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EFFECT OF FREQUENCY ON HUMAN UNIPEDAL HOPPING[1]

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Summary.—All mature forms of locomotion involve periods of unilateral stance. Unipedal hopping may provide useful information about the neuromuscular and biomechanical capabilities of a single lower extremity in adults. This study investigated whether hopping influenced vertical stiffness and lower extremity angular kinematics during human unipedal hopping. Vertical force and two-dimensional kinematics were measured in 10 healthy males hopping at three frequencies: preferred, +20%, and −20%. At +20%, compared to preferred, vertical stiffness increased 55% as hip flexion, knee flexion, and ankle dorsiflexion decreased, while at −20% vertical stiffness decreased 39.4% as hip flexion, knee flexion, and ankle dorsiflexion increased. As in bipedal hopping, the force-displacement relationship was more springlike at the preferred rate and +20% than at −20%. Given the prevalence of unilateral stance during walking, running, and skipping, findings related to unipedal hopping may be useful in the rehabilitation or conditioning of lower extremities.

Hopping, developmentally similar to the skip and gallop (Clark & Whitall, 1989), is a unipedal or bipedal bouncing motion that requires strength, balance, coordination, and control. Although the biomechanical and neuromuscular aspects of hoping in adults are similar to children (Moritani, Oddsson, Thorstensson, & Åstrand, 1989), hopping is less frequently observed and often recreational in adults (Burton, Garcia, & Garcia, 1999). Despite the fact that unipedal hopping involves a unilateral stance similar to walking, running, and skipping, few studies have addressed unipedal hopping (Mellvill-Jones & Watt, 1971; Austin, Garrett, & Tiberio, 2002).

Prior works support a simple mass-spring model during running, trotting, and hopping (Cavagna, Saibene, & Margaria, 1964; Cavagna, Heglund, & Taylor, 1977; Cavagna, Franzetti, Heglund, & Willems, 1988; Dalleau, Belli, Bourdin, & Lacour, 1998). In such a model, stiffness (k) is a function of force (F) and displacement (Δx), i.e., $-k = F/\Delta x$, and the period (τ) of the oscillation is a function of mass (m), and stiffness, i.e., $\tau = 2\pi\sqrt{m/k}$. In bipedal hopping vertical stiffness is related to frequency (Farley, Blickhan, Saito, & Taylor, 1991; Farley, Glasheen, & McMahon, 1993; Farley & Gonzalez,

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1996; Ferris & Farley, 1997). However, Farley, *et al.* (1991) found behavior like a simple mass-spring system only at frequencies at or above 2.2 Hz. Despite the predominance of unilateral stance, it is unknown whether the same is true of unipedal hopping.

During bipedal hopping, the ankle plays a greater role than the knee and hip (Fukashiro & Komi, 1987; Dyhre-Poulsen, Simonsen, & Voigt, 1991; Fukashiro, Komi, Jarvinen, & Miyashita, 1995; Farley, Houdijk, Van Strien, & Louie, 1998; Farley & Morgenroth, 1999). Apart from the related finding that increased knee flexion during running decreased vertical stiffness (McMahon, Valiant, & Frederick, 1987), there are no reports on the kinematics of unipedal hopping. Unipedal hopping places greater demands on the limb with respect to balance, control, coordination, and strength than bipedal hopping. Specifically, we hypothesized that (1) a hopping frequency 20% greater than preferred would result in an increase in vertical stiffness and a decrease in angular displacement, (2) a hopping frequency 20% less than preferred would result in a decreased vertical stiffness and an increase in angular displacement, and (3) the hip, knee, and ankle would contribute equally across frequencies.

Method

Materials

The sample consisted of 10 healthy male volunteer participants between 18 and 30 years of age (height: *M* 1.79 m, range 1.71–1.88; body mass: *M* 76.43 kg, range 63.74–84.99). With approval by the Committee on the Use of Human Subjects in Research at the University of Connecticut, all participants read and signed a statement of informed consent prior to participation.

Procedure

To assess the effects of frequency on unipedal hopping, we measured hip flexion, knee flexion, ankle dorsiflexion, and vertical stiffness at three hopping frequencies: preferred, +20%, and −20%. Individual preferred hopping frequency was determined using the mean of five 10-sec. trials at a self-selected preferred pace. The overall preferred hopping frequency, 2.03 Hz, was consistent with previous work on unipedal hopping (Melvill-Jones & Watt, 1971) and bipedal hopping (Blickhan, 1989; Dyhre-Poulsen, *et al.*, 1991; Farley, *et al.*, 1991).

Reflective markers were attached to the skin overlying the following anatomical landmarks on the right side of the body: acromion process, greater trochanter, lateral femoral condyle, lateral melleolus, calcaneus, and fifth metatarsal. Participants hopped vertically on the right leg while maintaining synchrony with an electronic metronome. To ensure vertical motion, a 15- ×

15-cm square was outlined on the force platform and contacts outside this were not analyzed. Each trial lasted 5 sec., and participants rested for 60 sec. between trials to minimize fatigue. The participants performed five trials at each of the three hopping frequencies, and trials were randomized across frequencies.

Two-dimensional videographic data were sampled at 60 Hz and a shutter rate of .001 sec. using the Peak Performance Motion Measurement System (Peak Performance Technologies, Inc., Englewood, CO). Following automatic digitization, the raw data were smoothed using optimal filtering parameters (Jackson, 1979). Angular displacements of the hip, knee, and ankle joints were calculated from the smoothed data. Maximal angular displacements for ankle dorsiflexion, knee flexion, and hip flexion were measured during the contact phase of the hopping cycle.

Vertical force data were sampled at 500 Hz during a 5-sec. sampling period using an AMTI OR6-5-2000 biomechanical force platform (Advanced Mechanical Technologies, Inc., Watertown, MA). Vertical displacement of the center of mass was estimated from double integration of the vertical force signal after the body mass and added mass were subtracted (Cavagna, 1975; Blickhan & Full, 1992). Displacement of the center of mass was measured during each contact phase (defined as the time on the force platform from initial contact to takeoff). For each cycle a line of best fit through the force-displacement data was found using a least squares method. Vertical stiffness for each trial was computed from the slope of the force-displacement relation for the first four cycles of that trial (Rack & Westbury, 1974, 1984).

Analysis

Two-way (Frequency × Trials) repeated-measures analyses of variance tested for differences across Frequencies and Trials for vertical stiffness, hip flexion, knee flexion, and ankle dorsiflexion. Tukey's *HSD post hoc* comparison was used to test for significant pairwise differences when significant main effects or interactions were present. Following a Bonferroni-type correction, the overall alpha level for all statistical tests was .0125 (i.e., .05/4).

Results

Vertical stiffness changed significantly across the three hopping frequencies ($F_{2,18}=61.54$, $p<.01$); see Table 1. Compared to the preferred frequency, vertical stiffness decreased 39.4% at −20% and increased 55% at +20%. The force and displacement relationship was typically linear at both preferred and +20% approximating a simple mass-spring; see Fig. 1. At −20%, however, the force-displacement relation was less than linear, uncharacteristic of a simple mass-spring, and thus, vertical stiffness at −20% may be overestimated; see Fig. 1.

TABLE 1
MEAN KINETIC AND KINEMATIC MEASURES AT THREE HOPPING FREQUENCIES

Measure	Hopping Frequency					
	−20%		Preferred		+20%	
	M	*SD*	*M*	*SD*	*M*	*SD*
Vertical Stiffness, kN/m	8.9[a]	6.0	14.6	6.6	22.7[b]	6.6
Hip Flexion, °	31.7[b]	9.8	25.3	6.9	22.8[a]	8.9
Knee Flexion, °	60.6[b]	8.4	53.6	9.8	47.0[a]	10.2
Ankle Dorsiflexion, °	26.9[c]	5.6	24.3	6.9	20.8[a]	6.0

[a] Significantly less than Preferred ($p<.01$). [b] Significantly greater than Preferred ($p<.01$). [c] Significantly greater than Preferred ($p<.05$).

Hopping frequency significantly altered lower extremity angular kinematics; see Table 1. Ankle dorsiflexion ($F_{2,18}=21.29$, $p=.00002$), knee flexion ($F_{2,18}=47.15$, $p=.000001$), and hip flexion ($F_{2,18}=30.54$, $p=.000002$) were inversely related to hopping frequency. Compared to the preferred frequency, ankle dorsiflexion increased 3.6°, knee flexion increased 6.9°, and hip flexion increased 6.4° at −20%. At +20% ankle dorsiflexion decreased 3.5°,

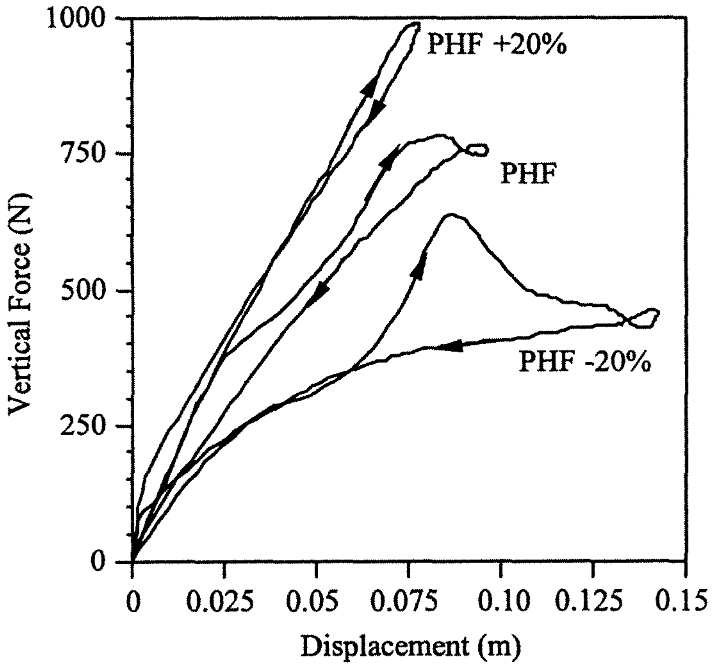

FIG. 1. Representative force-displacement relationships for Subject 10 at the three hopping frequencies. Arrows designate progression of movement during contact phase from initial contact to take-off. PHF is preferred hopping frequency.

knee flexion decreased 6.7°, and hip flexion decreased 3.4° compared to the preferred frequency. The overall difference between −20% and +20% was 7.1° at the ankle, 13.6° at the knee, and 9.8° at the hip. For all trials the mean hopping frequency was within ±4% of the metronomic frequency. There was no significant main effect for trials or a significant interaction between trials and frequency.

Discussion

This study found that hopping frequency predictably affected vertical stiffness and lower extremity angular kinematics during unipedal hopping. Furthermore, the effect of hopping frequency on vertical stiffness is similar to bipedal hopping (Farley, *et al.*, 1991, 1993; Farley & Gonzalez, 1996; Ferris & Farley, 1997). Vertical stiffness at the preferred frequency (14.63 kN/m) was less than at +20% (22.68 kN/m) and greater than at −20% (8.86 kN/m). Although vertical stiffness increased during hopping at +20%, the force-displacement relationship for preferred and +20% resembled a simple mass-spring (see Fig. 1), in which the ideal force-displacement relationship is expressed by a straight line. Consistent with prior findings (Farley, *et al*, 1991), the force-displacement relation during unipedal hopping at frequencies equal to or greater than the preferred frequency is strongly linear, characteristic of a simple mass-spring.

At −20%, however, the force-displacement relation is less than linear and no longer consistent with a simple mass-spring. Unlike the force-displacement relation at preferred or +20%, landing and propulsion are more distinct at −20%; see Fig. 1. During landing, the descent of the center of mass is accompanied by increased and decreased force. In a physical simple mass-spring system this is not possible, but humans can modulate stiffness (Melvill-Jones & Watt, 1971; Nichols & Houk, 1971, 1976; Hoffer & Andreasson, 1981; Sinkjaer, Toft, Andreasson, & Hornemann, 1988; Dyhre-Poulsen, *et al.*, 1991). Possible mechanisms contributing to this atypical force-displacement relation include the stretch reflex and viscoelastic musculotendinous unit. Another possible description of the uncharacteristic force-displacement relation may be the two mass-spring system (Greene & McMahon, 1979; Alexander, Bennett, & Ker, 1986; Alexander, 1988; Farley, *et al.*, 1991). Such a system consists of two distinct masses and deceleration of the first mass leads to a force peak, followed by second peak associated with deceleration of the other mass. Although it is possible the nonweightbearing leg functions as a second mass, this same force-displacement relation exists when both legs are weight-bearing, as in bipedal hopping (Farley, *et al.*, 1991).

Changes in lower extremity geometry across frequencies corresponded with changes in vertical stiffness. At +20% increased vertical stiffness was associated with decreased hip flexion, knee flexion, and ankle dorsiflexion. These findings suggest that, as in other tasks (Greene & McMahon, 1979; Mussa-Ivaldi, Hogan, & Bizzi, 1985; McMahon, *et al.*, 1987; Flash & Mussa-Ivaldi, 1990; Farley & Gonzalez, 1996; Newman, Jackson, & Bloomberg, 1997), limb geometry may contribute to increased vertical stiffness. Decreased stiffness at −20% was accompanied by increased hip flexion, knee flexion, and ankle dorsiflexion. These findings substantiate previous conjectures, based on displacement of the center of mass, about the lower extremity kinematics during landing (Farley, *et al.*, 1991, 1993). Finally, unlike bipedal hopping (Fukashiro & Komi, 1987; Dyhre-Poulsen, *et al.*, 1991; Fukashiro, *et al.*, 1995; Farley, *et al.*, 1998; Farley & Morgenroth, 1999), unipedal hopping requires significant responses from all joints of the lower extremity to hop at frequencies other than the preferred one.

In using the simple mass-spring model, several assumptions are made. The first and most significant is that the lower extremity functions as a linear spring, which was not noted at hopping frequencies less than the preferred one. Second, the contributions of movement by the arms and opposite leg are difficult to assess. Although the role of the arms remains questionable, similarities in the force-displacement relationship for bipedal and unipedal hopping would suggest the effect of the opposite leg would be negligible. Third, despite efforts to constrain horizontal motion, the hopping motion is not purely vertical, thus there is a small but commonly ignored horizontal force component. Lastly, despite the common element of unilateral stance, predictions about walking and running from unipedal hopping should be made cautiously due to subtle, but substantial, mechanical differences. Specifically, foot contact during hopping is almost exclusively on the forefoot, whereas walking and running typically involve rearfoot to forefoot progressions.

Little is known regarding the effect of frequency on unipedal hopping. In this study, vertical stiffness and lower extremity angular kinematics changed predictably across frequencies. The cyclic, bouncing, and constrained nature of hopping may prove useful during rehabilitation or conditioning. Unipedal hopping may offer a viable alternative for assessing and training lower extremity strength, balance, coordination, and control. Training may involve variables such as bipedal vs unipedal, frequency, surface characteristics, height, and mass.

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