Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

An optimized transversely isotropic, hyper-poro-viscoelastic finite element model of the meniscus to evaluate mechanical degradation following traumatic loading

Benjamin B. Wheatley [a], Kristine M. Fischenich [b], Keith D. Button [c], Roger C. Haut [c,d], Tammy L. Haut Donahue [a,b,*]

[a] Department of Mechanical Engineering, Colorado State University, Fort Collins, CO, USA
[b] School of Biomedical Engineering, Colorado State University, Fort Collins, CO, USA
[c] Orthopaedic Biomechanics Laboratories, College of Osteopathic Medicine, Michigan State University, East Lansing, MI, USA
[d] Department of Radiology, Michigan State University, East Lansing, MI, USA

ARTICLE INFO

Article history:
Accepted 15 February 2015

Keywords:
Meniscus
Knee
Inverse
Nonlinear
Osteoarthritis

ABSTRACT

Inverse finite element (FE) analysis is an effective method to predict material behavior, evaluate mechanical properties, and study differences in biological tissue function. The meniscus plays a key role in load distribution within the knee joint and meniscal degradation is commonly associated with the onset of osteoarthritis. In the current study, a novel transversely isotropic hyper-poro-viscoelastic constitutive formulation was incorporated in a FE model to evaluate changes in meniscal material properties following tibiofemoral joint impact. A non-linear optimization scheme was used to fit the model output to indentation relaxation experimental data. This study is the first to investigate rate of relaxation in healthy versus impacted menisci. Stiffness was found to be decreased ($p=0.003$), while the rate of tissue relaxation increased ($p=0.010$) at twelve weeks post impact. Total amount of relaxation, however, did not change in the impacted tissue ($p=0.513$).

1. Introduction

Menisci are wedge shaped semilunar biphasic tissues composed of approximately 70% fluid and 30% organic matter (Mow, 1993). Material testing can be performed on the meniscus to determine a number of properties including the compressive, tensile, and shear modulus, as well as Poisson's ratio and tissue permeability (Joshi et al., 1995; LeRoux and Setton, 2002; Martin Seitz et al., 2013; Moyer et al., 2012; Proctor et al., 1989). These material properties are governed by the arrangement and quantity of the various organic tissue components, including collagen, proteoglycans, elastin, and other glycoproteins (Mow, 1993). The matrix is predominantly type I collagen fibrils oriented in the circumferential direction that results in high tissue anisotropy (Neogi and Zhang, 2013; Proctor et al., 1989). Compressive loading from the femoral condyles is converted to tensile hoop stresses (Upton et al., 2006), which is supported by the circumferential collagen fibers. Proteoglycans help retain interstitial fluid during loading, which is important in load support and load distribution (MacConaill, 1950). However, these tensile and compressive properties are not fully independent (Sanchez-Adams et al., 2011), and characterizing the relationship between tissue structure and mechanical behavior is inherently important for understanding tissue function.

Identifying mechanical changes in meniscal tissue is of particular interest with respect to osteoarthritis (OA) research. OA is a degenerative disease that affects all tissues within the joint, and is one of the leading causes of adult joint pain and disability (Neogi and Zhang, 2013). Two critical components of OA research are developing models that mimic degradation in humans and characterizing changes to the tissue as a result of OA. Experimental animal models have been used extensively to simulate osteoarthritic changes within the knee, but very few have investigated meniscal changes (Adams et al., 1983; Fischenich et al., 2014a,b; Hellio Le Graverand et al., 2001). The majority of these studies have used a traditional anterior cruciate ligament transection (ACLT) model. ACLT models have been shown to induce osteoarthritic type changes that appear to differ when compared to OA-type changes induced by traumatic loading (Killian et al., 2010). A traumatic loading model may more directly recapitulate the common jump landing injury (Levine et al., 2013; Meyer and Haut, 2005; Yeow et al., 2009), and joint injury has been shown to increase the risk of OA development (Blagojevic et al., 2010).

Directly evaluating mechanical properties from experimental data provides useful insight into tissue behavior, but often yields limited results such as elastic modulus (Fischenich et al., 2014a,b;

* Correspondence to: Department of Mechanical Engineering, Colorado State University, 1374 Campus Delivery, Fort Collins, CO 80523, USA. Tel.: +1 970 491 1319.
E-mail address: Tammy.Donahue@ColoState.edu (T.L. Haut Donahue).

http://dx.doi.org/10.1016/j.jbiomech.2015.02.028
0021-9290/

Proctor et al., 1989) and testing procedures can be destructive in nature (Mansour and Mow, 1976). Inverse finite element (FE) approaches are an effective method to evaluate a range of material properties through parameter optimization (Korhonen et al., 2002; Lei and Szeri, 2007; LeRoux and Setton, 2002) and can be utilized with non-destructive experimental analyses. Previous FE modeling efforts of the meniscus have employed a combination of a biphasic approach (Haemer et al., 2012; Proctor et al., 1989; Spilker et al., 1992), three dimensional geometry (Haut Donahue et al., 2002; LeRoux and Setton, 2002; Peña et al., 2006, 2005), and non-linear behavior (Haemer et al., 2012), in addition to known transverse isotropy (LeRoux and Setton, 2002; Proctor et al., 1989). Although these models are successful in studying the effect of particular changes within the meniscus, they often fail to combine all aspects of tissue behavior. Viscoelasticity, for example, is seldom included in meniscal models, despite the studies that show the meniscus exhibits viscoelastic behavior in shear (Anderson et al., 1991; Zhu et al., 1994).

The goal of the current study was to develop and employ a finite element model of the meniscus which would utilize a novel constitutive model to predict tissue behavior under non-destructive indentation relaxation testing. The development of this model would be beneficial in multiple ways: (1) it would show the effectiveness of characterizing material behavior of the meniscus, (2) it could be used to evaluate potential changes in the material properties of healthy and traumatized menisci, and (3) it could provide a tool to further study how individual constituents affect the structural response of the menisci.

2. Methods

2.1. Mechanical testing

Six skeletally mature Giant Flemish rabbits (5.7 ± 0.2 kg) were used in this study. Following anesthesia (2% isoflurane and oxygen), a 1.75 kg mass was dropped from 70 cm in height onto the distal femur with the right tibiofemoral joint at 90° flexion, causing ACL rupture and acute damage to surrounding tissues (Fischenich et al., 2014a,b). Left limbs remained non-impacted as paired controls. Animals were euthanized twelve weeks post injury. This procedure was approved by Michigan State University and Colorado State University All-University Committees on Animal Use and Care. A licensed veterinary technician monitored the rabbits and administered buprenorphine (0.3 mL/KG-BW) for pain every 8 h for 72 h post impact.

Tissue harvest was performed immediately following euthanasia, and all samples were refrigerated (1.7° to 3.3 °C) before testing within 18 h of sacrifice. Each meniscus (lateral and medial for each joint) was sectioned into three regions: anterior, central, and posterior (Fig. 1). A 1.59 mm diameter sphere indenter tip was displaced 0.25 mm normal to the tissue surface in ~1 s and then held for 900 s of relaxation. All testing was completed with specimens submerged in 0.9% phosphate buffered saline (PBS) using a servo-hydraulic testing system (MTS Corp, Eden Prairie, MN).

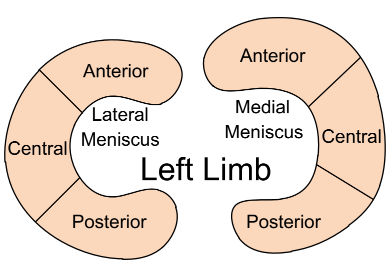

Fig. 1. Schematic of how experimental samples were sectioned from menisci. Each limb yielded six specimens, three lateral (anterior, central, and posterior) and three medial (similarly).

2.2. Constitutive model

The constitutive model utilized in this study combined a strain energy density function, the interaction between the solid portion (matrix and fibers) and the saturating fluid through a poroelastic approach, and the inherent viscoelasticity of the solid portion of the tissue. Thus, the meniscus was modeled as hyper-poro-viscoelastic. As the meniscus has been shown to be anisotropic (LeRoux and Setton, 2002; Proctor et al., 1989), non-linear (Abraham et al., 2011; Danso et al., 2014), and highly compressible (Sweigart et al., 2004), a hyperleasitc, transversely isotropic strain energy density (SED) function was developed and employed (Holzapfel, 2000) (Eq. (1)). In this expression $\bar{\boldsymbol{I}}_1$ and $\bar{\boldsymbol{I}}_2$ are the first and second modified invariants of the left Cauchy-Green deformation tensor $\mathbf{B}$ which characterize tension/compression and shear, respectively. J is the Jacobian or volume change ratio, and $\mathbf{I}_4$ is a pseudo-invariant characterizing the transverse isotropy or circumferential fiber stiffness. This function incorporated both the volume-preserving ($\bar{\boldsymbol{I}}_1$, $\bar{\boldsymbol{I}}_2$, $\boldsymbol{I}_4$) and volume-changing (J) responses of the material. Thus, three parameters (a_2, b_2, d_2) were used to characterize the isochoric response of the solid, while a fourth parameter (c_2) was used to characterize compressibility.

$$\Psi = a_2(\bar{\boldsymbol{I}}_1 - 3)^2 + b_2(\bar{\boldsymbol{I}}_2 - 3)^2 + c_2(J-1)^2 + d_2(\boldsymbol{I}_4 - 1)^2 \tag{1}$$

A three-term Prony series viscoelastic expansion was applied to the solid phase of the tissue (Eq. (2)), where $g_R(t)$ is the time dependent shear relaxation modulus, g_n are fractions of this shear modulus, and τ_n are time constants (Dassault Systèmes, 2012). The application of a three term series is common in studies incorporating the Prony method, as it provides a smooth curve shape without a large number of parameters (Kalyanam et al., 2009; Seifzadeh et al., 2012). The Prony series was applied to the constitutive relation at each increment (Eq. (3), where τ and γ are shear stress and strain, respectively). Additionally, for the material to retain positive stiffness, all g_n terms must sum to less than one.

$$g_R(t) = 1 - \sum_{n=1}^{3} g_n \left[1 - \exp\left(-\frac{t}{\tau_n} \right) \right] \tag{2}$$

$$\frac{\partial \tau}{\partial \gamma}(t, \tau) = \frac{\partial \tau}{\partial \gamma}(\gamma)\, g_R(t) \tag{3}$$

The inclusion of poroelasticity decoupled the total stress of a porous material (Eq. (4)) into the solid phase (where $\boldsymbol{\sigma}^{\text{solid}}$ is determined from Eqs. (1) and (3)) and fluid phase (where p is the pressure of the fluid and $\boldsymbol{I}$ is the identity matrix). In the case of meniscus, these pores were assumed to be fully saturated with interstitial fluid. The pressure of the saturating fluid is a function of the permeability, as shown in Darcy's Law (Eq. (5)), where k is the permeability, ∇p is the pore pressure difference within regions the tissue (and is the same pressure as Eq. (4)), μ is the fluid viscosity, and q is the fluid flow per unit area through the tissue. The permeability controls how quickly the fluid pressure reaches equilibrium following loading, which manifests itself in the overall relaxation rate of the tissue.

$$\boldsymbol{\sigma}^{\text{total}} = \boldsymbol{\sigma}^{\text{solid}} + p\boldsymbol{I} \tag{4}$$

$$q = \frac{-k}{\mu} \nabla p \tag{5}$$

However, permeability is also dependent on the voids ratio of the material (Eq. (6)) where V_V is the volume of the voids space (in this case full saturation was assumed) and V_S is the volume of only the solid. Haemer et al., 2012 utilized a permeability function (Eq. (7)) where the voids ratio is determined from the strain state of the material. In Eq. (7), $k(e)$ is permeability as a function of voids ratio, k_0 is the permeability of the undeformed material, e_0 is the undeformed voids ratio, and η and M are unit-less material parameters (Haemer et al., 2012).

$$e = \frac{V_V}{V_S} \tag{6}$$

Table 1
Material parameters, including both fixed vales and those that were varied.

Model	SED function				Prony series		Darcy's law & permeability				
Parameter	a_2	b_2	c_2	d_2	g_n $(n=3)$	τ_n $(n=3)$	k_0	μ	e_0	κ	M
Value	Varied				Varied		Varied	981	3	0.0848	4.638
Units	Pa				Unit-less		m^4/N s	Pa s	Unit-less		

Please cite this article as: Wheatley, B.B., et al., An optimized transversely isotropic, hyper-poro-viscoelastic finite element model of the meniscus to evaluate mechanical degradation.... Journal of Biomechanics (2015), http://dx.doi.org/10.1016/j.jbiomech.2015.02.028