The Interday Measurement Consistency of and Relationships Between Hamstring and Leg Musculo-articular Stiffness

By: Justin P. Waxman, Randy J. Schmitz, Sandra J. Shultz

Waxman JP, Schmitz RJ, Shultz SJ. The Interday Measurement Consistency of and Relationship Between Hamstring and Leg Musculoarticular Stiffness. Journal of Applied Biomechanics. 2015; 31(5):340-348.

Abstract:

Hamstring stiffness (K_{HAM}) and leg stiffness (K_{LEG}) are commonly examined relative to athletic performance and injury risk. Given these may be modifiable, it is important to understand day-to-day variations inherent in these measures before use in training studies. In addition, the extent to which K_{HAM} and K_{LEG} measure similar active stiffness characteristics has not been established. We investigated the interday measurement consistency of K_{HAM} and K_{LEG}, and examined the extent to which K_{LEG}predicted K_{HAM} in 6 males and 9 females. K_{HAM} was moderately consistent day-to-day ($ICC_{2,5}$ = .71; SEM = 76.3 $N{\cdot}m^{-1}$), and 95% limits of agreement (95% LOA) revealed a systematic bias with considerable absolute measurement error (95% LOA = 89.6 ± 224.8 $N{\cdot}m^{-1}$). Day-to-day differences in procedural factors explained 59.4% of the variance in day-to-day differences in K_{HAM}. Bilateral and unilateral KLEG was more consistent ($ICC_{2,3}$ range = .87–.94; SEM range = 1.0–2.91 $kN{\cdot}m^{-1}$) with lower absolute error (95% LOA bilateral= –2.0 ± 10.3; left leg = –0.36 ± 3.82; right leg = –1.05 ± 3.61 $kN{\cdot}m^{-1}$). K_{LEG} explained 44% of the variance in K_{HAM} ($P < .01$). Findings suggest that procedural factors must be carefully controlled to yield consistent and precise K_{HAM} measures. The ease and consistency of K_{LEG}, and moderate correlation with K_{HAM}, may steer clinicians toward K_{LEG} when measuring lower-extremity stiffness for screening studies and monitoring the effectiveness of training interventions over time.

Keywords: spring-mass model | free-oscillation | musculo-articular stiffness | leg stiffness | musculoskeletal

Article:

Measures of lower-extremity stiffness have often been examined for their relative contribution to performance and injury risk.[1–4] Stiffness describes the relationship between an applied load and the amount of elastic deformation that occurs within a given structure.[4] In terms of the human body, stiffness can be described from the macroscopic level of the whole body all the way to the microscopic level of a single muscle fiber. Although a number of in vitro and in vivo methods exist, 2 in vivo measures commonly employed are leg stiffness (K_{LEG}) and hamstring musculo-articular stiffness (K_{HAM}).

K_{HAM} and K_{LEG} are measures of active stiffness that rely on modeling the hamstring muscle-tendon unit and entire lower extremity, respectively, as mass-spring systems.[5–8] K_{HAM} is assessed using the free-oscillation technique,[6] whereby the hamstring muscle-tendon unit represents the massless linear spring and the lower leg (shank) and applied load represent the mass; a perturbation is then applied to the joint and the ensuing oscillations are measured.[7,8] Similarly, K_{LEG} is commonly assessed using functional bilateral or unilateral hopping test protocols, whereby the leg(s) represents the massless linear spring and the rest of the body represents the mass.[5,9] Given the in vivo nature of these methods, the values obtained represent global measures of stiffness for the system being modeled, which includes the stiffness of the muscle-tendon unit(s), skin, ligaments, bones, and articular capsule(s).[1,10]

Evidence suggests that these measures of lower-extremity stiffness are related to both athletic performance and injury risk. K_{LEG} has previously been shown to be positively related to running performance.[11,12] In addition, both K_{LEG} and K_{HAM} are positively related to concentric and eccentric rate of force development.[13–16] It has also been demonstrated that anterior cruciate ligament (ACL) deficient individuals with higher K_{HAM} possess greater knee functional stability than those with more compliant hamstrings,[8] and that uninjured healthy individuals with higher K_{HAM} display characteristics associated with reduced ACL loading during controlled perturbations and dynamic landing tasks.[17,18] Further, ACL-reconstructed individuals with higher levels of K_{LEG} have been reported to be more functional and able to participate in more demanding physical activity with fewer symptoms of joint pain and instability.[19] Thus, it appears that higher levels of K_{LEG} and K_{HAM} may be beneficial for enhancing performance and reducing injury risk. However, other research suggests that there may actually be an optimal range for stiffness, where either too much or too little stiffness may lead to increased risk of bony injuries[20–22] or soft tissue injuries,[23,24] respectively. Although additional work is needed to gain a greater understanding of how stiffness measures are related to performance and injury risk, the ability to modify this neuromechanical property makes stiffness an important variable to consider from injury prevention, rehabilitation, and performance perspectives.

Because of the potential for stiffness to be modified through training,[25–27] it is important to understand the inherent day-to-day consistency and precision of these measures before progressing to intervention strategies. The reliability of K_{LEG} has previously been established, with reported interday intraclass correlation coefficients (ICCs) and coefficients of variation (CV) ranging from .82 to .94 and from 4.2% to 13.9%, respectively, during bilateral and unilateral hopping tasks.[28–30] However, less is known about the reliability of K_{HAM}. Of the 5 investigations that have examined KHAM measurement consistency, 4 studies[17,25,31,32] reported intraday consistency, while only 2 studies[16,25] examined interday consistency. Blackburn and Norcross25 examined the day-to-day measurement consistency of K_{HAM} over a 6-week time period and reported a reliability estimate (ICC) of .86 and measurement precision (standard error of measurement [SEM]) of 1.69 $N/m \cdot kg^{-1}$. Similarly, Ditroilo et al[16] examined the day-to-day measurement consistency of K_{HAM} within a 1-week time period and reported ICCs ranging from

.66 to .89. Ditroilo et al[16] also reported the CV as an index of measurement precision, which ranged from 5.1% to 13.1% under different methodological conditions. Given the more moderate estimates of K_{HAM} day-to-day measurement consistency and precision compared with K_{LEG}, methodological factors inherent to the measurement of K_{HAM} (ie, subtle variations in perturbation magnitude and hip and knee joint angles) may, in part, impact measurement precision. To the best of our knowledge, however, the influence of such methodological factors on the measurement precision of K_{HAM} has not yet been investigated. Furthermore, we are unaware of any studies that have examined the extent to which K_{HAM} and K_{LEG} are related. Although K_{HAM} and K_{LEG} are assessed using fundamentally different procedures (eg, closed vs. open kinetic chain), both measures are based on the underlying construct that the muscle-tendon unit(s) can be independently or collectively modeled as a linear mass-spring system.[7] Further, because K_{HAM} is inherently a more difficult measurement technique for both the participant and researcher, having a simpler, clinically accessible measure of lower-extremity stiffness could be of great value when attempting to evaluate large samples in a minimal amount of time. Therefore, the primary purposes of this study were to: (1) assess the interday measurement consistency of K_{HAM} and K_{LEG} and (2) examine the relationships between these measures. A secondary purpose of this study was to examine potential procedural factors that could introduce sources of measurement error in K_{HAM}. Based on previously published work, we hypothesized that K_{HAM} and K_{LEG} would both display acceptable interday measurement consistency but that K_{LEG} would have superior consistency given the greater relative simplicity of the measure. In addition, we hypothesized that K_{HAM} and K_{LEG} would be moderately correlated with one other given that both measures are based on the same underlying construct.

METHODS

Participants

Fifteen healthy college-aged individuals (6 male, 9 female; age = 21.5 ± 2.5 years, height = 1.7 ± 0.1 m, mass = 69.3 ± 11.0 kg) volunteered to participate. Participants were: (1) recreationally active (≥ 90 minutes physical activity per week), (2) free from lower-extremity injury for a minimum of 6 months before participation, (3) free from any history of lower-extremity surgery, (4) without known medical conditions that would affect connective tissue, and (5) without any history of cardiovascular or pulmonary problems. To ensure that all participants met the inclusion criteria, physical activity and health history questionnaires were administered before enrollment in the study. Before participation, participants signed an informed consent approved by the university's institutional review board.

Procedures

All participants visited the laboratory for testing on 2 separate occasions separated by 2–5 days to minimize the risk of carryover effects between testing sessions (eg, muscle fatigue, soreness).[33] All testing sessions were administered by the same investigator (JW). Upon arrival to each testing session, participants were outfitted with laboratory compression shorts and an athletic top, and measures of body height and mass were obtained. Participants then performed a 5-minute warm-up on a stationary cycle ergometer (Life Fitness, Schiller Park, IL) at a cadence of 70–80 revolutions per minute and a target rating of perceived exertion of ≥ 3 on a Borg CR-10 scale.[34] After the warm-up, the following assessments were performed in identical fashion during both testing sessions: (1) K_{HAM}, (2) bilateral K_{LEG}, and (3) unilateral K_{LEG}.

K_{HAM} was measured using the free-oscillation technique, whereby the leg is modeled as a single degree of freedom mass-spring system. The damping effect that the hamstring muscles impose on oscillatory flexion/extension of the knee joint is then quantified, following a perturbation.[8,18,31] Because K_{HAM} does not differ between limbs in healthy individuals,[35] all data were obtained from the left leg. Participants were instrumented with clusters of 4 optical LED markers (Phase Space, San Leandro, CA), each placed on the lateral thigh and shank of the left limb. Three-dimensional kinematic data were obtained via an 8-camera IMPULSE motion tracking system (Phase Space, San Leandro, CA). Participants were positioned prone, with the trunk and thigh supported in 30° of hip flexion and the lower leg free to move (Figure 1). A thermoplastic splint was secured to the plantar aspect of the foot and posterior shank to standardize ankle position at approximately 90°. A 10% body mass load was then attached to the distal shank, at the level of the malleoli, using cuff-style ankle weights. The investigator then passively positioned the participant's shank parallel to the floor, placing the knee in approximately 30° of flexion, and the participant was required to hold this position via isometric hamstring contraction. Within 5 seconds of the participant holding this position, an anterior perturbation was manually applied to the posterior aspect of the calcaneus, resulting in slight knee extension and subsequent damped oscillatory knee flexion and extension.[17,18,31,36] This damped oscillatory motion was characterized as the tangential acceleration of the shank and foot segment, captured via a triaxial accelerometer (sensor dimensions: 2.54 × 2.54 × 1.91 cm; NeuwGhent Technology, Lagrangeville, NY) attached to the thermoplastic splint. Participants were verbally instructed not to interfere with or voluntarily produce the oscillations following the perturbation, and to attempt to keep the hamstring muscles active only to a level necessary to support the mass of the shank and foot segment, and the applied load, in the testing position.[17,31] Following 3–5 practice trials, 5 test trials were recorded for analysis. Test trials were separated by 30-second rest intervals to reduce the likelihood of fatigue.

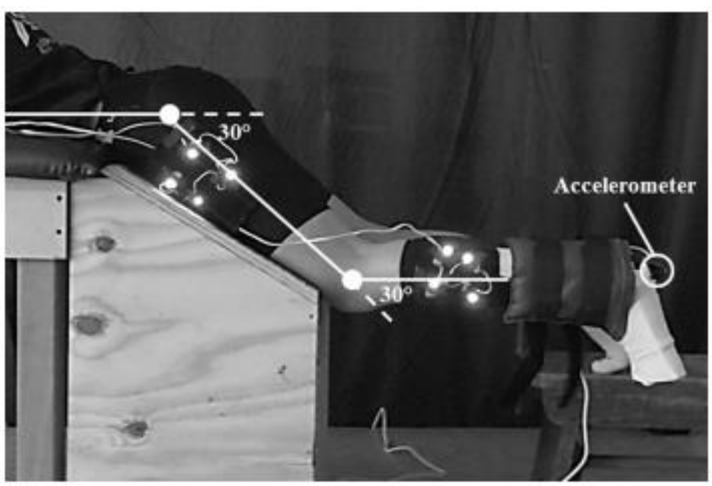

Figure 1 — Participant positioning during hamstring musculo-articular stiffness (K_{HAM}) assessments.

K_{LEG} was assessed via barefoot hopping in place on a force plate (Type 4060–130; Bertec Corporation, Columbus, OH) at a hopping frequency of 2.2 Hz under 3 different conditions: (1) bilateral, (2) unilateral left leg, and (3) unilateral right leg. Bilateral hopping was always performed first. Unilateral hopping order was assigned in a counterbalanced fashion; once order was assigned, participants alternated legs each trial to minimize the likelihood of fatigue. Participants performed unilateral hopping trials in the same order during both testing sessions. Participants were verbally instructed to stand tall with their hands placed on their hips and their eyes looking straight ahead. Once the metronome began to sound, participants were instructed to begin hopping while attempting to synchronize their hops with the tone. Because variations in ground contact time can affect stiffness regulation at a given hopping frequency,[37] participants were asked to hop with as short a ground contact time as possible. Before data collection, participants were allowed to practice until they felt comfortable with the task. Three 10-second trials, separated by 30-second rest intervals, were then recorded for each condition. At 2.2 Hz, the interval between vertical ground reaction force peaks should be 455 milliseconds. Peak ground reaction force intervals that fell within 5% (± 23 milliseconds) of 455 milliseconds were considered valid data to be used for comparisons.[38]

Data Sampling and Reduction

Accelerometer and force plate data were sampled at 1000 Hz, whereas kinematic data were sampled at 240 Hz. Kinetic and kinematic instrumentation was time synchronized and interfaced with Motion Monitor software (Innovative Sports Training, Chicago, IL). Left limb kinematics were modeled using Motion Monitor software. All data were later exported from Motion Monitor software and processed using custom written Matlab code (Mathworks, Inc., Natick, MA). Accelerometer and force plate data were low-pass filtered at 10 Hz, while kinematic data were low-pass filtered at 12 Hz, using a fourth-order zero-lag Butterworth filter.

Hamstring Stiffness (K_{HAM}). The time interval between the first 2 oscillatory peaks (t_1 and t_2) of the accelerometer time series was used to calculate the damped frequency of oscillation ($1/[t_2 - t_1]$) for each trial (Figure 2). K_{HAM} was then calculated using the equation, $K_{HAM} = 4\pi^2 m f^2$, where m is the summed mass of the shank and foot segment (6.1% body mass)[39] and the applied load (10% body mass), and f is the damped frequency of oscillation.[35] In addition, the following kinetic and kinematic variables were obtained from each K_{HAM} trial to examine the influence of procedural factors on measurement error: (1) perturbation magnitude, (2) initial knee flexion angle, and (3) initial tibia rotation angle. Perturbation magnitude was defined as the product of the peak downward acceleration immediately following perturbation onset and the total system mass (shank and foot segment weight + applied load [N]). Perturbation onset was defined as the instant at which downward acceleration of the shank and foot segment exceeded 3 standard deviations of the mean acceleration before the perturbation. Initial knee flexion and tibia rotation angles were defined as the average angles (in degrees) obtained 100 milliseconds before perturbation onset. All variables were then averaged across 5 trials and used for analysis.

Leg Stiffness (K_{LEG}). K_{LEG} was calculated from the vertical ground reaction force and the effective ground contact time.[9] Effective contact time is defined as the amount of time that the vertical ground reaction force is greater than body weight during the stance phase of hopping, and is expressed as *T/2*, where *T* equals the period of oscillation (Figure 3).[3,9,23] From this, the natural frequency of oscillation (ω) and spring constant of the spring-mass system were calculated as, $K_{LEG} = Mg\omega^2$, with $\omega = 2\pi/T$ and where M is the participant's body mass. Data collected during 5 consecutive hops, between the sixth and tenth hop of 15 hops, were used to calculate K_{LEG} for each trial.[40,41] Ground contact time (t_c) and flight time (t_f) were determined using the vertical ground reaction force. The beginning and end of ground contact was defined as the instants at which the vertical ground reaction force was above and below 5 N. The 3 trials for each condition were then separately averaged and used for analysis.

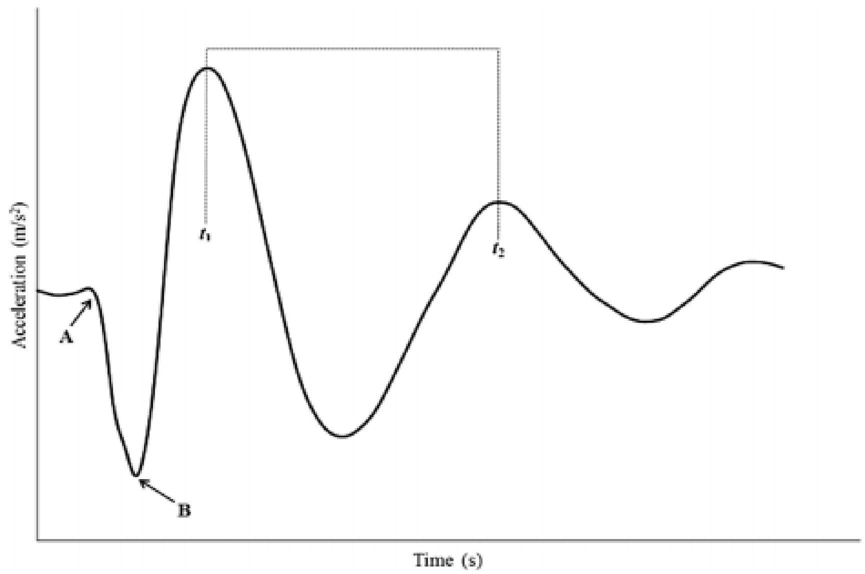

Figure 2 — Example time series data obtained from the accelerometer during a hamstring musculo-articular stiffness (K_{HAM}) assessment. Point A represents perturbation onset. Point B represents peak downward acceleration resulting from perturbation. t_1 and t_2 represent the time points at which the first 2 oscillatory peaks occur; these time points are then used to calculate the frequency of oscillation.

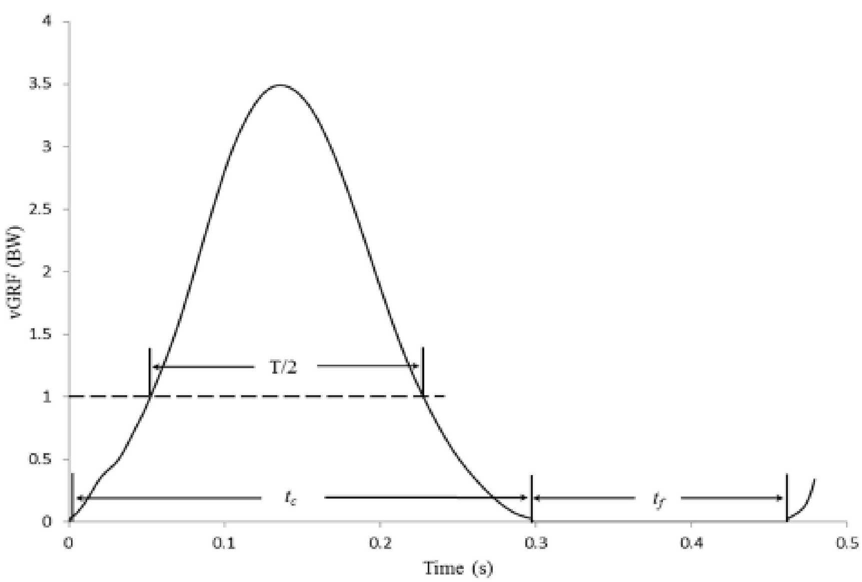

Figure 3 — Typical example of a vertical ground reaction force (vGRF) time curve during hopping at 2.2 Hz. K_{leg} was calculated by measuring one-half of a resonant period ($T/2$) and then using the equation listed above. The $T/2$ is the amount of time that the vGRF is greater than one body weight (BW). Ground contact time (t_c) and aerial/flight time (t_f) were also calculated for each hop to evaluate hopping frequency.

Statistical Analysis

All statistical analyses were performed using SPSS version 20.0 for Windows (IBM Inc., Chicago, IL). Interday measurement consistency was determined using a repeated-measures analysis of variance (ANOVA) and calculating ICCs, using the $ICC_{2,k}$ model as described by Shrout and Fleiss.[42] Although there is no clear consensus on ICC interpretation, it is generally accepted that ICC values greater than .75 represent good measurement consistency, whereas ICCs less than .75 reflect moderate to poor consistency.[43] Precision of measurement was then evaluated by computing the SEM and LOA for each variable. SEM was calculated as: $SEM = SD\sqrt{(1-ICC)}$, where SD is the sample standard deviation. LOA were obtained by calculating the mean and standard deviation of the differences (SD_{diff}) between testing sessions (ie, day 2 – day 1). We then calculated 68% and 95% confidence limits around the mean difference as the mean difference $\pm 1 \cdot SD_{diff}$ and $\pm 1.96 \cdot SD_{diff}$, respectively.[44] To examine the extent to which day-to-day differences in K_{HAM} could be attributed to day-to-day differences in procedural factors (ie, perturbation magnitude and initial knee flexion and tibia rotation angles), a multiple linear regression analysis was conducted on the test-retest difference scores (day 2 – day 1) for each measure. Finally, simple linear regression was used to determine the extent to which K_{LEG} predicted K_{HAM} on the first day of testing. This regression analysis was performed on body mass normalized values to account for the influence of body size. In addition, because K_{HAM} was assessed only on the left leg, K_{LEG} during unilateral left leg hopping was used as the predictor.

RESULTS

Interday measurement consistency ($ICC_{2,k}$) and precision (SEM and LOA) and test-retest means ± standard deviations for K_{HAM} and K_{LEG} are presented in Table 1 and Table 2, respectively.

K_{HAM} demonstrated moderate interday measurement consistency ($ICC_{2,5} = .71$) with an SEM (76.27 $N \cdot m^{-1}$) that represented 7.67% of the overall mean value (993.82 $N \cdot m^{-1}$). LOA analysis (absolute measurement error) revealed a relatively large systematic bias (mean difference from day 1 to day 2 = 89.57 $N \cdot m^{-1}$), with considerable variability (± 224.8 $N \cdot m^{-1}$) around this mean difference (Table 1, Figure 4). Specifically, in 95% of the cases, the actual test-retest difference for KHAM could be expected to range from 135.3 $N \cdot m^{-1}$ lower to 314.4 $N \cdot m^{-1}$ higher on day 2 as compared with day 1. In contrast, K_{LEG} demonstrated good to excellent measurement consistency ($ICC_{2,3}$ range = .87–.94) and relatively low SEMs (range = 1.00–2.91 $kN \cdot m^{-1}$) across all 3 hopping conditions (Table 2). LOA analysis revealed little systematic bias between testing sessions (bilateral = –2.0, left leg = 0.4, right leg = –1.1 $kN \cdot m^{-1}$), and in 95% of the cases (95% LOA), the actual test-retest differences for bilateral and unilateral K_{LEG} ranged from being 12.3 $kN \cdot m^{-1}$ and 4.7 $kN \cdot m^{-1}$ lower to 8.3 $kN \cdot m^{-1}$ and 2.5 $kN \cdot m^{-1}$ higher on day 2 as compared with day 1 (Table 2, Figure 5).

Table 1 Interday reliability results for hamstring musculo-articular stiffness (K_{HAM}) variables

	Day 1 Mean ± SD	Day 2 Mean ± SD	$ICC_{2,5}$	SEM	Mean Difference 68% LOA	95% LOA
K_{HAM} ($N \cdot m^{-1}$)	949.03 ± 134.13	1038.61 ± 141.83	.71	76.27	89.57 ± 114.70	89.57 ± 224.82
Perturbation magnitude (N)	139.60 ± 22.86	138.59 ± 22.33	.99	0.94	–1.01 ± 1.61	–1.01 ± 3.17
Initial knee flexion angle (°)	38.38 ± 4.89	41.73 ± 5.85	.82	2.48	3.35 ± 3.22	3.35 ± 6.32
Initial tibial rotation angle (°)	–0.27 ± 8.54	0.92 ± 8.63	.93	2.34	1.19 ± 4.48	1.19 ± 8.78

Abbreviations: SD = standard deviation; ICC = intraclass correlation coefficient; SEM = standard error of measurement; LOA = limits of agreement.
Note. The term mean difference denotes day 2 – day 1 mean differences.

Table 2 Interday reliability results for bilateral and unilateral leg stiffness (K_{LEG}) measures

	Day 1 Mean ± SD	Day 2 Mean ± SD	$ICC_{2,3}$	SEM	Mean Difference 68% LOA	95% LOA
K_{LEG} ($kN \cdot m^{-1}$)						
Bilateral	37.23 ± 8.13	35.23 ± 7.99	.87	2.91	–2.00 ± 5.27	–2.00 ± 10.32
Left leg	25.45 ± 4.13	25.09 ± 4.24	.94	1.00	–0.36 ± 1.95	–0.36 ± 3.82
Right leg	25.65 ± 4.16	24.59 ± 4.57	.94	1.10	–1.05 ± 1.84	–1.05 ± 3.61

Abbreviations: SD = standard deviation; ICC = intraclass correlation coefficient; SEM = standard error of measurement; LOA = limits of agreement.
Note. The term mean difference denotes day 2 – day 1 mean differences.

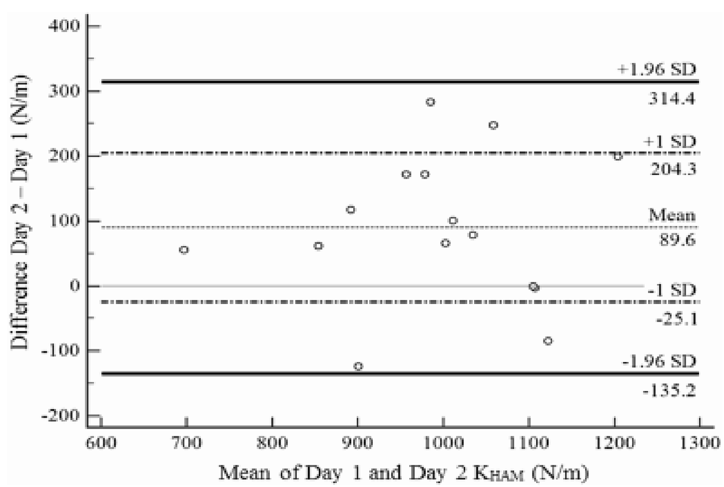

Figure 4 — Interday reliability for hamstring musculo-articular stiffness (K_{HAM}). This Bland-Altman plot shows the difference between values for the first and second day of testing on the vertical axis, and the mean of the first and second day of testing on the horizontal axis for each participant during K_{HAM} assessments. The thin dashed line represents the mean difference (bias) between testing days (day 2 – day 1), while the thick dashed lines and solid lines represent the 68% and 95% limits of agreement, respectively. The mean difference is greater than zero, indicating systematic bias between the first and second day of testing.

All procedural factors associated with the K_{HAM} assessment demonstrated good to excellent day-to-day measurement consistency (Table 1). Although we attempted to carefully standardize these procedural factors during the assessment of K_{HAM}, repeated measures ANOVA and LOA analyses revealed a small systematic bias for peak perturbation magnitude and initial knee flexion angle, where peak perturbation magnitude values were on average 1.0 N lower (P = .03) and initial knee flexion angles were on average 3.4° higher (*P* < .01) on day 2 as compared with day 1 (Table 1). In 95% of the cases, the actual test-retest differences for peak perturbation magnitude and initial knee flexion angle could be expected to range from 4.2 N lower to 2.2 N higher, and from 3.0° lower to 9.7° higher, respectively.

When examining the extent to which day-to-day measurement error in K_{HAM} could be attributed to day-to-day changes in procedural factors, multiple linear regression analysis revealed that the linear combination of the difference scores (day 2 – day 1) for peak perturbation magnitude, and initial knee flexion and tibia rotation angles, explained 59.4% of the variance in the difference score for K_{HAM} ($P = .02$). The regression equation for the 3-predictor model was: predicted $\Delta K_{HAM} = 72.06 + 36.66(\Delta\text{perturbation magnitude}) + 21.60(\Delta\text{initial knee flexion angle}) - 15.05(\Delta\text{initial tibia rotation angle})$. All 3 predictors uniquely contributed to the overall model after controlling for all other predictors (Table 3).

When examining the relationship between K_{HAM} and K_{LEG}, K_{LEG} explained 44.2% of the variance in K_{HAM} ($P < .01$), with a prediction equation of $K_{HAM} = 2.093 + 0.665(K_{LEG})$.

Table 3 Results of multiple linear regression analysis examining the extent to which day-to-day differences in musculo-articular stiffness (K_{HAM}) could be attributed to day-to-day differences in procedural factors

	Unstandardized Coefficients			Correlations		
Model	**B**	**Standard Error**	***P***	**Zero-Order**	**Partial**	**Semipartial**
Constant	72.057	35.648	.068			
Δ Perturbation magnitude	36.662	14.441	.028	.404	.608	.488
Δ Initial knee flexion angle	21.599	7.386	.014	.453	.661	.562
Δ Initial tibia rotation angle	–15.050	5.578	.021	–.188	–.631	–.518

Multiple $R^2 = .594$; $F(3,14) = 5.369$; $p = .016$.
Δ = day 2 – day 1 difference.

DISCUSSION

Previous research suggests that higher levels of hamstring and leg musculo-articular stiffness (K_{HAM} and K_{LEG}, respectively) may be beneficial for: (1) enhancing athletic performance through increased concentric and eccentric rate of force development[11,12,14–16] and (2) reducing lower-extremity injury risk by helping counteract deleterious loading and helping shield the ligaments from bearing the full responsibility of dynamic joint stability.[8,17–19] Fortunately, stiffness can be modified though training,[25–27] and thus may be an important factor to consider from both injury prevention and rehabilitation perspectives. Therefore, obtaining consistent measures of K_{HAM} and K_{LEG} is essential for the ability to distinguish between random variation in the measurement (random error) and changes attributable to external factors such as training interventions or injury over time.45 Although the interday measurement consistency for the assessment of K_{LEG} during bilateral and unilateral hopping has previously been established, less is known about the measurement consistency of K_{HAM}.

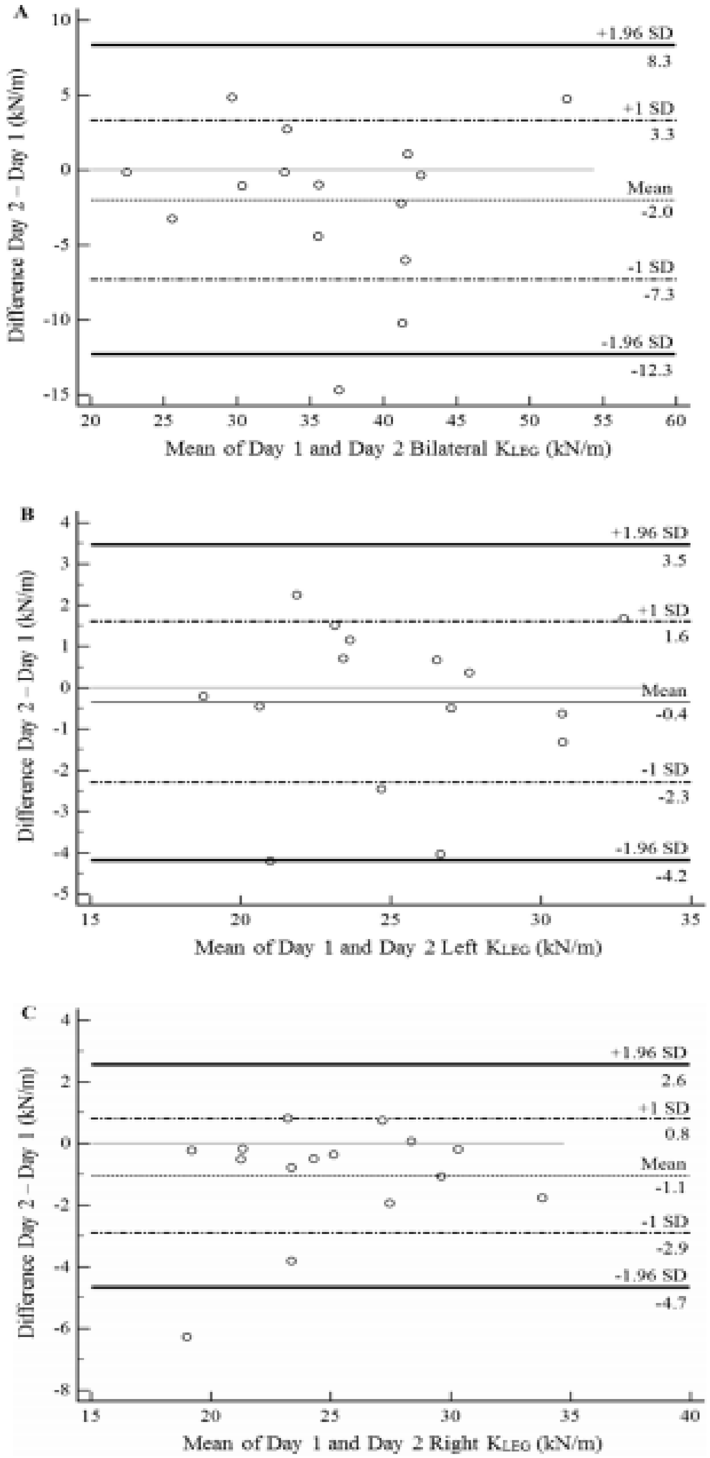

Figure 5 — Interday reliability of bilateral and unilateral leg stiffness (K_{LEG}) measurements: (A) bilateral, (B) left, and (C) right. These Bland-Altman plots show the differences between values for the first and second day of testing on the vertical axis, and the mean of the first and second day of testing on the horizontal axis for each participant during each of the 3 hopping conditions. The thin dashed line represents the mean difference (bias) between testing days (day 2 – day 1), while the thick dashed lines and solid lines represent the 68% and 95% limits of agreement, respectively.

Our primary findings were that K_{HAM} and K_{LEG} both demonstrated acceptable interday measurement consistency, but that K_{HAM} was less consistent than K_{LEG}. To our knowledge, only 2 studies[16,25] have previously examined the interday measurement consistency of K_{HAM}, and these studies have reported ICCs ranging from .66 to .89. In addition, previous investigations on the interday measurement consistency of K_{LEG} have reported ICCs of .93 and .94[29,30] during bilateral barefoot hopping, .8228 during bilateral shod hopping, and .84 to .8928 during unilateral shod hopping, at similar frequencies (2.0–2.5 Hz). Thus, our current findings are in agreement with prior work and support our hypothesis that, although K_{HAM} and K_{LEG} would both display acceptable interday measurement consistency, KLEG would display superior measurement consistency compared with K_{HAM}.

We also constructed Bland-Altman plots and calculated the 68% and 95% LOA for KHAM (Table 1, Figure 4) and K_{LEG} (Table 2, Figure 5) to gain a better picture of the magnitude of absolute measurement error (ie, systematic and random error) that one could expect when performing repeated measurements over time, in the absence of any training intervention or injury. When considering the magnitude of the day-to-day mean difference (systematic error = 89.6 $N \cdot m^{-1}$) for K_{HAM} relative to the deviation in scores obtained on day 1 (134.1 $N \cdot m^{-1}$), this yields an estimated effect size of 0.67. This moderately large estimated effect size for K_{HAM} may mask a potential treatment effect, or suggest that a treatment effect has occurred when in fact there is not one. Thus, future studies aiming to assess longitudinal changes in K_{HAM} should include a control group. In contrast, expressing the magnitude of the day-to-day mean differences for bilateral and unilateral K_{LEG} relative to the deviation in scores obtained on day 1 (Table 2) resulted in much smaller estimated effect sizes (bilateral = 0.25, left = 0.09, right = 0.25), which suggests that the simpler assessment of K_{LEG} may be a more sensitive measure for detecting changes over time.

The lower measurement consistency and precision observed for K_{HAM} is likely due to its more complex procedural characteristics. Our results revealed that the linear combination of test-retest differences for 3 procedural factors explained 59.4% of the variance in the difference in scores from day 1 to day 2 for K_{HAM}. Given that such a large amount of the day-to-day variability in K_{HAM} could be attributed to these procedural factors, it is plausible that magnitude of absolute measurement error could be dramatically reduced in future research efforts if: (1) attempts are made to better standardize the application of the manual perturbation and (2) stricter control is placed over initial knee joint positioning. However, it is important to note that 40.6% of the variance remained unexplained by these factors. Carryover effects, such as effects due to familiarization, muscle fatigue, or muscle soreness, may have also contributed to the magnitude of absolute measurement error observed in this study. Thus, measurement consistency of K_{HAM} may also be improved by allowing additional time for participants to become adequately familiarized to the task in future work. Furthermore, it has previously been demonstrated that the method by which the assessment load is determined also contributes to K_{HAM} variability.[46] Different methods of assigning the applied limb load include: (1) standardizing the load as 10% of the participant's body mass (as in the current study),[31,32] (2) standardizing the load as a percentage of the participant's maximal isometric voluntary contraction (MVIC),[16,25] or (3) using a single fixed load for all participants (eg, 6.5 kg).[16] Ditroilo et al[46] reported that assigning load as a percentage of the participant's MVIC introduces greater variability to the measurement of K_{HAM} when compared with the use of a single fixed load. Therefore, future studies should also carefully consider the method by which limb load is assigned when attempting to assess longitudinal changes in K_{HAM}.

Because of the complexity and equipment needed to measure K_{HAM}, we were also interested in examining the extent to which the more clinically-accessible and global measure of K_{LEG} was associated with K_{HAM}. K_{HAM} and unilateral K_{LEG} measures on day 1 were found to be moderately correlated with one another ($R = .67$, $P < .01$), with K_{LEG} predicting 44% of the variance in K_{HAM}. While these findings suggest that K_{LEG} and K_{HAM} are related measures, we acknowledge that this finding is based on a relatively small sample size (n = 15) which includes both male (n = 6) and female (n = 9) participants in the same analysis. It has previously been shown that K_{HAM} and K_{LEG}, as well as underlying factors that contribute to these measures, can differ in males and females.[25,47,48] Hence, in certain situations, between-sex differences can create artificial relationships when male and female participants are grouped into a single analysis. However, there were no between-sex differences observed for K_{HAM} ($P = .34$) or K_{LEG} ($P = .24$), and within-sex correlations were similar (males: $R = .66$, $P = .15$; females: $R = .63$, $P = .07$), confirming that this relationship was not simply driven by between-sex differences.

As mentioned previously, the procedures associated with the assessment of K_{HAM} can be somewhat cumbersome, and the need for adequate familiarization can be time-consuming. In contrast, the assessment of K_{LEG} requires little familiarization, and collecting data on 3 trials in each condition (ie, bilateral, left, right) can easily be accomplished in under 10 minutes. Although these 2 measures are assessed using fundamentally different procedures, both assessment methods are based on the underlying construct that the muscle-tendon unit(s) can be independently or collectively modeled as a linear mass-spring system.[7] Given the impact that procedural factors had on the day-to-day measurement consistency of K_{HAM}, it is possible that improving the measurement precision of K_{HAM} may reveal a stronger relationship than what we observed in the current study. In addition, participants were instructed to minimize quadriceps muscle activation during the K_{HAM} assessment, whereas cocontraction of the quadriceps and hamstring muscles is inherent to the assessment of K_{LEG}.[49] Hence, differences in cocontraction magnitude may also contribute to this unexplained variance. Because of the potential for K_{LEG} to be used in large screening studies, future research should continue to explore these factors to determine the extent to which K_{LEG} may be appropriately representative of K_{HAM}.

In summary, findings of the current study show that K_{LEG} is a consistent measure during bilateral and unilateral hopping at 2.2 Hz, and that K_{LEG} can adequately be assessed with relatively little familiarization. In contrast, K_{HAM} demonstrated lower interday measurement consistency, which was found to be largely due to day-to-day variations in the magnitude of the manual perturbation as well as initial knee joint positioning. Other factors, such as familiarization, muscle fatigue, or muscle soreness, and simple performance inconsistencies in isometrically holding the load may also have contributed to the lower interday measurement consistency observed for K_{HAM}. As such, researchers interested in assessing longitudinal changes in K_{HAM} are encouraged to allow time for additional familiarization, attempt to standardize the manual perturbation, place strict control over initial knee joint positioning, and include a control group in future work. In addition, researchers may find these results helpful for sample size estimation or when comparing day-to-day measurement consistency using alternative procedures. Furthermore, this study shows a relationship between K_{HAM} and K_{LEG}, which suggests that the simpler measure of K_{LEG} has the potential to serve as a reliable, time-efficient, and clinically accessible measure of lower extremity stiffness that could be of great value when attempting to evaluate large samples with limited time (eg, when conducting preseason athlete screenings). Further work is needed to examine the extent of the K_{HAM} and K_{LEG} relationship after addressing methodological factors that may introduce error (thus unexplained variance) in the measure, such as subtle trial-to-trial variations in knee joint positioning and perturbation magnitude.

ACKNOWLEDGMENTS

The authors thank Mridula Manoj and Hsin-Min Wang for their valuable assistance with this project.

REFERENCES

1. Butler RJ, Crowell HP, Davis IM. Lower extremity stiffness: Implications for performance and injury. *Clin Biomech (Bristol, Avon).* 2003;18(6):511–517.

2. Serpell BG, Ball NB, Scarvell JM, Smith PN. A review of models of vertical, leg, and knee stiffness in adults for running, jumping or hopping tasks. *J Sports Sci.* 2012;30(13):1347–1363.

3. Brughelli M, Cronin J. A review of research on the mechanical stiffness in running and jumping: Methodology and implications. *Scand J Med Sci Sports*. 2008;18(4):417–426.

4. Ditroilo M, Watsford M, Murphy A, De Vito G. Assessing musculoarticular stiffness using free oscillations: Theory, measurement and analysis. *Sports Med.* 2011;41(12):1019–1032.

5. Blickhan R. The spring-mass model for running and hopping. *J Biomech*. 1989;22(11-12):1217–1227.

6. Cavagna GA. Elastic bounce of the body. *J Appl Physiol*. 1970;29(3):279–282.

7. Shorten MR. Muscle elasticity and human performance. *Med Sport Sci*. 1987;25:18.

8. McNair PJ, Wood GA, Marshall RN. Stiffness of the hamstring muscles and its relationship to function in anterior cruciate ligament deficient individuals. *Clin Biomech (Bristol, Avon)*. 1992;7(3):131– 137.

9. Farley CT, Blickhan R, Saito J, Taylor CR. Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits. *J Appl Physiol*. 1991;71(6):2127–2132.

10. Granata KP, Wilson SE, Massimini AK, Gabriel R. Active stiffness of the ankle in response to inertial and elastic loads. *J Electromyogr Kinesiol*. 2004;14(5):599–609.

11. Chelly SM, Denis C. Leg power and hopping stiffness: relationship with sprint running performance. *Med Sci Sports Exerc*. 2001;33(2):326–333.

12. Arampatzis A, Brüggemann GP, Metzler V. The effect of speed on leg stiffness and joint kinetics in human running. *J Biomech*. 1999;32(12):1349–1353.

13. Walshe AD. The validity and reliability of a test of lower body musculotendinous stiffness. *Eur J Appl Physiol Occup Physiol*. 1996;73:332–339.

14. Wilson GJ. Murphy a J, Pryor JF. Musculotendinous stiffness: its relationship to eccentric, isometric, and concentric performance. *J Appl Physiol*. 1994;76(6):2714–2719.

15. Watsford M, Ditroilo M, Fernández-Peña E, D'Amen G, Lucertini F. Muscle stiffness and rate of torque development during sprint cycling. *Med Sci Sports Exerc*. 2010;42(7):1324–1332.

16. Ditroilo M, Watsford M, De Vito G. Validity and inter-day reliability of a free-oscillation test to measure knee extensor and knee flexor musculo-articular stiffness. *J Electromyogr Kinesiol*. 2011;21(3):492– 498.

17. Blackburn JT, Norcross MF, Cannon LN, Zinder SM. Hamstrings stiffness and landing biomechanics linked to anterior cruciate ligament loading. *J Athl Train.* 2013;48(6):764–772.

18. Blackburn JT, Norcross MF, Padua DA. Influences of hamstring stiffness and strength on anterior knee joint stability. *Clin Biomech (Bristol, Avon).* 2011;26(3):278–283.

19. Bryant AL, Kelly J, Hohmann E. Neuromuscular adaptations and correlates of knee functionality following ACL reconstruction. *J Orthop Res.* 2008;26(1):126–135.

20. Williams DS, Davis IM, Scholz JP, Hamill J, Buchanan TS. Higharched runners exhibit increased leg stiffness compared to low-arched runners. *Gait Posture*. 2004;19(3):263–269.

21. Williams DS, McClay IS, Hamill J. Arch structure and injury patterns in runners. *Clin Biomech (Bristol, Avon).* 2001;16(4):341–347.

22. Milner CE, Ferber R, Pollard CD, Hamill J, Davis IS. Biomechanical factors associated with tibial stress fracture in female runners. *Med Sci Sports Exerc.* 2006;38(2):323–328.

23. Granata KP, Padua DA, Wilson SE. Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks. *J Electromyogr Kinesiol*. 2002;12:127–135.

24. Watsford ML, Murphy AJ, McLachlan KA, et al. A prospective study of the relationship between lower body stiffness and hamstring injury in professional Australian rules footballers. *Am J Sports Med.* 2010;38(10):2058–2064.

25. Blackburn JT, Norcross MF. The effects of isometric and isotonic training on hamstring stiffness and anterior cruciate ligament loading mechanisms. *J Electromyogr Kinesiol*. 2014;24(1):98–103.

26. Burgess KE, Connick MJ, Graham-Smith P, Pearson SJ. Plyometric vs. isometric training influences on tendon properties and muscle output. *J Strength Cond Res*. 2007;21(3):986–989.

27. Kubo K, Morimoto M, Komuro T, et al. Effects of plyometric and weight training on muscle-tendon complex and jump performance. *Med Sci Sports Exerc.* 2007;39(10):1801–1810.

28. Joseph C, Bradshaw E, Clark R. Inter-Day Reliability Leg, Knee and Ankle Stiffness Measures During Hopping. *Port J Sport Sci*. 2011;11(2):989–992.

29. McLachlan KA, Murphy AJ, Watsford ML, Rees S. The interday reliability of leg and ankle musculotendinous stiffness measures. *J Appl Biomech*. 2006;22(4):296–304.

30. Lloyd RS, Oliver JL, Hughes MG. Williams C a. Reliability and validity of field-based measures of leg stiffness and reactive strength index in youths. *J Sports Sci*. 2009;27(14):1565–1573.

31. Blackburn JT. Sex comparison of extensibility, passive, and active stiffness of the knee flexors. *Clin Biomech (Bristol, Avon).* 2004;19(1):36– 43.

32. Bell DR, Blackburn JT, Norcorss MF, et al. Estrogen and muscle stiffness have a negative relationship in females. *Knee Surg Sports Traumatol Arthrosc*. 2012;20(2):361–367.

33. Bompa TO, Haff G. *Periodization: Theory and Methodology of Training*. 5th ed. Champaign, IL: Human Kinetics; 2009.

34. Borg G. *Borg's perceived exertion and pain scales*. Champaign, IL: Human Kinetics; 1998.

35. Jennings AG, Seedhom BB. The measurement of muscle stiffness in anterior cruciate injuries - An experiment revisited. *Clin Biomech (Bristol, Avon)*. 1998;13(2):138–140.

36. Granata KP, Wilson SE, Padua DA. Gender differences in active musculoskeletal stiffness. Part I. Quantification in controlled measurements of knee joint dynamics. *J Electromyogr Kinesiol*. 2002;12(2):119–126.

37. Arampatzis A, Schade F, Walsh M, Brüggemann GP. Influence of leg stiffness and its effect on myodynamic jumping performance. *J Electromyogr Kinesiol*. 2001;11(5):355–364.

38. Bishop M, Fiolkowski P, Conrad B, Brunt D, Horodyski M. Athletic footwear, leg stiffness, and running kinematics. *J Athl Train*. 2006;41(4):387–392.

39. Winter DA. Anthropometry. In: *Biomechanics of Human Movement*. New York, NY: John Wiley & Sons; 1990:51–74.

40. Hobara H, Kimura K, Omuro K, et al. Determinants of difference in leg stiffness between endurance- and power-trained athletes. *J Biomech*. 2008;41:506–514.

41. Hobara H, Inoue K, Kobayashi Y, Ogata T. A comparison of computation methods for leg stiffness during hopping. *J Appl Biomech*. 2014;30(1):154–159.

42. Shrout PE, Fleiss JL. Intraclass correlations: uses in assessing rater reliability. *Psychol Bull*. 1979;86(2):420–428.

43. Portney LG, Watkins MP. Reliability. In: *Foundations of clinical research: applications to practice*. 2nd ed. New Jersey: Prentice Hall Health; 2000:53–68. 44. Bland JM, Altman DG.

44. Bland JM, Altman DG. Statistical methods for assessing agreement between two methods of clinical measurement. *Lancet*. 1986;1:307– 310.

45. Atkinson G, Nevill AM. Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Med.* 1998;26(4):217–238.

46. Ditroilo M, Watsford M, Murphy A, De Vito G. Sources of variability in musculo-articular stiffness measurement. *PLoS One*. 2013;8(5):e63719.

47. Hobara H, Kato E, Kobayashi Y, Ogata T. Sex differences in relationship between passive ankle stiffness and leg stiffness during hopping. *J Biomech*. 2012;45(16):2750–2754.

48. Blackburn JT, Bell DR, Norcross MF, Hudson JD, Kimsey MH. Sex comparison of hamstring structural and material properties. *Clin Biomech (Bristol, Avon).* 2009;24(1):65–70.

49. Hobara H, Inoue K, Muraoka T, Omuro K, Sakamoto M, Kanosue K. Leg stiffness adjustment for a range of hopping frequencies in humans. *J Biomech.* 2010;43(3):506–511.