Measurement of varus–valgus and internal–external rotational knee laxities in vivo—Part I: assessment of measurement reliability and bilateral asymmetry

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Abstract:

We examined the capabilities of the Vermont Knee Laxity Device (VKLD) in measuring varus (VR)–valgus (VL) and internal (INT)–external (EXT) rotational laxities by quantifying measurement consistency and absolute measurement error ($N = 10$). Based on the expected measurement error, we then examined side-to-side differences ($N = 20$). For all measures, the knee was flexed 20°, the thigh securely fixed, and counterweights applied to the thigh and shank to create an initial zero shear and compressive load across the tibiofemoral joint. A 10-Nm torque was applied to the knee for VL and VR during non-weight-bearing, and a 5-Nm torque was applied for INT and EXT during non-weight-bearing and weight-bearing conditions. Position sensors measured angular displacements (deg). Except for INT during weight bearing, measurement consistency was good to excellent (range, 0.68–0.96), with absolute measurement errors generally less than 2° for VR–VL and 3–4° for INT–EXT. Although side-to-side differences were observed, they did not exceed absolute measurement errors. The VKLD yields reliable measures of VR–VL and INT–EXT laxities, with sufficient measurement precision to yield clinically relevant differences.

Keywords: bilateral asymmetry | measurement agreement | frontal plane knee laxity | transverse plane knee laxity

Article:

INTRODUCTION

Joint laxity has received attention as a possible risk factor for knee joint trauma,[1–4] and more specifically anterior cruciate ligament (ACL) injury.[5–8] Greater knee laxity is thought to be associated with an increased demand of the leg musculature to maintain joint stability.[9] This is supported by studies demonstrating increased anterior translation of the tibia relative to the femur when transitioning from non-weight bearing to weight bearing,[10] and greater biceps femoris activation during weight-bearing tasks.[11,12] However, when examining these relationships, knee laxity has primarily been characterized in the sagittal plane using common clinical measurement methods (e.g., general joint laxity, genu recurvatum, and anterior knee laxity), which may or may not be representative of joint motion in the frontal or transverse planes. Varus–valgus (VR–VL) and internal–external (INT–EXT) rotational knee laxities are rarely measured or investigated for their effects on knee joint function or injury risk, yet the capsulo-ligamentous restraints that limit these motions are different.[13] Further, the knee is frequently subject to varus–valgus and internal–external torques during sport, and most ACL injury mechanisms are reported to include components of valgus and either internal or external rotational torques at the knee.[14,15] Hence, quantifying VR–VL and INT–EXT laxities may also be important to understand how knee laxity (both within and between sexes) may impact knee joint function, and the potential for traumatic injury. Further, understanding the extent to which these values may differ from side to side may help us determine if the laxity measured on one limb (e.g., uninjured) may be sufficiently representative of the other (e.g., injured) limb.

Various custom devices have been used to measure VR–VL[16–18] and INT–EXT[19] laxities in healthy knees in vivo, as commercial clinical devices are not readily available. Prior to our initiating studies that examined the relationships between VR–VL and INT–EXT knee laxities and knee joint function, our purpose was to establish the capabilities of the Vermont Knee Laxity Device (VKLD) in measuring VR–VL and INT–EXT laxities by quantifying day-to-day measurement consistency and absolute measurement error. A secondary purpose was to quantify side-to-side differences in these laxity values, and compare these differences with the absolute measurement error.

METHODS

Twenty University students between the ages of 18 and 30 years (10 males, 27.3 ± 3.4 years, 177.3 ± 6.8 cm, 81.1 ± 7.0 kg; 10 females, 22.9 ± 1.5 years, 169.0 ± 7.1 cm, 66.1 ± 11.4 kg) participated. Subjects had no history of knee ligament injury or surgery, were free of other lower extremity injury or chronic pain for the past 6 months, and were otherwise healthy. Subjects were informed of the study risks and signed a consent form approved by the University's institutional review board. Measures were taken on both left and right knees. Varus (VR) and valgus (VL) rotation laxity was always tested first, followed by internal (INT) and external (EXT) rotation laxity during both non-weight-bearing (INT_{NWB}, EXT_{NWB}) and weight-bearing (INT_{WB}, EXT_{WB}) conditions. The first leg (left, right) and the first direction of applied torque (i.e., varus, valgus; internal, external) were counterbalanced across subjects. To quantify measurement consistency, 10 subjects (5 males, 5 females) were tested a second time, 24 to 48 h later. Two testers were required for data collection; each performed their respective procedures on both days, and both were blinded to the results from the previous session.

Procedures

VR–VL and INT–EXT knee laxities were measured with the Vermont Knee Laxity Device (VKLD; University of Vermont, Burlington, VT) (Fig. 1). Subjects were positioned supine with the foot of the test limb strapped to the foot cradle, and the flexion axes of the ankle and hip joints aligned with the mechanical axes of rotation of the VKLD counterweight system. For all measures, the knee was flexed 20°, the thigh securely fixed, and counterweights applied to the thigh and shank. The counterweights offset the effect of gravity acting on the lower extremity, creating an initial zero shear load across the tibiofemoral joint that was used as a measurement reference. Thigh and shank counterweights and their respective locations were selected using the model described by Zatsiorsky et al.[20] to estimate segment masses and center of mass locations. To minimize femur movement when applying torques to the knee, the thigh was tightly clamped on the medial and lateral sides just proximal to the femoral epicondyles with densely padded Plexiglas™ plates that conformed to the thigh shape (see Fig. 1). To prevent movement of the foot and ankle within the foot cradle, subjects were tightly fitted with an ankle brace (Ankle Lok,® Swede-O Inc., North Branch, MN), and excess space between the foot and foot cradle was filled with high density padding. We chose this brace over taping and a semi-rigid brace because it offered the best combination of standardized support while still allowing accurate palpation of the malleoli for digitization of the ankle joint center.

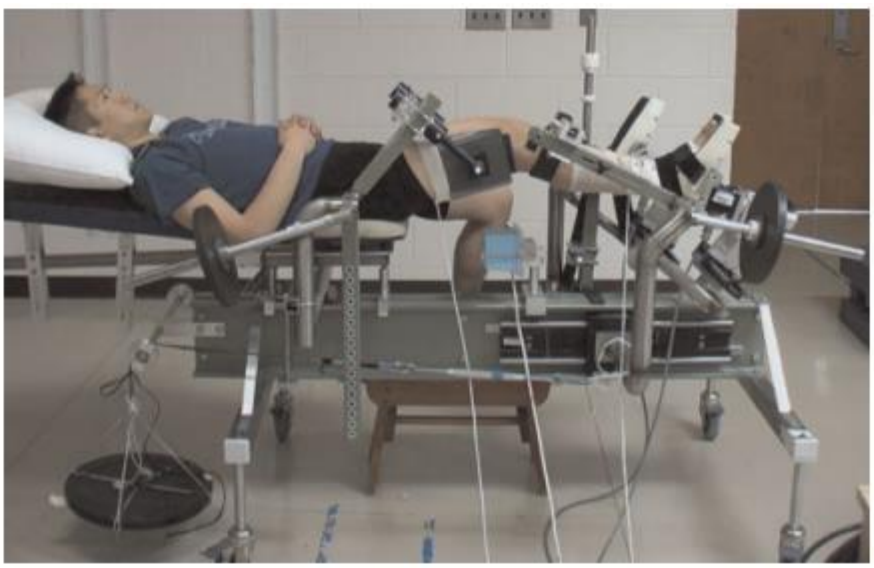

Figure 1. Vermont Knee Laxity Device set up for varus–valgus testing (VKLD; University of Vermont, Burlington, VT).

Prior to data collection, electromagnetic position sensors (Mini Birds, Ascension Technologies, Colchester, VT) were attached to the lateral aspect of the subject's thigh (just proximal to the thigh clamp along the iliotibial band) and the tibial shaft (distal to the shank strap). The VKLD was constructed with fiberglass-reinforced plastic and nonmagnetic 300 series stainless steel to minimize the amount of metal that could potentially interfere with the signal from the sensors. Sensor placements were chosen after qualitatively evaluating a variety of sensor locations during the application of torques to the knee joint, and identified as the locations that resulted in the least skin movement between the sensors and the bone, and therefore best represented the movements of the femur and tibia. Hip, knee, and ankle joint centers of rotation were estimated as previously described using the centroid method.[10] Segmental coordinate systems were constructed by digitizing the greater trochanter and lateral and medial femoral epicondyles for the thigh, and the most medial and lateral parts of the tibial plateau and the medial malleolus for the shank. Following digitization of joint centers and anatomical landmarks, VR–VL and INT–EXT measures where then obtained.

VR–VL Laxity

Prior to testing each subject, a neutral limb position was established. In reference to the knee, the foot cradle was locked in the transverse plane so that the second metatarsal was aligned parallel to the frontal plane, and unlocked in the frontal and sagittal planes to allow free movement of the lower leg. Subjects were asked to straighten their knee and then relax into flexion. The foot cradle was then locked with the knee in 20° flexion, prohibiting motion in all three planes. Knee flexion and VR–VL angles were confirmed (±5°) with a handheld goniometer and real-time angle data obtained from the motion sensors. This position defined the neutral (08) limb position from which all VR–VL measurements began.

VR–VL laxity was measured by unlocking the foot cradle in the frontal plane, and applying force to the medial and lateral aspect of the distal tibia (three finger breaths proximal to the maleoli) with a handheld force transducer (Model SM-50, Interface, Scottsdale, AZ) (Fig. 2). For each subject, the distance from the tibiofemoral joint line to the point of applied force at the tibia was marked and measured (mean distance = 34.2 ± 1.9 cm) to calculate the magnitude of force (mean force = 29.3 ± 1.7 N) necessary to create a 10-Nm VR or VL torque about the knee. The force transducer was attached to a concave molded splint to insure good contact between the tibia and the transducer, and the force was applied at eye level to insure that the direction of the force was directed horizontally and perpendicular to the long axis of the tibia. The signal from the force transducer was amplified through a strain gauge transducer (Model 9820, Interface Advanced Force Measurement) and interfaced with the data collection software. After calibrating the force transducer, three alternating VR–VL loads were applied to the knee to familiarize the subjects with the loading procedure. Three VR–VL loading cycles were then collected with the knee NWB. The limb was returned to the neutral position between each cycle.

INT–EXT Laxity

Neutral limb position was achieved using a similar procedure as previously described. In this case, the foot plate was also unlocked in the transverse plane, allowing unrestricted axial rotation of the tibia. Once subjects straightened and relaxed their knee into 20° of knee flexion, the foot plate was locked in all three planes of motion and the neutral limb position defined. INT–EXT laxity was measured by unlocking the foot cradle to allow transverse plane knee motion, and INT and EXT torques from 0–5 Nm were applied to the knee using a T-handle connected to a six degree-of-freedom (6 DOF) force transducer (Model MC3A, Advanced Medical Technology, Inc.; Watertown, MA), firmly fixed to the foot cradle. (Fig. 3) Through pilot testing, we determined that 5 Nm was the maximum torque that subjects could comfortably tolerate. For INT_{WB} and EXT_{WB}, the foot cradle attachment to the base of the VKLD was unlocked in the horizontal plane, and a compressive force equal to 40% of body weight was applied to the subject's foot prior to applying the INT–EXT torques. This required the subjects to actively maintain the knee in 20° of flexion while the INT–EXT torques were applied. The force transducer was interfaced with the data collection software to simultaneously record force and position data. Three alternating INT–EXT loading cycles were recorded following three familiarization cycles. Prior to each cycle, the limb was returned to the neutral position as previously described.

Figure 2. A handheld force transducer applied resistance to the medial and lateral aspect of the distal tibia to create a 10 Nm varus and valgus torque at the knee.

Data Reduction and Analyses

Position and force data were collected at 100 Hz using commercially available software (Motion Monitor, Innovative Sports Training; Chicago, IL). The signal from the position sensors and both the handheld and 6 DOF force transducers were low-pass filtered at 10 Hz and 20 Hz, respectively, using a 4th order zero lag Butterworth filter. For each limb segment, the +Y axis was directed superiorly, +Z axis directed laterally for the right leg and medially for the left leg, and +X axis directed anteriorly. Euler's equations describe joint motion about the knee using a rotational sequence of Z Y' X". [21]

VR and VL were calculated as the angular displacements (°) produced between 0 and 10 Nm of torque. INT and EXT were calculated as the angular displacements produced between 0 and 5 Nm. Measures were averaged over three cycles. To examine day-to-day measurement consistency, repeated measures ANOVA for each direction (VR, VL, INT_{NWB}, EXT_{NWB}, INT_{WB}, EXT_{WB}) and total motion (VR–VL, INT–EXT_{NWB}, INT–EXT_{WB}) were used to calculate intraclass correlation coefficients ($ICC_{2,k}$) and standard error of measurements (SEM). Absolute measurement errors were quantified by subtracting day 2 from day 1 measures, and constructing 68% and 95% confidence intervals (CI) around the mean difference.[22] Side-to-side differences were examined by subtracting the left from the right value, and constructing 68% and 95% CIs around the mean difference (*N* = 20). To adjust for small sample sizes (i.e., less than 30 subjects), CIs were constructed using the *t*-distribution for 10 subjects, and calculated as follows: 68% CI = (mean difference) ± 1.049 (SD of the difference scores) and 95% CI = (mean difference) ± 2.262 (SD of the difference scores).[22]

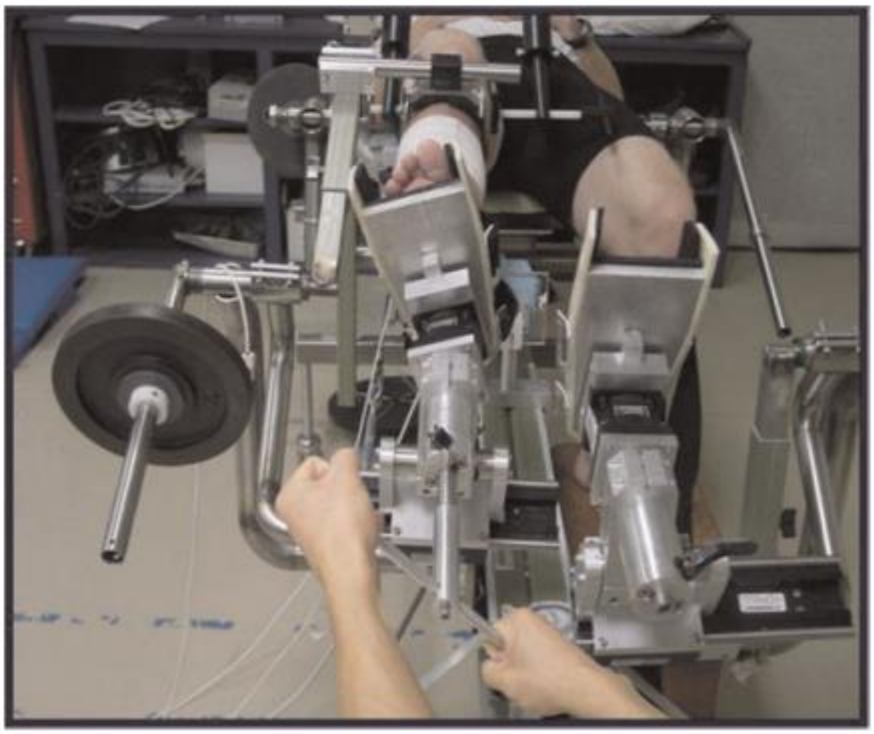

Figure 3. Internal and external rotation torques from 0–5 Nm were applied to the knee using a T-handle connected to a six degree-of-freedom force transducer firmly fixed to the foot cradle.

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RESULTS

Table 1 provides the means, reliability coefficients, and 68% and 95% CIs for all measures on the first 10 subjects. With the exception of INT_{WB} (0.20– 0.32), ICCs ranged between good and excellent (range, 0.68–0.96). Absolute measurement errors were smallest for VR and VL, with 68% and 95% of the subjects having day-to-day measurement differences less than ±1.2° and ±2.5°, respectively. For INT and EXT, absolute measurement errors in 68% and 95% of the subjects, respectively, were less than ±3° and ±7° during non-weight bearing and less than ±2.5° and ±5.5° during weight bearing. The mean absolute difference for all variables from day 1 to day 2 was generally close to zero, indicating little to no measurement bias from one day to the next. Table 2 lists the means and 68% and 95% CIs for side-to-side differences in laxity values for all 20 subjects. Mean left–right differences were also generally close to zero, indicating no systematic left–right differences. In 68% and 95% of the cases, respectively, actual side-to-side differences were less than 1.2° and 2.7° for VR and VL, less than 3° and 7° for INT_{NWB} and EXT_{NWB}, and less than 3° and 6° for INT_{WB} and EXT_{WB}.

Table 1. Means, Standard Deviations (SD), Reliability Coefficients, and 68% and 95% Confidence Intervals for Day-to-Day Measures of VR–VL and INT–EXT Laxities[a]

Variable	Day 1 Mean ± SD (°)	Day 2 Mean ± SD (°)	ICC	SEM (°)	Mean Difference ± 68% CI (°)	Mean Difference ± 95% CI (°)
Valgus						
Right	5.0 ± 1.9	5.4 ± 2.3	0.92	0.66	0.4 ± 1.2	0.4 ± 2.5
Left	5.4 ± 2.1	5.6 ± 1.9	0.96	0.43	−0.2 ± 0.9	−0.2 ± 1.9
Varus						
Right	4.0 ± 1.6	4.2 ± 0.9	0.91	0.47	−0.2 ± 0.8	−0.2 ± 1.7
Left	4.5 ± 1.1	5.0 ± 1.2	0.68	0.68	0.5 ± 1.2	0.5 ± 2.5
Total varus–valgus						
Right	8.9 ± 3.3	9.6 ± 3.0	0.96	0.67	0.6 ± 1.2	0.6 ± 2.7
Left	9.9 ± 2.7	10.5 ± 2.9	0.91	0.87	0.7 ± 1.6	0.7 ± 3.6
Internal Rotation (NWB)						
Right	8.2 ± 3.7	9.0 ± 4.8	0.93	1.28	−0.8 ± 2.3	−0.8 ± 5.0
Left[b]	10.6 ± 3.8	10.4 ± 5.2	0.89	1.56	−0.2 ± 3.1	−0.2 ± 6.7
External Rotation (NWB)						
Right	12.1 ± 4.0	11.9 ± 3.8	0.88	2.50	−0.2 ± 2.8	−0.2 ± 6.1
Left	11.8 ± 5.1	13.2 ± 3.5	0.86	2.99	−1.5 ± 3.0	−1.5 ± 6.6
Total INT–EXT Rotation (NWB)						
Right	20.3 ± 6.4	20.9 ± 7.1	0.91	2.14	−0.6 ± 4.2	−0.6 ± 9.1
Left[b]	22.1 ± 8.3	23.0 ± 6.7	0.89	2.80	0.9 ± 5.2	0.9 ± 11.2
Internal Rotation (WB)						
Right	3.6 ± 1.6	2.9 ± 1.1	0.32	1.80	0.7 ± 2.2	0.7 ± 4.7
Left	3.4 ± 2.0	3.6 ± 1.5	0.20	1.70	0.3 ± 2.5	0.3 ± 5.3
External Rotation (WB)						
Right	6.9 ± 3.7	5.7 ± 2.8	0.85	1.45	−1.2 ± 2.4	−1.2 ± 5.1
Left	5.9 ± 3.7	6.0 ± 3.4	0.88	1.26	−0.1 ± 2.5	−0.1 ± 5.4
Total INT–EXT Rotation (WB)						
Right	10.6 ± 4.9	8.7 ± 2.8	0.70	2.64	1.9 ± 3.8	1.9 ± 8.2
Left	9.3 ± 5.3	9.6 ± 4.5	0.75	2.67	0.4 ± 4.8	0.4 ± 10.3

[a]All values represent angular displacements (degrees). $N = 10$.
[b]$N = 9$; one subject excluded due to poor tibial internal rotation data.

Table 2. Mean Differences and 68% and 95% Confidence Intervals for Left-to-Right Measurement Differences for VR–VL and INT–EXT Knee Laxities[a]

Variable	Right Side Mean ± SD (°)	Left Side Mean ± Sd (°)	Mean Difference ±68% CI (°)	Mean Difference ±95% CI (°)
Valgus	4.9 ± 1.8	5.6 ± 2.1	−0.8 ± 1.1	−0.8 ± 2.4
Varus	4.1 ± 1.4	4.7 ± 1.0	−0.6 ± 1.2	−0.6 ± 2.7
Total varus–valgus	9.0 ± 2.9	10.3 ± 2.6	−1.4 ± 1.6	−1.4 ± 3.4
Internal rotation (NWB)	9.9 ± 4.5	10.8 ± 4.0	−1.0 ± 3.1	−1.0 ± 6.8
External rotation (NWB)	13.6 ± 3.8	13.4 ± 4.6	0.3 ± 2.8	0.3 ± 5.9
Total internal–external rotation (NWB)	23.5 ± 7.1	24.2 ± 7.7	−0.7 ± 5.4	−0.7 ± 11.7
Internal rotation (WB)	4.0 ± 1.8	3.9 ± 2.1	0.1 ± 1.7	0.1 ± 3.6
External rotation (WB)	7.0 ± 3.5	6.5 ± 3.8	0.5 ± 2.8	0.5 ± 6.0
Total internal–external rotation (WB)	11.0 ± 5.1	10.4 ± 5.5	0.6 ± 3.9	0.6 ± 8.4

[a]All values represent angular displacements (degrees). $N = 20$.

DISCUSSION

This study introduces the use of the VKLD to quantify VR–VL and INT–EXT laxities in vivo. Our primary findings confirm that consistent measures can be obtained from one day to the next, with absolute measurement errors less than 2.5° for VR and VL, and less than 5–7° for INT and EXT in 95% of the subjects. Although side-to-side differences were observed, they did not exceed the absolute measurement errors calculated for each variable.

Before initiating studies that use the VKLD to measure VR–VL and INT–EXT laxities, our desire was to confirm its measurement capabilities, given the reported challenges in stabilizing the thigh within the measurement system in vivo compared to cadaveric studies where the femur and tibia can be rigidly fixed.[17] Measurement consistency was good-to-excellent for all measures (0.68–0.96), with the exception of INTWB which was substantially lower (0.32 and 0.20), and likely affected reliability of total INT–EXT WB (0.70 and 0.75). The only other ICC below 0.85 was left VR (0.68). ANOVA results used to compute the ICCs for INT_{WB} and left VR indicated that poor reliability was primarily due to a small proportion of between-subjects variance (i.e., the range in subject values was less than 58for INT_{WB}, and 4° for left VR), rather than an increase in systematic or random measurement error. In fact, the measurement errors for INT_{WB} and left VR (determined both by SEM and the absolute measurement errors) were quite consistent with other INT/EXT and VR/VL measures, respectively (see Table 1). The lower ICC for left compared to right VR may simply be due to the slightly smaller range in subject values for left versus right, which would effectively increase the proportion of variance due to error on the left side. These findings suggest that the VKLD yields equally repeatable measures of INT_{WB} and left VR within subjects across time when compared to other laxity measures, but that comparison between subjects may be more challenging due to the smaller variations in subject scores. In these cases, total INT–EXT_{WB} and VR–VL laxity may yield more clinically useful comparisons.

Our sample was relatively small (N = 10), and it is possible that the distribution of scores in our sample are not reflective of the larger population, which may affect some of the ICC values. Hence, we also quantified absolute measurement error by constructing 68% and 95% CIs around the mean difference in values between tests, which are not dependent on the distribution of scores in the sample.[23,24] These CIs are useful when making clinical decisions as to whether the magnitude of the measurement error is acceptable based on the expected range in the value. VL and VR laxities varied less than 2.4° and 2.7°, respectively, from one test to the next in 95% of the cases. Measurement errors for INT and EXT laxities were higher, with 95% of the cases generally varying less than 6° from one test to the next. However, it is important to note that these values represent the *largest* expected measurement error, with the majority of the subjects (~70%) having absolute measurement errors that were much lower (see 68% CIs). With the exception of INT_{WB}, these absolute measurement errors were well within the clinical ranges for each value (typically representing less than 10%–20% of the total range of motion), and suggest the VKLD has sufficient measurement precision to identify true clinical differences of approximately 1–2° in VR and VL motion and approximately 2–3° in INT and EXT motion in 68% of the cases. Power calculations based on the most conservative estimates (i.e., using the highest standard deviation of the two samples, and taking the unusual step of correcting for the reliability of the measure[25]) confirmed this, revealing minimum detectable mean differences for between-subject comparisons of 1–2° for VR and VL, 3–4° for INT_{NWB} and EXT_{NWB}, and ~3° for INT_{WB} and EXT_{WB}, and for within-subject comparisons across time of < 1° for VR and VL, 1.5–2.5° for INT_{NWB} and EXT_{NWB}, 2–3° for INT_{WB}, and 1.5° for EXT_{NWB} (calculations based on 80% power and a fixed sample size of $N = 20$). Except for INT_{WB}, the precision of these measures appear to be adequate and allow the detection of clinically meaningful differences with reasonable sample sizes. Even so, we are working to further improve measurement precision by increasing the amount of familiarization so that subjects have the opportunity to become comfortable with the testing procedures prior to the first test session. This may be particularly helpful in improving measurement stability for INT–EXT_{WB}, where subjects are required to actively maintain the knee at 20° flexion while INT–EXT torques are applied to the knee.

To further analyze the quality of our data, we compared our laxity values with cadaveric reports where rigid fixation of position sensors to the femur and tibia could be achieved. Our VR and VL measures were very consistent with Markolf et al.[26] who reported approximately 10° of total VR–VL laxity at 20° knee flexion with a 10-Nm load, and with Hsu et al.[27] who reported similar VL values (males = 4.0°; females = 5.7°). While our values are greater than other in vivo studies measuring total VR–VL at 20° of knee flexion (range, 3–6.5°),[16,18] it is difficult to compare findings because different measurement devices and forces were used. For total INT–EXT_{NWB}, our values were somewhat lower than those reported by Markolf et al. at 20° knee flexion with a 5-Nm load (34.9 ± 3.6),[26] but were consistent with other measures with the knee flexed 20–30° (range, 18.6–27°).[13,27–29] Our NWB findings also compared favorably with in vivo work using computed tomography (INT = 10.88; EXT = 7.48).[19] When comparing INT–EXT_{NWB} and INT–EXT_{WB}, we observed a 58%–60% reduction in laxity values with joint loading, which was consistent with one cadaveric study (58%)[28] but substantially greater than others (range, 20%–25%).[26,29] Greater reductions would be expected in vivo, given the added contribution of muscle activity. Collectively, these findings provide good support that the VKLD yields both valid and reliable measures of VR–VL and INT–EXT laxities.

Assessing bilateral symmetry allows one to determine if the laxity measured on one limb is sufficiently representative of the contralateral limb. In approximately 70% of our subjects, side-to-side differences were < 1.6° for VR–VL, and < 2–3° for INT–EXT_{NWB} and INT–EXT_{WB}. While our VR–VL results are in close agreement with Markolf et al.[17] who reported left–right differences of 1–2° in 70% of the subjects (measured at full knee extension), we found no comparative studies reporting side-to-side differences in INT–EXT laxities in healthy knees.

Comparison of Tables 1 and 2 indicates that the magnitude of side-to-side differences were indistinguishable from the absolute measurement errors recorded for each variable. Hence, it is difficult to determine from these data if true side-to-side differences exist, or whether these differences simply reflect measurement error. Based on our data, side-to-side differences would need to exceed approximately 2.5° for VR and VL and 4–7° for INT and EXT for one to be 95% confident they reflect true differences. Because the majority of our subjects had side-to-side differences less than these values, these data support that VR–VL and INT–EXT laxities taken on one side of the body are adequately representative of the contralateral side when using these measurement techniques. Larger sample sizes should be studied to confirm these findings.

In conclusion, the VKLD yields consistent measures of VR–VL and INT–EXT laxities with adequate measurement precision to identify clinically meaningful differences. Although side-to-side differences were observed, the magnitude of these differences did not exceed the absolute measurement error. These findings are limited to 20° knee flexion, and different results may exist at other knee angles. While we took great care to achieve good immobilization of the femur and tibia during these tests, we cannot rule out that some motion may have occurred within the stabilizing constraints of the device. While this is a known limitation of in vivo models,[17] our results suggest this did not produce a major source of measurement error, as our values were quite reproducible from day to day, and were in general agreement with values obtained from cadaveric models where motion sensors are rigidly fixed to the bone.

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