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Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks

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Abstract

Leg stiffness was compared between age-matched males and females during hopping at preferred and controlled frequencies. Stiffness was defined as the linear regression slope between the vertical center of mass (COM) displacement and ground-reaction forces recorded from a force plate during the stance phase of the hopping task. Results demonstrate that subjects modulated the vertical displacement of the COM during ground contact in relation to the square of hopping frequency. This supports the accuracy of the spring–mass oscillator as a representative model of hopping. It also maintained peak vertical ground-reaction load at approximately three times body weight. Leg stiffness values in males (33.9 ± 8.7kN/m) were significantly ($p < 0.01$) greater than in females (26.3 ± 6.5kN/m) at each of three hopping frequencies, 3.0, 2.5 Hz, and a preferred hopping rate. In the spring–mass oscillator model leg stiffness and body mass are related to the frequency of motion. Thus male subjects necessarily recruited greater leg stiffness to drive their heavier body mass at the same frequency as the lighter female subjects during the controlled frequency trials. However, in the preferred hopping condition the stiffness was not constrained by the task because frequency was self-selected. Nonetheless, both male and female subjects hopped at statistically similar preferred frequencies (2.34 ± 0.22Hz), therefore, the females continued to demonstrate less leg stiffness. Recognizing the active muscle stiffness contributes to biomechanical stability as well as leg stiffness, these results may provide insight into the gender bias in risk of musculoskeletal knee injury.

Keywords

Gender; Stiffness; Muscle; Knee

1. Introduction

Epidemiological research reveals that females have a greater risk of lower extremity musculoskeletal injuries during functional activities than males [1-3]. Specifically, females sustain a greater incidence of knee related injuries [3-6] including knee sprains [7,8], anterior cruciate ligament (ACL) injuries [3,4,9,10], meniscal and cartilaginous tears [4], and patello–femoral disorders [11-13,14]. This trend has been demonstrated within several comparable sports. Female high school basketball players suffer 3.8 times the risk of ACL injury then males. Female collegiate soccer and basketball players demonstrate twice the risk of ACL and cartilage knee injury rates compared to their male counterparts [3,4,13]. Although the gender bias has yet to be explained, musculoskeletal joint stability has been implicated as contributing factor to knee injury [15,16]. To improve gender equity in leisure, athletic and work-related activities it is necessary to understand potential gender differences relating to musculoskeletal stability.

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Active muscle stiffness is essential for the maintenance of joint stability [17,18]. At low applied loads the passive structures of the knee provide sufficient stability and resistance to anterior tibial translation [19,20]. During weight-bearing and sporting activities joint forces go well beyond the stabilizing capacity of the joint capsule and ligaments and require assistance from active muscles to stabilize the joint. Hence, muscles serve as the primary active stabilizers of the knee during functional loading conditions, protecting against musculoskeletal injury [21-23]. Recent evidence demonstrates less stiffness in active muscles of females when compared to age-matched male subjects. This has been demonstrated by Wojtys [24] who concluded that active muscle co-contraction reduced anterior tibial translation more in men than in women. Similar results were demonstrated in tibial rotation [25]. During controlled measurements of knee kinematics following mechanical perturbation during active flexion and extension exertions, it has recently been demonstrated that women demonstrate less than 57% of the active muscle stiffness compared to males [26]. This may contribute to gender differences in musculoskeletal stability of the knee. It is unknown whether the lower stiffness measured in these controlled experiments translates to equivalent reductions in functional performance parameters such as leg stiffness during running and hopping.

Non-contact mechanisms of injury associated with jumping and landing are implicated in many lower extremity related injuries [7,12,27]. Active muscle stiffness contributes to leg stiffness and can be measured during functional tasks such as running and hopping have been reported [28-30]. Although it can be debated whether these measurements of leg stiffness truly record mechanical stiffness [31,32], research illustrates that the dynamics of hopping and running can be accurately represented using an equivalent mass–spring model [33]. The leg stiffness is attributed to the active muscle stiffness of the controlling joints [34] thereby influencing biomechanical stability. Unfortunately, there are no reported measures specifically examining gender differences in leg stiffness during functional tasks such as hopping.

We have previously observed that women demonstrate less active muscle stiffness during controlled open-chain measurements of the isolated in vivo knee compared to males. However, during functional performance tasks neuromotor control can voluntarily and reflexively modulate muscle stiffness [35,36]. This can be achieved through muscle recruitment strategies [37] or postural adaptation [34]. The goal of the present study was to evaluate whether female subjects also demonstrate less leg stiffness in functional tasks, specifically during two-legged hopping. It was hypothesized that male subjects must recruit greater leg stiffness during hopping than female subjects in order to drive their larger body mass during controlled frequency tasks. It was further hypothesized that lower active muscle stiffness in female subjects observed from previous analyses would translate to lower leg stiffness in hopping tasks at preferred frequency hopping tasks compared to the male subject group.

2. Materials and methods

2.1. Procedures

Measurements of leg stiffness were determined by requiring subjects to perform two-legged hopping on a force platform. Approximately 30-hops were performed at each hopping frequency. Vertical ground-reaction force was recorded from a force platform (Kistler/Bertec 6700, natural frequency 400 Hz, linearity ±0.2% full scale), sampled at 1000 Hz via an analogue to digital converter (RunTech, Laguna Hills, CA) and stored on a personal computer for post-test analysis. Subjects were asked to hop in place without shoes and with their hands on their hips at three separate hopping frequencies. Hopping was performed first at their preferred rate, then at 2.5 and finally at 3.0 Hz. Controlled frequency hopping was easily achieved by performing the tasks in time with a digital metronome. Subjects were instructed that each hop must be a continuous motion and were allowed as much practice as needed until they felt comfortable at the designated frequency. Hopping frequencies of 2.5 and 3.0 Hz were selected

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in addition to the preferred hopping frequency as they have been previously determined to be higher than the average preferred hopping frequency for humans and thus reveal greater leg stiffness values [29].

2.2. Subjects

Fifteen male and fifteen female volunteers with no reported knee abnormalities or recent musculoskeletal injuries participated in this study. Subjects ranged in age from 21 to 62 years with no significant difference in age between genders (Table 1). A separate validation study was performed using a similar protocol on an independent set of subjects who were generally younger and more physically active. Subjects in this group ranged in age from 21 to 31 years with no significant difference in age between genders (Table 1). In this second effort eleven male and ten female subjects performed the two-legged hopping at their preferred rate and at 3.0 Hz. Informed consent approved by the Human Investigations Committee, University of Virginia, was obtained from all subjects.

Ten acceptable hopping trials were used for analysis at each frequency condition. Hopping trials were determined to be acceptable based on two criteria. First, only those trials where the subject's hopping frequency was within 5% of the designated metronome frequency were accepted for data analysis. During preferred hopping steady-state behavior was assured by requiring the period of individual hops be within 5% of the mean value. Second, the correlation between vertical displacement and vertical ground-reaction force during the ground-contact phase of hopping must have been greater than $r = 0.80$ to be accepted for data analysis (Fig. 1). Hopping trials unable to meet these specified criteria were not used for data analysis.

Two independent methods were implemented in customized software to compute the leg stiffness during each hop. First, leg stiffness was determined by comparing the vertical ground-reaction force with the vertical displacement of the center of mass (COM) during the ground-contact phase as described by McMahon and Cheng [30]. Briefly, vertical acceleration of the COM was determined from the ground-reaction force and the subject's body mass as described by Cavagna [38]. Vertical displacement of the COM was calculated from numeric double integration of the acceleration data. Integration constants for velocity were based upon steady-state performance criteria wherein the mean vertical COM velocity is zero. Since the goal was to determine COM displacement, the integration constant for position was set arbitrarily to zero. Stiffness was determined from the regression slope of the profile when vertical ground-reaction force was plotted versus COM displacement [28] (Fig. 1). A second method of calculating stiffness was achieved by determining the natural frequency of the equivalent mass–spring system as described by Farley [29]. The ground-contact time during which the vertical force was greater than body weight represents a half-period of the harmonic oscillation ($T / 2$). Leg stiffness, k, can be calculated from the resonant period, T, and total body mass, M, recorded from static force plate measurements

$$k = M(2 \ / \ T)^2 \tag{1}$$

2.3. Statistical analysis

Leg stiffness values recorded during functional hopping tasks from each subject were averaged across the acceptable trials for each hopping frequency condition and analyzed using a mixed-model, 2-factor, repeated measures ANOVA with gender as the between-subject factor and hopping frequency as the within-subject factor. Separate one-way ANOVA were performed to evaluate differences in body mass, standing height, age, and preferred hopping frequency variables between genders. Repeated-measured ANOVA was used to assess duty cycle (ground-contact time as a percent of total hopping cycle) at each hopping frequency. The two studies were evaluated in separate statistical analyses to permit comparison of results and validation of conclusions. Statistical significance was set at $\alpha < 0.05$ for all analyses.

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In an attempt to understand factors contributing to the gender effects, multiple regression was performed incorporating the independent variables of gender, body mass, standing height, subject age, duty cycle and preferred hopping frequency.

3. Results

The method of leg stiffness calculation including regression slope and harmonic period methods generated near identical results. Therefore, only results from the regression slope method for calculating leg stiffness are presented.

A total of 19-trials were removed from the data set prior to analysis (10 at preferred, 2 at 2.5 and 7 at 3.0 Hz hopping conditions). Trials were removed from the data set when the subject's measured hopping frequency was not within 5% of specified hopping frequency or the correlation between vertical displacement and vertical ground-reaction force during the ground-contact phases of hopping was $r < 0.80$. Average correlations between vertical displacement and vertical ground reaction (Fig. 1) for the accepted trials were high for each of the hopping conditions ($r_{\mathrm{pref}} = 0.92 \pm 0.01$; $r_{2.5\mathrm{Hz}} = 0.94 \pm 0.01$; $r_{3.0\mathrm{Hz}} = 0.96 \pm 0.03$) (Fig. 1).

Results from the statistical analyses revealed significant main effects for gender ($p = 0.004$) and hopping frequency ($p = 0.0001$). Mean leg stiffness values were 30.1 ± 11.0kN/m and agree with values from the published literature [29]. Stiffness values were significantly greater in males (33.9 ± 8.7kN/m) than in females (26.3 ± 6.5kN/m), with significant differences at each of the hopping frequencies. Both male and female subjects hopped at statistically similar preferred frequencies (2.34 ± 0.22Hz) (Table 2).

Vertical displacement of the COM during ground contact was modulated in proportion to the square of hopping frequency. Peak-to-peak motion of the COM during ground contact significantly ($p < 0.0001$) declined with increasing hopping frequency, 8.4 ± 2.0, 7.1 ± 0.8, and 5.1 ± 0.4cm for the preferred, 2.5 and 3.0 Hz tasks, respectively. Analyses demonstrated this amplitude was proportional to hopping frequency raised to the power −1.92 ± 0.43, explaining 89% of the peak COM variability. This suggests subjects may have attempted to maintain peak acceleration and ground-contact force independent of hopping frequency by regulating the amplitude of the motion. Peak vertical ground-reaction force was significantly greater in males than in females but the difference was attributable to the increased body mass in the male population. Peak force normalized to body weight was not significantly influenced by gender and varied approximately 4% between frequency conditions.

The relation between gender and stiffness was determined to be a covariate of body mass. Multiple regression revealed that variance in leg stiffness was explained predominately by body mass, hopping frequency, and duty cycle; with the models accounting for 85–87% of the total variance (Table 3). Gender did not enter the regression as a significant variable, rather the differences in body mass explained the gender effect on hopping leg stiffness. Although subjects were constrained to hopping at specified frequencies in two of the three conditions, time spent in ground contact versus time in flight phase was not controlled. Nonetheless, both males and females hopped with statistically similar duty cycles (64 ± 8%). Hopping frequency and duty cycle combine to form the natural harmonic frequency of motion, i.e. these two factors describe ground-contact time. Recognizing that body mass was significantly less in women than in the men, Eq. (1) dictates that women must demonstrate less leg stiffness to hop at the appropriate controlled frequency. However, in the preferred hopping frequency condition the female subjects could easily have recruited stiffness identical to the value demonstrated by male subjects. Instead, the data demonstrates that women consistently chose to hop at a statistically similar preferred frequency as the male subjects by recruiting a lower stiffness ($k_{\mathrm{Pref}} = 26.3 \pm 9.1$Hz). Thus, when motion constraints were removed, the women continued to demonstrate lower leg stiffness.

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To validate these results, a second study using a younger, more physically active population was performed with 21 subjects. The results of the second study demonstrated comparable findings to the first. A total of 10-trials were removed from the data set prior to analysis (6 at preferred hoping frequency and 4 at 3.0 Hz hopping conditions) when failing to meet the performance criteria. Average correlations between vertical displacement and vertical ground reaction for the accepted trials were high ($r_{\mathrm{pref}} = 0.94 \pm 0.01$; $r_{3.0\mathrm{Hz}} = 0.95 \pm 0.01$). There was a significant main effect for gender ($p < 0.011$) as stiffness values were again significantly greater in males (34.7 ± 9.8kN/m) than females (28.4 ± 9.3kN/m). In addition there was a significant main effect for hopping frequency ($p < 0.001$) as stiffness values were greater during 3.0 Hz conditions than preferred hopping frequencies. Multiple regression analyses also revealed that primarily body mass, hopping frequency, and duty cycle explained leg stiffness, none of the variance was explained by gender.

4. Discussion

Active muscle stiffness contributes to the biomechanical stability of the knee [17]. By improving stability active muscle stiffness [18,24] may contribute to the prevention of musculoskeletal injury. Passive knee instability measured in terms of laxity, i.e. the relation between anterior tibial distraction and distraction force, is correlated with the risk of musculoskeletal injury [15,16]. This knee laxity is greater in females compared to equivalently trained males [39]. Recognizing that laxity is the inverse of passive stiffness, these results indicate that females have less distraction stiffness in the knee. Active muscle contribution to stability may also be influenced by gender [24,25] with recent evidence indicating that women demonstrate less than 57% of the active muscle stiffness compared to males during non-weight-bearing measurements of the in vivo knee [26].

The objective of the current study was to examine the role of gender on leg stiffness during functional weight-bearing tasks, specifically two-legged hopping. The hopping task was selected based on epidemiologic injury data suggesting that one of the primary non-contact injury mechanisms for the lower extremity is landing from a jump [7,12,27]. Epidemiologic data have established that females are at a greater risk for a variety of lower extremity related injuries than their male counterparts, most notably ACL injuries [3,4,9,10]. Musculoskeletal stability may contribute to this difference. Active muscle stiffness is the primary control variable in musculoskeletal stability [40,41]. Thus, it was hypothesized that female subjects would demonstrate lower leg stiffness during the hopping tasks

Leg stiffness in the female subjects during the hopping task was approximately 77% of the leg stiffness in the male subjects. These findings were confirmed in the separate investigation where leg stiffness in females was 81% of the leg stiffness observed in males. During the controlled hopping frequency conditions the gender difference was explained primarily by body weight. Female subjects weighed an average of 83% of the male subject population in the two populations. In a mass–spring model of harmonic motion the stiffness must change in proportion to the system mass in order to maintain a constant frequency. Similarly, when frequency of hopping increases stiffness must also increase to oscillate the mass. Empirical results confirm this requirement, i.e. vertical leg stiffness increased with frequency (Fig. 1), and agree with the literature [28,29]. In fact, the stiffness increased in proportion to the square of frequency supporting the validity of the spring–mass oscillator as a model of leg behavior in hopping. Other factors that may have contributed to the gender difference in leg stiffness was muscle recruitment strategies and leg posture during hopping. Functional leg stiffness is influenced by knee angle [30,34,42]. Kinematic data describing landing strategies suggest possible gender differences in knee and ankle flexion angles at contact [43-46]. These strategies may be employed to tune the mass–spring system to achieve the prescribed hopping frequency and will be examined in future research. However, body mass explained the majority of the

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gender effect during the controlled hopping conditions. Body mass does not adequately explain gender factors during the preferred hopping frequency condition.

Preferred hopping frequencies described in the scientific literature were similar to our value of 2.34 ± 0.22Hz (65% duty cycle). Jones and Watt [47] reported preferred hopping frequency values 2.06 ± 0.07Hz but these were representative of one-legged hopping. Cavagna [33] demonstrated that leg stiffness during two-legged hopping was greater than for one-legged hopping as performed by our subjects. Farley et al. [29] reported preferred hopping frequency values of 2.17 ± 0.07Hz (65% duty cycle) in two-legged hopping, within one standard deviation of our results. However, their subjects tended to be younger and lighter (21 years, 63.5 kg) than our subject population (32 years, 77.2 kg) possibly contributing to the small difference in the preferred frequencies. In our second or validation study the age and anthropometry of the subject population was comparable to Farley et al. [29] and we observed preferred hopping frequencies identical to the values recorded with the first set of subjects (2.32 ± 0.35Hz, 66% duty cycle).

Women and men demonstrated similar hopping frequencies when permitted to perform at self-selected rates. When frequency of hopping was not constrained, neither was the leg stiffness. The female subjects could have chosen to hop with similar leg stiffness as the male group in these preferred hopping conditions by performing the task at higher self-selected frequencies. Instead, the women choose to hop at a similar preferred frequency as the comparatively heavier men by recruiting less leg stiffness. Why did the women hop with identical preferred frequency and duty cycle as the men? One possible explanation is inherent active muscle stiffness differences between genders. Other explanations include reflex factors and energy conservation.

Preferred frequencies of hopping may be explainable in terms of reflex and active force latency. The human soleus muscle behaves similar to a second order low-pass filter wherein the maximum energy efficiency occurs at a natural motion frequency of approximately 2 Hz [48]. Jones and Watt [47] similarly proposed that the preferred frequency is tuned to the polysynaptic reflex latency. They observed low amplitude monosynaptic reflex 40 ms after initial Achilles tendon stretch, a larger amplitude polysynaptic reflex at 120 ms and an electromechanical delay of 30 ms in the non-weight-bearing ankle. Thus, total time from initial Achilles stretch to the polysynaptic reflex force was 150 ms, occurring near-synchronously with the lowest position in hopping stance phase. This timing may optimize hopping efficiency [49]. Ground contact time in the current study was 284 ± 48ms during preferred hopping. The acceleration peak during ground contact from our measurements was roughly tuned, i.e. within 10 ms, to polysynaptic reflex force latency reported in the literature. This reflex latency is independent of gender or body mass, suggesting the preferred hoping frequency should also be independent of body mass. Further research is necessary to evaluate the gender factors in reflex contribution to hopping function.

An alternative explanation for the preferred frequency is the efficiency derived by tuning leg stiffness to the intrinsic properties of muscle. Active muscle stiffness is proportional to the applied moment or bias force [50-52]. Empirical measurements indicate this relationship is approximately linear [53-55] and has been expressed as

$$k = qF_0 \tag{2}$$

where F_0 is the equilibrium force in the leg and q a constant of linear proportionality relative to the equilibrium length [56]. Wilson et al. [55] conclude the nonlinear behavior is less than 7% over the full range of force suggesting that this linear approximation is a reasonable description of muscle stiffness. The equilibrium equation can be expressed as the sum of the inertial load, $M\omega^2 x$, elastic forces, kx, gravitational load, Mg and applied muscle force F_0,

$$M\omega^2 x + kx + Mg + F_0 = 0 \tag{3}$$

where M is the system mass, ω is the oscillation frequency, x the displacement amplitude, k the elastic stiffness, g the acceleration of gravity. The minimum energy level of the dynamic system is achieved when the equilibrium force is equal and opposite to the static load, $F_0 = Mg$ requiring the system to oscillate at the natural frequency, $\omega^2 = k/M$. Combining Eqs. (2) and (3), it can be shown that the natural frequency, i.e. the state of minimum dynamic energy, is independent of system mass

$$\omega^2 = qg \tag{4}$$

If the preferred frequency of hopping is related to energy cost then preferred hopping frequency may be independent of body mass as demonstrated by our results. Further research is necessary to evaluate the intrinsic stiffness properties of the hopping system in relation to the preferred frequency of hopping. Further study is also necessary to evaluate the interaction between this behavior and the reflex tuning response described above.

A common trend in both men and women was the behavior of the COM during ground contact. Research suggests peak-to-peak COM motion is modulated with hopping frequency or running cadence [28]. By modifying COM displacement it was possible to maintain a constant value of peak acceleration and peak ground-contact force independent of hopping frequency. Control of ground-contact force is necessary to limit peak joint load and may be necessary to prevent articular and osteoligamentous injury. Although ground-contact force was statistically influenced by frequency the difference between the 2.5 and 3.0 Hz conditions was merely 4% of the peak value. Had the subjects failed to modulate the COM displacement this difference would have exceeded 44%. Instead the COM displacement amplitude was reduced with the square of frequency to maintain the peak force at approximately 3 times body weight. Results demonstrated that the square of frequency explained 89% of the variability in amplitude of COM motion. This suggests an attempt to limit ground-contact force or peak joint load by modulating COM displacement through the control of leg stiffness.

5. Conclusion

Healthy women demonstrated lower leg stiffness during functional hopping tasks compared to age-matched men. This difference was necessary to oscillate the lighter body mass of the female participants at the same frequency as the heavier male subjects. However, during preferred hopping conditions the female subjects were not constrained to recruit lower stiffness. Nonetheless, the women continued to demonstrate lower leg stiffness then the men, hopping at similar preferred frequencies. Potential explanations for the mass-independent selection of preferred frequency are proposed.

Epidemiologic injury data has established that females are at a greater risk for musculoskeletal injuries than their male counterparts. Active muscle stiffness contributes to the biomechanical stability and may contribute to the prevention of musculoskeletal injury. The results of recent non-weight-bearing measurements suggest biomechanical stability may be challenged in women as a result of less active muscle stiffness. The current study illustrates the lower active stiffness is also evident in functional tasks. To control gender bias in injury risk further research is necessary to evaluate the affect of the active stiffness and stability as a risk factor in injury.

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Biographies

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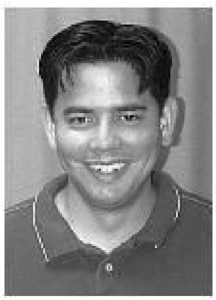

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Fig. 1.
Subjects hopped in place on the force platform at preferred, 2.5, and 3.0 Hz hopping frequencies. Stiffness was calculated from vertical force and COM displacement.

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Table 1

Subject characteristics. Male subjects were significantly taller and heavier than female subjects. Preferred hopping frequency for the male and female subjects were not significantly different. (mean±SD)

	Study #1		**Study #2**	
	Men	**Women**	**Men**	**Women**
# Subjects	15	15	11	10
Age (yrs)	32.1±8.3	32.6±9.7	27.8±4.3	24.1±4.3
Height (m)	1.80±0.08*	1.69±0.06	1.76±0.05	1.68±0.07
Weight (kg)	84.1±12.3*	70.4±15.2	80.1±9.2*	66.9±12.3
Preferred hop frequency (Hz)	2.38±0.24	2.30±0.21	2.33±0.34	2.31±0.35

* =Males significantly greater than females.

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Table 2

Equivalent vertical leg stiffness, hopping duty cycle, ground reaction force (GRF) and center of mass (COM) displacement by gender (mean±SD)

	Hopping frequency	Males	Females
Stiffness (kN/m)	Preferred*	26±9	19±8
	2.5 Hz*	31±8	24±5
	3.0 Hz*	43±8	35±7
Duty cycle (%)	Preferred	65±8	65±8
	2.5 Hz	62±8	63±8
	3.0 Hz	62±8	64±7
Peak vertical GRF (Body weight)	Preferred	3.0±0.5	2.9±0.5
	2.5 Hz	3.2±0.5	3.1±0.4
	3.0 Hz	3.1±0.3	3.0±0.3
COM displacement (cm)	Preferred	8.2±2.0	8.6±2.1
	2.5 Hz	7.4±0.8	6.8±0.9
	3.0 Hz	5.1±0.5	5.0±0.3

* =Males significantly greater than females

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Table 3

Multiple regression revealed that gender effects on equivalent stiffness were attributable to body mass. Hopping frequency, hopping duty cycle, and body mass explained approximately 86% of the variability in measured stiffness

	Intercept	Gender	Mass	Height	Age	Duty cycle	Preferred frequency
Preferred hopping condition							
Regression coefficient	−53.51	0.013	0.227	0.174	0.002	−53.35	26.84
SE	27.71	2.33	0.069	0.038	0.106	13.48	3.80
t-score	−1.93	0.006	3.270	1.260	0.028	−3.96	7.054
p-value	0.069	0.995	0.004	0.223	0.978	0.001	0.001
	Multiple R^2=0.860, F=18.456, p=0.001						
2.5 Hz Hopping condition							
Regression coefficient	−8.33	1.72	0.362	0.244	0.099	−65.36	
SE	21.70	1.980	0.060	0.114	0.087	10.250	
t-score	−0.384	0.871	6.030	2.150	1.146	−6.370	
p-value	0.705	0.395	0.001	0.045	0.266	0.001	
	Multiple R^2=0.870, F=25.468, p=0.001						
3.0 Hz Hopping condition							
Regression coefficient	50.27	1.070	0.513	0.037	−0.151	−85.190	
SE	26.930	2.520	0.076	0.141	0.099	13.210	
t-score	1.867	0.428	6.740	0.260	−1.530	−6.448	
p-value	0.077	0.674	0.001	0.794	0.142	0.001	
	Multiple R^2=0.847, F=21.102, p=0.001						