LEG STIFFNESS AND THE METABOLIC COST OF HOPPING WITH DIFFERENT EXOSKELETON SPRING STIFFNESS PROFILES IN PARALLEL TO THE LEGS

By

STEPHEN ALLEN

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This thesis entitled: Leg Stiffness and the Metabolic Cost of Hopping With Different Exoskeleton Spring Stiffness Profiles in Parallel to the Legs written by Stephen Allen has been approved for the Department of Integrative Physiology

Alena Grabowski, Ph.D.

Rodger Kram, Ph.D.

Kota Takahashi, Ph.D.

Date_____

The final copy of this thesis has been examined by the signatories, and we find that both the content and the form meet acceptable presentation standards of scholarly work in the above mentioned discipline.

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Abstract

Allen, Stephen (M.S., Integrative Physiology)

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Leg Stiffness and the Metabolic Cost of Hopping With Different Exoskeleton Spring Stiffness Profiles in
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Parallel to the Legs

Thesis directed by Assistant Professor Alena Grabowski

A previous study found that when humans hop on both legs with exoskeletal springs in parallel with the legs, net metabolic power decreases compared to normal hopping. Further, they retained near constant overall vertical stiffness. Here, I quantified the biomechanics and metabolic costs of 10 subjects (3F) who hopped on both legs normally and using a passive-elastic exoskeleton with three different spring stiffness profiles in parallel to the legs at 2.4-3.0 Hz. The springs had degressive (DG - stiff then compliant), linear (LN), or progressive (PG - compliant then stiff)stiffness. Compared to normal hopping (NH) at 2.4 - 3.0 Hz, use of the exoskeleton with DG stiffness reduced net metabolic power (P_{met}) by 13-24%, LN stiffness reduced P_{met} by 4-12%, and PG stiffness increased P_{met} by 0-8%. P_{met} was significantly reduced when using the exoskeleton with DG stiffness compared to NH at 2.4-2.6 Hz ($p \le 0.0135$). Dimensionless vertical stiffness remained invariant while hopping with an exoskeleton compared to NH, except when using the exoskeleton with DG and LN spring stiffness at 2.8 Hz (p<0.005). Peak vertical ground reaction force was 9-24% lower ($p \le 0.0008$) and center of mass displacement was 6-12% lower ($p \le 0.0013$) at 2.4-3.0 Hz when using the exoskeleton with DG stiffness compared to NH. Hopping with an exoskeleton with DG stiffness provided the greatest elastic energy return (EE), followed by LNand PG (p<0.001). Future designs of passive-elastic exoskeletons used for bouncing gaits should consider using DG or LN stiffness profiles rather than PG stiffness to minimize metabolic costs.

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Chapter I:

Introduction

During bouncing gaits such as hopping and running, the mechanics of the body have been well-described and quantified by a spring-mass model (Blickhan, 1989; Farley & Gonzalez, 1996; Farley, Houdijk, Van Strien, & Louie, 1998; Farley & Morgenroth, 1999; Ferris, Bohra, Lukos, & Kinnaird, 2006; Ferris & Farley, 1997; McMahon & Cheng, 1990). In this model, the center of mass is characterized as a point mass and the legs as massless, linear springs. These springs represent the combined action of the muscles, tendons and ligaments of the leg. During the first half of ground contact, the kinetic and gravitational potential energy of the center of mass act to compress the leg spring, and this energy is stored as elastic energy (Figure 1A). During the second half of ground contact, a large portion of this elastic energy is returned and converted back into kinetic and potential energy to move the center of mass upward into the aerial phase. The recoil of the tendons within the leg spring returns ~93% of the energy used to stretch them, but muscles must generate force to allow energy storage within the tendons and to support body weight (Alexander, 2002; Ker, 1981).

When hopping or running on surfaces with different stiffnesses, humans maintain and control their center of mass dynamics by adjusting their leg stiffness, such that the combined stiffness of the leg and the in-series surface are invariant across a wide range of surface stiffnesses (Farley et al., 1998; Ferris & Farley, 1997; Ferris, Louie, & Farley, 1998; Kerdok, Biewener, McMahon, Weyand, & Herr, 2002). Stiffness refers to the ability of an object to resist displacement and equals the peak force divided by the change in displacement. On a compliant surface, leg stiffness is adjusted inversely with surface stiffness, so that the leg becomes more compliant on stiff surfaces and more stiff on compliant surfaces (Ferris & Farley, 1997; Kerdok et al., 2002). When hopping on surfaces with different stiffness, leg stiffness increases by almost 3-fold (17.8-

53.3 kN/m) over a 1000-fold decrease in surface stiffness (35,000-27 kN/m) (Ferris & Farley, 1997). The stiffness of the leg is adjusted through changes in muscle force, which incurs a metabolic cost, such that over a 12.5-fold decrease in surface stiffness, runners increase leg stiffness by 29% and decrease metabolic cost by 12% (Kerdok et al., 2002). Further, decreases in surface stiffness appear to improve 1-mile performance by 2-3% compared to a hard surface (McMahon & Greene, 1979).



Figure 1: A) Spring-mass model of hopping with a spring in series (surface spring) with the leg spring, where total displacement (ΔY_{tot}) equals the sum of the surface and center of mass displacement (ΔL). B) Spring-mass model of hopping with a spring in parallel (exoskeleton spring) with the leg spring, where ΔL equals the sum of the leg and exoskeleton displacement. The panels depict the first half of the ground contact phase from initial contact to maximum center of mass displacement from left to right. Body mass is represented as a point-mass, and the leg as a massless linear spring. Below each panel is the equation describing total stiffness (k_{tot}) as a function of leg stiffness (k_{leg}) and surface stiffness (k_{surf}) or exoskeleton stiffness (k_{exo}).

More recently, studies have examined the effects of manipulating spring stiffness in parallel to the leg (Figure 1B). In-parallel stiffness can be adjusted by using a springy full-leg exoskeleton that is worn by an individual. Passive-elastic exoskeletons that have springs placed in parallel to the ankle or the leg have reduced the metabolic cost of human hopping (Farris, Hicks, Delp, & Sawicki, 2014; Farris, Robertson, & Sawicki, 2013; Farris & Sawicki, 2012; Grabowski & Herr, 2009; Robertson, Farris, & Sawicki, 2014). Such devices store elastic energy within the springs when they are compressed by the user's body weight during the first half of ground contact, and then return a large portion of this elastic energy during the second half of ground contact. Although the stiffness of the exoskeleton in parallel with the leg proportionally reduces leg stiffness when exoskeleton stiffness is lower than leg stiffness, humans retain the overall stiffness of their leg plus the exoskeleton. As such, the force required from the leg muscles is reduced during hopping with an in-parallel springy exoskeleton that is less stiff than the leg.

Previously, Grabowski and Herr (2009) designed a passive-elastic, in-parallel, full-leg exoskeleton that when used for hopping, reduced net metabolic power by an average of 24% at 2.0-2.6 Hz compared to normal hopping. They used two configurations of fiberglass springs within the exoskeleton: a single leaf spring with a non-linear stiffness and a multiple leaf spring that was stiffer than the single leaf and also had a non-linear stiffness. Both exoskeletons had a degressive (DG) spring rate, whereby the exoskeletal spring was stiff upon initial compression and then less stiff as it was further compressed. Comparatively, a linear spring (LN) has a spring rate that is constant throughout compression. Finally, a progressive spring (PG) has a spring rate that, like tendon, is compliant during initial compression and then becomes increasingly stiff (Figure 2A). The area under the force-displacement curve represents the total elastic energy that can be stored and returned by each spring type. For a given displacement, DG can store the most elastic energy,

followed by *LN* and *PG*. Grabowski and Herr (2009) tested two *DG* spring rates, one compliant and the other stiff. They found that the compliant spring returned 44% of the total energy required, while the stiff spring returned 52% of the total energy required to hop at 2.0-2.6 Hz. Despite the capacity to store and return a greater amount of elastic energy, the net metabolic power for hopping with the stiffer spring was only reduced at 2.2 Hz by 12%, whereas the net metabolic power for hopping with the more compliant spring was reduced by 19-28% compared to normal hopping across all frequencies. This suggests that simply hopping with a spring capable of storing and returning large amounts of elastic energy does not guarantee large reductions in net metabolic cost. Because the leg has been modelled as a linear spring during bouncing gaits (Blickhan, 1989; Farley, Blickhan, Saito, & Taylor, 1991; Ferris & Farley, 1997; Kerdok et al., 2002; McMahon & Cheng, 1990), it is possible that an in parallel spring stiffness profile other than a *DG* spring stiffness may elicit different overall stiffness and/or further reductions in metabolic cost.



Figure 2: A) Representative degressive (DG), linear (LN), and progressive (PG) stiffness profiles for each of the springs within the exoskeleton. B) The DG stiffness profile was implemented using a fiberglass leaf spring. Exoskeleton leg length was selected based on biological leg length, and the thickness of the fiberglass was adjusted to achieve the target stiffness, which depended on each subject's mass. The LN stiffness profile was created using linear coil springs with even spacing between the coils, which results in a constant stiffness rate. The PG stiffness profile was implemented using coil springs with variable spacing between the coils. As the PG spring is compressed, compliant coils with the smallest space between them "close up" first, followed by stiffer, wider-spaced coils. The end result is stiffness that increases as the spring is compressed.

The goal of the present study was to determine how humans adjust their leg stiffness to accommodate an exoskeleton with different stiffness springs in-parallel to the legs, and the exoskeleton spring stiffness profile that is metabolically optimal (lowest) for hopping, a proxy for running. I determined the biomechanics and metabolic costs of hopping in-place on two feet with an exoskeleton that had DG, LN, and PG springs in parallel to the legs compared to normal hopping. I hypothesized that (1) net metabolic power will be lower when using an exoskeleton with each set of springs compared to normal hopping; (2) for a given displacement, the elastic energy that can be stored and returned differs between the spring stiffness profiles, therefore, net metabolic power will be lowest when using an exoskeleton with DG springs, followed by LN and PG springs, respectively; (3) dimensionless vertical stiffness (K) will be invariant when using an exoskeleton with each set of springs compared to normal hopping.

Methods and Materials

Exoskeleton

To create DG stiffness springs, I custom-made fiberglass leaf-springs (Gordon Composites, GC-67 UB, Montrose, CO), similar to previous studies (Grabowski & Herr, 2009). The "thigh" and "shank" portions of the DG leaf-spring were joined at a rigid "knee" joint to create the exoskeleton legs. The knee joint was fixed at a 165° angle to ensure that the springs always bowed forward relative to the subject. I used evenly coiled compression springs (Century Spring, Commerce, CA) to achieve a LN stiffness. The PG stiffness was achieved with custom-made variable pitch compression springs (Rockford Spring Company, Rockford, IL), where the space (or pitch) between coils varied. LN and PG springs were placed between plates mounted to two

telescoping rods to create exoskeleton "legs" that could be placed in parallel with the biological legs (Figure 2B).

Each exoskeleton leg was attached to an adjustable aluminum frame that encompassed the subject's waist. The aluminum frame was attached to each subject via a padded waist-belt and thigh straps. The exoskeleton legs were connected to the aluminum frame through a 3-degree of freedom "hip" joint and to the shoes via a 1-degree of freedom "ankle" joint. The ankle joint was mounted on a small plate secured to the bottom of clipless mountain biking shoes (MTB/Spin Cycling Shoe, Gavind/Vilano, Elkton, FL, USA; Trace MTB shoe, Diamondback, Kent, WA, USA). On average, the mass of the exoskeleton with the *DG* springs was ~5.0 kg, with *LN* springs was ~6.0 kg, and with *PG* springs was ~5.3 kg. The exoskeleton mass does not include shoe mass because subjects wore the same appropriately sized shoes in every condition.

I measured the stiffness of each exoskeleton leg using a materials testing machine (Instron Series 5859, Norwood, MA, USA). I chose to characterize the exoskeleton leg stiffness with a 10 cm maximum compression, as this approximates center-of-mass displacement during hopping at a preferred frequency (Farley et al., 1998). I determined each exoskeleton leg stiffness from the maximum measured force divided by 10 cm of displacement. All exoskeleton legs were assembled and tested with three cycles of compression and extension. Exoskeleton legs were compressed at a rate of 40 cm/min. I performed the compression tests on one exoskeleton leg, and then multiplied by 2 to obtain the stiffness of both exoskeleton legs (k_{exo}).

Similar to a previous study (Grabowski & Herr, 2009), I approximated k_{exo} a priori based on each subject's mass; therefore, I performed a pilot study to determine the method of estimating k_{exo} . Grabowski and Herr (2009) observed that an exoskeleton stiffness greater than that of the leg compromised balance and control for subjects while hopping. Therefore, I ensured that k_{exo} was less stiff than the biological leg stiffness. In addition, previous studies have shown that leg stiffness increases with hopping frequency (Farley et al., 1991). Therefore, I calculated k_{exo} based on hopping at 2.4 Hz. I asked one pilot subject (59.6 kg) to hop at 2.4 Hz using the exoskeleton with a range of different *DG* stiffness (4.6, 5.8, 7.8, 8.8 kN/m). I chose 2.4 Hz because Grabowski and Herr (2009) estimated that net metabolic power was minimized at ~2.5 Hz while hopping with an exoskeleton that had *DG* springs in-parallel to the legs. I used an exoskeleton with *DG* springs to maintain similar fitting methods to the previous study that used a similar exoskeleton (Grabowski & Herr, 2009). The stiffness values were selected based on the subject's leg length (*l*), measured as the distance between the greater trochanter and the floor, and the thickness of the fiberglass.

For the pilot study and the experimental study, all standing and hopping trials were 5 minutes long with 5 minutes rest between trials. I measured the rates of oxygen consumption and carbon dioxide production via indirect calorimetry (TrueOne 2400, ParvoMedics, Sandy, UT) throughout each trial. Average steady-state metabolic power was determined from the average metabolic rates during the last 2 minutes of each 5 minute trial, and calculated using a standard equation (Brockway, 1987). During each trial, I measured ground reaction forces (GRFs; Bertee, Columbus, OH) for 15 seconds at 1000 Hz twice during minutes 3-5 to ensure that the subject maintained the target frequency. GRFs were filtered with a fourth-order, zero-lag Butterworth filter with a 30 Hz cut-off using a custom MATLAB script (Mathworks, Natick, MA). I used a 10 N threshold to detect ground contact events. The average of 15 consecutive hops was used for analysis.

A spring-mass model compresses and extends at its natural frequency with minimal energy input. Therefore, I calculated the natural frequency (ω_o) (Equation 1) from the exoskeleton with a

DG spring stiffness (k_{exo}) that resulted in the lowest metabolic cost (7.8 kN/m) for the pilot subject during hopping:

$$\omega_o = \sqrt{\frac{k_{exo}}{m}}$$
(Equation 1)

where ω_o is in rad/s, *m* is the subject's mass, and k_{exo} is the *DG* exoskeleton stiffness at 10 cm of displacement. I assumed that the calculated ω_o (11.54 rad/s) could predict the optimal exoskeleton stiffness for all other subjects based on their body mass, and verified this by testing a second pilot subject (69.4 kg).

Subjects

10 healthy subjects (3F, 8M; Table 1), who were physically active at least 3 times per week, provided written informed consent to participate in the study according to the University of Colorado Institutional Review Board. Subjects reported no cardiovascular, neurological or musculoskeletal impairments.

Subject	Age (years)	Ht. (m)	Mass (kg)	<i>l</i> (m)	k _{exo} (kN/m)	<i>k_{DG}</i> (kN/m)	<i>k_{LN}</i> (kN/m)	<i>k_{PG}</i> (kN/m)
1	23	1.61	68.5	0.84	9.2	9.0	9.4	9.0
2	27	1.75	67.1	0.89	9.0	8.8	8.4	9.0
3	24	1.73	71.4	0.91	9.6	8.8	9.4	10.0
4	21	1.62	60.8	0.87	8.0	7.8	8.4	8.0
5	26	1.71	70.1	0.88	9.4	8.8	9.4	9.0
6	31	1.77	76.3	0.91	10.2	10.6	10.6	10.0
7	28	1.84	76.0	0.96	10.2	11.0	10.6	10.0
8	33	1.68	56.8	0.88	7.6	7.8	7.6	8.0
9	19	1.75	69.3	0.91	9.2	8.0	9.4	9.0
10	25	1.81	74.9	0.92	10.0	10.6	10.6	10.0
Avg.	25.7	1.73	69.1	0.89	9.2	9.1	9.4	9.2
SD	4.2	0.07	6.4	0.33	0.9	1.2	1.0	0.8

Table 1: Subject age, height (Ht.), body mass, leg length (l), calculated stiffness (k_{exo}) for 2.4 Hz, and actual exoskeleton spring stiffness for degressive (DG), linear (LN), and progressive (PG) springs with 10 cm of displacement (k_{DG} , k_{LN} , k_{PG}).

Protocol

Subjects hopped in place on two feet under four conditions: normal hopping (*NH*), and hopping with an exoskeleton that had degressive (*DG*), linear (*LN*), and progressive (*PG*) springs at four frequencies (2.4, 2.6, 2.8, 3.0 Hz). I chose this range of frequencies because Grabowski and Herr (2009) estimated that net metabolic power for hopping with an exoskeleton with *DG* springs is minimized at 2.5 Hz and for *NH* is minimized at 2.9 Hz. To ensure that a subject could hop stably at each desired frequency, I selected exoskeleton stiffnesses that were less than leg stiffness during hopping (Grabowski & Herr, 2009). During all conditions, I asked subjects to hop to the beat of the metronome and to maintain an aerial phase, but did not enforce hop height. Each subject completed 16 trials over the course of three experimental sessions where condition order, and frequency order within the condition, were randomized. Each experimental session included 4-6 trials, was at the same time of day to minimize day-to-day variability in metabolic cost, and was separated by at least one day to account for fatigue.

At the beginning of each session, subjects stood quietly while I measured their rates of oxygen consumption and carbon dioxide production via indirect calorimetry. Average steady-state metabolic power was determined in the same manner as the pilot study. I assumed subjects primarily utilized aerobic metabolism as indicated by respiratory exchange ratios (RER) less than 1.0 throughout all trials (Ferrannini, 1988). Net metabolic power (P_{met}) was calculated by subtracting standing from gross metabolic power for each condition and normalized to the subject's body mass.

I collected GRFs in the same manner as the pilot test. From the GRFs, I determined average hopping frequency (*Hf*), ground contact time (t_c), and peak vertical GRF (F_{peak}) for 15 consecutive hops. I calculated vertical acceleration, velocity, and center of mass displacement (ΔL) during

ground contact according to Cavagna (Cavagna, 1975). I used ΔL to approximate the compression of the legs and exoskeleton springs. In order to estimate the elastic energy returned (*EE*) from each exoskeleton, I matched ΔL to the corresponding force from individual force-displacement curves attained from the materials testing machine and calculated the integral. On average, the *DG*, *LN*, and *PG* springs had 5, 3, and 8% hysteresis, respectively, which were included in the estimates of *EE*. In order to compare vertical stiffness across subjects of different masses and leg lengths, I calculated dimensionless stiffness (*K*) by normalizing F_{peak} to bodyweight (*BW*), and center of mass displacement (ΔL) to leg length (*l*) (Equation 2):

$$K = \frac{F_{peak}/BW}{\Delta L/l}$$
(Equation 2)

I calculated hop height (*ht*) based on the vertical velocity of the center of mass (Equation 3):

$$ht = \frac{V_{off}^2}{2g}$$
(Equation 3)

where V_{off} is the velocity of the center of mass at toe-off and g is the acceleration due to gravity. Statistical Analysis

I used a one-way, repeated measures ANOVA to test for statistical differences in P_{met} between conditions at each frequency, and performed a Tukey's HSD post-hoc test if applicable. Significance was set as p<0.05. To test for a difference in mechanical variables between *NH* and the exoskeleton conditions, I used paired t-tests with a Bonferroni correction for multiple comparisons (p=0.0167; RStudio, Boston, MA).

Results

The average hopping frequency for all trials was within 2% of the metronome frequency. One outlier subject was removed from analysis at 2.4 Hz because *K* was greater than 3 standard deviations from the mean. All results for 2.4 Hz are reported from 9 subjects, while results from 2.6-3.0 Hz are reported from 10 subjects.

Exoskeletons

I was able to use springs that accurately matched k_{exo} for all subjects. On average, the percent difference of the actual exoskeleton spring stiffness from k_{exo} was only -1.4 ± 6.6% (p=0.80) for *DG* springs, 0.7 ± 4.8% (p=0.70) for *LN* springs, and -0.3 ± 3.0% (p=0.92) for *PG* springs (Table 1).

Metabolic Rates

When subjects hopped using a passive-elastic, full-leg exoskeleton in parallel to the legs, the exoskeletons with DG and LN spring stiffness profiles resulted in numerically lower P_{met} over all hopping frequencies and the exoskeleton with a PG spring stiffness profile resulted in numerically higher P_{met} (Figure 3). At 2.4 Hz, P_{met} for the exoskeleton with DG stiffness was 22% lower than NH (p=0.0006), but with LN and PG stiffness, P_{met} was not significantly different (p=0.15 and p=0.79, respectively). At 2.6 Hz, P_{met} for the exoskeleton with DG stiffness was 14% lower than NH (p=0.011), but with LN and PG stiffness, P_{met} was not significantly different (p=0.73 and p=0.72, respectively). At 2.8 and 3.0 Hz, none of the exoskeleton conditions were statistically different from NH (p=0.079 and p=0.063, respectively).

When comparing P_{met} between exoskeletons, the exoskeleton with DG stiffness consistently had the lowest P_{met} , while the exoskeleton with PG stiffness consistently had the highest P_{met} . However, there were no differences in P_{met} between the exoskeletons with DG and LN stiffness at any hopping frequency (p \geq 0.097). The exoskeleton with DG stiffness resulted in 26%, 18%, and 18% lower P_{met} compared to the exoskeleton with PG stiffness at 2.4, 2.6, and 2.8 Hz (p \leq 0.0008), respectively. The exoskeleton with *LN* stiffness resulted in 15% and 13% lower P_{met} than the exoskeleton with *PG* stiffness at 2.4 (p=0.0224) and 2.8 Hz (p = 0.0152), respectively.



Figure 3: Average \pm SE net metabolic power (P_{met}) normalized to body mass for normal hopping (*NH*, white squares), and with different exoskeleton spring stiffness profiles: progressive springs (*PG*, black circles), linear springs (*LN*, gray diamonds), and degressive springs (*DG*, black triangles) across hopping frequencies. A significant difference is indicated as * compared to *NH*, ^ compared to *LN*, # compared to *PG* and † compared to *DG*. P_{met} results are described for each spring by the following equations, where Hf is hopping frequency: $PG = 231 \text{Hf}^2 - 1376 \text{Hf} + 2492$ ($R^2 = 0.14$); $NH = 468 \text{Hf}^2 - 2620 \text{Hf} + 4086$ ($R^2 = 0.12$); $LN = 115 \text{Hf}^2 - 691 \text{Hf} + 1438$ ($R^2 = 0.04$); $DG = -5.1 \text{Hf}^2 + 34.8 \text{Hf} + 322$ ($R^2 = 0.001$).

Biomechanics

For each hopping frequency, ground contact time (t_c) did not differ across conditions (p \ge 0.047; Table 2). Despite the invariant ground contact time, F_{peak} was 24%, 16%, 12%, and 9% lower for the exoskeleton with DG stiffness compared to NH at 2.4, 2.6, 2.8 and 3.0 Hz (p \le 0.0008), respectively. F_{peak} was 15% and 9% lower for the exoskeleton with LN stiffness compared to NH at 2.4 and 2.8 Hz (p \le 0.009), respectively. Use of the exoskeleton with DG and LN stiffness resulted in less center of mass displacement (ΔL) compared to NH (Table 2). ΔL for the exoskeleton with DG stiffness was 12%, 9%, 7%, and 6% lower than NH for 2.4, 2.6, 2.8, and 3.0 Hz (p \le 0.0013), respectively. ΔL for the exoskeleton with LN stiffness was 5% lower than NH at 2.4 and 2.6 Hz (p \le 0.003). With two exceptions, these adjustments to F_{peak} and ΔL resulted in constant dimensionless vertical stiffness (K) at each frequency when hopping with an exoskeleton with different spring stiffness profiles in parallel to the legs (Figure 4 and 5; Table 2). At 2.8 Hz, K was 5% lower when using the exoskeleton with DG and LN stiffness compared to NH (p \le 0.0048), but all other comparisons were not different (p \ge 0.02; Table 2).

The estimated amount of elastic energy returned (*EE*) by each exoskeleton differed between spring types ($p \le 0.001$; Figure 5; Table 2). The exoskeleton with *DG* stiffness had the greatest elastic energy return, while the exoskeleton with *PG* stiffness had the least elastic energy return. From 2.4-3.0 Hz, the exoskeletons with *DG*, *LN*, and *PG* stiffness accounted for 65-74%, 37-51%, and 12-24%, of the total elastic energy return for hopping, respectively. Hopping with an exoskeleton also decreased the total hysteresis, or energy loss, compared to *NH* at every hopping frequency, except for hopping with an exoskeleton with a DG stiffness profile at 3.0 Hz ($p \le 0.014$; Figure 3; Table 2). From 2.4 – 3.0 Hz, the exoskeletons with *DG*, *LN*, and *PG* stiffness reduced total hysteresis by 44-61%, 28-51%, and 32-53%, respectively, compared to *NH*. Subjects were encouraged to hop with an aerial phase, but hop height (*ht*) was not enforced. With exception of two trials, when hopping with an exoskeleton, *ht* was not different from *NH*. At 2.6 Hz, *ht* for the exoskeleton with *LN* stiffness averaged 0.4 cm greater than *NH* (0.8 cm; p = 0.012). At 2.8 Hz, *ht* for the exoskeleton with the *DG* stiffness averaged 0.3 cm greater than *NH* (0.6 cm; p = 0.007; Table 2).



Figure 4: Normalized force in body weight (BW) as a function of dimensionless displacement (ratio of center of mass displacement [ΔL] to leg length [I]) for normal hopping (NH), and hopping with an exoskeleton with degressive (DG), linear (LN), and progressive (PG) springs. As subjects increased hopping frequency, dimensionless stiffness (K), or the slope of the lines, increased in all conditions. Traces are the average of 15 consecutive hops from all subjects. Hopping with an exoskeleton using each spring stiffness profile reduced overall hysteresis at each hopping frequency.



Figure 5: Average \pm SE A) total dimensionless stiffness (*K*) and B) exoskeleton elastic energy return (*EE*) across hopping frequencies during normal hopping (*NH* - white squares), with an exoskeleton with degressive (*DG* - black triangles), linear (*LN* - grey squares), and progressive (*PG* - black circles) springs. A significant difference is indicated as * compared to *NH*, ^ compared to *LN*, # compared to *PG* and † compared to *DG*.

Table 2: Mean \pm SD contact time (t_c), peak GRF (F_{peak}) in units of body weight (BW - including the weight of the exoskeleton in the respective conditions), center of mass displacement (ΔL), dimensionless stiffness (K), hop height (ht), elastic energy return (EE), and total hysteresis (E_{hyst}) at each hopping frequency during normal hopping (NH) and hopping with an exoskeleton with degressive (DG), linear (LN) and progressive (PG) springs. All values are the average of 15 consecutive hops from each subject. A significant difference is indicated as * compared to NH, ^ compared to PG and † compared to DG.

		NH	DG	LN	PG
	2.4 Hz	0.28 ± 0.04	0.30 ± 0.03	0.30 ± 0.03	0.30 ± 0.03
4 (a)	2.6 Hz	0.27 ± 0.02	0.28 ± 0.02	0.26 ± 0.03	0.27 ± 0.02
$l_c(\mathbf{S})$	2.8 Hz	0.25 ± 0.02	0.25 ± 0.02	0.26 ± 0.02	0.26 ± 0.01
	3.0 Hz	0.23 ± 0.02	0.23 ± 0.02	0.24 ± 0.02	0.24 ± 0.01
	2.4 Hz	3.4 ± 0.6	$2.6 \pm 0.3*$	$2.9 \pm 0.4*$	3.2 ± 0.5
F_{peak}	2.6 Hz	3.3 ± 0.3	$2.8 \pm 0.4*$	3.3 ± 0.5	3.3 ± 0.4
(BW)	2.8 Hz	3.3 ± 0.3	$2.9 \pm 0.4*$	$3.0 \pm 0.3*$	3.2 ± 0.3
	3.0 Hz	3.2 ± 0.2	$2.9 \pm 0.3*$	3.0 ± 0.3	3.1 ± 0.2
	2.4 Hz	10.9 ± 0.6	$9.6 \pm 0.7*$	$10.4 \pm 0.5*$	10.9 ± 0.6
ΔL	2.6 Hz	9.5 ± 0.4	$8.6 \pm 0.6*$	$9.0 \pm 0.5*$	9.4 ± 0.4
(cm)	2.8 Hz	8.2 ± 0.4	$7.6 \pm 0.5*$	7.9 ± 0.4	8.2 ± 0.4
	3.0 Hz	7.0 ± 0.3	$6.6 \pm 0.2*$	7.0 ± 0.3	7.1 ± 0.2
	2.4 Hz	26.70 ± 5.85	23.74 ± 2.48	24.27 ± 3.59	25.57 ± 4.50
V	2.6 Hz	31.25 ± 3.58	28.95 ± 2.76	31.92 ± 5.53	31.52 ± 3.56
K	2.8 Hz	36.02 ± 3.19	$34.33 \pm 3.47*$	$34.04 \pm 3.46*$	35.39 ± 2.85
	3.0 Hz	40.73 ± 3.71	39.62 ± 3.58	39.10 ± 3.94	39.92 ± 2.71
	2.4 Hz	1.3 ± 0.8	1.2 ± 0.6	1.1 ± 0.5	1.1 ± 0.7
ht	2.6 Hz	0.8 ± 0.4	0.9 ± 0.5	$1.2 \pm 0.6*$	0.9 ± 0.4
(cm)	2.8 Hz	0.6 ± 0.4	$0.9 \pm 0.4*$	0.7 ± 0.3	0.8 ± 0.3
	3.0 Hz	0.5 ± 0.4	0.7 ± 0.3	0.7 ± 0.3	0.6 ± 0.2
	2.4 Hz	83 ± 14	59 ± 10#^	$44 \pm 7\#$ †	21 ± 1^†
EE	2.6 Hz	70 ± 9	$53 \pm 10 \#^{10}$	$37 \pm 7\#$ †	$17 \pm 2^{+}$
(Nm)	2.8 Hz	60 ± 8	$43 \pm 8 \#^{-1}$	27 ± 5#†	$10 \pm 2^{+}$ †
	3.0 Hz	52 ± 6	$36 \pm 6 \#^{-1}$	21 ± 4#†	7 ± 1^†
	2.4 Hz	11.90	4.62*	6.89*	7.15*
Ehyst	2.6 Hz	12.76	6.44*	9.15*	8.78*
(Nm)	2.8 Hz	11.18	6.27*	5.45*	5.22*
	3.0 Hz	9.28	6.45	5.05*	5.96*

Discussion

The present study examined the metabolic and biomechanical effects of normal hopping and hopping with a springy exoskeleton in parallel to the legs with three different spring stiffness profiles. I hypothesized that hopping with an exoskeleton would lower P_{met} compared to NH regardless of the spring stiffness profile. The metabolic results do not support this hypothesis because hopping with the exoskeleton with *DG* stiffness resulted in lower P_{met} at slower hopping frequencies compared to *NH*. But, hopping with the exoskeleton with *LN* and *PG* spring stiffness profiles did not change P_{met} compared to *NH*. The metabolic results of the present study at 2.4 and 2.6 Hz are consistent with a previous study that utilized a similar full-leg, passive-elastic exoskeleton with a *DG* spring stiffness profile (Grabowski & Herr, 2009). They found 19-22% reductions in P_{met} when subjects used an exoskeleton with *DG* stiffness compared to *NH* at 2.4-2.6 Hz, whereas I found 14-22% reductions in P_{met} when subjects used an exoskeleton with *DG* stiffness compared to *NH* at 2.4-2.6 Hz.

Hopping with an exoskeleton with *PG* stiffness resulted in no change in P_{met} compared to *NH*. On average, across 2.4-3.0 Hz, P_{met} was approximately 5% numerically higher when using the exoskeleton with *PG* springs compared to *NH*. This response is similar to previous studies showing that hopping with 9% of added body weight elicits a 13% increase in P_{met} compared to *NH* (Allen & Grabowski, 2017), and running with 10% added weight increases P_{met} by 14% compared to normal running (Teunissen, Grabowski, & Kram, 2007). The exoskeleton with *PG* springs is about 7.6% of subjects' body weight, which would elicit an approximate 10% metabolic penalty, assuming a linear relationship between added weight and P_{met} during bouncing gaits. Therefore, the assistance provided from the exoskeleton with *PG* springs. Future designs of full-leg, passive-elastic exoskeletons for bouncing gaits that are aimed at reducing metabolic cost

should not utilize PG stiffness profiles, but should consider lightweight exoskeletons with DG or LN stiffness profiles.

Previous research has shown that the preferred step frequency in running minimizes metabolic cost (Cavagna, Mantovani, Willems, & Musch, 1997), however the metabolic cost of hopping is not minimized at the preferred hopping frequency ($\sim 2.0-2.2$ Hz) (Farley et al., 1991; Melvill-Jones & Watt, 1971). Instead, Grabowski and Herr (2009) estimated the minimum metabolic cost of NH occurs at 2.9 Hz. While I did not specifically seek to determine the hopping frequency that minimizes metabolic cost, I did incorporate a wider range of hopping frequencies, and found that hopping at ~ 2.8 Hz minimized the metabolic cost of NH (Figure 3). Previously, Grabowski and Herr (2009) estimated the minimum metabolic cost of hopping with an exoskeleton with DG springs to be 2.5 Hz. I was unable to estimate the hopping frequency that minimized metabolic cost when using an exoskeleton with DG stiffness because the 2nd order polynomial characterizing P_{met} for hopping frequencies of 2.4-3.0 Hz had a negative coefficient and a weak correlation (Figure 3). However, to estimate the minimum metabolic cost, I combined the averaged data from Grabowski and Herr (2009) with those of the current study, and estimated that with a DG exoskeleton, metabolic cost is minimized when hopping at 2.8 Hz ($DG = 124 Hf^2 - 701 Hf +$ 1369; Hf is hopping frequency). Additionally, I estimated that metabolic cost is minimized at ~3.0 Hz for exoskeletons with LN or PG springs.

The underlying reason that the metabolic cost is minimized at 2.8-3.0 Hz while hopping normally or with a passive-elastic, full-leg exoskeleton still remains to be determined. Farris and Sawicki (2012) found that an elastic, exoskeleton that spanned the ankle minimized the metabolic cost of hopping at 2.6-2.7 Hz. In addition, Farris et al. (2014) suggested the same device caused changes in plantar-flexor muscle force-length or force-velocity curves. However, this ankle-only

exoskeleton required the ankle to be in a more plantar-flexed position compared to normal hopping. Thus, the ankle-only exoskeleton may have decreased the average plantar-flexor muscle operating lengths and may have increased muscle fiber shortening velocity. Hopping with a full-leg exoskeleton may elicit different changes in joint position, and subsequent changes to the muscle force-length and force-velocity relationships. Future research is needed to determine the underlying changes in the muscle force-length and force-velocity relationships when hopping with a full-leg exoskeleton.

I further hypothesized that P_{met} would be lowest when hopping with the exoskeleton with DG springs, followed by with LN and PG springs. The results partially support this second hypothesis because use of the exoskeleton with DG springs substantially reduced P_{met} by 18-26% compared to PG springs at 2.4-2.8 Hz, and use of the exoskeleton with LN springs reduced P_{met} by 13-15% compared to PG springs at 2.4 and 2.8 Hz. However, I found no difference in P_{met} between hopping with an exoskeleton with DG or LN springs at 2.4-3.0 Hz (Figure 3). These results suggest that the spring stiffness profile plays an important role in eliciting metabolic reductions while hopping with springs in parallel to the legs.

When compressed, the biological leg and exoskeletal springs store elastic energy during downward displacement and then return most of this energy to the user during recoil. Overall, hopping with an exoskeleton minimizes the total hysteresis, or energy loss, compared to *NH* (Figure 4; Table 2). The elastic energy return (*EE*) from the exoskeletal springs likely reduce the muscular demands and force generated by the legs, which reduces P_{met} , during hopping. I found large differences in *EE* between exoskeleton spring conditions, and estimated that *EE* from the *DG* and *LN* springs was at least twice as much as the *EE* from the *PG* springs (Figure 5; Table 2). I estimated *EE* to be 53-59, 37-44, and 17-21 J/hop with the *DG*, *LN*, and *PG* springs, respectively.

The estimates of *EE* are much higher than those reported in previous work (Grabowski & Herr, 2009), where an exoskeleton with DG springs in parallel with the legs had an average EE of 23-30 J/hop at 2.4-2.6 Hz. This difference is likely explained by the method of determining spring compression in each study and energy loss due to the non-rigid interface between the exoskeleton and the user (Yandell, Quinlivan, Popov, Walsh, & Zelik, 2017). I used ΔL to calculate the exoskeleton compression during ground contact and assumed a rigid connection between the human/exoskeleton interface, whereas Grabowski and Herr (2009) used reflective markers placed on the "hip" and "ankle" joints of the exoskeleton to determine exoskeleton leg compression. Thus, I likely over-estimated EE. Grabowski and Herr (2009) reported 5.5-6.7 cm of exoskeleton spring compression, approximately 67% of the center of mass displacement. I used 5.5-6.7 cm as an estimate for spring compression, re-analyzed EE for the DG springs at 2.4 and 2.6 Hz, and calculated *EE* to be 37.0 J/hop and 27.6 J/hop, respectively, which is similar to the previously reported values. There is a need to improve the interface between the exoskeleton and the user so that the *EE* can be better utilized and not dissipated (Herr, 2009; Young & Ferris, 2017); such improvements could further reduce the metabolic cost of hopping.

Finally, the results support the hypothesis that *K* is invariant when hopping with different exoskeletal spring stiffness profiles in parallel to the legs and does not differ from *NH*. On average, I found that *K* for each exoskeleton condition was within 1-8% of *NH*. These results are similar to previous research, which has shown that humans maintain overall stiffness when hopping on a wide range of elastic or dampened surfaces (Ferris & Farley, 1997; Kerdok et al., 2002; C. Moritz & Farley, 2005), or with springs placed in parallel to the legs (Grabowski & Herr, 2009) or the ankle joints (Ferris et al., 2006). The present findings contribute to the hypothesis that humans maintain spring-mass dynamics through biological leg stiffness adjustments to springs in series or

in parallel, which may be a primary neuromuscular control strategy for bouncing gaits (Farley et al., 1998; Ferris & Farley, 1997; C. T. Moritz & Farley, 2003). In order to maintain constant stiffness across conditions, biological leg stiffness (k_{leg}) must decrease when hopping with an exoskeleton (Figure 6). The overall central nervous system (CNS) strategy for adjusting k_{leg} likely changes in response to different exoskeleton spring stiffness profiles, especially at the joint level. Future research exploring the relationship between joint stiffness and k_{leg} while hopping with an exoskeleton with different spring stiffness profiles in parallel to the legs may provide further insight into neuromuscular strategies that are implemented during bouncing gaits.



Figure 6: Total (solid black lines), exoskeleton (Exo - solid gray lines) with degressive (DG), linear (LN), and progressive (PG) springs (solid gray lines) and estimated biological leg (Leg - dashed black lines) force versus center of mass displacement. Total force is measured from the vertical GRF, and Exo force is estimated from the force and displacement data collected from the materials testing machine. I estimated biological leg force by subtracting Exo force from total force for each center of mass displacement. Each line represents the average of 15 consecutive hops from each subject.

Previously, Kram and Taylor (Kram & Taylor, 1990) found that the metabolic cost of running is determined by the rate of muscular force production and the magnitude of force required to support the body, which are directly related to $1/t_c$. Presumably, a shorter contact time requires force to be produced from less economical muscle fibers with faster shortening velocities, cross bridge cycling, and rates of ATP consumption (Huxley, 1974). The data from the present study show that hopping with an exoskeleton did not change contact time compared to *NH*, thus the rate of generating force $(1/t_c)$ was unchanged and does not explain differences in metabolic cost. Despite the added mass of the exoskeletons (5-6 kg), F_{peak} was lower when hopping with an exoskeleton with *DG* or *LN* stiffness. This suggests that the muscles produced less force while hopping with an exoskeleton with *DG* or *LN* springs. Indeed, when humans hop with an ankleonly exoskeleton, metabolic cost is reduced and the soleus muscle produces less force compared to *NH* (Farris et al., 2013).

Hopping with an exoskeleton with *DG* or *LN* stiffness reduced F_{peak} and ΔL . These variables were reduced near proportionally so that *K* was maintained when hopping with an exoskeleton compared to *NH* (Table 2). In addition to the *EE* returned by the springs, P_{met} while hopping with an exoskeleton with *DG* or *LN* springs may also be lower due to ΔL and its effects on effective mechanical advantage (EMA). EMA is the ratio of the muscle-tendon moment arm and ground reaction force moment arm about a joint, and provides an estimate of the volume of active muscle that generates force to perform the movement. Studies on running and walking have shown that as EMA decreases, the volume of active muscle required to produce force increases, which ultimately increases metabolic cost (Biewener, Farley, Roberts, & Temaner, 2004; Carrier, Heglund, & Earls, 1994; Kipp, 2017; McMahon, Valiant, & Frederick, 1987). During hopping, 99% of the ground reaction force (GRF) is directed vertically (Veilleux, Rauch, Lemay, & Ballaz,

2012), and the ankle, knee and hip joints initially flex to lower the center of mass; therefore, changes in EMA are likely due to changes in joint angle rather than changes in GRF direction. I found that center of mass displacement was reduced when subjects hopped with an exoskeleton with DG or LN springs as compared to NH (Table 2). This indicates that the joints are not as flexed when hopping with an exoskeleton with DG or LN springs, and EMA was likely greater compared to NH, thus reducing the volume of active muscle and metabolic cost.

The biomechanics of hopping are a proxy for the biomechanics of running. Thus, running with springs in parallel to the legs may elicit similar changes in biomechanics and metabolic cost. Step frequency is approximately 2.8 Hz when running at 3 m/s (Cavagna et al., 1997; Kaneko, Matsumoto, Ito, & Fuchimoto, 1987). Thus, running with an exoskeleton with *DG* stiffness would likely reduce P_{met} more than with *LN* or *PG* stiffness. In addition, use of a passive-elastic, full-leg exoskeleton that reduces the forces borne by the legs during running could decrease the risk of musculoskeletal injury. Finally, because metabolic cost increases with added weight during bouncing gaits, the assistance provided by a passive-elastic, full-leg exoskeleton may allow populations such as military, first-responders, recreational hikers or obese individuals to run with decreased effort.

Limitations

I determined that the stiffness profile influences metabolic cost while hopping with a passive elastic exoskeleton in parallel to the legs. I varied DG spring stiffness by manipulating both the length and thickness of the fiberglass by 2.54 cm and 0.5 cm increments, respectively, and provided subjects with the spring combination that was the closest match to their leg length and the calculated exoskeleton stiffness. These incremental changes in fiberglass thicknesses resulted in an average of 0.9 kN/m increments in DG spring stiffness that could be utilized for each

subject. The study reasonably approximated optimal exoskeleton spring stiffness, but future studies may be needed to test a range of DG springs with finer stiffness resolution than 0.9 kN/m increments. In addition, more refined stiffness profiles could provide insight into how initial DG stiffness and the inflection point within the stiffness profile affect the metabolic cost of hopping, and determine if further reductions in metabolic cost are possible.

I measured the metabolic costs and biomechanics for hopping at 2.4-3.0 Hz. The metabolic and biomechanical effects of using an exoskeleton with different spring stiffness profiles may depend on hopping frequency. For example, hopping with an exoskeleton with *DG* springs might result in a lower metabolic cost at slower hopping frequencies, whereas hopping with an exoskeleton with *LN* springs might result in a lower metabolic cost at faster hopping frequencies. As stated previously, I estimated elastic energy return of the exoskeleton springs based on center of mass displacement rather than spring compression, which likely overestimates elastic energy return. Future studies should examine how the movement of the exoskeleton relative to the person using the device or human-device interface influences elastic energy return and metabolic cost.

Conclusion

I determined the metabolic and biomechanical effects of hopping with a passive-elastic, full-leg exoskeleton with different spring stiffness profiles in parallel to the legs. Hopping while using an exoskeleton with degressive springs reduces metabolic cost at 2.4-2.6 Hz by up to 22% compared to normal hopping. Use of an exoskeleton with degressive or linear spring stiffness profiles numerically reduced the metabolic cost of hopping at 2.4-3.0 Hz. In addition, dimensionless vertical stiffness remains invariant when hopping while using an exoskeleton with three different spring stiffness profiles compared to hopping normally. I found that the spring stiffness profile used in a passive-elastic, full-leg exoskeleton affects the metabolic cost of

hopping; information that can be used in the future design of exoskeletons for bouncing gaits such as running.

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References

- Alexander, R. (2002). Tendon elasticity and muscle function. *Comparative Biochemistry and Physiology. Part A, Molecular & Integrative Physiology, 133*(4), 1001-1011.
- Allen, S., & Grabowski, A. (2017). Hopping with a full-leg exoskeleton lowers metabolic cost and muscle activity [abstract]. *41st American Society of Biomechanics Annual Meeting*.
- Biewener, A. A., Farley, C. T., Roberts, T. J., & Temaner, M. (2004). Muscle mechanical advantage of human walking and running: implications for energy cost. [Research Support, U.S. Gov't, Non-P.H.S. Research Support, U.S. Gov't, P.H.S.]. *Journal of Applied Physiology*, 97(6), 2266-2274. doi: 10.1152/japplphysiol.00003.2004
- Blickhan, R. (1989). The spring-mass model for running and hopping. *Journal of Biomechanics*, 22(11-12), 1217-1227.
- Brockway, J. (1987). Derivation of formulae used to calculate energy expenditure in man. *Human Nutrition: Clinical Nutrition, 41*, 463-471.
- Carrier, D., Heglund, N., & Earls, K. (1994). Variable gearing during locomotion in the human musculoskeletal system. *Science*, *265*(5172), 651-653.
- Cavagna, G. (1975). Force platforms as ergometers. *Journal of Applied Physiology*, 39(1), 174-179.
- Cavagna, G., Mantovani, M., Willems, P., & Musch, G. (1997). The resonant step frequency in human running. *Pflugers Archiv: European journal of physiology*, 434(6), 678-684.
- Farley, C., Blickhan, R., Saito, J., & Taylor, C. (1991). Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits. *Journal of Applied Physiology*, 71(6), 2127-2132.
- Farley, C., & Gonzalez, O. (1996). Leg stiffness and stride frequency in human running. *Journal* of *Biomechanics*, 29(2), 181-186.
- Farley, C., Houdijk, H., Van Strien, C., & Louie, M. (1998). Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. *Journal of Applied Physiology*, 85(3), 1044-1055.
- Farley, C., & Morgenroth, D. (1999). Leg stiffness primarly depends on ankle stiffness during human hopping. *Journal of Biomechanics*, *32*, 267-273.
- Farris, D., Hicks, J., Delp, S., & Sawicki, G. (2014). Musculoskeletal modelling deconstructs the paradoxical effects of elastic ankle exoskeletons on plantar-flexor mechanics and energetics during hopping. [Research Support, N.I.H., Extramural Research Support, Non-U.S. Gov't]. *Journal of Experimental Biology*, 217(Pt 22), 4018-4028. doi: 10.1242/jeb.107656

- Farris, D., Robertson, B., & Sawicki, G. (2013). Elastic ankle exoskeletons reduce soleus muscle force but not work in human hopping. J Appl Physiol, 115, 579-585. doi: 10.1152/japplphysiol.00253.2013.-Inspired
- Farris, D., & Sawicki, G. (2012). Linking the mechanics and energetics of hopping with elastic ankle exoskeletons. J Appl Physiol, 113, 1862-1872. doi: 10.1152/japplphysiol.00802.2012.
- Ferrannini, E. (1988). The Theoretical Bases of Indirect Calorimetry: A Review. *Metabolism*, 37(3), 287-301.
- Ferris, D., Bohra, Z., Lukos, J., & Kinnaird, C. (2006). Neuromechanical adaptation to hopping with an elastic ankle-foot orthosis. [Clinical Trial Research Support, N.I.H., Extramural]. *J Appl Physiol (1985), 100*(1), 163-170. doi: 10.1152/japplphysiol.00821.2005
- Ferris, D., & Farley, C. (1997). Interaction of leg stiffness and surface stiffness during human hopping. *Journal of Applied Physiology*, 82(1), 15-22.
- Ferris, D., Louie, M., & Farley, C. (1998). Running in the real world: adjusting leg stiffness for different surfaces. *Proceedings. Biological sciences Royal Society*, 265(1400), 989-994.
- Grabowski, A., & Herr, H. (2009). Leg exoskeleton reduces the metabolic cost of human hopping. [Research Support, Non-U.S. Gov't]. *Journal of Applied Physiology, 107*(3), 670-678. doi: 10.1152/japplphysiol.91609.2008
- Herr, H. (2009). Exoskeletons and orthoses: classification, design challenges and future directions. [Research Support, Non-U.S. Gov't]. J Neuroeng Rehabil, 6, 21. doi: 10.1186/1743-0003-6-21
- Huxley, A. (1974). Muscular Contraction. Journal of Physiology, 243(1), 1-43.
- Kaneko, K., Matsumoto, M., Ito, A., & Fuchimoto, T. (1987). Optimum stride frequency in human running. In B. Johnson (Ed.), *Biomechanics X-B* (Vol. 6.B, pp. 803-807). Champaign, IL: Human Kinetics.
- Ker, R. (1981). Dynamic tensile properties of the plantaris tendon of sheep (ovis aries). *Journal of Experimental Biology*, 93, 283-302.
- Kerdok, A., Biewener, A., McMahon, T., Weyand, P., & Herr, H. (2002). Energetics and mechanics of human running on surfaces of different stiffnesses. [Research Support, Non-U.S. Gov't]. J Appl Physiol (1985), 92(2), 469-478. doi: 10.1152/japplphysiol.01164.2000
- Kipp, S. (2017). Why Does Metabolic Rate Increase Curvilinearly with Running Velocity?

Kram, R., & Taylor, R. (1990). Energetics of running: a new perspective. Nature, 346, 265-267.

- McMahon, T., & Cheng, G. (1990). The mechanics of running: How does stiffness couple with speed? *Journal of Biomechanics*, 23, 65-78.
- McMahon, T., & Greene, P. (1979). The influence of track compliance on running. *Journal of Biomechanics*, 12, 893-904.
- McMahon, T., Valiant, G., & Frederick, E. (1987). Groucho Running. Journal of Applied Physiology, 62, 2326-2337s.
- Melvill-Jones, G., & Watt, D. (1971). Observations on the control of stepping and hopping movements in man. *Journal of Physiology*.
- Moritz, C., & Farley, C. (2005). Human hopping on very soft elastic surfaces: implications for muscle pre-stretch and elastic energy storage in locomotion. [Comparative Study Research Support, Non-U.S. Gov't Research Support, U.S. Gov't, P.H.S.]. *J Exp Biol*, 208(Pt 5), 939-949. doi: 10.1242/jeb.01472
- Moritz, C. T., & Farley, C. T. (2003). Human hopping on damped surfaces: strategies for adjusting leg mechanics. [Research Support, U.S. Gov't, P.H.S.]. *Proc Biol Sci*, 270(1525), 1741-1746. doi: 10.1098/rspb.2003.2435
- Robertson, B., Farris, D., & Sawicki, G. (2014). More is not always better: modeling the effects of elastic exoskeleton compliance on underlying ankle muscle-tendon dyamics. *Bioinspiration & Biomimetics*, 9. doi: 10.1088/1748-3182/9/4/04601810.1088/1748-3190/10/1/019601
- Teunissen, L., Grabowski, A., & Kram, R. (2007). Effects of independently altering body weight and body mass on the metabolic cost of running. [Clinical Trial]. *J Exp Biol, 210*(Pt 24), 4418-4427. doi: 10.1242/jeb.004481
- Veilleux, L.-N., Rauch, F., Lemay, M., & Ballaz, L. (2012). Agreement between vertical ground reaction force and ground reaction force vector in give common clinical tests. *Journal of Musculoskeletal and Neuronal Interactions*, 12(4), 219-223.
- Yandell, M., Quinlivan, B., Popov, D., Walsh, C., & Zelik, K. (2017). Physical interface dynamics alter how robotic exosuits augment human movement: implications for optimizing wearable assistive devices. *J Neuroeng Rehabil*, 14(1), 40. doi: 10.1186/s12984-017-0247-9
- Young, A., & Ferris, D. (2017). State of the Art and Future Directions for Lower Limb Robotic Exoskeletons. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 25(2), 171-182.