COMPUTER-ASSISTED MEASUREMENTS OF CORONAL KNEE JOINT LAXITY IN-VITRO ARE RELATED TO LOW STRESS BEHAVIOUR RATHER THAN STRUCTURAL PROPERTIES OF THE COLLATERAL LIGAMENTS

Running Header: Ligament Properties and Coronal Knee Laxity

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ABSTRACT

The relationship between coronal knee laxity and the restraining properties of the collateral ligaments remains unknown. This study investigated correlations between the structural properties of the collateral ligaments and stress angles, used in total knee arthroplasty (TKA) and sports injuries, measured with an optically-based navigation system. Ten fresh-frozen cadaveric knees (age 81±11 years) were dissected to leave the menisci, cruciate ligaments, posterior joint capsule and collateral ligaments. The resected femur and tibia were rigidly secured within a test system which permitted kinematic registration of the knee using a commercially available image-free navigation system. Frontal plane knee alignment and varus-valgus stress angles were acquired. The force applied during varus-valgus testing was quantified. Medial and lateral bone–collateral ligament–bone specimens were then prepared, mounted within a uniaxial materials testing machine and extended to failure. Force and displacement data were used to calculate the principal structural properties of the ligaments. There was no significant difference in the mean varus (4.3±0.6º) and valgus (4.3±2.1º) laxity or the corresponding force applied (9.9±2.5N and 11.1±4.2N, respectively). While measures of knee laxity were independent of the ultimate tensile strength and stiffness of the collateral ligaments, there was a significant correlation between the force applied during stress testing and the instantaneous stiffness of the medial ($r = 0.91$, $p = 0.001$) and lateral ($r = 0.68$, $p = 0.04$) collateral ligaments. These findings suggest that clinicians may perceive a rate of change of ligament stiffness as the end-point during assessment of collateral knee laxity.

Key words: Collateral Knee Ligaments, Navigated TKA, varus/valgus laxity.

INTRODUCTION

Total Knee Arthroplasty (TKA) has become the standard procedure for the management of degenerative joint disease, providing pain relief, correction of deformity, and restoration of function. However its success depends on several factors, including patient selection, appropriate implant design, effective perioperative care and correct surgical technique [1]. In particular, the importance of post-operative coronal alignment of the knee on clinical outcomes has been well documented [2-4]. More recently however, it has been reported that alignment alone may not be an accurate predictor of implant function and longevity, but rather a combination of accurate alignment and soft tissue balancing is required to optimise the clinical outcome [5, 6].

Often, in knees with large varus and valgus deformities, the soft tissue structures have contracted, meaning that it is not possible to achieve a neutral alignment through bone cuts alone [7]. Functionally, the medial collateral ligament (MCL), particularly its superficial portion, acts as the primary restraint to valgus angulation, and the lateral collateral ligament (LCL) serves as the primary restraint to varus angulation [8, 9]. Therefore, the collateral ligaments may be moderately or extensively released in order to achieve a well balanced joint [10]. Soft tissues should be balanced in both flexion and extension and tensioned sufficiently to produce stability, but without limiting the range of movement [6]. Conventionally, the technique has involved a subjective assessment of laxity [11], but with increasing patient expectations and greater mechanical demands placed on the prosthesis [12], quantitative assessment may be required.

Computer navigation systems have been recently introduced into TKA surgery [13-15] and the use of computer-assisted surgery has been associated with better clinical outcomes when compared to conventional techniques [16-21]. Several studies have independently demonstrated that these systems improve the accuracy of both component placement and alignment [22-27]. However, they are also particularly useful for the intra-operative measurement of passive kinematic joint characteristics, as they can provide accurate assessment of coronal knee laxity [28], and can be used to ensure that the quantified joint laxity is acceptable [29]. Navigation-based measures of varus and valgus stress angles have been used to define a number of algorithms for management of soft tissue in TKA [29-33]. This provides an objective assessment of laxity which, in turn, will allow for more precise soft tissue balancing.

However, the extent to which such navigation-derived stress angles reflect the restraining properties of the collateral ligaments of the knee remains unknown. The aim of this study was to assess the relationship between traditional measures of the structural properties of the collateral ligaments of cadaver knees and the stress angles measured with an optically-based navigation system. It would be expected that these ligaments, as major stabilising structures of the knee, would have a significant role in resisting the moments applied in measuring stress angles. Therefore the hypothesis was that a relationship would exist between the structural properties of the collateral ligaments and the coronal laxity of the knee.

MATERIALS AND METHODS

Specimens

This study received Institutional Review Board approval. Dissections were performed on ten fresh frozen human cadaver knees (nine left and one right) obtained from five male and five female donors, mean age 81 years (range, 62 – 92 years). All knees were free of gross musculoskeletal trauma and pathology and had been resected approximately 10cm either side of the joint line. The dissection technique was developed using previously described methods [34]. Specimens retained the collateral ligaments, cruciate ligaments, menisci and posterior capsule in order to retain structural stability of the knee. All other remaining soft tissues of the femur, tibia and fibula were removed.

< Figure 1 >

Assessment of knee laxity using computer navigation system

As the specimens were resected 10cm proximal and distal to the knee joint line, a custom-made aluminium replica of the lower limb was manufactured. The dimensions of the femoral and tibiofibular-foot components were based on average segment lengths reported within the literature [35]. The resected femoral and tibial shafts were rigidly secured within the femoral and tibiofibular components, respectively (Figure 1). The model allowed kinematic registration of the knee using the standard workflow of a commercially available image free navigation system [15, 36]. Data was collected using the Orthopilot® Computer Tomography free navigation system (Aesculup/B. Braun, Tuttlingen, Germany). Two passive infrared trackers were affixed, one to the femoral component and one to the tibiofibular component and the

workflow followed until the monitor screen displayed the coronal and sagittal femorotibial angle (FTA). The knee was placed in extension and the coronal mechanical FTA recorded using the navigation system. A varus and valgus stress force was then applied at the ankle with the knee joint held at 3° flexion. Force was manually applied at the distal end of tibiofibular component using the right hand of the principle investigator. The left hand was used to stabilise the knee joint. The force applied was a subjective evaluation of the force required to reach a perceived ‘end-point’. The maximum coronal FTA attained during varus and valgus stress testing was recorded. The magnitude of the force was quantified with a dynamometer attached to the distal tibial shaft. Varus and valgus stress testing was repeated three times. Knee laxity was calculated as the average peak angle, in degrees, that the knee deviated from its initial coronal plane alignment during application of varus and valgus stress force.

< Figure 2 >

Mechanical testing of collateral ligaments

Following assessment of knee laxity, specimens were prepared for mechanical testing of the structural properties of the collateral ligaments. The methodology is described in detail elsewhere [34]. In brief, the specimen was sagittally sectioned and remnants of the cruciate ligaments, posterior capsule and any other remaining tissue were dissected free to leave two bone-collateral ligament-bone specimens (Figure 2). Each specimen was then secured in a uniaxial materials testing machine and following 10 preconditioning cycles (3.5 % strain at 1 Hz), extended to failure at a rate of 500 $mm.min^{-1}$. Force and displacement data were used to calculate the ultimate tensile strength and stiffness of the two ligaments. Ultimate tensile

strength was defined as the maximum force reached during testing. Stiffness was derived via two methods. A traditional measure of ligament stiffness (tangent stiffness) was determined for each ligament by calculating the gradient of the linear section of the force-displacement curve (Figure 3). In addition, instantaneous stiffness of each ligament was estimated at forces corresponding to the peak moment applied manually to the knee during the clinical laxity testing. Instantaneous stiffness was calculated by analytically determining the first order derivative of polynomial functions fit to the force-displacement data [37].

< Figure 3 >

Statistics

Statistical analysis was carried out using the Statistical Package for the Social Sciences (SPSS Inc, Chicago, IL USA). Paired t-tests were used to compare normally distributed variables (varus and valgus stress angles, force applied during stress testing). Pearson's correlations were used to investigate potential relationships between dependent variables. For all tests, significance was assessed at an alpha level of 0.05.

RESULTS

The MCL and LCL were found to be well-defined structures on dissection and free of obvious musculoskeletal pathology. One of the ten specimens failed via fracture of the femur during mechanical testing, and consequently was excluded from the study, giving a total of nine knees for subsequent analysis.

Assessment of knee laxity

The median coronal alignment of the knees in extension was 2º valgus, and ranged between 3º varus and 8º valgus. Six of the nine knees were valgus (FTA > 0°), one was varus (FTA < 0°), while two were neutrally aligned (FTA = 0°).

The mean varus laxity for the nine knees was 4 ± 1 º (range 3 – 6 º), while the mean valgus laxity was 4 ± 2 º (range 2 – 8 º). There was no statistically significant difference between varus and valgus laxity (p=0.97). However, valgus laxity was more variable than varus laxity as demonstrated by a greater standard deviation and range (Table 1).

< Table 1 >

The mean force applied during all laxity tests was 10 ± 3 N. There was no statistically significant difference between the force applied in either direction during varus and valgus laxity testing (p=0.17). Similarly, there was no significant correlation between the force applied during laxity testing and the resultant angular deviation in a varus (r = 0.24) or valgus direction (r = 0.13).

Collateral Ligament Properties

The principal structural properties of the collateral ligaments are shown in Table 1. While the ultimate tensile strength of the MCL was significantly greater than that of the LCL ($p<0.001$), there was no statistically significant difference in the tangent stiffness or instantaneous stiffness of the two ligaments at forces equivalent to those applied during laxity testing.

< Table 2 >

Measures of varus and valgus laxity were not significantly correlated with the tangent stiffness or ultimate tensile strength of the collateral ligaments (Table 2). However, there was a significant correlation between the force applied during stress testing and the instantaneous stiffness of the medial ($r = 0.92$, $p = 0.001$) and lateral ($r = 0.69$, $p = 0.04$) collateral ligaments. (Figure 4)

< Figure 4 >

DISCUSSION

This study tested the hypothesis that the mechanical properties of the collateral ligaments of the knee would govern varus and valgus stress angles measured with the aid of an optically-based navigation system. Contrary to our hypothesis, we found that the magnitude of stress angles was neither correlated with the force applied during laxity testing nor the structural properties of the restraining collateral ligaments. As shown in Figure 3, the magnitude of the

force applied during clinical laxity testing was small (10 N) and well below those used to determine the key structural properties of ligaments (ie the linear section of the force-displacement curve, >200 N). Thus, clinical tests of knee laxity do not give an indication of the true elastic or restraining properties of the collateral ligaments.

While it may be argued that measures of laxity reflect the low stress behaviour of the ligaments, equivalent to the so called ‘toe region’ of the force-displacement curve, angular deviation was also unrelated to the instantaneous stiffness of the collateral ligaments at forces equivalent to those applied during laxity testing. Thus, varus and valgus stress angles of the knee may be influenced by deformation of other soft tissue structures, and some variable degree of laxity may be possible without eliciting strain on the collateral ligaments.

Although varus and valgus stress angles were unrelated to the applied force or restraining properties of the collateral ligaments, the force applied during manual laxity testing was significantly correlated with the instantaneous stiffness of the two ligaments. We hypothesise, therefore that during manual testing, the operator may have sensed a change in the stiffness of these ligaments as the perceived end-point of the movement. Speculatively, this perception may be surgeon specific and may explain previous reports of large inter-surgeon variation when assessing laxity *in vivo* using a non-invasive navigation system [38]. Further research evaluating factors related to the determination of a perceived endpoint of clinical laxity testing, therefore, appears warranted.

The mean force applied during clinical laxity testing in this study (10N) is considerably lower than the values reported elsewhere for examination of the joint *in vivo* [39]. This would be expected, due to the resistance provided by active structures and other anatomical restraints which were removed from our specimens. A clinical examination of knee laxity on a patient is an attempt to isolate and test only the passive restraints provided by the ligaments and joint geometry, not the active restraints or joint compressive forces. Such isolation is not possible *in vivo*, so only a limited evaluation of the ligaments that provide joint stability is possible by the clinical laxity tests [40].

This study has a number of limitations which should be considered when interpreting the results. Measurements of initial coronal alignment are a result of the test set up which does not perfectly re-create the natural anatomy of the lower limb. Hence initial alignment is used solely as a neutral reference point for subsequent determination of laxity. Furthermore, this study focused on individual structures and is, therefore, not a true representation of knee behaviour *in vivo*. While we have chosen to focus on the primary ligamentous stabilisers of the extended knee during coronal laxity testing, other structures undoubtedly act as secondary restraints [8]. Thus, the experiment represents an idealised version of the *in vivo* condition and further study would be required to investigate the relationships between laxity and the properties of the other stabilising structures. Also, all laxity tests in this study were carried out with the knee joint held in three degrees of flexion, but there is evidence to suggest that measurements of laxity change with flexion of the knee [28]. Therefore further work is required to assess any potential influences of laxity over a range of knee flexion angles.

Nonetheless, the findings of this study suggest that angular deviation of the knee during varus and valgus laxity testing is not related to traditional structural properties of the restraining collateral ligaments. However, the force applied by operators during clinical testing, although low, is positively correlated to the instantaneous stiffness of the ligament, suggesting operators may perceive a change in the low-stress behaviour of these ligaments.

CONCLUSION

The findings of the current study suggest that clinically assessed knee laxity is independent of traditional measures of structural properties of the collateral ligaments. However, the force applied during manual stress testing of the knee was correlated with the instantaneous stiffness of the collateral ligaments. Therefore, clinicians may perceive an “end-point” based on the low-stress behaviour of these ligaments, particularly a rate of change of stiffness. Further work is required to understand the relationships between applied force, resultant stress angles and clinical outcomes for TKA, in order to make full use of the quantitative data afforded by current navigation systems.

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DECLARATION OF INTEREST

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Table 1: Mean (±SD) knee joint laxity, manually applied force, and structural properties of the MCL and LCL for each specimen.

	Laxity (degrees)		Applied Force (N)		UTS (N)		Tangent Stiffness (N/mm)		Instantaneous Stiffness (N/mm)	
Knee	**Valgus**	**Varus**	**Valgus**	**Varus**	**MCL**	**LCL**	**MCL**	**LCL**	**MCL**	**LCL**
1	4	4	11	10	619	288	42	47	37	38
2	4	5	17	12	1128	460	63	58	52	49
3	3	5	15	12	925	432	63	51	60	45
4	3	4	15	12	548	189	67	41	54	44
5	3	4	4	4	793	549	96	74	21	26
6	2	3	7	10	1076	394	61	63	34	42
7	3	4	8	10	632	384	53	78	32	54
8	8	5	10	9	642	373	63	60	42	51
9	7	4	12	10	661	315	56	45	37	46
Mean ± SD	4 ± 2	4 ± 1	11 ± 4	10 ± 3	780 ± 214	376 ± 104	63 ± 15	57 ± 13	41 ± 12	44 ± 8
p-value	0.86		0.22		<0.001*		0.34		0.46	

* indicates a statistically significant difference (p<0.05)

Table 2: Correlation coefficients and corresponding p-values between measures of varus and valgus knee laxity and the principal structural properties of the corresponding ligament.

	Valgus laxity (MCL)		**Varus laxity (LCL)**	
Structural property	**Pearson's Co-efficient (r)**	**p-value**	**Pearson's Coefficient (r)**	**p-value**
Instantaneous stiffness	0.02	0.97	0.35	0.36
Tangent stiffness	-0.19	0.63	-0.13	0.74
Ultimate tensile strength	-0.37	0.33	0.21	0.58

* indicates a statistically significant difference (p<0.05)

FIGURES

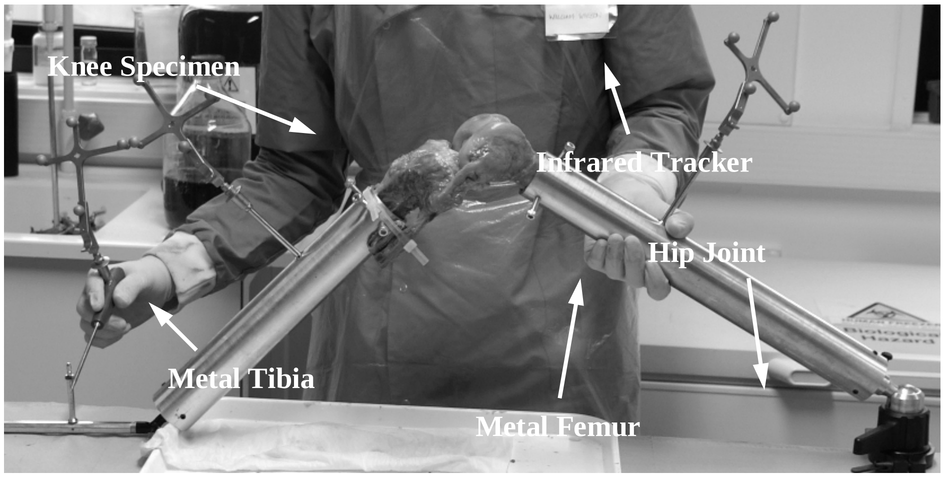

Figure 1. Photograph of the simulated lower limb during the registration process for computer navigation system.

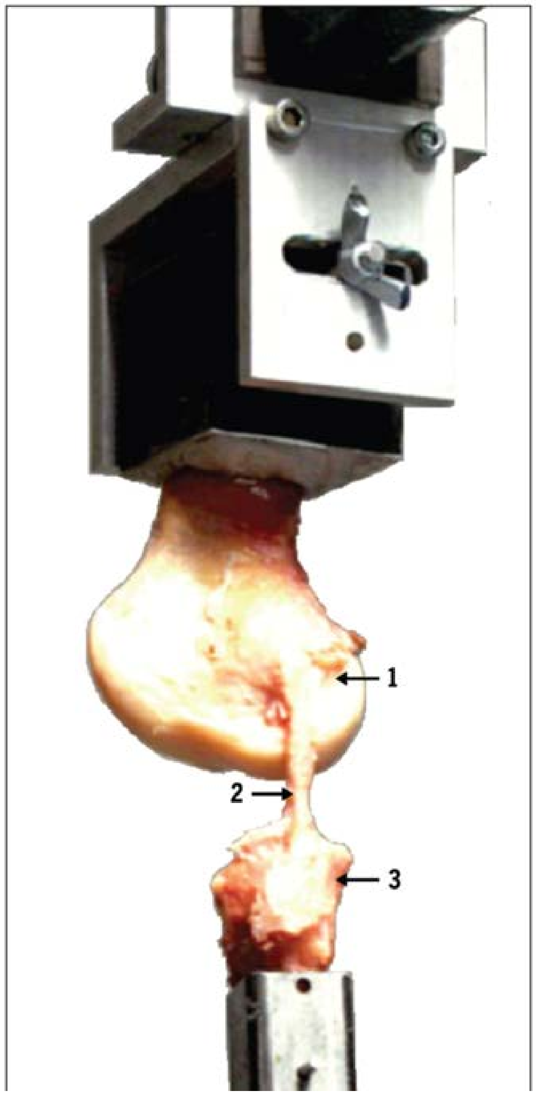

Figure 2. Picture of a bone-ligament-bone preparation mounted in the tensile testing apparatus.

(1) Femur; (2) Lateral collateral ligament; (3) Fibula.

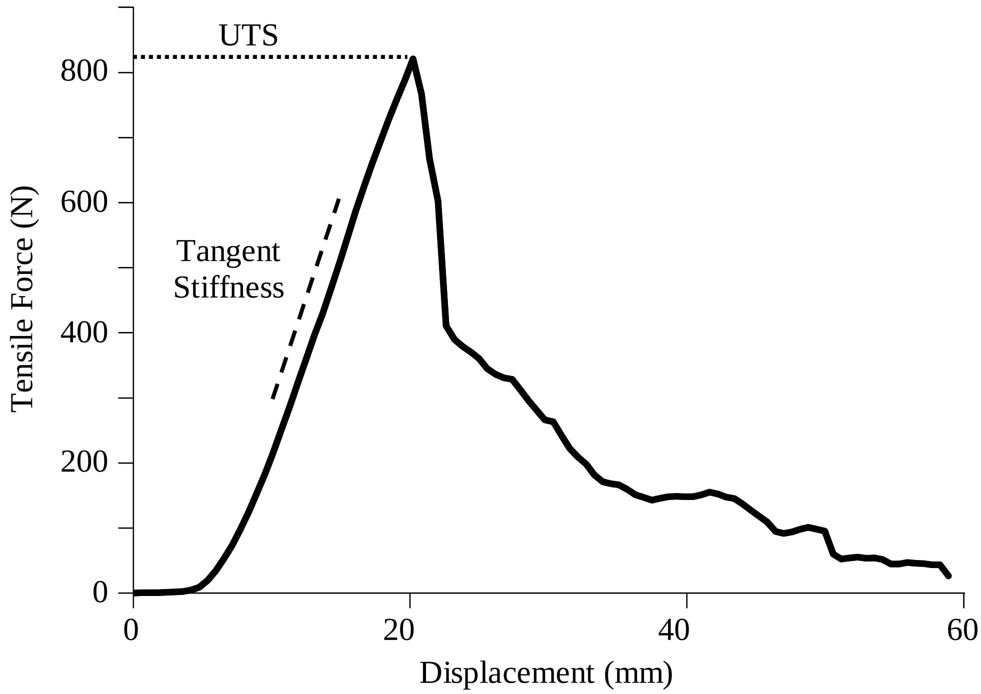

Figure 3. Illustration of tangent stiffness and ultimate tensile strength parameters derived from a typical force-displacement curve for the MCL. Tangent stiffness was defined by the slope of the linear section of the force-displacement curve (offset for illustrative purposes), while ultimate tensile strength was defined as the peak force during failure testing. Note that the force applied during manual laxity testing (N) lies within the 'toe' region of the force-displacement curve, and is well below the tensile strain at which structural properties are typically calculated.

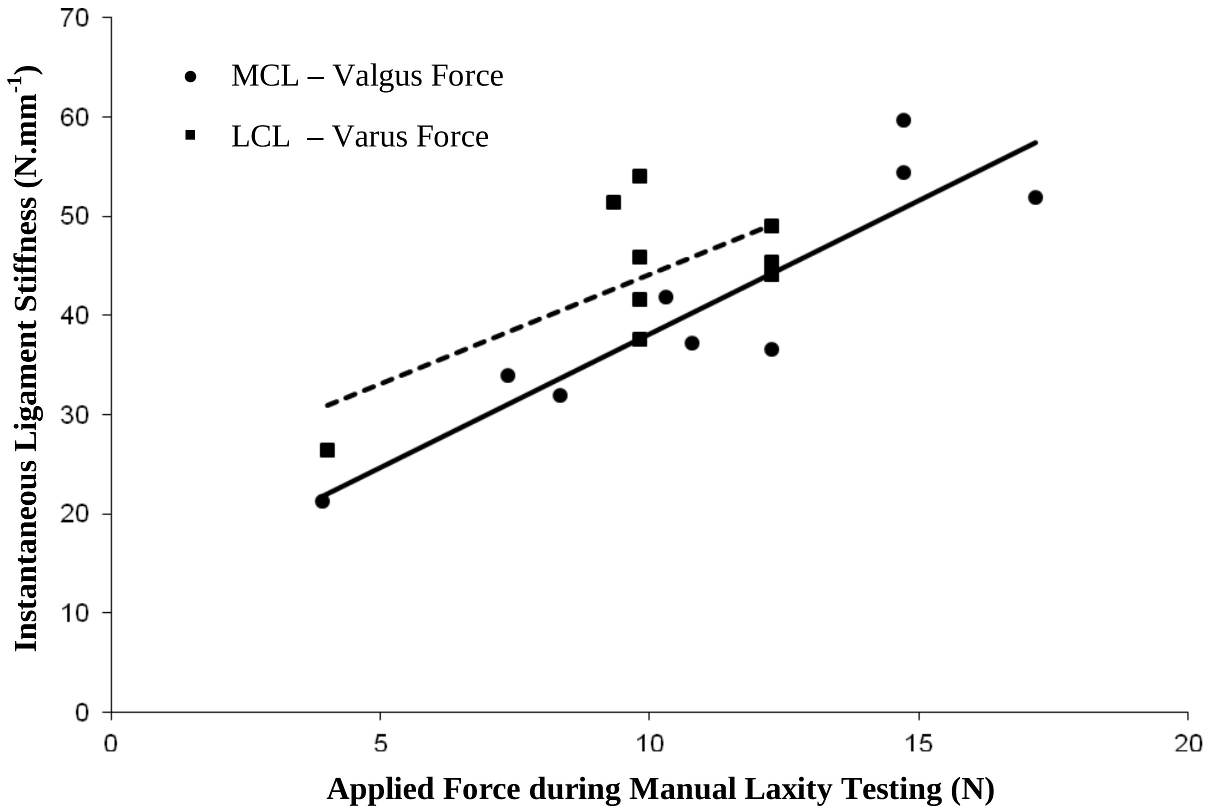

Figure 4. Correlation between applied force and ligament instantaneous stiffness at that force for valgus / MCL and varus / LCL tests