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Pertinence of Sheep Knee Joint for Calibration of Ligaments' Constitutive Equations; Experimental and Theoretical Study

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ABSTRACT: The knee joint is one of the most complex joints in human body because of its complex geometry and articulations. On the other hand, due to many practical constraints for studying the anatomy and biomechanics of the human knee, in vivo and in vitro animal models have been widely used. Based on this fact, an objective comparison of the sheep samples especially from mechanical behavior point of view is needed. Therefore, a purpose of the present study is to evaluate priority of usage of sheep specimens via comparing the biomechanical differences of normal ligaments between sheep and human. To this end, some experimental tensile tests have been done on the different knee ligaments of sheep including hyperelastic behavior of the anterior cruciate ligament, medial collateral ligament, posterior cruciate ligament, and lateral collateral ligament. So, an objective comparison of the sheep and human samples has been done. Furthermore, the magnitude of material constants of different hyperelastic constitutive equations including 3rd order Ogden, Yeoh and Fung–Demiray models, as well as the maximum experienced stress by the knee ligaments have been considered.

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1- Introduction

One injury that imposes a serious hurdle to one's daily activities is knee injuries [1]. Knee joint injuries accounted for approximately 40% of sports-related injuries, and the knee injury threat in recreational and competitive sports was 10 times higher than in commuting and lifestyle activities [2]. Now it is obvious that ligaments play main roles in maintaining the stability and restrict degree of freedom of human joints. During the recent decades, many studies have focused on the function, injury, and healing of ligaments in order to make more accurate models. Historically, due to many practical constraints for studying the anatomy and biomechanics of the human knee, in vivo and in vitro animal models have been used. The reports have been based on studies performed with many species such as mice [3], rats [4], rabbits [5, 6], goats [7], pig [8], monkey [9, 10] and dogs [11, 12]. These studies would help to find methods of how to clinically manage ligament healing, and it could be achieved by characterizing and simulating these processes with employing animal models. Among many animal models sheep stifle joint ligaments was selected for conduct this research because of its similarity in anatomy and normalized value [13]. The properties and function of ligaments are almost nonlinear and cannot easily be simulated. On the other hand accurate modeling of the realistic mechanical behavior of biological tissues is of great importance for surgery simulation, finite element modeling of soft tissues and clinical applications [14]. While the theory of nonlinear elasticity using hyperelastic models, could describe accurately the nonlinear mechanical behavior of tissues, finding the most appropriate material model and its constants are almost problematic. This process would lead to further animal models advances and better understanding of mechanical behavior. From a variety of different methods, numerical simulation, especially the Finite Element Method (FEM), has been employed more widely [15]. This method gives users the capability of obtaining the stress-strain curve and simulating many and various situations with low cost. To get accurate simulated results by using this method, the constitutive equations of ligaments have been considered as one of the principal factors [16]. In addition, the constitutive model parameters could be fitted by different methods. Morrow et al. [17] have found that it would be more accurate to average data from multiple trials than averaging the material coefficients fitted by the experiment data of each specimen. Wan et al. [18] provided an overview of the current research on ligament constitutive relations on the macro, meso and micro levels as well as the anatomy and histological structure of ligament. They presented a discussion based on the research on ligament constitutive relation in the past three decades and proposed a new constitutive relation. Tanaka et al. [19] investigated the human knee joint sound during the Lachman test based on comparison between healthy and Anterior Cruciate Ligament (ACL)-deficient knees. Also, Wan et al. [20] modified a constitutive relation for ligament tissues based on the previous constitutive relation by considering the effects of collagen types. In another study, Oskui et al. [21] formulated a viscohyperelastic constitutive model with the use of the internal variables approach to evaluate the nonlinear elastic and time dependent anisotropic mechanical behavior of the periodontal ligament. Marchi et al. [22] presented a constitutive theory to address these issues-predominately assuming transverse isotropy with the preferred material direction aligned with structural collagen of Medial Collateral Ligament (MCL). They showed how recent data of the MCL

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fails to be accurately represented using previously validated and generally accepted constitutive theories.
Pierrat et al. [23] developed and validate an experimental testing machine to standardized human limb by using a finite element approach, and then to use this machine to characterize the efficiency of different categories of orthoses. Moreover, Coles et al. [24] evaluated the suitability of a novel knee simulator for investigating patella femoral joint biomechanics facilitating the extended assessment of joint biomechanics under physiological levels of loading. Furthermore, Lowry et al [25] developed a crouching simulator, based on the Oxford-type machine, with novel features including a synthetic knee including ligaments. Lots of researchers have used animal models including sheep knee joint to investigate the behavior of the human knee ligaments. Few studies related to the mechanical properties of the sheep joint ligaments are present in literatures. Just some studies were performed on ACL of sheep whereas different studies were performed on dog, monkey, pig and rabbit [26-28].
Although significant studies have been done on the identification and modeling of soft tissues from mechanical behavior point of view, there is still considerable challenges in this regard. Without enough related information, using animal ligament testing instead of human ones cannot be much confident.
Based on these facts, assessing the applicability of the animal test results is challenging because of the lack of suitable animal models. Therefore, a purpose of the present study is to evaluate the priority of usage of sheep specimens via comparing the biomechanical differences of normal ligaments between sheep and human. To this end, an objective comparison of the sheep samples especially from mechanical behavior point of view is needed. The purpose of this paper is to determine the parameters of three mainly used constitutive models by fitting experiments data of sheep joint ligaments and a comparison with coefficients of constitutive models of human knee ligaments. In order to investigate it, some experimental mechanical tests have been done on the different knee ligaments of sheep, including the ACL, MCL, Posterior Cruciate Ligament (PCL), and Lateral Collateral Ligament (LCL). On the other hand, the magnitude of material constants of different hyperelastic constitutive equations including 3rd order Ogden [29], Yeoh [30] and Fung–Demiray [31-33] Models, as well as the maximum experienced stress by knee ligaments have been considered.

2- Materials and Methods

Over a limited range of joint motion, ligaments resist increasingly to tensile loading, this behavior permits joints to move easily within normal motion limits, so the mechanical behavior in tensile loading is of great importance. The ligaments constitute of collagen, elastin, glycoproteins, protein polysaccharides, water, glycolipids, and cells (mostly fibrocytes). Collagen and ground substance form the greatest amount of ligaments, Water in ligament is associated with the ground substance and constitute about 60 to 80 percent of the wet weight of ligaments.70 to 80 percent of the remaining weight is made up by the fibrillar protein collagen, type I collagen formed the majority of collagen in ligament which is also found in tendon, skin and bone. The ground substance matrix surrounds the collagen; this connective tissue is to some degree responsible for grasping the collagen together. The chief constituent of the ground substance matrix is proteoglycan. These molecules have a very important role in ligament function; however, constitute just less than 1% of the ligament's total dry weight [34]. Based in such complex structural constituent finding a constitutive equation for modeling realistically the mechanical behavior could be challenging [35-37].

2- 1- Specimen preparation and test method

Harvested three rear legs from adolescent sheep (6-8 month, 45-50 kg), and fifteen specimens remained intact totally. The stifle joints selected were not necessarily harvested from the same sheep. Ligaments of the sheep stifle joint were selected because of their similarity in morphology, size and structure to the human knee joint [13]. The fat, muscle and joint capsule were removed by sharp cutter, and the ligaments dissected carefully to avoiding tear as it shown in Fig. 1.

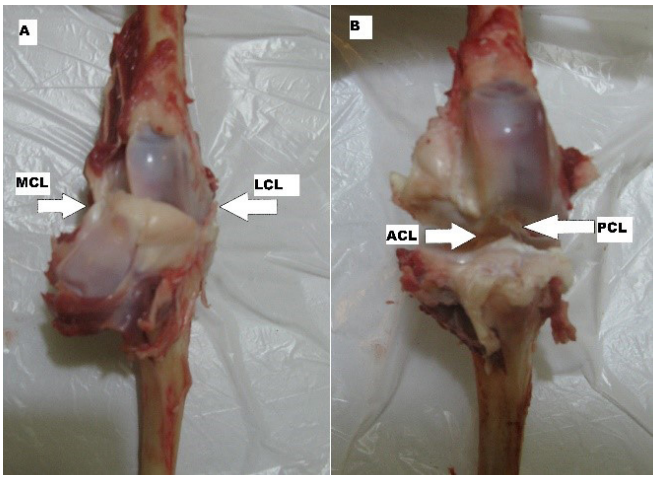

Fig. 1. (A) Sheep knee joint including MCL, LCL ligaments, (B) ACL, PCL ligaments of sheep knee joint

Physiological 0.9% saline used to keep all ligament specimens moist during the extraction time to prevent dehydration. The isolated ligaments were fixed with cyanoacrylate and sandpaper as in Fig. 2, the specimens then were wrapped in a sterile saline-moisten gauze, which were remoistened frequently, sealed in a polyethylene bag, labeled and stored at -20 °C. It was demonstrated by Woo et al. [38] that the biomechanical properties of ligaments are not affected by prolonged freezing.
Twelve to twenty-four hours before each tensile test, the specimens were removed from the freezer and allowed to thaw gradually at 4°C in a refrigerator. The specimen was then submerged in physiological 0.9% saline (at room temperature) for a minimum of 15 min. The ligaments' length and circumference were measured multiple times using a caliper (0.1 mm division) and average the results for improving the reliability before each test, ligaments are relatively flat and, therefore, a rectangular cross section was assumed. Then the specimens attached to a 5 kN load cell of Hounsfield H10KS universal testing machine. Initial gage length of all specimens adjust to a definitive amount equals to 13 mm, after mounting the specimen between the top and bottom grips securely and completely vertical, force and extension of tensile test balanced to zero to be ready for starting the test. The load was applied in the direction parallel

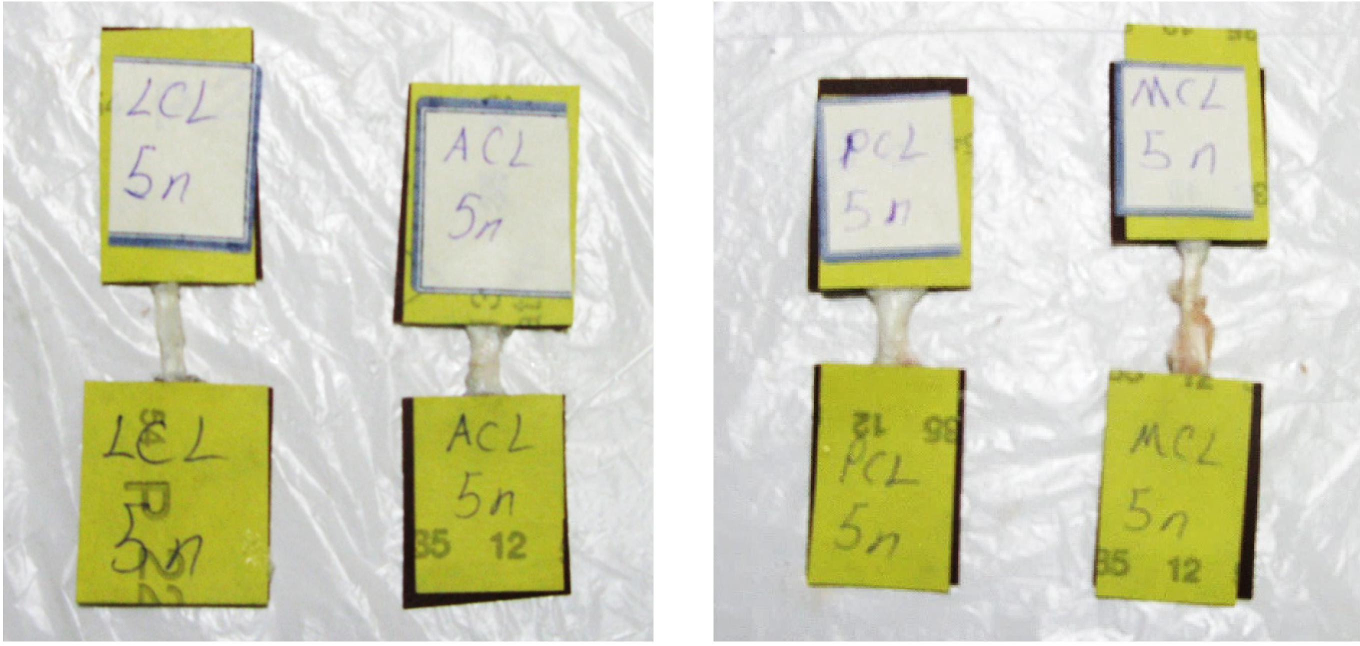

Fig. 2. Prepared ligaments specimens

to the long axis of the ligaments, and all tests were conducted with a crosshead speed of 20 mm/min (Fig. 3). The slow loading rate was selected to eschew the possible errors as inaccurate strain measurements related to fast strain rate and to simulate quasi-static behavior. Testing was performed at room temperature, and during the test, samples were wetted using a saline spray in order to preclude dehydration and the resulting dimensional changes. The force-extension data was automatically collected by the testing software, exported to Microsoft Excel, engineering stress-strain curve was calculated and its diagram was constructed.

On the other hand in order to compare the sheep result to human ones two different experimental test results of the human specimens reported in related published reference are selected. Preparations and test method of the specimens of human knee ligaments recall as Human-1 and Human-2 are as follow.

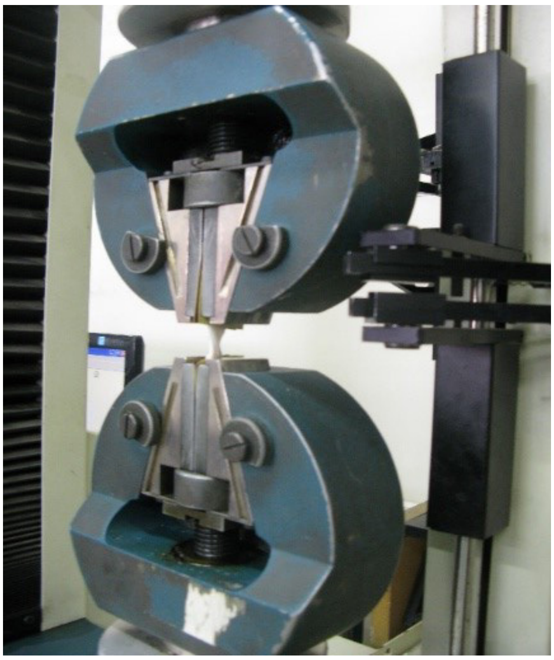

Fig. 3. LCL ligament on tensile test machine

Human-1 test data selected from reference [20] in which Winkler's conservation solution used to moist them and then refrigerated at +3 °C for preparing ligaments. Ligaments were dissected and then were embedded in resin from the bone insertions. Tests were held along fiber orientations with velocities of 1.98 m/s, and at room temperature with a 5-25 N preload [20].

Human-2 test data selected from reference [39] in which human knee joints were obtained from patient suffered from pelvic tumor. The joints were double-wrapped using gauze soaked with phosphate buffered solution and stored at -20 °C. A day before uniaxial tensile tests knee joints were thawed at room temperature. The tests were carried out along fiber directions with velocities of 10 mm/min, and inside a bath chamber with a constant temperature of 37 °C and subjected to a 2 N preload [39].

2- 2- Constitutive relations for ligaments and parameters estimation

For modeling the rubber-like material behavior such as biological soft tissues the hyperelasticity theory is employed. These types of materials portray completely elastic mechanical behavior. The stress-strain relation relies only on the current level of strain and insensitive to its history [40]. Strain energy function W (per unit reference volume) is used to define mechanical behavior of hyperelastic materials. When a system deforms, the energy stored by that system represented by the strain energy function and when the load is eliminated, strain energy is gradually gets free and the system returns to its initial state in such a hyperelastic material. The strain energy function, W, is a function of the deformation gradient tensor, F, because the model material presumed homogeneous. Components of the deformation gradient, F, based on both the initial and current position are represented as:

$$F_{ij} = \frac{\partial x_i}{\partial X_j} \tag{1}$$

where x and X as a point in the reference position and in the current position, respectively and i and j are coordinates directions. The deformation gradient tensor allows us to take into account, large deformations and rotations which are not considered in linear elasticity theories. The Jacobean of motion is defined as $J=det(F)$ and is a measure for the volume dilatation near a point x between instants t_0 and t. Most of the soft tissues are nearly incompressible (i.e. volumetric dilatation is not considerable in the continuum and can be neglected), hence $J=det\ F=1$ is defined as the incompressibility constraint [41].
The right and left Cauchy-Green deformation tensors are measures of the strain the body experiences and are defined as:

$$C = F^T F \tag{2}$$

$$B = FF^T \tag{3}$$

where C and B are the right Cauchy-Green and the left Cauchy-Green deformation tensors respectively and the superscript T represent the transpose. Also the modeled material presumed to be isotropic. This means that the material acts uniform in all directions. For isotropic materials, the strain energy function, $W= W(F)$, is a function of invariants I_1, I_2, and I_3, which are defined as [14]:

$$I_1 = tr\left(C\right) = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \tag{4}$$

$$I_2 = 0.5(I_1^2 - tr\left(C^2\right)) = \lambda_1^2\lambda_2^2 + \lambda_1^2\lambda_3^2 + \lambda_2^2\lambda_3^2 \tag{5}$$

$$I_3 = det\left(C\right) = \lambda_1^2\lambda_2^2\lambda_3^2 \tag{6}$$

For incompressible materials, $det(C) = \lambda_1^2 \lambda_2^2 \lambda_3^2 =1$, therefore the strain energy function is a function of only two invariants $W=W(I_1, I_2)$.
There is a variety of strain energy functions expressed in different forms [14]. Here, three of the most commonly used forms for modeling biological materials including Ogden, Yeoh and Fung-Demiraywill be introduced and then fit experimental data. There were two important assumptions for these constitutive relations of ligaments. First, some material characteristics of ligaments depending on time such as viscoelasticity, relaxation and creep were neglected because of the insensitivity of the constitutive behavior to strain rate [42, 43]; secondly, ligaments under high strain rates were not observed failure.

2- 2- 1- Ogden model

The Ogden strain energy function [29] is based on principle stretches and represents acceptable correlation with the experimental data [44]. The strain energy function in the Ogden model is described as:

$$w = \sum_{i=1}^{N} \frac{\mu_i}{\alpha_i}\left(\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3\right) \tag{7}$$

where μ_i and α_i are material constants and i defines the number of terms included in the summation and represents the order of the Ogden model. However, for practical purposes in many simulations, a third order equation of Ogden model is sufficient. The expression for W is derived using the stretch ratios instead of the invariants of the Cauchy deformation tensor. Therefore, the Ogden's model is often classified as a stretch-based model [14]. For consistency with the classical theory, the constants must satisfy the requirement [44]. Shear modulus of material (μ) in the Ogden model results from the expression:

$$\sum_{i=1}^{N} \mu_i \alpha_i = 2\mu \tag{8}$$

For an incompressible material the incompressibility constraint could be written as $J = \lambda_1 \lambda_2 \lambda_3 = 1$. Rearranging the Ogden strain energy function using this constraint we can obtain:

$$w = \sum_{i=1}^{N} \frac{\mu_i}{\alpha_i}\left(\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \frac{1}{\left(\lambda_1\lambda_2\right)^{\alpha_i}} - 3\right) \tag{9}$$

2- 2- 2- Yeoh model

A polynomial form of the strain energy function is the Yeoh [30] model, also called the reduced polynomial model. Yeoh developed a hyperelastic material model that only depends on the first invariant of the Cauchy-Green deformation tensor, I_1. It has the following form [30]:

$$W = \sum_{i=1}^{N} C_i \left(I_1 - 3\right)^i + W_{vol} \tag{10}$$

where W_{vol} is volumetric part of the strain energy density. For the incompressible material it has the following form:

$$W = C_{10}(I_1 - 3) + C_{20}\left(I_1 - 3\right)^2 + C_{30}\left(I_1 - 3\right)^3 \tag{11}$$

Shear modulus of material (μ) in the Yeoh model results from the expression:

$$\mu = 2C_{10} \tag{12}$$

Yeoh hyperelastic model includes the second and the third orders of I_1, which can be a more accurate representation of tissue properties [14].

2- 2- 3- Fung-Demiray model

An exponential form of the strain energy function was introduced first in the 1960s by Fung [31] and then by Demiray [32]. The Fung-Demiray model is only based on the first strain invariant. It has the following form:

$$W = \frac{a}{b}\left(e^{\frac{b}{2}(I_1 - 3)} - 1\right) \tag{13}$$

where a (stress-like material parameter) and b (dimensionless material parameter) are material constants, and $a, b > 0$.

3- Results and Discussion

The most common and widely accepted forms of the hyperelastic strain energy functions for biological tissue modeling as Ogden, Yeoh and Fung-Demiray [14] are described and implemented to compare their ability to characterize the nonlinear rate-independent mechanical behavior of knee ligaments as well as investigating the priority of using sheep samples instead of human ones. Mechanical tests were performed to determine the material coefficients of the constitutive relation for ligaments of sheep knee joint in order to validate the feasibility and accordance of results in comparison to the human ones.

The data from 16 biomechanical tensile tests of ligaments were available for analysis while the remaining data sets were not used due to gripper failure during testing, and early sample failure around the sand paper or tearing during dissection process. The time to failure of the ligaments ranged from 6 seconds to 12 seconds. The longest ligament was found to be medial with 43±7 mm, and the lateral was 31±9 mm, while the anterior and posterior lengths were almost identical with 26±2 mm and 30±2 mm, respectively.

The structural properties of isolated ligaments are determined using tensile tests. In this test ligament is subjected to tensile force applied at constant rate. The stress-stress curve is initially upwardly concave, but the slope becomes nearly linear in the prefailure phase of tensile loading. The geometry of the specimen tested including length and cross-section can affect the curve configuration. The stress–strain curves obtained from test of different ligaments of sheep knee joint shown in Fig. 4.

Considering the stress-strain curves obtained from tensile test to failure of the sheep ligaments shown in Fig. 4, there are four kinds of behavior during the loading progress. Firstly, a non-linear increase in load called "toe region" could be seen in which the tissue elongates. In the second region, a linear behavior is seen during the loading. Isolated collagen fibers are disrupted and begin to fail in the third part of the curve and the ligament completely ruptures in the last region of the curve.

The comparison of the mechanical properties and size of sheep stifle joint ligaments and human knee ligaments [45] summarized in Table 1. In simple tension test, the behaviors of the ACL, PCL and MCL specimens are generally similar in median range of maximum strains as well as maximum forces. On the other hand, the order of values and behavior of LCL vary significantly for sheep and human specimens.

3- 1- Curve fitting of parameters in constitutive models

For identifying parameters of hyperelastic constitutive models (Ogden, Yeoh and Demiray), an optimization algorithm for curve fitting has been used. Experimental data of each ligament were averaged and then the differences with theoretical results were minimized. The optimization method is Nelder-Mead, also known as Downhill simplex [46].

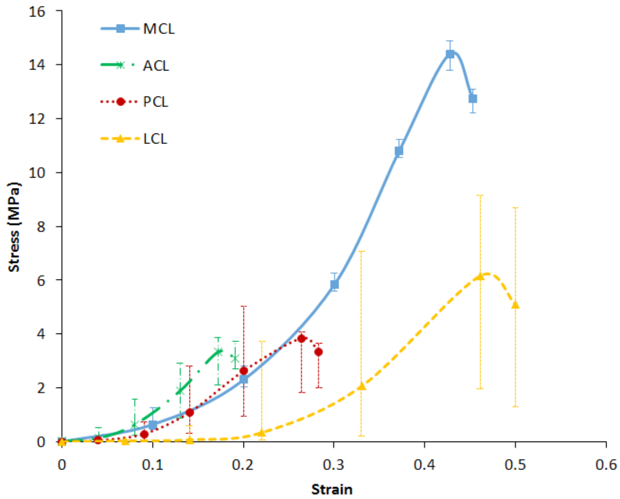

Fig. 4. The nominal stress–strain curves of sheep ligament specimens from the uniaxial tensile test

All the parameters of the three constitutive models for sheep specimens as well as two different human ligaments demonstrated as Human-1 and Human-2 reported by published references [20] and [39] respectively, shown in Tables 2 to 4. In order to obtain parameters of the hyperelastic material models, the experimental data up to the maximum stress level have been considered in curve fitting.

Considering the mechanical properties via material parameters of Ogden model, it is clearly seen in Table 2 that parameters remained similar in different ligaments of sheep and human specimens, except in PCL of human results of Human-1 from reference [20]. A significant difference could be also seen in predicted parameters of MCL in human ligaments of Human-2 from reference [39].

The material properties of the ligament are expressed in terms of a stress-strain relationship. A tissue's material properties may be obtained from force-elongation data by dividing the recorded force by the original cross-sectional area to give stress, and by dividing the difference between the specimen length and its original length by its original length to give strain.

On the other hand, comparison of the results for material constant in Yeoh and Demiray model in Tables 3 and 4 reveal

Table 1. Comparison between human and sheep knee ligaments

		Max Strain (%)	Max stress (MPa)	Max Force (N)	Lengths (mm)
ACL	Human [45]	18-24	-	75.5-605	-
	Sheep	14-35	3-7.2	55.2-121.2	23.5-28
PCL	Human [45]	18-24	-	158-505	-
	Sheep	21-31	2.22-5.22	63.6-125.1	25-31.5
MCL	Human [45]	21-38	-	160-350	-
	Sheep	8-37	13.7-14.9	113.2-169.2	36.7-50.6
LCL	Human [45]	21-38	-	155-400	-
	Sheep	30-60	1.9-10.2	49.8-61.6	35.7-39.6

Table 2. Material constants based on Ogden model for human and sheep ligaments

		μ_1	α_1	μ_2	α_2	μ_3	α_3
ACL	Sheep	4.89	7.036	0.581	-1.423	23.597	-1.474
	Human-1	5.88	20.91	10.18	-9.46	36.79	-0.085
	Human-2	7.32	6.36	1.47	-1.65	13.46	-3.144
PCL	Sheep	9.21	0.636	178.32	-1.12	307.403	0.617
	Human-1	0.3	52.1	4.8	-2.8	-7.2	0.07
	Human-2	9.18	0.50	153.75	-0.97	306.15	0.511
MCL	Sheep	0.579	10.16	203	-0.618	709	0.167
	Human-1	13.6	14.5	217.03	-1.7	715.3	0.3
	Human-2	1.33	-9.41	12.40	-0.51	729.45	0.043
LCL	Sheep	22.92	5.46	220.25	4.18	-240.24	4.3
	Human-1	31.78	7.51	128.99	7.27	-466.45	2.549
	Human-2	26.05	4.93	215.81	4.88	-242.36	4.86

Table 3. Material constants based on Yeoh model for human and sheep ligaments

		C_{10}	C_{20}	C_{30}
ACL	Sheep	0.325	30.52	-68.21
	Human-1	9.67	738.45	-20.41
	Human-2	3.52	14.0	63.07
PCL	Sheep	1.16	2.65	14.28
	Human-1	0.16	606.7	-588
	Human-2	3.64	2.65	32.19
MCL	Sheep	0.538	7.464	1.881
	Human-1	14	771	-66.9
	Human-2	3.28	1.32	-5.96
LCL	Sheep	-0.44	2.85	1.34
	Human-1	17.5	42	35.5
	Human-2	0.67	11.89	-23.92

Table 4. Material constants based on Fung-Demiray model for human and sheep ligaments

		a	b
ACL	Sheep	2.3	32.8
	Human-1	22.6	158.06
	Human-2	7.06	16.99
PCL	Sheep	2.24	12.72
	Human-1	4.7	359.56
	Human-2	6.83	8.21
MCL	Sheep	3.44	7.4
	Human-1	26.6	152.39
	Human-2	6.52	0.1
LCL	Sheep	0.73	9.28
	Human-1	37.9	193.8
	Human-2	3.09	6.37

that the values are greatly different in human and sheep specimens in magnitude and signs.

In order to validate the applicability of calibrated coefficient of constitutive equations based on in vitro sheep knee joint tests for predicting the same mechanical behavior of human knee ligaments, related stress-strain curves drawn in Figs. 5 to 7 could be give a clear demonstration. Also, both averaged measuring results of the longitudinal tensile specimens of human and simulated results based on the constitutive models are shown in the same figures. For comparing sheep and human ligaments, the results of the longitudinal tensile test of human ligaments were extracted from the study of the references [20] and [39]. As it is clear the initial toe portion of the curve characterized by lower stiffness in sheep results and it followed by a stiffer region in that by increasing strain, stiffness began to increase rapidly, presumably due to the straightening of the collagen fibers. Fitting of these constitutive models on experimental results of sheep and human knee ligaments could be clearly seen in these figures.

Significant differences between the magnitude of maximum strain experienced by the sheep ACL, MCL, LCL and PCL when compared with the Human-1specimens are obviously seen.

During routine daily activities such as walking and standing, ligaments are loaded to less than one fourth their ultimate tensile load. During strenuous activities such as fast cutting during intense running, loading levels may enter into region 3 where isolated fiber damage takes place. Based on this facts there is no need to consider region 4 of the curve in fitting hyperelastic models.

As it can be seen in these figures, in the toe region of the curve increasing the strain results in uncurling of the crimp pattern of collagens. After that, the collagen fibers are stretched in linear part of the curve. While the ligament is more strained ligament fibers begin to rupture and then complete ligament fail occurs.

It is believed that ligaments are not generally loaded above one-fourth of their ultimate tensile load during these daily activities. In the upper operating range of the ligaments especially ACL during strenuous activities as might be experienced during fast cutting or pivoting while running.

The strains in the ACL ligaments are approximately the same in the Human-2 knees and the sheep stifle, while the maximum stress in is ligaments are roughly different. The same

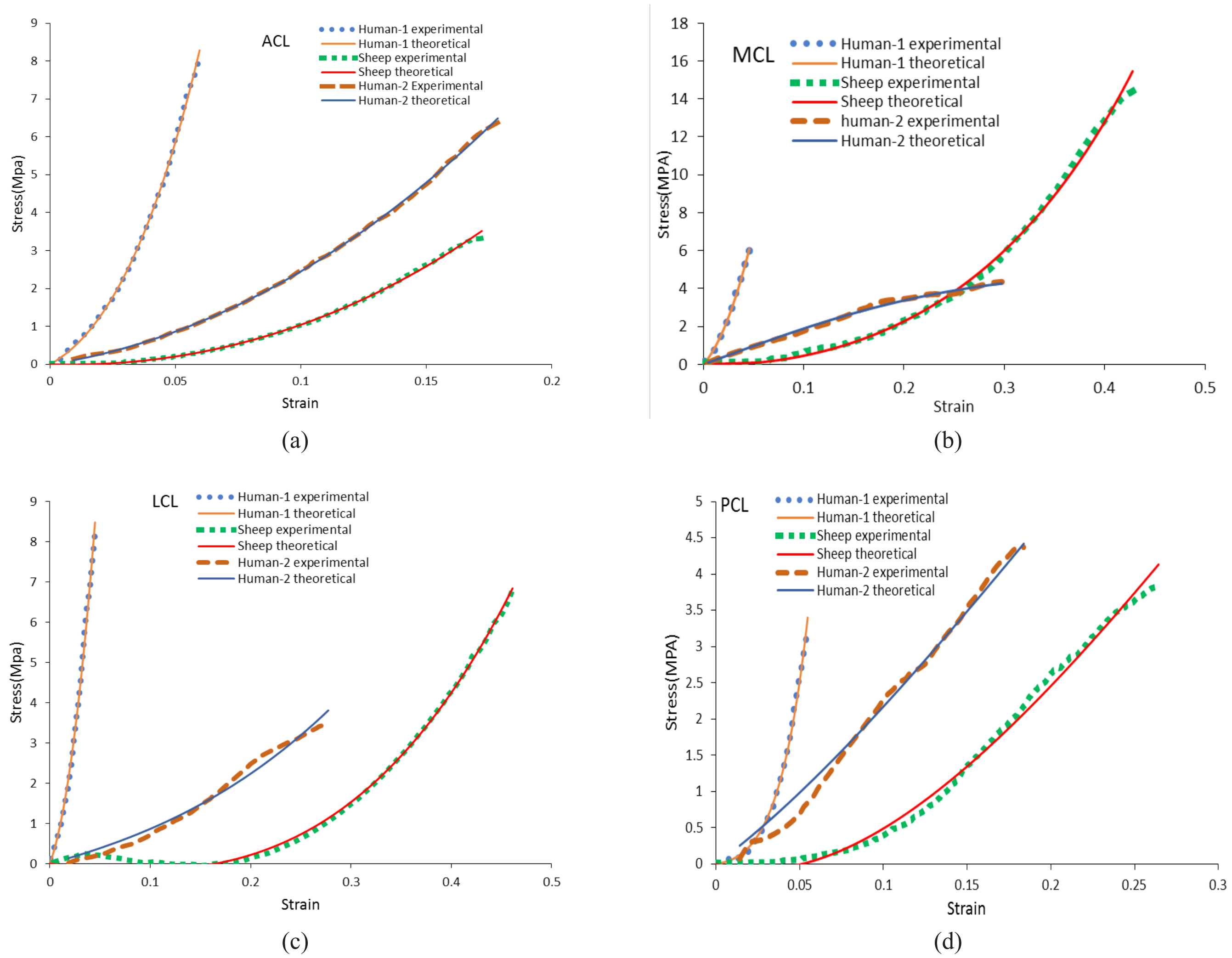

Fig. 5. Sheep and human ligaments mechanical behavior obtained by Ogden model and experimental test results, a)ACL, b)MCL, c) LCL, d)PCL

comparison for MCL and PCL reveals both stress and strain maximum values are different while the hyperelastic models could roughly predict the true behavior. In PCL specimens almost the same maximum stress could be seen beside the different values of maximum strain at the same stress level. It should be noted that by comparing the accordance of different theoretical mechanical behavior predicted 3rd order Ogden model is considerably more appropriate in both cases of sheep and human specimens.

It should be noted that based on anatomic configuration of LCL, the in situ forces in the whole LCL were approximately the same in the human knees and the animal knees. While the human species tended to carry a larger portion of this force in its bundle than the animal models. The direction of the in situ force in the whole human LCL bundle also differed significantly from that of the sheep knees. The differences between the LCLs of various species may be due to variations in the LCL bundle as well as whole LCLs, as significant differences found the whole LCLs. We postulate that this finding may be secondary to the anatomy of the knees or to differences in the anatomical division between the LCL bundles of the different species. Also, variations in shape and size of the tibial plateau affect the relative positions of the LCL insertions, in turn affecting the force of the LCL and its bundles obtained in tests. Another explanation for the association of significant differences found between the results in the LCL may be that measurements were taken at only parallel position in current tests.

There are limitations to this study which should be taken into consideration when using these data as criteria for selecting sheep model. Firstly, the measurements taken under only 1 degree of freedom (DOF), which are parallel to the ligaments collagen fibers and shear tests, were not held. Secondly, the range of stifle flexion is different in the sheep stifle and human knee, because of sheep stifle joint inability to reach full extension. Also, it should be considered that sheep spend much more time of their life in deeper flexion than their human counterparts.

4- Conclusion

In this study, uniaxial tensile behavior of knee joint ligaments of sheep was determined based on experimental test. At the same time investigation of common constitutive equations based on the continuum mechanics approach for modeling the rate-independent mechanical behavior of sheep and human knee ligaments have been considered. Three frequently used isotropic hyperelastic constitutive relations of ligaments including Ogden, Yeoh and Fung-Demiray were fitted by

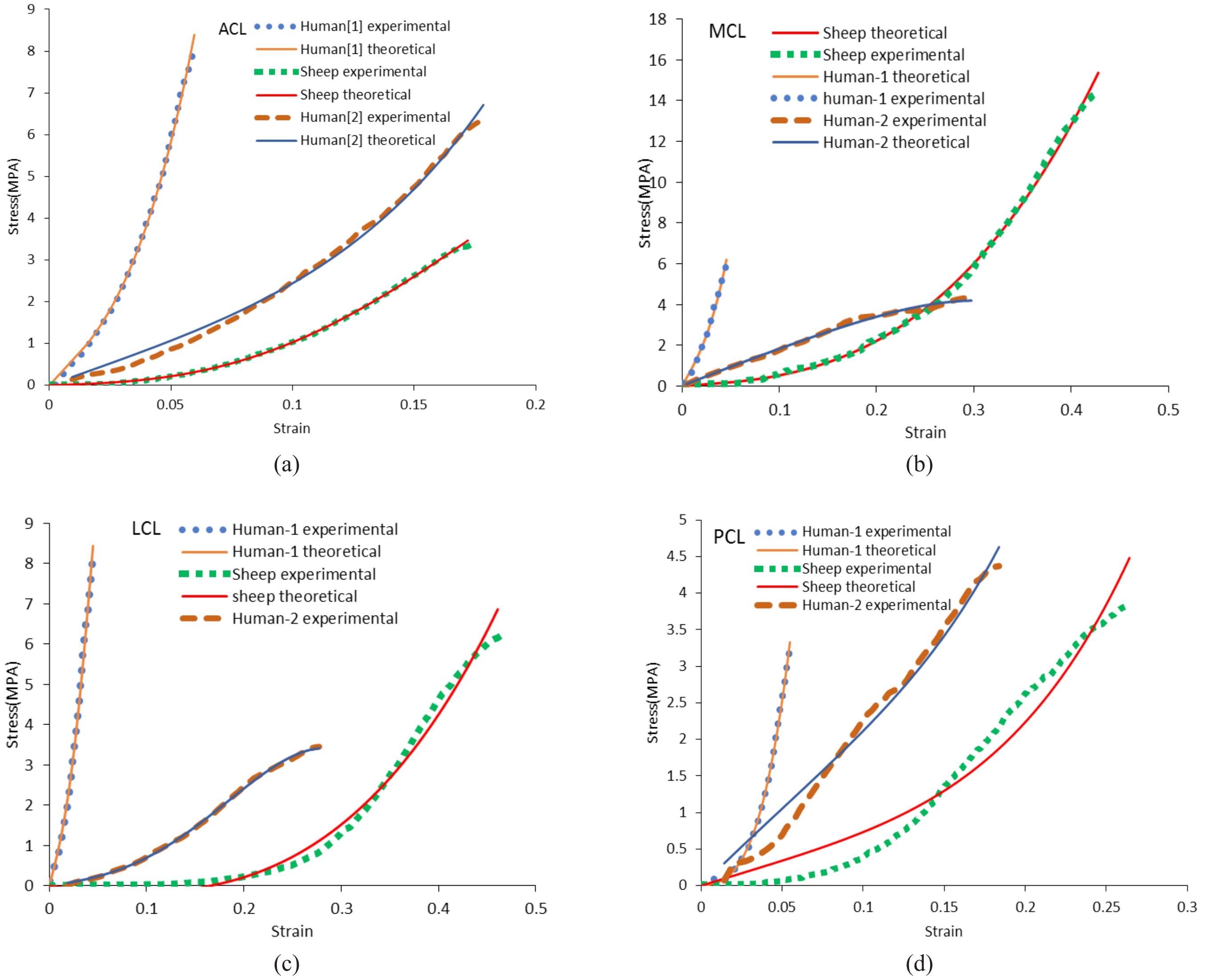

Fig. 6. Sheep and human ligaments mechanical behavior obtained by Yeoh model and experimental test results, (a)ACL, (b)MCL, (c) LCL, (d)PCL

the longitudinal stress–strain data of human and sheep knee ligaments. It was found that the parameters of ACL, MCL and LCL of sheep in the Ogden model are analogous with their human counterparts, while the ligaments of sheep and human response no similarity in Yeoh and Demiray models. The different behavior of the sheep and human ligaments was also compared. The data obtained in this study suggest that it would be better to use just ACL, MCL and LCL Ogden constitutive equation instead of human ACL, MCL and LCL, while it is not recommended in other situations.

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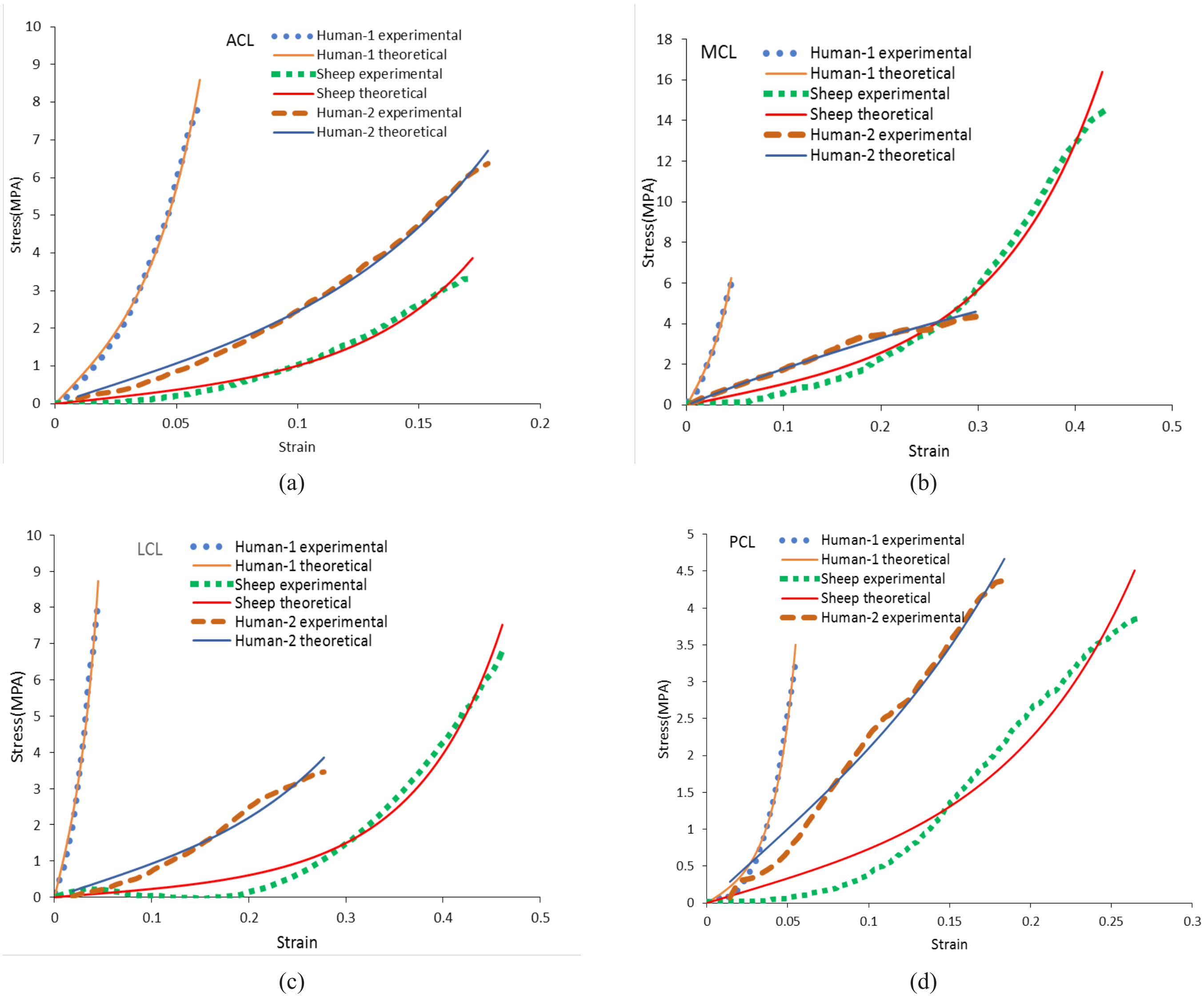

Fig. 7. Sheep and human ligaments mechanical behavior obtained by Fung-Demiray model and experimental test results, (a)ACL, (b) MCL, (c)LCL, (d)PCL

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