Does anterior cruciate ligament rupture lead to changes in knee joint mechanics distribution? A single framework for the knee finite element musculoskeletal study

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Introduction

Anterior Cruciate Ligament (ACL) rupture is the commonly encounter form of knee rupture faced by athletics participating in high demand sports. The mechanical properties of ligaments during movement are difficult to measure directly, making computer modeling methods a viable alternative for predicting ligament mechanics. However, simplistic joint models cannot accurately simulate real movements as they do not account for muscle activity. There is an interplay between muscle activity and joint motion in vivo. To address this, we have developed a musculoskeletal model that includes joint soft tissues, ligaments, and muscles to infer the mechanical properties of ligaments during movement.

Methods

The study utilized a finite element musculoskeletal (FEMS) model of the right lower limb, including a healthy knee, which had been previously developed and analyzed using FEBio software (MRL Utah University, Salt Lake, UT, USA). The model's construction and validation were discussed in previous works by Wang et al. [1,2]. The FEMS model consisted of the geometries of all the bones in the right lower limb, as well as the articular cartilages and meniscus of the knee. These structures were manually segmented and reconstructed from computed tomography and magnetic resonance imaging scans.

Various joints were included in the model: a spherical joint with three degrees of freedom represented the hip, joints with six degrees of freedom represented the tibiofemoral and patellofemoral joints, and a hinge joint with one degree of freedom represented the ankle joint.

The ligaments were modeled as nonlinear spring bundles. The force-strain relationship for each ligament was described by Equation (1), where the force sustained by the ligament depended on its current length, slack length, strain, and stiffness coefficients [3].

$$f_i = \begin{cases} 0 & \varepsilon_i \leq 0 \\ {}^1k_i(l_i - {}^0l_i)^2 & 0 < \varepsilon_i \leq 2{}^l\varepsilon \\ {}^2k_i[l_i - (1 + {}^l\varepsilon){}^0l_i] & 2{}^l\varepsilon < \varepsilon_i \end{cases} \quad (1)$$

where f_i is the force sustained by the ith ligament, l_i is the current length of the ith ligament, ${}^0l_i$ is the slack length of the ith ligament, ${}^l\varepsilon$ is the strain assumed to be constant at 0.03, and ${}^1k_i$ and ${}^2k_i$ are the stiffness coefficients of the spring elements representing the ith ligament for the nonlinear and the linear regions, respectively.

The FEMS model also included 20 representative one-dimensional linear element muscles. Muscles wrapping around the femoral and tibial bones was considered to represent their paths. The mechanical properties of the muscles were represented using a Hill-type dynamic model, which incorporated both active and passive elements.

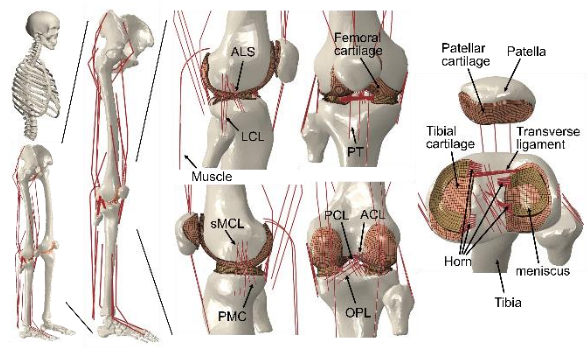

Figure 1: A musculoskeletal model containing a lower limb finite element musculoskeletal model includes a 12-DOF knee joint

Results and Discussion

The interaction between the muscle level and joint level that exists in vivo is an important relationship that influences the biomechanics of the musculoskeletal system and can only be described in the concurrent framework. Our model allows for accurate inference of the mechanical properties of ligaments during movement.

Acknowledgments

Research supported by Japan Society for the Promotion of Science [Grant-in-Aid for Research Activity Start-up 22K20503].

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