Passive Anterior-Posterior Knee Stability After Unconstrained Unicompartmental Arthroplasty

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***Abstract*—These Passive anterior-posterior (A-P) stability of the knee, measured in terms of joint laxity, is considered important for good clinical outcome following knee arthroplasty. *In vitro* and *in vivo* studies measured the laxity at selected joint positions in the intact and replaced knees. However, analyzing the effects of surgical techniques on the joint stability is difficult to implement experimentally.**

In the present study a mathematical model of the knee with unconstrained unicompartmental arthroplasty was used to study relative translations of the bones during a simulated A-P laxity test over 0-120^o flexion. The knee ligaments were modeled as bundles of non-linear elastic fibers. Anatomical data, material properties of ligaments, geometries of the prosthetic components and guidelines for component placement on the bones were taken from literature. The model calculations for tibial translations resulting from ±130 N A-P forces were compared with the experimental measurements of Lo *et al*. (2010), reported as mean of 14 cadaver knees. Further, the effects of component placement on the bones were also studied.

The model calculations agreed in general with the experimental measurements showing similar patterns during flexion. The joint laxity first increased from 0^o to about 45^o flexion and decreased thereafter. An increase in the A-P force resulted in uniform increase in laxity over flexion. A change of 1 mm in the placement of femoral component affected the laxity by nearly 3 mm near extension. However, this effect of change varied significantly with flexion. Such effects can alter the joint kinematics and may be clinically significant. The analysis has clinical relevance .

***Index Terms*—Component Placement, Knee Mechanics, Knee Stability, Unicompartmental Knee Arthroplasty.**

I. Introduction

PASSIVE A-P stability of the knee is estimated from laxity tests where known magnitudes of A-P forces are applied on the tibia at fixed flexion angles and corresponding tibial translations are recorded. Such translations of the tibia in the absence active muscle forces are considered important clinical measure to assess knee function after joint replacement [1], [2]. In the sagittal plane, primary restrain to anterior tibial translation is the anterior cruciate ligament, while primary restrain to the posterior tibial translation is the posterior cruciate ligament. Unicompartmental arthroplasty with unconstrained prosthetic surfaces depends on the knee ligaments for A-P stability [3]. Therefore, appropriate ligament balancing during surgery can play significant role in determining kinematics of the replaced knee [1], [3]. Incorrect placement of the prosthetic components can affect the ligament lengths and result in altered kinematics of the joint.

In vitro and *in vivo* experimental studies have analyzed the passive knee stability in the intact and replaced knees at selected flexion angles and external loads [4], [5]. However, analyzing the effects of surgical techniques such as placement of the prosthetic components is difficult to implement experimentally. An understanding of these effects, particular for minimally invasive surgery, is important as they can influence patient outcome [6]–[9]. Theoretical models validated with the experimental observations are useful tools to study such effects [9], [10].

In the present study a sagittal plane mathematical model of the knee is used to analyze the passive stability of the joint in terms of anterior and posterior translations of the tibia after unicompartmental replacement with intact ligaments. The model is also used to analyze the effects on the A-P translations resulting from incorrect placement of the prosthetic components.

II. Methods

The knee with intact cruciate and collateral ligaments and unconstrained prosthetic components was modeled in the sagittal plane. The ligaments were modeled as bundles of non-linear elastic fibers [11]. A flat tibial and a circular femoral component were used. Anatomical data, attachments of the ligaments and their material properties as well as geometries of the prosthetic components and surgical guidelines for their attachment on the bones were taken from literature [3], [11], [12]. The components were attached to the bones such that no fiber of any ligament was significantly stretched at 0 or 90^o flexions. This position of the prosthetic components was defined as the *correct position* (or *correct placement*). The knee motion was defined during 0–120^o flexion at 5^o interval.

At each flexion position, an A-P laxity test was simulated by applying a known anterior (+) or posterior (-) force on the tibia while the femur was held fixed and the resulting tibial translations were noted. The model calculations were compared with *in vitro* experimental measurements from Lo *et. al.* [4]. Further, the femoral component was positioned 1 mm proximal and 1 mm distal to its *correct position*. The A-P simulation was repeated and total A-P translations of the tibia were noted.

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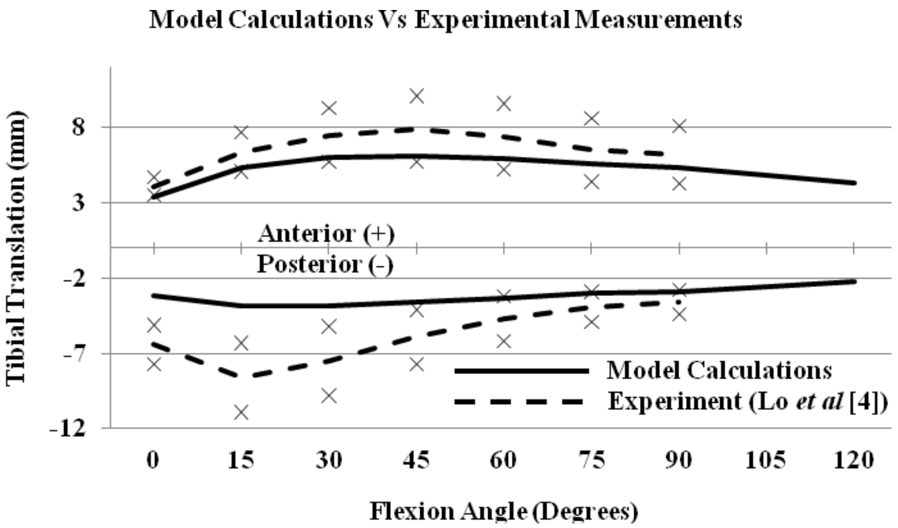

Fig. 1. A-P tibial translation under ±130 N A-P force over the flexion range. Model calculations (continuous lines) are compared with the experimental measurements of Lo *et al*. [4] (dashed lines correspond to the mean values (at 15^o interval) and '×' show standard deviation).

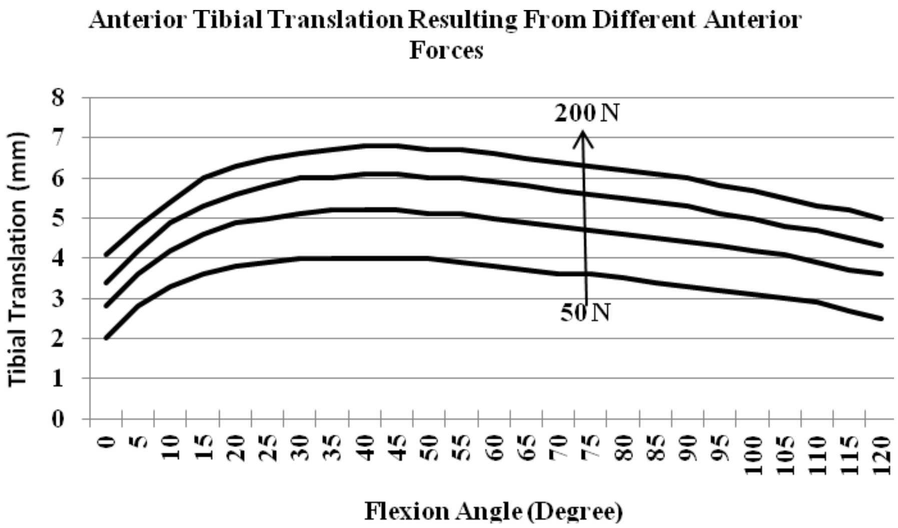

Fig. 2. Model calculations for anterior tibial translation corresponding to 50, 100, 150 and 200 N anterior force.

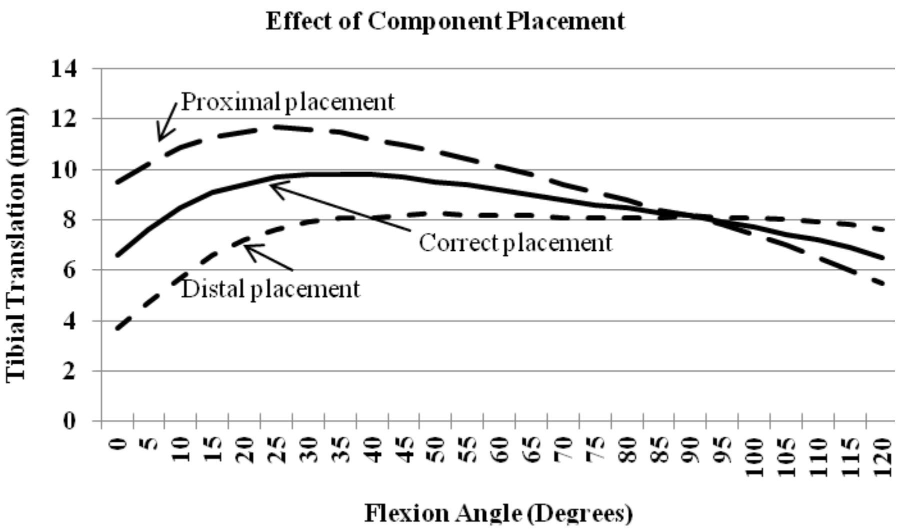

Fig. 3. Model calculations for total A-P translation under ±150 N force, comparing *correct placement* with 1mm proximal or 1 mm distal placement of the femoral component.

III. Results

Figure 1 shows a comparison between A-P tibial translations under ±130 N force calculated from the model during 0–120^o flexion and those reported by Lo *et. al*. [4] given as mean values from 14 intact cadaver knees during 0–90^o flexion (at 15^o interval).

Figure 2 gives anterior tibial translations as calculated from the model corresponding to 50, 100, 150 and 200 N anterior force on the tibia.

Figure 3 shows a comparison of the total A-P translations for the *correct placement* and for incorrect placement of the femoral component by 1 mm proximal and 1 mm distal to its correct position.

IV. Analysis

From figure 1, the patterns of A-P tibial translations calculated from the model show reasonable agreement with those from the experiment. The model calculations for anterior translations are similar to the mean experimental values and are within the reported standard deviation. The model calculations for posterior translations for angles greater than 45^o are similar to the mean experimental values and are within the reported standard deviation. For 0–45^o flexion positions, the model calculations for the posterior translation underestimate the experiment.

The anterior translation at any load (figure 2), first increases with flexion angle from 0 to about 45^o and then decrease in higher flexion. Similarly, the posterior translation first increases from 0 to about 30^o and then decreases in higher flexion. These patterns of variation with flexion are in agreement with the experimental measurements (figure 1).

Figure 2 shows that the tibial translations increased uniformly with the magnitude of the external applied load.

Figure 3 gives the effects over the flexion range of a femoral component placed 1 mm proximal or 1 mm distal to the *correct position*. Total A-P translations under ±150 N forces are plotted over the flexion range. For each incorrect placement, the variations in translation were flexion dependent, except at 90^o. Also, the patterns of change in the tibial translation reversed beyond 90^o flexion. For example, the tibial translation due to a proximal placement resulted in 2.9 mm increase at 0^o and 1 mm decrease at 120^o flexion. On the contrary, the tibial translation due to a distal placement resulted in 2.9 mm decrease at 0^o and 1.1 mm increase at 120^o flexion. There was no change at 90^o. Such variations altered the joint kinematics and, therefore, may be clinically significant for the patient outcome.

V. Conclusion

Reasonable agreement between the model calculations and experimental measurements shows that unconstrained prosthetic components with intact ligaments can reproduce the patterns of A-P translations similar to those observed in the intact knee. However, the model simulations also show that the surgical placement of prosthetic components can affect the joint kinematics significantly and variably over the flexion range. This may influence the patient outcome. The analysis has clinical relevance.

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