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The low permeability of healthy meniscus and labrum limit articular cartilage consolidation and maintain fluid load support in the knee and hip

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ABSTRACT

The knee meniscus and hip labrum appear to be important for joint health, but the mechanisms by which these structures perform their functions are not fully understood. The fluid phase of articular cartilage provides compressive stiffness and aids in maintaining a low friction articulation. Healthy fibrocartilage, the tissue of meniscus and labrum, has a lower fluid permeability than articular cartilage. In this study we hypothesized that an important function of the knee meniscus and the hip labrum is to augment fluid retention in the articular cartilage of a mechanically loaded joint. Axisymmetric hyperporoelastic finite element models were analyzed for an idealized knee and an idealized hip. The results indicate that the meniscus maintained fluid pressure and inhibited fluid exudation in knee articular cartilage. Similar, but smaller, effects were seen with the labrum in the hip. Increasing the fibrocartilage permeability relative to that of articular cartilage gave a consolidation rate and loss of fluid load support comparable to that predicted by meniscectomy or labrectomy. The reduced articular cartilage fluid pressure that was calculated for the joint periphery is consistent with patterns of endochondral ossification and osteophyte formation in knee and hip osteoarthritis. High articular central strains and loss of fluid load support after meniscectomy could lead to fibrillation. An intact low-permeability fibrocartilage is important for limiting fluid exudation from articular cartilage in the hip and knee. This may be an important aspect of the role of fibrocartilage in protecting these joints from osteoarthritis.

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1. Introduction

The anatomy of fibrocartilaginous structures in joints has been known for many years, but the mechanisms by which these structures perform their functions are not fully understood.

The load-distributing function of the knee meniscus has been described in experimental (Ahmed and Burke, 1983; Anderson et al., 1993; Ihn et al., 1993; Newman et al., 1989) and numerical studies (Adeeb et al., 2004; Donahue et al., 2002; Donzelli et al., 1999; Pena et al., 2006; Wilson et al., 2003). The meniscus also has a stabilizing function in the knee joint (Bendjaballah et al., 1998) and is important in guiding the relative position of the femur and tibia (Netravali et al., 2010). Osteoarthritis after injury or removal of the meniscus occurs in humans and animals (Cox et al., 1975; Englund et al., 2001; Fauno and Nielsen, 1992; Ghosh et al., 1990; Huang et al., 2003; Jackson, 1968; Oakley et al., 2004).

The function of the hip labrum is less well understood. In contrast to the knee, the labrum does not significantly redistribute articular contact stress (Konrath et al., 1998). The labrum, however, has been found to have a sealing function. If this seal is disrupted, hip intraarticular pressure during compressive joint loading decreases (Ferguson et al., 2003) and cartilage consolidation is increased (Ferguson et al., 2003; Ferguson et al., 2000b). The labrum may thus protect the hip from osteoarthritis development, but clinical and animal studies on this have not been as definitive as in the knee (Appleyard et al., 1999; Englund, 2004; Englund et al., 2003, 2001; Oakley et al., 2004; Roos et al., 1998).

The meniscus and labrum each have a permeability that is one-half to one-tenth that of articular cartilage (Fithian et al., 1990). The relatively impermeable nature of these tissues may be important in protecting the joint from developing osteoarthritis. In this study, we hypothesized that (1) the meniscus and labrum seal the articular cartilage from fluid efflux during loading, augmenting fluid pressure in the articular cartilage and reducing articular cartilage consolidation in the knee and hip; (2) changes in patterns of fluid pressure and fluid flow after removal of the meniscus and the labrum are related to observed patterns of osteoarthritis development.

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2. Methods

2.1. Material properties

The articular cartilage was modeled as isotropic hyperporoelastic (Guilak and Mow, 2000; Holmes and Mow, 1990). The material's solid phase was implemented as an isotropic hyperelastic solid. The strain energy density function (Ψ) had the form of Eq. (1) (Holmes and Mow, 1990), where I, II, and III are the principal strain invariants, and β is a nonlinearity exponent. α_0, α_1 and α_2 were derived from the Lamé constants μ and λ, and the non-linearity exponent β using the following three equations: $\lambda=4\alpha_0\alpha_2$, $\mu=2(\alpha_1+\alpha_2)\alpha_0$, and $\beta=\alpha_1+2\alpha_2$.

The material properties for cartilage (Table 1) were adapted from those given by Wu and Herzog (2000). The strain dependent permeability, k, was adapted from that given by Federico et al. (2005) in Eq. (2). In Eq. (2), M is a material constant which has been determined for cartilage to be 4.638 (Holmes, 1986). e is the void ratio and e_0 is the void ratio in the undeformed state.

$$\Psi=\alpha_0\frac{\exp(\alpha_1(\mathrm{I}-3)+\alpha_2(\mathrm{II}-3))}{\mathrm{III}^{\beta}} \tag{1}$$

$$k(e)=k_0\left(\frac{e}{e_0}\right)^{\kappa}\exp\left\{\frac{M}{2}\left[\left(\frac{1+e}{1+e_0}\right)^2-1\right]\right\} \tag{2}$$

The meniscus and labrum were modeled as transversely isotropic, hyperporoelastic with strain dependent permeability. The aggregate modulus in the transverse plane (perpendicular to the circumferential fibers of the meniscus or labrum) was comparable to that of articular cartilage (0.55 MPa) (Ateshian et al., 1997; Athanasiou et al., 1995; Joshi et al., 1995) while the tensile modulus in the direction of fiber orientation was set at 200 MPa, roughly one-fourth the tensile modulus of tendon (Gibbons, 1976). The permeability of the meniscus and the labrum were set at one-sixth of articular cartilage, which is comparable to what has been found experimentally (Ferguson et al., 2001; Fithian et al., 1990; Joshi et al., 1995; Proctor et al., 1989). The form of the strain energy density function is that described by Almeida (1995).

The strain-dependence of meniscus permeability has not yet been characterized. Since the transverse properties of cartilage and meniscus are of the same order, the form of strain dependence was adapted from that for cartilage. Material properties (Table 2) were adapted from analogous linear poroelastic properties given Spilker et al. (1992). Ferguson et al. found properties of the labrum to be comparable to those of the meniscus (Ferguson et al., 2001), thus the same material model was used for both materials.

To understand the sensitivity of the model to chosen material properties, and to investigate the relative importance of permeability and fiber stiffness to meniscal and labral function, additional models of the meniscus and labrum were created where the circumferential modulus was cut to half and to one-quarter of the original value, and the permeability of the meniscus or labrum was increased to a value comparable to that of articular cartilage.

2.2. Model components

Axisymmetric finite element models of idealized knee and hip joints were developed (Fig. 1). The underlying bone was modeled as rigid and impermeable (Adeeb et al., 2004; Ferguson et al., 2000a, 2000b). This assumption is supported by observations that perforations in the subchondral bone are rare (Clark and Huber, 1990). Fluid was free to flow from open surfaces of the articular cartilage, and the contact formulation allowed fluid flow between opposing surfaces of articular cartilage or fibrocartilage. A load, approximating a single-legged stance for each model, was applied to the femoral side of the joint while the tibial or acetabular side remained fixed. The load was ramped over 1 s, and held constant for 1000 s. The loading scheme was devised to produce initial physiologic cartilage loads, followed by a period to characterize consolidation rates

Table 1
Material properties for cartilage.

Model formulation	Hyperelastic Exponential strain energy density Isotropic
Lamé constant analogs at infinitesimal deformation	
μ	0.339 MPa
λ	0.013
Nonlinearity exponent	
β	0.761
Permeability parameters	
k_0	7.60E-003 mm^4/N s
e_0	4
κ	0.0848
M	4.638

Table 2
Material properties for meniscus and labrum.

Model Formulation	Hyperelastic Exponential strain energy density Transversely isotropic Circumferential (θ) fiber orientation
Elastic property analogs at infinitesimal deformation	
E_f (fiber)	200 MPa
E_t (transverse)	0.55 MPa
ν_{ft}	0.05
ν_{tt}	0.05
G_{ft}	0.026 MPa
Nonlinearity parameters	
a_0	0.08
n	0.6
Permeability properties	
k_0	1.26E-003 mm^4/N s
e_0	3
κ	0.0848
M	4.638

The intact knee joint model (Fig. 1A) was symmetric about the horizontal plane, designated as the dotted line "A" in the figure. The articular cartilage had a central thickness of 1.5 mm and the meniscus was congruent to the articular cartilage. The geometry was an idealization of a sheep knee imaged during in vitro studies of cyclically loaded joints (Song et al., 2006). It should be noted that both tibial condyles of the sheep tibia are convex (Song et al., 2006). Modeling contact between the femur, tibia, and meniscus can present difficulties, as portions of the joint transition from cartilage–cartilage contact to cartilage–meniscus contact (Wilson et al., 2003). To accommodate this behavior, in the center of contact, a small strip of cartilage material was extended from the edge of the meniscus to the axial centerline of the joint. This simplification made the contact problem tractable and allowed portions of the articular cartilage to transfer to contacting the meniscus. The knee was loaded on the femoral side to 250 N, representing the bodyweight of a sheep distributed over two femoral condyles.

The meniscectomized knee model is shown in Fig. 1B. The tibia was fixed and the femur was loaded to 250 N. Modeling large deformations in the cartilage of a meniscectomized knee required special consideration of the fluid boundary conditions, as portions of the articular surface transition into contact. The method developed here allowed contact-dependent flow between cartilage layers and between unaligned meshes. A user-defined seepage flow coefficient was specified on conditionally free-flowing surfaces: portions of the joint that came into contact over the course of the analysis became free flowing, via a high seepage flow coefficient. The contact state of each node on the contact slave surface was tracked, and the positions of integration points on either side of the joint were checked for being in contact. When portions of the joint came into contact, the seepage flow coefficient was set to zero. After coming into contact, the default biphasic contact formulation governed contact between bodies, ensuring continuity of stress and fluid flow across the contacting interface (Hibbit and Sorensen, 2004).

Models of the hip are shown in Fig. 1C and D. The hip was idealized as being fully congruent, and the fibrocartilage labrum was modeled as being in continuity with the articular cartilage (Seldes et al., 2001). The thickness of the articular cartilage was 3 mm, and the radius of the femoral head was 26 mm, approximating the dimensions of a human hip and similar to those used in a published idealized model of fluid flow in the hip (Ferguson et al., 2000b). The hip was loaded to 3 kN, corresponding to the joint reaction force in single-legged stance (Konrath et al., 1998).

3. Results

3.1. Patterns of pressure, strain, and fluid flow

Peak fluid pressure after one second of ramp load in the intact knee was over 1 MPa and pressurization occurred throughout the articular cartilage to the edge of the meniscus (Fig. 2A). Fluid pressure in the meniscectomized knee reached 4 MPa under the center of contact and fell to zero at the edge of the contact (Fig. 2B). In the hip, fluid pressure occurred throughout the contacting regions, and the presence of the labrum slightly extended the region of pressurization (Fig. 2C). Peak fluid