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Computational and Experimental Study of Articular Cartilage Thickness on Biomechanical Behavior

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Abstract

Degeneration of articular cartilage in synovial joint has long been recognized as the main source of osteoarthritis (OA). It is generally accepted that the biomechanical properties of articular cartilage was more sensitive to pathologies changes of the tissue. In previous studies, the biomechanical properties of the cartilage have been characterized based on the idealized physical conditions of the cartilage. This may contribute to inaccuracy of the results because the articular cartilage are geometrically inhomogeneous across the synovial joint. Therefore, the aimed of this study is to examine the effect of the cartilage thickness on biomechanical behavior of the cartilage tissue using computational and experimental methods. Cartilage specimens obtained from bovine humeral heads were used to conduct creep indentation tests. In computational study, axisymmetry poroelastic finite element (FE) models were developed based on the measured thickness and curvature for each of the specimen to observe the contact pressure and pore pressure of the cartilage. Based on the individual thickness of the specimen, the cartilage biomechanical properties of elastic modulus and permeability were then characterized by integrating the FE model and creep indentation test. The FE analysis shows that the cartilage thickness was observed to be more significant effect on the pore pressure compared to the contact pressure. Moreover, the cartilage thickness was crucial in characterizing the properties where the effect on the characterized elastic modulus and permeability could reached to 150% and 119% respectively. This findings show the importance of the cartilage thickness in order to study the biomechanical behavior of articular cartilage across synovial joint.

Keywords: Articular cartilage, thickness, biomechanical properties, finite element

INTRODUCTION

Osteoarthritis (OA) also known as degenerative joint disease is one of the common major causes of disability [1],[2],[3]. It was reported that 15% of the world's population suffered from OA where 25% of the patients were over 55 years old [4]. The typical early symptom of OA is the limitation of the joint movement where the patient have difficulties in their movement in daily activities which then lead to chronic joint pain [2]. Although there are various causes of OA, it is well known that degeneration of articular cartilage is the most common cause of OA [5],[6]. The treatments of OA are based on the condition of the OA patient from physical activities, medications and physiotherapies. However, severe OA patient requires prosthetic replacement of the articular surface known as arthroplasty [7].

Articular cartilage is a thin layer tissue which covers the end surface of articular bone in a synovial joint. It has no blood supply or nerve system and serves an important role as load absorption in the human body [5]. Through the thickness of the cartilage, there are four distinct zones with different composition and characteristic at each layer [8]. The superficial zone at the outer layer is 10-20% of articular cartilage thickness and consists the highest water content compare to other zones [9],[7]. The middle zone represents 40-60% of the cartilage volume and contains more proteoglycans which provides swelling pressure and highly stable hydrated structure [10]. The deep zone represent 30% of the cartilage volume is responsible in providing the greatest resistance to compressive forces, while the last layer is the calcified zone that plays an integral role in securing the cartilage to bone by anchoring the collagen fibrils to subchondral bone [9]. The flow of the water through the cartilage is to provide lubrication and help to transport and distribute nutrients. Further experimental studies have shown that the articular cartilage consists of two phases which are 15-22% solid phase and 60-85% fluid phase [11]. This structure contributes to inhomogeneous and uniqueness of the cartilage behavior.

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In previous computational models, various constitutive material formulations have been used to describe cartilage in finite element (FE) models from single-phase to multiphase. These material formulations were applied to study the mechanical function of synovial joint under different loading and pathological conditions. The single-phase could describe the mechanical behavior of the cartilage under static, instantaneous and equilibrium conditions [12]. However it is unable to describe the time-dependent creep and stress-relaxation behaviors of the cartilage. The behavior of the cartilage is best understood when the tissue is represented as biphasic material which consists of solid and liquid phases [13], [14].

Two important biomechanical properties of cartilage in biphasic theory are elastic modulus and permeability. Permeability indicates the rate of fluid flow of the tissue and it is not constant throughout the tissue, where it is higher near the cartilage surface and lower in the deep zone. Elastic modulus is the measure of the stiffness of cartilage tissue. It is important to characterize the biomechanical properties of cartilage accurately because it provide reference data for tissue development and regeneration, biomaterial development, the functional assessment of engineered and repaired tissues, and functional imaging [15]. Furthermore, the properties can be used to monitor the conditions and changes in cartilage before any visual changes can be detected [16].

The biphasic biomechanical properties have been characterized using a combination of experiment and numerical methods. Recent computational advancement has utilized the FE method to integrate with the experiment method [14], [17]. The experiment methods that are commonly used are unconfined compression, confined compression and indentation test. However, the indentation test is widely used because it provides a simple specimen preparation procedure and is close to physiological constraints in deformation and fluid flow as it is still attached to the supporting bone [13], [18].

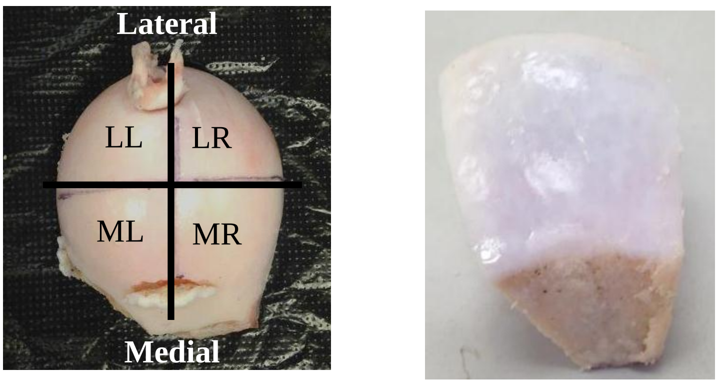
In previous studies, computational models were based on generalized cartilage thickness [19],[20]. Moreover, the cartilage was also represented as idealized geometrical shape to characterize the biomechanical properties [19]. Studies have found that the thickness of the cartilage varies at different synovial joint and different locations in the same joint [21]. This may contribute to inaccuracy of the computational results and characterized properties because of the physical nature and inhomogeneity across articular cartilage. Therefore, the aim of this study is to examine the effect of thickness on biomechanical behavior of cartilage using computational and experimental methods.

METHODOLOGY

The biomechanical behavior of cartilage tissue was examined using both experiment and computational methods. For the experimental study, the creep indentation test was utilized while the cartilage thickness and curvature measurements were carried out mainly to develop FE models. In computational method, FE models were developed to simulate the indentation test using the Abaqus 6.9-1 (DS Simulia Corp., Providence, RI, USA) software. The biomechanical properties of cartilage were characterized using the combination of the data from creep indentation test and FE simulation.

Specimen Preparation

Twelve cartilage specimens (*n=12*) abstracted from bovine humeral head aged between three to five years old obtained from local abattoir within 24 hours after slaughtered. The bovines are local breed species (Kedah Kelantan, KK) with an average weight of 200-250 kg. The femoral head was then cut into four cartilage specimens and labelled as lateral left (LL), lateral right (LR), medial left (ML) and medial right (MR) as shown in Figure 1. The cartilage surface was kept moist with phosphate buffered saline (PBS) solution during the cutting process in order to avoid sample dehydration. The specimens were then stored at 7ºC in moist condition within 48 hours prior testing.



(a) (b)

Figure 1: (a) Articular cartilage of femoral head (b) Cartilage specimen.

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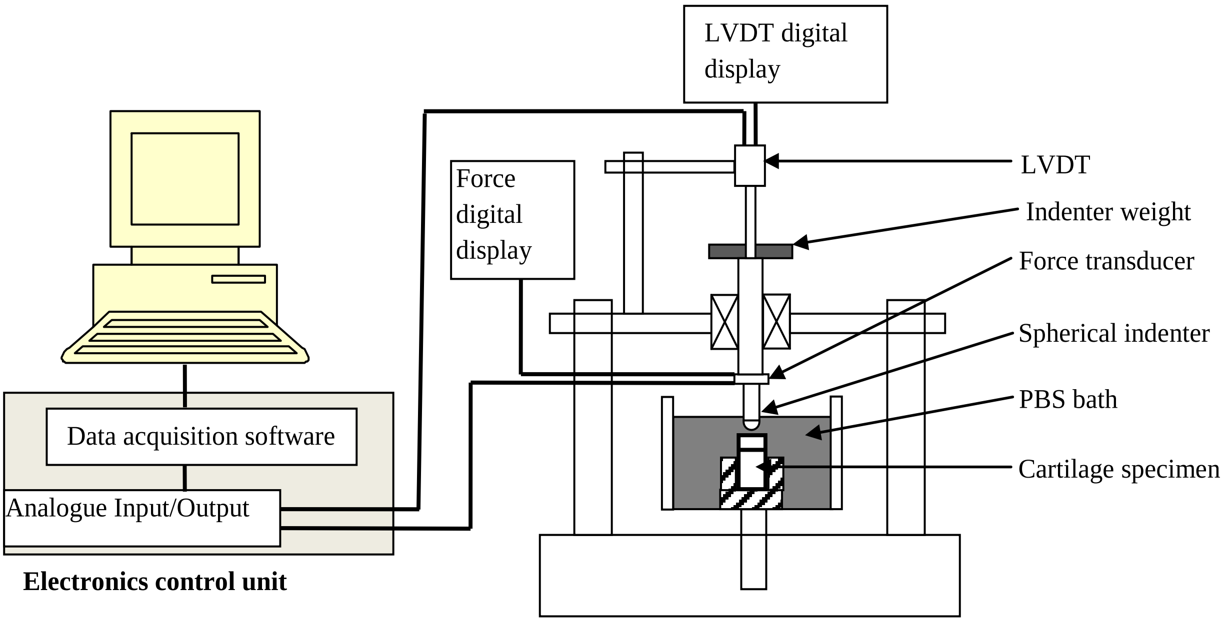


Figure 2: Schematic diagram of indentation test apparatus.

Cartilage Creep Indentation Test

Creep indentation test was carried out using the indentation apparatus (Figure 2) and the procedure have been described previously [22], [23]. The cartilage specimen was immersed in PBS solution during the entire period of the testing to avoid any dehydration and to maintain the environment of the cartilage in synovial joint. Impermeable spherical indenter with 4 mm diameter and 0.38 N compression load were used to intend the cartilage surface which yielded 10% to 20% cartilage strain of the cartilage thickness. The displacement was recorded continuously at a sampling time of 0.01 seconds for 1000 seconds where the displacement had reached the equilibrium state. All the data were recorded using the data acquisition software (LabVIEW 8.5.1, National Instrument Corporation, Austin, TX, USA). The deformation and time data of cartilage will be used to characterize its biomechanical properties of elastic modulus and permeability.

Cartilage Thickness Measurement

The specimen was allowed to equilibrate in PBS solution for an hour to ensure that the cartilage reforms to its initial shape after the creep indentation test. The thickness measurement was carried out subjected to 3.16 N loads indented to the point marked on the cartilage surface using needle indenter. The sharp needle indenter was used to penetrate the cartilage tissue until it reaches the underlying bone. The displacement of the needle and the load response were recorded to determine the cartilage thickness. The displacement and load reading were recorded every sampling 0.001 seconds to enable an accurate measurement of the cartilage thickness. The thickness was obtained by determining the difference of the displacement between the contact points of the needle to the cartilage surface and to the subchondral bone which have been described previously [22].

Finite Element Modelling

The articular cartilage and bones were modelled as axisymmetric biphasic poroelastic model was developed to replicate the creep indentation test using the Abaqus 6.9-1 (DS Simulia Corp., Providence, RI, USA) as shown in Figure 3. The cartilage models were developed based on the measured radius ranging from 0.4 mm to 1.4 mm and measured thickness with 5 mm width. The 4 mm diameter spherical indenter was modelled as an analytical rigid surface. The element types for cartilage were set as four node bilinear displacement and pore pressure elements (CAX4P), while four-node bilinear displacement and pore pressure elements (CAX4) were used to represent the underlying bone [20],[24].

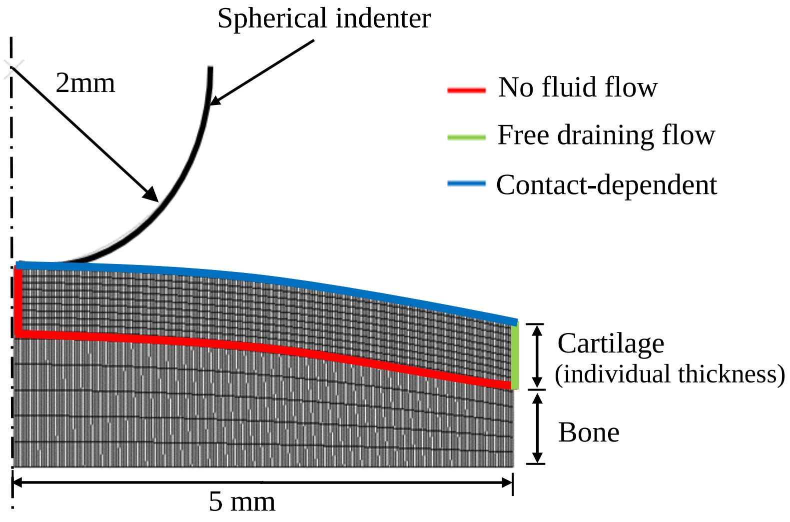


Figure 3: Axisymmetric FE model of curved-surface cartilage with flow boundary conditions.

The boundary and interface conditions were applied on the cartilage and indenter to replicate the experiment creep indentation test set up. The bottom nodes 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 allowed to move in the vertical direction, while the horizontal and rotational movements were constrained. The contact-dependent flow condition was applied at the top surface of the cartilage [20]. The fluid flow was prevented at the bottom and vertical symmetry axis of the cartilage surface whilst the outer edge nodes of the cartilage were maintained at zero pore pressure to allow unrestricted fluid flow. This was implemented at each increment such that flow was prevented only in the region when it was in contact with the cartilage surface. The material properties applied in the FE model are shown in Table 1.

Table 1: Material properties for the FE model [24]

Parameter	Value
Young's Modulus, $E_{cartilage}$	0.54 MPa
Poisson's ratio, $v_{cartilage}$	0.08
Permeability, $k_{cartilage}$	4.0×10^{-15} m^4/Ns
Young's Modulus, E_{bone}	2000 MPa
Poisson's ratio, v_{bone}	0.2

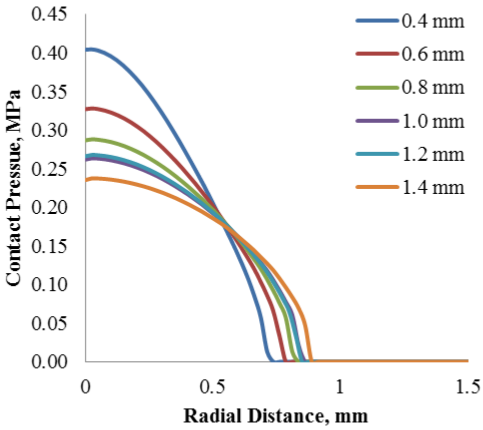
In order to simulate the creep-deformation phenomenon, 0.38 N ramp load based on the indentation test experiment was applied on the indenter for 2 seconds and maintained for a further 1000 seconds. The 2 seconds ramp period is an instantaneous effect was based on experimental studies, which found that the minimum time at which creep compression of the cartilage could be compared reliably was 2 seconds after the load was released. Automatic time increment was used with UTOL parameter, which specified the allowed maximum change in pore pressure in one increment at 600 kPa to produce acceptable results [20]. The verification of the model was performed and described previously [24],[13].

Characterization of Biomechanical Properties of the Articular Cartilage

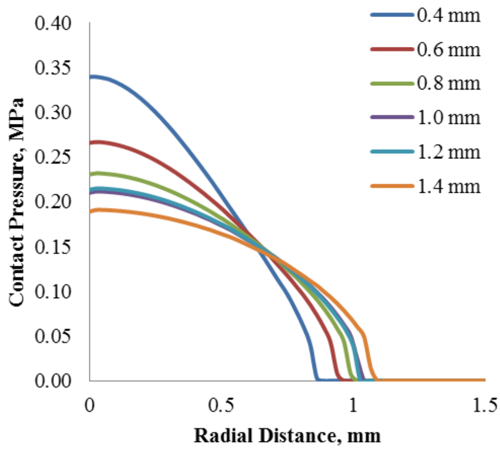
Data generated from the FE model and creep indentation test were integrated in order to characterize the biomechanical properties of elastic modulus and permeability. The initial values of the properties were iteratively altered until the deformation-time curve generated from the FE model matched to the experiment results. Non-linear least-square method was utilized to perform the curve-fitting using 'lsqnonlin' function in Matlab software (V7.12.0 R2011a, MathWorks Inc, MA, USA).

RESULTS AND DISCUSSION

The thickness of the cartilage were measured at lateral left lateral right, medial right and medial left sides on the humeral head in order to examine the thickness variation across the articular cartilage. Based on the result shown in Table 2, it was observed that the thickness varies across the articular cartilage ranging from 0.69±0.38 mm to 0.90±0.47 mm. The average thickness of the cartilage was 0.85±0.42 mm which is within the range to the previous studies [25], [26].



(a)



(b)

Figure 4: Distribution of contact pressure at the cartilage surface for different thickness after (a) 2 seconds (b) 1000 seconds

Table 2: Articular cartilage thickness of bovine humeral head.

Position	Thickness (mm)
Lateral left (LL)	0.70±0.33
Lateral right (LR)	0.83±0.51
Medial left (ML)	0.69±0.38
Medial right (MR)	0.90±0.47

The range of the thickness was then used in FE analysis to examine the biomechanical behavior. Figure 4 shows the FE analysis of contact pressure at 2 seconds and 1000 seconds. Based on the results, it was found that the 0.4 mm cartilage thickness generated 52% higher compared with 1.4 mm at 2 seconds. This trend was similar for 1000 seconds where the percentage is slightly higher with difference of 56%. The graph show the maximum thickness has the lowest contact pressure for both 2 seconds and 1000 seconds. This is due to the higher water content that decreased the concentration of collagen in the load-bearing region.

The effect of thickness on pore pressure was also observed in FE analysis to examine the state of the fluid flow throughout the cartilage tissue. The 0.4 mm cartilage thickness generated 63% difference compared to 1.4 mm at 2 seconds as shown in Figure 5. However, at 1000 seconds, opposite trend was observed where the 1.4 mm thickness generated the highest pore pressure. This is due to the fluid flow spread towards the cartilage surface and attempts to flow out of the tissue, which pressurizes the fluid and allows the load-bearing. Based on these results, the pore pressure was observed to be more significant effect on the cartilage thickness as compared to the contact pressure.

Further investigation was carried out to examine the characterized biphasic biomechanical properties of the cartilage where the average elastic modulus and permeability of the cartilage were 1.23±0.91MPa and 1.33±0.54 $\times 10^{-15}$ m^4/Ns respectively. These properties were within the range compared with the previous reported values [27], [28]. The characterized properties based on the thickness of the cartilage are shown in Figure 6. The cartilage thickness does give an effect to the characterized biomechanical properties where the elastic modulus increased when the thickness was increased and the difference of the value could reached to 150% between cartilage thicknesses. However, opposite trend was found for the permeability where the difference of the value could reached to 119% between cartilage thicknesses.

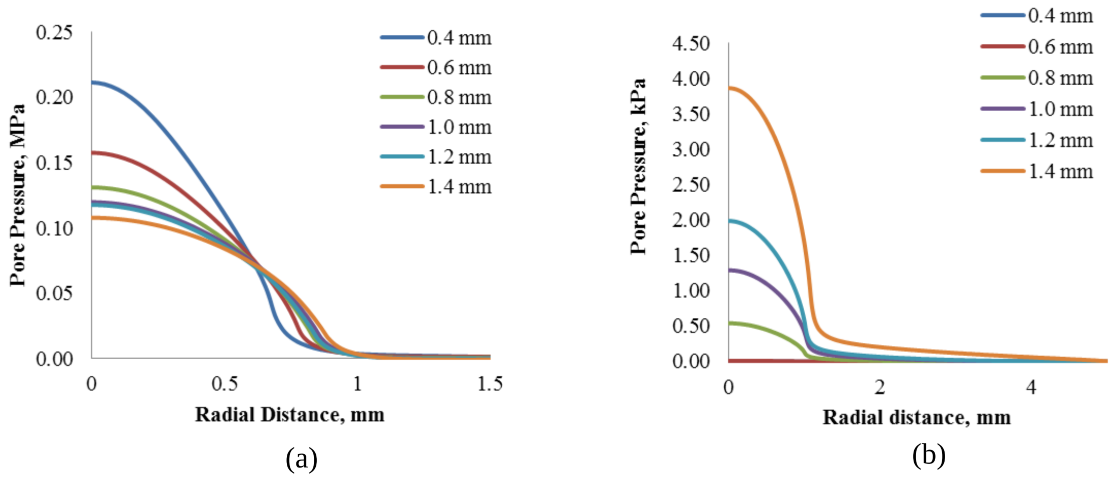


Figure 5: Distribution of pore pressure at cartilage surface for a different thickness after (a) 2 seconds (b) 1000 seconds

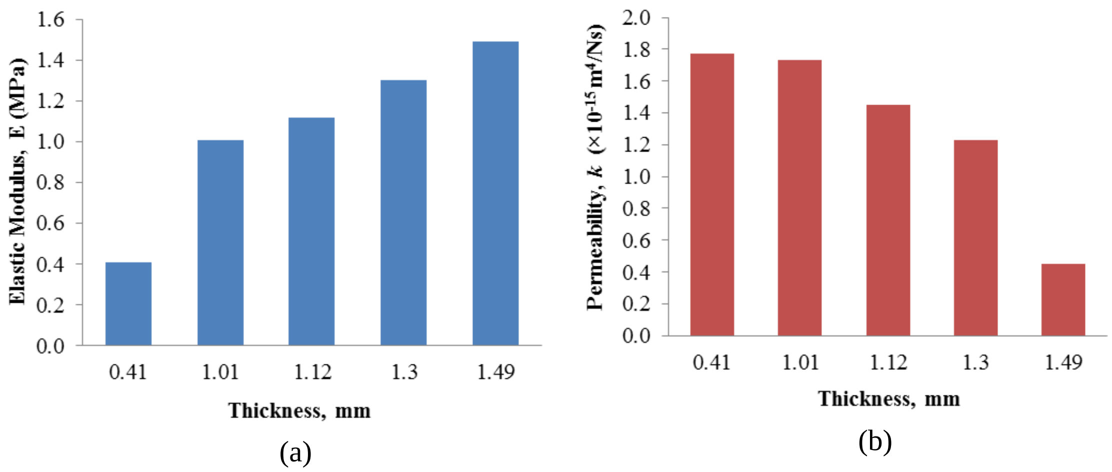


Figure 6: Cartilage biphasic properties characterized from different cartilage thickness (a) elastic modulus (b) permeability

These indicate that it is important to consider the actual thickness of articular cartilage in FE modelling and could be crucial in characterizing the biomechanical properties. Furthermore, studies have found that the thickness of the cartilage varies at different synovial joint and different locations in the same joint [21], [29].

CONCLUSIONS

This study has examine the biomechanical behavior of cartilage tissue across a synovial joint of bovine humeral head using experimental and computational methods. The findings of this study show that the thickness of cartilage does have an effect on the biomechanical properties of articular cartilage across a synovial joint. By considering the thickness and curvature of the cartilage, the biomechanical properties are able to be characterized more accurately. The good sensitivity of the cartilage thickness could be used for further investigation into the pathology of cartilage degeneration.

NOMENCLATURE

OA : Osteoarthritis

FE : Finite Element

PBS : Phosphate Buffered Saline

LL : Lateral Left

LR : Lateral Right

ML : Medial Left

MR : Medial Right

k : Permeability

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