# Material Characterization of Human Medial Collateral Ligament

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# Abstract

The objectives of this study were to determine the longitudinal and transverse material properties of the human medial collateral ligament (MCL) and to evaluate the ability of three existing constitutive models to describe the material behavior of MCL. Uniaxial test specimens were punched from ten human cadaveric MCLs and tensile tested along and transverse to the collagen fiber direction. Using load and optical strain analysis information the tangent modulus, tensile strength and ultimate strain were determined. The material coefficients for each constitutive model were determined using nonlinear regression. All specimens failed within the substance of the tissue. Specimens tested along the collagen fiber direction exhibited the typical nonlinear behavior reported for ligaments. This behavior was absent from the stress-strain curves of the transverse specimens. The average tensile strength, ultimate strain and tangent modulus for the longitudinal specimens was 38.6±4.8 MPa, 17.1±1.5% and 332.2±58.3 MPa, respectively. The average tensile strength, ultimate strain and tangent modulus for the transverse specimens was 1.7±0.5 MPa, 11.7±0.9% and 11.0±3.6 MPa, respectively. All three constitutive models described the longitudinal behavior of the ligament equally well. However, the ability of the models to describe the transverse behavior of the ligament varied.


# Introduction

Despite the large amount of research into the mechanical properties of knee ligaments, reports of the material properties of the human medial collateral ligament (MCL) were not found in the literature. These data are necessary to understand the normal function of the MCL in comparison to other knee ligaments, to quantify pathologies in the structure, and to develop constitutive models to represent its material behavior. For the development of computational models of ligaments or joints, often a simple hyperelastic representation is sufficient to describe the quasistatic material behavior. However, the stress and strain distribution in the human MCL is highly inhomogenous, even under simple knee motions such as knee flexion [Gardiner and Weiss, 1997, Kennedy et al., 1976]. Further, as the knee is forced to follow injurious motions such as anterior-posterior tibial displacement or varus-valgus rotation, significant loads are transferred from the MCL to the surfaces of the femur and tibia through contact with the bones [Weiss et al., 1996]. The geometry of the ligament insertion sites results in a complex pattern of load transfer to the bones. To study the effects of these factors on knee mechanics using a computational model, a three-dimensional representation is necessary for the ligaments. With an accurate description of MCL material behavior, the finite element (FE) method can be utilized to create a computational model of the MCL. Such a FE model could be used to describe the variation in stresses or material properties of the tissue that occurs as a function of joint orientation, external joint loading, injury, growth and healing.

To describe the anisotropic material behavior of human MCL, multiaxial material property data are necessary. Multiaxial material property data for human MCL are not available in the literature. However, studies have been conducted to determine the uniaxial tensile properties of human ACL, PCL, LCL and patellar tendon (PT) [Butler et al., 1985, Butler et al., 1986, Noyes et al., 1984, Woo et al., 1991, Yoshitsugu et al., 1994]. These studies provided one-dimensional material properties of the ligament along the long axis of the ligament through tensile testing the bone-ligament-bone complex. Only one study has reported any tensile properties for the human MCL. Kennedy et al. measured the ultimate load of human MCL [Kennedy et al., 1976]. However, uniform loading of the ligament was not achieved and cross-sectional area measurements of the ligaments were not included in the report. Thus, material properties could not be extracted from the study. Also, data are not available for ligament tensile properties transverse to the collagen fiber direction for any knee ligament.

The objectives of this research were to determine the longitudinal and transverse tensile material properties of the human MCL and to evaluate the ability of three existing constitutive models to describe the material behavior of the human MCL. To achieve these objectives, tensile tests were conducted with human cadaveric MCL and stress-strain curves were created from the tensile data. The material properties of the human MCL were determined from the curves. The stress-strain curves were transformed into Cauchy stress-stretch curves. Using a nonlinear least squares method, the constitutive models were fit to the experimental data. The ability of each model to describe the ligament behavior was compared based on the value of the correlation coefficient, $R^2$. It was hypothesized that the longitudinal material properties would be significantly greater than the transverse material properties.

# Methods

The methods are divided into two distinct sections - "Material Properties" and "Parameter Estimation for Material Models." The first section describes the specimen preparation, testing protocol, data reduction and statistical analysis performed to obtain the material properties of human MCL. The second section briefly reviews the constitutive models under study and describes the parameter estimation methods.

## Material Properties

**Specimen Preparation.** Ten human cadaveric MCLs were tested (age=62±18 years, 9 males, 1 female). The MCLs were harvested from fresh frozen cadaver legs. The soft tissues surrounding each ligament were carefully dissected away from the ligament. To minimize end effects from clamping, two custom hardened steel punches were used to cut dog-bone shaped specimens from the center section of each ligament (Figure 1). One specimen was cut with the long axis of the punch parallel to the collagen fiber orientation, and the other specimen was cut with the long axis of the punch transverse to the collagen fibers (Figure 2). A total of ten longitudinal specimens and nine transverse specimens were tested. The gauge dimensions were 10 x 2 mm and 6 x 4 mm for the longitudinal and transverse punches, respectively. These dimensions were chosen based on the average dimensions of human MCL and the resolution of the load cell. The length of the transverse punch was chosen based on the width of the ligament while the width of the punch was chosen so that the failure load of the specimen would be large enough to be measured by the load cell.

The ends of each specimen were wrapped in saline soaked gauze and mounted into a set of custom clamps. The clamps were designed such that the face of the clamps in contact with the tissue had teeth to grip the gauze-wrapped tissue ends. The teeth were slightly rounded so that they did not cut the gauze and tissue. The clamp and tissue assembly was loaded into a material test machine (MTS, Eden Prairie, Minnesota). A 2 N tare load was applied to the specimen so that the ligament was taut. This enabled the application of two black markers (1.4 mm diameter, 0.2 mm thickness) to the central section of the specimen using cyanoacrylate. These markers formed a gauge length for analysis of the tissue strain using a video motion analysis system (Peak Performance, Englewood, CO). The zero load position for each ligament was found by successively applying and removing a 1 N load to determine the position of the MTS crosshead at which the ligament was stress-free. At the zero load position the specimen width and thickness were measured using digital calipers. Each measurement was taken three times at the center of the gauge length and averaged (Table 1). The cross-sectional area of the specimen was assumed to be rectangular. During all dissections and testing, the tissue was continuously kept moist with 0.9% normal saline. All testing was conducted at room temperature.

**Test Protocol.** After the zero load was established, a 1N or 2N preload was applied to the transverse or longitudinal specimen, respectively. The specimen was preconditioned by cycling to a maximum amplitude of 0.5 mm at a rate of 10mm/min for 10 cycles. This yielded a strain rate of approximately 1%/s for the longitudinal specimens and 1.7%/s for the transverse specimens. After preconditioning, the zero load was reestablished and the specimen was immediately loaded to failure using the same elongation rates. Load and crosshead position were digitally acquired at 60 Hz using data acquisition software (LABTECH, Wilmington, MA). The load cell was accurate to ± 0.1% full scale (±0.1N). The elongation data were recorded at 60 Hz by a video camera. The boundaries of the black markers were thresholded on the videotape such that the coordinates of their centroids could be automatically tracked through time. Errors in linearity and accuracy of the video system were determined to be less than ± 0.2% strain full scale. Using custom software, strain-time data were generated from the initial and deformed gauge lengths determined using the video motion analysis system.

**Data Reduction and Analysis.** The load-time data from the load cell and strain-time data from the video motion analysis were interpolated at 0.1% strain increments to generate a load-strain curve. Stress-strain ($s-\epsilon$) and Cauchy stress-stretch ($\sigma-\lambda$) curves were created by noting that

$$s = \frac{L}{A}, \quad \epsilon = \frac{l - l_0}{l_0}, \tag{1}$$

$$\sigma = \frac{L\lambda}{A}, \quad \lambda = 1 + \epsilon. \tag{2}$$

where $L$ is the length between the clamps at the zero load, $\lambda$ is the stretch ratio, $A$ is the initial cross-sectional area measured at the center of the gauge length, $l_0$ is the initial distance between the markers and $l$ is the final distance between the markers. The incompressibility assumption has been invoked to determine Cauchy stress from engineering stress.

Experimental data from nine of the longitudinal samples and seven of the transverse samples were used to calculate the material properties for human MCL. Difficulties encountered during the testing of one longitudinal sample and two transverse samples prohibited the inclusion of their values in the calculations of the average material properties. Stress-strain ($s - \epsilon$) curves for the longitudinal and transverse specimens were created from the data collected during the tensile tests of the individual specimens. The average stress-strain curve was created by averaging the stresses from the individual specimens at 1% strain increments over the common strain range. For both the longitudinal and transverse specimens, the stress-strain curves were used to find the tangent modulus for the linear section of each stress-strain curve. The tangent modulus was the slope of the stress-strain curve in the linear portion of the curve and was determined by linear regression. The tensile strength and ultimate strain were determined at the maximum load sustained by the sample. The values for the tangent modulus, tensile strength and ultimate strain were determined for each specimen and then averaged to report the average material properties for human MCL. Because the experimental data did not have a normal distribution, the nonparametric Mann Whitney ranked sum test was used to compare the tensile strength, ultimate strain and tangent modulus for the longitudinal and transverse specimens. Statistical significance was set at $p \leq 0.05$.

## Parameter Estimation for Material Models

Three hyperelastic constitutive models were evaluated to determine their ability to describe the quasistatic pseudoelastic material behavior of human MCL. These models were chosen because they provide a hyperelastic, three-dimensional, anisotropic description of the tissue, capturing the most important effects of the ligament behavior. None of the models allowed for any time- or rate-dependence or multiphasic behavior, however they all could be extended to include such effects (see, i.e., [Puso and Weiss, 1997]). The first two constitutive models were referred to as the "One Coefficient" and "Two Coefficient" material models [Weiss, 1994]. These names refer to the number of material coefficients that were required to describe the behavior of the ground substance matrix, since the collagen fibers had the same representation for both models. The third model, proposed by Lanir [Lanir, 1983] was referred to as the "Lanir" material model. These models are briefly described below. Full descriptions of the models can be found in the original papers.

**One Coefficient and Two Coefficient Models.** For these models, it was assumed that the elastic response of the ligament could be described as originating from the resistance of a composite of collagen fibers, ground substance matrix and their interaction. Further, it was assumed that the collagen fibers were responsible for the transverse isotropy of the ligament and that the ground substance matrix was incompressible and isotropic. For a transversely isotropic, hyperelastic, incompressible material, the strain energy equation was written as

$$W = F_1(I_1, I_2) + F_2(\lambda) + F_3(I_1, I_2, \lambda) \tag{3}$$

where $I_1$ and $I_2$ were the invariants of the right Cauchy stretch tensor and $\lambda$ was the stretch along the collagen fiber direction. The function $F_1$ represented the behavior of the ground substance matrix while $F_2$ represented the behavior of the collagen fibers and $F_3$ represented an interaction, presumed to take the form of a shear coupling, between the matrix and the fibers. For uniaxial testing, there was not any shear in the matrix with respect to the fibers, and thus the $F_3$ component was negligible.

For both the One Coefficient and Two Coefficient models, the behavior of the collagen fibers was modeled by the same equations. However, in the One Coefficient model the behavior of the ground substance matrix was described by the Neo-Hookean material model

$$F_1 = \frac{C_1}{2}\left(I_1 - 3\right), \tag{4}$$

whereas in the Two Coefficient model the behavior of the ground substance matrix was described by Mooney-Rivlin material model

$$F_1 = \frac{C_1}{2}\left(I_1 - 3\right) + \frac{C_2}{2}\left(I_2 - 3\right). \tag{5}$$

The Cauchy stress can then be written as

$$\boldsymbol{\sigma} = 2\left\{(W_1 + I_1 W_2)\mathbf{B} - W_2\mathbf{B}^2\right\} + \lambda W_\lambda \mathbf{a} \otimes \mathbf{a} + p\mathbf{1}, \tag{6}$$

where $\mathbf{B}$ is the left deformation tensor [Spencer, 1980], $W_1$, $W_2$ and $W_\lambda$ are strain energy derivatives with respect to $I_1$, $I_2$ and $\lambda$, respectively, $\mathbf{a}$ is a unit vector field representing the fiber direction in the deformed configuration and $p$ is a hydrostatic pressure.

Some assumptions were made about the behavior of the collagen fibers and incorporated into the function for $F_2$. First, collagen did not support a compressive load. Second, the tensile stress-stretch relationship for collagen was approximated by an exponential for the toe region and a straight line for the linear region. Thus, the strain energy for the collagen fibers was

$$\lambda W_\lambda = 0, \quad \lambda < 1, \tag{7}$$
$$\lambda W_\lambda = C_3 \left(\exp\left(\mathrm{C}_4 \left(\lambda - 1\right)\right) - 1\right), \quad \lambda < \lambda^*,$$
$$\lambda W_\lambda = C_5 \lambda + C_6, \quad \lambda \geq \lambda^*.$$

For these equations, $\lambda^*$ was the stretch value where the collagen fibers were straightened, $C_3$ scaled the exponential stresses, $C_4$ controlled the collagen fiber uncrimping rate, and $C_5$ was the modulus of the straightened collagen. $C_6$ was determined from the condition that ensured that the exponential and linear regions were $C^0$ continuous at $\lambda^*$

$$C_6 = C_3 \left(\exp\left(\mathrm{C}_4 \left(\lambda^* - 1\right)\right) - 1\right) - C_5 \lambda^*. \tag{8}$$

**Lanir Model.** Lanir proposed a simple three-dimensional model for fiber-reinforced collagenous tissues that described the stress in ligament as a function of the tensile stretch in the collagen fibers and a hydrostatic pressure from the incompressible ground substance matrix [Lanir, 1983]. This model represented a composite of undulating collagen fibers embedded in a fluid matrix. The model assumed that the collagen fibers did not support a compressive load and the unfolding of the fibers during deformation squeezed the matrix, resulting in an internal hydrostatic pressure:

$$\boldsymbol{\sigma} = \lambda W_\lambda \mathbf{a} \otimes \mathbf{a} + p\mathbf{1}. \tag{9}$$

Using this model to represent a uniaxial tensile test along the fiber direction, the entire stress was on the collagen fibers; thus, $\lambda W_\lambda$ was described by (7b and 7c) and $p = 0$. Transverse stretching of the structure would not cause any stress in the tissue with this simple representation for the ligament. Thus, the Lanir model was only used to describe the longitudinal tensile data.

**Parameter Estimation Methods.** The Cauchy stress-stretch curves were used to determine the material coefficients for each material model studied. Paired data (a longitudinal and a transverse stress-strain curve for each specimen) were necessary for evaluating the One Coefficient and Two Coefficient material models. Seven sets of paired data were used for the parameter estimation. A procedure was developed to determine the material coefficients from the experimental data using an unconstrained nonlinear least squares method [Marquart, 1963]. It can be summarized as follows:

- The matrix coefficients, $C_1$ and $C_2$ were determined using nonlinear regression with the transverse tensile data. (The Lanir model did not describe the behavior of the ground substance matrix.)

- The uncrimped stretch, $\lambda^*$, the transition point from the exponential curve to the linear region, was estimated by inspection of the longitudinal Cauchy stress-stretch curve. Using a straight edge, a line was drawn through the linear region of the curve. The point where the experimental data deviated from the line was called $\lambda^*$.

- The toe region coefficients, $C_3$ and $C_4$, from (7b) were determined using nonlinear regression with the longitudinal data below $\lambda^*$ and $C_1$ and $C_2$.

- The linear region coefficient, $C_5$, from (7c) was determined using nonlinear regression with the longitudinal data above $\lambda^*$ and the coefficients $C_1 - C_4$.

- The experimental data were plotted with the material model fits and the regression values, $R^2$, for the curve fits were calculated to compare the ability of the models to describe the experimental data.

- An analysis was performed to assess the inter- and intra-observer variability in the determination of $\lambda^*$ and the effects on the material coefficients in the material models. In this study, $\lambda^*$ for the average Cauchy stress-stretch curve was varied by $\pm 0.005$ and the coefficients $C_1$ - $C_5$ for each model were recalculated.

# Results

## Material Properties

During a typical tensile test, the ligament material was uniformly stretched as the clamps moved apart. Failure of the ligament occurred abruptly. Visibly, failure in the longitudinal specimens could be defined by the collagen fibers in the ligament slipping past each other. Further stretching of the ligament resulted in a gross disruption of fibers within the ligament substance. All of the specimens used in this study failed in the central part of the structure and no slipping of the ligament between the clamps was observed on the video or in the experimental data. Failure of the ligament occurred within the accurate range of the load cell for all specimens.

The longitudinal average stress-strain curve exhibited the typical nonlinear behavior that has been reported for other ligaments (Figure 3) [Butler et al., 1986, Daniel et al., 1990, Noyes et al., 1984, Woo et al., 1990, Woo et al., 1991]. The initial toe region of the curve was characterized by a low tangent modulus. With the application of additional load, the modulus of the curve rapidly increased. After approximately 5% strain, the curve became linear and reached its highest modulus. The toe region was absent from the stress-strain curves of the transverse specimens. Nine specimens were used to calculate the average longitudinal material properties while seven specimens were used to calculate the transverse material properties. For the individual and average stress-strain curves, the longitudinal tensile strength and ultimate strain were significantly greater than the transverse tensile strength ($p=0.001$) and ultimate strain ($p=0.03$) (Table 2). The linear region of the stress-strain curves was typically between 7-9 % strain for the longitudinal samples and 4-6 % strain for the transverse samples. The modulus of the longitudinal specimens was over an order of magnitude higher than the transverse specimens ($p=0.002$). The material properties for the longitudinal MCL specimens were comparable to those reported by Butler et al. [Butler et al., 1986] for human ACL, PCL and LCL.

## Parameter Estimation for Material Models

Although the regression values, $R^2$ were all greater than 0.9, the graph of the curve fits reveals that the Two Coefficient model provided a better description of the transverse data than the One Coefficient model (Figure 4A, Tables 3 and 4). The $C_1$ and $C_2$ coefficients obtained from the One Coefficient and Two Coefficient material models described the behavior of the ground substance matrix (Tables 3 and 4). There was more variability between the coefficients for the Two Coefficient model. Because the Two Coefficient model more closely described the behavior of the ligament, this indicates that there was a large variability in the behavior of the specimens tested.

The variation in $\lambda^*$ had a small effect on the outcome of $C_1$ - $C_5$ for the One Coefficient material model. A 0.5% error in determining $\lambda^*$ for the average Cauchy stress-stretch curve caused a 19% error in the $C_3$ coefficient, a 13% error in the $C_4$ coefficient and a 1% error in the $C_5$. There was no change for the $C_1$ coefficient. Similar errors for the $C_3$, $C_4$ and $C_5$ coefficients were seen for the Two Coefficient and Lanir material models. The $C_1$ and $C_2$ coefficients did not change for the Two Coefficient material model and the Lanir material model did not have $C_1$ and $C_2$ coefficients.

Using $C_1$, $C_2$ and $\lambda^*$, the remaining coefficients $C_3$, $C_4$ and $C_5$ describing the collagen fiber material behavior were determined from the longitudinal Cauchy stress-stretch curves (Tables 3, 4, and 5). Regardless of which model was used to describe the matrix, there was close agreement between the experimental data and the models for the longitudinal data (Figure 4B). The coefficients $C_3$ and $C_4$ were the material coefficients for the collagen fibers in the toe region of the longitudinal curve while $C_5$ was the material coefficient that described the modulus of the straightened collagen fibers in the linear region of the longitudinal curve. The $C_5$ coefficients obtained from each model for the average Cauchy stress-stretch curve were similar to the average tangent modulus reported above (Table 2). There was very little difference between the coefficients obtained by the different models (Tables 3, 4, and 5). This was attributed to the low tensile stress contributed by the matrix in comparison to the collagen fibers. The regression values for the toe and linear regions for all models were greater than 0.98. However, there was a large variation of the coefficients for both the toe and linear regions. Because the models fit the experimental data well, the variation in the coefficients indicated that there was a large variation in the material properties between the specimens.

# Discussion

The uniaxial tensile material properties of human MCL were determined both along and transverse to the collagen fiber direction, and the ability of three material models to describe the material behavior of human MCL was assessed. The results of this study showed that the material properties along the collagen fiber direction were significantly greater than the transverse material properties. For the transverse specimens, the Two Coefficient material model described the material behavior better than the One Coefficient model. The three models described the behavior of the longitudinal specimens equally well.

**Material Properties.** Only one other study has investigated the strength of the human MCL [Kennedy et al., 1976]. Tensile tests of the entire human MCL were conducted. However, only the load at which the ligaments failed was reported (467.46±33.32 N). The area of the MCL was not measured and so the tensile strength could not be estimated for comparison to the tensile strength values from the current study. No other material properties of human MCL were available to compare with the results of this study.

Butler et al. [Butler et al., 1986] reported the material properties of human ACL, PCL, LCL and PT. The properties for the ACL, PCL and LCL were similar and thus reported as one value (mean±standard error). The material properties for the human MCL along the fiber direction were comparable to the values reported by Butler et al. They found the average tensile strength of ACL, PCL and LCL was 36.4±2.5 MPa, the ultimate strain was 15.0±0.8% and the modulus was 345.0±22.4 MPa. In the present study of the MCL, the average tensile strength was 38.6±4.8 MPa, the ultimate strain was 17.1±1.5% and the modulus was 332.2±58.3 MPa for the longitudinal specimens.

The present work is the first to provide data on the uniaxial tensile material properties of ligament transverse to the collagen fiber direction. Thus, other data are not available for comparison. The results of this study have shown that the ligament can resist a small transverse tensile load. However, at physiological strains, $< 4\%$, the load in the transverse direction is not insignificant. Since the collagen is the primary loadbearing component along the long axis of the MCL, the longitudinal tensile tests mainly reflect the behavior of the collagen. However, the transverse tests measure the contributions from the ground substance matrix (and possible interfiber and fiber-matrix interactions, which were not included in the material models studies). Quantification of interactions between the matrix and fibers would require more detailed material testing to assess, and conclusions regarding these interactions cannot be drawn from the uniaxial tensile tests performed in this study.

It is possible that removing a test sample from the MCL may alter the overall mechanical response of the tissue by cutting the collagen fibers and breaking interfiber and transfiber bonds. However, it is impossible to uniformly load the entire MCL. As such, to achieve a uniformly loaded test sample, dog-bone shaped punches were used to cut samples from the ligament tissue. The dog-bone shape was used to ensure uniform loading of the specimens and to minimize the end effects from clamping the test specimens.

**Parameter Estimation for Material Models.** The One Coefficient, Two Coefficient and Lanir material models were good descriptors of the tensile test data. The longitudinal tensile data were fit equally well by all three models but the transverse data were fit best by the Two Coefficient model. Thus, the Two Coefficient model provided the best overall description of the three-dimensional behavior of the MCL.

A previous study tested the ability of the One Coefficient and Two Coefficient material models to describe and predict the three-dimensional behavior of human fascia lata

[Weiss, 1994]. In this study, both uniaxial tensile tests and strip biaxial tests were used to determine and validate the material coefficients of the One Coefficient and Two Coefficient models for human fascia lata. The coefficients for human fascia lata were of the same magnitude as the average coefficients for human MCL. The coefficients that described the ground substance behavior for the fascia lata and the MCL were similar. However, the coefficients that described the collagen fiber behavior were somewhat different. The $C_3$ coefficient for fascia lata and MCL was similar but the $C_4$ coefficient was over two times greater for the fascia lata than for the MCL and the $C_5$ coefficient was three times greater for the fascia lata than for the MCL.

This evaluation of the One Coefficient, Two Coefficient and Lanir material models provided material coefficients to be used in constitutive models of human MCL. The One Coefficient and Two Coefficient models described the three-dimensional behavior of the MCL while the Lanir model described the behavior of the collagen fibers only. This study also provided investigators with information on $\lambda^*$, the point at which the collagen fibers were uncrimped. The variation in both the coefficients and $\lambda^*$ illustrated variability of the MCL specimens. However, the variability of the material coefficients was also dependent on the determination of $\lambda^*$ as shown in the Results section. The $C_1$ and $C_2$ coefficients were not affected by variation in $\lambda^*$. This was expected because these coefficients described the material behavior of the ground substance matrix and were not dependent on $\lambda^*$. The largest error was in $C_3$ coefficient, followed by the $C_4$ coefficient and then the $C_5$ coefficient. If the errors in the determination of $\lambda^*$ amount to 0.5% strain or less, then all the coefficients had an error of 20% or less for any of the material models.

An investigation of material inhomogeneities within the MCL was not performed. It is likely that the properties are dependent on position within the structure, as the material did not appear uniform in thickness or composition during dissection. Specimens in this study were harvested from an area of the ligament that consistently had uniform thickness and color between specimens. However, future work should attempt to address the material inhomogeneities within this ligament and other ligamentous structures.

This study provided measurements of the longitudinal and transverse tensile material properties of human MCL. This study also analyzed the ability of three constitutive models to describe the material behavior of human MCL. It established coefficients that describe the one- and three-dimensional behavior of the MCL. These data are necessary to develop three-dimensional material representations for knee ligaments, which will allow simulations of the mechanical behavior. Also, this study provided information on the interspecimen variability of human MCL. Finally, it established a range of values for the uncrimped stretch ratios to be used in constitutive modeling efforts. The material data and test techniques developed in this work will form the basis for ongoing computational studies of the stresses in the human MCL as a function of joint flexion angle, valgus rotation, and anterior-posterior knee translation.

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Table 1: Physical dimensions of the MCL test specimens measured in the zero load configuration (mean $\pm$ standard deviation).

| Specimen Type | Initial Length (mm) | Width (mm) | Thickness (mm) |
|---|---|---|---|
| Longitudinal | $15.07 \pm 2.46$ | $1.75 \pm 0.21$ | $1.57 \pm 0.36$ |
| Transverse | $9.70 \pm 2.20$ | $3.56 \pm 0.26$ | $1.29 \pm 0.34$ |

Table 2: Average material properties of human MCL along (n=9) and transverse (n=7) to the collagen fiber direction (mean $\pm$ standard error).

| Property | Longitudinal | Transverse |
|---|---|---|
| Tensile Strength (MPa) | $38.56 \pm 4.76$ | $1.69 \pm 0.53$ |
| Ultimate Strain (%) | $17.11 \pm 1.53$ | $11.7 \pm 0.93$ |
| Tangent Modulus (MPa) | $332.15 \pm 58.27$ | $11.02 \pm 3.57$ |

Table 3: Material coefficients for the One Coefficient material model for the seven specimens tested and for the average Cauchy stress-stretch curve. The $R^2$ values ($R_T^2$ for the transverse data, $R_L^2$ for the longitudinal data) reflect the ability of the model to describe the experimental data. "Avgcurv" indicates coefficients from tha average Cauchy stress-stretch curve (Figure 4).

| Specimen | $C_1$ (MPa) | $C_2$ (MPa) | $R_T^2$ | $\lambda^*$ | $C_3$ (MPa) | $C_4$ | $C_5$ (MPa) | $R_L^2$ |
|---|---|---|---|---|---|---|---|---|
| 28433 | 1.1 | - | 0.9727 | 1.035 | 2.7 | 46.6 | 513.7 | 0.9940 |
| 33199 | 3.8 | - | 0.9136 | 1.02 | 0.4 | 108.3 | 284.7 | 0.9896 |
| 33623 | 10.4 | - | 0.9899 | 1.05 | 0.1 | 66.8 | 94.6 | 0.9940 |
| 32326 | 1.9 | - | 0.9747 | 1.12 | 1.5 | 17.6 | 187.3 | 0.993 |
| 32339 | 1.9 | - | 0.9649 | 1.065 | 2.0 | 32.6 | 621.4 | 0.9892 |
| 32651 | 1.08 | - | 0.9868 | 1.055 | 2.9 | 38.0 | 617.5 | 0.9986 |
| 32652 | 12.8 | - | 0.9813 | 1.09 | 2.1 | 25.3 | 406.1 | 0.9994 |
| avgcurv | 4.6 | - | 0.9742 | 1.055 | 2.4 | 30.6 | 323.7 | 0.9990 |

Table 4: Material coefficients for the Two Coefficient material model for the seven specimens tested and for the average Cauchy stress-stretch curve. The $R^2$ values ($R_T^2$ for the transverse data, $R_L^2$ for the longitudinal data) reflect the ability of the model to describe the experimental data. "Avgcurv" indicates coefficients from tha average Cauchy stress-stretch curve (Figure 4).

| Specimen | $C_1$ (MPa) | $C_2$ (MPa) | $R_T^2$ | $\lambda^*$ | $C_3$ (MPa) | $C_4$ | $C_5$ (MPa) | $R_L^2$ |
|---|---|---|---|---|---|---|---|---|
| 28433 | 6.5 | -5.7 | 0.9981 | 1.035 | 2.7 | 46.4 | 512.4 | 0.9943 |
| 33199 | 47.0 | -46.0 | 0.9992 | 1.02 | 0.5 | 102.3 | 279.4 | 0.9875 |
| 33623 | 31.2 | -22.1 | 0.9935 | 1.05 | 0.1 | 59.5 | 85.5 | 0.9946 |
| 32326 | 9.4 | -8.2 | 0.9971 | 1.12 | 1.6 | 17.2 | 183.6 | 0.9993 |
| 32339 | 13.6 | -12.6 | 1 | 1.065 | 2.1 | 32.2 | 618.2 | 0.9984 |
| 32651 | 1.9 | -1.2 | 0.9979 | 1.055 | 2.9 | 37.9 | 617.9 | 0.9986 |
| 32652 | 49.3 | -40.0 | 0.9948 | 1.09 | 2.4 | 23.9 | 392.9 | 0.9995 |
| avgcurv | 30.1 | -27.1 | 0.9989 | 1.055 | 2.6 | 29.5 | 317.9 | 0.9991 |

Table 5: Material coefficients for the Lanir material model for the seven specimens tested and for the average longitudinal Cauchy stress-stretch curve, describing the material behavior of the collagen fibers. "Avgcurv" indicates coefficients from tha average Cauchy stress-stretch curve (Figure 4).

| Specimen | $\lambda^*$ | $C_3$ (MPa) | $C_4$ | $C_5$ (MPa) | $R^2$ |
|---|---|---|---|---|---|
| 28433L | 1.035 | 2.8 | 46.1 | 517.1 | 0.9944 |
| 33199L | 1.02 | 0.5 | 100.5 | 296.3 | 0.9896 |
| 33623L | 1.05 | 0.9 | 31.2 | 126.6 | 0.9995 |
| 32326L | 1.12 | 1.9 | 16.4 | 193.7 | 0.9993 |
| 32339L | 1.065 | 2.2 | 31.7 | 627.2 | 0.9984 |
| 32651L | 1.055 | 3.0 | 37.6 | 620.5 | 0.9986 |
| 32652L | 1.09 | 4.0 | 20.7 | 446.8 | 0.9995 |
| avgcurv | 1.055 | 3.0 | 28.4 | 337.9 | 0.9990 |

Figure 1: Hardened steel punches were used to cut the longitudinal and transverse dog-bone shaped test specimens from the ligament.

Figure 2: Location of the harvest locations for longitudinal and transverse tensile test specimens of human MCL. Longitudinal specimens were taken from the anterior-distal portion of the MCL, while transverse specimens were taken from the anterior-central two-thirds of the MCL.

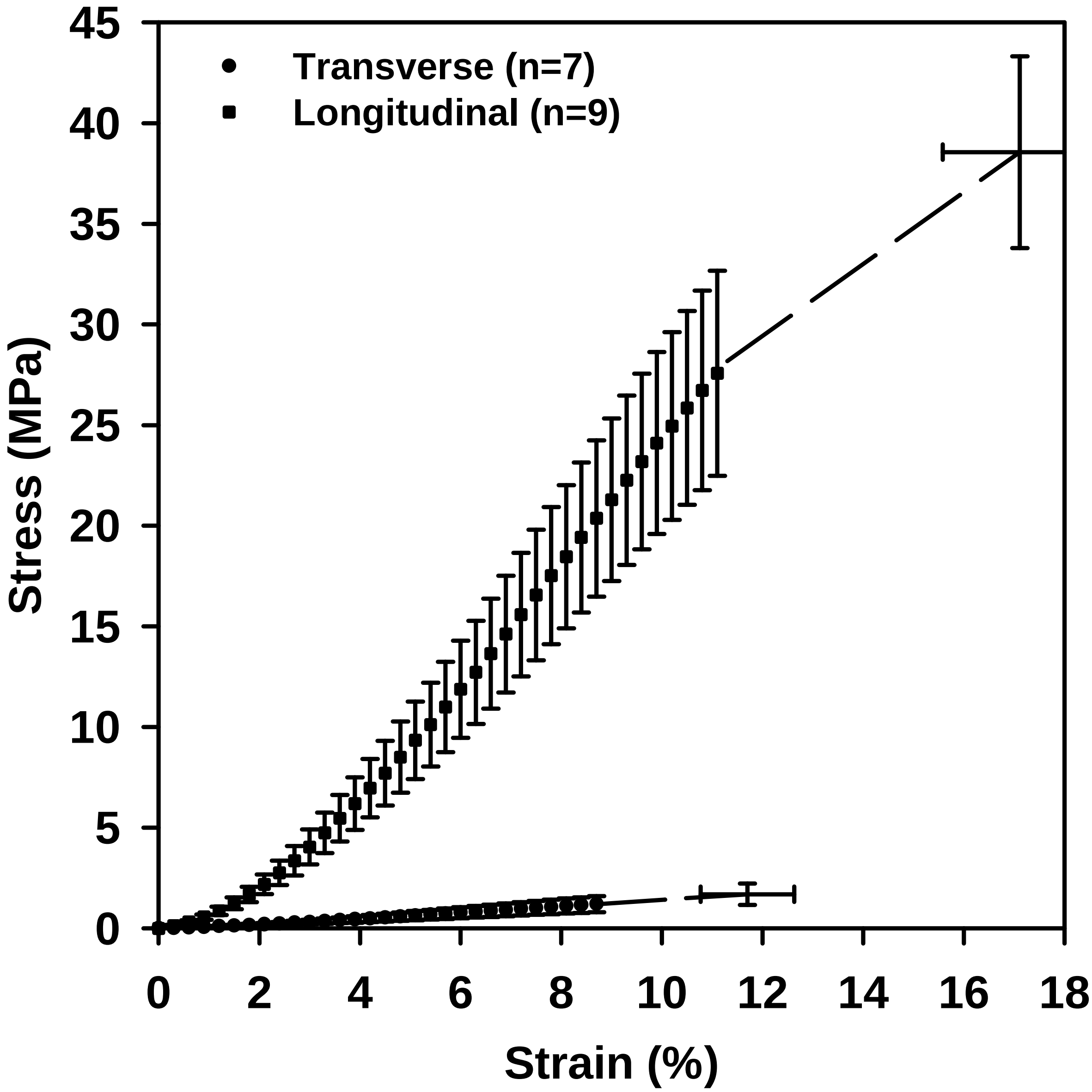


Figure 3: Stress-strain curves for human MCL along (longitudinal) and transverse to the collagen fiber direction. The final point in each curve was the average failure point for the specimens. The error bars indicate the standard error for the average ultimate strain and tensile stress. The tangent modulus of the longitudinal curve was over an order of magnitude greater than the tangent modulus of the transverse curve.

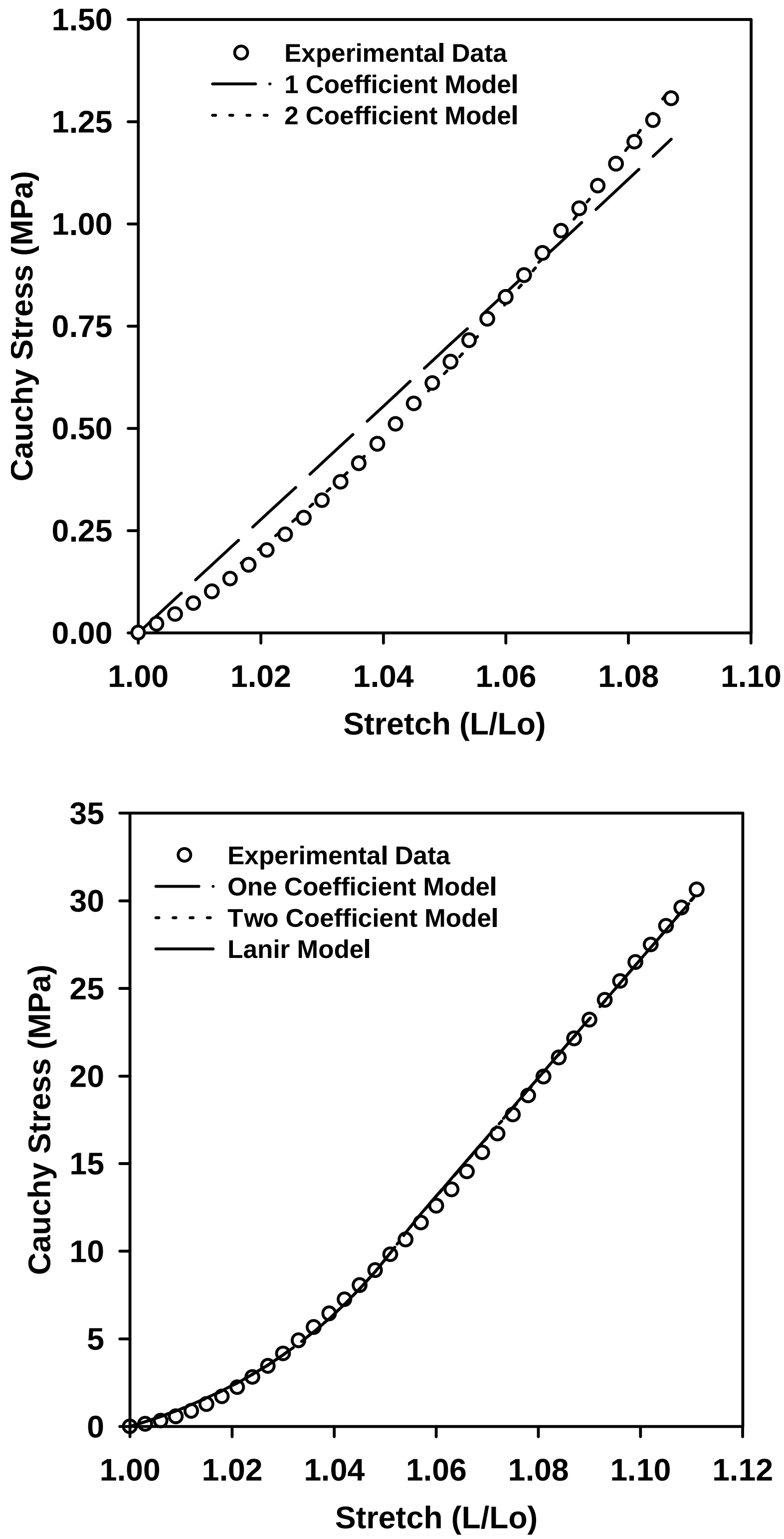


Figure 4: A.) Average Cauchy stress-stretch curves for the specimens transverse to and B.) along the collagen fiber direction, and the material model curve fits. The Two Coefficient model provided a better description of the transverse data, while all three models provided a good description of the longitudinal data. In 4B the lines representing the model fits are coincident.