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Technical note

A method for quantifying the anterior load–displacement behavior of the human knee in both the low and high stiffness regions

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Abstract

The anterior load–displacement behavior of the human knee with an intact ACL is characterized by a very low stiffness region initially and a high stiffness region that develops as anterior load is increased. Although this behavior has been well recognized for some time, a method for quantitatively describing the behavior in these two regions based on limits of motion at specific values of anterior/posterior force has not yet been developed. Thus, the purposes of this study were to describe and justify such a method for measuring the laxity and stiffness in both of these regions in the intact knee.

Unique to this study, low stiffness and high stiffness laxities were computed based on three limits of motion for seven cadaveric knees tested at flexion angles ranging from 0° to 90°. Defining the reference position of the tibia relative to the femur, one limit was the 0 N posterior limit which was determined using a specially designed load cycle to reduce uncertainty in establishing a reference position. Defining the upper bound of the load–displacement curve, a second limit was the 225 N anterior limit. A third intermediate limit was the 45 N anterior limit, which was the load that represented the transition from the low stiffness to the high stiffness region. Stiffnesses corresponding to each of the two regions were computed using regression analysis and also estimated based on the laxities. Comparison between the computed and estimated stiffnesses demonstrated that the stiffnesses in both the low and high stiffness regions can be estimated reasonably accurately based on the laxities. Therefore, the 0 N posterior limit and the two laxities are the three quantities needed to describe the load–displacement behavior of the normal knee.

Keywords: Knee; Anterior force; Anterior displacement; Non-linear; Stiffness; Laxity

1. Introduction

Since the anterior cruciate ligament (ACL) is frequently torn thus requiring reconstructive surgery to restore anterior knee stability, the surgical variables that affect stability have been and continue to be widely studied through in vitro experiments on cadaveric knees with the ACL reconstructed. Among the surgical variables of interest are the graft material, the sites of the tibial and femoral tunnels, the fixation method stiffness, and graft initial tension. Regardless of the surgical variable of interest, to determine whether the anterior load–displacement behavior of a reconstructed knee has been restored to normal requires a method that comprehensively describes the anterior load–displacement relation.

Developing a method that comprehensively describes the anterior load–displacement behavior of the knee is challenging because the load–displacement curve has different regions. A typical curve has an initial low stiffness linear region and a terminal high stiffness linear region with a transitional non-linear region in between (Markolf et al., 1976). Thus, the anterior load–displacement curve of the knee has distinctive behavior within each region. It follows that a method for quantifying both the laxity and stiffness of the low and high stiffness regions may be useful for assessing how well normal load–displacement behavior is restored in a reconstructed knee.

For the most part, previous studies have not quantified the low and high stiffness regions, but instead have measured total anterior laxity as a basis for evaluation (Markolf et al., 1978, 1984; Shino et al.,

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1987; Dahlkvist and Seedhom, 1990). Total anterior laxity has been defined as the displacement of the tibia with respect to the femur from a reference position to the position achieved under a defined anterior load. The limitation of relying on just total anterior laxity to assess anterior load–displacement behavior is that two knees can have the same total anterior laxity but fundamentally different load–displacement curves because of differences in laxity and stiffness within the low and high stiffness regions.

Two studies that determined laxity in more than one region were those by Daniel et al. (1985a, b) and Edixhoven et al. (1987, 1989). The former study used an intermediate load value of 67 N and a maximum load value of 89 N, where as the latter study used an intermediate load value of 90 N and a maximum load value of 180 N. However, the intermediate values appeared to be arbitrary rather than determined from the load–displacement behavior of the knee. Considering that a regional description of load–displacement behavior is needed, and that no previous study known to the authors has satisfied this need, the purposes of this study were to describe and justify a method for measuring the laxity and stiffness in each of the two regions.

2. Methods and materials

Seven fresh-frozen cadaveric knees obtained from two female and five male donors ranging in age from 70 to 79 years (mean 75 years) were tested. On the day of testing each knee was thawed, potted, aligned, preconditioned, and tested in a six degrees-of-freedom, computer-controlled, load application system designed and built in our laboratory using a previously described protocol (Bach and Hull, 1995). This load application system allowed unconstrained motion in all degrees of freedom except flexion/extension which was constrained to a prescribed flexion angle. The knee was preconditioned by applying five anterior and five posterior load cycles to 250 N at 0° and 90° flexion to precondition the anteromedial and posterolateral bundles of the ACL (Bach et al., 1995). Zero degrees of knee extension was defined as the position of the knee with an extension moment of 2.5 N m (Markolf et al., 1990; Goss et al., 1998).

Following preconditioning, an anterior/posterior/anterior load cycle was applied at 0°, 30°, 60°, and 90° of flexion in a random order to determine the load–displacement behavior of the intact knee. From the load–displacement curve at each flexion angle, three limits of motion were derived (Fig. 1). A limit of motion was defined as the equilibrium position of the tibia relative to the femur at a defined load (Haimes et al., 1994). From an arbitrary resting position, an anterior load of 45 N was applied to the tibia and then removed.

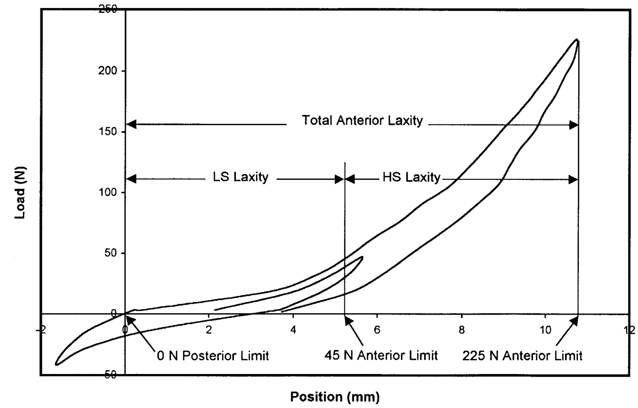

Fig. 1. Limits of motion and laxities from a typical load cycle.

Next a posterior load of 45 N was applied and removed. The position of the tibia with respect to the femur after removal of the posterior load was defined as the 0 N posterior limit. Finally, an anterior load of 225 N was applied because this is in the range of loads applied by the manual maximum test (Staubli et al., 1992) and because differences in anterior laxity are best observed when a relatively high anterior force is applied (Markolf et al., 1984). The method used to determine the third limit of motion will be described in the following paragraph.

At each flexion angle the load-deflection data were processed to compute approximate linear stiffnesses for each of the low and high stiffness regions and to derive a load for defining the third limit of motion. The stiffness of the low stiffness region was computed as the slope of a line fitted to the points on the load cycle curve from 30 N posterior load to 30 N anterior load using simple linear regression (Fig. 2). The stiffness of the high stiffness region was computed as the slope of a line fitted to the points on the load cycle curve from 105 N anterior load to 225 N anterior load using simple linear regression. The range of anterior loading greater than 30 N and less than 90 N was omitted from the regression

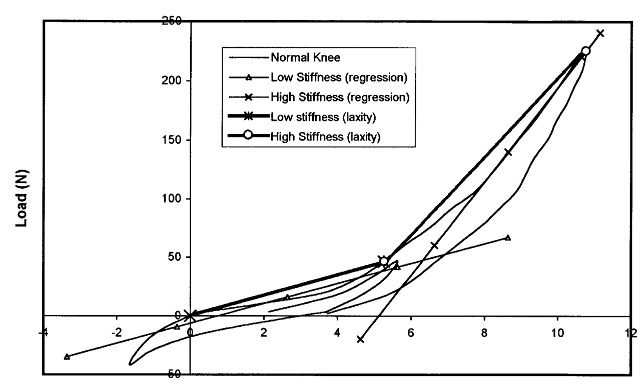

Fig. 2. The low and high stiffnesses computed from linear regression with an overlay of the stiffnesses estimated from the LS and HS laxities for a typical load cycle.

Table 1
Intersection loads (N) of the low and high stiffness slopes computed from regression analysis of the intact knee at four flexion angles

Flexion angle	Avg	(S.D.)	Min	Max
0°	49.7	(9.8)	35.4	66.1
30°	42.5	(12.8)	23.5	62.2
60°	50.1	(14.6)	30.7	74.8
90°	54.8	(13.0)	40.6	72.3
All four angles averaged	49.3		32.5	68.9

analyses because the load-deflection behavior was markedly non-linear in this range (Fig. 1). Since the load at the intersection point of the low and high stiffness regression lines for the normal knee for all flexion angles averaged 49 N anterior (range 24–75 N) (Table 1), the position of the tibia at an anterior load of 45 N was chosen as the third limit of motion. The limit was chosen as 45 N rather than 49 N to account for the small amount of friction inherent in the load application system (Bach and Hull, 1995).

Three laxities were determined from the three limits and the two stiffnesses were estimated. The laxity of the low stiffness region, or LS laxity was defined as the difference in tibial displacement between the 45 N anterior limit and the 0 N posterior limit. The laxity of the high stiffness region, or HS laxity was defined as the difference in tibial displacement between the 225 N anterior limit and the 45 N anterior limit. Total anterior laxity was defined as the difference in tibial displacement between the 225 N anterior limit and the 0 N posterior limit. The stiffness in the low stiffness region was estimated as 45 N divided by the low stiffness laxity and the stiffness in the high stiffness region was estimated as 180 N (i.e. 225 N–45 N) divided by the high stiffness laxity.

Procedures of statistical analysis were used to analyze the data in relation to the objective of the study. These procedures were applied to six dependent variables (i.e. the 0 N posterior limit, the laxity and stiffness of the low and high stiffness regions, and the total anterior laxity). One factor, repeated measures ANOVA where flexion angle was the independent variable (i.e. four levels: 0°, 30°, 60°, and 90°), was used to determine which dependent variable varied significantly between flexion angles for the intact knee. A post hoc Tukey's multiple comparisons test was used to identify the flexion angles at which the dependent variables were significantly different. Differences were significant when $p<0.05$. To determine if the estimated and computed stiffnesses in each of the low and high stiffness regions were significantly different, a paired t-test was performed.

3. Results

The 0 N posterior limit of the intact knee relative to the limit at 0° moved posteriorly by 0.9 mm as the knee was flexed to 30° and anteriorly by 1.0 mm as flexion continued to 90° (Table 2). Due to the variability of the 0 N posterior limit between flexion angles, however, these differences were marginally not significant ($p = 0.0518$).

In the low stiffness region, the LS laxity varied significantly between flexion angles ($p = 0.0104$) with the LS laxity at 30° of 5.6 mm being significantly greater than the LS laxity at both 0° and 90° of flexion (Table 2). Consistent with these results, the average stiffness of the intact knee was lowest at 30° of flexion and varied significantly between flexion angles ($p = 0.0176$) with the stiffness at 0° being significantly greater than the stiffness at 30° and 60° of flexion.

In the high stiffness region, the average HS laxity was also greatest at 30° flexion (Table 2) and varied significantly between flexion angles ($p = 0.0001$) with the HS laxity of 7.3 mm at 30° being significantly greater than the HS laxity at 0° and 90° of flexion. Similarly, the stiffness was lowest at 30° flexion and varied significantly between flexion angles ($p = 0.0001$) with the stiffness at 0° being significantly greater than the stiffness at 30° and 60° of flexion.

The total anterior laxity varied significantly between flexion angles ($p = 0.0009$) and was greatest at 30° flexion (Table 2). The total laxity at 0° of flexion was significantly less than the total laxity at 30°, 60°, and 90°, and the total laxity at 30° was significantly greater than the total laxity at 90° of flexion.

Table 2
Limits of motion, laxities, and stiffnesses computed from regression analysis of the intact knee at four flexion angles

	0°			30°			60°			90°		
Dependent variable	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max
0 N posterior limit (mm)	0.0 (0.0)	0.0	0.0	−0.9 (2.2)	−4.7	1.8	−0.4 (1.4)	−2.6	1.5	1.0 (1.7)	−0.7	4.1
Total anterior laxity (mm)	7.1 (1.9)	4.3	10.8	11.5 (2.0)	8.4	14.7	10.1 (1.8)	7.5	12.1	8.8 (1.9)	6.3	11.2
LS laxity (mm)	2.6 (1.6)	0.9	6.1	5.6 (1.9)	2.4	8.9	4.3 (1.3)	2.1	6.0	3.8 (1.3)	1.8	5.0
HS laxity (mm)	4.5 (0.5)	3.4	4.9	5.9 (0.5)	5.1	6.6	5.8 (0.7)	4.8	6.7	5.1 (0.8)	4.4	6.6
Stiffness in LS region (N/mm)	19.5 (11.0)	5.9	42.0	7.3 (2.3)	4.4	10.6	11.0 (4.1)	7.2	15.9	13.6 (5.7)	7.6	20.8
Stiffness in HS region (N/mm)	44.9 (5.8)	40.4	54.5	35.7 (3.1)	31.4	40.1	36.1 (3.9)	32.2	42.1	41.7 (5.7)	32.4	49.8

Table 3
Stiffness estimated from LS and HS laxity and the difference from the stiffness computed regression for the intact knee at four flexion angles

Stiffness of the low and high stiffness regions estimated from LS and HS laxities for the intact knee

	0 degrees			30 degrees			60 degrees			90 degrees		
Estimated variable	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max	Avg (S.D.)	Min	Max
Stiffness from LS laxity (N/mm)	22.2 (12.6)	7.4	47.8	9.2 (4.3)	5.0	18.4	11.6 (4.7)	7.5	21.0	13.6 (5.9)	8.9	24.6
Stiffness from HS laxity (N/mm)	41.0 (5.8)	36.7	52.9	30.7 (2.7)	27.4	35.2	31.6 (3.8)	26.8	37.5	36.1 (5.2)	27.2	41.0

Difference between the stiffness estimated from the LS and HS laxities and the stiffness computed from regression for the intact knee

	0°	30°	60°	90°
Estimated stiffness computed variable	Avg	Avg	Avg	Avg
stiffness difference in LS region (N/mm)	2.8	1.9	0.5	0.1
Stiffness difference in HS region (N/mm)	−3.9	−5.0	−4.5	−5.6

The comparison of the estimated versus computed stiffness revealed no significant difference between the estimated and computed stiffness in the low stiffness region at 30° ($p = 0.2136$), 60° ($p = 0.6243$), and 90° ($p = 0.9584$) of flexion (Table 3). At 0° of flexion the estimated stiffness was significantly greater than the computed stiffness by an average of just 2.8 N/mm ($p = 0.0322$). In the high stiffness region, however, the estimated and computed stiffness were significantly different at 0° ($p = 0.0032$), 30° ($p = 0.0005$), 60° ($p = 0.0008$), and 90° ($p = 0.0003$) of flexion (Table 3). Although the differences were statistically significant, the graphic appearance of the estimated and computed stiffness was similar (Fig. 2) because the computed stiffness was only 7–13% (3.9 to 5.6 N/mm) greater than the estimated stiffness.

4. Discussion

Recognizing that a method for characterizing the anterior load–displacement behavior is necessary for evaluating variables important to ACL reconstructive surgery, and that no previous study has characterized the regional behavior based on limits of motion, the goals of this study were to develop such a method and demonstrate its use. Implicit to the approach taken to characterize regional behavior was that a two-piece linear approximation of an inherently non-linear curve (Fig. 1) is useful for comparative analysis of load–displacement behavior. This approximation is useful for two reasons. First, the two-piece linear approximation represents most of the curve since the non-linear transition region between the low and high stiffness regions is the smallest of the three regions (Fig. 2). Second, the piecewise linear approximation reduces the complex non-linear curve to a set of six measurable quantities amenable to comparison through statistical analysis. The six quantities included the 0 N posterior limit, the three laxities, and the two stiffnesses. Without a reduction to these quantities, it would be difficult to compare two non-linear curves and draw meaningful conclusions in a clinical context.

Since the computations of the laxities depended on the limits of motion, the effect that any mechanical friction in the load application system had on establishing these limits should be considered. Of the three limits, friction would affect the 0 N posterior limit the greatest since the load was the lowest. Any friction inherent in the load application system would establish a more posterior reference position of the tibia with respect to the femur than if the frictional effect was negligible. A more posterior position of the tibia might inflate the low stiffness laxity and the total laxity (Fig. 2).

Recognizing that friction introduces errors in two of the laxity quantities, a pilot study was conducted to quantify the error. In the pilot study, the same knees were subjected to the same load cycle applied with the friction inherent to the load application system both present and absent. In comparing the laxities for the two cases, the differences were limited to about 10% for both the low stiffness laxity and total laxity. Furthermore, the differences were not systematic. The low relative error and the lack of any systematic effect indicate that the friction inherent to the load application system was not an important source of error in laxity measurements.

The definition of a reference position based on a prescribed loading cycle is important to avoid errors with the resting or neutral reference position used in other studies as the reference position to calculate laxity quantities (Markolf et al., 1984; Grood et al., 1988; Dahlkvist and Seedhom, 1990; Haimes et al., 1994). The resting or neutral position has a random error caused by two factors: the potential for the resting or neutral reference position to lie anywhere within the primary laxity region and the high coefficient of variation (30%) or low repeatability of the primary laxity region. Primary laxity or the amount of "play" has been defined as the amount of movement present at low force levels where only frictional and viscous forces need to be overcome to translate the knee (Dahlkvist and Seedhom, 1990). The random error is influenced by the interaction between the applied load, friction in any load application system, and viscous forces within the knee. If the applied loads are not great enough to overcome the frictional and viscous forces, then the tibial position may be anywhere within limits of the primary laxity region. Since the resting or neutral position depends on the load applied prior to the knee reaching an equlibrium state, the position of the tibia would more likely be in the anterior portion of the primary laxity region if the tibia had been subjected to an anterior load and conversely for a posterior load. Considering that the primary laxity at 30° of flexion in our study was 3.4 mm, the uncertainty inherent in using the resting or neutral position as the reference position becomes apparent. The use of the 0 N posterior limit provides a repeatable reference position that avoids the random error associated with the resting or neutral reference position.

Although the 0 N posterior limit provides a repeatable reference position, laxity calculations based on this limit may be difficult to compare in cadaveric studies that use different methods of load application because of two factors. One factor is friction since it can inflate two of the laxity quantities as noted above. Although the load application system used in the present study controlled the friction to acceptable levels, if the friction were to increase, then low stiffness and total laxity would increase accordingly. For meaningful comparisons to be made between studies, the effect of friction on the 0 N posterior limit must be assessed critically. Since any friction would introduce a systematic change in the two laxity quantities, comparisons within a study could still be made meaningfully however.

The other factor is the weight whose effect depends both on the orientation of the shank in the gravity field and on any method of load application used to balance the weight. With the load application system used in the present study, the knee specimen was in the prone position, the shank segment remained horizontal, and the weight of the shank was balanced so that the weight did not affect the 0 N posterior limit.

If this method is used in the clinical setting, then caution should be exercised in interpreting laxity measurements based on the 0 N posterior limit. If the posterior cruciate ligament is damaged, then the 0 N posterior limit may be more posterior than if the posterior cruciate ligament is healthy (Edixhoven et al., 1987). Consequently, both the low stiffness laxity and total laxity would be overestimated hence, possibly leading to a false positive in the diagnosis of ACL deficiency. In tests on ACL deficient knees not reported herein, however, both the low stiffness and high stiffness laxities approximately doubled over those of the intact knee (Eagar et al., 1999). Since the high stiffness laxity does not depend on the 0 N posterior limit, the high stiffness laxity can serve as a viable indicator of ACL deficiency in its own right (Daniel et al., 1985a). Moreover, because damage to the posterior cruciate ligament is much less common than damage to the ACL, the 0 N posterior limit will serve as a meaningful reference in most cases.

Not only is a reference position an essential quantity in determining laxity, but also it is an important quantity in its own right. This is because when anterior laxity is reported without an accompanying reference position, even if the load–displacement behavior of one knee has the same curve shape as another knee, then there is still no assurance that the load–displacement behavior is identical for the two knees. For example a total anterior laxity equal to the total laxity of the normal knee can be established if the tibia position was shifted 2 mm posterior to both the 0 N posterior limit and the 225 N anterior limit. Such a posterior shift might be expected in cases of ACL reconstruction where the graft is pretensioned excessively; therefore, a reference position such as the 0 N posterior limit defined herein should be included in load–displacement evaluations of reconstructed knees. Of course, this statement applies only to cadaveric studies and not clinical studies since there is no way to determine a priori a reference position for the intact joint in the clinical setting.

One interesting observation from this study is that the change in the 0 N posterior limit over a wide range of flexion (i.e. 0–90°) was bounded by about ± 1 mm on average (Table 2). This limit remained relatively constant because of the kinematics inherent to flexion/extension motion of the knee in conjunction with the method used to align the joint with the axes of motion in the load application system. As demonstrated by Hollister et al. (1993), the functional flexion/extension axis is fixed to the femur and translations of the tibia in the plane perpendicular to this axis are minimal. Using a load application system structured around the coordinate system of Grood and Suntay (1983), the knee specimens used in this study were aligned such that the functional axes in flexion/extension and axial rotation were collinear with those of the load application system. Accordingly, the coupled anterior/posterior translation in the plane of flexion vanished (Bach and Hull, 1995) so that the 0 N posterior limit remained relatively constant as the flexion angle was varied.

Although the 225 N anterior limit in conjunction with the 0 N posterior limit bound the anterior load–displacement curve, unique to this study was the determination of an intermediate third limit. This limit was required to describe the shape of the curve because of its hardening non-linear behavior. In an earlier study, Markolf et al. (1976) performed a piecewise linear approximation to the A/P load–displacement curve by placing a line tangent to the inflection point of the curve at no load and a second line tangent to the terminal load points (100 N). The intersection formed by these two tangents was defined as the upper bound of the "neutral stiffness" region from which laxity was measured. However, the load at which this intersection point occurred varied with changes in the stiffness. Therefore, the corresponding laxity was not determined at a consistent load. Since total anterior laxity is determined at consistent load limits which can be controlled experimentally, it is more appropriate to measure intermediate laxities also from consistent load limits, such as the 45 N anterior limit used for this study. Note that the 45 N anterior limit determined from analysis of the anterior load-deflection behavior is substantially lower than intermediate anterior load values of 67 N (Daniel et al., 1985a, b) and 90 N (Edixhoven et al., 1987, 1989) selected arbitrarily by others.

With the 45 N anterior limit defined as the limit of motion that bounds the low stiffness region of the load–displacement curve, the basic shape of the curve can be characterized in terms of two laxity values. By defining LS laxity as the displacement from the 0 N posterior limit to the 45 N anterior limit, the low stiffness region of the curve can be identified and the slope of the line connecting these two points can be used to estimate the stiffness in the low stiffness region (Fig. 2). Similarly, by defining HS laxity as the displacement from the 45 N anterior limit to the 225 N anterior limit, the high stiffness region of the curve can be identified and the slope of the line connecting these two points can be used to estimate the stiffness in the high stiffness region. Thus with the 0 N posterior limit, LS laxity, and HS laxity, the load–displacement curve can be located on the displacement axis and the shape of the curve can be approximated with two line segments.

In summary, this study developed a method that described the regional behavior of the anterior load–displacement curve of the human knee. The 0 N posterior limit reduced the uncertainty in defining a reference position and the 45 N anterior limit was a reasonable upper bound for the low stiffness region and a reasonable lower bound of the high stiffness region. This method may prove useful both in the scientific study of independent variables (e.g. graft material, tunnel positions, pretension, and fixation methods) which affect the anterior load–displacement of reconstructed knees and in the clinical assessment of anterior stability.

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