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Review

Finite element modeling of soft tissues: Material models, tissue interaction and challenges

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ABSTRACT

Background: Musculoskeletal soft tissues, such as articular cartilage, ligaments, knee meniscus and intervertebral disk, have a complex structure, which provides elasticity and capability to support and distribute the body loads. Soft tissues describe an inhomogeneous and multiphasic structure, and exhibit a nonlinear, time-dependent behavior. Their mechanical response is governed by a substance composed of protein fiber-rich and proteoglycan-rich extracellular matrix and interstitial fluid. Protein fibers (e.g. collagen) give the tissue direction dependent stiffness and strength. To investigate these complex biological systems, the use of mathematical tools is well established, alone or in combination with experimental in vitro and in vivo tests. However, the development of these models poses many challenges due to the complex structure and mechanical response of soft tissues.
Methods: Non-systematic literature review.
Findings: This paper provides a summary of different modeling strategies with associated material properties, contact interactions between articulating tissues, validation and sensitivity of soft tissues with special focus on knee joint soft tissues and intervertebral disk. Furthermore, it reviews and discusses some salient clinical findings of reported finite element simulations.
Interpretation: Model studies extensively contributed to the understanding of functional biomechanics of soft tissues. Models can be effectively used to elucidate clinically relevant questions. However, users should be aware of the complexity of such tissues and of the capabilities and limitations of these approaches to adequately simulate a specific in vivo or in vitro phenomenon.

1. Introduction

The mechanical function of musculoskeletal soft tissues, such as articular cartilage, ligaments, meniscus or intervertebral disk, is in supporting and distributing the loads generated by muscles, body weight and inertial forces and allowing for the controlled motion of the human body. In order to investigate the mechanics of these biological systems, mathematical tools, such as finite element methods act as a complementary approach to in vivo and in vitro studies. It provides valuable insights into the behavior of the structures' function and are helpful for explorative preclinical investigations.

As a matter of fact, soft tissues exhibit mechanical characteristics with a higher complexity than most engineering materials and structures. For modeling purposes it is crucial to select a material law as well as material parameters that well describe the mechanical properties of the soft tissue with regard to the investigation scope.

The mechanical behavior of soft tissues is governed mainly by the major phases of these materials. The solid phase with collagen fibers, proteoglycans, other proteins and cells as well as an interstitial fluid phase composed of water and electrolytes (Mow and Huiskes, 2005). While substances like proteoglycans bind water to form a firm gel and give the tissue its resiliency, the collagen fibers give the tissue its tensile strength. The specific arrangement and hierarchical organization of the fibers have major influences on the mechanical behavior of the tissue (Fung, 1993). It is naturally optimized to fulfill the specific mechanical function of each tissue (Schneck and Bronzino, 2003). Hyaline cartilage for example is highly hydrated and is comprised of fine collagen fibrils oriented isotropically in planes parallel to the articular contact. This specific orientation gives the cartilage its high resistance to compressive loads and provides good lubrication to highly mobile joint surfaces (Bell et al., 2006). Differently, soft tissues made of fibrocartilage, e.g. annulus fibrosus of the intervertebral disk, knee meniscus or temporomandibular joint, are made up of compact collagen fiber bundles oriented in the circumferential direction of the tissue. However, a closer look at the fiber distribution reveals not only circumferentially oriented fibers. For example, the meniscus also

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Table 1
Material formulations for statically responding simulations.

Linear material formulations (Hooke's law)

General formulation (*anisotropic material*)

$\sigma = C \cdot \varepsilon$

with

$$\sigma = \begin{bmatrix} \sigma 11 \\ \sigma 22 \\ \sigma 33 \\ \sigma 23 \\ \sigma 13 \\ \sigma 12 \end{bmatrix}, \varepsilon = \begin{bmatrix} \varepsilon 11 \\ \varepsilon 22 \\ \varepsilon 33 \\ 2\varepsilon_{23} \\ 2\varepsilon_{13} \\ 2\varepsilon_{12} \end{bmatrix}, C = \begin{bmatrix} C_{1111} & C_{1122} & C_{1133} & C_{1123} & C_{1113} & C_{1112} \\ C_{2211} & C_{2222} & C_{2233} & C_{2223} & C_{2213} & C_{2212} \\ C_{3311} & C_{3322} & C_{3333} & C_{3323} & C_{3313} & C_{3312} \\ C_{2311} & C_{2322} & C_{2333} & C_{2323} & C_{2313} & C_{2312} \\ C_{1311} & C_{1322} & C_{1333} & C_{1323} & C_{1313} & C_{1312} \\ C_{1211} & C_{1222} & C_{1233} & C_{1223} & C_{1213} & C_{1212} \end{bmatrix}$$

σ: stres vector; ε: strain vector; C: 4th-order stiffness tensor

For orthotropic, transversal isotropic and isotropic materials:

$$C = \begin{bmatrix} C_{1111} & C_{1122} & C_{1133} & 0 & 0 & 0 \\ C_{2211} & C_{2222} & C_{2233} & 0 & 0 & 0 \\ C_{3311} & C_{3322} & C_{3333} & 0 & 0 & 0 \\ 0 & 0 & 0 & C_{2323} & 0 & 0 \\ 0 & 0 & 0 & 0 & C_{1313} & 0 \\ 0 & 0 & 0 & 0 & 0 & C_{1212} \end{bmatrix}$$

Orthotropic material	Transversal isotropic material	Isotropic material
$C_{2211} = C_{1122}$ $C_{3311} = C_{1133}$ $C_{3322} = C_{2233}$	$C_{2211} = C_{1122}$ $C_{3311} = C_{3322} = C_{1133} = C_{2233}$ $C_{1111} = C_{2222}$ $C_{2323} = C_{1313}$ $C_{1212} = \frac{1}{2}(C_{1111} - C_{1122})$	$C_{1111} = C_{2222} = C_{3333}$ $C_{2323} = C_{1313} = C_{1212}$ $C_{1122} = C_{1133} = C_{2233} = C_{2211} = C_{3311} = C_{3322}$

Nonlinear fiber reinforced material formulations

$\sigma = \sigma_{gs} + \sigma_{fib}$ with $E_{fib} = \begin{cases} E^0_{fib} + E^{\varepsilon}_{fib}\varepsilon_{fib} & for\ \varepsilon_{fib} > 0 \\ 0 & for\ \varepsilon_{fib} < 0 \end{cases}$

σ_{gs}: ground substance stress; σ_{fib}: fibril stress; E_{fib}: fibril tensile strain dependent modulus; ε_{fib}: fibril strain; E^0_{fib}: initial fibril stiffness; E^{ε}_{fib}: fibril stiffness

consists of a few radial tie fibers in the central portion (Petersen and Tillmann, 1998). In the superficial layer the fibers are randomly distributed (Petersen and Tillmann, 1998). Such locally varying collagen fiber arrangement can also be found in other soft tissues and are of major interest in modeling soft tissues since the collagen fibers are quantitatively the major organic component (Mow and Huiskes, 2005). Proteoglycans are the second most abundant organic component (Mow and Huiskes, 2005). With their fixed charge density, they are involved in osmotic swelling (Schneck and Bronzino, 2003), which regulates the pressure and tissue hydration, hence, contributing to the viscoelastic nature of the soft tissue. The fluid phase also plays an important role in terms of viscoelasticity. The pressure of the interstitial fluid due to osmotic imbibition and mechanical loads creates a stress in the solid phase, which contributes to the stiffness and the apparent incompressibility of the soft tissue (Mow and Huiskes, 2005).

Simulations of musculoskeletal structures are usually not limited to one specific soft tissue. Instead, most studies are focused on the interaction between the various anatomical components, such as contact between cartilage layers or ligament wrapping, which introduces a nonlinearity in a numerical model and can be particularly critical in biphasic models or when considering friction.

Moreover, simulations of musculoskeletal tissues targeted to achieve a clinical impact are not usually limited to modeling the complexity of the response of these materials or their interaction. Modeling of pathologies, like continuous degeneration, damage, failure or crack propagation introduces geometric discontinuity in the tissue and constitutes another challenge.

The present non-systematic review discusses different modeling strategies with associated material properties, contact interactions between articulating tissues, validation and sensitivity analyses of soft tissues and provides some salient clinical findings of reported finite element simulations with special focus on the soft tissues of the knee and the intervertebral disk. Nevertheless, as soft tissues exhibit analogous structure and properties in all anatomical regions, modeling approaches and material constitutive laws can be adapted to other soft tissues with little effort.

2. Material laws for soft tissues

2.1. Static response

Static analyses can be used to investigate the behavior of the soft tissue at one certain time point, for example instantaneous or equilibrium response. In this case the time-dependent behavior, e.g. creep or relaxation, of the soft tissue is neglected.

2.1.1. Isotropic, linear elasticity

The simplest way to model the mechanical response of a soft tissue is to use a linear elastic and isotropic formulation (Table 1). In such case it is assumed that the stress–strain curve is linear and not dependent on the direction of load. Due to their simplicity, linear elastic isotropic materials can be used to limit numerical difficulties in challenging contact formulations when the investigation of the response of the tissues in the proximity of the contact is not the main focus, e.g. in kinematic analyses of the knee, as well as for debugging purposes of more complex models.

One of the first finite element simulations of the meniscus used a linear elastic and isotropic material model in a strongly geometrically simplified, axisymmetric model (Sauren et al., 1984). Later studies also included linear elastic isotropic material formulations for e.g. articular cartilage, menisci (Beillas et al., 2001; Pena et al., 2005a; Perie and Hobatho, 1998) or nucleus pulposus of the intervertebral disk (Ueno and Liu, 1987) and considered a more detailed geometry. In a sensitivity study, in which the linear elastic material properties of e.g. cartilaginous and meniscal tissue were varied in a physiological range (Beillas et al., 2007) it was shown that the variation of articular cartilage properties highly influenced the cartilage contact pressure response. In general, higher elasticity and Poisson's ratios of the cartilage tissue caused higher maximal contact pressure (Beillas et al., 2007).