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Leg exoskeleton reduces the metabolic cost of human hopping

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Grabowski AM, Herr HM. Leg exoskeleton reduces the metabolic cost of human hopping. *J Appl Physiol* 107: 670–678, 2009. First published May 7, 2009; doi:10.1152/japplphysiol.91609.2008.— During bouncing gaits such as hopping and running, leg muscles generate force to enable elastic energy storage and return primarily from tendons and, thus, demand metabolic energy. In an effort to reduce metabolic demand, we designed two elastic leg exoskeletons that act in parallel with the wearer's legs; one exoskeleton consisted of a multiple leaf (MLE) and the other of a single leaf (SLE) set of fiberglass springs. We hypothesized that hoppers, hopping on both legs, would adjust their leg stiffness while wearing an exoskeleton so that the combination of the hopper and exoskeleton would behave as a linear spring-mass system with the same total stiffness as during normal hopping. We also hypothesized that decreased leg force generation while wearing an exoskeleton would reduce the metabolic power required for hopping. Nine subjects hopped in place at 2.0, 2.2, 2.4, and 2.6 Hz with and without an exoskeleton while we measured ground reaction forces, exoskeletal compression, and metabolic rates. While wearing an exoskeleton, hoppers adjusted their leg stiffness to maintain linear spring-mass mechanics and a total stiffness similar to normal hopping. Without accounting for the added weight of each exoskeleton, wearing the MLE reduced net metabolic power by an average of 6% and wearing the SLE reduced net metabolic power by an average of 24% compared with hopping normally at frequencies between 2.0 and 2.6 Hz. Thus, when hoppers used external parallel springs, they likely decreased the mechanical work performed by the legs and substantially reduced metabolic demand compared with hopping without wearing an exoskeleton.

biomechanics; spring-mass model; leg stiffness; locomotion; elastic energy

HUMANS use the storage and return of elastic energy during bouncing gaits such as hopping and running (9, 14, 23), where the overall mechanics of the musculoskeletal system have been well characterized and predicted using a spring-mass model (4, 6, 14, 15, 22, 29). This simple model consists of a linear leg spring attached to a point mass that is equivalent to body mass. During the first half of the stance phase in a bouncing gait, the kinetic and gravitational potential energies of the center of mass are stored as elastic energy in the compliant leg spring. Then, during the second half of the stance phase, this elastic energy is returned to the model's point mass. In vivo data of muscle-tendon force and displacements during bouncing gaits have suggested that the storage and return of elastic energy by the compliant musculoskeletal structures of the leg, primarily tendons, greatly reduces the metabolic costs required for bouncing gaits (2, 3, 27, 28). However, the muscles must generate force during each stretch-shorten cycle of the series tendons and compensate for energy losses due to damping (1, 10), thus accruing a metabolic cost.

To accommodate different mechanical tasks and perturbations during hopping and running, humans may or may not adjust leg stiffness. Previous studies (13, 16) have shown that humans can voluntarily adjust leg stiffness to change hopping frequency and running stride frequency when running over level ground. However, runners maintain near-constant leg stiffness across a range of speeds on stiff level surfaces, despite necessary changes in ground reaction forces (GRFs) (13, 16, 22) and metabolic cost (34). Instead of changing leg stiffness, humans achieve faster running speeds by increasing vertical stiffness, decreasing vertical center of mass displacement and contact time, and increasing the angle swept by the leg.

While hopping or running on compliant surfaces, humans modulate leg stiffness, yet maintain peak vertical GRF, contact time, and vertical displacement of the center of mass (14, 20, 21, 26). Hoppers and runners use the rebound of the surface, change leg stiffness, and maintain the spring-like mechanics of the leg and in-series surface combination (31). Because changes in leg stiffness require hoppers and runners to adjust the rate and amount of force generated by the leg muscles, the metabolic energy required for the task is also altered. Thus, changes in leg stiffness considerably affect the overall biomechanics and metabolic energy required to produce bouncing gaits.

We posited that if elastic energy can be efficiently stored and returned in an exoskeleton positioned in parallel to the human leg, rather than in the musculoskeletal system, wearing this exoskeleton could augment human hopping performance by reducing the force requirements of the leg and the metabolic energy required for hopping. We designed two wearable exoskeletons with elastic legs that function as external elastic components acting in parallel to the wearer's legs and transmitting the weight of the body through the exoskeleton directly to the ground (Fig. 1). One exoskeleton comprised a set of multiple leaf springs (MLE), and the other exoskeleton comprised a set of single leaf springs (SLE). Both exoskeletons exhibited a spring stiffness that was nonlinear (Fig. 2). However, we deliberately designed the MLE to approximate a leg spring stiffness (k) that we calculated a priori from the spring-mass model and a hopping frequency of 2.0 Hz.

We had four main hypotheses. First, we hypothesized that human hoppers would adjust their leg stiffness while hopping in each of the exoskeletons so that the combination of the hopper and exoskeleton would behave as a linear spring, with a total stiffness similar to that of normal hopping at the same frequency. Second, we anticipated that hoppers would decrease the rate and amount of muscular force generated by their legs while wearing an elastic leg exoskeleton and thus hypothesized that they would require less metabolic energy to hop over a range of frequencies when wearing either exoskeleton compared with hopping normally without wearing an exoskeleton. Third, since the biological leg essentially behaves as a linear spring during normal hopping, we predicted that an exoskele-
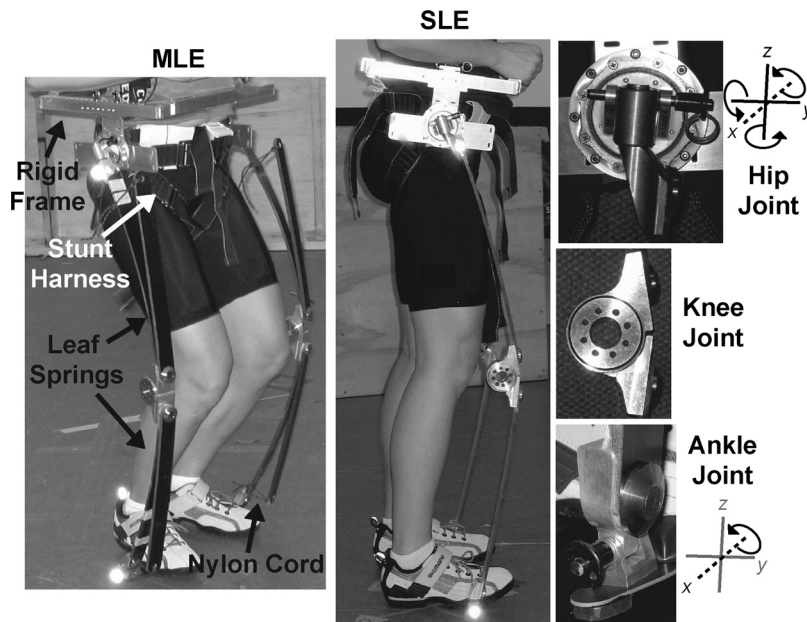
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Fig. 1. Multiple leaf exoskeleton (MLE; *left*), single leaf exoskeleton (SLE; *middle*), 3 degree of freedom "hip" joint (*top right*), fixed "knee" joint (*middle right*), and 2-degree of freedom "ankle" joint (*bottom right*). We engaged an additional set of leaf springs in the MLE to stiffen the exoskeletal leg.

ton with a spring stiffness closer to that of the biological leg would afford hoppers the greatest reduction in metabolic demand. Thus, we hypothesized that hopping while wearing the MLE would require less metabolic energy than hopping while wearing the SLE. Finally, we anticipated that hoppers would require the least metabolic power at the hopping frequency where the greatest amount of elastic rebound work is done by the exoskeleton. This frequency could be predicted using the spring-mass model, where the mass of the system equals the mass of the exoskeleton plus the hopper and the spring stiffness equals the exoskeletal leg stiffness. For each study participant, we arbitrarily designed the MLE to have a spring stiffness based on a 2.0-Hz hopping frequency and thus hypothesized that while wearing the MLE, metabolic cost would be minimized at 2.0 Hz.
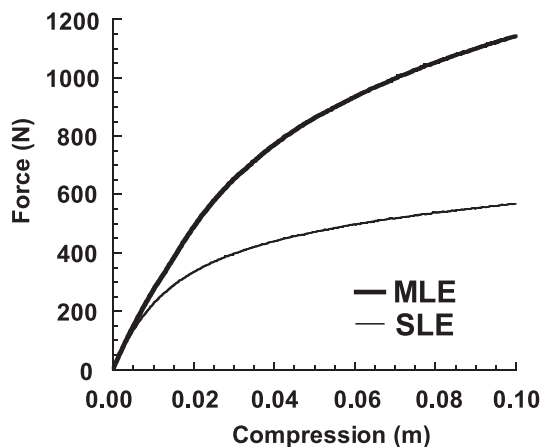

Fig. 2. Representative vertical force versus compression traces from the Instron materials testing machine for a MLE and SLE designed for a 59-kg subject. By engaging two sets of leaf springs (MLE) and compressing them by 0.10 m, we obtained a MLE leg spring stiffness of 11.44 kN/m, which nearly equaled the desired leg spring stiffness of 11.13 kN/m, calculated from a 2.0-Hz hopping frequency and body mass. Using a single set of leaf springs (SLE), we obtained a much more nonlinear force-displacement profile compared with the MLE.

We tested our hypotheses by measuring GRFs, exoskeletal compressions, and metabolic rates while subjects hopped on both legs at 2.0, 2.2, 2.4, and 2.6 Hz. Hopping was an ideal choice for our study because it was easier to design the exoskeletons, the metabolic energy required to hop in place has not been previously reported, and the effects of wearing an exoskeleton during hopping can likely predict the biomechanical and metabolic effects of wearing an exoskeleton during running.

METHODS

Exoskeleton stiffness. We custom made each MLE such that the exoskeletal stiffness approximated a k value that we calculated a priori. To calculate k, we assumed spring-mass mechanics during the rebound stance phase of hopping with a nonzero aerial phase. Using the approach outlined by McMahon et al. (30), we solved the equation of motion for a spring-mass model striking the ground with a vertical velocity (u) upon first impact to get the following expression for GRF (F) as a function of time during a single rebound of the spring-mass system:

$$F/mg = (u\omega_0/g)\sin \omega_0 t + 1 - \cos \omega_0 t \qquad (1)$$

where ω_0 is the natural frequency ($\sqrt{k/m}$), m is total body mass, g is the acceleration due to gravity, and t is the time from first impact. Because the vertical force equals zero at the beginning and end of ground contact (set F = 0 in *Eq. 1*), the total phase of ground contact can be related to u by the following equation:

$$(u\omega_0/g)\sin \omega_0 t_c = \cos \omega_0 t_c - 1. \qquad (2)$$

To estimate contact time (t_c) and u a priori, we chose a hopping frequency of 2.0 Hz and assumed an aerial time (t_a) of 0.14 s based on a previous study of hopping (18). We calculated t_c (0.35 s) from t_a and hopping frequency and then calculated u at the beginning and end of ground contact (0.687 m/s) from $u = t_a g/2$. We then solved *Eq. 2* numerically to determine pairs of values for $\omega_0 t_c$ and $u\omega_0/g$ over the ranges of 2π to π and 0 to ∞, respectively. With knowledge of t_c and u, we determined ω_0 and included m to finally calculate stiffness from $k = m\omega_0^2$ (see Table 1 for values).

We designed each MLE to have an exoskeletal spring stiffness approximately equal to the calculated k and a length approximately

Table 1. *Subject masses and k values*

Subject	Mass, kg	$k_{2.0}$, kN/m	$k_{2.2}$, kN/m	$k_{2.4}$, kN/m	$k_{2.6}$, kN/m	k_{calc}, kN/m	k_{MLE}, kN/m
1	59.5	11.24	13.50	15.59	20.91	11.13	11.44
2	85.0	14.08	18.03	20.95	25.15	15.90	16.61
3	92.3	15.51	17.66	24.11	29.05	17.26	17.49
4	70.3	12.50	15.73	20.12	21.95	13.15	12.79
5	86.8	15.39	19.87	23.47	26.50	16.24	16.71
6	70.8	11.77	14.37	17.98	21.37	13.23	11.63
7	62.3	13.45	15.29	18.83	22.81	11.65	11.61
8	66.4	12.02	14.33	17.70	22.03	12.41	12.47
9	50.9	8.69	12.15	15.30	20.12	9.52	11.44
Mean	71.6±130.8	12.74±20.16	15.66±20.44	19.34±30.13	23.32±20.96	13.39±20.59	13.58±20.57

Shown are subjects' masses, leg spring stiffness during normal hopping at 2.0–2.6 Hz ($k_{2.0-2.6}$), calculated leg spring stiffness for 2.0 Hz (k_{calc}), and multiple leaf exoskeleton (MLE) leg spring stiffness (k_{MLE}) determined from Instron machine measures.

equal to each subjects' intact leg length. We varied exoskeletal spring stiffness by varying the thickness of the fiberglass leaf springs and included two leaf springs on each thigh and shank segment (Fig. 1). The additional set of leaf springs was placed on top of the initial set, attached at the exoskeleton "knee," and tethered to the "hip" and "ankle" joints with a nylon cord (Fig. 1). We adjusted the length of the nylon cord so that the additional leaf springs were engaged just before the base set of leaf springs began to soften; thus, each MLE exoskeletal leg behaved more closely to a linear stiffness leg spring, in contrast to the distinctly nonlinear stiffness response of only single leaf springs (Fig. 2). We also sought to address the effects of an exoskeleton that was less stiff, had a more nonlinear leg spring stiffness, and was slightly lighter than the MLE. Thus, we custom designed a SLE for each subject by removing the extra set of leaf springs from the MLE. On average, the weight of each MLE was 67.5 N, ~10% of each subject's body weight (BW), and the weight of each SLE was 62.6 N, ~9% of each subject's BW.

To confirm that the actual spring stiffness of each exoskeletal leg was nearly the same as the desired spring stiffness, and to determine the exoskeletal force at various levels of compression, we used an Instron materials testing machine (Instron 5580 with a 100-kN load cell, Norwood, MA) to measure vertical force and displacement. We compressed each exoskeletal leg a total of 10 cm while we measured force at 0.33-mm increments of displacement (Fig. 2). We chose a total compression of 10 cm because this represents a hopper's typical center of mass displacement while hopping at a preferred frequency (17). We then determined the actual MLE spring stiffness (k_{MLE}) by assuming a linear stiffness and used a measured peak vertical force (F_{MLE}) and peak displacement (z_{MLE}) of 10 cm, or:

$$k_{MLE} = F_{MLE}/z_{MLE}. \qquad (3)$$

From our measures of all nine MLE forces and compressions, we found that the average k_{MLE} equaled 13.58 kN/m and from our spring mass calculations the average k equaled 13.39 kN/m, a difference of only 1.4%. Thus, we assumed that each custom-made MLE represented a spring-mass system for hopping at a frequency of 2.0 Hz . In contrast, the spring stiffness of the SLE was much more nonlinear and less stiff; the stiffness was relatively high upon initial compression and then became less stiff as the exoskeletal leg was further compressed (Fig. 2).

Subjects. Nine healthy recreational runners [5 women and 4 men, age: 27.3 yr (6.0), body mass: 71.58 kg (13.83), and leg length: 0.929 m (0.055)] participated in the study. All subjects gave informed written consent before participation according to a Massachusetts Institute of Technology Committee on the Use of Humans as Experimental Subjects-approved protocol. Each participant completed three experimental sessions. During the first session, subjects did not wear an exoskeleton. During the second and third sessions, subjects wore a MLE and SLE. The order of the MLE and SLE sessions was random. Before data were collected from the hopping trials during the second

and third sessions, subjects were given as much time as they deemed necessary for accommodation to the exoskeletons (~10 min). Each session evoked a moderate level of exercise intensity; therefore, sessions were separated by at least 2 days to allow the subjects adequate rest and recovery.

Protocol. While subjects stood quietly and then hopped in place on a force platform at 2.0, 2.2, 2.4, and 2.6 Hz, we measured GRFs and
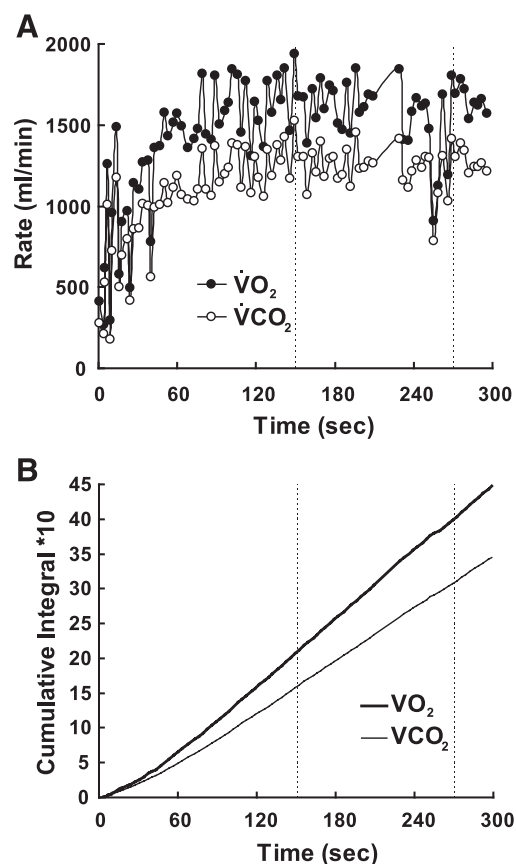

Fig. 3. *A*: example from a 2.4-Hz hopping trial showing rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$). For all subjects during all trials, metabolic rates reached a plateau after ~2 min of hopping. Thus, we averaged $\dot{V}O_2$ and $\dot{V}CO_2$ during *minutes 2.5–4.5* (shown by vertical dotted lines) of each trial and used these average values to calculate metabolic power. *B*: cumulative integrals of the rates of $\dot{V}O_2$ and $\dot{V}CO_2$ calculated from the data shown in *A*. The slope of the cumulative integral with respect to time was nearly constant ($R^2 > 0.994$) after 90 s of hopping, implying that this subject had attained a steady-state metabolic rate. Dotted vertical lines show *minutes 2.5* and *4.5*.

metabolic rates during all sessions and compression of the exoskeletal legs during *sessions 2* and *3*. We followed the same trial order of ascending frequencies during all three sessions to ensure consistency between sessions. Before beginning a session, we instructed subjects to hop in place on both feet, to leave the ground between hops, and to match the beat of a metronome. Subjects wore their own shoes during the first session, and the exoskeletal shoes described subsequently for *sessions 2* and *3*. Each trial lasted 5 min with at least 3 min of rest between trials.

GRF. Subjects stood and hopped in place during all trials on a force platform (AMTI, Watertown, MA) that was mounted flush with the floor. Between *minutes 3* and *5* of each trial, we collected 10 s of GRF data, sampled at 1,080 Hz, and then processed this data using a customized Matlab program (MathWorks, Natick, MA). We filtered the raw force data with a fourth-order zero-lag Butterworth filter that had a cutoff frequency of 30 Hz. Next, we detected the instant of initial foot-ground contact and the instant of toe-off from the vertical GRF data using a threshold of 10 N. These instances allowed us to calculate contact time, aerial time, hopping frequency, and peak GRF for 15 consecutive hops. During all trials, subjects hopped at frequencies that were within ±3% of the desired hopping frequencies.

We then calculated the peak vertical displacement of the center of mass (z_{vert}) during the stance phase according to Cavagna (8). First, we subtracted BW (including the weight of the exoskeleton for *sessions 2* and *3*) from the vertical GRF and divided by m to obtain vertical acceleration. Then, we integrated vertical acceleration to obtain vertical velocity. The required integration constant for vertical velocity was calculated by assuming that the mean vertical velocity per step was equal to zero (7, 8). Finally, we integrated vertical velocity to obtain z_{vert}. We calculated total stiffness (k_t) from the peak vertical GRF (F_{vert}) and peak z_{vert} using the following equation:

$$k_t = F_{vert}/z_{vert}. \qquad (4)$$

To normalize and compare k_t across subjects of varied body mass and height, we also calculated a dimensionless stiffness (k_{dim}) by converting F_{vert} into units of BW and calculating displacement as the ratio of z_{vert} to l, measured as the length from the greater trochanter to the lateral malleolus, or:

$$k_{dim} = (F_{vert}/BW)/(z_{vert}/l). \qquad (5)$$

Exoskeletal compression. During each trial of *sessions 2* and *3*, we used a three-dimensional motion capture system (Vicon 512 System, Oxford, UK) to determine exoskeletal leg compression from the distance between two reflective markers. One marker was placed at

the top of the thigh leaf spring segment, and the other marker was placed on the exoskeletal "ankle" joint. The markers also represented the attachment points that were used previously during the Instron measurements. Marker position data during hopping were sampled at 120 Hz and were captured simultaneously with the GRF data. We calculated the stance phase average kinematic and kinetic values from 15 consecutive hops for each subject and trial by resampling the motion capture and GRF data so that each stance phase included 100 data points.

Description of the exoskeletons. We attached each exoskeleton to subjects via an adjustable stunt harness (Climbing Sutra, Las Vegas, NV) and standard clipless mountain biking shoes (Shimano MTB SH-MT40, Sakai, Osaka, Japan). Each exoskeleton was a custom-made passive-elastic device that mainly consisted of fiberglass (Gordon Composites, GC-67 UB, Montrose, CO) leaf spring "legs" that acted in parallel to the wearer's legs. Each exoskeletal leg was composed of a shank and thigh leaf spring segment, a 2-degree of freedom "ankle" joint, a fixed "knee" joint, and a 3-degree of freedom "hip" joint (Fig. 1). The "ankle" joint was a pin joint attached to the cleat of the biking shoe distally and to the shank leaf spring proximally. The shank leaf spring was attached to the thigh leaf spring via a rigid "knee" joint that was fixed at an 165° angle. This angle choice was arbitrary but assured us that the exoskeletal legs would always flex forward and would retain predictable stiffness characteristics. Proximally, the thigh segment leaf spring was attached to a 3-degree of freedom "hip" joint that was mounted to the stunt harness. The stunt harness and "hip" joint were attached to and held in place by an adjustable (to accommodate different waist sizes), yet rigid, metal frame. The overall configuration of the exoskeleton allowed the transfer of the weight of the wearer through the exoskeleton to the ground instead of this force being completely borne by the hopper's legs.

Energy return. By combining force and displacement data collected from the Instron machine with the exoskeletal compression data collected during the hopping trials, we estimated the amount of energy return provided by each exoskeleton during each hop by calculating the cumulative integral of the exoskeletal force and displacement during spring recoil, or the area under the force displacement curve. Similarly, to determine the total energy return from the combination of the hopper and exoskeleton, we calculated the cumulative integral of the GRF and exoskeletal displacement during the period of biological and exoskeletal leg recoil. We estimated the force provided by the biological leg by subtracting the force delivered by the exoskeleton from the vertical GRF and then calculated the cumulative integral
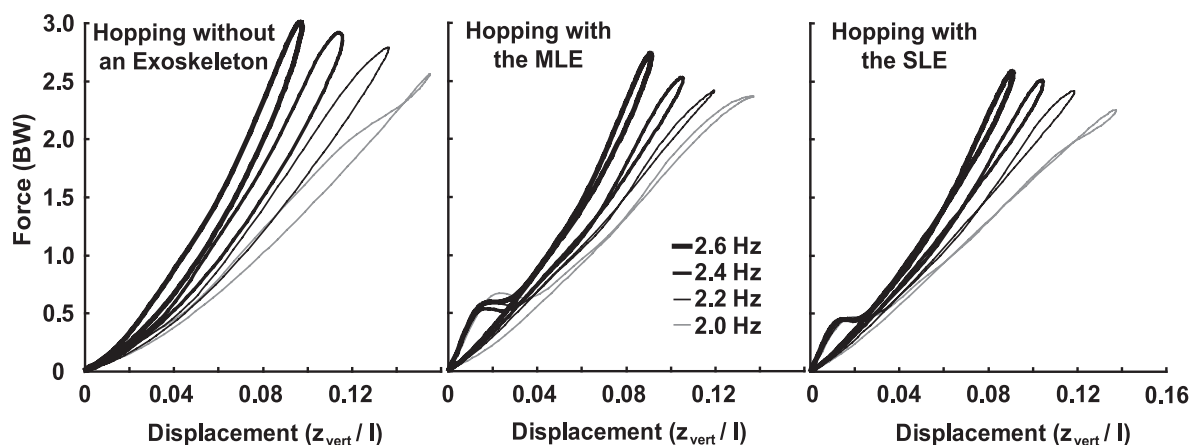
Fig. 4. Dimensionless ground reaction force (GRF) versus displacement of the center of mass at each hopping frequency with and without wearing an exoskeleton. Dimensionless force is achieved by dividing GRF by body weight (BW), and displacement equals center of mass displacement (z_{vert}) normalized by leg length (l). As subjects increased hopping frequency, they increased dimensionless stiffness (the ratio of peak force and displacement). The initial spring stiffness of the exoskeletons appeared to influence force and displacement during initial displacements of 0–0.015, thus displaying a steeper initial slope. All traces represent the average of 15 consecutive hops from all subjects.

of biological leg force and exoskeletal displacement to estimate the energy return of the biological leg.

Metabolic rate. We measured rates of oxygen consumption and carbon dioxide production during each trial using a portable metabolic analysis system (Cosmed K4b^2, IT). Before each session, we calibrated the gas analyzers using reference gases and the flow-rate transducer using a 3-liter syringe. Subjects completed all three sessions at the same time of day to reduce day-to-day variability in metabolic rates.

We confirmed that each subject reached steady-state metabolic rates after ~2 min of hopping (Fig. 3*A*) by calculating the cumulative integrals for rates of oxygen consumption and carbon dioxide production (Fig. 3*B*) and the linear regression equations for the slopes of the integrals with respect to time. For all subjects and all hopping trials, the slopes of the integrals during *minutes 2.5–4.5* had R^2 values of >0.994. We also visually checked the metabolic data to ensure that each subject's rates of metabolism reached a plateau within 2 min and remained consistent between *minutes 2.5* and *4.5*. Thus, to determine the metabolic cost of hopping, we averaged the rates of oxygen consumption and carbon dioxide production during *minutes 2.5–4.5* of each trial. We then used these average rates to calculate metabolic power (in Watts) using a standard equation (5). We did not normalize metabolic power to BW because we sought to determine how each exoskeleton affected metabolic cost irrespective of each exoskeleton's added weight. We calculated net metabolic power for each hopping trial by subtracting standing metabolic power from gross metabolic power. During all trials and for all subjects, respiratory exchange ratios (RERs) were <1.1, indicating that metabolic energy was primarily supplied by oxidative metabolism. The average RER values for all trials at 2.0, 2.2, 2.4, and 2.6 Hz were 0.91, 0.86, 0.85, and 0.85, respectively.

Statistics. We calculated kinematic and kinetic variables from an average of 15 consecutive hops per subject per condition. We used repeated-measures ANOVAs with Tukey's HSD followup tests when warranted ($P < 0.05$) to compare hopping frequency, k_{dim}, t_c, hopping height, distance between consecutive hops (d_{hops}), total energy return, exoskeleton energy return, leg energy return, and net metabolic power (NMP) at each hopping frequency with and without wearing an exoskeleton (JMPin4, SAS Institute, Cary, NC). To ensure that the integrals of metabolic rates were linear during *minutes 2.5–4.5* and that stiffness was linear, we calculated linear regression equations and correlations (R^2). We used repeated-measures ANOVAs with Tukey's honestly significant difference followup tests when warranted ($P < 0.05$) to compare the R^2 values from the linear regression equations of the total, exoskeletal, and biological leg force versus displacement curves. To establish the hopping frequency with the minimum metabolic cost, we determined curvilinear equations and correlations for each condition using standard statistical software (JMPin4).

RESULTS

Hoppers adjusted their biological leg stiffness while hopping in each exoskeleton so that the combination of the hopper and exoskeleton behaved as a linear spring with a total stiffness similar to that of normal hopping at the same frequency (Figs. 4 and 5 and Table 2). While wearing both the SLE and MLE, hoppers had to make moderate adjustments in their biological leg stiffness to maintain overall linear spring-like behavior. We compared the R^2 values of the linear regression equations for each subject's total, exoskeletal, and biological leg force versus displacement curves (stiffness) at each hopping frequency while hopping in the MLE and SLE. We found that at all hopping frequencies, the mean R^2 values for total stiffness while hopping in the both exoskeletons and for exoskeletal stiffness while hopping in the MLE were 0.97–0.99, implying that stiffness was near linear. While hopping with the SLE at all frequencies,
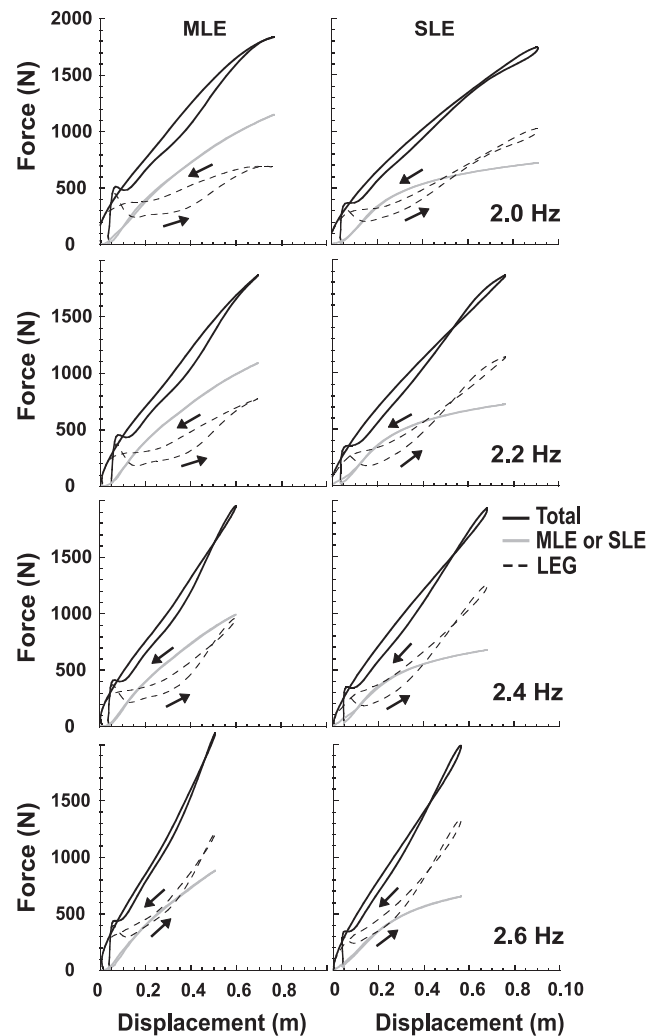

Fig. 5. Total, exoskeleton (MLE or SLE), and biological leg force versus displacement while subjects hopped at 2.0–2.6 Hz wearing a MLE (*left*) and SLE (*right*). Total force equals the measured vertical GRF. Exoskeletal force was estimated from exoskeletal leg compression during hopping and force data collected from the Instron materials testing machine. We estimated biological leg force by subtracting exoskeletal force from GRF. Vertical displacement equals exoskeletal leg compression during hopping. Hoppers adjusted leg stiffness while hopping in an exoskeleton so that the combination of their leg and the parallel spring of the exoskeletal leg behaved as a linear stiffness spring. Lines represent the average of 15 hops from all subjects. Arrows show the direction of biological leg force and displacement.

the R^2 values for both exoskeletal and biological leg stiffness were significantly lower ($P < 0.004$) than the R^2 values for k_t, implying that both the exoskeletal and biological leg had a more nonlinear spring stiffness.

There were no significant differences in hopping frequency, t_c, or hopping height between hopping with or without wearing an exoskeleton (Table 2); thus, our efforts to encourage subjects to maintain amplitude (asking subjects to jump with an aerial phase) and frequency (matching the beat of a metronome) when hopping appeared to be successful. Not surprisingly, total k_{dim} was not significantly different while wearing the MLE than while hopping without an exoskeleton at all frequencies (Figs. 4 and 5 and Table 2). While wearing the SLE, total k_{dim} was not significantly different from normal hopping at all but one frequency. At 2.2 Hz, k_{dim} was signif-

Table 2. k_{dim} and kinematics of hopping with and without a leg exoskeleton

Condition	Hopping Frequency, Hz	k_{dim}	t_c, s	h_{vert}, cm	d_{hops}, cm
Normal	2.01±0.01	16.80±10.60	0.379±00.027	1.14±00.61	4.32±20.02
MLE	2.02±0.01	17.87±20.76	0.358±00.041	1.78±10.40	6.40±30.35†
SLE	2.01±0.01	16.40±10.48*	0.383±00.024	1.43±00.60	3.60±20.01*
Normal	2.20±0.02	20.70±10.28	0.340±00.023	0.98±00.37	4.07±20.57
MLE	2.21±0.02	20.51±10.90	0.339±00.025	1.36±00.67	5.64±30.02†
SLE	2.21±0.02	20.48±20.36†	0.346±00.024	1.12±00.38	3.85±10.87*
Normal	2.41±0.01	25.58±10.78	0.314±00.019	0.60±00.25	4.68±20.20
MLE	2.41±0.02	24.50±10.87	0.321±00.015	0.81±00.39	4.12±20.12
SLE	2.42±0.01	24.55±10.78	0.321±00.018	0.82±00.41	3.53±10.80
Normal	2.61±0.02	31.07±10.87	0.286±00.026	0.57±00.50	4.37±20.38
MLE	2.61±0.01	30.33±20.75	0.289±00.017	0.80±00.37	4.23±20.19
SLE	2.61±0.02	29.13±20.52*	0.300±00.014*	0.63±00.27	2.92±10.69*

Values are means ± SD. Shown are hopping frequency, dimensionless stiffness (k_{dim}), contact time (t_c), hopping height (h_{vert}), and mean distance between consecutive hops (d_{hops}) averaged from 15 hops/subject with and without (normal) wearing a leg exoskeleton (MLE and SLE). k_{dim} equals the ratio of peak ground reaction force (GRF) in units of body weight and peak center of mass displacement normalized by leg length. *Significant difference ($P < 0.05$) between hopping with the MLE and SLE; †significant difference ($P < 0.05$) between hopping with and without (normal) an exoskeleton.

icantly lower ($P < 0.05$) while wearing the SLE than while hopping without an exoskeleton, but, curiously, this stiffness difference was not reflected by significant differences in t_c or impact velocity (u; Table 2).

Wearing an exoskeleton while hopping demanded considerably less metabolic power than hopping without wearing the exoskeleton (Fig. 6), despite the added weight of each exoskeleton. While hopping at 2.2 Hz wearing the MLE, subjects demanded 12% less Net metabolic power (NMP) compared with hopping without wearing the exoskeleton ($P < 0.05$). While wearing the SLE, subjects demanded significantly less metabolic power at all hopping frequencies compared with hopping without wearing an exoskeleton ($P < 0.005$). NMP while wearing the SLE was 28%, 28%, 22%, and 19% lower
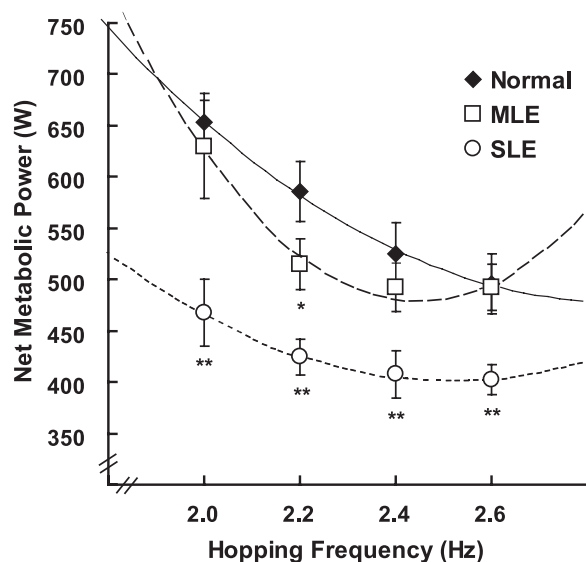

Fig. 6. Average net metabolic power (error bars represent SEs) required to hop at 2.0–2.6 Hz without an exoskeleton (normal), with a MLE, and with a SLE. Wearing a MLE demanded significantly less metabolic power at 2.2 Hz (*$P < 0.05$). Wearing a SLE demanded significantly less metabolic power at all hopping frequencies (**$P < 0.005$) compared with hopping normally and to hopping while wearing the MLE.

than normal hopping at 2.0, 2.2, 2.4, and 2.6 Hz, respectively. Subjects also demanded significantly less metabolic power at all hopping frequencies while wearing the SLE compared with wearing the MLE ($P < 0.005$), which was contrary to our third hypothesis. Thus, in distinction to hopping while wearing an exoskeleton with a spring stiffness that approximated leg stiffness, wearing an exoskeleton with a less stiff and more nonlinear spring incurred a substantially lower metabolic cost.

The hopping frequency that elicited the minimum metabolic cost of hopping while wearing the MLE did not equal the spring-mass model-predicted frequency of 2.0 Hz. We estimated the hopping frequency that required the minimum NMP using a curvilinear equation fitted to the NMP versus hopping frequency data for each condition. While hopping without wearing an exoskeleton, NMP decreased while subjects increased hopping frequency from 2.0 to 2.6 Hz (Fig. 6). These results are described by the following equation: NMP (in Watts) = $236f^2 - 1,353f + 2,416$ ($R^2 = 0.37$), where f is hopping frequency in Hz. From this equation, we calculated that the minimum NMP is required while hopping normally at 2.86 Hz. From the curvilinear equations describing NMP and hopping frequency for the MLE and SLE ($NMP_{MLE} = 720f^2 - 3,531f + 4,808$, $R^2 = 0.27$, and $NMP_{SLE} = 240f^2 - 1,211f + 1,929$, $R^2 = 0.14$), we calculated that the minimum NMP is required while hopping at 2.45 and 2.52 Hz, respectively. Thus, hopping while wearing either exoskeleton shifted the minimum NMP to a lower value and a lower frequency.

DISCUSSION

Hoppers adjusted their leg stiffness while hopping in an exoskeleton so that the combination of their leg and the parallel spring of the exoskeletal leg behaved as a linear spring (Figs. 4 and 5 and Table 2). Our results are similar to results from previous research that showed that hoppers maintain linear spring-mass dynamics when hopping on a wide range of elastic and damped surfaces placed in series with the hopper's legs (17, 20, 31–33). Thus, our findings also support the idea that maintaining linear spring-mass dynamics of the center of mass, by dramatically altering biological leg mechanics in the pres-

ence of series or parallel impedance perturbations to the leg, may be a primary neuromuscular control strategy during bouncing gaits (17, 20, 31, 32). Further research exploring the broader impact of variable stiffness surfaces and exoskeletons is necessary to determine the conditions under which the leg is unable to maintain overall linear spring-mass behavior. Such research may help us to better understand how underlying control strategies are used to achieve bouncing gaits.

Although we tried to ensure that subjects maintained similar mechanics and timing while hopping with or without an exoskeleton and found no significant differences in the tasks and timing of hopping, subjects may have altered hopping frequency, t_c, or hopping height between conditions. The overall effects of these small alterations may have contributed to the significant differences that we found in k_{dim} between hopping normally and with the SLE at 2.2 Hz and between hopping with the SLE and MLE at 2.0 and 2.6 Hz (see Table 2). For example, when subjects hopped while wearing the SLE, they had greater t_c values than while hopping normally or with the MLE. If a hopper maintains hopping frequency, longer t_c values and thus shorter t_a values would result in a lower total spring stiffness.

Our leg exoskeletons appreciably reduced the metabolic cost of hopping, likely by transferring the weight of the wearer through the exoskeleton to the ground surface, storing elastic energy in the fiberglass leaf springs, and then returning this energy to the wearer. We calculated an average energy return of 23.3–50.5 J/hop from the MLE while hopping at 2.0–2.6 Hz, which comprised an average of 52% of the total energy returned per hop. The SLE returned an average of 23.1–45.5 J/hop at 2.0–2.6 Hz, which comprised an average of 44% of the total energy returned per hop (Table 3). Thus, we found that the biological legs had to generate seemingly more energy while subjects hopped wearing the SLE compared with wearing the MLE, yet the SLE provided a much more substantial metabolic benefit. Additionally, there were no differences in the vertical center of mass displacement between exoskeletons (Fig. 4); therefore, neither energy return nor vertical displacement seems to explain the significant metabolic differences resulting from wearing the stiffer MLE compared with the softer SLE.

Although hoppers reduced their metabolic cost while wearing the MLE, they only experienced a significant reduction at

2.2 Hz compared with hopping normally. In contrast, the SLE significantly reduced metabolic cost at all frequencies compared with hopping normally and hopping while wearing the MLE. The differences between the spring stiffness of each exoskeleton may have caused these metabolic differences. We built each MLE to have a spring stiffness based on a hopping frequency of 2.0 Hz (assuming a t_a of 0.14 s). This predetermined spring stiffness may have been too stiff, in that it impaired hoppers' balance and sensory feedback. Before our experimental trials, we had designed a MLE with an exoskeletal spring stiffness based on a frequency faster than 2.0 Hz. But, while wearing this exoskeleton during pilot trials, the hopper's ability to hop was greatly compromised; the hopper was noticeably uncomfortable, was unstable, could not adequately match the beat of the metronome, was not able to hop consistently for >10 s, and could not stay on the force plate while hopping in place. Thus, our final design of the MLE had a lower spring stiffness calculated from a 2.0-Hz hopping frequency, but balance and/or sensory feedback may still have been compromised during our experimental trials, resulting in a higher metabolic cost when subjects used the MLE compared with the less stiff SLE.

As a proxy measure of a subject's balance, comfort, and ability to effectively hop in place, we calculated the location of the center of pressure from the force platform data at touchdown and then calculated the mean distance between consecutive hops (d_{hops}) for normal hopping, hopping while wearing the MLE, and hopping while wearing the SLE (Table 2). We found that d_{hops} was significantly lower when subjects wore the SLE compared with the MLE during hopping at all but the 2.4-Hz hopping frequency. Additionally, d_{hops} was significantly greater while subjects hopped at 2.0 and 2.2 Hz while wearing the MLE compared with normal hopping. This biomechanical effect, likely caused by the MLE's spring stiffness and counteracted by increased muscle activity, may help to explain the greater metabolic cost of hopping while wearing this exoskeleton compared with wearing the SLE. The effects of balance and sensory feedback on the metabolic cost of hopping deserve further attention and may be important considerations for future research in exoskeletal design and performance. Future studies are also needed to determine how different stiffness characteristics of a parallel exoskeleton af-

Table 3. *Energy return while subjects hopped with a leg exoskeleton*

Condition	Hopping Frequency, Hz	E_t, J	E_{exo}, J	E_{leg}, J
MLE	2.02±0.01	89.83±210.12	50.45±100.79	39.38±160.45
SLE	2.01±0.01	94.80±220.79	45.45±120.32	49.35±130.38
MLE	2.21±0.02	74.07±140.04	42.52±90.99	31.55±100.54
SLE	2.21±0.02	81.30±160.51*	37.17±120.30	44.14±90.63*
MLE	2.41±0.02	61.66±90.48	32.33±60.40	29.33±80.49
SLE	2.42±0.01	69.44±130.18*	30.32±90.83	39.12±80.32*
MLE	2.61±0.01	54.05±80.59	23.82±60.06	30.23±80.81
SLE	2.61±0.02	58.66±90.50*	23.07±80.15	35.59±60.54*

Values are means ± SD. Shown are hopping frequency, total energy return (E_t), exoskeletal energy return (E_{exo}), and biological leg energy return (E_{leg}) while subjects hopped while wearing a MLE and SLE. Energy return was calculated from the integral of force and displacement, where displacement was based on exoskeletal compression. *Significant difference ($P < 0.05$) between hopping with the MLE and SLE.

fect spring-mass mechanics, muscle activity, joint stiffness, and metabolic cost during bouncing gaits.

The lighter weight of the SLE (62.6 N) compared with the MLE (67.5 N) may have resulted in a lower metabolic cost of hopping. However, the weight difference between exoskeletons was only 4.9 N, on average, and therefore was not likely to have a substantial effect on our results. A previous study (35) has shown that during running, the addition of 10% BW, the approximate weight of each of our exoskeletons, results in a 14% increase in metabolic rate. However, because the weight of each exoskeleton was self-supported during ground contact, this added weight might have had a smaller overall effect on metabolic cost than a weight fully supported by the biological legs. In fact, even with the added weight of the SLE, hoppers experienced substantial reductions in metabolic cost. Future studies are warranted to determine if and how exoskeletal weight affects metabolic cost during hopping and running.

We assumed that the leg exoskeletons had a negligible amount of hysteresis due to dynamic tests performed on the Instron machine, the material properties of the fiberglass leaf springs, the rigid joints of the exoskeletal legs, and the attachment to the wearer. Although we firmly fixed the harness and shoes onto each subject, it is probable that a small amount of movement at the interfaces between the subject and each exoskeleton was not accounted for in our calculations. Due to our assumptions and potential interface movement, we may have overestimated the amount of mechanical energy transfer from the exoskeletons to the hopper. It is likely that the interface movement was small yet consistent between exoskeletons, and, thus, an overestimate of energy transfer should not affect our overall conclusions. In future designs of the exoskeleton, we plan to minimize any movement occurring at the interfaces by improving the fit of the harness, increasing the surface area of the harness, and improving the shoe's rigid connection to the "ankle" joint. These improvements should facilitate better transfer of mechanical energy between the exoskeleton and the wearer and will improve the accuracy of our predictions regarding mechanical energy transfer.

Based on the curvilinear equations that describe our results, we estimated that hoppers minimized metabolic cost at hopping frequencies of 2.45 and 2.52 Hz while wearing the MLE and SLE, respectively, which are frequencies included within the range of our data. However, while hopping without wearing an exoskeleton, we estimated that the metabolic cost of hopping was minimized at 2.86 Hz, an extrapolation beyond our experimental frequency range. Previous studies have shown that runners likely minimize metabolic cost at preferred step frequencies (12, 25), which range from 2.6 to 2.9 Hz for speeds of 1.5–4.6 m/s (10, 11, 25). Thus, 2.86 Hz seems to be a reasonable estimate of the hopping frequency that minimizes metabolic cost, but additional research is necessary to confirm this value and establish the metabolic cost of hopping over a wider range of frequencies.

We custom built each MLE to represent a spring-mass system based on a 2.0-Hz hopping frequency and 0.14-s aerial phase and assumed the exoskeleton would do all of the rebound work so that metabolic cost could be minimized at this frequency. However, the biological leg also contributed to rebound work while subjects hopped while wearing the MLE. The work provided by the biological leg likely helped hoppers to maintain balance and sensory feedback, as discussed previ-

ously. Therefore, our assumption that the exoskeleton would do all of the rebound work was unfounded, and the hopping frequency that required the minimum metabolic power, 2.45 Hz, reflects the combined effects of the exoskeleton and biological leg.

Previous studies have shown that leg stiffness is primarily adjusted by changes in ankle stiffness during human hopping (17, 18) and that the triceps surae muscles play a major role in leg stiffness regulation during hopping (19, 24). Similar to the results of hopping with an elastic ankle-foot orthosis (19), our leg exoskelton likely reduces biological leg stiffness by reducing ankle stiffness via the muscular activity of the triceps surae muscles. We did not measure leg muscle activity while subject's hopped in our exoskeleton but plan to do so in future work. Elucidating leg muscle activity changes while hopping in an exoskeleton may provide further insight into the functional role of specific muscles and may allow a better understanding of the mechanistic determinants of metabolic cost during bouncing gaits. We anticipate that hoppers experienced significant cocontraction to maintain balance while hopping in the MLE; thus, measures of leg muscle activity would also support or refute this speculation.

In addition to improving hopping performance by decreasing metabolic demand, parallel spring exoskeletons have the potential to vastly change the biomechanics and metabolic costs of running. Based on the results and conclusions of this study, we plan to design a leg exoskeleton that facilitates running and hypothesize that this exoskeleton will significantly improve metabolic running economy. In conclusion, we found that our leg exoskeletons can substantially decrease the metabolic demands of human hopping by effectively and efficiently transferring the weight of the wearer through an exoskeleton to the ground instead of this force being completely borne by the hopper's legs.

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