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Sex-based differences in the tensile properties of the human anterior cruciate ligament

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Abstract

After immense amounts of research, the root cause for the significantly higher rates of anterior cruciate ligament (ACL) failure incidents in females as compared to males still remains unknown and the existing sex-based disparity has not diminished. To date, the possibility that the female ACL is mechanically weaker than the male ACL has not been directly investigated. Although it has been established in the literature that the female ACL is smaller in size, the differences in the structural and material properties of the ACL between sexes have not been studied. The aim of this cadaveric study was to determine if any sex-based differences in the tensile properties of the human ACL exist when considering age as well as ACL and body anthropometric measurements as covariates. Ten male and 10 female unpaired cadaveric knees (mean age 36.75 years) were used for this study. The geometry of the ACL (including length, cross-sectional area, and volume) was analyzed using a 3-D scanning system. The femur-ACL tibia complex was tested to failure along the longitudinal axis of the ligament in a tensile testing machine. The structural properties of the ACL as well as its mechanical properties were determined. Analysis of covariance was performed to assess the effect of sex on tensile properties. The female ACL was found to have a lower mechanical properties (8.3% lower strain at failure; 14.3% lower stress at failure, 9.43% lower strain energy density at failure, and most importantly, 22.49% lower modulus of elasticity) when considering age, ACL, and body anthropometric measurements as covariates.

Keywords: Sex; ACL injury; Mechanical properties; ACL size; Cadaveric

1. Introduction

Risk factors explaining the increased incidence of anterior cruciate ligament (ACL) injury in female athletes are commonly classified as anatomic (Ireland, 2002), neuromuscular (Anderson et al., 2001), and hormonal (Liu et al., 1996; Slauterbeck et al., 1997). Among the reported anatomic risk factors, differences in ACL size (length, cross-sectional area, and volume) have been reported in a cadaveric study using a 3-D imaging system by Chandrashekar et al. (2005); in an in vivo study using MRI measurements by Anderson et al. (2001); and in a cadaveric study using a molding technique by Muneta et al. (1997). All three studies report on a smaller female ACL size. Assuming that the ACL's tensile properties are independent of variables such as sex, race, or body size, it is reasonable to believe that smaller ligaments (lower length, area, and/or volume) are more prone to injury when compared to larger ligaments while subjected to similar load levels.

Sex-based differences in material properties of the human ACL have not been investigated. To date,

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studies reporting on the material properties of the human ACL have assumed the properties to be similar between the sexes. The two landmark papers that report on the mechanical properties of the human ACL (Noyes and Grood, 1976; Woo et al., 1991) used a sample population that included both male and female specimens. If sex differences exist in the mechanical properties of the human ACL, then failure characteristics between the sexes could be very different. This could provide a core factor (or root cause) predisposing females involved in sports activities to injury.

We propose that the core factor in determining why ACL injury incidence is higher in females is that the female ACL has lower material properties. Thus, during non-contact circumstances such as a "plant and cut", the stresses or strains experienced by the ACL are high enough to exceed critical levels in females but not high enough to exceed the critical levels in males of similar stature. We therefore hypothesize that the female ACL is of lower mechanical and structural properties compared to the male ACL when covariates such as age, body and ACL anthropometric measurements are considered in the analysis.

2. Methods

Twenty unpaired knees (10 male and 10 female) were obtained from donors within 12 hours after death and were frozen with all tissue intact at $-20\,^{\circ}C$ until the day of the experiment. The primary causes of death for the majority of donors were acute illness or trauma; none were excessively obese or thin. Knees with any signs of abnormality were excluded from this research. The donor anthropometry data are given in Table 1. On the day of the experiment, the knees were thawed and all soft tissues other than the ACL were dissected. The tibia and femur were cut approximately 3 in from the attachment sites of the ACL. The medial and lateral portions of the bones were cut parallel to the longitudinal axis of the ACL (Fig. 1a), to isolate the ACL with bone plugs.

A photographic 3-D scanner was used to create the 3-D model of the ACL. This 3-D model was exported to the 3-D Doctor software (Able Software Corp., USA) and the minimum cross-sectional area and volume of the ACL were found (Hashemi et al., 2005a). The lengths of anteromedial and posterolateral bundles of the ACL were measured using a vernier caliper. The averages of the two lengths represented the length of the ACL (Noyes and Grood, 1976).

The bone plugs were potted in aluminum cylinders of 2.5 in internal diameter using a low melting point alloy, Cerrobend. Care was taken to assure that the molten metal did not come into contact with the ACL by wrapping the ACL with saline-soaked gauze. To prevent thermal injury to the ACL, the potted Femur-ACL-Tibia complex (FATC) was rapidly cooled in a freezer for 5 min. The FATC was then mounted on a custom made jig fitted to a servohydraulic tensile testing machine (Instron 8500+). The ACL was aligned along its longitudinal or ligamental axis. To facilitate this, the tibia was oriented at 45° to the loading axis of the ACL (Fig. 1c) since the angle of insertion of the ACL into the tibia is approximately 45°. The femur was oriented at 45° (Noyes and Grood, 1976) to the tibial axis (Fig. 1b, c). The orientation of the FATC is now similar to the one used by Woo et al. (1991) except that the flexion angle is 45° with the coronal rotation (Fig. 1b) being implemented through cutting direction (Fig. 1a).

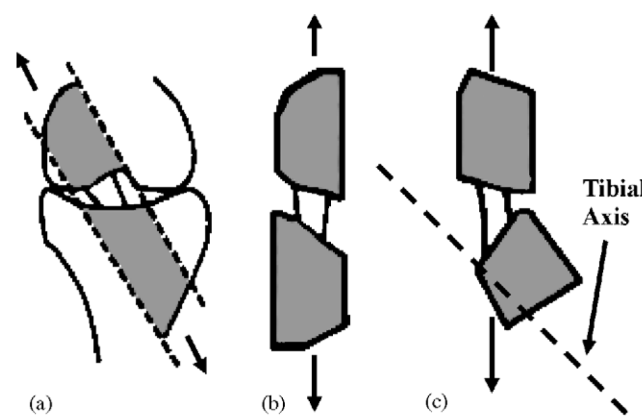

Fig. 1. Schematics showing the anatomic orientation of the ACL with the loading axis. Arrows represent the loading direction (ACL axis). (a) The direction of cut on the femoral condyles and tibial bone shown by dotted lines. (b) The frontal view of the ACL orientation: the bone plugs are rotated in the coronal plane to ensure anatomic orientation. (c) Saggittal view of the ACL orientation: Tibial bone plug oriented 45° to loading direction and femoral bone plug flexed 45° to the tibial axis.

Table 1
The mean age and body anthropometry data of donors

	Age (years)	Body mass (kg)	Height (cm)	Body mass index (kg/m^2)
Male	39 (26–50)	78.07 (54–121.5)	177.01 (167.64–182.88)	24.82 (19.21–37.35)
Female	37.7 (17–50)	70.38 (49.5–111.6)	164.4 (152.4–175.26)	26.13 (19.95–44.42)

The range is also given in parenthesis.