (75% interstitial fluid)	[3]
Young's modulus, E_{bone}	1510 MPa	[11]
Poisson's ratio, ν_{bone}	0.3	[11]

In order to simulate the creep-deformation phenomenon, a ramp load from 0 to 0.24 N was applied on the indenter for 2 s, and the load was then maintained at 0.24 N for a further 1000 s [10]. The 2 s ramp period was based on an experimental study, which found that the minimum time at which creep compression test of the cartilage could be compared reliably was 2 s after the application of the load [12]. The 2 s ramp load was also used in previous computational studies [6,9,10]. Although the automatic time increments were applied in the model, the increments were controlled using the UTOL parameter, which specified the allowed maximum pore pressure change in one increment at a typically small value of 600 kPa so as to produce acceptable results [6,9,10].

3. Results and Discussion

The cartilage surface curvature study was carried out to observe the effects on the cartilage contact stress and pore pressure during creep indentation test using axisymmetric FE model. Both concave and convex curvatures were evaluated with the cartilage surface radii from 10 mm and 50 mm. Fig. 3 and Fig. 4 show the variation in contact pressure and pore pressure over the cartilage surface at 2 s and 1000 s respectively. The negative and positive values of the radius represent the concave and convex curves respectively.

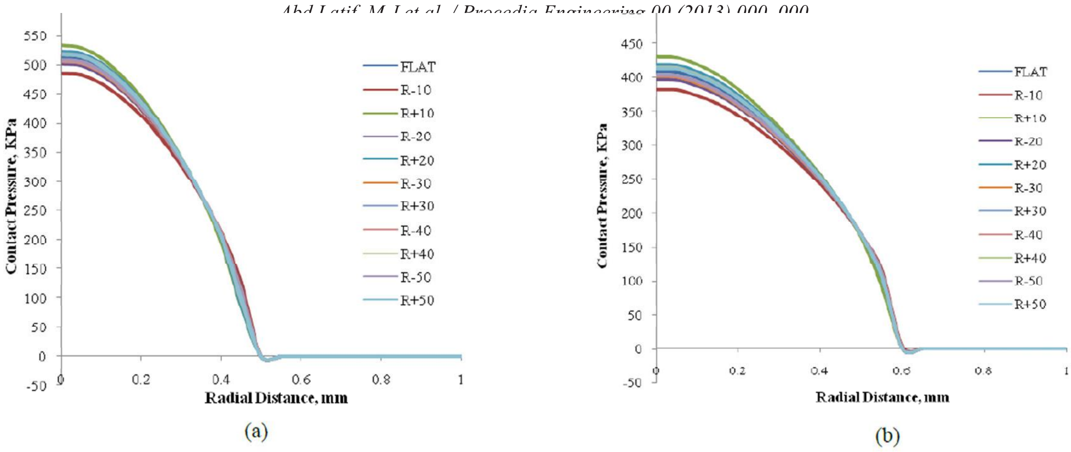

Fig.3. Distribution of contact pressure at the cartilage surface for different cartilage surface radii after (a) 2 s and (b) 1000 s.

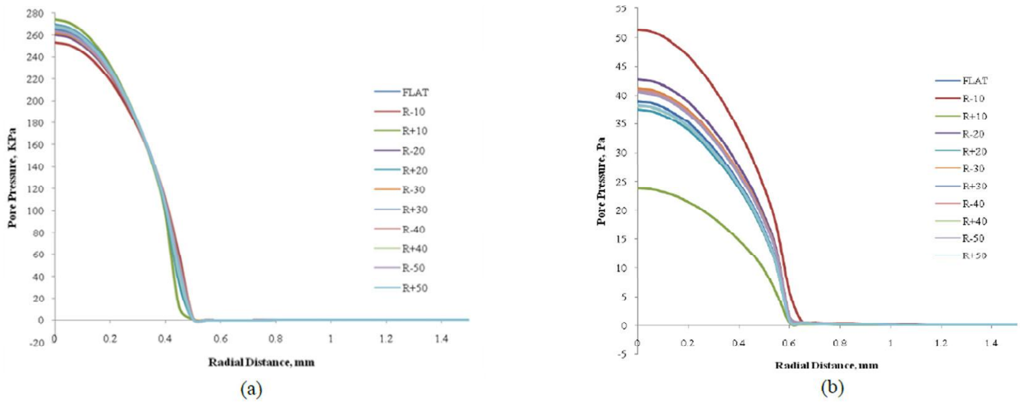

Fig.4. Distribution of pore pressure at the cartilage surface for different cartilage surface radii after (a) 2 s and (b) 1000 s.

Based on the results, the cartilage curvature had an effect on the contact pressure and pore pressure of the cartilage. The smaller cartilage surface of 10 mm radius produced higher difference of the cartilage behavior compared to the 50 mm radius. At 1000 s, the 10 mm radius cartilage surface generated 6% difference in the contact pressure and 39% difference in the pore pressure, compared to the flat cartilage. This is in agreement with a computational study carried out by Holzapfel and Stadler (2006), which reported that the cartilage curvature was crucial and played an important role in the load-bearing characteristics study [13].

The results indicate that the cartilage pore pressure would be most affected by the cartilage curvature during the indentation test. This could be due to the frictional drag force of interstitial fluid flow being the dominant factor controlling compressive creep behavior. The biphasic model described the nature of the cartilage in solid and fluid phases where the water content could reach to 85% [1,2]. Due to the very low permeability of the cartilage, large drag forces were generated from the fluid flow which maintained the high fluid pressure over a long period of time [1,2]. Fig. 5 shows the distribution of the pore pressure throughout the cartilage layer for flat and curve surfaces.

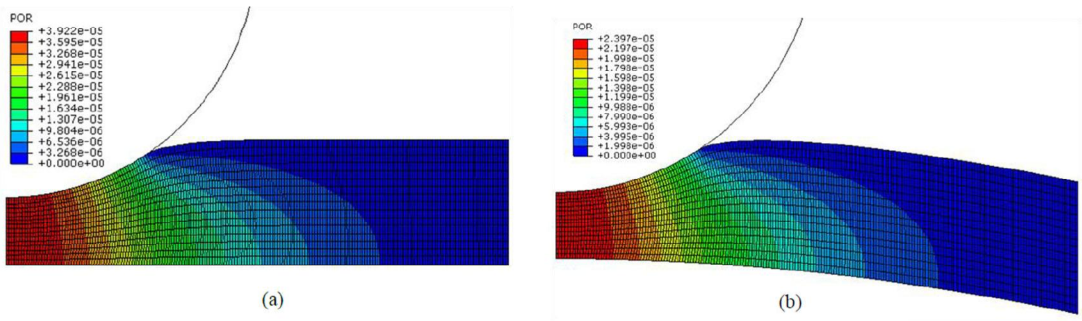

Fig.5. Distribution of pore pressure throughout the cartilage layer at 1000 s (a) flat surface (b) convex surface with 10 mm radius.

From the axisymmetric FE model, the cartilage behavior were affected at the cartilage surface radius of 20 mm was applied and became more severe when the radius was decreased. At this stage, it is not known at which specific level of curvature modeling becomes necessary, and further investigation is required. However, the axisymmetric model used to characterise the properties was seen to include some error in the estimated cartilage properties since it failed to represent the actual curvature of the cartilage surface. Therefore, further study is needed using specimen-specific model to replicate the actual three-dimensional curvature of the cartilage surface to simulate the indentation test.

4. Conclusion

This study presented the effect of the cartilage surface curvature to the contact stress and pore pressure of the cartilage during indentation test using axisymmetric FE model. The assumption that the cartilage pin specimen has a flat surface in the axisymmetric FE model will limit the accuracy of the cartilage properties. In this study, the difference of the cartilage behavior in the indentation test simulated using the curved surface with 10 mm radius were found to be 39% for the pore pressure and 6% for the contact pressure, compared to flat surface. Based on these results, it clearly indicates that the cartilage curvature does affect the cartilage behavior in the creep-deformation simulation. However, the curvature produced from the axisymmetric model does not well represent the three-dimensional curvature of the actual cartilage pin specimen. This study could be extended using specimen-specific model to model the actual three-dimensional curvature of the cartilage pin.

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