INTRODUCTION

It has been reported that the anterior cruciate ligament (ACL) may be injured due to its impingement against the intercondylar notch [1]. To determine the possible role of impingement before and after ACL injuries, it is necessary to quantify the interaction between the intercondylar notch and the ACL. This knowledge could be useful to estimate the tension in the ACL during the impingement and provide better understanding of the mechanisms of ACL injury and effects of ACL reconstruction on grafts impingement. In this study, the impingement of the ACL against the intercondylar notch under axial tibial weightbearing was investigated.

METHODS

Nine subjects, 4 males - 5 females, aged 23 to 48 years old, with healthy knees were recruited under the guidance of IRB. Consent forms were collected from all subjects. Each knee (chosen randomly, 5 right and 4 left knees) was scanned in a relaxed, extended position using a 3.0 Tesla MR Scanner in both sagittal and coronal planes based on an existing protocol established in our laboratory [2]. The 3D anatomic models of the tibia and femur were created using these MR images. The insertion sites of the ACL on the MR images were outlined on both the tibia and femur to determine the tibial and femoral insertion areas on the 3D anatomic bone models. The anteromedial (AM) and posterolateral (PL) bundles were defined on each end of the ACL on the tibia and femur.

The same knee was then imaged at four different flexion angles (0°, 15°, 30° and 45°) using the dual-orthogonal fluoroscopy technique [2] under two different in-vivo axial tibial loading: (i) minimum load <10N, (ii) full body weight (BW). Using a force plate, the value of the load applied to the subject's leg was recorded.

The 3D knee model and the fluoroscopic images were used to reproduce the in-vivo positions of the tibia and femur using a solid modeling software. At each position, a 3D volume model of the ACL was created using the ACL insertion areas on the tibia and femur. The impingement of the ACL was defined as the penetration of the surface of the ACL through the 3D surface of the femur (Fig. 1). Next, the location of the maximum impingement (t) was determined and the value of maximum impingement was measured. At the location of maximum impingement, the impingement ratio was defined as the ratio of maximum impingement (t) to the diameter of the ACL (D).

φ was defined and measured as the angle between the vertical (anterior-posterior) axis in the notch view and the axis connecting the origin of the clock to the location of maximum impingement in the notch view (Fig. 2).

RESULTS

No ACL impingement was observed at 30° and 45° of flexion. However, ACL had anterior impingement against the femoral intercondylar notch at 0° and 15° of flexion, both under minimum and full BW (Fig. 3).

Maximum impingement (t):

At full extension, the maximum impingement was 1.7 ± 0.7 mm under minimum tibial load (<10N), and increased to 2.1 ± 0.9 mm at full weight bearing (Fig. 3-A). However, the maximum impingement against the femoral notch decreased at 15°. It was measured 0.7 ± 0.3 mm at minimum load and 0.9 ± 0.3 mm at full BW.

The ratio of impingement (t/D) was about 30% at full extension and about 15% at 15° of flexion (Fig. 3-B). By applying full BW, the impingement ratio increased from 26.6 ± 9.0% (min. load) to 32.5 ± 9.4% (full BW) at full extension, and from 10.9 ± 6.7% to 15.3 ± 6.1% at 15° of flexion.

Angle φ:

At full extension, the location of the maximum impingement with respect to the clock coordinates in the notch view (angle φ, fig. 2) was 32.4° ± 9.4° at minimum load and decreased to 30.5° ± 9.4° under full BW (Fig. 3-C). At 15° of flexion, angle φ was 46.0° ± 10.0° and 42.7° ± 10.6° under minimum load and full BW, respectively. This means that the loading changed the location of the maximum impingement medially (towards the center of the notch) at both 0° and 15° (about 1.9° and 3.3° respectively).

DISCUSSION

Since the impingement ratio was greater at full extension and the medial shift was smaller compared to 15°, it might imply that the ACL is tighter and likely under more tension under full body weight at full extension compared to 15° of flexion.

If an optimal graft material was designed which would replicate the exact mechanical function of the native ACL then, based on our present findings, a notchplasty can be avoided. In this manner the normal in-vivo impingement with its inherent stability could be maintained. However, with current techniques, reshaping of the notch is still advisable so that the graft fixation is protected from excessive impingement at lower flexion angles. In addition, the rehabilitation under weightbearing at full extension should be delayed, to avoid extra tension on the graft.

Future studies should focus on quantifying of impingement force of the ACL using either in-vitro experiments or 3D finite element modeling.

REFERENCES