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Effect of Added Mass on Human Unipedal Hopping

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EFFECT OF ADDED MASS ON HUMAN UNIPEDAL HOPPING

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Summary.—Although hopping is considered a children's activity, it can be used to provide insight into the neuromuscular and biomechanical performance of adults. This study investigated whether mass added during unipedal hopping altered the vertical stiffness, hopping period, and angular kinematics of the lower extremity of adults. Measures of two-dimensional kinematics and vertical force were made from 10 healthy men during hopping at a preferred period under three conditions: Body Mass, Body Mass + 10%, and Body Mass + 20%. Adding mass significantly increased hopping period and hip flexion without significantly affecting vertical stiffness, ankle dorsiflexion, or knee flexion. Overall, the findings agreed with predictions based on a simple-mass spring model. The results indicate unique kinetic and kinematic responses to increased mass during hopping may have potential application in neuromuscular assessment and training for the lower extremities.

The development of hopping typically occurs between 4 and 7 years of age (Clark & Whitall, 1989), soon after galloping and before skipping. Constraints upon the development of hopping in children include adequate strength, balance, control, and coordination. Although hopping in adults is often recreational (Burton, Garcia, & Garcia, 1999), the neuromuscular and biomechanical patterns of adults and children are similar during hopping (Moritani, Oddsson, Thorstensson, & Åstrand, 1989). Hopping may provide an opportunity to assess the neuromuscular and biomechanical performance of the lower extremities in adults.

Using the mass-spring model, several studies of bipedal hopping have measured stiffness (the force-displacement relationship) while manipulating such variables as imposed hopping frequency (Farley, Blickhan, Saito, & Taylor, 1991; Farley, Glasheen, & McMahon, 1993; Farley & Gonzalez, 1996; Ferris & Farley, 1997), surface characteristics (McMahon & Greene, 1979; Ferris & Farley, 1997), and hopping height (Farley, et al., 1991). Related investigations have focused on the effects of added mass on energy expenditure, electromyographic activity, and kinematics during walking (Alexander,
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1986; Maloiy, Heglund, Prager, Cavagna, & Taylor, 1986; Kram, 1991) and running (Pandolf, Givoni, & Goldman, 1977; Cureton, Sparling, Evans, Johnson, Kong, & Purvis, 1980; Bobet & Norman, 1984). Kram (1991) found that walking with an additional 19% of the body mass on a springy pole supported across the shoulders decreased flight period and increased contact period and vertical ground reaction forces.

However, previous research has not addressed two unique aspects of hopping, (a) the mechanics of unipedal hopping and (b) the effect of an added mass on hopping mechanics. Therefore, this project seeks to address the effect of added mass on the mechanics of unipedal hopping. Important predictions are possible based upon the mass-spring model. In comparison with unweighted hopping, we hypothesized that hopping with an additional 10% and 20% of body weight should result in either (a) an increase in the preferred hopping period and unchanged vertical stiffness or (b) an increase in the vertical stiffness while preferred hopping period remains unchanged. Finally, we hypothesized that the angular kinematics of the lower extremity would be altered by the added mass.

METHOD

Materials

Ten healthy male participants between the ages of 18 and 30 years participated in this study on a voluntary basis (height: $M = 1.79$ m, range 1.71–1.88, weight: $M = 76.43$ kg, range 63.74–84.99). In accordance with approval by the Committee on the Use of Human Subjects in Research at the University of Connecticut, all participants read and signed an informed consent form prior to participation.

Procedure

To assess the effects of mass on performance of unipedal hopping we measured the vertical stiffness and angular displacement of the hip, knee, and ankle under conditions of Body Mass, Body Mass +10%, and Body Mass +20%. During the conditions with added mass participants wore a weight vest (Perform Better, Cranston, RI). Prior to data collection we measured each subject’s body mass and attached spherical reflective markers to the skin overlying the following anatomical landmarks on the right side of the body: fifth metatarsal, calcaneus, lateral malleolus, lateral femoral condyle, greater trochanter, and acromion process. Participants hopped vertically on the right leg at a self-selected, preferred pace on the force platform. To ensure vertical motion a 15- × 15-cm square was outlined on the surface of the force platform and contacts outside these boundaries were not analyzed. The participants performed five trials in each of the three mass conditions, and trials were randomized across conditions. A single trial lasted 5 sec., and participants rested for 60 sec. between trials to minimize fatigue.
Vertical force data were captured at a sampling frequency of 500 Hz for a sampling period of 5 sec. using an AMTI OR 6-5-2000 biomechanical force platform (Advanced Mechanical Technologies, Inc., Watertown, MA). Vertical displacement of the center of mass was determined 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, that is, from initial contact to takeoff). Hopping period was defined as the time between consecutive contacts. For each cycle, a line of best fit through the force-displacement data was found using a least squares method. The vertical stiffness was computed from the slope of the force-displacement relation (Rack & Westbury, 1974, 1984). The vertical stiffness for each trial was calculated over the first four hop cycles of the trial.

Two-dimensional videographic based kinematic data were collected 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, all raw data were smoothed using optimal filtering parameters, as measured by the Jackson “knee” method (Jackson, 1979). From the smoothed raw data, angular displacements of the hip, knee, and ankle joints were calculated. Maximal angular displacements for ankle dorsiflexion, knee flexion, and hip flexion were determined during the contact phase of the hopping cycle.

Analysis

Two-way (Load x Trials) repeated-measures analyses of variance were performed to test for differences across Trials and Loads for vertical stiffness, hopping period, and angular displacements for the three lower extremity articulations. When significant main effects or interactions were present, a Tukey HSD post hoc comparison tested for significant pairwise differences. Following a Bonferroni correction to the overall alpha level for all statistical tests was .01, i.e., .05/5.

Results

Vertical stiffness did not significantly change across the three loads ($F_{2,18} = .76$, $p = .48$), although there was a small but nonsignificant drop in stiffness at Body Mass + 20%; see Table 1. However, hopping period varied significantly across the three loads ($F_{2,18} = 9.52$, $p = .002$). Specifically, hopping period varied directly with load (see Table 1) and was significantly greater at Body Mass + 20% than at Body Mass + 10% and Body Mass. Hopping periods at Body Mass and Body Mass + 10% were not significantly different from one another.

The angular kinematics of the lower extremity during hopping were al-
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tered by the load. Hip flexion increased significantly with loads greater than Body Mass \((F_{2,18} = 18.89, p = .00003)\); see Table 1. Neither the 1.5° increase in knee flexion \((F_{2,18} = 3.01, p = .07)\) nor the 1° increase in ankle dorsiflexion

<table>
<thead>
<tr>
<th>Measure</th>
<th>Load</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Body Mass</td>
</tr>
<tr>
<td></td>
<td>(M)</td>
</tr>
<tr>
<td>Vertical Stiffness, kN/m</td>
<td>15.9</td>
</tr>
<tr>
<td>Hopping Period, msec.</td>
<td>476</td>
</tr>
<tr>
<td>Hip Flexion, deg.</td>
<td>21.9</td>
</tr>
<tr>
<td>Knee Flexion, deg.</td>
<td>51.5</td>
</tr>
<tr>
<td>Ankle Dorsiflexion, deg.</td>
<td>25.1</td>
</tr>
</tbody>
</table>

\(^a\)p < .01.

(F\(_{2,18} = 4.65, p = .02\)) across loads were significant. Hip flexion varied directly with the load, and both Body Mass +10% and Body Mass +20% were significantly greater than at Body Mass, although the two added mass conditions were not significantly different from one another. None of the omnibus tests resulted in a significant main effect for trials or a significant interaction between trials and frequency.

**DISCUSSION**

This investigation asked whether added mass during unipedal hopping affects vertical stiffness, preferred hopping frequency, and angular kinematics of the lower extremity. With the addition of 10% and 20% of the body mass, hopping period and hip flexion increased while vertical stiffness, ankle dorsiflexion, and knee flexion did not change. Findings are noteworthy considering the relationship among stiffness \((k)\), mass \((m)\), and period \((\tau)\) as defined by the period of oscillation for a mass-spring system, \(\tau = 2\pi\sqrt{m/k}\). Based upon Hooke's Law, \(F = -k\Delta x\), stiffness is the relationship between force and displacement. The ability of humans to modulate force and, therefore, stiffness complicates predictions based upon the simple mass-spring system, a clear limitation of such a model. Accordingly, increasing the mass of the system can either increase the hopping period (for a constant stiffness) or decrease the vertical stiffness (for a constant hopping period). The findings here suggest that, when presented with an additional load, although capable of modulating stiffness, participants maintain vertical stiffness and use a new preferred hopping period. Although limited in scope of analysis, a contributor to this strategy may include increased hip flexion in the absence of significant changes in ankle dorsiflexion and knee flexion.

Studies of landings and bipedal hopping on surfaces of various stiffness have yielded similar responses (Sanders & Wilson, 1992; McNitt-Gray,
Yokoi, & Millward, 1993, 1994; Sanders & Allen, 1993; Ferris & Farley, 1997; Ferris, Louie, & Farley, 1998). The stiffness of the overall body-surface system remains constant across variations in the stiffness of the surface. The findings here suggest the same may be true of an overall body-mass system during unipedal hopping across variations in mass. Although previous work suggests that the body may adopt a preferred frequency, our findings suggest a tendency for a preferred stiffness (Corlett & Mahaveda, 1970; Salventy & Pilitsis, 1971; Molen, Rozendale, & Boon, 1972; Zarrugh, Todd, & Ralston, 1974; Zarrugh & Radcliffe, 1978; Cavanagh & Williams, 1982; Sargent & van der Woude, 1988; Holt, Hamill, & Andres, 1990, 1991; Holt, Jeng, & Fetters, 1991; Cavagna, Mantovani, Willems, & Musch, 1997).

Far less work has been done regarding the effect of additional mass on the temporal nature of cyclical movements. However, the limited findings show cycle period increasing as the mass of the system increases, consistent with the predictions based upon a simple mass-spring system. During walking while carrying a 19% load on a springy pole, African men decreased their cycle period by 9.6% (Kram, 1991). Similarly, in this study a 20% increase in the mass of the system resulted in a 3.98% (19 msec.) increase in hopping period, while vertical stiffness remained unchanged.

The effects of the increased mass on the kinematics of the lower extremity were nearly opposite those found during bipedal hopping at various frequencies. Whereas previous studies have indicated an effect of frequency on ankle kinematics, there was no evidence of this here (Fukashiro & Komi, 1987; Dyhre-Poulsen, Simonsen, & Voigt, 1991; Farley, Houdijk, Van Strien, & Louie, 1998; Farley & Morgenroth, 1999). In fact, ankle dorsiflexion remained nearly constant (1° increase) across the three different loads. Similarly, in contrast to previous findings (Fukashiro & Komi, 1987; Dyhre-Poulsen, et al., 1991; Fukashiro, Komi, Jarvinen, & Miyashita, 1995), knee flexion did not vary (1.5° increase) with changes in load. In a study of “Groucho,” or crouched running, increased knee flexion decreased the effective vertical stiffness of the subject by 18% compared to normal running but increased the energy demands (McMahon, Valiant, & Frederick, 1987). In contrast to these findings, in our study knee flexion and vertical stiffness remained unchanged, perhaps due to the participants being instructed to self-select a preferred pace. Overall, our findings suggest a kinematic response by the lower extremity to increased mass which is opposite to that of increased frequency, i.e., a greater role for the hip than for the knee and ankle.

The simple mass-spring model used here is based upon several assumptions, the most significant being the assumption that the lower extremity functions as a conservative, linear spring, which is not likely the case. Despite attempts to constraint horizontal motion, hopping in place is not pure-
ly vertical; thus, there is a small but commonly disregarded horizontal force component. The contribution of movement by the arms, opposite leg, and weight vest was also assumed to be negligible and was not evaluated. Although enticing, comparisons with walking and running should be made judiciously. The main reason for caution is the difference in foot fall. Whereas walking and running are typically rearfoot to forefoot progressions, foot strike during hopping is entirely on the forefoot.

There is little information available regarding unipedal hopping and the effect of mass on cyclical movement. In this study, varying the mass during unipedal hopping resulted in no significant change in vertical stiffness, increased hopping period, increased hip flexion, and no significant change in knee flexion or ankle dorsiflexion. These results suggest that increases in mass may place increased demand on the hamstrings and gluteals as a function of increased hip flexion. Unipedal hopping may offer a viable alternative for assessing and training lower extremity strength, balance, control, and coordination.

REFERENCES


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