Identifying Dominant Structural and Material Contributors to Enthesis Stress Redistribution Via Cotter's Method supplementary information

Model Material Parameters and Justification in Structural-intensive sets (VS,TS1 and TS2)

Cotter's Analysis Variable Parameters: These parameters acted as variables for the 3 sets of Cotter's analysis. Thus, they have two alternate values to them, which was applied in the models as described in Cotter's Method Calculation.

- Overall Enthesis Geometry (TS1, VS): For the OFF value of the analysis, the overall geometry was a 700x200unit rectangle. All the regions had the same width, the length of each region were as followed: Ligamentous region 340 units, UFC: 110 units, CFC 60 units, SB 190 units. For ON value, a bottle shaped structure based on Bovine medial meniscal enthesis[1] was used. Total length and length of individual regions were the same. The enthesis was then modelled a semi-hourglass figure with widths changing as spline in the following intervals: 200 units at beginning of ligamentous region, 150 units at Lig/UFC interface, 60 units at tidemark, 130 units at cement line, 230 units at end of SB. All of the models with distinct geometrical properties can be found by contacting PI at darryl.dickerson@fiu.edu .

- Mean fiber orientation in ligamentous region: The fiber orientation was provided by changing material orientation of the section with respect to origin. The two values were -25º and -45º, based on an average orientation of Rabbit Achilles tendon[2]
- Fiber Disorganization in ligamentous region: Represented by the unitless parameter kappa. The discrete values 1/30 and 1/15 were used based on ranges of dispersion noted in Rabbit Achilles tendon[2] which was compared for appropriate kappa values in literature(Gasser, Ogden and Holzapfel, 2006, Fig 5)
- Intra-fiber dispersion in ligamentous region: This was represented by the unitless parameter k2. Based on intra- dispersion ranges noted in literature[4] in collagen fibers in the interstitial aorta, the values 5 and 35 were chosen. Results from set TS2 were given greater weight when deducing the effect of k2.
- Mean fiber orientation in UFC: The fiber orientation was provided by changing material orientation of the section with respect to origin. The two values were -52º and -48º, based on an average orientation of Human meniscal enthesis[5]
- Fiber Disorganization in UFC: Represented by the unitless parameter kappa. The discrete values 1/18 and 1/15 were used based on ranges of dispersion noted in Rabbit Achilles tendon[5] which was compared for appropriate kappa values in literature(Gasser, Ogden and Holzapfel, 2006, Fig 5)
- Intra-fiber dispersion in UFC: This was represented by the unitless parameter k2. Based on intra- dispersion ranges noted in literature[4] in collagen fibers in the interstitial aorta, the values 5 and 35 were chosen.

- Mean fiber orientation in CFC: The fiber orientation was provided by changing material orientation of the section with respect to origin. The two values were -42° and -46°, based on an average orientation of Human meniscal enthesis[5]
- Fiber Disorganization in CFC: Represented by the unitless parameter kappa. The discrete values 8/75 and 2/15 were used based on ranges of dispersion noted in Rabbit Achilles tendon[5] which was compared for appropriate kappa values in literature(Gasser, Ogden and Holzapfel, 2006, Fig 5)
- Geometry of Ligamentous/UFC interface: This parameter was controlled by manipulating the geometry of border separating the Ligamentous region and the UFC region. For Cotter sets TS2 and TS1, the OFF value was a straight line. For Set TS1, the ON value was a sinusoidal curve of amplitude and period of 1 unit, running parallel to the Y-axis with the X coordinate of the straight line previously mentioned as the Center. For set TS2, it was a curve with amplitude and period of 10 units.
- Geometry of Tidemark interface: This parameter was controlled by manipulating the geometry of border separating the CFC region and the UFC region. For Cotter sets VS, TS2 and TS1, the OFF value was a straight line. For Set TS1 and VS, the ON value was a sinusoidal curve of amplitude and period of 1 unit, running parallel to the Y-axis with the X coordinate of the straight line previously mentioned as the Center. For set TS2, it was a curve with amplitude and period of 10 units.
- Geometry of Cement line interface: This parameter was controlled by manipulating the geometry of border separating the SB region and the CFC region. For Cotter sets TS2 and TS1, the OFF value was a straight line. For Set TS1, the ON value

was a sinusoidal curve of amplitude and 1 unit and period of 10 units, running parallel to the Y-axis with the X coordinate of the straight line previously mentioned as the Center. For set TS2, it was a curve with amplitude and period of 10 units.

Invariant Parameters: These parameters are kept same all over the test cases

- Overall geometry (TS2): TS2 is the expanded version of a 390 by 30 unit slice taken from sets TS1 and magnified by 2, resulting in a 780x60 unit rectangle. The lengths of Ligamentous, UFC, and SB regions are 220 units and CFC is 120 units.
- Fiber Stiffness in Ligamentous region: This is represented in HGO model via parameter k1. A value of 128MPa based on findings on ACL fiber stiffness[6].
- Ground Matrix Stiffness in Ligamentous region: Represented by parameter C10 in HGO model. A value of 10MPa was chosen based on interpolation of ground matrix stiffness on unmineralized region of meniscal enthesis via nanoindentation away from the tidemark[5]
- Compressibility of Ligamentous region: Represented by dimensionless parameter D. Value of 0 was given due to absence of collagen type II which indicates non-compression under tensile or shear forces[7–9]
- Fiber Stiffness in UFC : A lower value of 80MPa was assigned to parameter k1, due to presence of Collagen type II in UFC[9], which has been shown to decrease stiffness compared to collagen fibers made of purely Collagen type I[10].
- Matrix Stiffness in UFC: Represented by parameter C10. Values, obtained via nanoindentation[5], was modelled as a linear decrease from 8MPa at the tidemark to 2MPa at the ligamentous/UFC interface.

- Compressibility of UFC: Represented by dimensionless parameter D. Value of 1e-6 was given due to presence of collagen type II which indicates non-compression under tensile or shear forces[8,9]
- Fiber Stiffness in CFC: The fibers in the CFC are linearly mineralized from tidemark to cement line[11]. Based on this, the value of parameter k1 ranged linearly from fiber stiffness at UFC of 80MPa (Described before) to fiber stiffness value of 3900MPa at SB (Virtual fibers, result from literature[12]).
- Matrix Stiffness in CFC: This is represented by the parameter C10. The value was linearly increased from 150MPa at the tidemark to 600MPa at the cement line. This increase is based on nanoindentation results[5] and linear increase in mineralization at the CFC[11].
- Intra-fiber dispersion in CFC: Since the fibers in CFC are increasingly mineralized, an arbitrary value of 2.5 was given to the dimensionless parameter k2 to ensure highest failure stress based on literature findings[4].
- Compressibility of CFC: Represented by dimensionless parameter D. Value of 1e-6 was given due to presence of collagen type II which indicates non-compression.
- Fiber Stiffness in SB: Since fibers in the SB region were considered as virtual fibers, their stiffness was the same as Matrix stiffness, 3.9GPa, based on indentation values[12].
- Matrix Stiffness in SB: Value of 3.9GPa was used, based on indentation values[12].

- Mean fiber orientation in SB: Fiber orientation of 0^{0} was inputted. Since the SB region was modelled as isotropic, this value has not significant effect other than completing the model values.
- Fiber Disorganization in SB: Represented by the dimensionless parameter kappa. A value 0.3 was used to indicate isotropy[4].
- Intra-fiber dispersion in SB: Since virtual fibers were employed in the SB region, an arbitrary value of 2.5 was given to the dimensionless parameter k2 to ensure highest failure stress based on literature findings[4].
- Compressibility of SB: Represented by dimensionless value D. Value of 0 was given due to absence of collagen type II which indicates non-compression under tensile or shear forces[7,9]

Model Material Parameters and Justification in Composition-intensive sets (TS3)

Testing parameters: These values have an ON or OFF value based on test case number in accordance with Cotter's statistical method protocols:

- Parameters changed in an identical manner in set TS2
    - Geometry of Lig/UFC interface
    - Geometry of Tidemark
    - Geometry of Cement line

- Fiber Stiffness in Ligamentous region: This is represented in HGO model via parameter k1. The values of 100 MPa and 120MPa were chosen based on recorded value value of 128MPa based on findings on ACL fiber stiffness[6].
- Ground Matrix Stiffness in Ligamentous region: Represented by parameter C10 in HGO model. The ON/OFF values of 5MPa and 10MPa were chosen based on value of 10MPa was chosen based on interpolation of ground matrix stiffness on unmineralized region of meniscal enthesis via nanoindentation away from the tidemark[5]
- Fiber Stiffness in UFC : A lower ON/OFF values of 80MPa/60MPa was assigned to parameter k1, due to presence of Collagen type II in UFC[9], which has been shown to decrease stiffness compared to collagen fibers made of purely Collagen type I[10].
- Matrix Stiffness in UFC: Represented by parameter C10. Values, obtained via nanoindentation[5], ON/OFF values were modelled as a linear decrease [from 14MPa to 2MPa] and [from 8MPa to 2MPa] from tidemark to the ligamentous/UFC interface.
- UFC region fiber disorganization gradient: The kappa value was made temperature dependent over the ranges [1/30 to 1/15] and [1/30 to 1/10] from Lig/UFC interface to the tidemark. Temperature varied linearly as explained later. This replaced the disorganization magnitude parameter in set TS2.
- Fiber Stiffness in CFC: The fibers in the CFC are linearly mineralized from tidemark to cement line[11]. Based on this, the OFF value of parameter k1 ranged

linearly from fiber stiffness at UFC of 80MPa (Described before) to fiber stiffness value of 3900MPa at SB (Virtual fibers, result from literature[12]). The ON value was of parameter k1 ranged linearly from fiber stiffness at UFC of 80MPa (Described before) to fiber stiffness value of 7720MPa at SB (Virtual fibers), to match with 7720MPa SB stiffness value.

- Matrix Stiffness in CFC: This is represented by the parameter C10. The OFF value was linearly increased from 150MPa at the tidemark to 600MPa at the cement line. This increase is based on nanoindentation results[5] and linear increase in mineralization at the CFC[11]. The ON value was a theoretical increase to 1050MPa at the cement line
- Overall stiffness in SB: Controlled by simultaneous change of the following two parameters:
    - Fiber Stiffness in SB: Since fibers in the SB region were considered as virtual fibers, their stiffness was the same as Matrix stiffness, 3.9GPa, based on indentation values[12]. A theoretical ON value of 7.72GPa was used to provide contrast.
    - Matrix Stiffness in SB: Value of 3.9GPa was used, based on indentation values[12]. A theoretical ON value of 7.72GPa was used to provide contrast.

Invariant Parameters: These parameters have been kept the same over all the test cases

- Parameters kept constant in an identical manner in set TS2
    - Overall Geometry
    - Compressibility of Ligamentous region

  - Compressibility of UFC
  - Intra-fiber dispersion in CFC
  - Compressibility of CFC
  - Mean fiber orientation in SB
  - Fiber disorganization in SB
  - Intra-fiber dispersion in SB
  - Compressibility of SB
- Mean fiber orientation in ligamentous region: The fiber orientation was provided by changing material orientation of the section with respect to origin. The value of 35°, based on an average orientation of Rabbit Achilles tendon[2]
- Fiber Disorganization in ligamentous region: Represented by the unitless parameter kappa. The discrete values 1/20 were used based on ranges of dispersion noted in Rabbit Achilles tendon[2] which was compared for appropriate kappa values in literature(Gasser, Ogden and Holzapfel, 2006, Fig 5)
- Intra-fiber dispersion in ligamentous region: This was represented by the unitless parameter k2. Based on intra- dispersion ranges noted in literature[4] in collagen fibers in the interstitial aorta, the value 35 was chosen. Results from set TS2 were given greater weight when deducing the effect of k2.
- Mean fiber orientation in UFC: The fiber orientation was provided by changing material orientation of the section with respect to origin. The chosen value was -50°, based on an average orientation of Human meniscal enthesis[5]

- Intra-fiber dispersion in UFC: This was represented by the unitless parameter k2. Based on intra- dispersion ranges noted in literature[4] in collagen fibers in the interstitial aorta, the value 35 was chosen.
- Mean fiber orientation in CFC: The fiber orientation was provided by changing material orientation of the section with respect to origin. The two values were -44^{o}, based on an average orientation of Human meniscal enthesis[5]
- Fiber Disorganization in CFC: Represented by the unitless parameter kappa. The discrete value 0.12 was used based on ranges of dispersion noted in Rabbit Achilles tendon[5] which was compared for appropriate kappa values in literature(Gasser, Ogden and Holzapfel, 2006, Fig 5)

HGO constitutive equations

The HGO model is based on the strain energy function (U) of the general form, where C is the right Cauchy-Green deformation tensor and $\boldsymbol{n}$ is fiber direction vector :

$$U = U_{iso}(C) + U_{anio}(C, \boldsymbol{n}) \tag{1}$$

The overall function form for HGO is provided as such (Taken from Abaqus documentation: https://abaqus-docs.mit.edu/2017/English/SIMACAEMATRefMap/simamat-c-anisohyperelastic.htm#simamat-c-anisohyperelastic-holzapfel)

$$U = C_{10}(\bar{I} - 3) + \frac{1}{D}\left[\frac{[J^{el}]^2 - 1}{2} - lnJ^{e}l\right] + \frac{k_1}{2k_2}\sum_{\alpha=1}^{N}\{\exp[k_2\langle\bar{E}_\alpha\rangle^2] - 1\} \tag{2}$$

With

$$\bar{E}_\alpha \stackrel{\text{def}}{=} \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)\left(\bar{I}_{4(\alpha\alpha)} - 1\right) \tag{3}$$

where U is the strain energy per unit of reference volume; C_{10}, D, k_1, k_2, and κ are temperature-dependent material parameters; N is the number of families of fibers ($N \leq 3$); $\bar{I}_1$ is the first deviatoric strain invariant; $J^{e}l$ is the elastic volume ratio as defined below and $\bar{I}_{4(\alpha\alpha)}$ are *pseudo-invariants* of the distortional part of the Right Cauchy-Green strain tensor and reference configuration vectors of the characterized fibers (***n***).

The model assumes that the directions of the collagen fibers within each family are dispersed (with rotational symmetry) about a mean preferred direction. The parameter κ ($0 \leq \kappa \leq 1/3$) describes the level of dispersion in the fiber directions. If ρ(θ) is the orientation density function that characterizes the distribution (it represents the normalized number of fibers with orientations in the range [θ, θ +d θ] with respect to the mean direction), the parameter κ is defined as:

$$\kappa = \frac{1}{4}\int_{0}^{\pi} \rho(\theta)\sin^3\theta d\theta \tag{4}$$

It is also assumed that all fiber bundles have same mechanical properties and same dispersion, with κ =0 being no dispersion and κ =1/3 being completely dispersed.

Thus, $\bar{I}_{4(\alpha\alpha)}$ term controls the direction of mean fiber orientation and κ term controls the dispersion of the fibers about mean direction.

The second Piola-Kirchoff stress tensor *S* is then taken via

$$S = 2\frac{\partial U}{\partial C} \tag{5}$$

Then the second Piola-Kirchoff stress tensor *S* is transformed to the Cauchy stress tensor σ

$$\sigma = \frac{1}{J} F S F^{\mathrm{T}} \tag{6}$$

Where *F* is the deformation gradient, and *J* is the determinant of *F*.

The deviatoric part of σ, ***s*** is calculated (**I** being the identity matrix)

$$\boldsymbol{s} = \sigma - \frac{1}{3}\mathbf{tr}(\sigma)\mathbf{I} \tag{7}$$

Finally, the Von-Mises stress σ_m calculated by

$$\sigma_m = \sqrt{\frac{3}{2}(\boldsymbol{s}:\boldsymbol{s})} \tag{8}$$

Where $(\boldsymbol{s}:\boldsymbol{s})$ is the double dot product of ***s*** with itself.

Modelling SB region as isotropic

Let us recall that

$$U = C_{10}(\bar{I} - 3) + \frac{1}{D}\left[\frac{[J^{el}]^2 - 1}{2} - ln J^{\mathrm{e}}l\right] + \frac{k_1}{2k_2}\sum_{\alpha=1}^{N}\{\exp[k_2\langle\bar{E}_\alpha\rangle^2] - 1\} \tag{2}$$

With

$$\bar{E}_{\alpha} \stackrel{\text{def}}{=} \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)\left(\bar{I}_{4(a\alpha)} - 1\right) \tag{3}$$

For completely randomly oriented fibers, the following applies[13], where C is the Right Cauchy-Green tensor and *n*0 is the unit vector of fibers

$$\bar{I}_1 = \mathrm{tr}(\mathrm{C}) = 3 \tag{9}$$

And

$$\bar{I}_{4(a\alpha)} = n0.\mathrm{C}.n0 = 1 \tag{10}$$

Thus,

$$\bar{E}_{\alpha} = 0 \tag{11}$$

Which leads to

$$U = C_{10}(\bar{I} - 3) + \frac{1}{D}\left[\frac{[J^{el}]^2 - 1}{2} - ln J^{\mathrm{e}} l\right] + \frac{k_1}{2k_2}(0) \tag{12}$$

Or

$$U = U_{iso}(C) \tag{13}$$

In practical implementation of the model, the virtual fibers were provided the same stiffness as the ground matrix to 1) complete the material property inputs and 2) prevent simulation failure due to inequality occurring in the nodes during calculation.

Model Geometry Details

The different model files have been included and can be found by contacting PI at darryl.dickerson@fiu.edu. The resources include only the models that have distinct geometric differences, be it through overall geometry or geometry of inter-region sectioning. The distinct sets that are represented by the models are: [VS_0], [VS_1], [VS_2], [VS_3], [VS_4, TS1_13], [VS_5], [VS_6, TS1_21], [TS1_0-TS1_7,TS1_11,TS1_12], [TS1_8], [TS1_9], [TS1_10], [TS1_14-TS1_19,TS1_23-TS1_25,VS_7], [TS1_20], [TS1_21], [TS1_22], [TS2_0 – TS2_8, TS3_0-TS3_8], [TS2_9,TS3_9], [TS2_10,TS3_10], [TS2_11,TS3_11], [TS2_12 – TS2_19, TS2_23, TS3_12-TS3_19,TS3_23], [TS2_20, TS3_20], [TS2_21,TS3_21], [TS2_22,TS3_22]. The models within each set do not differ through model geometry, but through changes in material properties or material orientation.

Each model was meshed using adaptive meshing to generate TRI3 and QUAD4 elements, with average mesh size of 10 units. The generated mesh were manually checked for unstable nodes and the particular region was modified manually (via node repositioning, as well conversion of adjacent TRI3 elements to unified QUAD4 element or vice-versa). Number of elements ranged from 460 to 730 in sets TS2 and TS3, while it ranged from 1400 to 10300 elements in set TS1. The large variance arose due to geometric complexities introduced by the sinusoidal interface geometries in all sets. Furthermore, TS1 had distinct macroscale geometry differences between samples.

Model Material property gradient

The gradient of stiffness across the UFC and CFC regions were implemented through usage of temperature dependent material properties and temperature fields across the regions. The temperature field equations were set up so that the temperature increased linearly across the length from 0°C at the Lig/UFC interface to 5°C at the tidemark to 10°C at the cement line, for each height. Since these variations are dependent on the geometry of the interfaces UFC and CFC were bounded by, 4 variations of the gradient equations were used for each region. The variations are as follows:

- UFC (Xl= x-coordinate of Tidemark in straight line, Xh= x-coordinate of Tidemark in straight line, A = Amplitude of sinusoid)
    - Lig/UFC interface straight, Tidemark straight
      ((5/((Xh-Xl)-Y*0))*(X -(Xl+(Y*0))))
    - Lig/UFC interface sinusoidal, Tidemark straight
      (5 / (Xh - (Xl-(2 * sin ((A/ (2* pi)) * (- Y)))))) * (X - (Xl- (2 * sin ((A / (2* pi)) * (- Y)))))
    - Lig/UFC interface straight, Tidemark sinusoidal
      (5 / ((Xh-(2 * sin ((A / (2* pi)) * (- Y))))- Xl) * (X - Xl)
    - Lig/UFC interface sinusoidal, Tidemark sinusoidal
      (5 / ((Xh-(2 * sin ((A / (2* pi)) * (- Y))))- (Xl-(2 * sin ((A/ (2* pi)) * (- Y)))))) * (X - (Xl- (2 * sin ((A / (2* pi)) * (- Y)))))
- CFC (Xl= x-coordinate of Tidemark in straight line, Xh= x-coordinate of Cement line in straight line, A = Amplitude of sinsoid) [Provided as input in the software]

- Tidemark straight, Cement line Straight:

  5+((5/((Xh-Xl)-Y*0))*(X -(Xl+(Y*0))))

- Tidemark sinusoidal, Cement line Straight

  5 + (5 / (Xh - (Xl-(2 * sin ((A/ (2* pi)) * (- Y)))))) * (X - (Xl-(2 * sin ((A / (2* pi)) * (- Y)))))

- Tidemark straight, Cement line sinusoidal

  5 + (5 / ((Xh-(2 * sin ((A / (2* pi)) * (- Y))))- Xl) * (X - Xl)

- Tidemark sinusoidal, Cement line sinusoidal

  5 + (5 / ((Xh-(2 * sin ((A / (2* pi)) * (- Y))))- (Xl-(2 * sin ((A/ (2* pi)) * (- Y)))))) * (X - (Xl- (2 * sin ((A / (2* pi)) * (- Y)))))

Model Step, Constraints and Load Settings

All models were given a static general step following the initial step, which had nlgeom On, Direct equation solver with solver default matrix storage, using Full Newton solution technique, at 1sec time period, with automatic incrementation with an initial increment of 0.1 seconds. In this step, the following outputs were requested: E,MISESONLY,NT,PEEQ,S,TEMP,U,UR,UT.

In the initial step, the temperature fields for UFC and CFC were set as described before, and the outer edge of the Ligamentous region were Encastred (No displacement and no rotation). To simulate uniform loading along the edge, as well as provide directional change with minimal difference, general surface traction that increased linearly to 50N/unit length of undeformed surface was applied to outer edge of the SB region, without following rotation. The only difference between the 3 load cases was the direction vectors, being [1,0,0], [0,1,0], and [(√2)/2, (√2)/2, 0]. These directions are meant to represent purely tensile loading, purely shear loading and a combination of the two (Complex loading).

Model Verification

All of the Models under each loading case were simulated with at least 3 mesh size model for verification studies. The average stress due to loading at tidemark was used to assess verification. The verification graphs can be accessed by contacting PI at darryl.dickerson@fiu.edu or the first author mzahi002@fiu.edu .

Names of files

The file name prefixes for the testing models were based on a previous naming scheme. **In this scheme: PY = VS, CY= TS1, MY = TS2 and MMY = TS3.** The file names were not changed to maintain simulation integrity

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Parameter Justification (Next Page)

Parameter Name	Justification
Overall Geometry	Based on dimensions recorded transverse section at mid depth of a bovine medial anterior meniscal enthesis [1], and an equivalent cuboid
Fiber orientation Ligamentous region	Based on fiber orientation in Rabbit Achilles tendon [2] and reasonable deviation
Fiber disorganization in ligamentous region	Based on disorganizations noted in Achilles tendon collagen compared with reference graph[2,3]
Fiber orientation UFC region	Based on mean fiber orientation in Human Medial Anterior Meniscal enthesis [14] and reasonable deviation
Fiber disorganization in UFC region	Based on disorganizations noted in Human Medial Anterior Meniscal enthesis compared with reference graph[3,14]
Fiber orientation CFC region	Based on mean fiber orientation in Human Medial Anterior Meniscal enthesis [14] and reasonable deviation
Fiber disorganization in CFC region	Based on disorganizations noted in in Human Medial Anterior Meniscal enthesis compared with reference graph [3,14]
Intra-fiber dispersion ligamentous region	Based on range observed in interstitial aorta walls [4]
Intra-fiber dispersion UFC region	Based on range observed in interstitial aorta walls [4]
Geometry of Cement line	Based on experimental observation in Meniscal enthesis [14] and adjusted for Abaqus modelling
Geometry of Tidemark	Based on experimental observation in Meniscal enthesis[14] and adjusted for Abaqus modelling
Geometry of Lig/UFC interface	Original assumption of Lig/UFC interface being similar to the other two interface
Presence of UFC and CFC	Used as validation parameter in set VS and compared with results in literature [15,16]

Ligamentous region Ground matrix stiffness	Ground matrix stiffness assumed close to UFC ground matrix stiffness based on nanoindentation result[14]
Ligamentous region fiber stiffness	Based on ACL fiber stiffness [17]
UFC region Ground matrix stiffness Gradient	Lower value based on nanoindentation results [14]
UFC region Fiber stiffness magnitude	Oscillation of Arbitrary value chosen based on observation that inclusion of Collagen type II in a collagen type I hydrogel reduces stiffness[10] and applying that to previous value based [17]
UFC region fiber disorganization gradient	Based on fiber dispersion noted in [18] and assuming linear dispersion
CFC region Ground matrix stiffness Gradient	Based on nanoindentation results [14] and stiffness change[11] for lower value, and arbitrary for higher value
CFC region Fiber rate of crystallization	Lower value based on tidemark result [14] and linearity of stiffness and mineral content across length[11]. Higher value theoretical for variation purpose
SB region stiffness magnitude	Based on experimental data [12] and arbitrary value to match calcification of CFC

Bibliography

[1] D.F. Villegas, T.A. Hansen, D.F. Liu, T.L. Haut Donahue, A quantitative study of the microstructure and biochemistry of the medial meniscal horn attachments, Ann. Biomed. Eng. 36 (2008) 123–131. https://doi.org/10.1007/s10439-007-9403-x.

[2] J. Du, A.J.-T. Chiang, C.B. Chung, S. Statum, R. Znamirowski, A. Takahashi, G.M. Bydder, Orientational analysis of the Achilles tendon and enthesis using an ultrashort echo time spectroscopic imaging sequence, Magn. Reson. Imaging 28 (2010) 178–184. https://doi.org/10.1016/j.mri.2009.06.002.

[3] T.C. Gasser, R.W. Ogden, G.A. Holzapfel, Hyperelastic modelling of arterial layers with distributed collagen fibre orientations, J. R. Soc. Interface 3 (2006) 15–35. https://doi.org/10.1098/rsif.2005.0073.

[4] U. Huh, C.-W. Lee, J.-H. You, C.-H. Song, C.-S. Lee, D.-M. Ryu, Determination of the Material Parameters in the Holzapfel-Gasser-Ogden Constitutive Model for Simulation of Age-Dependent Material Nonlinear Behavior for Aortic Wall Tissue under Uniaxial Tension, Appl. Sci. 9 (2019) 2851. https://doi.org/10.3390/app9142851.

[5] A.C. Abraham, T.L. Haut Donahue, From meniscus to bone: A quantitative evaluation of structure and function of the human meniscal attachments, Acta Biomater. 9 (2013) 6322–6329. https://doi.org/10.1016/j.actbio.2013.01.031.

[6] N. Chandrashekar, H. Mansouri, J. Slauterbeck, J. Hashemi, Sex-based differences in the tensile properties of the human anterior cruciate ligament, J. Biomech. 39 (2006) 2943–2950. https://doi.org/10.1016/j.jbiomech.2005.10.031.

[7] A. Rufai, J.R. Ralphs, M. Benjamin, Ultrastructure of fibrocartilages at the insertion of the rat Achilles tendon., J. Anat. 189 (1996) 185–191.

[8] M. Benjamin, J.R. Ralphs, Fibrocartilage in tendons and ligaments - An adaptation to compressive load, J. Anat. 193 (1998) 481–494. https://doi.org/10.1017/S0021878298004300.

[9] J. Gao, Immunolocalization of types I, II, and X collagen in the tibial insertion sites of the medial meniscus, Knee Surg. Sports Traumatol. Arthrosc. 8 (2000) 61–65. https://doi.org/10.1007/s001670050013.

[10] N. Vázquez-Portalatín, C.E. Kilmer, A. Panitch, J.C. Liu, Characterization of Collagen Type I and II Blended Hydrogels for Articular Cartilage Tissue Engineering, Biomacromolecules 17 (2016) 3145–3152. https://doi.org/10.1021/acs.biomac.6b00684.

[11] D. Qu, S.D. Subramony, A.L. Boskey, N. Pleshko, S.B. Doty, H.H. Lu, Compositional mapping of the mature anterior cruciate ligament-to-bone insertion, J. Orthop. Res. 35 (2017) 2513–2523. https://doi.org/10.1002/jor.23539.

[12] P.L. Mente, J.L. Lewis, Elastic modulus of calcified cartilage is an order of magnitude less than that of subchondral bone, J. Orthop. Res. 12 (1994) 637–647. https://doi.org/10.1002/jor.1100120506.

[13] G.A. Holzapfel, T.C. Gasser, R.W. Ogden, A New Constitutive Framework for Arterial Wall Mechanics and a Comparative Study of Material Models, J. Elast. Phys. Sci. Solids 61 (2000) 1–48. https://doi.org/10.1023/A:1010835316564.
[14] A.C. Abraham, T.L. Haut Donahue, From meniscus to bone: A quantitative evaluation of structure and function of the human meniscal attachments, Acta Biomater. 9 (2013) 6322–6329. https://doi.org/10.1016/j.actbio.2013.01.031.
[15] A.J. Boys, M.C. McCorry, S. Rodeo, L.J. Bonassar, L.A. Estroff, Next generation tissue engineering of orthopedic soft tissue-to-bone interfaces, MRS Commun. 7 (2017) 289–308. https://doi.org/10.1557/mrc.2017.91.
[16] M.C. McCorry, M.M. Mansfield, X. Sha, D.J. Coppola, J.W. Lee, L.J. Bonassar, A model system for developing a tissue engineered meniscal enthesis, Acta Biomater. 56 (2017) 110–117. https://doi.org/10.1016/j.actbio.2016.10.040.
[17] N. Chandrashekar, H. Mansouri, J. Slauterbeck, J. Hashemi, Sex-based differences in the tensile properties of the human anterior cruciate ligament, J. Biomech. 39 (2006) 2943–2950. https://doi.org/10.1016/j.jbiomech.2005.10.031.
[18] A.J. Boys, J.A.M.R. Kunitake, C.R. Henak, I. Cohen, L.A. Estroff, L.J. Bonassar, Understanding the Stiff-to-Compliant Transition of the Meniscal Attachments by Spatial Correlation of Composition, Structure, and Mechanics, ACS Appl. Mater. Interfaces 11 (2019) 26559–26570. https://doi.org/10.1021/acsami.9b03595.