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Gender Differences in Leg Stiffness and Stiffness Recruitment Strategy During Two-Legged Hopping

Darin A. Padua
Department of Exercise and Sport Science University of North Carolina Chapel Hill

Brent L. Arnold
Department of Exercise Science Virginia Commonwealth University Richmond

Christopher R. Carcia
Department of Physical Therapy Duquesne University Pittsburgh, PA

Kevin P. Granata
Department of Engineering Science and Mechanics Virginia Polytechnic Institute and State University Blacksburg

Abstract

The authors compared leg stiffness (K_{VERT}), muscle activation, and joint movement patterns between 11 men and 10 women during hopping. Physically active and healthy men and women performed continuous 2-legged hopping at their preferred rate and at 3.0 Hz. Compared with men, women demonstrated decreased K_{VERT}; however, after the authors normalized for body mass, gender differences in K_{VERT} were eliminated. In comparison with men, women also demonstrated increased quadriceps and soleus activity, as well as greater quadriceps-to-hamstrings coactivation ratios. There were no significant gender differences for joint movement patterns ($p > .05$). The relationship between the observed gender differences in muscle recruitment and the increased risk of anterior cruciate ligament injury in women requires further study.

Keywords

ACL; quadriceps; coactivation; hop; jump

Females' anterior cruciate ligament (ACL) injury rates are 2.0 to 9.7 times greater than are males' ACL injury rates (Arendt, Agel, & Dick, 1999; Bjordal, Arnly, Hannestad, & Strand, 1997; Cox & Lenz, 1984; Gomez, DeLee, & Farney, 1996; Malone, Hardaker, Garrett, Feagin, & Bassett, 1993; Messina, Farney, & DeLee, 1999). That difference has been demonstrated in several comparable sports and activity levels (Arendt et al.; Gomez et al.; Messina et al.) and in nonathletic populations of similarly trained backgrounds (e.g., military-related training; Cox & Lenz; Gwinn, Wilckens, McDevitt, Ross, & Kao, 2000). Biomechanical factors are believed to partially explain the gender bias in ACL injury rates (L. Y. Griffin et al., 2000), and stiffness of the musculoskeletal system is one such biomechanical factor that may influence the gender bias in ACL injury rates.

Correspondence address: Darin A. Padua, Department of Exercise and Sport Science, 216 Fetzer CB#8700, University of North Carolina, Chapel Hill, NC 27599-8700, USA. E-mail address: dpadua@email.unc.edu.

The term *stiffness* describes the force response that results from and resists mechanical stretch. Stability requires active muscle stiffness (Duan, Allen, & Sun, 1997; Wagner & Blickhan, 1999) and may ultimately influence musculoskeletal injury (McGill, 2001). However, the contribution of gender differences in stiffness properties to the increased ACL injury rates observed in women has been largely overlooked. During controlled open-chain measurements of the isolated in vivo knee, women demonstrate less active muscle stiffness than men do (Blackburn, Riemann, Padua, & Guskiewicz, 2004; Granata, Wilson, & Padua, 2002). Granata, Padua, and Wilson (2002) observed similar findings during closed-chain, functional tasks such as two-legged hopping. Reduced stiffness properties in women may result in decreased stability and may potentially influence their elevated risk of ACL injury.

The stiffness behavior of the lower extremity during functional loading conditions is complex. Lower extremity stiffness during functional tasks represents the average stiffness of the musculoskeletal system and thus depends on the torsional stiffness of the joints (torsional joint stiffness) during ground contact (Arampatzis, Bruggemann, & Klapsing, 2001; Arampatzis, Bruggemann, & Metzler, 1999; Farley, Houdikj, Strien, & Louie, 1998; Farley & Morgenroth, 1999; Greene & McMahon, 1979; McMahon, Valiant, & Frederick, 1987). Torsional joint stiffness is controlled by several biomechanical factors, including muscle activation and force (Hunter & Kearney, 1982; Julian & Sollins, 1975; Lacquanti, Licata, & Soechting, 1982; Weiss, Hunter, & Kearney, 1988; Zhang, Nuber, Butler, Bowen, & Rymer, 1998), reflexes (Houk, 1979; Kearney, Stein, & Parameswaran, 1997; Nichols & Houk, 1976), antagonist muscle coactivation (Agarwal & Gottlieb, 1977; Cannon & Zahalak, 1982; Lacquanti et al.), and lower extremity kinematics during ground contact (Farley & Morgenroth; Greene & McMahon; McMahon et al.; Zhang et al.). As such, one can modulate lower extremity stiffness during functional loading conditions through different muscle activation and movement strategies. In a multijoint system with several strategies available to modulate torsional joint stiffness, the potential stiffness recruitment strategies available to modulate lower extremity stiffness are limitless (Farley et al.; Farley & Morgenroth). *Stiffness recruitment strategy* may be operationally defined as the multijoint coordination (joint kinematics) and muscular recruitment plan (muscle activation) an individual executes to modulate joint torsional stiffness and lower extremity stiffness and, hence, to satisfy the objectives of the functional task (Farley et al.; Hortobagyi & DeVita, 1999, 2000). Women may use altered stiffness recruitment strategies (muscle activation, movement strategies, or both) to compensate for inherent reductions in stiffness properties during functional loading conditions.

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Investigations of gender differences in muscle activation and movement strategies have revealed that women repeatedly demonstrate greater reliance on their quadriceps muscles and move in a more erect posture (increased knee and hip extension) than do their male counterparts (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Hewett, Stroupe, Nance, & Noyes, 1996; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Wojtys, Huston, Taylor, & Bastian, 1996). Quadriceps and gastrocnemius contractions increase anterior tibial shear forces that magnify ACL strain (Beynnon et al., 1995; Durselen, Claes, & Kiefer, 1995; Fleming et al., 2001; Hirokawa, Solomonow, Lu, Lou, & D'Ambrosia, 1992; Li, Sakane, Kanamori, Ma, & Woo, 1999; K. L. Markolf, Gorek, Kabo, & Shapirt, 1990; Renstrom, Arms, Stanwyck, Johnson, & Pope, 1986). Rotary stresses at the knee are also known to facilitate ACL strain (Arms et al., 1984; K. Markolf et al., 1995). As such, imbalanced recruitment between the medial and lateral muscles crossing the knee (e.g., quadriceps, hamstrings, and gastrocnemius) may influence the magnitude of the rotary stresses at the knee joint (Arms et al.; K. Markolf et al.). Because excessive strain is the ultimate cause of ACL injury, greater quadriceps and gastrocnemius activation or imbalanced activation of medial and lateral knee musculature may increase ACL injury risk. Quadriceps- and gastrocnemius-induced ACL strain is amplified when the knee joint is in a more extended position and the hamstrings are unable to counteract the force generated by those muscles

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(Beynnon et al.; Renstrom et al.). Increased quadriceps and gastrocnemius activity with the knee in more extended postures may be an effective stiffness recruitment strategy for increasing lower extremity stiffness during functional tasks. However, such stiffness recruitment strategies may increase ACL loading and strain and, hence, possibly facilitate increased ACL injury risk.

The previously identified gender differences in kinematics (Malinzak et al., 2001; McNitt-Gray, Yokoi, & Millward, 1993) and muscular recruitment (Hewett et al., 1996; Huston & Wojtys, 1996; Malinzak et al.) may represent women's use of altered stiffness recruitment strategies to modulate or compensate for reduced stiffness properties when performing functional loading tasks. Unfortunately, we have not been able to find reports of specific examinations of gender differences in lower extremity stiffness and stiffness recruitment strategies during functional loading conditions.

Gender differences in lower extremity stiffness, stiffness recruitment strategies, or both, during functional loading conditions may play a role in the elevated ACL injury rates observed in women. One can assess lower extremity stiffness and the associated stiffness recruitment strategies during functional loading conditions such as hopping. Thus, we hypothesized that women would demonstrate less lower extremity stiffness and altered stiffness recruitment strategies than would men during a functional, closed-chain task (two-legged hopping). Specifically, we hypothesized that women would demonstrate greater quadriceps and gastrocnemius activity, reduced hamstrings activity, an increased coactivation ratio of the quadriceps and hamstrings, and less knee flexion and ankle plantarflexion in comparison with those of men. We further hypothesized that gender differences in quadriceps, hamstrings, and gastrocnemius muscle activation would be influenced by muscle side because we believed that women may use an imbalanced muscle recruitment strategy between the medial and lateral muscles within a muscle group.

Method

Participants

Physically active men (n = 11, age = 27.81 ± 4.35 years [$M \pm SD$], height = 176.54 ± 7.54 cm, weight = 80.11 ± 9.21 kg) and women (n = 10, age = 24.10 ± 3.75 years, height = 168.50 ± 5.91 cm, weight = 66.92 ± 12.39 kg) volunteered to participate in this study. All participants had previous recreational experience in jumping and landing sports (basketball, volleyball, and soccer) as documented through a questionnaire. No participants enrolled in the study had previous history of significant knee ligament trauma. In addition, no participant reported any type of vestibular disorder. Written informed consent approved by the university's Human Investigations Committee was obtained from all participants.

Testing Procedures

Upon arrival at the laboratory, participants completed a verbal questionnaire so that we could ensure compliance with the inclusion criteria. Before testing, participants received an explanation of all testing procedures and were allowed practice trials to become acquainted with the testing procedures. The dominant lower extremity limb served as the test limb for all muscle activity and kinematic data. We determined limb dominance by having participants perform a single-leg landing from a 30-cm-high box. We defined the dominant limb as the self-selected lower extremity limb on which the participant landed. We performed all testing during a single session in the Motion Analysis and Motor Performance Laboratory.

Two-Legged Hopping

To enable us to assess lower extremity stiffness and stiffness recruitment strategies, participants performed repetitive two-legged jumping in place during two different conditions. To be consistent with previous investigations of near identical tasks, we refer herein to repetitive two-legged jumping as *hopping*. Because participants performed two-legged hopping for all trials, lower extremity stiffness was equivalent to the combined stiffness for both legs. Given the lack of studies of gender differences in lower extremity stiffness and stiffness recruitment strategies, we felt it important to investigate a highly controlled and well-documented functional task such as two-legged hopping.

Participants performed two separate hopping frequency conditions on the same occasion. During all hopping trials, participants maintained their trunk in an upright position, with their hands on their hips, and wore no shoes. Participants were allowed to self-select their preferred knee and ankle movement patterns during the hopping trials. They hopped at their preferred, self-selected rate ($FREQ_{PREF}$) and at a controlled hopping rate of 3.0 Hz ($FREQ_{3.0}$). We chose those hopping frequencies to investigate lower extremity stiffness and stiffness recruitment strategies at two different hopping frequencies. The results of previous research have demonstrated that individuals' preferred hopping frequency is approximately 2.2 ± .07 hops/s ($M \pm SE$; Farley, Blickhan, Saito, & Taylor, 1991). In addition, lower extremity stiffness increases in proportion to hopping frequency (Farley et al.). In the current study, men and women displayed near identical preferred hopping frequencies (men, $M = 2.30 \pm .35$ Hz; women, $M = 2.30 \pm .35$ Hz), which are comparable with previous results. In addition, the magnitude of change in hopping frequency from $FREQ_{PREF}$ to $FREQ_{3.0}$ hopping conditions was also similar for men ($M = 22\% \pm 11\%$) and women ($M = 22\% \pm 11\%$). Assessment of the $FREQ_{PREF}$ hopping condition allowed us to investigate gender differences in lower extremity stiffness and stiffness recruitment strategies during conditions in which participants were able to self-select their lower extremity stiffness behavior and corresponding preferred hopping frequency. We were also able to investigate whether the men and women adjusted their lower extremity stiffness and stiffness recruitment strategies similarly during the faster $FREQ_{3.0}$ hopping conditions. We achieved controlled frequency hopping trials ($FREQ_{3.0}$) by having participants hop in time with a digital metronome. They were instructed that each hop must be a continuous motion and were allowed as much practice as needed until they felt comfortable performing each of the hopping conditions.

We determined during preliminary testing that use of the digital metronome during $FREQ_{3.0}$ hopping influenced the preferred hopping rates ($FREQ_{PREF}$). When participants performed $FREQ_{3.0}$ hopping before $FREQ_{PREF}$ conditions, their $FREQ_{PREF}$ was increased compared with their initial $FREQ_{PREF}$. Therefore, participants performed $FREQ_{PREF}$ hopping conditions followed by $FREQ_{3.0}$ hopping conditions—approximately 45 continuous hops in each of the hopping conditions.

Data Processing and Analysis

We acquired all data by using the Datapac III Version 2000 data-collection hardware and software systems (Run Technologies; Laguna Hills, CA), and we stored the data in a personal computer for later analysis by using customized software developed in MATLAB Version 6.1 (The Math-Works, Natick, MA). We sampled all data at 1000 Hz; thus, all electromyographic (EMG), force plate, and electrogoniometer data were synchronized.

Data Selection

We used the first 10 acceptable hopping trials from each of the hopping conditions for analysis. We determined hopping trials to be acceptable on the basis of two criteria. First, we accepted for analysis only those trials in which participants' hopping frequency was within 5% of the designated metronome frequency ($FREQ_{3.0}$) or average self-selected hopping rate ($FREQ_{PREF}$). We selected the 5% criterion on the basis of previous research results demonstrating vertical leg stiffness to be directly related to hopping frequency (Farley & Morgenroth, 1999; Granata, Padua et al., 2002). Hopping frequencies slower or faster than 5% of the designated or self-selected hopping rate would likely result in significantly different vertical leg stiffness values. We used vertical ground reaction force profiles to determine which trials were within the acceptable hopping frequency range. Second, the linear correlation between vertical center of mass (COM) displacement and vertical ground reaction force during the ground-contact phases of hopping had to be greater than $r = .80$ to be accepted for analysis. It is assumed in the vertical leg stiffness term, K_{VERT}, that the lower extremity behaves like a simple spring-mass system. To evaluate that assumption, we calculated the linear relationship between vertical ground reaction force and vertical COM displacement during the ground-contact phase of the hopping trials. Because we accepted only trials in which the correlation between vertical ground reaction force and vertical COM displacement was $r > .80$, we examined only those trials in which the lower extremity behaved like a simple spring-mass system (Farley & Morgenroth; Granata, Padua, et al.). We did not use for data analysis hopping trials that were unable to meet those specified criteria.

We used the aforementioned data-selection criteria to ensure that gender comparisons were made across similar loading conditions. We also investigated the flight time and the duty cycle of the acceptable hopping trials to further ensure similar loading conditions across genders. Similar flight times would indicate equivalent hopping heights for men and women. We defined *duty cycle* as the ground-contact time divided by the total hop time (sum of ground-contact time and flight time as determined from force plate data).

Vertical Leg Stiffness Calculation

To measure K_{VERT}, we modeled the lower extremity as a simple spring-mass system as participants performed continuous two-legged hopping on a force platform (Kistler/Bertec 6700, natural frequency 400 Hz, linearity ± 0.2% full scale), sampling at 1000 Hz. We calculated K_{VERT} during each hop from the regression slope of the profile when vertical ground reaction was plotted versus the vertical displacement of the individual's COM during the ground-contact phase in kN/m (see Figure 1; McMahon & Cheng, 1990). Briefly, we determined vertical acceleration of the COM from the ground reaction force and the participant's body mass measured during static calibration trials (Cavagna, 1975). We calculated vertical displacement of the COM during ground-contact periods from numerical double integration in the time domain of the acceleration-time data. The acceleration-time curve was generated from the vertical ground reaction force. We based the integration constants for velocity upon steady-state performance criteria in which the mean vertical COM velocity was zero. Because our goal was to determine COM displacement, we set the integration constant for position arbitrarily to zero. We assumed that the vertical velocity of the COM was zero at the time when the COM reached its peak downward displacement.

Previous research results have demonstrated lower extremity stiffness to be strongly related to participant size (Farley, Glasheen, & McMahon, 1993). On the basis of that knowledge and the knowledge that the men were significantly taller, $F(1, 19) = 7.294$, $p = .014$, and heavier, $F(1, 19) = 7.708$, $p = .012$, than were the women in this study, we normalized each participant 's K_{VERT} values by dividing by the respective body weight ($K_{VERT\text{-}NORM}$ [N]). $K_{VERT\text{-}NORM}$ values for each participant were averaged across the acceptable trials for each hopping frequency.

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Muscle Activity and Coactivation

We used an eight-channel, telemetry EMG system (Noraxon, Scottsdale, AZ) sampling at 1000 Hz to record peak muscle activity and coactivation. Unit specifications included a differential amplifier gain of 1,000 fixed, a frequency bandwidth of 16–500 Hz, a common mode rejection ratio = 114 dB, and input resistance from 20 MΩ to 1 GΩ. We placed bipolar silver/silver-chloride surface electrodes (Medicotest, Rolling Meadows, IL) measuring 10 mm in diameter with a center-to-center distance of approximately 2.0 cm in parallel arrangement over the muscle bellies of the rectus femoris (RF), vastus medialis (VM), medial hamstring (MH), lateral hamstring (LH), medial gastrocnemius (MG), lateral gastrocnemius (LG), soleus (SO), and anterior tibialis (AT) according to Cram and Kasman (1998). The participant's skin was shaved and cleaned with isopropyl alcohol before we applied surface electrodes. We confirmed all electrode placements with manual muscle testing and checked for cross-talk. We checked cross-talk through visual inspection of the EMG data collected during manual muscle testing to ensure that no cross-talk occurred between antagonist muscle groups (D. Winter, Fuglevand, & Archer, 1994). We further secured the surface electrodes with an elastic bandage to prevent cable tensioning and movement artifact during hopping. Muscle activity was collected from surface electrodes via a battery-operated FM transmitter/amplifier (Noraxon, Scottsdale, AZ) worn by the participant. From the transmitter, the signal was telemetered to the computer where the raw EMG data were stored for later analysis. Postacquisition, we low-pass filtered at 250 Hz, high-pass filtered at 30 Hz, rectified, and smoothed all EMG data by using a Hanning integrator set to 20 points.

Once we had achieved proper placement of surface electrodes, participants sat on a commercial isokinetic dynamo-meter (Biodex Medical Systems, Shirley, NY) and we asked them to perform maximal voluntary isometric contractions (MVIC) for each of the muscles tested (Yang & Winter, 1984). All MVIC testing was performed in standardized joint positions for the specific muscle group. We established MVIC levels for the eight muscles tested for each participant by collecting three maximal 5-s trials. We removed the first and last second of the MVIC trials from the data to assure only steady-state results during MVIC trials. We averaged the peak activity across the three trials for each muscle and then used the average peak muscle activity during MVIC trials to normalize all EMG data collected during hopping. Thus, EMG data are expressed as a percentage of MVIC (% MVIC).

We assessed muscle activity by averaging the peak muscle activation amplitude during the preparatory response (PR) and loading response (LR) phases from the first 10 acceptable hopping trials for each muscle tested. Similarly, we recorded coactivation ratios during the same phases while participants were hopping. We defined the *PR phase* as the 50 ms preceding the instant of ground contact, as determined from the vertical ground reaction force (Figure 2). PR-phase muscle activation is believed to represent the individual's preprogrammed muscle recruitment strategy for modulating lower extremity stiffness and joint stability during ground contact. In previous investigations of preparatory muscle activation during jumping tasks from a fixed height, time windows of 100–200 ms before ground contact were found (Horita, Komi, Nicol, & Kyrolainen, 1999;Hortobagyi & DeVita, 2000;Mortiani, Oddson, & Thorstensson, 1990). In the current study, we were unable to use such a long time window to determine preparatory activation, given the participants' flight time between successive hops ($FREQ_{PREF}$, $M = 146.42 \pm 17.57$ ms; $FREQ_{3.0}$, $M = 106.76 \pm 9.10$ ms). The use of longer time windows to calculate preparatory muscle activation would have resulted in our collecting data during the ground-contact phase or the upward movement of the individual's COM during the flight phase of the previous hop. We defined the *LR phase* as the 50-ms interval immediately following ground contact (Figure 2). We selected that time interval in an attempt to assess the muscle activation response immediately following perturbation from ground contact and to be consistent with previous research investigations of muscle response activation following landing (Avela & Komi, 1998;Avela, Kyrolainen, Komi, & Rama, 1999;Maton & Pellec, 2001;Nicol, Komi, Horita, Kyrolainen, & Takala, 1996). In addition, Boden and colleagues (Boden, Dean, Feagin, & Garrett, 2000;Boden, Griffin, & Garrett, 2000) indicated that leg collapse following ACL injury occurs near foot strike; however, they did not describe a specific temporal window. Leg collapse following ACL injury likely results from a painful response caused by the injury. Although the exact moment of injury following foot strike has not been determined, the temporal window for ACL injury following foot strike appears to be quite small. We believe that assessment of muscle activation during the 50-ms time window following foot strike (LR phase) may offer insight into the muscle recruitment strategy used to stiffen and stabilize the lower extremity during periods when ACL injury is estimated to occur.

We determined coactivation ratios for the quadriceps and hamstrings (Q:H) as well as the triceps surae (MG, LG, and SO muscles) and anterior tibialis (TS:AT) muscle groups. For our purposes in this article, we considered a muscle according to the role it plays during the execution of motion (Hortobagyi & DeVita, 2000). Thus, during hopping, the Q and TS are agonists undergoing eccentric contraction and absorbing energy during the loading phase, and the antagonists H and AT stabilize the knee and ankle joints (Hortobagyi & DeVita). We computed Q:H coactivation as the sum of quadriceps (RF and VM) activity divided by the sum of hamstrings (MH and LH) activity. We computed the TS:AT coactivation ratio as the sum of TS activity divided by AT activity.

Knee and Ankle Kinematics

We assessed knee (flexion and extension) and ankle (plan-tarflexion and dorsiflexion) motion by using electrogoniometers (Penny and Giles Biometrics Ltd, Cwmfelinfach, Gwent, UK). Electrogoniometer specifications included an accuracy of ±2° over a 90° range of motion. The electrogoniometers weighed 19 g and 17 g for the knee and ankle, respectively. For assessment of knee motion, we placed the electrogoniometer over the lateral aspect of the dominant leg, using the joint line as the axis of rotation and lines drawn from the greater trochanter to the lateral femoral condyle, and from the head of the fibula to the lateral malleolus. We assessed ankle motion by placing the electrogoniometer over the dorsum of the foot in line with the third metatarsal and along the anterior shaft of the tibia. We attached electrogoniometers to the participant's skin with double-sided medical tape and positioned them over the lateral aspect of the knee to measure knee flexion and extension and over the dorsum of the foot and anterior aspect of the tibia for ankle plantarflexion and dorsiflexion measures. We further secured the electrogoniometers by using an elastic bandage to prevent movement artifact during hopping.

We computed knee (ANG_{KNEE}) and ankle (ANG_{ANKLE}) joint positions at the moment of initial ground contact. Joint position at initial ground contact has been revealed to influence leg stiffness as well as ground reaction forces (DeVita & Skelly, 1992; McMahon et al., 1987). In addition, we assessed knee (ROM_{KNEE}) and ankle (ROM_{ANKLE}) joint excursion. We defined ROM_{KNEE} and ROM_{ANKLE} as the range of joint motion occurring from the time of initial ground contact until reaching the position of maximal joint flexion during the ground-contact portion of hopping. That definition of joint excursion has been used in previous studies (Hortobagyi & DeVita, 1999; Lephart et al., 2002). Knee and ankle joint excursions have been found to influence leg stiffness and ground reaction forces during hopping and jumping maneuvers (Arampatzis et al., 1999; Farley & Morgenroth, 1999; Hortobagyi & DeVita).

Statistical Analyses

The basic research design was multivariate repeated measures analyses of variance (ANOVAs). In all analyses, gender was the only between and independent variable (two levels: men and women), whereas the number of within and repeated variables differed depending on the parameters tested. Analyses of K_{VERT}, ANG_{KNEE}, ANG_{ANKLE}, ROM_{KNEE}, and ROM_{ANKLE} involved a 2 × 2 (Gender × Hopping Frequency) multivariate repeated measures ANOVA with only one within and repeated variable: hopping frequency ($FREQ_{PREF}$ and $FREQ_{3.0}$). Muscle activation amplitude of the SO and AT, as well as Q:H and TS:AT coactivation ratios, involved a 2 × 2 × 2 (Gender × Hopping Frequency × Phase) multivariate repeated measures ANOVA with two within and repeated variables, including hopping frequency and phase (PR and LR). Muscle activation amplitude of the VM and RF, MH and LH, and MG and LG involved a 2 × 2 × 2 × 2 (Gender × Hopping Frequency × Phase × Side) multivariate repeated measures ANOVA with three within and repeated variables, including hopping frequency, phase, and muscle side (M and L). We used Box's *M* test to check for homogeneity of the covariance matrices of the dependent variables. When Box's *M* test was significant, we adjusted the significance level by using the Hyunh-Feldt technique. As previously indicated, we also performed statistical analyses, using independent *t* tests, to determine if there were gender differences in height and weight. Statistical significance was set a priori at $\alpha < .05$ for all analyses. To investigate significant main effects and interactions, we performed Tukey's post hoc analyses. SPSS for Windows software (Version 10.0, SPSS Inc., Chicago, IL) was used for all statistical analyses.

Results

We removed 13 of the 420 trials from the data set before analysis (8 at $FREQ_{PREF}$ and 5 at $FREQ_{3.0}$ hopping conditions). Trials were removed from the data set when the participant's measured hopping frequency was not within 5% of specified hopping frequency or when the correlation (*r*) between vertical COM displacement and vertical ground reaction force during the ground-contact phases of hopping was less than .80. Average correlations between vertical COM displacement and vertical ground reaction for the accepted trials were high for each of the hopping conditions ($FREQ_{PREF}$, $M = .99 \pm .02$; $FREQ_{3.0}$, $M = .99 \pm .01$). That highly linear relationship demonstrated that the lower extremity behaved like a simple spring-mass system during the acceptable hopping trials.

Means and standard deviations for the dependent measures are presented by gender during the $FREQ_{PREF}$ and $FREQ_{3.0}$ hopping conditions in Tables 1 through 4. Specifically, the following dependent measures are presented: (a) K_{VERT} and $K_{VERT\text{-}NORM}$, (b) knee and ankle joint landing angles and excursions, (c) EMG activity during PR and LR phases for thigh and ankle musculature, and (d) coactivation ratios during PR and LR phases.

Vertical Leg Stiffness (K_{VERT})

Statistical analysis revealed significant main effects for gender, $F(1, 19) = 7.875$, $p = .011$, and hopping frequency, $F(1, 19) = 48.968$, $p = .001$, but no significant Gender × Hopping Frequency interaction, $F(1, 19) = .003$, $p = .960$. On average, the women's K_{VERT} values were 18% less than were those of the men across both hopping conditions (Figure 3). The main effect for gender was not influenced by preferred hopping rate, flight time, or duty cycle because those variables were equivalent between men and women ($p > .05$). Once normalized for body mass, $K_{VERT\text{-}NORM}$ values were nearly identical between genders, and the gender differences were no longer apparent (Table 1), $F(1, 18) = .002$, $p = .962$.

Joint Position and Range of Motion

There were no significant main effects or interactions involving gender for ANG_{KNEE}, ANG_{ANKLE},ROM_{KNEE}, and ROM_{ANKLE}. Those findings indicate that joint angles at initial ground contact as well as joint excursions of the knee and ankle joints were similar between genders during $FREQ_{PREF}$ and $FREQ_{3.0}$ hopping conditions (Table 2). There were significant main effects involving hopping frequency for both ROM_{KNEE}, $F(1, 19) = 41.766$, $p = .001$, and ROM_{ANKLE}, $F(1, 19) = 30.626$, $p = .001$. Those findings indicated that joint excursions during $FREQ_{3.0}$ were significantly decreased in comparison with their values during $FREQ_{PREF}$ hopping conditions by 66% and 35% at the knee and ankle joints, respectively (Table 2).

Muscle Activity

Quadriceps Activity—A significant main effect for gender was demonstrated for quadriceps activity, $F(1, 19) = 4.981$, $p = .038$ (Figure 4). Overall, women participants recruited 46% greater quadriceps activity than did the men across both hopping conditions. In addition, there was a significant main effect for phase of muscle firing, revealing that quadriceps activity was significantly greater during the LR phase than during the PR phase, $F(1, 19) = 42.210$, $p = .001$. On average, there was a 48% increase from the PR to the LR phase for both men and women. The observed gender difference in quadriceps activation was not influenced by the phase of muscle firing (PR and LR) as shown by a nonsignificant Gender × Phase interaction ($p = .096$). We observed that the increase in quadriceps activity from the PR to the LR phase was statistically similar between genders because quadriceps activity increased by 49% and 47% for men and women, respectively (Table 3).

Hamstrings Activity—No significant main effects or interactions involving gender were demonstrated ($p > .05$; see Figure 4). Thus, men and women used similar levels of hamstrings activation during the hopping tasks (Table 3).

Gastrocnemius Activity—There were no significant main effects or interactions involving gender (Figure 4), but there were significant main effects for both phase, $F(1, 19) = 47.706$, $p = .001$, and hopping frequency conditions, $F(1, 19) = 6.789$, $p = .017$. There was a 25% increase in gastrocnemius activity from the PR to the LR phase, and overall gastrocnemius activity was increased by 16% during $FREQ_{3.0}$ conditions compared with that in the $FREQ_{PREF}$ hopping (Table 3).

Soleus Activity—Significant main effects were found for phase, $F(1, 19) = 37.284$, $p = .001$, hopping frequency condition, $F(1, 19) = 20.304$, $p = .001$, and gender, $F(1, 19) = 6.883$, $p = .017$. Overall, female participants had 37% greater SO activity than did the men (Figure 4), and there was a 50% increase in SO activity from the PR to the LR phase. Similar to gastrocnemius activity, during $FREQ_{3.0}$ conditions, SO activity increased 33% over that displayed when individuals' hopped at their $FREQ_{PREF}$ (Table 3).

Anterior Tibialis Activity—There were no significant main effects or interactions involving gender (Figure 4) or hopping frequency, but there was a significant main effect for phase, $F(1, 19) = 18.810$, $p = .001$. Activation of the AT during the LR phase increased by 30% from that observed during the PR phase (Table 3).

Q:H Coactivation Ratio—Significant main effects for gender, $F(1, 19) = 4.969$, $p = .038$, and phase, $F(1, 19) = 42.035$, $p = .001$, were demonstrated. The main effect for gender revealed that Q:H coactivation ratios were significantly greater in women than in men (Figure 5). In comparison with the men (Q:H = 1.54), the women's Q:H values were significantly greater; their quadriceps activation was 2.01 times greater than their hamstrings activation for both hopping conditions. The significant main effect for phase showed that Q:H coactivation ratios significantly increased from the PR to the LR phase of hopping. The observed gender differences in Q:H coactivation ratios were not significantly affected by phase (PR and LR), as evidenced by the nonsignificant Gender × Phase interaction ($p = .097$). Thus, Q:H coactivation ratios increased in similar fashion from the PR to the LR phase for men and women (Table 4).

TS:AT Coactivation Ratio—There were no significant main effects or interactions involving gender, but there was a significant main effect for phase, $F(1, 19) = 19.315$, $p = .001$ (Table 5). TS:AT coactivation ratios during the LR phase increased from those observed during the PR phase.

It should be noted that there were no significant main effects or interaction involving muscle for each of the muscles tested in this study ($p > .05$). As such, our hypothesis that gender differences in muscle activation strategy are influenced by muscle side was not supported.

Discussion

Our objective in the current study was to examine the influence of gender on K_{VERT} and stiffness recruitment strategy during a functional weight-bearing task—specifically two-legged hopping. Our results demonstrated that women have reduced K_{VERT} in comparison with men; however, that difference was no longer evident once we normalized for body mass ($K_{VERT\text{-}NORM}$). Women also used a different stiffness recruitment strategy than men. Those results suggest that women place greater reliance on their quadriceps and soleus muscles to modulate the torsional joint stiffness about the knee and ankle joints, respectively. The increased quadriceps activity observed in women was not associated with greater scaling of hamstrings activity. The observed gender differences in muscle activation strategy were not influenced by phase (PR vs. LR) or muscle side (M vs. L). We speculate that the observed gender differences in stiffness recruitment strategy may have implications for the greater incidence of noncontact ACL injuries observed in women. Our speculation is based on previous research demonstrating the influence of muscle activation and movement patterns on ACL loading and strain.

Vertical Leg Stiffness

One of the findings of this research was that women demonstrated 18% less K_{VERT} than did their male counter-parts. Those findings were supported by the results of an earlier investigation performed by Granata, Padua, et al. (2002) in which lower extremity stiffness was 23% less in women than in men when they were tested across three separate hopping frequencies (preferred, 2.5, and 3.0 Hz). As in the current study, the gender differences in lower extremity stiffness during hopping were no longer significant after we normalized for body mass. Our findings agree with the results of other studies indicating that individuals with greater body mass exhibit increased leg spring stiffness (Farley et al., 1993; Greene & McMahon, 1979) because stiffness of the leg spring is directly proportional to body mass (Farley et al.). According to the spring-mass model of harmonic motion, K_{VERT} must change in proportion to the mass of the system so that a constant hopping frequency (ω) can be maintained, because $K_{VERT\text{-}NORM}$ is proportional to ω^2. Because hopping frequencies were nearly identical between genders in the present study, it is logical and necessary that their $K_{VERT\text{-}NORM}$ was also identical. Therefore, gender differences in K_{VERT} observed in this study appear to be the result of differences in body mass because, on average, the women weighed 17% less than men. It remains to be determined why men and women have identical preferred hopping frequencies.

Our results support previous research findings indicating that leg spring stiffness varies in accordance with hopping frequency (Farley et al., 1991; Farley & Gonzalez, 1996). While hopping at the faster, controlled frequency of $FREQ_{3.0}$, participants exhibited greater K_{VERT} in comparison with that in the slower, uncontrolled $FREQ_{PREF}$ hopping conditions. On the basis of the simple spring-mass model of harmonic motion, we expected that K_{VERT} would increase when participants hopped at $FREQ_{3.0}$ in comparison with their K_{VERT} in $FREQ_{PREF}$ conditions. A system with constant mass must increase K_{VERT} at faster hopping frequencies to maintain functional performance.

Those results suggest that the basic mechanism for increasing K_{VERT} at faster hopping frequencies was to increase the torsional stiffness about the knee and ankle joints. That conclusion is based on our findings, which demonstrated significant decreases in both ROM_{KNEE} and ROM_{ANKLE} from the slower $FREQ_{PREF}$ to the faster $FREQ_{3.0}$ hopping conditions (Table 2). Although we did not specifically measure torsional joint stiffness properties, a reduction in total knee and ankle joint excursion during ground contact suggests an increase in torsional knee and ankle joint stiffness, assuming similar joint moments between $FREQ_{PREF}$ and $FREQ_{3.0}$ hopping conditions.

Stiffness Recruitment Strategy

Knee and Ankle Kinematics—The absence of a kinematic strategy difference between genders was somewhat of a surprise because previous researchers have demonstrated more extended knee angles at ground contact when women performed cutting maneuvers or landed from a jump (Malinzak et al., 2001; McNitt-Gray et al., 1993). The previously identified kinematic differences between genders were observed during physical tasks that involved rapid deceleration as the individual attempted to halt motion. Deceleration during two-legged hopping is probably much less demanding than cutting and landing maneuvers, hence potentially explaining the difference between current and previous research findings. We believe that the previously reported gender differences in lower extremity kinematics may represent an altered strategy used to modulate K_{VERT} in women during more strenuous functional tasks (cutting and landing from a jump). Performing those functional tasks with greater knee extension would serve as an effective mechanism to increase K_{VERT} with little extra work from surrounding knee musculature. However, that type of movement strategy during strenuous physical tasks may compromise overall knee stability because greater ground-reaction and resultant knee joint forces accompany that type of functional posturing (DeVita & Skelly, 1992; McMahon et al., 1987). The absence of a gender difference in kinematic strategies suggests that during the less demanding hopping tasks, the primary distinguishing factor in stiffness recruitment strategy between genders lies within the muscle activation strategies they use to modulate K_{VERT}.

Muscle Activity and Coactivation—A major finding of the present work was that although there were no gender differences in K_{VERT} once we normalized it for body mass, women performed two-legged hopping with substantially greater quadriceps and soleus muscle activity than did the men. It is interesting that although the women were of lesser mass, they still recruited greater quadriceps activity than the men because they scaled their preparatory quadriceps activity in anticipation for ground contact to a larger extent (~2 times greater) than the men did. The greater reliance on quadriceps activation displayed by women was further emphasized during the LR phase because the absolute increase in quadriceps activation during the LR phase was 31% in women but was only 17% in men (Table 3). Those data suggest that women attempt to modify K_{VERT} in part through greater recruitment of the quadriceps muscles. That type of quadriceps-dominant recruitment strategy has been previously identified in women during controlled (Huston & Wojtys, 1996) and functional (jumping and cutting) tasks (Hewett et al., 1996;Malinzak et al., 2001). This is the first report of a quadriceps-dominant strategy in women during a hopping task. A quadriceps-dominant profile for women has now been identified in several different research studies that incorporated different measurement techniques and physical tasks. We believe the quadriceps-dominant profile in women may influence the gender bias in noncontact ACL injury rates.

Although greater quadriceps activation may serve as an effective mechanism for modulating K_{VERT} during two-legged hopping, its effects on knee joint stability are potentially injurious. Large quadriceps forces result in increased anterior tibial shear forces that cause anterior translation of the tibia with respect to the femur, placing increased forces and strain on the

ACL (Beynnon et al., 1995; Durselen et al., 1995; Fleming et al., 2001; Hirokawa et al., 1992; Li et al., 1999; K. L. Markolf et al., 1990; Renstrom et al., 1986). Excessive strain is the ultimate cause of ACL injury; thus, a quadriceps-dominant recruitment strategy may place women closer to their injury threshold by facilitating quadriceps-induced ACL strain.

Minimizing ACL strain and injury risk requires an increase in antagonist hamstrings activation to counteract quadriceps muscle contraction (Draganich, Jaeger, & Kralj, 1989; Li et al., 1999; Pandy & Shelburne, 1997; Renstrom et al., 1986; Shelburne & Pandy, 1998). In this study, women used an imbalanced recruitment between the quadriceps and hamstrings (Q:H = 1.7). In contrast, men demonstrated a relatively balanced recruitment strategy (Q:H = 1.3; Figure 5). During the LR phase, both genders exhibited significantly increased Q:H ratios because quadriceps activity increased (47% increase from PR to LR) with no significant change in hamstrings activation (PR = 23.0%, LR = 23.8%) to stiffen the knee joint and support the external knee flexion moment. During the LR phase, the quadriceps activity of women was 2.3 times greater than that of their hamstrings, whereas in men, quadriceps activity was only 1.7 times that of their hamstrings. Thus, the Q:H ratio in men during the weight-bearing LR phase was essentially identical to that of women during the non–weight-bearing PR phase. We were unable to find previous research investigations of gender differences in Q:H coactivation; thus, we were unable to compare our findings with those of previous research.

Overall, soleus muscle activation was 38% greater in women than in men. Greater reliance on soleus muscle activity is an efficient strategy for controlling K_{VERT}. However, it is assumed that the soleus muscle is unable to protect the ACL because of its anatomical location. Women also demonstrated a trend for greater gastrocnemius activity than men (25% greater in women), but that difference was not significant (p = .277). We calculated mean overall effect size of 0.64 for that comparison, indicating that, although not statistically significant, the greater gastrocnemius activity in women may be of clinical significance. Recent researchers have demonstrated that the gastrocnemius is an antagonist to the ACL by increasing ACL strain (Fleming et al., 2001). That research revealed that the greatest ACL strain was produced during simultaneous activation of the gastrocnemius and quadriceps muscles (Fleming et al.). The exact mechanism by which gastrocnemius contraction facilitates ACL strain is unknown. It is speculated, however, that because of its anatomical attachment the gastrocnemius creates a posterior shear force on the femur, which may result in posterior translation of the femur relative to the tibia. Essentially, because that creates anterior translation of the tibia relative to the femur, it is known to increase ACL strain. Women, who recruit significantly greater quadriceps activation and tend to recruit greater gastrocnemius activation in comparison with men, may use a muscle recruitment strategy that facilitates ACL strain and places them at risk for ACL injury.

Therefore, women appear to use a different stiffness recruitment strategy than men during hopping. In comparison with men, women use a quadriceps-dominant and an ankle-dominant stiffness recruitment strategy that involves significantly greater quadriceps and soleus activity, a tendency for greater gastrocnemius activity, and minimal coactivation of the hamstrings. A quadriceps- and ankle-dominant stiffness recruitment strategy will efficiently modulate K_{VERT} so that the functional demands of the physical task can be met and sustained. We observed that strategy because, once we accounted for gender differences in body mass, the women displayed equivalent lower extremity stiffness values. However, we speculate that the stiffness recruitment strategy observed in women may potentially influence the gender bias associated with ACL injury risk.

Potential explanations for gender differences in muscle activation include gender differences in knee joint moment during hopping, neuromuscular control, strength, rate of force production, and active muscle stiffness. We do not feel that gender differences in knee joint moment can explain the differences in muscle activation because there were no differences between men and women in joint angles or duration of activity (hopping frequencies) and the data were scaled to anthropometrics. The reduced muscular strength of women has been documented (J. Griffin, Tooms, Zwaag, Bertorini, & O'Toole, 1993; Hakkinen, Kraemer, & Newton, 1997; Huston & Wojtys, 1996; Kanehisa, Okuyama, Ikegawa, & Fukunaga, 1996; Miller, MacDougall, Tarnopolsky, & Sale, 1993), as have their reduced rate of muscular force production (Bell & Jacobs, 1986; Hakkinen & Hakkinen, 1991; Komi & Bosco, 1978; Viitasalo & Komi, 1978; E. M. Winter & Brookes, 1991) and active muscle stiffness compared with those of men (Blackburn et al., 2004; Granata, Wilson, et al., 2002). Although women use greater quadriceps activation, it is possible that the relative force acting at the knee and ankle may not differ between men and women. The observed gender differences in muscle activation may represent a feedforward neuromuscular control strategy whereby women compensate for decreased muscular strength, rate of force production, and active muscle stiffness by increasing activation of the quadriceps and soleus muscles. It is important to note that speculation concerning muscle force solely on the basis of muscle activity during dynamic motion is tenable. Although muscle activity level is an important factor in determining a muscle's output, the resultant contractile force is also influenced by muscle length, contractile velocity, and contraction mode (isometric, concentric, and eccentric). Therefore, we do not suggest that muscle activity level is a direct representation of the resultant muscle force at the joint. That limitation should be considered when interpreting the results of this study. We are unable to definitively explain the quadriceps- and ankle-dominant stiffness recruitment strategy used by women during hopping.

Conclusions

Women demonstrated reduced K_{VERT} during the functional hopping tasks in comparison with that of men. We attribute the gender difference in K_{VERT} to the lighter body mass observed in women because once we normalized for body mass there were no significant differences in $K_{VERT\text{-}NORM}$ between men and women. That result indicates that the gender differences in K_{VERT} during a functional hopping task are likely functions of anthropometric differences.

When comparing the stiffness recruitment strategy between genders during a functional hopping task, we revealed that female participants recruited significantly greater quadriceps and soleus activity that was not associated with increased hamstrings activity. In theory, the recruitment strategy may efficiently modulate K_{VERT}. However, it may compromise stability at the knee joint. We are unable to explain why the women used a different stiffness recruitment strategy than men did, yet demonstrated equivalent lower extremity stiffness once we had normalized for body mass. However, similar gender differences in muscle activation strategies have been reported previously. Future research is necessary to determine the factors contributing to the observed gender differences in stiffness recruitment strategy. In addition, whether the quadriceps-dominant and ankle-dominant strategies used by women actually place the ACL at greater risk for injury is still unknown and requires further study.

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FIGURE 1.

Lower extremity spring force (vertical ground reaction force) plotted as a function of vertical center of mass (COM) displacement during the ground-contact phase of a single trial at the preferred ($FREQ_{PREF}$) and 3.0-Hz ($FREQ_{3.0}$) hopping frequencies. The linear relationship is an indicator that the lower extremity behaved like a simple spring-mass system at both hopping frequencies. The slope (dashed line) of the force versus displacement curves represents the vertical leg stiffness during hopping. As hopping frequency increased, the slope, that is, vertical leg stiffness (K_{VERT}), also increased. The graph represents the exemplar records from a single participant.

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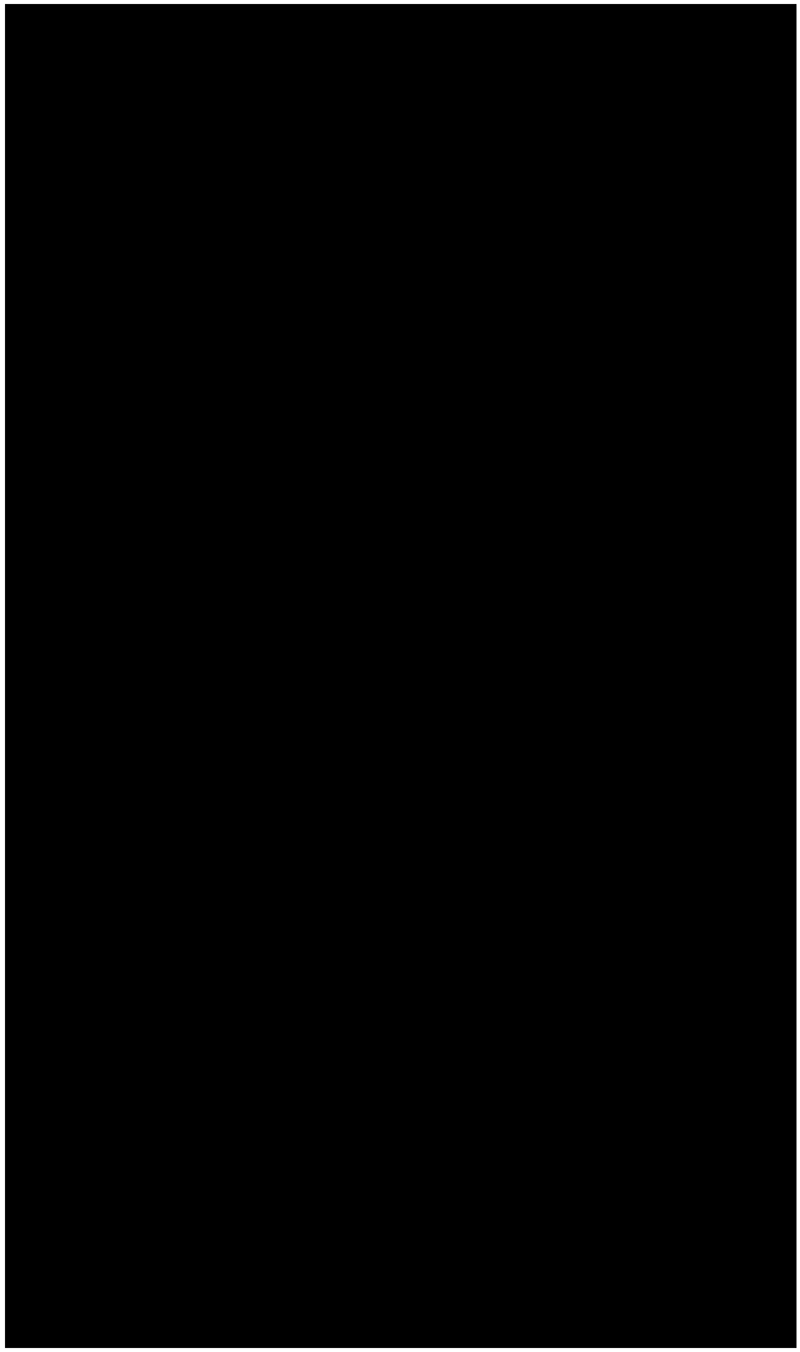

FIGURE 2.
Representation of vertical ground reaction force (VGRF) and electromyographic (EMG) activity normalized to maximal voluntary isometric contraction (MVIC) for the quadriceps, hamstrings, gastrocnemius, and soleus muscles for 1 male and 1 female participant during the preparatory (PR) and loading (LR) response phases. The typical EMG records for all participants are shown.

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FIGURE 3.
Effect of gender on vertical leg stiffness (K_{VERT}). Data were pooled across the preferred ($FREQ_{PREF}$) and 3.0-Hz ($FREQ_{3.0}$) hopping frequency conditions. The men's K_{VERT} was increased in comparison with the women's. When K_{VERT} was normalized for body mass ($K_{VERT\text{-}NORM}$), gender differences were no longer significantly different. Asterisk (*) indicates significantly greater K_{VERT} ($p < .05$).

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FIGURE 4.
Effect of gender on muscle activity. Data were pooled across preferred ($FREQ_{PREF}$) and 3.0-Hz ($FREQ_{3.0}$) hopping frequency conditions, preparatory (PR) and loading (LR) response phases, and muscle side (e.g., rectus femoris and vastus medialis). Asterisk (*) indicates significantly greater muscle activity in female participants ($p < .05$). MVIC = maximal voluntary isometric contraction.

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FIGURE 5.
Effect of gender on quadriceps:hamstrings coactivation ratios (Q:H). Data were pooled across preferred ($FREQ_{PREF}$) and 3.0-Hz ($FREQ_{3.0}$) hopping frequency conditions and preparatory (PR) and loading (LR) response phases. Asterisk (*) indicates significantly greater Q:H in women than in men ($p < .05$).

TABLE 1

Average Vertical Leg Stiffness (kN/M) for Men and Women During Preferred Frequency and 3.0-Hz Hopping Conditions (*M* ± SD)

	$FREQ_{PREF}$		$FREQ_{3.0}$	
Gender	**M**	**SD**	**M**	**SD**
		K_{VERT}		
Men	28.02	7.55	41.52	5.16
Women	21.61	8.62	35.32	5.28
		$K_{VERT\text{-}NORM}$		
Men	0.36	0.11	0.52	0.04
Women	0.33	0.15	0.53	0.07

Note. K_{VERT} = Vertical leg stiffness; $K_{VERT\text{-}NORM}$ = vertical leg stiffness normalized to body mass; $FREQ_{PREF}$ = preferred hopping frequency; $FREQ_{3.0}$ = 3.0-Hz hopping frequency.

TABLE 2

Average Landing Angle (Degrees) and Excursion Values (Degrees) for the Knee and Ankle During Preferred Frequency and 3.0-Hz Hopping Conditions

	$FREQ_{PREF}$				$FREQ_{3.0}$			
	Landing angle		Excursion		Landing angle		Excursion	
Gender	M	SD	M	SD	M	SD	M	SD
Knee								
Men	18.24	4.04	18.24	8.19	18.88	4.29	6.57	2.93
Women	21.47	4.28	21.02	10.31	18.64	7.39	6.63	3.98
Ankle								
Men	17.41	8.34	28.01	14.07	19.92	9.66	17.09	8.64
Women	17.34	8.33	24.44	13.91	15.66	6.02	17.44	11.31

Note. $FREQ_{PREF}$ = preferred hopping frequency; $FREQ_{3.0}$ = 3.0-Hz hopping frequency.

TABLE 3

Average (and Standard Deviation) Thigh and Ankle Muscle Activity Values (% MVIC) for the Preparatory and Loading Response Phases During Preferred Frequency and 3.0-Hz Hopping Conditions

	Preparatory				Loading			
	$FREQ_{PREF}$		$FREQ_{3.0}$		$FREQ_{PREF}$		$FREQ_{3.0}$	
Gender	**M**	**SD**	**M**	**SD**	**M**	**SD**	**M**	**SD**
				Quadriceps				
Men	16.80	18.66	19.60	18.97	37.70	8.26	34.40	29.09
Women	31.00	20.56	39.50	20.89	67.10	42.12	65.30	32.17
				Hamstrings				
Men	19.00	10.44	23.20	14.55	25.80	10.44	27.30	10.75
Women	23.30	11.28	21.90	15.92	24.00	11.61	22.60	11.61
				Gastrocnemius				
Men	86.10	67.99	104.70	89.81	135.50	89.81	144.50	71.78
Women	118.20	74.62	160.70	98.84	160.00	98.84	186.60	78.94
				Soleus				
Men	44.60	26.56	121.30	72.73	75.80	62.30	166.00	92.97
Women	74.60	29.19	185.00	9.93	160.20	68.65	232.30	102.48
				Anterior tibialis				
Men	21.10	13.60	22.10	12.97	28.80	20.87	30.10	19.92
Women	29.80	14.92	29.90	14.26	45.40	23.22	42.70	21.89

Note. $FREQ_{PREF}$ = preferred hopping frequency; $FREQ_{3.0}$ = 3.0-Hz hopping frequency; % MVIC = percentage of maximal voluntary isometric contraction.

TABLE 4

Average Agonist:Antagonist Coactivation Ratios for the Preparatory and Loading Response Phases During Preferred Frequency and 3.0-Hz Hopping Conditions

	Preparatory				Loading			
	$FREQ_{PREF}$		$FREQ_{3.0}$		$FREQ_{PREF}$		$FREQ_{3.0}$	
Gender	**M**	**SD**	**M**	**SD**	**M**	**SD**	**M**	**SD**
				Q:H				
Men	1.33	0.28	1.39	0.22	1.75	0.47	1.68	0.35
Women	1.62	0.48	1.79	0.52	2.35	1.06	2.31	0.80
				TS:AT				
Men	1.21	0.06	1.22	0.10	1.28	0.10	1.30	0.13
Women	1.29	0.19	1.30	0.17	1.46	0.30	1.43	0.27

Note. Q:H = Quadriceps relative to hamstrings coactivation ratio; TS:AT = Triceps surae relative to anterior tibialis coactivation ratio; $FREQ_{PREF}$ = preferred hopping frequency; $FREQ_{3.0}$ = 3.0-Hz hopping frequency.