An estimation of mechanical properties of articular cartilage for biphasic finite element analyses

Nobuo SAKAI*, Yuichiro HAGIHARA**, Chie HASHIMOTO*, Mochimitsu KOMORI*, Yoshinori SAWAЕ*** and Teruo MURAKAMI****

*Graduate School of Engineering, Kyushu Institute of Technology
1-1 Sensuicho, Tobata-ku, Kitakyushu 804-8550, Japan
E-mail: sakai.nobuo@iise.kyutech.ac.jp
**JFE Engineering Corporation
1-8-1 Marunouchi, Chiyoda-ku, Tokyo 100-0005, Japan
***Faculty of Engineering, Kyushu University
744 Motooka, Nishi-ku, Fukuoka 819-0395, Japan
****Research Center for Advanced Biomechanics, Kyushu University
744 Motooka, Nishi-ku, Fukuoka 819-0395, Japan

Received 7 April 2015

Abstract
The purpose of this study is to estimate material properties of articular cartilage by curve fitting method using finite element (FE) analysis. While various material tests have been conducted to predict the behavior of articular cartilage, one of the recent interests was the accurate estimation of material properties under physiological and dynamic conditions. In this study, cylindrical indentation was experimentally conducted in high compressive amount and high compression rate in considering the physiological condition. Each single specimen was sequentially exposed to compressive tests at definite deflection with different compressive amount and different compression rates and compressive creep test. The time-dependent compressive force given by a precise compression tester was utilized for estimation of material properties by curve fitting method with FE analysis. Five typical material properties, which represented total apparent Young's modulus, strain dependent permeability and fibril reinforcement by collagen network, were selected for the estimation process with depth-dependency of Young's modulus. In the curve fitting processes by FE analysis, each material property had specific roles on reproducing experimental time-dependent reactional force. A single material property set estimated in this study successfully reproduced the four different experimental time-dependent behaviors.

Key words: Articular cartilage, Biphasic compression, Finite element model, Material property

1. Introduction

In human body, diarthrodial joints especially in lower extremity have to manage large contact force up to more than 10 times body weight, while relative rubbing speed subsides nearly stationary. Even in these severe conditions, the soft articular cartilage can operate with low friction coefficient of normally less than 0.03. The study on the mechanics of the cartilaginous tissue is usefully applicable to medical insight for joint diseases, cultivating conditions of regenerative tissue and engineering interest as a load bearing system. Articular cartilage is a soft gel material containing high water content (70-80 % by weight) in intact state. This water content is partly derived from aggregate proteoglycan, which induces osmotic pressure for swelling behavior. The matrix contents are enmeshed in type II collagen fibril network. The proteoglycan matrix mainly supports compressive load in equilibrium condition, while the collagen fibril network resists tensile load. Tensile stiffness of articular cartilage is experimentally much higher than compressive stiffness in equilibrium condition (Huang et al., 2003; Chahine et al., 2004). Articular cartilage shows compressive strain inhomogeneity (Schinagl et al., 1997; Erne et al., 2005) and anisotropy (Jurvelin et al., 2003; Wang et al., 2003) along...
depth direction.

Articular cartilage has several mechanisms to reduce friction coefficient by synergistic cooperation of lubrication modes, which is called "adaptive multimode lubrication" (Murakami et al., 1998). The surface constitution of synovial or regenerated cartilage enhances tribological properties (Graindorge et al., 2006; Klein, 2006; Katta et al., 2008; Murakami et al., 2009; Yarimitsu et al., 2009; Murakami et al., 2011; Nansai et al., 2011). Bulk property of articular cartilage also contributes frictional behavior. The time-dependent compressive behavior was well explained by "biphasic model" (Mow et al., 1980), in which the gel matrix was expressed by solid and fluid phases. Further biphasic complex lubrication system of articular cartilage is called the "biphasic lubrication mechanism" (Ateshian, 2009). When the counter surface contacts to the biphasic material, the interstitial fluid is trapped within an apparent contact area and pressurized by the contact load. As the interstitial fluid begins flowing along pressure gradient, the fluid presses the solid phase. However the tissue in the contact area is going to stretch laterally, the solid phase is well reinforced to the tensile direction by collagen network. This situation causes interstitial fluid pressurization, which supports considerable proportion of total load even in unconfined condition (Park et al., 2003). The reduction of load support by the solid phase subsequently causes low friction coefficient (Ateshian, 1997). This phenomenon was experimentally proved by the direct measurement of fluid pressure (Krishnan et al., 2004).

In addition to experimental studies, biphasic finite element (FE) analysis advanced the understandings of the material behavior and its frictional functionality. While more detailed models have been continuously researched, the experimental material tests in these studies were often carried out in small compressive amount or slow speed. Behavior of articular cartilage in physiological condition was also concerned beyond the material testing level. In experimental studies, physiological compressive speed and load was applied to assess cartilage viability (Luzchinetti et al., 2002). Except for impact tests, dynamic cyclic load was used to assess the relation between the load and friction coefficient (Krishnan et al., 2005). By advance of experimental apparatus, several experimental reports especially emphasized the importance of the high-speed compression, e.g. walking condition (June and Fyhrie, 2010; Espino et al., 2014). We should note that 3-dimensional FE model is also one of the physiological conditions. The 3-dimensional FE analyses have been studied with their own experimental material test (Gu and Li, 2011; Pawaskar et al., 2011; Henak et al., 2014; Osawa et al., 2014). The sliding condition with migrating contact area (Pawaskar et al., 2007; biphasic FE model; Caligaris and Ateshian, 2008: experimental study) is the other dynamic condition.

While many reports on the FE modeling of articular cartilage were conducted by small compression amount or slow speed, Li et al. used initially verified material properties, which captured the "trends" of the experimental tangential stiffness varying 0-20 % strain at 0.5-50 %/s strain rate by a FE model (Li et al., 2003). Goreham-Voss employed experimental cyclic load in 0.1-40 Hz for verifying a FE model of which the material properties were gathered elsewhere (Goreham-Voss et al., 2007). Mäkelä et al. used well defined FE model to estimate the mechanical properties of osteoarthritic articular cartilage in 5% strain step at 100 %/s (Mäkelä et al., 2012). Deneweth et al. proposed a FE model for curve fitting of 16 % experimental stress-strain curve at 100 %/s. (Deneweth et al., 2013). On the other hand, Gao employed the creep behavior for analytical model fitting (Gao et al., 2014), as past used (Soltz and Ateshian, 1998). In our previous study, experimental compressive behaviors of 10 % strain at strain rate of 5, 2 and 1 %/s were used to obtain material properties (Sakai et al., 2012a), because this research specially focused on specific sliding speed to compare with the other report. Although Soltz and Ateshian employed an experimental data set including 10 % strain and the creep behavior for curve fitting (Soltz and Ateshian, 1998), the compressive speed was at 0.25 μm/s. To represent more various compressive behaviors by FE model, we experimentally conducted a precise compression test, which included different compressive amount, different speed and creep behavior. The time-dependent experimental data set was applied for curve fitting of a FE model. The result showed a single material property set for the FE model successfully reproduced the experimental data set.

2. Methods

We briefly summarize the experimental procedure before moving to detailed descriptions follows. In experimental study, bar shaped specimen of articular cartilage was prepared without any freezing process. In compressive experiment, a cylindrical indenter was employed for further experimental sliding test (Pawaskar et al., 2007). The compressive test was executed by high-precision and high-speed compression tester equipped on confocal laser scanning microscope. One specimen was exposed to 4 sequential experiments in considering different compressive
amount, different compressive rates and the creep compression. The time-dependent reactional forces were gathered and averaged for the curve fitting of the FE analysis. The depth-dependency of compressive modulus determined in our previous report was applied to the model. In the curve fitting procedure, a single specific material property set was identified as a result.

2.1 Experimental apparatus and materials

To reproduce physiological high-speed compressive speed, a self-made high-precision and high-speed compressive tester with force control was developed, as shown in Fig. 1. Both of the compressive plates were made from impermeable alumina ceramics (CERATIP SNGN120704, Kyocera Corporation; material: KA30), which were rigidly fixed in mechanical parts. The cylindrical indenter with 5 mm radius, which was made from stainless steel (SUS316L, impermeable solid) with mirror finish, was fixed on one side of the compressive plates. The compressive plate was driven by a ball screw actuator (KR-15, THK) with a DC servomotor (MR-J2-JR, Mitsubishi Electronic Corporation; power: 20 W). The reactional force was measured by a load cell (LUR-A-50NSA1, Kyowa Electric Instruments), which was serially placed between the compressive plate and the actuator. Since load cell principally detects force by its deformation, which disturbs accuracy of the position of the compressive plate, eddy current displacement sensor (E2CA-X1R5A, Omron Corporation; operating range: 1000 µm) was used to detect true position of the plate. Position feedback control with the displacement sensor precisely actuates the compressive plate with accuracy of 0.5 µm and maximum speed of 4000 µm/s in any loading condition. For the creep test, additional force feedback control was implemented to generate constant compressive force, in which the compressive plate follows compressive deformation of the specimen. Sampling rate of the data acquisition and the feedback control was set at 333 Hz. The compressive tester was installed on the self-made steady X-Y stage of a confocal laser scanning microscope (CLSM, C1-Plus, Nikon Corporation) to measure initial depth of each tissue and to observe strain field of the tissue through a glass plate. To eliminate frictional artifact, the clearance between the glass plate and moving part was modulated to about 20 µm. The friction of linear slider for mounting moving part was minimized by linear ball slide (LS852, THK; µ = 0.0006-0.0012 from data sheet).

Experimental specimen was dissected from distal end of femur of porcine knee joint (6-8 months old) without any freezing process to prevent a certain effects on the mechanical properties due to freezing (Willett et al., 2005). In this study, the cylindrical indenter was chosen for further sliding condition. So, a bar shaped osteochondral explant with the dimension of 2.5 mm width and 10 mm length was dissected by a scalpel blade with a brief jig, shown in Fig. 2. The bar shaped explant was cut with the longitudinal direction of the specimen being parallel to physiological anterior-posterior line in chondyle. The specimens were shortly rinsed in PBS (Phosphate Buffered Saline, Invitrogen Corporation; pH 7.4) solution, and living cells were stained with Calcein-AM (C-3099, Molecular Probes) to measure certain depth of each tissue. Then, the specimens were incubated for 30 min staining and washed by PBS solution.
again. The specimens were kept in PBS solution before and after a single experiment. In the compressive experiment, the specimen was also kept in well water-soaked condition of PBS solution throughout the experiment to maintain hydration and osmotic phenomena. The compressive experiment and reservation state between experiments were executed in air-conditionated room temperature of 23°C, in which it was reported that articular cartilage drastically changes mechanical stiffness by exposing to around 60°C (June et al., 2010).

2.2 Method for cylindrical indentation test

The cylindrical indentation was conducted to obtain time-dependent reaction force and the creep deformation. In this configuration, the effect of surface seepage from intact tissue surface would be involved in the reactional response. The osteochondral specimen was put on the glass plate, in which the intact surface of the cartilaginous tissue faced to the indenter. The specimen was soaked in PBS solution during experiment. Prior to the compression test, the depth of the cartilaginous tissue of each specimen was measured by the image length of microscopic view as shown in Fig.2 (b), because the specimen was dissected with subchondral bone. The depth of cartilaginous tissue was defined by the distance between the initial surface and the one-third position from the bottom of the tidemark. The average depth of cartilaginous tissue of the specimens was 1.48±0.23 mm (Mean±SD). Measured depth was used to calculate compressive displacement for specific compressive strain. The drawing of Fig.2 (b) shows the compressive setup from the microscopic view shown in Fig.1. So the curved surface of the cylindrical indenter pressed the intact surface of the specimen. Just before the compression, pre-compression by the peak reaction force of 0.05 N was allowed for 3 s for a stable contact, because the surface of the tissue involves inevitable small shape undulation or disorder along 2.5 mm width direction. Then, the initial load was removed, and the compressive test was immediately executed. In the definite displacement compression, the indenter compressed the specimen with the prescribed amount at specified strain rate. After the compression phase, the position of the indenter was kept constant by the feedback position control during the stress relaxation period. In the entire definite compression test, the compressive load or reaction force was recorded by control computer. For the creep test, initial setup was the same way with definite compression test. The indentation force was linearly raised up to the specified value in the first 1s and kept constant force by the force feedback control. Through the creep test, the displacement of the indenter by the force feedback control was recorded by control computer.

The compressive conditions were selected as follows. From our previous study (Sakai et al., 2012b), the peak stress in the definite compression did not increase with strain rate in the fast strain rate of more than 10 %/s. In this situation, it was thought that the interstitial fluid could not flow enough in the tissue within short compression time, and it suggested that the tissue behaved like a simple elastic material. So, the strain rate for the fast compression in considering physiological or dynamic condition was selected by 10 %/s. In the walking condition, the peak contact deformation of the articular cartilage of knee joint ranged from 7 % to 23 % (Liu et al., 2010). The compressive strain at the strain rate of 10 %/s were selected by 15 % and 10 % strain in considering deterioration of the tissue due to the

Fig. 2
(a) Shape of the experimental specimen.
(b) Configuration of cylindrical indentation from microscopic view.

A bar shaped osteochondral explant with the dimension of 2.5 mm width and 10 mm length was dissected by a scalpel blade. The bar shaped explant was cut with the longitudinal direction of the specimen being parallel to physiological anterior-posterior line of the chondyle. (b) Prior to the compression test, the depth of the cartilaginous tissue of each specimen was measured by the image length from surface to tide mark in microscopic view.
sequential experiment design, in which a single specimen was exposed to 4 sequential different compression. The compressive displacement means the displacement at the cylindrical apex of the indenter. Slow strain rate was set at 2 %/s, while compressive displacement remained by 10 %. Therefore, the total compression time of 15 % strain at 10 %/s, 10 % strain at 10 %/s and 10 % strain at 2 %/s were 1.5, 1.0 and 5 s, respectively. In this study, the total experimental time was limited within 30 s in considering tissue deterioration. The other experimental condition was the creep compression. The compressive force of 1 N/mm was selected, in which the force value was determined prior to the experiment in considering adequate compressive deformation for the curve fitting of subsequent FE analysis. Although the larger indentation force might partly reduce the variance, the deformation would also increase more than 20% beyond physiological deformation. Width of each specimen was measured to calculate the commanding force. It seemed that the creep test was the most severe condition within our experiment from the viewpoint of tissue deformation. The experimental sequence was arranged by the order of (1) 10 % strain at 2 %/s strain rate, (2) 10 % strain at 10 %/s, (3) 15 % strain at 10 %/s and (4) 1 N/mm creep test in considering the tissue deterioration. After each compression test, the specimen was well soaked in PBS more than 30 min to recover the shape and water content. The number of the specimen was N = 7.

2.3 Estimation of material properties by finite element analysis

The purpose of this study was to identify a single material property set, which could simultaneously represent the 4 different compressive behavior. From the experimental time-dependent data set, material properties of articular cartilage were estimated by curve fitting method using FE analysis. A commercial package of ABAQUS (6.8-4) was employed for FE analysis. It is thought that one of the physiological conditions is 3-dimensional model constructed by medical scanning images, in which the 3-dimensional model could deliver the effect of joint morphology. On the other hand, the comparison between experimental sliding test and FE analysis with a dynamic sliding motion would also still extract realization mechanism of functionalities of articular cartilage as a load bearing system. In considering further experiment, 2-dimensional model was accepted to easily introduce the sliding motion of cylindrical indenter in FE analysis. Dimension of the FE model was 1.5 mm thickness and 10 mm length as a typical dimension of the experimental specimens, as shown in Fig. 3. The cylindrical indenter was modeled by geometrical rigid body with 5 mm radius. The biphasic tissue was represented by CPE4RP (four-node bilinear displacement and pore pressure, reduced integration with hourglass control, plane strain-based element) elements with 0.1 mm square. So, the cartilage
was modeled by 100×15 elements. The boundary condition of the bottom of the cartilage model was positionally constrained and impermeable. Another surfaces were not constrained and were permeable except for contact area. Management of surface seepage in contact area was implemented by the user subroutine of the ABAQUS package using FLOW function, which can control surface seepage coefficient. If a node was contacted to the cylindrical indenter, the surface seepage coefficient was set to zero to prevent flowing. In the non-contact nodes, the surface seepage was set to one as a sufficient large value (Pawaskar et al., 2007). Compressive condition in FE analysis was as same as the experimental one. The equilibrium friction coefficient $\mu_{eq}$ (Ateshian, 1997) was set to zero, assuming that functional boundary lubrication remained well in the experimental specimens.

The inhomogeneity along the depth direction promotes the superficial fluid load support and improves frictional properties (Krishnan et al., 2003). The depth-dependency of the Young's modulus could be obtained by the depth-dependent strain in equilibrium condition (Schinagl et al., 1997; Erne et al., 2005). In equilibrium condition in uniaxial compression test, apparent strain $\varepsilon_0$ was experimentally measured with the relationship of

$$\sigma_0 = E_0 \varepsilon_0,$$

where $\sigma_0$ is apparent stress and $E_0$ is apparent Young's modulus. Under an assumption that local stress $\sigma(x)$ in the equilibrium condition has a constant value $\sigma_0$ in any depth position $x$ ($0 < x < 1$; the normalized position: $x = 0$ means the surface), the relationship between apparent strain $\varepsilon_0$ and local strain $\varepsilon(x)$ is given by

$$E_0 \varepsilon_0 = E(x) \varepsilon(x),$$

where $E(x)$ is the depth-dependent Young's modulus. Once the local strain $\varepsilon(x)$ was obtained by a microscopic observation, $E(x)$ was given by

$$E(x) = \frac{\varepsilon_0}{\varepsilon(x)} E_0 .$$

The apparent strain $\varepsilon_0$ is the value when $\varepsilon(x)$ was obtained. In the previous study (Hosoda et al., 2008), the local strain $\varepsilon(x)$ was experimentally acquired by

$$\varepsilon(x) = 0.462e^{-6.53x} + 0.0284 .$$

In equilibrium condition, the superficial tissue (in $x = 0$) was deformed with more than 40% strain by Eq. (2). Articular cartilage can aggregate by drawing interstitial fluid, where the solid phase would be assumed as a sponge material. So, we accepted that the solid phase was a linear elastic material as an apparent behavior up to 40% compression. The depth dependent Young's modulus (or compressive modulus) of solid phase was applied to the FE model. Poisson's ratio of the solid phase was set to 0.125 as a typical value (Sakai et al., 2012a). Indeed, the small Poisson's ratio have been applied or estimated as the value nearly 0 (Warner et al., 2001: as 0.08, Jurvelin et al., 2003: as 0.158 ± 0.148; Graindorge et al., 2006: as 0, Pawaskar et al., 2007: as 0).

Permeability of the interstitial fluid is known as a well strain dependent property. The relationship between the strain of the solid phase $\varepsilon$ and current void ratio $e$ is written by

$$\varepsilon = (e - e_0)/(1 + e_0)$$

with initial void ratio $e_0$. Strain-dependent permeability $k$ was incorporated in the FE model by

$$k = k_{min} + k_0 \exp \left( \frac{M e - e_0}{1 + e_0} \right),$$

where $k_{min}$ is a lower limit of the permeability, $k_0$ is a component of strain dependent permeability and $M$ represents the compaction effect. The second term of Eq. 3 is widely utilized as a typical model (Li et al., 1999). In the past, the lower limit of the permeability was proposed for estimating material properties of articular cartilage (Jurvelin et al., 2003). In this study, $k_{min}$ was introduced mainly for acceptable curve fitting of the creep behavior as a minimum value. The initial void ratio $e_0$ was set to 4.0, which means 80% interstitial fluid content.

Collagen fibril reinforcement brings the sharp peak stress in the definite compression (Li et al., 1999). The fibril in the matrix was assumed to resist only tensile strain of the solid phase. It was reported that deep vertical collagen fibril plays an important role especially in dynamic condition (Shirazi and Shirazi-Adl, 2008; Hosoda et al., 2010). In this study, horizontal and vertical fibrils were represented by spring element SPRINGA (axial spring between two nodes). The spring stiffness $K(\varepsilon)$ had the linear dependence of strain $\varepsilon$, as written by
where the value $K_0$ is spring stiffness in $\varepsilon = 1$. The strain dependence of spring stiffness was expected to generate more sharp peak stress than constant value. The spring stiffness $K(\varepsilon)$ was simplified as the uniform in the tissue both in horizontal and vertical direction.

We summarize the material properties subjecting for the estimation. For Eq. 1, the depth-dependent strain $\varepsilon(x)$ and apparent strain $\varepsilon_0$ was experimentally obtained in the previous study. So, the apparent Young's modulus $E_0$ was the first material property for the estimating. For Eq. 3, initial void ratio $e_0$ was specified as a typical value of 4.0. Therefore, $K_{\text{min}}$, $k_0$, and $M$ were subjected for the estimation. However, $k_{\text{min}}$ was chosen as a minimum value. The other estimating material property was $K_0$ in Eq. 4. In rewording the estimation of material properties, we tried to reduce the representation of the time-dependent behaviors of the 4 different experimental tests into 5 variables. The curve fitting was executed by visual confirmation with superposing calculation result into the experimental data. The NLGEOM in the software package was turned on for geometrical nonlinearity. The UTOL threshold for automatic time length control was specified by 10 kPa.

3. Results

Because of the cylindrical indentation, the time-dependent reaction was shown by the force per width (N/mm). Figure 4 shows the result of the experimental compression test. In Fig. 4 (a), there was large difference of the peak stress between 10% and 15% strain. This might be caused by the difference of increase rate of penetration volume and contact area in comparing with flat face indentation. The same situation was applicable in the result of (b) creep test. Namely, the experimental results of the cylindrical indentation configuration would include a complex and a lot of information in terms of the contact condition and related bulk behavior.

Figure 5 shows variance of experimental data and the results of curve fitting by the FE analysis. The resultant time-dependent curves of the FE analysis were calculated by a single parameter set. The curve fitting procedure involved some insights on how each parameter functionally represented the time-dependent behaviors of 4 different compressive conditions. In equilibrium condition, reactional force was assumed to occur by elasticity of the solid phase. The apparent Young's modulus $E_0$ was roughly estimated by the reactional force at the end of the definite compression. Then, the depth-dependent Young's modulus $E(x)$ was distributed by Eq. 1. Because the peak reaction force was apparently sensitive to the spring stiffness component $K_0$, the value $K_0$ was approximately estimated here. The components of the strain dependent permeability $k_0$ and $M$ were examined by the rapid force reduction just after the peak reactional force. The permeability also well affected the creep test. In this study, the lower limit of permeability $k_{\text{min}}$ was selected as a minimum value to fit the creep behavior. In a feasibility trial within the definite

![Graph](image)
compression tests, we easily got an acceptable curve fitting without the value $k_{\text{min}}$. Once the creep test was incorporated, we obviously could not fit both the rapid force reduction of the definite compression and the creep behavior simultaneously. As shown in Fig. 5 (d), the tissue was compressed almost 20 %. Without $k_{\text{min}}$ in Eq. 3, the strain dependent permeability $k$ was considerably reduced with the exponential function. In the definite compression, the strain dependence of the spring stiffness, shown in Eq. 4, was also effective. However, the line of the creep test without $k_{\text{min}}$ stayed under the experimental line, because of excessive small value of the permeability $k$. So, we decided to introduce the lower limit of the permeability $k_{\text{min}}$. Another component $M$ for the strain dependent permeability regulated stress relaxation of latter time period mainly in the definite compression, in which the value $M$ decreases the relative large value of $k_0$ after the introduction of $k_{\text{min}}$.

On the base of these insights, repetition of the procedure resulted in an acceptable curve fitting between the experimental results and the FE analysis, as shown in Fig. 5. The resultant parameters are listed in Table 1. From our experimental data, a relative large permeability as the initial value of $k_0 + k_{\text{min}}$ was estimated in comparing with the other past reports. This large initial permeability is shortly reduced with a compressive strain by the exponential function with the compaction effect $M$, as shown in Eq. 3. Since we used the cylindrical indenter, the contact area was small in early compression period. The large permeability in small compression test, which means quasi-initial state, was past reported with the similar definition of the strain dependent permeability (Jurvelin et al., 2003).

4. Discussion

The purpose of this study was to estimate the material properties of articular cartilage by the experimental compression data obtained in more physiological condition. One of the physiological conditions is 3-dimensional model by medical scanning images. The findings of the 3-dimensional regional severity of the tissue would provide the

![Graph](#)

Fig. 5 Variance of experimental data in each compressive conditions and the results of curve fitting by the finite element analysis. The data from finite element analysis were reproduced by a single parameter set for representing material properties. All of the FE analysis data were within the variance of the experimental data with an acceptable fitness for both each time-dependency of the reaction force and the different compressive conditions, which included the creep test.
Table 1  Material properties of the articular cartilage. The values with asterisk mark (*) were estimated by the curve fitting of the finite element analysis.

| Parameter                                      | Value                      |
|------------------------------------------------|----------------------------|
| Young's modulus ($E_0$)                        | 0.83 MPa                   |
| Poisson's ratio                                | 0.125                      |
| Lower limit of permeability ($k_{min}$)        | $5.0 \times 10^{-15}$ m$^4$/Ns |
| Base of strain dependent permeability ($k_0$)  | $58.86 \times 10^{-15}$ m$^4$/Ns |
| Compaction effect on permeability ($M$)        | 22                         |
| Initial void ratio ($e_0$)                     | 4.0 (80 % interstitial fluid content) |
| Spring stiffness component ($K_0$)             | 17.5 MPa                   |

pathological insight (Li et al., 2014). On the other hand, we thought that the study in material testing level would also provide the better understandings for regenerative or artificial cartilage. The knowledge on what and/or how the constitutive components of the tissue realize its synergistic functionalities should bring important provisions from the engineering viewpoint. While various studies on modeling of articular cartilage have been conducted with their specific concern, several studies emphasized the importance of the physiological dynamic condition. These studies selected special experimental conditions including high compression rate, high compressive amount and sliding with migrating contact area for investigating additional tissue behavior.

The geometrical complex of the indenter and the compressive rate influences the estimation of material properties (Warner et al., 2001; Julkunen et al., 2008). The plate-to-plate compression does not include the effect of the surface seepage. The flat face indenter involves an edge loading in indenter circumference. In this study, the cylindrical indentation was selected for further 2-dimensional sliding test with migrating contact area. In this configuration, the penetration volume was not proportional to the penetration distance. In the compression test of 10% strain at 2 %/s strain rate and the creep test, the fluid near the indenter seeps from surface, because the compressive time period is enough long to flow out. Since the cylindrical indentation involved several complex phenomena, the resultant time-dependent behavior also includes plenty information, which would reduce the conflict among the material properties in the curve fitting. The curve fitting in 10% strain at 2 %/s and the creep test, shown in Fig. 5 (c) and (d), should capture these complex condition. Additional force feedback control was implemented in the self-made compressive tester with the high-speed precise positional control. The first result in this study was the experimental data set with 4 different conditions by the single testing machine, in which a single specimen sequentially exposed to different conditions in considering physiological condition. This sort of experimental data set is rarely seen in the literature.

The estimation of material properties by the curve fitting method was achieved with visually good conformation. In this study, 5 constitutive material properties were identified by the curve fitting onto the 4 different experiments. Though the material parameters in this study might not represent all behaviors, the 4 different behaviors were reduced into 5 constitutive material parameters. The roles of the constitutive material properties were ascertained in curve fitting technique, as described in the result section. The lower limit of permeability $k_{min}$ was incorporated for the strain dependent permeability $k$. The time constant of osmotic behavior is very longer than a physiological dynamic load, in which the recovery of the tissue shape commonly take more than 30 min. With the exception of 2-dimensional limitation, the necessity of the lower limit of permeability might be derived from the quasi-static osmotic effect, which caused the tissue to prevent severely compressed condition and maintained the permeability within a certain level. In this situation, the static osmotic pressure was represented by the apparent Young's modulus $E_0$ in the equilibrium condition. Though the large permeability in the initial swollen condition was past reported in a literature (Jurvelin et al., 2003), this was partly caused by the source of the tissue. The specimen used in this study was harvested from porcine knee joint of 6-8 month old from regional meet processor without any freezing process. The specimen might not be a mature state enough, because Salter's incremental line was sometimes not fully closed.

We should note the limitation of this study. In the recursive curve fitting procedure, the resultant conformity was validated by a visual agreement. Thought the more distinct methodology should be incorporated, e.g. a least-square
method, the other concern was to understand how the material properties realized its functionalities as a load bearing material. As noted in the result section, the curve fitting procedure showed interesting insights of the role of each material property. For this study, we did not experienced inevitable interference among the material parameters except for the lower limit of permeability $k_{\text{min}}$, which was selected as a minimum value to reproduce acceptable fitting for mainly the creep behavior. This meant that the swelling effect on the permeability in quasi-static condition could not be negligible factor. The effect of the swelling factor would be indirectly managed by the resultant effect of the value $k_{\text{min}}$. Today, more detailed FE models were proposed with their interest. While we believe that additional component for the modeling shows more exact resultant behavior, the study on the realization mechanism of the functionalities of each material property also brings insightful information for the development of artificial or regenerative cartilage. One of our next interest on the discrition of the cartilaginous model would be anisotropy and inhomogeneity of the collagen network structure. As the anisotropy and the inhomogeneity of the mechanical properties of collagen network were not fully revealed by experimental studies, this structure should bring an impact on the prediction of the functionality of articular cartilage. From an engineering viewpoint, the incorporation of the anisotropic fabric structure of the collagen network would be significant even if the mechanical properties still involves an assumption. In this study, 2-dimensional model rather than 3-dimensional model was accepted for the further sliding material test to reduce computational cost. Though the reduction to the 2-dimensional model might involve not a little difference from the real phenomena, we convinced that it did not completely change the resultant insight. The resultant absolute values are often influenced by the source of the specimen, etc. In this study, the experimental data set was obtained by the compressive tests of high compressive amount at high-speed compression and the creep test, which were expected to range from quasi-static to dynamic situation in considering physiological condition. We tried to reduce the representation of the time-dependent behaviors of the 4 different experiments into these 5 variables. The more distinct method for curve fitting and the examination with the 3-dimensional model will be the future work at this stage. The experimental data sets for curve fitting procedure were firstly averaged with some variance, as shown in Fig. 5. This methodology concealed the potentialities for investigating the specimen-specific differences and following variance of each parameter. The findings of the specimen-specific material properties might bring a sensitivity study of each material property in the reactional behaviors.

5. Conclusion

In the compressive experiment, the cylindrical indenter was employed to capture complex contact condition. The experimental data set was obtained by the compression tests of different compressive amount at different compressive rates and the creep test. The material properties of articular cartilage were identified by the curve fitting procedure. The identified material properties well reproduced the experimental data set, which ranged from quasi-static to dynamic condition.

Acknowledgement

This study was financially supported by the Grant-in Aid for Specially Promoted Research of Japanese Society of the Promotion of Science (KAKENHI: 23000011).

References

Ateshian, G. A., A theoretical formulation for boundary friction in articular cartilage, Journal of Biomechanical Engineering: Transactions of ASME, Vol.119 (1997), pp.81-86.
Ateshian, G. A., The role of interstitial fluid pressurization in articular cartilage lubrication, Journal of Biomechanics, Vol.42 (2009), pp.1163–1176.
Caligaris, M. and Ateshian, G. A., Effect of sustained internal fluid pressurization under migrating contact area, and boundary lubrication by synovial fluid, on cartilage friction, Osteoarthritits and Cartilage, Vol.16 (2008), pp.1220–1227.
Chahine N. O., Wang C. C. B., Hung C. T. and Ateshian G. A., Anisotropic strain-dependent material properties of bovine articular cartilage in the transitional range from tension to compression, Journal of Biomechanics, Vol.37 (2004), pp.1251-1261.
Deneweth J. M., McLean, S. G. and Arruda, E. M., Evaluation of hyperelastic models of the non-linear and non-uniform high strain-rate mechanics of tibial cartilage, Journal of Biomechanics, Vol.43 (2013), pp.1604-1610.
Espino, D. M., Shepherd D. E. T, Hukins, D. W. L., Viscoelastic properties of bovine knee joint articular cartilage: dependency on thickness and loading frequency, BMC Musculoskeletal Disorders, Vol.15 (2014), pp.205-213.
Erne O. K., Reid J. B., Ehmke L. W., Sommers M. B., Madey S. M. and Bottlang, M., Depth-dependent strain of patello femoral articular cartilage in unconfined compression. Journal of Biomechanics, Vol.38 (2005), pp.667-672.

Gao, L., Zhang, C., Gao, H., Liu, Z. and Xiao, P., Depth and rate dependent mechanical behavior of articular cartilage: Experiments and theoretical prediction, Materials Science and Engineering C, Vol.38 (2014), pp.244-251.

Goreham-Voss, C. M., McKinley, T. O. and Brown, T. D., A finite element exploration of cartilage stress near an articular incongruity during unstable motion, Journal of Biomechanics, Vol.40 (2007), pp.3438-3447.

Graindorge, S., Fernandez, W., Ingham, E., Jin, Z., Twigg, P. and Fisher, J., The role of the surface amorphous layer of articular cartilage in joint lubrication, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.220 (2006), pp.597-607.

Gu, K. B. and Li, L. P., A human knee joint model considering fluid pressure and fiber orientation in cartilages and menisci, Medical Engineering & Physics, Vol.33 (2011), pp.497-503.

Henak, C. R., Kapron, A. L., Anderson, A. E., Ellis, B. J., Maas, S. A. and Weiss, J. A., Specimen-specific predictions of contact stress under physiological loading in the human hip: validation and sensitivity studies, Biomechanics and Modeling in Mechanobiology, Vol.13 (2014), pp.387-400.

Hosoda N., Sakai N., Sawae Y. and Murakami T., Depth-dependence and time-dependence in mechanical behaviors of articular cartilage in unconfined compression test under constant total deformation, Journal of Biomechanical Science and Engineering, Vol.3, No.2 (2008), pp.209-220.

Hosoda N., Sakai N., Sawae Y. and Murakami T., Finite element analyses of articular cartilage models considering depth-dependent elastic modulus and collagen fiber network, Vol.5, No.4 (2010), pp.437-448.

Huang C. Y., Soltz M. A., Kopacz M., Mow V. C. and Ateshian G. A., Experimental verification of the roles of intrinsic matrix viscoelasticity and tension-compression nonlinearity in the biphasic response of cartilage, Journal of Biomechanical Engineering: Transactions of ASME, Vol.125 (2003), pp.84-93.

Julkunen, P., Korhonen, R. K., Herzog, W. and Jurvelin, J. S., Uncertainties in indentation testing of articular cartilage: A fibril-reinforced poroviscoelastic study, Medical Engineering and Physics, Vol.30 (2008), pp.506-515.

June, R. K. and Fyhrie, D. P., Temperature effects in articular cartilage biomechanics, Journal of Experimental Biology, Vol.213 (2010), pp.3934-3940.

Jurvelin, J. S., Bushmann M. D. and Hunziker E. B., Mechanical anisotropy of the human knee articular cartilage in compression, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.217 (2003), pp.215-219.

Katta, J., Jin, Z., Ingham, E. and Fisher, J., Biotribology of articular cartilage - a review of the recent advances, Medical Engineering and Physics, Vol.30 (2008), pp.1349–1363.

Klein, J., Molecular mechanisms of synovial joint lubrication, Proceedings of the Institution of Mechanical Engineers - Part J: Journal of Engineering Tribology, Vol.220 (2006), pp.691-710.

Krishnan, R., Park, S., Eckstein, F. and Ateshian, G. A., Inhomogeneous cartilage properties enhance superficial interstitial fluid support and frictional properties, but do not provide a homogeneous state of stress, Journal of Biomechanical Engineering: Transactions of ASME, Vol.125 (2003), pp.569-577.

Krishnan, R., Kopacz, M. and Ateshian, G. A., Experimental verification of the role of interstitial fluid pressurization in cartilage lubrication, Journal of Orthopaedic Research, Vol.22 (2004), pp.565-570.

Krishnan, R., Marinier, E. N. and Ateshian, G.A., Effect of dynamic loading on the frictional response of bovine articular cartilage, Journal of Biomechanics, Vol.38 (2005), pp.1665-1673.

Li, J., Hua, X., Jin, Z., Fisher, J. and Wilcox, R. K., Biphaptic investigation of contact mechanics in natural human hip during activities, Procedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.228 (2014), pp.556-563.

Li, L. P., Soulhat, J., Buschmann, M. D. and Shirazi-Adl, A., Nonlinear analysis of cartilage in unconfined ramp compression using a fibril reinforced poroelastic model, Clinical Biomechanics, Vol.14 (1999), pp.673-682.

Li, L. P., Buschmann, M. D., Shirazi-Adl A., Strain-rate dependent stiffness of articular cartilage in unconfined compression, Journal of Biomechanical Engineering: Transactions of ASME, Vol.125 (2003), pp.161-168.

Liu, F., Kozanek, M., Hosseini, A., In vivo tibiofemoral cartilage defromation during the stance phase of gait, Journal of Biomechanics, Vol.43 (2010), pp.658-665.

Lucchinetti, E., Adams, C. S., Horton Jr, W. E. and Torzilli, P. A., Cartilage viability after repetitive loading: a preliminary report, Osteoarthritis and Cartilage, Vol.10 (2002), pp.71-81.

Mäkelä, J. T. A., Huttu, M. R. J. and Korhonen, R. K., Structure-function relationship in osteoarthritic human hip joint articular cartilage, Osteoarthritis and Cartilage, Vol.20 (2012), pp.1268-1277.

Mow, V. C., Kuei, S. C., Lai, W. M., Armstrong, C. G., Biphaptic creep and stress relaxation of articular cartilage in compression: theory and experiments, Journal of Biomechanical Engineering: Transactions of ASME, Vol.102 (1980), pp.73-84.

Murakami T., Higaki H., Sawae Y., Ohtsuki N., Moriyama S. and Nakanishi Y., Adaptive multimode lubrication in natural synovial joints and artificial joints, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of
Murakami, T., Nakashima, K., Sawae, Y., Sakai, N., Yarimitsu, S. and Hosoda, N., Roles of adsorbed film and gel layer in hydration lubrication for articular cartilage, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.223 (2009), pp.287-295.

Murakami, T., Nakashima, K., Yarimitsu, S., Sawae, Y. and Sakai, N., Effectiveness of adsorbed film and gel layer in hydration lubrication as adaptive multimode lubrication mechanism for articular cartilage, Proceedings of the Institution of Mechanical Engineers - Part J: Journal of Engineering Tribology, Vol.225 (2011), pp.1174-1185.

Nansai, R., Suzuki, T., Shimomura, K., Ando, W., Nakamura, N. and Fujie, H., Surface morphology and stiffness of cartilage-like tissue repaired with scaffold-free tissue engineered construct, Journal of Biomechanical Science and Engineering, Vol.6 (2011), pp.40-48.

Osawa, T., Moriyama, S. and Tanaka, M., Finite element analysis of hip joint cartilage reproduced from real bone surface geometry based on 3D-CT image, Journal of Biomechanical Science and Engineering, Vol.9 (2014), 8 pages.

Park, S., Krishnan, R., Nicoll, S. B. and Ateshian, G. A., Cartilage interstitial fluid load support in unconfined compression, Journal of Biomechanics, Vol.36 (2003), pp.1785-1796.

Pawaskar, S. S., Jin, Z. M., Fisher, J., Modelling of fluid support inside articular cartilage during sliding, Proceedings of the Institution of Mechanical Engineers - Part J: Journal of Engineering Tribology, Vol.221 (2007), pp.165-174.

Pawaskar, S. S., Grosland N. M., Ingham E., Fisher, J. and Jin, Z., Hemiarthroplasty of hip joint: An experimental validation using porcine acetabulum, Journal of Biomechanics, Vol.44 (2011), 1536-1542.

Sakai, N., Hagihara, Y., Furusawa, T., Hosoda, N., Sawae, Y. and Murakami, T., Analysis of biphasic lubrication of articular cartilage loaded by cylindrical indenter, Tribology International, Vol.46 (2012a), pp.225-236.

Sakai, N., Hosoda, N., Hagihara, Y., Sawae, Y. and Murakami, T., A study on the realization mechanism of the functionality of articular cartilage, Biomechanism, Vol. 21 (2012b), pp.251-263, in Japanese.

Schinagl R. M., Gurskis D., Chen A. D. and Sah, R. L., Depth-dependent confined compression modulus of full-thickness bovine articular cartilage, Journal of Orthopaedic Research, Vol.15 (1997), pp.499-506.

Shirazi, R. and Shirazi-Adl, A., Deep vertical collagen fibrils play a significant role in mechanics of articular cartilage, Journal of Orthopaedic Research, Vol.26 (2008), pp.608-615.

Soltz M. A., Ateshian, G. A., Experimental verification and theoretical prediction of cartilage interstitial fluid pressurization at an impermeable contact surface in confined compression, Journal of Biomechanics, Vol.31 (1998), pp.927-934.

Wang, C. C. B., Chahine N. O., Hung C. T. and Ateshian, G.A., Optical determination of anisotropic material properties of bovine articular cartilage in compression, Journal of Biomechanics, Vol.36 (2003), pp.339-353.

Warner, M. D., Taylor, W. R., Clift, S. E., Finite element biphasic indentation of cartilage: a comparison of experimental indenter and physiological contact geometry, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.215 (2001), pp.487-496.

Willet, T. L., Whiteside, R., Wild, P. M., Wyss, U. P., Anastassiades, T., Artefacts in the mechanical characterization of porcine articular cartilage due to freezing, Proceedings of the Institution of Mechanical Engineers - Part H: Journal of Engineering in Medicine, Vol.219 (2005), pp.23-29.

Yarimitsu, S., Nakashima, K., Sawae, Y. and Murakami, T., Influences of lubricant composition on forming boundary film composed of synovia constituents, Tribology International, Vol.42 (2009), pp.1615-1623.