Anterior Cruciate Ligament Fibres – Effects of Tibial Translation During Flexion at the Knee

Ahmed Imran, *Member, IAENG*

Abstract— **Anterior Cruciate Ligament (or ACL) of the knee is frequently injured during activities involving large relative translations of the bones. Contributions of the knee ligaments in stabilizing the joint have been studied. Separately, tightening or slackening of the ACL fibres during flexion has also been analyzed. However, there is a need to understand the role of different fibre bundles in the ligament during activities. In the present study, a planar mathematical model of the knee is used to analyze the patterns of changes in the ACL fibres during passive motion and because of anterior tibial translation, which stretch the ligament fibres.**

The joint ligaments were modeled as bundles of non-linear elastic fibres. The bones were frictionless and impenetrable. Anatomical data and material properties were obtained from literature. An anterior laxity test was simulated.

During passive flexion over 0-120^o range, the most anterior fibre of ACL remained isometric while other fibres slackened. Tibial translation during the simulated test stretched the fibres variably over the flexion range. For example, at 45^o flexion, 130N anterior load on the tibia resulted in 6.1 mm tibial translation and stretched the anterior fibre by 11.5% and the intermediate fibre by nearly 1.2% while the posterior fibre remained slack.

The model calculations showed reasonable agreement with the experimental observations from literature. The analysis suggests that injuries can affect different fibres of the ligament depending on the relative positions of bones at the time of injury. Such effects may be clinically significant and may help in explaining complex functional behavior of the ligament during activity and injury.

Index Terms— **knee biomechanics, biomechanics of ligament injury, anterior cruciate ligament, ACL fibre length.**

I. Introduction

THE anterior cruciate ligament (ACL) of the knee is frequently injured, partially or fully, particularly during activities involving large anterior translations of the lower bone or tibia relative to the upper bone or femur. ACL is the primary restrain to anterior tibial translation (ATT) relative to the femur [1–4]. In clinical practice, the ligament integrity is estimated using an anterior laxity test in which ATT is measured corresponding to an applied external load while the joint angle is maintained [5].

Further, contributions of the knee ligaments in stabilizing the intact or replaced joint have been studied in experiments on cadaver knees [1–3, 6] and in model analysis [7] by recording changes in relative movements of the bones in response to externally applied loads while the ligaments were resected sequentially. Separately, investigators have analyzed tightening or slackening patterns of the ligament fibre bundles during flexion [3, 4, 8, 9]. The ACL shows a complex functional behavior mainly resulting from variations in material properties and geometric patterns of stretching and slackening of different fibre bundles during motion, with influence on the joint mechanics [1–4, 6–10]. The ligament fibres stretch or slacken depending on the relative positions of their attachments on the two bones [3, 8, 9]. Therefore, more detailed analyses are needed to understand the role of ACL in the knee mechanics during activity as well as in understanding the mechanics during ligament injury and after replacement.

In the present study, a planar mathematical model of the knee is used to analyze the patterns of changes in lengths of different fibres of the ACL during passive flexion and after tibial translation at several flexion positions of the joint.

II. Methods

The knee joint was modeled in the sagittal plane with intact cruciate and collateral ligaments represented as bundles of non-linear elastic fibres [11]. A flat tibial surface and a compatible polycentric, polyradial femoral surface were used assuming the articulation to be frictionless and the bones to be impenetrable [12]. Passive motion of the joint in the absence of any muscle force or external load was defined during 0-120^o flexion at 15^o interval such that selected fibres in the cruciate ligaments remained isometric while no other fibre in any ligament stretched [11, 12].

Distance between the tibial and femoral attachments of a ligament fibre gave its length. Reference lengths of the fibres were defined at 0^o flexion, where all the ACL fibres were just taut (without any stretch or slackness). Changes in the fibre lengths were calculated as percentage of their reference lengths for the anterior, intermediate and posterior fibres.

An anterior laxity test was simulated at each flexion position by applying a known anterior force on the tibia along with a balancing moment to maintain the joint angle fixed during the test [5]. ATT resulting from the applied load were recorded and compared with *in vitro* experimental observations of Lo *et al* [6].

Anatomical data for attachments of the ligament fibres, position and tilt of the tibial surface as well as material properties of the ligaments fibre bundles were taken from literature [11, 12].

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A. Imran is with the Department of Biomedical Engineering, Ajman University of Science & Technology, Ajman, U.A.E. (phone: ++971-50-2850131; fax: 971-6-7438888; e-mail: ai_imran@yahoo.com or ajac.ai_imran@ajman.ac.ae).

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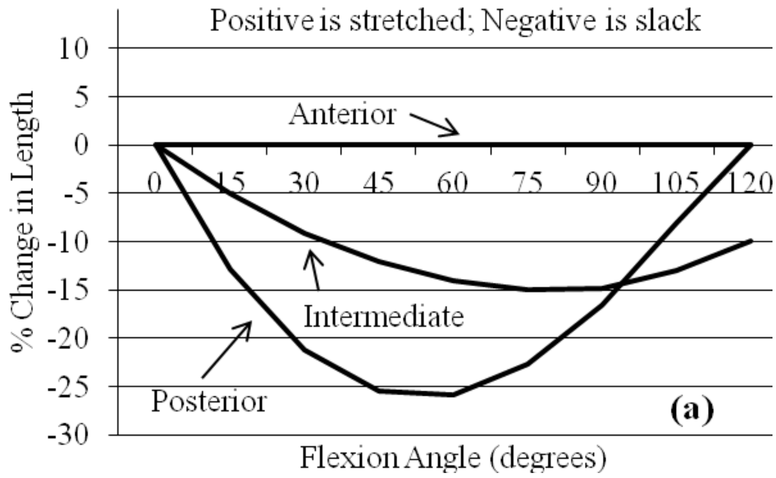

Fig. 1(a). Change in length (as percentage of reference length) over the flexion range calculated for the three fibres of ACL during passive flexion.

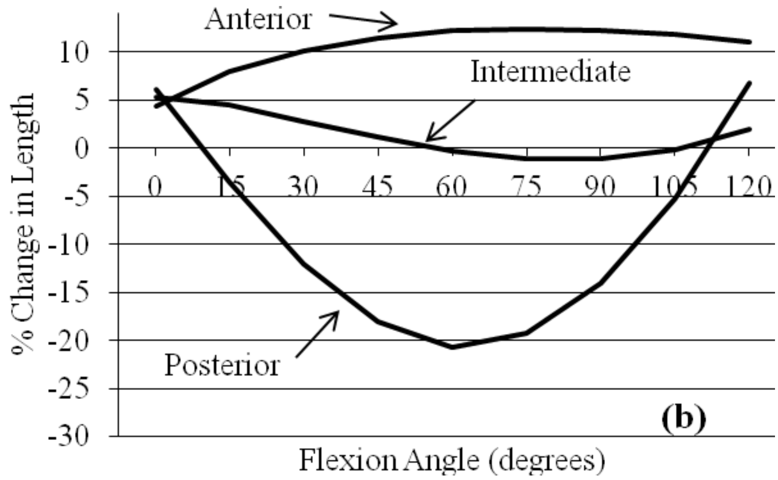

Fig. 1(b). Change in length (as percentage of reference length) over the flexion range calculated for the three fibres of ACL after ATT due to 130N anterior laxity test.

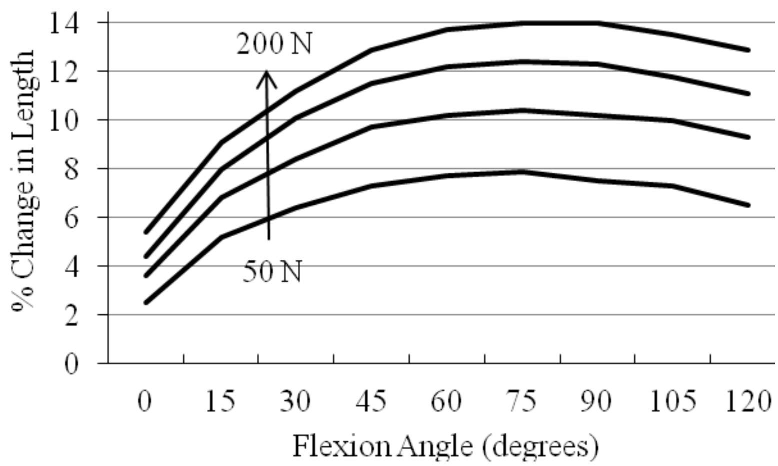

Fig. 2. Change in length (as percentage of reference length) over the flexion range calculated for the anterior fibre of the ligament after ATT due to 50, 100, 150 and 200N anterior laxity test.

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III. Results

Table 1 gives ATT resulting from 130 N anterior laxity test simulated at several flexion positions of the joint. The model calculations are compared with the experimental observations of Lo *et al* [6] on 14 intact cadaver knees given here as mean values and standard deviations.

Figure 1 gives calculated results for % change in lengths of the ACL anterior, intermediate and posterior fibres plotted over 0-120^o flexion range during passive flexion (figure 1(a)) and after ATT due to 130N laxity test (figure 1(b)).

Figure 2 gives calculated results for % change in lengths of the ACL anterior fibre due to ATT following the laxity test with 50, 100, 150 and 200N anterior force on the tibia.

Table 1. Comparison between model calculations and experimental measurements (Lo *et al*. [6]). ATT due to 130N anterior laxity test is given for different flexion positions of the joint.

Flexion angle (Degrees)	**Anterior Tibial Translation (ATT) (mm)**	
	Model Calculations	**Experiment [6] Mean (SD)**
0	3.4	4.1 (0.6)
15	5.3	6.4 (1.3)
30	6.0	7.5 (1.8)
45	6.1	7.9 (2.2)
60	5.9	7.4 (2.2)
75	5.6	6.5 (2.1)
90	5.3	6.2 (1.9)
105	4.8	--
120	4.3	--

IV. Analysis

From table 1, ATT calculated from the model showed reasonable agreement with the experimental measurements of Lo *et al* [6]. For all flexion positions, the calculated values are close to the mean experimental values and are within the reported standard deviation. From both, the model and experiment, ATT first increased for 0–45^o and then decreased for higher flexion.

Figure 1(a) for passive flexion shows that the anterior fibre in the ACL maintained its length over flexion, while the intermediate and posterior fibres remained slack. Further, for the intermediate fibre, slackness increased for 0–75^o range and then decreased, while for the posterior fibre, slackness increased for 0–60^o and then decreased to its reference length at 120^o. In an experimental study on cadaver knees with 0^o flexion as reference, Amis [3] reported that the antero-medial bundle of the ACL remained slack for nearly 70^o and then stretched in higher flexion, while the intermediate and postero-lateral bundles of the ligament remained slack throughout flexion. These patterns of variation in the ligament fibre lengths agree with the model calculations except for the anterior fibre in higher flexion.

Figure 1(b) shows that ATT from 130N anterior test resulted in increased stretch for all the fibres near extension and in high flexion. The anterior fibre was stretched for all flexion angles, while the intermediate fibre remained stretched for 0–60^o range, slackened, and then stretched again in higher flexion. The posterior fibre stretched only at 0 and 120^o. Maximum ATT was achieved at 45^o flexion as 6.1 mm from the model calculations and 7.9 mm (SD =2.2) from the experiment. With this level of ATT, the model anterior fibre stretched by 11.5% and the intermediate fibre by 1.2%, while the posterior fibre remained slack. For the human cruciate ligament tissue, reported strain at failure is around 20% [1, 13] or 27% [8].

Figure 2 shows that stretch in the anterior fibre of the ACL increased with ATT as the external anterior load was increased. An increase in the load from 50 to 200N at 0^o flexion resulted in change in length from 2.5 to 5.4% , with a difference of 2.9%. This difference increased progressively with flexion angle and reached to 6.4% at 120^o flexion. This observation suggests that the effect of increment in load on the fibre length increases with flexion angle.

V. Conclusion

The model calculations showed reasonable agreement with experimental measurements. During passive flexion of the joint, anterior fibre remained isometric while the other fibres remained slack with reference to 0^o flexion. Tibial translation due to the simulated laxity test stretched the fibres variably over the flexion range. The anterior fibre achieved the highest stretch while the posterior fibre achieved the highest slackness.

The patterns of variation shown by different fibres of the ligament with flexion and with tibial translation suggest that injuries can affect different fibres depending on the position of the bones at the time of injury. Analysis of such effects is clinically important in order to understand complex functional behavior of the ligament and the mechanics of injury.

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