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DETAILED FINITE ELEMENT MODELING OF FIBER-REINFORCED TISSUES

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INTRODUCTION

Fiber-reinforced tissues, including tendons, ligaments, and the annulus fibrosus (AF) of the intervertebral disc, sustain large tensile stresses during physiological loading. These materials consist of collagen fibers embedded within a proteoglycan rich matrix, and previous constitutive models have shown that fiber-matrix interactions may play an important role in stress distribution; however, these properties are difficult to measure experimentally [1, 2].

Finite element modeling (FEM) is widely used as a complementary approach to experiments. FEMs provide detailed information of stress strain distributions that are difficult to measure directly, including stress distributions and failure behavior throughout complex tissues [3, 4]. For fiber-reinforced biological tissues, most FEM methods describe the fibers and matrix components as either fibers fixed within a matrix [5, 6] or a homogenized volume [7, 8]. However, these two methods cannot provide information about the fiber-matrix interactions, which may be crucial for understanding failure mechanics [9]. Alternatively, a more intuitive and biomechanically correct method describes collagen fibers as a separate material from the extrafibrillar matrix. Separation of fibers and matrix materials has been rarely used, likely due to difficulties of fiber bundle modeling and computational time.

The objective of this study was to compare the mechanical behavior of AF tissue under uniaxial tension using the three FEM methods described above. We based the model architecture on the AF due to the complex fiber architecture, where collagen fibers are oriented at ±45-65º to the vertical axis of the spine (Fig. 1). However, the methods applied here are easily applicable to understanding fiber-matrix interactions of other fiber-reinforced tissues, such as tendons and ligaments.

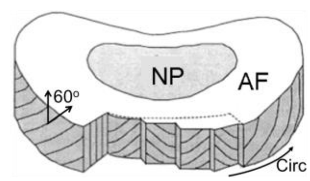

Fig. 1: Schematic of disc.

METHODS

Single lamella models were developed based on AF anatomy reported in Marchand *et al*. [10]. Specifically, each lamella is 0.2 mm thick with fibers described as cylinders through the length of the tissue (radius = 0.059 mm & 0.22 mm between fibers). Three FEM descriptions were assessed, including (1) embedded smeared (EMB), (2) homogenized hyper-elastic (HOM), and (3) fiber-matrix separated (SEP).

The solid matrix for all three models was described using a Mooney-Rivlin material description and all fibers were described as tension-only materials (Table 1). In the EMB model, the fibers were described as being an embedded rebar layer (material coefficients: E = 500 MPa, ν = 0.3, which was first reported in [5] and latter widely used in [6, 11, 12]). In the HOM and SEP models, an exponential-linear description was employed for the fibers, where the stress-stretch response increases exponentially at lower strains, then increases linearly past a defined strain threshold. Single lamellae tensile testing data in [13] was used to determine material coefficients for HOM and SEP models.

The EMB model was developed and solved by Abaqus (6.11) using model parameters reported in the literature for the AF [5]. The HOM model was developed using Preview. The SEP model was developed in SolidWorks (2015) and meshed by Abaqus and Ansa (v16.0.0), and then imported into Preview. HOM and SEP models were evaluated using the FEBio solver.

	EMB(Matrix)	HOM	SEP(Matrix)	SEP(Fiber)
C_1 (MPa)	0.7	0.5	0.75	0.75
C_2 (MPa)	0.2	0.1	0.1	0.1
K (MPa)	N/A	50	50	50
D	~0	N/A		
C_3 (MPa)		0.05		0.21
C_4	N/A	78	N/A	98
C_5 (MPa)		70		380
λ		1.017		1.025

Table 1: Material parameters for FEM models.

Once material coefficients were determined, uniaxial tensile simulation results of double-layer models with fibers in ±65° orientation were compared to multi-layered uniaxial tensile data reported by Ebara *et al.* [14]. HOM and EMB double-layer models had ~20,000 elements, while SEP has ~2-3 x 10^6 elements. Finally, the SEP model was used to evaluate the effect of fiber orientation on tissue-level mechanics. Fiber bundles were arranged at ±45º, ±50º, ±55º, or ±60º, representing the change in fiber orientation from the inner to the outer AF.

RESULTS

Material properties for the HOM and SEP models were able to closely match single-lamella data (Fig. 2A) [13], while the linear description of fibers in the EMB model resulted in a poor fit to the data (Fig. 2 – green line). Model validation of the HOM and SEP models using a multi-layered description with alternating fiber orientation showed a close fit to experimental data (Fig. 2B; R^2 = 0.64 and 0.99, respectively) [14], and EMB almost had a linear behavior.

Once validated, the SEP model was used to describe uniaxial tensile with fiber orientation varying from ±45º to ±65º. A decrease in fiber orientation reduced tissue modulus by approximately 64% (Fig. 3A – slope of purple line *versus* blue line). Tensile stress resulted in fiber re-orientation along the direction of loading (Fig. 3B), which matched well with experimental data of fiber re-orientation [15].

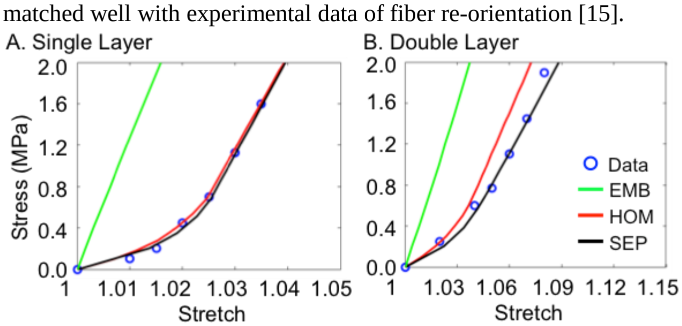

Fig. 2: (A) Stress-strain under 2 MPa uniaxial tensile for (A) single-lamella models and (B) double-lamella models (fiber orientation = ±65°).

DISCUSSION

In this study, we developed a separated fiber-matrix model description (SEP) and compared it to more commonly used FEM for fiber-reinforced tissues (EMB and HOM). Describing collagen fibers as a separate material from the extrafibrillar matrix is more consistent with the disc anatomy, but is rarely used due to computational requirements (run time: 2 minutes for HOM, 30 minutes for SEP). Recent work has demonstrated the importance of fiber-matrix interactions under physiological loading conditions [2, 16], and it is likely that these interactions play an important role in understanding the failure mechanics. SEP model demonstrated inhomogeneous stress distribution between the matrix and fibers, due to the large differences in material properties (Fig. 4)

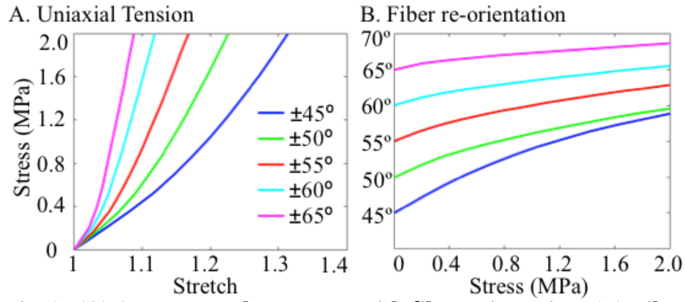

Fig. 3: (A) Stress-stretch response with fiber orientation. (B) Fiber re-orientation under tension.

The healthy AF is noted by transition of collagen fiber orientation and collagen composition (type I to II) from the outer AF to the inner AF. We used the SEP model to evaluate the effect of collagen fiber orientation while maintaining fiber strength. As expected, decreasing the fiber's initial orientation from ±65º (i.e. more aligned with the direction of loading) to ±45º resulted in a decrease in tissue stiffness. The decrease in Young's modulus from 21.8 MPa to 7.5 MPa in the inner AF follows a similar trend to experimental values (17.22 ± 7.69 MPa and 2.65 ± 1.03 MPa, respectively) [17]. However, the predicted Young's modulus of the inner annulus was 3X greater than reported values, which is likely due to the differences in collagen type.

Most embedded model descriptions for fiber-reinforced biological tissues use the linear elastic model description for collagen fibers (i.e. EMB model), with disc joint nonlinearity in compression provided by the disc geometry with fibers of various orientations [5]. However, both our single-lamella and double-layer embedded model showed an approximately linear behavior under a physiological level of strain (Fig. 2). This implies that geometric nonlinearity is not significant enough to cause stress-stretch nonlinearity in tension.

In conclusion, the findings of this study demonstrate that the SEP model has advantages over the EMB and HOM models and will be valuable for studying tissue failure. Future work will incorporate tissue-swelling behavior of the extrafibrillar matrix. Few FEMs of disc tissue include tissue swelling; however, it is likely that matrix swelling will allow for more homogenized stress distributions during loading [18].

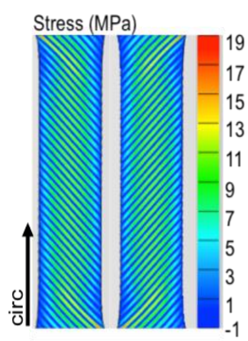

Fig. 4: Circumferential stress in multi-lamellae AF (±65°, 2 MPa applied stress). Higher stresses observed in fibers (blue).

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