1. Preliminary Material

1.1 Introduction

Stability of the knee is instrumental during basic activities of daily living and recreation, such as descending stairs, running, and jumping. To accommodate this need for stability without inhibiting overall motion of the individual, the knee incorporates ligaments as soft tissue constraints [7]. Ligaments derive their mechanical properties primarily from biochemical components, such as collagen fibers [29]. An injury to one or more of the knee ligaments permits undesired relative motion between the tibia and femur. Therefore, a primary goal in the treatment of an injury of this nature is the restoration of intact knee kinematics [13].

The anterior cruciate ligament (ACL), which limits anterior displacement and internal rotation of the tibia [8,9,15,16,22,30,32,34], is the most commonly injured knee ligament [18,20]. Unlike extrasynovial ligaments, an injured ACL does not respond well to primary repair [4,14,28,36]. Hence, a ruptured ACL may require reconstruction with a substitute structure [6]. A bone-patellar tendon-bone autograft is arguably the most common structure used for ACL reconstruction [6,19]. During the procedure, the selected graft is fixed under tension between the femur and tibia in place of the ruptured ACL.

Before implantation, patellar tendon exhibits mechanical, biochemical, and histological properties dissimilar to the ACL. The stiffness and ultimate load of the graft complex are two to three times greater than that of the femur-ACL-tibia complex [12,25,26,35]. Patellar tendon contains greater than 95% type I and only traces of type III collagen. The ACL, on the other hand, contains 10%-14% type III, in addition to type I collagen [2]. Fibrillar crimp and fibroblast morphology also differ between patellar tendon and ACL tissues [1].

After implantation, the graft undergoes a remodeling process [2,3,11,27,31]. Within days of reconstruction, the graft experiences a period of avascular necrosis, followed by cellular proliferation. Eventually, the fibroblast morphology and collagen content of the tissue approach those of the ACL. However, graft stiffness and ultimate load significantly decrease immediately upon implantation [5,23,24,33]. Although there is some recovery over time, graft stiffness and ultimate load reach levels of only about 25% of those of the femur-ACL-tibia complex. Joint laxity remains, and the long-term results of ACL reconstruction fall short of ideal.

Specific reasons for mediocre results following ACL reconstruction are not known. One factor that may influence the outcome is graft pretension, which significantly affects joint kinematics [10]. A graft pretension may be selected to best restore intact knee kinematics [17]. However, because of their viscoelastic nature, grafts undergo stress relaxation, and knee kinematics may thus change over time. In fact, all grafts may relax to the same tension level regardless of initial setting [21], although researchers have not adequately demonstrated this behavior.

Graft pretension also significantly affects the tissue remodeling process. After a 3 month recovery period, Yoshiya et al. [37] found focal degeneration in grafts pretensioned to 39 N, but the authors noted no apparent damage in grafts pretensioned set to 1 N. Collagen fibers in the higher tensioned graft were replaced by a myxoid extracellular matrix. Therefore, it may be desirable to place a greater influence on pretension selection to avoid damaging the graft rather than kinematic considerations, especially if all grafts relax to the same tension.

1.2 Statement of the Problem

The purpose of this dissertation was to study the effects of graft pretension in ACL reconstruction. To achieve this goal, a series of experiments were designed using both in vitro and in vivo models. First, an in vitro study of human knee kinematics was conducted to determine if intact knee kinematics could be restored when using the ideal graft—the intrinsic ACL. Second, the in vitro kinematics of human and porcine knees were compared and contrasted. Third, the mechanical characteristics of porcine patellar tendons, the graft used in the in vivo model, were investigated. Fourth, in vivo ACL reconstructions were performed in a porcine model using a specially designed load cell/telemetry system to monitor graft load following surgery.

1.3 Organization

Following this introductory chapter, a review of literature pertinent to the structure and function of the knee is presented in Chapter 2. Two in vitro studies of knee kinematics are detailed in Chapter 3. The mechanical characteristics of porcine patellar tendon are described in Chapter 4. An in vivo study of ACL reconstruction in a porcine model is detailed in Chapter 5. The conclusions from this collection of work are summarized in Chapter 6. Finally, literature supplemental to studies described in Chapters 3 and 4 is offered in Appendix A.