Anterior and Posterior Bundles of Superficial Medial Collateral Ligament of Knee Joint: Achieving Balanced Structural Behaviors with Different Material Properties

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Research Article

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Abstract

Background

In the knee joint simulations under accidents or physiological loadings, the mechanical properties of the superficial medial collateral ligament (sMCL) are essential. The common practice for sMCL material parameters in simulations is to either assign different structural properties to the fiber bundles of the sMCL in 1D models or give homogeneous material properties in 2D or 3D models. The sMCL can be functionally divided into anterior bundle (AB) and posterior bundle (PB), which are alternatively loaded during walking. However, little data are available on the mechanical properties of the two bundles; consequently, the rationality of the material assignment method used in current computational studies is questionable.

Methods

Using porcine stifle joints as test subjects, we obtained the mechanical properties (i.e., threshold strain, failure strain, failure stress, and modulus) of the AB, PB, and whole ligament of the sMCL.

Results

The three subjects had significant differences in the modulus ($p < 0.05$), and the AB had higher failure strain ($p < 0.01$) and threshold strain ($p < 0.001$) than the PB. The ligament length may be the main reason for the modulus and failure strain differences. Meanwhile, the AB and PB had no significant differences in the structural properties (i.e., stiffness per unit area and failure displacement) as well as the failure stress. Based on the understanding of the balanced structural properties, we proposed a method to estimate the material properties of the AB and PB from those of the whole ligament to reduce the experimental cost for simulations.

Conclusions

This study suggests that, contrary to the current material assignment method, the AB and PB of the sMCL have different material properties and similar structural properties. The AB and PB should be treated separately and assigned different properties according to their geometry to realize the biomechanics of native knee joints in simulations.

Background

The function and stability of the knee joints are dictated by the restraint of the ligaments and tendons. An important restraint to anteromedial instability is the superficial medial collateral ligament (sMCL), the largest structure of the MCL [1, 2]. The MCL is one of the most frequently injured knee ligaments in sports and vehicle accidents [3, 4]. Thereby, understanding the biomechanics of the sMCL is vital for injury prevention, medical treatment, and rehabilitation.

The computational study plays an irreplaceable role in deepening the understanding of knee ligament behaviors. It can integrate the information obtained by other methods (e.g., animal and volunteer tests) and predict variables that are difficult to measure directly (e.g., stress distribution and loading of ligament and bone interaction) [5]. Moreover, it can also provide considerable insights into ligament changes under complex loading conditions that are difficult to realize experimentally (e.g., the injury evolution process in a pedestrian accident) [6]. It is essential to present material properties properly in computational studies [7]. The most widely used material assignment method for sMCL in simulations either assumes that the sMCL has homogeneous material in 2D or 3D models or gives different structural properties to the fiber

bundles of the sMCL in 1D models [8]. However, the rationality of this material assignment method is questionable due to the lack of experimental data [5, 8–12].

The mechanical properties of the sMCL are often measured using the whole ligament (WL) along its working load direction [13–16]. Most experiments tested the WL at a uniform load angle, where the fibers of the sMCL are said to be evenly taut (Fig. 1b) [14, 16, 17]. Others tested the WL under anatomical position [18, 19], which corresponds to the normal standing state of humans (i.e., the knee flexion angle is 0°) (Fig. 1c). Although it is difficult to distinguish from the exterior clearly, volunteer and cadaver tests have shown that the sMCL has functionally separated anterior bundle (AB) and posterior bundle (PB). The AB and PB are alternately loaded during walking (Fig. 1) [20, 21]. The similarly alternately loaded AB and PB of the anterior and posterior cruciate ligaments have been proved to have different material properties [22, 23]. Comparatively, there is little information regarding the mechanical properties of the two bundles of sMCL.

The study aimed to identify whether the material properties of the AB and PB of the sMCL are different from those measured from the WL, and whether the material assignment method used in current computational studies is rational. We measured and compared the material properties (i.e., threshold strain, Young's modulus, failure stress, and failure strain) of the AB, PB, and WL using porcine sMCLs. And specific for the porcine sMCL tests, an efficient, custom-made fixture was proposed without bone cement. The main reason for the material property differences was further founded by comparing the relevant structural properties and ligament geometry. Finally, the failure modes and morphology differences were examined.

Methods

Specimen preparation

We purchased the fresh stifle joints of the hind legs of mature Yorkshire pigs from a local market. The pigs were slaughtered according to the food production process in slaughterhouses; therefore, a review by the research ethics committee was not required. Only the stifle joints that could be tested on the day were purchased. The purchased stifle joints were dissected individually as soon as they were delivered to the laboratory. The stifle joint treated in this way did not go through any frozen or unfrozen process, which can ensure the freshness of the joints. Redundant soft tissue was removed to obtain the femur-sMCL-tibia complex. The specimens were randomly divided into three groups for testing. Ligaments of the “WL group”, like most mechanical tests of entire sMCLs, were tested under a uniform load angle. Specimens of the “AB group” and “PB group” were tested at 150° and 60° knee flexion angle, respectively. The knee flexion angles ensured the uniform load of the fibers of the AB or PB.

The femur and tibia were positioned at the tested flexion angle to carve marker lines (Fig. 2). Then along the marker lines, condyles were osteotomized using a table saw. Finally, the osteotomized flat surface on the condyle was attached to a custom-made fixture for loading (Section 2.3). To maintain the correct flexion angle during the loading, the red lines were carved as parallel as possible. Moreover, the blue lines indicated the maximum allowable diaphysis length that did not affect the connection between the fixture and test machine.

The ligament was hydrated by frequently spraying 0.9% saline during preparation and testing. The preparation and testing were carried out at room temperature (21–23°C).

Measurement of sample geometry

The attachments of the sMCL have complex shapes and are often described as ellipses (Fig. 3a) [24, 25]. The distance between the centroids of the tibial and femoral attachments is the ligament's initial length (i.e., effective length) [25]. The effective length is difficult to measure directly since the centroids are challenging to locate. Therefore, we simplified the shape of the attachments. And three directly measurable lengths of the simplified attachments were recorded to

characterize the sMCL geometry. Because tibial attachment behaves complicated in experiments, a triangular area ($\triangle BDC$) was retained as its main feature (Fig. 3a). The side BD was the separation line between the tibia and ligament, from which the ligament can be pulled outward. And point C was the farthest point at which the ligament extended on the tibia. The tibial attachment is much larger than the femoral attachment [24–26]. And the fibers of the femoral attachment blended well with the bone. So, the femoral attachment shape is difficult to distinguish. Thus, we ignored the femoral attachment size and simplified the separation between the stretchable ligament substance and femur as a straight line (side AE). Consequently, the sMCL surface was characterized as a trapezoid (AEDC) (Fig. 3b).

Based on the above characterization, the following three length values were measured: anterior length (side ED), full posterior length (side AC), and net posterior length (side AB). The effective length of AB, PB, and WL was defined as the anterior length, the average of the full posterior and net posterior lengths, and the average of the anterior and full posterior lengths, respectively. These effective length definitions approximated the distance between the centroids of the corresponding quasi-real attachments for each subject. The cross-section was assumed to be rectangular [27], and the midsubstance width and thickness were gauged. All the geometric dimensions were averaged using three independent measurements.

Test apparatus

Static tensile tests were conducted using a Zwick Z020TE test machine. The test machine clamped the femur-sMCL-tibia complex using a custom-made fixture (Fig. 4). The fixture has two metal boxes attached to the upper and lower clamps of the test machine. The box housed the femoral or tibial condyles. The condylar plane was in plane-to-plane contact with the bottom plate of the box. And the ligament substance passed through the hole on the bottom of the box. We carefully adjusted the position and orientation of the condyles to form a satisfactory lineup (i.e., the ligament substance was vertical in both the front and side views). Then, the condyles were fixed by the pointed screws, which had a 60° cone at the end. The screws were screwed into the condyle through the holes on the sidewalls of the box. Such mounted and fixed femur-sMCL-tibia complex was loaded when the upper box moved upward on the test machine. Consequently, the ligament was stretched under uniaxial tension.

The tensile force was measured using a commercial load cell (range, 20 kN) at the lower clamp. The displacement was recorded by the motion of the upper cross beam of the test machine and cross-checked using a camera that captured the movement of the fixture.

Test protocol

The test protocol is illustrated in Fig. 5. After mounting on the test machine, the AB and PB groups' sample was firstly given a 2 N preload force for cutting. The PB of the sMCL was cut off to test the mechanical properties of the AB. In turn, the AB of the PB group samples was also cut off. The cut portion was greater than half of the entire ligament width, which conservatively avoided the possible fiber overlap between the AB and PB. The remaining width of the uncut portion was measured as the effective width for the cross-sectional area (CSA) calculations. Moreover, the WL group tested the un-cut entire ligament. Next, the specimen was subjected to the preconditioning, 10 cycles of cyclic loading with a strain varying between 0% and 3% at 1.9 Hz, to reduce experimental variability [28, 29]. Finally, from a 2 N preload, the specimen was loaded to failure at 10 mm/min.

Data processing and analysis

Because the three subjects had different geometric dimensions, we compared their material properties (i.e., the stress and strain relationship). The engineering stress is the force divided by the initial CSA (i.e., the effective width multiplied by the thickness), and the engineering strain is the deformation (extension) divided by the effective length. Figure 6 shows a representative stress-strain curve of the ligament determined from the tests. The curve included a toe region at the beginning and a yield region near the peak (failure). Most of the middle portion of the curve was nearly linear, and its slope

was defined as modulus. We carefully described how to determine the modulus such that the determination method of the material properties was consistent between the tests (Fig. 6).

We used the data points from a starting point to an ending point along the curve for the initial linear fitting. The initial starting point may be a data point slightly beyond the toe region to eliminate the toe region's influence on the regression (e.g., 4% strain point). The initial ending point was the failure point (i.e., the maximum stress point). After that, new linear fittings were performed by moving the starting and ending points toward the middle of the curve at a certain increment. The number of data points per linear fitting was gradually decreased. The linear fitting was repeated until the coefficient of determination of the fitting reached 0.99. The data points used for the final linear fitting were defined as the linear region. The slope of the linear region was the modulus. Illustrated by the enlarged circle in Fig. 6, the threshold strain indicating the end of the toe region was defined as the strain point at which the difference between the linearly regressed stress and measured stress reaches 1% of the failure stress.

Statistical analyses were performed using SPSS software. The test matrix included the AB, PB, and WL groups. The experimental data are presented as mean $\pm$ standard deviation because all data were normally distributed (Shapiro-Wilk tests). The threshold strain, Young's modulus, failure strain, and failure stress were compared using the one-way ANOVA. LSD or Games-Howell multiple comparison methods were used based on Levene's tests. The failure mode was treated as a binary variable and compared using the chi-square test. The results were considered statistically significant when the significance level (i.e., p-value) was less than 0.05.

Results

Eighteen tests were conducted with six valid data in each group. The geometric dimensions and material properties of each specimen are listed in Table 1. As the stress-strain curves shown in Fig. 7, the trends among the three groups were prominent. In the order of PB, WL, and AB, the failure strain and failure stress became smaller, while the modulus became greater.

There were significant differences in the modulus values of the three groups (AB 319.18 ± 24.94 MPa, PB 481.38 ± 64.81 MPa, WL 377.72 ± 27.65 MPa; $p < 0.05$) and the failure strains of the AB and PB groups (AB $22.30 \pm 3.41\%$, PB $13.42 \pm 1.18\%$; $p = 0.002$). While the difference in the failure stress was small and insignificant in the three groups (AB 61.95 ± 9.24 MPa, PB 55.09 ± 2.36 MPa, WL 60.31 ± 10.97 MPa). As to the threshold strain, there were significant differences (AB $3.83 \pm 1.49\%$, PB $1.09 \pm 0.7\%$, WL $3.39 \pm 0.56\%$) between AB and PB groups ($p < 0.001$) and PB and WL groups ($p = 0.001$) but not AB and WL groups. The AB group had one modulus outlier and one threshold strain outlier; hence, we performed ANOVA again on the data without the outliers. The results did not change except for the specific p-value.

We observed two failure modes: midsubstance rupture and tibial avulsion. It is worth noting that all specimens in the AB group failed at the midsubstance, whereas the failure sites of samples of the PB and WL groups were relatively random. Thus, we analyzed the failure mode difference in the three groups using the chi-square test. The results showed no significant difference.

Discussions

The mechanical properties of the entire sMCL obtained in this study are comparable to those obtained by others (Table 2), demonstrating the effectiveness of the custom-made fixture. And the measured material properties of the AB and PB may be used as references for computational models of human knee joints. The simulation of normal physiological activities requires the threshold strain and modulus, and the injury simulation needs the failure stress or failure strain. The significant differences in the threshold strain, modulus, and failure strain between the AB and PB indicated that they should be treated as separate components and assigned different material parameter values in computational models.

Reasons for the differences in the material properties

The modulus values of the AB and PB were significantly different from that of the WL. Therefore, it is inappropriate to directly assign the modulus measured from WLs for ABs and PBs. However, it is often not convenient to perform tensile tests on AB and PB. To address this problem, we analyzed the main factors leading to the differences in the material properties of individual bundles, and proposed a method to estimate the material properties of the AB and PB from those of the WL.

Modulus and stiffness

The modulus can be written as:

$$E = l_e \frac{K}{CSA} \quad (1)$$

where E is the modulus (MPa), l_e is the effective length (mm), CSA is the cross-sectional area (mm²), and K is the stiffness (N/mm), which is the linear slope of the force-displacement curves. Since the effective width of the AB and PB groups was approximately halved compared to that of the WL group, their CSAs were evidently different. A linear correlation between the CSA and stiffness were verified by Chandrashekar et al. [31] and Blackburn et al. [32] using anterior cruciate ligament and hamstring tendon, respectively. Consequently, we regarded K/CSA as a single variable, called stiffness per unit area (SPA). Namely, the modulus can be decomposed into effective lengths and SPA. We then compared the effective lengths and SPA values of the three groups. The effective length (AB 49.92 ± 7.62 mm, PB 68.17 ± 4.02 mm, WL 67.13 ± 6.81 mm) differences between the AB and PB groups ($p < 0.001$) and the AB and WL groups ($p < 0.001$) were significant. And the SPA (AB 6.53 ± 1.15 N/mm³, PB 7.04 ± 0.62 N/mm³, WL 5.68 ± 0.74 N/mm³) between the PB and WL groups ($p = 0.016$) was significant as well. These led us to speculate that the effective length is the main reason for the modulus difference, and the SPA is the secondary reason. Pearson's correlation analysis further showed a high correlation between the effective length and modulus ($\rho = 0.63$; $p = 0.005$) (Fig. 8a).

We tested the three groups at their respective uniform load angles to load the individual bundles to their maximum capacities (Fig. 1). But the AB and PB are alternatively loaded in walking. It is not easy to find a knee flexion angle that all sMCL fibers are uniformly loaded simultaneously. Although the specimens of the WL group were tested at the uniform loading angle, some marginal fibers may not be fully engaged in the early stage of loading. Therefore, the proportion of the fully engaged fibers in the WL group may be less than those in the AB and PB groups. This may be why the SPA values of the AB and PB groups were greater than that of the WL group.

Failure strain and failure displacement

Similarly, the failure strain was analyzed. The failure strain is the failure displacement divided by the effective length. The reciprocal of the effective length was significantly different between the AB and PB groups ($p = 0.013$) and between the AB and WL groups ($p = 0.015$), while the failure displacement (AB 11.02 ± 1.71 mm, PB 9.12 ± 0.59 mm, WL 12.37 ± 3.73 mm) difference among the three groups was insignificant. And the correlation coefficient between the failure strain and effective length was -0.60 ($p = 0.008$) (Fig. 8b). Thus, the main reason for the difference in the failure strain should be the effective length.

Estimation of the material properties

Based on the above analysis, we hypothesized a balance mechanism in sMCL: by "adjusting" the material properties (i.e., modulus and failure strain), the structural properties (i.e., SPA and failure displacement) of individual bundles tend to be consistent, so that humans do not feel evident force changes when the AB and PB are alternatively loaded during walking. Based on the balance mechanism, we proposed a method for estimating the modulus and failure strain of the AB and PB

from the data of the WL to reduce experimental cost. According to the result of no significant difference between the three groups in failure displacement, we can assume the failure displacement of the three groups are equal. Thereafter, dividing the failure displacement of the WL by the effective lengths of the AB and PB can calculate the failure strains of the AB and PB, respectively. The effective lengths of the AB and PB were measured for the WL samples (Section 2.2). Similarly, considering the linear correlation between the stiffness and CSA, we further assumed the SPA values among the three groups were equal (discussed after the verification). Hence, according to Eq. (1), multiplying the SPA of the WL by the effective lengths of the AB and PB can speculate the modulus values of the AB and PB, respectively.

The estimation method was tested using the mean values of the measured data (Fig. 9). The estimation errors for the properties of the AB were approximately 11%, whereas the errors were about 20–35% for the PB. It is worth noting that using the estimated values for the AB and PB is better than directly using the material properties of the WL. The estimation may be regarded as a modification of the measured materials of the WL when using them for the AB and PB. In other words, between conducting tests of the AB and PB (the workload would double) and using property data of the WL directly, using the estimation method may be an alternative.

The relatively large error in estimating the failure strain of the PB (35%) may be due to the experimental measurement errors of ligament geometry. The difference in the effective lengths was the main reason for the differences in the material properties of the three groups. The anterior side of the tibial attachment is well-defined and can be simplified to a small connection point, whereas the posterior side of the tibial attachment occupies a large area, and its borderlines are not easy to identify (Section 4.2). Based on the definitions of the effective length described in Section 2.2 (Fig. 4), we can tell that the effective length of the AB is more reliable than those of the PB and WL. In fact, we can also use the estimation method in another way: estimating the modulus and failure strain of the PB and WL from the results of the AB. The resultant estimation errors were about 10–15% for the WL properties and about 7–20% for the PB properties.

Morphology and strength of attachments

The failure morphology showed that the tibial attachment roughly consisted of two areas: a relatively smooth area covered by the periosteum (i.e., periosteal area) and a rough area covered by the cartilage (i.e., cartilaginous area, close to the ligament midsubstance) (Fig. 10a). Consequently, based on whether the ligament was avulsed from the cartilaginous area, the tibial avulsion can be further divided into the cartilage avulsion (Fig. 10b) and the periosteum avulsion (Fig. 10c). This classification is consistent with Sekiguchi et al. [33].

All tibial avulsion samples in the WL group were avulsed from the posterior inferior side, so the periosteal area is supposed to be weaker than the cartilaginous area. The short AB mainly locates in the cartilaginous area, while the PB extends from the cartilaginous area to the periosteal area (Fig. 10a). Meanwhile, the failure stress difference between the AB and PB was small and insignificant. Therefore, the fact that all specimens in the AB group failed at the midsubstance indicated that the cartilaginous area is likely to be stronger than the ligament substance. Because the attachment strength was affected by the ratio of the two areas, the failure sites of the WL and PB group samples were randomly happened at the midsubstance or attachment.

Limitations

This study has several limitations. First, the porcine stifle joint has been a good substitute for the human knee joint, but they have differences. Human sMCLs have two tibial attachments, proximal and distal [34], while porcine sMCLs have only one tibial attachment. Although the strength of the proximal attachment is much smaller than that of the distal attachment [3], the extent to which the present results can be applied to human sMCLs remains to be further studied. Second, the estimation method based on the balanced structural properties was only applicable to the modulus and failure strain. The threshold strain of the AB and PB also had a significant difference. But because the end of the toe

region is related to the crimp of fibers, it is difficult to estimate the threshold strain according to the structural level response. The estimation for the threshold strain of different bundles may require further research. Third, this study divided the sMCL into AB and PB. However, the fiber bundles of the sMCL do not have a clear structural boundary. Others have suggested three bundle partitions for the sMCL [5, 10, 21, 35]. Further studies are needed to determine which segmentation is reasonable and meaningful for applications, and whether it still follows the hypothesis of the balance mechanism.

Conclusions

We found that the AB and PB of sMCL have different material properties, which is mainly due to the difference of ligament length. The ligament lengths and material properties jointly contribute to the balanced structural properties of the AB and PB. The different material properties and balanced structural properties are contrary to the material assignment method commonly used in simulating the sMCL, using homogeneous material properties or different structural properties. Moreover, the attachment strength of AB and PB to the condyles are also different. Therefore, we suggest that the AB and PB should be not only regarded as separate components but also given different properties according to their geometry to realize native knee joint behaviors in computational studies. And these computational studies may further help identify injury mechanisms under mechanical loading and improve the design of orthopedic devices.

Abbreviations

sMCL, superficial medial collateral ligament; WL, whole ligament; AB, anterior bundle; PB, posterior bundle; CSA, cross-sectional area; SPA, stiffness per unit area.

Declarations

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Availability of data and materials

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

Ethics approval and consent to participate

Not applicable

Consent for publication

All authors read, approved the final manuscript, and consented to publication.

Competing interests

None of the authors hold financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work.

Authors' contributions

WTC performed the experiments, analyzed the data, and drafted the manuscript. QZ checked the data, contributed to the interpretation of the data, and revised the manuscript. Both authors read and approved the final manuscript.

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Tables

Table 1 Mechanical and geometric properties of test samples ($\bar{x} \pm s$)

Mechanical properties						Geometry data				
		Modulus (MPa)	Threshold strain (%)	Failure strain (%)	Failure stress (MPa)	Failure site	I_a (mm)	I_{pf} (mm)	I_{pn} (mm)	CSA (mm ²)
AB group	1	365.83	3.68	27.03	74.16	Midsubstance	47	71	51	7.72
	2	296.07	4.57	20.36	55.67	Midsubstance	50	71	56	8.65
	3	304.18	5.52	24.13	67.97	Midsubstance	40	71	54	8.65
	4	326.37	4.04	21.40	59.79	Midsubstance	45	71	52	7.72
	5	311.34	1.09	17.22	48.42	Midsubstance	60	82	61	8.15
	6	311.30	4.08	23.69	65.69	Midsubstance	58	88	69	7.90
PB group	1	583.63	1.88	11.45	56.18	Midsubstance	45	78	57	9.78
	2	413.16	1.33	14.24	51.76	Midsubstance	48	67	57	8.15
	3	522.94	0.64	12.59	58.71	Tibial attachment	45	76	64	8.65
	4	431.08	0.01	14.38	53.63	Midsubstance	48	70	59	9.60
	5	443.10	0.74	14.26	54.65	Tibial attachment	45	79	60	8.42
	6	494.40	1.94	13.59	55.59	Midsubstance	52	80	61	10.60
WL group	1	384.35	2.96	13.53	48.51	Tibial attachment	46	85	73	19.02
	2	415.65	2.49	11.85	45.71	Tibial attachment	55	85	70	18.47
	3	396.01	3.55	21.96	66.06	Midsubstance	48	76	58	16.12
	4	364.80	3.91	17.97	60.27	Tibial attachment	40	75	55	16.71
	5	369.92	3.84	21.52	71.60	Midsubstance	59	94	65	21.60
	6	335.57	3.61	23.33	69.72	Midsubstance	52	91	64	18.13
I_a anterior length, I_{pf} full length of posterior side, I_{pn} net length of posterior side, CSA cross-sectional area.										

Table 2
Material properties of sMCL of Yorkshire pigs and humans ($\bar{x} \pm s$)

Mechanical properties	WL group	Yorkshire pig [14]	Human [30]
Modulus (MPa)	377.72 $\pm$ 27.65	400.0 $\pm$ 47.5 (R)/ 477.8 $\pm$ 110.7 (L)	332.15 $\pm$ 58.27
Failure stress (MPa)	60.31 $\pm$ 10.97	71.5 $\pm$ 15.6 (R)/ 88.5 $\pm$ 19.1 (L)	38.56 $\pm$ 4.76
Failure strain (%)	18.36 $\pm$ 4.76	21.0 $\pm$ 3.0 (R)/ 22.3 $\pm$ 2.8 (L)	17.11 $\pm$ 1.53
<i>R right, L left.</i>			

Figures

Figure 1

Human superficial medial collateral ligament (sMCL) states under different flexion angles. **a** Flexed knee: anterior bundle (AB) loaded. **b** Uniform load state. **c** Extended knee: posterior bundle (PB) loaded. The dashed red line indicates the approximate boundary between the AB and PB

Figure 2

Condyles before and after sawing

Figure 3

Simplified geometry of sMCL. **a** Photo. The blue area indicates quasi-real attachments. **b** Simplified schematic

Figure 4

Figure 5

Figure 6

Figure 7

Experimental stress-strain curves of the AB, PB and whole-ligament (WL) groups. The curves end at their respective failure points

Figure 8

Correlation between effective length and material properties

Figure 9

Verification of the estimation method

Figure 10

Cartilaginous area and periosteal area of tibial attachment. **a** Tibial attachment partitions. The dashed red line is an approximate boundary between the two areas. The green and purple areas represent the ligament substance and the simplified tibial attachment, respectively. **b** Cartilage avulsion. **c** Periosteum avulsion