Evaluation of the constitutive models of meniscus and their effect on
the knee joint biomechanics during physiologically relevant loading

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Abstract:

The meniscus is an integral part of the overall health of the knee joint. To evaluate medical procedures, finite element models can predict degenerative changes in the articular cartilage and meniscus. In those models, fibril-reinforced poroelastic (FRPE) material has been implemented to consider a realistic response of the meniscus during static and dynamic loading. However, such advanced FRPE formulation demands a high computational capacity. It limits the application of the FRPE model on a biomechanical assessment of multiple subjects for clinical proposes. Hence, it is relevant to utilize a simpler material representation which can capture these complex characteristics within optimized implementation time to be applied in many subjects. The aim of this study was to investigate more simple constitutive representation of meniscus that produces a similar mechanical response in articular cartilage during walking. Thus, an orthotropic poroelastic (OTP) material was considered to represent the meniscus during numerical analysis of knee joint. The OTP constitutive parameters were optimized based on numerical comparisons with respect to FRPE material. Then, these adjusted OTP meniscus properties were implemented into a 3D computational knee joint model to evaluate the total reaction force, stress, strain, pore pressure and fibril strain on the tibial cartilage. Finally, these results were compared with knee joint model with FRPE meniscus. The biomechanical response exhibited on the articular cartilage for both meniscus models were similar during the gait. In conclusion, OTP meniscus model can mimic the effect produced by the FRPE meniscus model on the tibial articular cartilage, reducing the pre-processing time to generate subject-specific models. Hence, the OTP model can be applied in the finite element analysis of the knee joint in routine clinical practice.
Abbreviations

ACL       Anterior cruciate ligament
BW        Body weight
ECM       Extracellular matrix
FEM       Finite element method
FRPE      Fibril reinforced poroelastic
GAG       Glycosaminoglycan
LCL       Lateral collateral ligament
MCL       Medial collateral ligament
OTP       Orthotropic Poroelastic
PCL       Posterior cruciate ligament
PGs       Proteoglycans
Symbols:

A \quad \text{Cross-sectional area}

C \quad \text{Stiffness matrix}

\sigma^E \quad \text{Effective solid stress}

\sigma_{\text{fluid}} \quad \text{Fluid stress}

\sigma_{\text{solid}} \quad \text{Solid matrix stress}

\phi_{\text{fluid}} \quad \text{Volume fraction for the liquid}

\phi_{\text{solid}} \quad \text{Volume fraction for the solid}

\varepsilon^E \quad \text{Elastic strain tensor}

E_1 \quad \text{Young’s modulus in X-axis (in the direction of 1)}

E_2 \quad \text{Young’s modulus in Y-axis (in the direction of 2)}

E_3 \quad \text{Young’s modulus in Z-axis (in the direction of 3)}

E_f \quad \text{Fibril network modulus}

E^{nf} \quad \text{Young’s modulus of non-fibrillar matrix}

E_p \quad \text{Young’s modulus in plane}

E_z \quad \text{Young’s modulus out of plane}

G_{12} \quad \text{Shear modulus between plane one and plane two}

G_{13} \quad \text{Shear modulus between plane one and plane three}

G_{23} \quad \text{Shear modulus between plane two and plane three}

G_m \quad \text{Shear modulus of the non-fibrillar matrix}

\epsilon_0 \quad \text{Initial void ratio}

k_0 \quad \text{Initial Permeability}

\nu_{12} \quad \text{Poisson’s ratio between plane one and two}

\nu_{23} \quad \text{Poisson’s ratio between plane two and three}

\nu_{13} \quad \text{Poisson’s ratio between plane one and three}

\nu^{nf} \quad \text{Poisson’s ratio of non-fibrillar matrix}
\( \nu_{zp} \)  Poisson’s ratio out of plane

\( \nu_p \)  Poisson’s ratio in plane

\( \nu_m \)  Poisson’s ratio of the non-fibrillar matrix

\( \varepsilon_f \)  Fibril strain

\( \sigma_f \)  Total fibril stress

\( \sigma_{nf} \)  Non-fibrillar matrix stress tensor

\( \Delta p \)  Pressure difference

I  Unit tensor

J  Determinant of deformation tensor

I  Unit tensor

K  Bulk Modulus

Q  Velocity of fluid flow

e  Current void ratio

k  Permeability

p  Fluid pressure

\( \eta \)  Damping coefficient
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# Contents

1. Introduction ......................................................................................................................... 1

2 Background .......................................................................................................................... 3

2.1 Knee Joint .......................................................................................................................... 3

2.2 Articular cartilage ............................................................................................................. 4

2.2.1 Composition ................................................................................................................ 5

2.3 Meniscus ........................................................................................................................... 6

2.3.1 Composition ................................................................................................................ 8

2.4 Biomechanics of meniscus ............................................................................................. 10

2.5 Constitutive modelling of meniscus .............................................................................. 12

2.5.1 Poroelastic theory ...................................................................................................... 12

2.5.2 Isotropic, linear elastic material behavior ............................................................... 14

2.5.3 Orthotropic constitutive model ................................................................................ 15

2.5.4 Fibril reinforced poroelastic constitutive model ....................................................... 17

2.5.5 Non-fibrillar part ...................................................................................................... 17

2.5.6 Fibrillar part ............................................................................................................. 18

3 Aim of the study and hypothesis ....................................................................................... 20

4 Material and methods ......................................................................................................... 21

4.1 Adjustment of OTP material parameters ...................................................................... 22

4.1.1 Boundary conditions of the cubic model ............................................................... 22

4.1.2 Mechanical response comparison between OTP and FRPE models ....................... 25

4.2 Implementation of OTP meniscus into global knee joint model ................................. 26

5 Results ............................................................................................................................... 28

5.1 Comparison between constitutive models in the global knee model ......................... 28

5.1.1 Reaction force on tibial cartilage .......................................................................... 30

5.1.2 Reaction force through the meniscus .................................................................... 32

5.1.3 Maximum principal stress ..................................................................................... 33

5.1.4 Maximum principal strain ..................................................................................... 34

5.1.5 Fibril strain ............................................................................................................. 35

5.1.6 Pore pressure ........................................................................................................ 36

5.1.7 Comparison cartilage-cartilage contact area ......................................................... 37
Chapter 1

Introduction

The meniscus is a multifunctional component in the knee joint that plays a vital role in loading transmission and distribution from the femur to tibia, though it was considered for a long time to be an evolutionary remnant. Structurally meniscus is a biphasic tissue. Also, a set of collagen fibrils are oriented in the circumferential direction together with a few radial fibrils present in the central and randomly oriented in the superficial layer, contributes tissue tensile stiffness. The orientation of collagen fibres makes meniscus anisotropic in compression, tension and shear [1–7]. During daily activities, meniscus prevents the articular cartilage from premature degeneration by avoiding abnormally high stresses due to good load transfer capabilities of the meniscus. When the meniscus is injured, clinical interventions as partial or total meniscectomy might transfer higher forces through articular cartilage between femur and tibia which might be the cause of osteoarthritis [8,9]. These physiological changes in the meniscus increase the stress distributions over the cartilage surfaces, reducing the ability of the tissue to resist these excessive loads.

On the other hand, computational approaches as the finite element method (FEM) allows to evaluate the stresses, strains and possible failure points in joint tissue with in-vitro analysis at different loading condition such as walking [10–12]. In addition, complex finite element models are sensitive to different variables such as material properties, boundary conditions, geometries and meshing techniques [13–16]. Hence, different (circumferential and radial) orientations of collagen structure should be considered in three-dimensional modelling of the tissue. Based on literature review, different material models exist to describe the mechanical behaviour of the meniscus in the finite element analysis (FEA) of the knee joint. For instance, the menisci had been modeled as a compressive spring element to represent the resistance of the meniscus [17–20]. Such a model implemented the motion of the knee joint, but it did not present stress and strain in the articular
cartilage [17,20]. On the other hand, numerical knee model was developed with the assumption that the menisci behaves as a single phase (solid phase); it was modeled as a linear elastic isotropic material with elastic modulus ($E=59$ MPa) and Poisson’s ratio ($\nu=0.49$) [21–28]. However, this assumption does not take into account the time-dependency response of the tissue caused by the interstitial fluid flow during compressive loads. Vaziri et al., [29] applied the poroelastic transversely isotropic meniscus model. He considered the response of both fluid and solid phase and the inherent contribution of collagen fibrils oriented along the circumferential direction. Similarly, many other studies were conducted with the transversely isotropic elastic material model of the meniscus [30,31] to reduce the model generation and simulation times, and it incorporates the compression and tension non-linearity during the dynamic loading condition. None of the early mentioned studies had been included the knee joint modelling with real gait cycle. In the FEA of the knee joint [11,12,32,33] a transversely isotropic elastic material for meniscus was extensively used. However, those studies do not incorporate the compression tension nonlinearity effect of the tissue.

In addition, the meniscus is considered as a fibril reinforced poroelastic material [16,34], and this characteristic has been included in various FEA studies of knee joint under different loading conditions [35–40]. The FRPE model can integrate the mechanical role of the collagen, proteoglycans and fluid in both dynamic as well as equilibrium loading condition [36,37,41]. However, FRPE model is computationally demanding with three-dimensional (3D) geometries, and its implementation might take time. All these limitations must be overcome for future clinical applications in the treatment of multiple subjects [42]. Particularly, simpler applications might expedite the generation and simulation of complex biomechanical models. In this sense, a good candidate can be the orthotropic poroelastic (OTP) constitutive model. This material has nine independent elastic constants which can represent the collagen architecture (radial tie and circumferential) as well as the role of the fluid flow in the tissue.

In this thesis, an OTP constitutive model was evaluated to reproduce the similar effect of FRPE meniscus model on articular cartilage. First, OTP properties were adjusted using a numerical cubic model, which represents a portion of the meniscus. FRPE model was used as a reference to optimize the OTP material parameters. The reaction force, maximum principal stress, maximum principal strain were compared with respect to FRPE cubic model results. Finally, OTP meniscus properties were applied in a 3D knee joint model and contrasted to FRPE meniscus model.
Chapter 2

Background

2.1 Knee Joint

The knee joint is the largest and the most complex synovial joint which performs simultaneously six degrees of freedom (three rotations and three translations). It consists of three bones, femur, tibia and patella. It is a modified hinge joint, and the joint movement is possible with a complex set of translation and rotation motions [43–45]. The knee joint contains two well-defined joints; tibiofemoral joint and patellofemoral joint. In former case, femur articulates over tibia, and in later case, patella articulates over the femur. The tibiofemoral joint comprises of four ligaments: posterior cruciate ligament (PCL), anterior cruciate ligament (ACL), medial collateral ligament (MCL) and lateral collateral ligament (LCL) (Figure 1). The smooth motion of knee transfers total joint loads from femur to the tibia where articular cartilage and meniscus reduce the friction and shear forces between the bones [4].
The knee joint acts as a pivot between the two longest bones femur and tibia where bones are connected with the four ligaments. The MCL and LCL connect the sides of the knee joint; stabilizing the joint in medial and lateral sides during motion. Similarly, ACL and PCL stop the sliding of the femur and tibia in the anterior and posterior direction. The knee joint is surrounded by the joint capsule filled with synovial fluid. The synovial fluid facilitates the cartilage lubrication and provides a frictionless sliding movement between femoral, tibial and patellar cartilages. Besides lubrication, synovial fluid provides nutrition to the articular cartilage [47–49].

2.2 Articular cartilage

The articular cartilage covers the ends of articulating bones in synovial joints (Figure 2). Articular cartilage is an avascular layer of fibrous connective tissue. It acts as a cushion which transmits and distributes articular forces over the bone surfaces, so the local stress concentrations are well distributed. It ensures the smooth and virtually frictionless movement of the joint surfaces together with the synovial fluid [50,51].
2.2.1 Composition

Articular cartilage is a fibril-reinforced, viscoelastic tissue with a highly heterogeneous composition. The essential biochemical building blocks of articular cartilage are chondrocytes and extracellular matrix (ECM). The chondrocytes are cartilage cells which fill 1% of the volume of the tissue [52]. They synthesize the ECM components such as proteoglycans, collagen, and glycoproteins and regulate the macromolecular content of cartilage. Moreover, chondrocytes respond to external stimulus and tissue damage in cartilage. The ECM has both fluid and solid phase. The solid phase is mainly composed of proteoglycans (PGs), collagen fibers and non-collagenous proteins which accounts for 20-40% of total cartilage wet weight. The PGs represents the second major component of the solid phase of the articular cartilage [53–56]. Similarly, the fluid phase mainly contains the interstitial fluid that occupies between 60% and 90% of the total tissue weight. The content of the interstitial fluid is higher in the superficial zone than in the deep zone. In fact, the fluid content in articular cartilage decreases with the function of tissue depth [52,57–61]. The main mechanical role of interstitial fluid is to control the dynamic and impact response of cartilage under loading [50,62]. It has been shown that due to low permeability of cartilage it can bear up to 90% of the total stresses with collagen fibril network under dynamic or impact loads [53]. PGs are macromolecules of protein with covalently attached negatively charged glycosaminoglycan (GAG) side chains. The proteoglycans determine the stiffness of the cartilage
in a mechanical equilibrium [51]. The collagen constitutes 15-22% of the wet weight of cartilage. Mostly, type-II collagen is found in cartilage tissue, and these fibrils are always under tension even it is free from loading because it opposes the tissue swelling caused by the negative charge of PGs [62]. They are distributed in different patterns in different zones. In the superficial zone, they are organized in parallel to the cartilage surface. In the middle zone, fibrils are located towards the deep zone, whereas in the deep zone, collagen fibrils are oriented in a normal direction to the cartilage surface. The collagen matrix regulates the stress and strain and provide high tensile stiffness, reducing the tissue deformation in fibril direction during joint loading [40,63].

The composition and structure of the immature cartilage is actively adapted to the loading conditions to which the joint is subjected. For instance, it has been shown that exceeded level of tissue stresses or strains may alter these compositions and structures, which eventually may lead to tissue degeneration. On the other hand, studies have reported that immobilization may impair mechanical properties in the cartilage tissue by altering composition and structure. As the mechanical properties of the articular cartilage entirely rely on the structure and composition of the tissue [64,65], it is crucial to maintain normal loading conditions during daily living activities.

2.3 Meniscus

Menisci are crescent-like fibro cartilaginous tissue in knee joint cavity. The knee joint contains meniscus in both the lateral and medial compartment rested between the corresponding tibial plateau and femoral condyle cartilages inside the knee joint capsule (Figure 3). The cross-section of the menisci is a wedge-like, and their shape is accommodated according to the tibial and femoral contact surfaces. The superior concave surface provides the effective articulation with the respective convex femoral condyles. Similarly, the inferior flat surface easily adapt to the tibial plateau [66–69]. Structurally, the meniscus is a complex tissue and a critical component for a normal knee joint function during daily living activities. Each meniscus is a glossy-white complex tissue composed of cells, specialized ECM molecules and region-specific innervation and vascularization. The ECM incorporates site-specific contents of PGs, collagen fibrils and interstitial fluid. [46].
The meniscus horns are joined to the tibia through four ligamentous attachments which support the meniscus in the joint and distribute the loads in the knee joint. Also, it prevents from extrusion during movement of condyles and provides functional integrity of the meniscus during physical activities. The main stabilizing ligaments in meniscus are a transverse ligament, medial collateral ligament and meniscofemoral ligaments [68,70,71]. Medial and lateral menisci have noticeably different structure and dimensions. The medial meniscus is semicircular and measures approximately 35 mm in length [66], and its body is wider and thicker posteriorly than anteriorly. The radius of curvature of the medial meniscus in the transverse plane is large [72].

The medial menisci are larger than its lateral counterpart. The approximate dimension of the medical meniscus is 40.5-45.5 mm long and 27 mm wide. The medial menisci cover grossly 51-74% portion of tibial cartilage medially [73,74]. The insertions of the medial meniscus are further apart than those of the lateral meniscus [72]. The anterior medial meniscus horn attaches medial meniscus to the anterior ridge of the tibia and intercondylar eminence. The medial meniscus connects tightly to the joint capsule and the posterior oblique ligament just behind the medial collateral ligament. Similarly, it has a connection with the anterior cruciate ligament. The lateral and medial menisci are attached to each other through the transverse ligament.

The lateral meniscus is broader and more symmetrical, and it conceals a more substantial portion of the tibial cartilage plateau, 75-93% laterally compared to the medial meniscus [75]. The posterior horn of the lateral meniscus inserts directly into the intercondylar prominence, and it is connected to the posterior cruciate ligament by the ligament of Wrisberg. The anterior horn of the lateral meniscus inserts anteriorly to the intercondylar eminence next to the anterior ACL attachment site. The lateral meniscus has more mobility than medial meniscus [72].

In summarize, the meniscus has important biomechanical functions in shock absorption, load transmission, avoiding stress concentration in articular cartilage as well as in subchondral bone and the stability in the tibiofemoral joint sections during static and dynamic loading of knee joint. These biomechanical functions of meniscus entirely depend on its material properties, structure and morphology [76–79].
2.3.1 Composition

Biochemically, the meniscus in knee joint represents a fibro cartilaginous tissue. 70% of the total weight of meniscus consists of water. The main structural components of the meniscus are collagen type I, elastin and proteoglycans [80–82].

2.3.1.1 Collagen

Collagen is the main fibrillar component of the meniscus. Type I collagen is the most abundant which occupies 15-25% of wet weight and 75% of dry weight in meniscus [64]. These fibrils are well organized for the conversion of a compressive load into circumferential stress [83]. Collagen fibrils inside the meniscal tissue are divided into three groups based on definite directions: radial, oblique and circumferential [1,84]. The radial fibrils are situated between the articular surface, and the capsular periphery (Figure 4). The circumferential fibrils are organized in bundles from the dorsal to the ventral horn of the meniscus. On the other hand, the oblique fibrils do not have a specific site and intersect with both groups of fibrils [85]. Depending on collagen orientations, the
meniscus can be divided into three distinct layers across the tissue; the superficial layer, lamellar layer and the central main layer (Figure 4).

The external network consists of a meshwork of fine fibrils with a random orientation that covers the femoral and tibial sides of human meniscus surface. The lamellar layer consists of a layer of lamellae of collagen fibrils just below the superficial network on the femoral and tibial sides. In this layer, a bundle of collagen fibril interconnects at different angles apart from external circumference of the posterior and anterior segments where the fibrils are arranged in a radial direction. The innermost layer is the central layer, and it has circumferentially oriented collagen fibril. It also contains the main portion of the collagen fibrils [84]. Collagen fibrils control the tensile and dynamic compressive properties of the meniscus. It also opposes the tissue swelling caused by the negatively charged of PGs, so that collagens are always under tension even it is free from loading [62]. Elastin is another important fibrillar part of the meniscus. However, its significance is not identified. The combination of both mature and immature elastin fibers are obtained in very low concentrations (<0.6%) in adult meniscus [86,87].

Figure 4: The orientation of collagen fibre. The arrowhead shows the orientation of radial tie-fiber that intervened the circumferential collagen fibre modified with the figure [84].
2.3.1.2 Proteoglycans

Proteoglycans are the glycosylated proteins that contain covalently bonded negatively charged glycosaminoglycan’s (GAG) which occupy 17% wet weight [88,89]. It is the most abundant component present in ECM of the meniscus. The normal menisci contain different types of GAGs which are chondroitin-6-sulfate, dermatan sulfate, chondroitin-4-sulfate and keratin sulfate [89]. The largest proteoglycan of the meniscus is aggrecan, and smaller are biglycan and decorin [90]. The presence of these molecules is regionally varied; the inner two-thirds contain a relatively higher proportion of proteoglycans than the outer one third. PGs with their fixed charge density engage in osmotic swelling [91] which regulates the tissue hydration and pressure. It also contributes to the viscoelastic nature of the soft tissue. The primary function of these molecules is to support the meniscus to absorb water whose confinement supports the tissue under compression [92]. The adhesion glycoproteins conjoin the ECM components and cells. The main adhesion glycoprotein found in the human meniscus is thrombospondins, fibronectin and collagen VI [86,93].

2.3.1.3 Interstitial fluid

The interstitial fluid is the most abundant component found in the ECM of the meniscus which accounts for the 60-75 % wet weight of the meniscus. The fluid content in meniscal tissue is arranged homogenously from superficial to deeper layers [88]. The fluid entrapment and its interactions with the ECM components enable the soft tissues to resist the fast-rate compression and recover the tissue to original shape after deformation. In addition to that, interstitial fluid pressurization properties influence the dynamic stiffness of the soft tissue. Similarly, the fluid flow also controls the prolonged loading (creep) and tissue time-dependent stress-relaxation behaviour [94–96]. Complementary, another role of the interstitial fluid is to provide nourishment and nutrients to the soft tissues [62].

2.4 Biomechanics of meniscus

The role of the meniscus within the knee joint is load bearing. Besides that, it is involved in different functions: load transmission, shock absorption and lubrication. Also, it provides nutrition to the
Articular cartilage [97,98]. The ability of the menisci to execute these mechanical functions is based on their fundamental material properties, their anatomic structure and attachments to the tibia [1]. According to the geometry of the knee joint, the medial compartment is more congruent than the lateral compartment. The medial femoral condyle articulates over a concave medial tibial plateau, and the lateral femoral condyle articulates over a flat and slightly convex lateral tibial plateau. The biomechanics of the meniscus in the tibiofemoral joint results from their shape, structure and attachments of the menisci and it is responsible for healthy knee joint. The wedge shape of the meniscus enables it to adapt and stabilize between the curved femoral condyle and the flat tibial plateau during the articulation.

During some daily activities, the knee joint is subjected to the compressive force in the axial direction, and this can be splitted into compressive and shear components [67]. The compressive force through the joint spreads on the articulating contact area and develops contact stress. The menisci optimize the load that is transferred through the tibiofemoral joint by enlarging the congruency of the articulation so that the contact area will increase and the contact pressure on the articulating surfaces decreases. When the femoral condyles exert the downward pressure (compressive force) on to the menisci, the wedged shape of the meniscus and its horn attachments to the tibia through anterior and posterior insertional ligaments serve to convert the part of compressive force (axial load) to the horizontal hoop stress (circumferential tension) to resist further deformation. In addition, the stiffness of the meniscus and presence of collagen fibre in the circumferential direction has great effect to generate the hoop stress. At the same time, the origin of the shear force between the collagen fibres deforms meniscus radially [99,100].

The biomechanical properties of the menisci are properly adjusted to bear the forces exerted on the tissue. The aggregate axial modulus of elasticity of the meniscus which resists the axial compression to 0.1-0.15 MPa [101]. The modulus of elasticity in the tissue varies between the radial and circumferential direction. The modulus of elasticity in the circumferential direction is approximately 100-300 MPa, and it is ten times lower than the radial one. The shear modulus of the meniscus is approximately 0.12 MPa [102]. On the other hand, menisci influence in the joint mobility, which allows to maximise the degree of conformity of articulation across knee flexion.

The load-bearing mechanism appears throughout the whole range of knee-joint flexion as the menisci are mainly attached to the tibia and allow the displacement in all directions. When the knee
joint fully extends, the large radius of curvature of the distal part of the femur (medial and lateral condyle) will contact to the whole area of the menisci from their anterior to the posterior aspect. On the other hand, when the flexion of the knee occurs, the contact area between the posterior aspect of the femoral condyles and meniscus will decrease, and it moves posteriorly towards and onto the posterior meniscal horns. Due to this movement, menisci get displaced posteriorly; resulting in the displacement of anterior horn posteriorly with 10 mm towards the lateral side. Some typical properties like geometry, attachments to the tibia and microstructure of the menisci enable proper functioning of the knee joint. If any of these three elements are disrupted, the biomechanics of the menisci will be hampered [76,103–105].

2.5 Constitutive modelling of meniscus

An appropriate constitutive and geometric representation of the meniscus is required to evaluate internal stress-strain, pore-pressure and resultant reaction force through the meniscus in the articular cartilage. For this purpose, the poroelastic models [106,107] have been used for the time-dependent mechanical behavior of the extracellular matrix of the menisci.

2.5.1 Poroelastic theory

The poroelastic theory considers the existence of solid (PG- collagen) phase and liquid (interstitial fluid) phase separately and the fluid flow in and out of the tissue because the solid phase is assumed to be porous and the fluid phase is non-viscous. All the poroelastic tissues show the fundamental incompressible and non-dissipative features. When these tissues are compressed or pulled rapidly in one direction a corresponding lateral response occurs, and the total volume remains the same. In poroelastic models, both solid and liquid phases are treated as the inherently incompressible, non-dissipative and immiscible. The fluid flow will give the time-dependent changes so that the tissue response will not linear under a step strain or stress. Moreover, the permeability of the porous media contributes to the equilibrium time.
According to the biphasic theory, the total stress $\sigma_{\text{total}}$ is given as the sum of solid matrix stress $\sigma_{\text{solid}}$ and fluid stress $\sigma_{\text{fluid}}$ stress tensors which are given as follows [50,108–110]:

$$\sigma_{\text{total}} = \sigma_{\text{solid}} + \sigma_{\text{fluid}},$$  \hspace{1cm} (1)

where $\sigma_{\text{solid}}$ and $\sigma_{\text{fluid}}$ are defined as follows:

$$\sigma_{\text{solid}} = -\phi_{\text{solid}} \ p \ I + \sigma^E,$$ \hspace{1cm} (2)

$$\sigma_{\text{fluid}} = -\phi_{\text{fluid}} \ p \ I,$$ \hspace{1cm} (3)

finally the total stress is defined

$$\sigma_{\text{total}} = -p I + \sigma^E,$$ \hspace{1cm} (4)

where $\phi_{\text{solid}}$ and $\phi_{\text{fluid}}$ are volume fraction for the solid and fluid phase respectively, $p$ is the fluid pressure, $I$ is the unit tensor and $\sigma^E$ is the effective solid stress. When tissue obtain an equilibrium state, all the fluid will flow out. At this stage, solid stress tensor restrict the further tissue deformation[50]. According to Hook’s law, the effective solid stress tensor for linear elastic materials can be written as follows:

$$\sigma^E = C \ \varepsilon^E,$$ \hspace{1cm} (5)

where $C$ is the stiffness matrix and $\varepsilon^E$ is the elastic strain tensor. The stiffness matrix is different for isotropic, transverse isotropic and orthotropic materials.

The velocity of fluid flow $Q$ through porous material is commonly described by the Darcy’s law. According to the Darcy’s law,

$$Q = A k \ \frac{\Delta p}{h},$$ \hspace{1cm} (6)

where $A$ is the cross-sectional area of fluid flow, $k$ is the permeability of the medium and $\Delta p/h$ is the ratio between the fluid pressure difference and the penetration depth.

Similarly, the permeability $k$ for the porous tissue is described by a non-linear relation with tissue porosity [111];
where $k_0$ is initial permeability, $M$ is material constant, $e_0$ is initial void ratio and $e$ is current void ratio. The void ratio is given by the fluid ($n^{\text{fluid}}$) and solid ($n^{\text{solid}}$) volume fraction as follows:

$$e = \frac{n^{\text{fluid}}}{n^{\text{solid}}}.$$  

(8)

2.5.2 Isotropic, linear elastic material behavior

Isotropic linear elastic is the simplest model to represent the mechanical response of the biological soft tissue where the material properties are the same in each direction. In this case, it is considered that the stress-strain curve is linear and independent of the direction of the load. Therefore, the stiffness matrix $C$ is described by

$$C = \frac{E}{(1+\nu)(1-2\nu)} \begin{pmatrix}
1 - \nu & \nu & \nu & 0 & 0 & 0 \\
\nu & 1 - \nu & \nu & 0 & 0 & 0 \\
\nu & \nu & 1 - \nu & 0 & 0 & 0 \\
0 & 0 & 0 & 1 - 2\nu & 0 & 0 \\
0 & 0 & 0 & 0 & 1 - 2\nu & 0 \\
0 & 0 & 0 & 0 & 0 & 1 - 2\nu
\end{pmatrix},$$  

(9)

where $E$ and $\nu$ are the Young’s modulus and the Poisson’s ratio respectively. The shear modulus ($G$) and the bulk modulus ($K$) are related with the Young’s modulus and the Poisson’s ratio with the following equation.

$$G = \frac{E}{(1+\nu)(1-2\nu)},$$  

(10)

$$K = \frac{E}{3(1-2\nu)}.$$  

(11)

The isotropic materials have two independent variables: Young’s Modulus $E$ and the Poisson’s ration $\nu$. In principle, the isotropic material does not take into account the anisotropy of tissue. For instance, the properties caused by the predominant natural collagen fiber orientation e.g.
circumferential and radial direction in meniscus and cartilage. Hence, isotropic models exhibit limitations to represent the material nonlinearities of menisci and articular cartilage.

2.5.3 Orthotropic constitutive model

The orthotropic material behavior shows that at least two orthogonal planes of symmetry and the material properties are independent of the direction within each plane. These materials require nine independent variables (elastic constants) in their constitutive matrices. $E_1$, $E_2$ and $E_3$ are Young’s modulus of elasticity, $v_{12}$, $v_{23}$, $v_{13}$ are three Poisson’s ratio and $G_{12}$, $G_{23}$ and $G_{31}$ are three shear moduli in three orthogonal plane-1(1-2), plane-2(2-3) and plane-3(1-3) respectively in an orthotropic cube model (Figure 5).

Figure 5: A cube with three orthogonal planes denoted as plane-1 (1), plane-two (2) and plane-three (3) with Young’s moduli of elasticity $E_1$, $E_2$ and $E_3$ for each plane.
The stiffness matrix for the orthotropic materials is given by the Equation 12;

\[
C = \begin{pmatrix}
1 - \nu_{23} \nu_{32} & \frac{\nu_{21} + \nu_{31} \nu_{23}}{E_2 E_3 \Delta} & \frac{\nu_{31} + \nu_{21} \nu_{31}}{E_2 E_3 \Delta} & 0 & 0 & 0 \\
\frac{\nu_{12} + \nu_{13} \nu_{32}}{E_3 E_1 \Delta} & \frac{1 - \nu_{31} \nu_{13}}{E_2 E_3 \Delta} & \frac{\nu_{32} + \nu_{31} \nu_{12}}{E_2 E_3 \Delta} & 0 & 0 & 0 \\
\frac{\nu_{13} + \nu_{12} \nu_{23}}{E_4 E_2 \Delta} & \frac{\nu_{23} + \nu_{13} \nu_{21}}{E_4 E_2 \Delta} & \frac{1 - \nu_{12} \nu_{21}}{E_4 E_2 \Delta} & 0 & 0 & 0 \\
0 & 0 & 0 & \frac{2G_{23}}{E_4 E_2 E_3} & 0 & 0 \\
0 & 0 & 0 & 0 & \frac{2G_{13}}{E_4 E_2 E_3} & 0 \\
0 & 0 & 0 & 0 & 0 & \frac{2G_{12}}{E_4 E_2 E_3}
\end{pmatrix},
\] (12)

where

\[
\Delta = \frac{1 - \nu_{12} \nu_{21} - \nu_{23} \nu_{32} - \nu_{31} \nu_{13} - 2\nu_{12} \nu_{23} \nu_{31}}{E_4 E_2 E_3}.
\] (13)

In fact, the stiffness matrix is symmetric, this leads to following relationships;

\[
\frac{\nu_{21} + \nu_{31} \nu_{23}}{E_2 E_3 \Delta} = \frac{\nu_{12} + \nu_{31} \nu_{13}}{E_3 E_1 \Delta},
\] (14)

\[
\frac{\nu_{32} + \nu_{31} \nu_{12}}{E_3 E_1 \Delta} = \frac{\nu_{32} + \nu_{13} \nu_{23}}{E_1 E_2 \Delta},
\] (15)

\[
\frac{\nu_{31} + \nu_{21} \nu_{32}}{E_2 E_3 \Delta} = \frac{\nu_{13} + \nu_{12} \nu_{23}}{E_1 E_2 \Delta},
\] (16)

\[
\frac{\nu_{21}}{E_2} = \frac{\nu_{12}}{E_1},
\] (17)

\[
\frac{\nu_{31}}{E_3} = \frac{\nu_{13}}{E_1}.
\] (18)
2.5.4 Fibril reinforced poroelastic constitutive model

Fibril reinforced poroelastic (FRPE) material considers the non-fibrillar and the collagen fibrils components as well as the poroelastic properties in the tissue. In addition, this formulation is able to take into account the anisotropy and the stress-strain nonlinearity of the soft tissue.

2.5.5 Non-fibrillar part

The non-fibrillar part represents the PGs and interstitial fluid in the tissue. The non-fibrillar part of a solid matrix can be assumed to be isotropic and linear elastic with Young’s modulus of elasticity $E$ and Poisson’s ratio $\nu$ so that it can be described with Hooke’s law [51,112]. The stiffness matrix $\mathbf{C}$ for the non-fibrillar part is given by,

$$
\mathbf{C} = \frac{E}{(1+\nu)(1-2\nu)} \begin{pmatrix}
1 - \nu & \nu & \nu & 0 & 0 & 0 \\
\nu & 1 - \nu & \nu & 0 & 0 & 0 \\
\nu & \nu & 1 - \nu & 0 & 0 & 0 \\
0 & 0 & 0 & 1 - 2\nu & 0 & 0 \\
0 & 0 & 0 & 0 & 1 - 2\nu & 0 \\
0 & 0 & 0 & 0 & 0 & 1 - 2\nu
\end{pmatrix}.
$$

However, soft tissues perform in a highly nonlinear manner within large deformations. For this reason, a Neo-Hookean formulation (hyperelastic) can be used to capture this nonlinear stress-strain behavior [113]. The following equation defines the neo-Hookean model,

$$
\sigma_{nf} = K \frac{\ln(f)}{f} \mathbf{I} + G \left( \mathbf{F} \mathbf{F}^T - f^2 \mathbf{I} \right),
$$

where $G$ and $K$ are shear and bulk modulus respectively, $\mathbf{F}$ is the deformation gradient tensor, $f$ is the determinant of the deformation tensor $\mathbf{F}$ and $\mathbf{I}$ is the unit tensor. In addition, $K$ and $G$ can be further expressed by Young’s Modulus ($E_{nf}$) and Poisson’s ratio ($\nu_{nf}$) of the non-fibrillar matrix;

$$
K = \frac{E_{nf}}{3(1 - 2\nu_{nf})},
$$

$$
G = \frac{E_{nf}^3}{(1 + 2\nu_{nf})^2(1 - 2\nu_{nf})}.
$$
\[ G = \frac{E^{nf}}{2(1 + v^{nf})}. \]  \hspace{1cm} (22)

2.5.6 Fibrillar part

The fibrillar part includes the collagen fibrils present in the cartilage and meniscus matrix. In case of the FRPE material, a linear strain-dependency of the collagen fibril network modulus \( E_f \) with tensile strain \( \varepsilon_f \) has been applied. Similarly, an FRPE material can be modelled as a system that consists of a linear spring \( E_0 \) connected to a nonlinear spring \( E_1 \) in parallel where, \( E_1 = E_\varepsilon \varepsilon_f \) in (Figure 6a). Then, the fibrillar stress \( \sigma_f \) can be defined as;

\[ \sigma_f = E_0 \varepsilon_f + \frac{1}{2} E_\varepsilon \varepsilon_f^2. \] \hspace{1cm} (23)

![Figure 6: (a) A schematic representation of elastic collagen fibrils and (b) Viscoelastic collagen fibrils. \( E_0 = \text{linear spring constants}, E_\varepsilon = \text{nonlinear spring constants, } \varepsilon_f = \text{fibril strain, } \eta = \text{damping coefficient, } \varepsilon_\varepsilon = \text{strain in nonlinear spring and } \varepsilon_\varepsilon = \text{strain in dashpot.} \]
\[
\sigma_f = E_0 \varepsilon_f + \eta \dot{\varepsilon}_f - \dot{\varepsilon}_e, \quad (25)
\]

where \( \dot{\varepsilon}_e \) indicates the time derivative of strain which is given by,

\[
\varepsilon_e = \sqrt{\frac{\sigma_e}{E_e}}, \quad (26)
\]

\[
\dot{\varepsilon}_e = \frac{1}{2\sqrt{\sigma_e E_e}} \dot{\sigma}_e, \quad (27)
\]

and

\[
\sigma_e = \sigma_f - E_0 \varepsilon_f. \quad (28)
\]

where \( \sigma_e \) = effective solid stress, \( \varepsilon_f \) = fibril tensile strain.

when, substituting equations (26), (27) and (28) in equation (25) then stress \( \sigma_f \) in the viscoelastic fibrils can be calculated:

\[
\sigma_f = -\frac{\eta}{2\sqrt{(\sigma_f - E_0 \varepsilon_f) E_e}} \dot{\sigma}_f + E_0 \varepsilon_f + \left( \frac{\eta E_0}{2\sqrt{(\sigma_f - E_0 \varepsilon_f) E_e}} + \eta \right) \dot{\varepsilon}_f, \quad (29)
\]

where \( \sigma_f = 0, \varepsilon_f \leq 0 \).
Chapter 3

Aim of the study and hypothesis

In FEA of knee joint, FRPE constitutive model has been utilized to investigate the role of the meniscus in the knee joint. This particular material considers the role of collagen fibrils as well as the fluid effect under dynamic loading during the stance phase of the gait cycle [35–40]. However, FRPE properties of the meniscus are complex, and their implementation is computationally demanding. As such, this approach might have limitations for effective clinical applications in day to day practices. Hence, this limitation motivates to find other suitable constitutive model to represent the meniscus that could mimic the similar mechanical response in the tibial cartilage.

The aim of this study is to propose a simplified material model of meniscus to be applied in FEA of the knee joint, instead of FRPE model, to obtain similar mechanical response in tibial cartilage under representative loading during the phase of stance. For this purpose, an OTP constitutive model was adjusted to produce the similar mechanical response of FRPE material properties. Then, OTP meniscus model was implemented into a 3D computational knee joint model and total reaction force, stress, strain, pore pressure and fibril strain on tibial cartilage were measured. Finally, these mechanical responses were compared with FRPE meniscus model. The hypothesis is that an OTP meniscus model can be suitable and reliable alternative for FEA analysis of the knee joint. It might allow to perform several subjects-specific knee joint models rapidly, avoiding the time limitations during clinical application.
Chapter 4

Material and methods

The workflow of this study is described in the Figure 7. As summary, OTP constitutive parameters were adjusted based on FRPE model. The models were compared with respect to the following results: reaction force, principal stress, principal strain and pore pressure. Finally, based on these analysis, OTP material parameters such as Poisson’s ratio, shear modulus and Young’s modulus were adjusted and implemented into a global knee joint model.

Figure 7: The workflow of the study. Adjustment of the OTP parameters was performed by contrasting the FRPE results. After this optimization, the OTP parameters were implemented into the meniscus in a 3D computational model of the knee joint.

FRPE Static/ Dynamic = fibril reinforced poroelastic cube model is subjected to 10% compression refers Static and 10% tension refers Dynamic, OTP Static/ Dynamic = orthotropic poroelastic cube model is subjected to 10% compression refers, Static, and 10% tension refers, Dynamic in the FEA.
4.1 Adjustment of OTP material parameters

A three-dimensional FEM of a cube that represents a building block of the meniscus was developed in Abaqus v6.14-3 (Dassault Systems, Providence, RI, USA). The geometry was meshed using 8-node brick element with trilinear displacement with pore pressure properties (type C3D8P). Then, FRPE properties published in the literature [116,117] were implemented into the model (Table 1). This FRPE cubic model was used as reference to adjust the OTP parameters. Finally, based on these comparisons, the OTP material parameters for instance Poisson’s ratio, Shear modulus and Young’s modulus were adjusted.

Table 1: FRPE material properties used as reference.

<table>
<thead>
<tr>
<th>Material properties</th>
<th>Fibril-reinforced poroelastic meniscus</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_0$ (MPa)</td>
<td>3.7</td>
</tr>
<tr>
<td>$E_E$ MPa</td>
<td>$1 \times 10^{-8}$</td>
</tr>
<tr>
<td>$E_m$ (MPa)</td>
<td>0.08</td>
</tr>
<tr>
<td>$k$ ($\times 10^{-15}$ m$^4$/Ns)</td>
<td>0.02</td>
</tr>
<tr>
<td>$\nu_m$</td>
<td>0.3</td>
</tr>
<tr>
<td>$M$</td>
<td>12.1</td>
</tr>
<tr>
<td>$k_0$ ($\times 10^{-15}$ m$^4$/Ns)</td>
<td>0.08</td>
</tr>
</tbody>
</table>

$E_0$ = initial fibril network modulus, $E_E$ = strain-dependent fibril network modulus, $E_m$ = non-fibrillar matrix modulus, $k_0$ = initial permeability and $M$ = exponential term for the strain-dependent permeability.

4.1.1 Boundary conditions of the cubic model

The boundary conditions are described in the Figure 8 (a), (b) and (c) for the compression through Y-axis, tension through Z-axis and shear through Z-axis, respectively. In detail, all the nodes from one side of the cube model were fixed to restrict all three translation and rotational motion (X, Y, and Z direction) and to the nodes of the opposite side where compression, tension and shear force
are applied were fixed to restrict three rotational motion. The fluid flow through the cube surface was allowed both in the dynamic and equilibrium loading. Compression, tension and shear forces were controlled with a displacement of 10% translation strain (Figure 8).
As was mentioned in a previous chapter, nine independent material constants describe an orthotropic material which are $E_1$, $E_2$, $E_3$, $v_{12}$, $v_{31}$, $v_{23}$, $G_{12}$, $G_{31}$ and $G_{23}$. Hence, Young’s modulus of elasticity in the circumferential plane in the fiber direction ($E_3$) and in the axial plane ($E_2$) were converted from FRPE model [116,117]. The remaining modulus of elasticity in the radial direction ($E_1$) was calculated based on the tensile test (Figure 8b). The shear moduli ($G_{12}$, $G_{31}$, $G_{23}$) were calculated based on the shear simulations (Figure 8c), and the Poisson’s ratios ($v_{23}$, $v_{12}$, $v_{31}$) were calculated based on the compression and tensile simulations (Figure 8a-b). In the results, the Young’s modulus in circumferential and axial directions were set to 184.63 MPa [1] and 0.08 MPa, respectively. In the beginning, the modulus of elasticity in the radial direction was obtained ~4 MPa based on the tensile test, but in order to have good match in terms of internal stresses and forces through the tissue, it was increased to 6 MPa. Finally, in the OTP model the modulus of elasticity in radial direction was adjusted to 6 MPa, which displayed similar mechanical contribution as regarding the effect of the secondary collagen fibrils in the FRPE model. A list of the OTP adjusted parameters is described in the Table 2.
Table 2: Orthotropic material properties.

<table>
<thead>
<tr>
<th>Material properties</th>
<th>Orthotropic poroelastic Meniscus</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_1$ (MPa)</td>
<td>6</td>
</tr>
<tr>
<td>$E_2$ (MPa)</td>
<td>0.08</td>
</tr>
<tr>
<td>$E_3$ (MPa)</td>
<td>184.63</td>
</tr>
<tr>
<td>$v_{12}$</td>
<td>0.4</td>
</tr>
<tr>
<td>$v_{13}$</td>
<td>0.02</td>
</tr>
<tr>
<td>$v_{23}$</td>
<td>0.0001</td>
</tr>
<tr>
<td>$G_{12}$ (MPa)</td>
<td>0.62</td>
</tr>
<tr>
<td>$G_{13}$ (MPa)</td>
<td>1.99</td>
</tr>
<tr>
<td>$G_{23}$ (MPa)</td>
<td>1.99</td>
</tr>
<tr>
<td>$k \times 10^{-15} \text{m}^4/\text{Ns}$</td>
<td>0.08</td>
</tr>
</tbody>
</table>

$E_1$ = modulus of elasticity in radial direction, $E_2$ = modulus of elasticity in axial direction, $E_3$ = modulus of elasticity in circumferential direction, $v_{12}$ = Poisson’s ratio between plane 1 and 2, $v_{13}$ = Poisson’s ratio between plane 1 and 3, $v_{23}$ = Poisson’s ratio between plane 2 and 3, $G_{12}$ = shear modulus between plane 1 and 2, $G_{13}$ = shear modulus between plane 1 and 3, $G_{23}$ = shear modulus between plane 2 and 3, and $k$ = permeability.

4.1.2 Mechanical response comparison between OTP and FRPE models

The OTP model was similar to the FRPE model with respect to the principal stress, principal strain, reaction force and pore pressure) for both dynamic and static conditions, especially along Y-axis direction, the reaction force was identical (Table 3)
Table 3: Comparison of the mechanical response between Orthotropic and FRPE constitutive models in dynamic and equilibrium state.

<table>
<thead>
<tr>
<th>Cubic Models</th>
<th>Stress</th>
<th>Strain</th>
<th>Reaction force on 10% Axial Compression</th>
<th>Pore Pressure</th>
</tr>
</thead>
<tbody>
<tr>
<td>OTP Dynamic Compression</td>
<td>0.37</td>
<td>0.01</td>
<td>0.30</td>
<td>0.1</td>
</tr>
<tr>
<td>FRPE Dynamic Compression</td>
<td>0.62</td>
<td>0.05</td>
<td>0.51</td>
<td>0.11</td>
</tr>
<tr>
<td>OTP Equilibrium Compression</td>
<td>0.00</td>
<td>0.01</td>
<td>0.00</td>
<td>0.00</td>
</tr>
<tr>
<td>FRPE Equilibrium Compression</td>
<td>0.00</td>
<td>0.01</td>
<td>0.00</td>
<td>0.00</td>
</tr>
</tbody>
</table>

OTP = Orthotropic poroelastic, FRPE = Fibril reinforced poroelastic

4.2 Implementation of OTP meniscus into global knee joint model

A finite element knee model developed by Halonen et al., 2013 and modified by Orozco et al., 2018 [118] (Figure 10) was utilized in this study. As briefly description, a knee joint of an anonymous male volunteer (28 years and 80kg) was imaged using magnetic resonance imaging (MRI). The model geometry was built with the segmentation of femoral, tibial and patellar cartilage and meniscus and meshed according to the study done by Halonen et al., 2013. Furthermore, this model also contains patellar and quadriceps tendons with six other ligaments. The tendons and ligaments were segmented and meshed as described in the study by Orozco et al., 2018. In this knee model, the cartilage was defined as fibril-reinforced poroviscoelastic material. The fibrillar matrix (collagen fibrils) was modeled with viscoelastic properties, and nonfibrillar part (fluid and proteoglycans) were modelled as a Neo-Hookean, nonlinear, hyperelastic material. In global knee model, the outward fluid flow from the meniscus and tibial cartilage was not consider because of the short gait cycle (0.55 s) and low permeability in those tissues.

FRPE and OTP constitutive materials were used for modelling the meniscus. For FRPE meniscus model, circumferential oriented collagen fibrils were implemented assuming elastic behaviour ($E_\varepsilon \approx 0$). The implemented FRPE material parameters were presented previously in the Table 1. In the global knee joint model with OTP meniscus, a cylindrical coordinate system was
created to implement the adjusted OTP material parameters (Figure 9). The translations, moments and forces in the joint as well as the contact properties between parts were similar to the previous developed models [36,118].

Figure 9: (a) The finite element model of knee joint with articular cartilages, menisci and ligaments, (b) the lateral and medial meniscus in the knee joint and (c) the cylindrical co-ordinate system for meniscus where radial, axial and circumferential axis are represented by θ, ρ and Z, respectively.
Chapter 5

Results

The total joint reaction force on the tibial cartilage was found maximum during early stance phase of gait cycle. Force distributions were similar between the FRPE and OTP meniscus models. The force transferred through the lateral meniscus was higher when compared to the force transferred through the medial meniscus during walking.

In the lateral meniscus, the maximum difference between the peak forces through both meniscus models happened during the beginning of the early stance (~5% fraction of stance), whereas the force transferred through the medial meniscus between the materials models was similar along the gait. On the medial tibial cartilage, the average value of the maximum principal stress, maximum principal strain, pore pressure and fibril strain were calculated from its surface, and they were consistently distributed between the FRPE and OTP meniscus models along the gait cycle. On the lateral tibial cartilage, the maximum principal stress, maximum principal strain, fibril strain and pore pressure were unequally distributed during the early stance (~5% fraction of stance) phase of gait cycle, but these responses were equally distributed after ~15% fraction of stance phase between both meniscus models.

5.1 Comparison between constitutive models in the global knee model

The joint reaction force, principal stress, principal strain, pore pressure and fibril strain in the medial and lateral tibial cartilage were contrasted throughout the gait cycle between OTP and FRPE meniscus models. Moreover, the reaction force through the meniscus were evaluated and compared between both meniscus models. The maximum principal stress on the medial and lateral side of
OTP and FRPE meniscus models and the tibial cartilage during the stance phase of gait was evaluated (Figure 10). It was noticed that the maximum principal stress on the lateral and medial tibial cartilage was similar between the meniscus material models from beginning to the end of the stance phase of the gait cycle. However, the stress on the meniscus was varied between the FRPE and OTP material models along the gait cycle.

![Figure 10: The maximum principal stress distribution on medial and lateral in orthotropic poroelastic and fibril reinforced poroelastic material model during the stance phase of gait (Lat: lateral; Med: medial).](image-url)
5.1.1 Reaction force on tibial cartilage

The total reaction force on the medial and lateral tibial cartilages was similar throughout the stance of the gait cycle (maximum relative difference 5.12%) between the OTP and the FRPE meniscus models (Figure 11). At the beginning of the stance for both models, the reaction force increased rapidly and showed the first peak around the quarter of the stance phase. The peak forces were 1975 N (2.46 × body weight (BW) for FRPE) and 1925 (2.40 × BW for OTP). In contrast, the minimum reaction force was observed on the half of the stance. Moreover, the second peak was displayed in the 80% of stance which was 975 N and 925 N for FRPE and OTP meniscus model, respectively.

![Total Reaction Force in tibial cartilage](image)

Figure 11: The total reaction force on the tibial plateau cartilage throughout the stance phase of gait.

The reaction force on the lateral and medial tibial cartilage was calculated and compared separately between OTP and FRPE meniscus model (Figure 12). The behavior of the reaction force through the entire surfaces of the lateral and medial compartment tibial cartilage was identical, whereas the maximum reaction force was observed on the medial compartment cartilage than lateral tibial cartilage between the models. In the lateral tibial compartment, the peak reaction force was found around 7% of the stance phase. After the peak reaction force, it gradually decreased till to the terminal stance phase where the force increased. On the other hand, in the medial tibial
compartment, the first peak reaction force was found around 20% of stance phase and the second peak was around terminal (~ 80% of) stance phase.

Figure 12: The distribution of reaction force on the tibial cartilage as a function of time: (a) Lateral tibial cartilage (b) Medial tibial cartilage.

(a) Lateral tibial cartilage  (b) Medial tibial cartilage

Figure 12: The distribution of reaction force on the tibial cartilage as a function of time: (a) Lateral tibial cartilage (b) Medial tibial cartilage.
5.1.2 Reaction force through the meniscus

The reaction force through the FRPE and OTP meniscus models were calculated (Figure 13). The reaction force transferred through the medial meniscus was 117 N (0.14× BW) for FRPE and 104 N (0.13 × BW) for OTP model at 20% of the stance phase (maximum difference 11.1%). On the other hand, at the lateral side, the first peak forces was 540 N (0.62 × BW) for FRPE and 425 N (0.53 × BW) for OTP (maximum difference 21.29%) approximately at 5% of the stance phase. Also, the difference in second peak force between the models was 4.16 % around the half of the stance of stance.

Figure 13: The reaction force passes through the meniscus as a function of time during the gait cycle between the FRPE and OTP material models.
5.1.3 Maximum principal stress

The distribution of the average maximum principal stress in the medial and lateral tibial cartilage in the tibiofemoral contact area showed a similar pattern in both meniscus constitutive models (Figure 14). In the lateral compartment of tibial cartilage, the maximum principal stress occurred in the first quarter of stance and reached the minimum (tends to zero) from 50% to 80% of stance. On the other hand, medial compartment exhibited a peak at the first quarter of stance (~ 20% of stance) and the second highest peak stress occurred at 85% of stance. A higher stress was observed in the medial than the lateral part of the tibial cartilage during the phase stance of gait cycle for both meniscus models. The highest difference regarding maximum principal stress in the medial tibial cartilage between FRPE and OTP meniscus models was within the 50% to 90% of stance.

![Figure 14: Average maximum principal stress in the tibial cartilage as a function of time (a) Lateral tibial cartilage and (b) Medial tibial cartilage with OTP and FRPE constitutive models.](image)
5.1.4 Maximum principal strain

In the tibiofemoral contact area, the average maximum principal strain in the medial and lateral tibial cartilage in FRPE and OTP showed a similar response throughout the stance (Figure 15 a, b). In both models, the highest maximum principal strain was observed at 20% stance of the gait in the lateral cartilage whereas in the medial it occurred at 80% stance of the gait cycle. In addition, there was not strain from 50% to 85% of the phase of stance in the lateral tibial cartilage, but in medial part, the maximum principal strain increased from 50% of stance and reached to the peak at the 85% of stance.

(a) Lateral tibial cartilage  (b) Medial tibial cartilage

Figure 15: Average maximum principal strains in the tibial cartilage as a function of time (a) Lateral tibial cartilage and (b) Medial tibial cartilage with OTP and FRPE constitutive material models.
5.1.5 Fibril strain

The average fibril strain in the lateral and medial part of the tibial cartilage in the tibiofemoral contact area showed a similar response in both models (Figure 16 a, b). In the lateral tibial cartilage, the fibril strain was highest at around 20% of stance, and it remained zero within 50% and 85% of stance phase approximately. Similarly, in the medial tibial cartilage, the fibril strain decreases rapidly in the beginning (5%) of stance phase, and it increased and finally reached at the maximum about 85% phase of stance in both models. The highest difference in fibril strain between the lateral and medial cartilage occurred from 50% to 85% of the stance phase for each meniscus model.

Figure 16: Average fibril strains in the tibial cartilage as a function of time (a) Lateral tibial cartilage and (b) Medial tibial cartilage with OTP and FRPE constitutive material models.
5.1.6 Pore pressure

In the tibiofemoral contact area, the average pore pressure in the lateral and medial compartment of the tibial cartilage (Figure 17 a, b) displayed similar pattern in the medial side during whole phase of stance between the models. But, in lateral side of tibial cartilage in the beginning (less than 20%) phase of stance the pore pressure behaves differently. In the medial compartment, the first peak of pore pressure came about 20% stance phase which was followed by the second peak of pore pressure approximately 80% stance phase in both models. In the lateral tibial cartilage, with FRPE meniscus model the pore pressure displayed as a protuberance, while it was rather smooth as compared to OTP meniscus model around the 15% stance phase. However, the pore pressure between the models was quite smooth after approximately 40% phase stance followed a peak at the end of stance at medial tibial cartilage. The pore pressure found zero approximately from 40% to 90% phase of stance in lateral tibial cartilage in both FRPE and OTP meniscus models during gait cycle.

Figure 17: Average pore pressure in the tibial cartilage as a function of time (a) Lateral tibial cartilage and (b) Medial tibial cartilage with OTP and FRPE constitutive material models.
5.1.7 Comparison cartilage-cartilage contact area

The tibiofemoral cartilage-cartilage contact area in the FRPE and OTP meniscus model was compared as the function of stance phase. The result showed that in the lateral compartment (Figure 18a), the maximum peak cartilage-cartilage contact area occurred in the beginning of the loading response (5% of stance phase). The highest peak area of contact for FRPE and OTP meniscus model was 160 mm$^2$ and 140 mm$^2$, respectively.

A minimal contact area was found from approximately mid fraction of stance to terminal loading phase. Similarly, in the medial compartment (Figure 18b), the distribution pattern of cartilage-cartilage contact area between models was similar throughout the stance phase of the gait cycle. The peak cartilage-cartilage contact area (230 mm$^2$) was measured at medial side ~20% of stance phase of gait cycle.

Figure 18: The cartilage-cartilage contact area as a function of stance phase (a) Lateral and (b) Medial side in the knee joint.
5.1.8 Comparison cartilage-meniscus contact area

The cartilage-meniscus contact area with FRPE and OTP models as the function of time are presented for lateral side (Figure 19a) and medial side (Figure 19b). On the lateral side, the distribution of the cartilage-meniscus contact area pattern was different at ~ 5% of stance phase in both models. The cartilage-meniscus contact area with OTP model showed a smooth transition in the beginning of the stance whereas in the FRPE model was irregular. Similarly, on the medial side, the distribution pattern of the cartilage-meniscus contact area was similar with a maximum contact area during the ~ 20% of stance phase in both models.

![Figure 19: The cartilage-meniscus contact area as a function of stance phase](image)

(a) Lateral side  
(b) Medial side
5.1.9 Displacement of the meniscus

The displacement of meniscus in tibiofemoral joint compartment was calculated during the ~ 5 % stance phase which was equivalent to the 0.029 s of total step time (Figure 20). A similar displacement of lateral meniscus was noticed in X-direction in both models. The displacement of medial meniscus did not occur at that time between these two models. The different displacement in the lateral meniscus between the models was observed in the Z-axis with that specific time.

Figure 20: The displacement of OTP and FRPE meniscus model in X-axis, Y-axis and Z-axis occurred at the step time 0.029 s which is equal to ~ 5 % of stance phase.
5.1.10 Maximum principal stress in meniscus

In the lateral meniscus, the maximum principal stress in the tibiomeniscal contact area (Figure 21 a) was compared between both models. The highest stress (1.6 MPa) was obtained with the FRPE model, and the relative difference between the models was 16.6%, during the beginning of the stance phase. Similarly, in the medial meniscus, the distribution pattern of the maximum principal stress between both meniscus models was similar along the stance phase (Figure 21 b).

![Graph showing the maximum principal stress in meniscus](image)

(a) Lateral meniscus  
(b) Medial meniscus

Figure 21: The maximum principal stress in the FRPE and OTP material model of meniscus as a function of stance phase (a) Lateral meniscus (b) Medial meniscus.
5.1.11 Maximum principal strain in meniscus

The maximum principal strain in the tibiomeniscal contact area was compared in the FRPE and OTP meniscus models on the lateral meniscus and medial meniscus (Figure 22 a, b). In the lateral meniscus, the high relative difference (40%) in the principal strain between models occurred at ~5 % of stance phase. The maximum principal strain distributed subsequently after ~5% of stance phase until end of the gait cycle with relative error of 10% between the models. On the other hand, in the medial meniscus a peak maximum principal strain obtained with FRPE model during 20% of stance phase.

Figure 22: The maximum principal strain occurred in the FRPE and OTP material model of meniscus as a function of stance phase (a) Lateral meniscus (b) Medial meniscus.
Chapter 6

Discussion

The motivation of this study was to propose a simplified material representation of meniscus to enhance the computational efficiency of computational models of the knee joint for clinical applications. For this purpose, OTP constitutive model was optimized and implemented into a 3D knee joint finite element model. In this study, a proposed OTP meniscus model was studied to reproduce the similar reaction force, the stress, strain, pore pressure and fibril strain on the tibial cartilage during normal gait cycle to the FRPE meniscus model. Indeed, the obtained result showed that OTP model was able to mimic the reaction force, stress, strain, fibril strain and pore pressure approximately in both medial and lateral tibial cartilage along the gait cycle within a shorter pre-processing step time. The study showed that OTP model could be a good choice to represent the mechanical behavior of meniscus during numerical simulations of the knee joint.

The magnitude and the distribution of the total reaction force on the tibial cartilage throughout the gait cycle was similar between OTP and FRPE meniscus models (Figure 11). The peak reaction forces on tibial cartilage were same using both meniscus material models during 20% and 80% of stance phase. The peak reaction force obtained in our study (result) was similar to the reported values [119–122].

The distribution of the reaction force through the menisci were similar between OTP and FRPE models throughout the stance phase of the gait. However, there was a difference in the peak force (maximum percentage difference, 21.29%) especially in the lateral compartment between the models at ~5% of stance phase (Figure 13). This might be due to the difference in cartilage-cartilage and cartilage-meniscus contact area between FRPE and OTP meniscus models which was clearly observed during the early loading response (~5 of fraction of stance phase) (Figure 18 -19, a). The
cartilage-cartilage contact area was found larger in OTP model than in FRPE model in the lateral compartment during 5% of stance phase (Figure 18a) so that cartilage might support large amount of force.

The deformation of the FRPE and OTP material meniscus in x-axis, y-axis and z-axis was evaluated at ~5% fraction of the stance phase which was equal to 0.029 s of gait cycle. The only difference in deformations was found along the z-axis. It was noticed that in OTP model, posterior horn displaced more laterally compared to FRPE model. This might be one reason behind why there was substantial differences between the models in all observed parameters at the tibiofemoral contact area at ~5% fraction of the stance phase.

The response of meniscus to the applied forces and moments depends on their attachments on the tibial bone, its geometry and microstructure. Generally, the lateral meniscus has more posterior translation than the medial meniscus due to the femoral condyle in knee joint flexion [67]. In our study, it was seen that the lateral meniscus was more responsible to carry the axial load than its counterpart medial meniscus during the normal gait cycle. The lateral meniscus bears the maximum load from the beginning to the terminal stance phase. The maximum posterior translation of lateral meniscus occurred at the beginning of loading stance phase followed by the heel strike (~ 5% stance phase). Therefore, most of the force was absorbed by the lateral meniscus at ~ 5% stance phase of gait cycle. Similarly, the second highest reaction force on the lateral meniscus was noticed at the 50% of stance phase, due to the posterior translation of the lateral meniscus, by the femur. Finally, minimum posterior translation of the lateral meniscus was seen at the 75% of stance. This caused decrease in reaction forces through lateral meniscus.

The tensile stiffness of the lateral meniscus is higher than the medial meniscus and it can bear hoop tension when it is compressed radially by the femoral condyles [123–125]. This might be due to tissue adaptation by the loading caused by daily physical activities. In our study, lateral meniscus distributed substantially higher loads compared to medial meniscus, which may be the reason why higher stiffness is found in lateral meniscus compared to medial meniscus. The medial meniscus carried only small fraction of those loads observed in lateral meniscus during entire stance phase of the gait. The difference in the distribution of the reaction force between the lateral and the medial meniscus in our study is consistent to the biomechanics of the meniscus in loading distribution
during gait cycle as mentioned in the studies [67,123] and the tibiofemoral kinematics and condylar motion, femorotibial contact patterns described in other studies [122,126,127].

On the medial side, the principal stress, strain, pore pressure and fibril strain on the cartilage (cartilage-cartilage contact area) were same between FRPE and OTP meniscus models throughout the gait cycle (Figure 14 - 17, b). Their values were high on the medial tibial cartilage than on the lateral tibial cartilage because higher load passes through the medial meniscus than lateral during walking, which is similar in other studies [128,129]. Similarly, high principal stress was measured during 80% of stance phase, though, a high reaction force (1400N) acted on medial tibial cartilage. It might be due to the ~30% less cartilage-cartilage contact area was measured at 80% of stance phase than 20 % of stance phase along the gait cycle. Hence, the relatively small cartilage-cartilage contact area at 80% of stance phase supporting the force and found under higher stress during 80% of stance phase than at ~20% of stance phase (Figure 18b). On the medial tibial cartilage, the pore pressure was increased by 27% at ~80% of stance phase (Figure 17b). The occurrence of high pore pressure during 80% of stance phase was due to the short time (0.44 s) available for the fluid to be squeezed out from cartilage matrix. In our simulation, the total time (0.55 s) available for the fluid to be squeezed out from cartilage matrix was found too short as compared to the total time needed for the fluid coming out from matrix which was observed in study [130]. Thus, most of the fluid will be retained inside the cartilage matrix during walking which results in increasing the pore pressure at the terminal (around 80% of stance) phase. On the medial tibial cartilage, the maximum principal strain and the fibril strain was increased by 33% during 80% of stance phase from their value measured at ~20% of stance phase (Figure 15b and 16b). The high fluid pressure expand the cartilage tissue horizontally and vertically results in increasing the strain [33]. The high joint reaction force results high stress and strain on the medial tibial cartilage in our study is consistent with earlier studies [10,131].

On the lateral side, the distribution of the stress, strain, pore pressure and the fibril strain were uneven at ~5% of stance phase (Figure 14-17, a). It might be due to the early described cartilage-cartilage contact area (Figure 18a) difference in the displacement of the meniscus (Figure 20). On the lateral side, the inclusion of the varus-valgus moment in our model might cause the discontinuity in the cartilage-cartilage contact from approximately mid stance to the terminal stance results the zero stress, strain, pore pressure and fibril strain on the lateral tibial cartilage. However,
similar behaviour of joint reaction forces is shown by the previous experimental measurement utilizing motion analysis[120]

This study has a few limitations. First, the tissue geometries and gait input data were taken from an earlier study, which consist of only one healthy male subject [35]. However, this is a methodological study, where the use of only one subject does not affect in the study as the basic material function of the menisci will be same. Second, the reaction force, maximum principal stress, maximum principal strain, and pore pressure in the lateral and medial tibial cartilage would vary person to person in their magnitude, but the differences between the models would remain small as presented in the current study. Third, the knee model was simulated only by the loading from walking, which is the most common typical type of movement. However, this approach can be easily applied for other daily motor task such as stair climbing, sit-to-stand, and squatting. Finally, other limitation of this study was the lack of direct experimentally validation of the OTP material properties of the meniscus and as far as we know this is the first study where such material properties of the meniscus was implemented in the finite element analysis of the knee joint with real human gait cycle. Experimental characterization of meniscus is needed to validate the adjusted OTP material parameters. This pending work would be part of future efforts in coming studies.
Conclusion

In summary, this study suggests that the OTP material model of meniscus is feasible to incorporate and approximate the contribution of the meniscus during normal gait cycle. In addition, the use of OTP model of meniscus speeds up model generation time as compared to FRPE model of meniscus. It is possible to obtain the joint reaction force as well as the cartilage response to the applied body load for instance, the maximum principal stress, strain, pore pressure and fibril strain in a similar way as the FRPE material model during normal walking. Thus, the use of OTP material model of meniscus allows to simulate the multiple knee joint in daily basis. Indeed, many patients will be benefited with the use of OTP material model in FEA of knee joint. The efficiency of the knee joint simulation can be substantially increased for instance with use of automatic segmentation of tibial and femoral cartilage [132,133] to create the knee model from MRI images and the application of simpler material model of ligaments (spring or solid constitutive) together with the OTP material model of meniscus with gait [118].
References


[38] Venäläinen, M. S., Mononen, M. E., Väänänen, S. P., Jurvelin, J. S., Töyräs, J., Virén, T.,


Mow, V. C., and Huiskes, R., 2005, Basic Orthopaedic Biomechanics & Mechano-Biology, Lippincott Williams & Wilkins.


Masouros, S. D., Bull, A. M. J., and Amis, A. A., 2010, “(I) Biomechanics of the Knee...


MONONEN, M., “Computational Modeling of Knee Joint Mechanics Under Impact and
Gait Cycle Loading.”


