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Linking the mechanics and energetics of hopping with elastic ankle exoskeletons

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Farris DJ, Sawicki GS. Linking the mechanics and energetics of hopping with elastic ankle exoskeletons. *J Appl Physiol* 113: 1862–1872, 2012. First published October 11, 2012; doi:10.1152/japplphysiol.00802.2012.—The springlike mechanics of the human leg during bouncing gaits has inspired the design of passive assistive devices that use springs to aid locomotion. The purpose of this study was to test whether a passive spring-loaded ankle exoskeleton could reduce the mechanical and energetic demands of bilateral hopping on the musculoskeletal system. Joint level kinematics and kinetics were collected with electromyographic and metabolic energy consumption data for seven participants hopping at four frequencies (2.2, 2.5, 2.8, and 3.2 Hz). Hopping was performed without an exoskeleton; with an springless exoskeleton; and with a spring-loaded exoskeleton. Spring-loaded ankle exoskeletons reduced plantar flexor muscle activity and the biological contribution to ankle joint moment (15–25%) and average positive power (20–40%). They also facilitated reductions in metabolic power (15–20%) across frequencies from 2.2 to 2.8 Hz compared with hopping with a springless exoskeleton. Reductions in metabolic power compared with hopping with no exoskeleton were restricted to hopping at 2.5 Hz only (12%). These results highlighted the importance of reducing the rate of muscular force production and work to achieve metabolic reductions. They also highlighted the importance of assisting muscles acting at the knee joint. Exoskeleton designs may need to be tuned to optimize exoskeleton mass, spring stiffness, and spring slack length to achieve greater metabolic reductions.

metabolic power; spring-loaded; locomotion; power; EMG

WHEN HUMANS UTILIZE BOUNCING gaits such as hopping and running, the supporting limb(s) during ground contact can be considered to act like a compressive spring (3, 13, 31). This has led to such motions commonly being modeled as a spring-mass system (12, 14, 19, 20, 31). The springlike behavior of the lower limb makes it possible for the muscle-tendon units acting around joints to store and return energy in elastic structures (1, 6, 26, 30). Predominantly, energy is stored and returned in tendons acting in series with the contractile elements of muscles (16, 34). The result of this is a reduction in the mechanical work requirements of contractile elements and substantial metabolic energy savings (2, 7). Some metabolic cost is still incurred owing to the contractile elements having to produce force to stretch tendons (2, 34).

The way in which the muscle-tendon units of the leg utilize elastic energy cycling to efficiently power bouncing gaits has inspired the design of orthoses and exoskeletons for the leg that seek to passively store and return energy during movement (8, 18, 22). Such devices might be useful for rehabilitating, restoring, and/or augmenting human locomotor performance by reducing musculoskeletal loading and reducing metabolic energy consumption. Loading of the leg is expected to be reduced by wearing assistive exoskeletons, because the device bears some of the load normally taken by the limb (18, 22). By using this loading to store and return energy in a spring, such devices may reduce the mechanical demands on biological tissues, which may, in turn, reduce metabolic energy consumption (22).

Previous studies of the effects of passive lower limb exoskeletons on human movement have typically used hopping motions because of the simple biomechanical goals of the task (8) and its similar spring-mass mechanics to running but less complicated joint kinematics (18). Grabowski and Herr (22) showed that a full leg exoskeleton crossing hip, knee, and ankle joints significantly reduced the metabolic power requirement of bilateral hopping (up to 30%) over a range of frequencies from 2.0 to 2.6 Hz. Their participants reduced their biological contribution to leg stiffness to keep the combined exoskeleton and biological stiffness similar to leg stiffness when hopping without an exoskeleton. This helped maintain consistent center of mass mechanics, which appears to be a goal of movement control for bouncing gaits (19, 20).

A large proportion ($\approx$45%) of the overall positive mechanical power output generated at leg joints during human running comes from muscle-tendon units acting at the ankle joint (17). Also, leg stiffness during hopping is predominantly determined by ankle joint stiffness (15). This highlights the importance of the ankle joint in maintaining the mechanics of bouncing gaits. Therefore, providing exoskeletal assistance at only the ankle joint could provide a significant proportion of the benefits achieved with a full leg device, while simplifying its design.

Both Ferris et al. (18) and Chang et al. (8) investigated the effects of wearing an ankle-foot orthosis that was spring-loaded to assist plantar flexion on joint and limb stiffness during unilateral hopping. Both studies showed that hoppers were able to reduce biological ankle stiffness to maintain overall limb stiffness when wearing the devices. Ferris et al. (18) also showed a concurrent reduction in the amplitude of electromyographic (EMG) signals from the plantar flexor muscles. This suggested that muscle activation was reduced to achieve lesser biological ankle joint stiffness when assistance was being provided by the device. While the studies of Ferris et al. (18) and Chang et al. (8) both suggest that spring-loaded ankle exoskeletons might unload biological tissues, it was not the aim of these studies to link the mechanics of hopping in ankle exoskeletons to the associated metabolic cost. Therefore, they did not determine the mechanical demands and metabolic power requirements of the task concurrently. Doing so would help to reveal if assisting only at the ankle joint could reduce the mechanical demands on leg muscles sufficiently to reduce the metabolic cost of a bouncing gait.

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The aim of this study was to investigate the effects of spring-loaded ankle exoskeletons on the metabolic cost and mechanics of bilateral human hopping. It was hypothesized that, when hopping with the spring-loaded ankle exoskeletons, participants would reduce plantar flexor muscle activity, and that this would reduce the mechanical power output by biological tissues (muscle-tendon units) at the ankle. Because of reduced mechanical demand placed on muscles, it was also hypothesized that hopping with spring-loaded exoskeletons would reduce metabolic power. Furthermore, it was expected that the ankle joint would dominate overall mechanical input to the task and thus metabolic reductions would be proportionally less than those previously achieved with a full-leg exoskeleton (22) (i.e., if the ankle joint provides 60% of the mechanical input, the observed reduction will be $\approx$60% of that previously reported for full leg assistance).

Glossary

d_{spring}	Moment arm of the exoskeleton spring
F_{spring}	Force on the exoskeleton spring
$\overline{\dot{M}}^+$	Average rate of joint moment production
$\overline{\dot{M}}^+_{ank}$	Average rate of joint moment production at the ankle
$\overline{\dot{M}}^+_{hip}$	Average rate of joint moment production at the hip
$\overline{\dot{M}}^+_{knee}$	Average rate of joint moment production at the knee
P_{exo}	Instantaneous exoskeleton power
$\overline{P}^+_{ank}$	Average positive power generated at the ankle joint
$\overline{P}^+_{bio}$	Average positive power generated by biological tissues at the ankle
$\overline{P}^+_{exo}$	Average positive power generated by the exoskeleton at the ankle
$\overline{P}^+_{hip}$	Average positive power generated at the hip joint
$\overline{P}^+_{joint}$	Average positive power generated at a generic joint
$\overline{P}^+_{knee}$	Average positive power generated at the knee joint
$\overline{P}^+_{mech}$	Average positive power
$\overline{P}^+_{tot}$	Total average positive power of all three joints
t_{hop}	Time for a complete hop to be completed
W^+	Positive work
ω_{ank}	Ankle angular velocity

MATERIALS AND METHODS

Seven male participants (mean $\pm$ SD, age = 28 $\pm$ 7 yr, height = 1.8 $\pm$ 0.06 m, mass = 80 $\pm$ 10 kg) gave written, informed consent to participate in this study. All participants were in good health and had no recent history of lower limb musculoskeletal injury. All procedures were approved by an institutional review board and complied with the guidelines for research involving human participants as set out in the Declaration of Helsinki.

Because metabolic power required for hopping and the effect of exoskeletons is frequency dependent (22), participants hopped in time with a metronome at four different frequencies (2.2, 2.5, 2.8, and 3.2 Hz). There were three different conditions: 1) without an exoskeleton (NE); 2) with an exoskeleton but with no spring (NS); and 3) with a spring-loaded exoskeleton to assist plantar flexion (S). Each trial lasted 4 min. Because of the physically challenging nature of the task, each condition was performed on a separate day to minimize fatigue effects. The order of conditions and frequencies was randomized. The intent of this experiment was to test the effects of the exoskeletons on naive users, so participants were given sufficient familiarization with hopping in the exoskeletons to feel comfortable, but not for long enough to introduce any training effects.

An example of the exoskeletons used is shown in Fig. 1. The exoskeleton consisted of a carbon fiber cuff around the upper shank, which was connected to a carbon fiber foot section via two aluminum bars, which had a freely rotating joint aligned with the participants' malleoli. The foot section was embedded in a training shoe, through the sole and around the heel. An extension spring could be attached to a bracket on the posterior aspect of the cuff and a bolt on the heel of the foot segment via a number of metal links. The number of links was adjusted for each participant, such that the resting length of the spring coincided with an ankle angle of 127°, which has been determined as the typical angle at ground contact in hopping (18). This same approach was used by Ferris et al. (18) for a similar exoskeleton. A compression load cell (Omegadyne, OH) was placed on the inferior side of the bolt at the heel of the foot segment and attached to the links in series with the spring. This was used to measure the forces in the spring. The stiffness of the spring in tension was 5 kN/m, and its moment arm about the joints was 0.135 m. This gave a rotational stiffness of 1.59 N·m/° (91 N·m/rad), which is $\sim$40% of ankle stiffness during unassisted hopping at preferred frequency (18).

An eight-camera motion analysis system (Vicon, Oxford, UK) was used to capture the three-dimensional positions of 22 reflective markers attached to the pelvis and right leg (modified Cleveland Clinic marker set). Raw marker positions were filtered using a second-order low-pass Butterworth filter with a cutoff of 10 Hz. A static standing trial was captured, and the positions of markers placed over segment end points (pelvis: left and right ASIS and greater trochanter) or lateral and medial aspects of joint flexion-extension axes (lateral only for the hip joint) were used to calibrate a four-segment (pelvis, thigh, shank, and foot) model for each subject using established inertial parameters (10). Clusters of three or four markers on rigid plates were attached to the pelvis, thigh, and shank segments to track segment motion during hopping. For the foot, a cluster of three markers was attached directly to the shoe. Joint angles for the hip knee and ankle were computed in three dimensions as the orientation of the distal segment with reference to the proximal segment and differentiated to calculate joint velocities.

Three-dimensional (3D) ground reaction forces applied to the left and right legs were computed during hopping using a split belt instrumented treadmill (Bertec, OH, USA). Participants hopped such that each foot was on a separate half of the treadmill, and thus the two
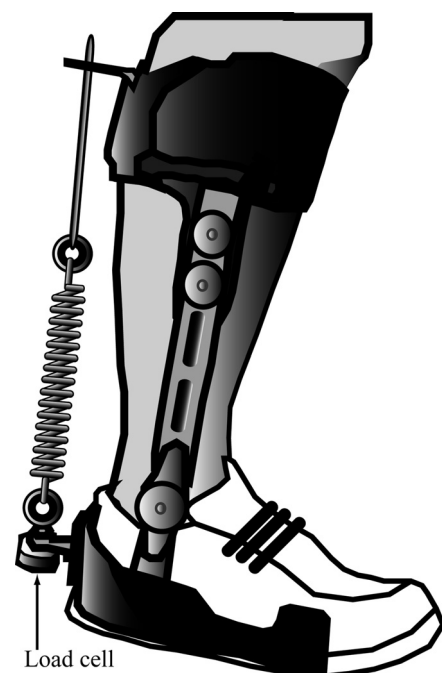

Fig. 1. Sketch of the spring-loaded exoskeleton and load cell.

3D force vectors could be attributed separately to the left and right legs. Raw analog force platform signals were filtered using a low-pass Butterworth digital filter with the cutoff set to 35 Hz. Inverse dynamic analyses (40) were then used to compute net joint moments, which were multiplied with joint velocities to calculate joint powers at the hip, knee, and ankle. Kinematics and kinetics were calculated for the right leg only, and it was assumed that the left leg behaved symmetrically. Inverse dynamic procedures were performed with Visual 3D software (C-motion, Germantown, MD).

The mechanical demands of hopping under different conditions was partly assessed by the average positive mechanical power ($\bar{P}^+_{mech}$) of the task. More details of the computation of $\bar{P}^+_{mech}$ can be found elsewhere (17). Briefly, it was calculated for each joint [ankle ($\bar{P}^+_{ank}$), knee ($\bar{P}^+_{knee}$) and hip ($\bar{P}^+_{hip}$)] as the sum of the integral of each period of instantaneous positive power for that joint (i.e., positive joint work) during a hop, divided by the time per hop (*Eq. 1*).

$$\bar{P}^+_{joint} = \frac{\sum W^+_1, W^+_2 \ldots W^+_n}{t_{hop}} \qquad (1)$$

where $\bar{P}^+_{joint}$ is the average positive power of the joint; $W^+_1 \ldots W^+_n$ are the values of positive work done during each period of positive instantaneous power during a hop; and t_{hop} is the time taken for a complete hop cycle.

$\bar{P}^+_{ank}$, $\bar{P}^+_{knee}$, and $\bar{P}^+_{hip}$ were summed to compute total average positive power ($\bar{P}^+_{tot}$) (*Eq. 2*).

$$\bar{P}^+_{tot} = \bar{P}^+_{ank} + \bar{P}^+_{knee} + \bar{P}^+_{hip} \qquad (2)$$

For the S condition, only a portion of $\bar{P}^+_{ank}$ was provided by biological tissues, and the rest came from the exoskeleton. To separate out these contributions, the power output of the exoskeleton was determined as follows. First, the force in the spring (measured by the load cell, Fig. 1) was multiplied by the spring's moment arm about the joint of the exoskeleton, to give the moment provided by the spring. This moment was multiplied by ankle angular velocity to compute the instantaneous power provided by the exoskeleton (*Eq. 3*).

$$P_{exo} = (F_{spring} \cdot d_{spring}) \cdot \omega_{ank} \qquad (3)$$

where P_{exo} is the instantaneous power provided by the exoskeleton about the ankle joint; F_{spring} is the force in the spring measured by the load cell; d_{spring} is the moment arm of the spring about the exoskeleton joint; and ω_{ank} is the angular velocity of the ankle.

The average positive power of the exoskeleton ($\bar{P}^+_{exo}$) was calculated as it was for joints. The contribution of biological tissues at the ankle ($\bar{P}^+_{bio}$) was calculated as $\bar{P}^+_{ank}$ minus $\bar{P}^+_{exo}$. All average powers were normalized to body mass (W/kg).

The rate at which muscles produce force may also be an important mechanical determinant of metabolic cost (29). Therefore, a rate of force metric was also computed. This metric was the average rate of joint moment production ($\dot{\bar{M}}^+$). Somewhat analogous to average positive power, this value was calculated for each joint [ankle ($\dot{\bar{M}}^+_{ank}$), knee ($\dot{\bar{M}}^+_{knee}$), and hip ($\dot{\bar{M}}^+_{hip}$)] by integrating the first derivative of each joint moment during periods of increasing moment production (i.e., when the derivative was positive), summing each of the integral values during each hop, and dividing the result by the time taken for an entire hop cycle. As for average positive power, $\dot{\bar{M}}^+_{ank}$ was reduced to separate exoskeleton and biological contributions for the S condition.

Surface EMG was used to record muscle activity from medial gastrocnemius (MG), lateral gastrocnemius (LG), soleus (SO) and tibialis anterior (TA). All four channels were recorded using wired electrodes (Biometrics, Newport, UK) that were carefully placed over muscle bellies after the skin surface was prepared by light abrasion and cleaned with an alcohol swab. DC offsets were removed from raw signals, which were then band-pass filtered (20–300 Hz). The data were then smoothed by calculating the root mean squared (RMS) value of the signals over a rolling window of 20 ms. The RMS of each muscle's signal was also calculated over the period of ground contact and the aerial phase of each hop as a metric of total activity over these two phases of the hop. For each session of data collection, the participant performed a set of 20 hops at 2.2 Hz while wearing exoskeletons without a spring. The maximum of the processed signals from each muscle for this condition was used to normalize EMG signals from the experimental conditions on that day. This was to avoid any effects of day-to-day variation in electrode placement on signal magnitudes.

Rates of oxygen consumption and carbon dioxide production during hopping trials were recorded using a portable metabolic system (Oxycon Mobile, Viasys Healthcare, CA). Before hopping, measurements were made during 5 min of quiet standing, and values from the last 2 min were averaged and used to calculate rates of metabolic energy consumption while standing. For the hopping trials, data from the last 2 of the 4 min were averaged for the calculation of metabolic rate. Visual inspection of rates of oxygen consumption with time (averaged over 30-s intervals) confirmed that participants were at steady state during this period, and the respiratory exchange ratio was never greater than one. Rates of oxygen consumption and carbon dioxide production were converted to metabolic powers using standard equations detailed by Brockway (5). Net metabolic powers during hopping were calculated by subtracting metabolic power during standing from metabolic power during hopping, and

Table 1. *Group hopping metrics at each frequency for each spring condition*

	ω, Hz	Hop Height, mm	Ground Contact Time, ms	Actual Frequency, Hz	Duty Factor	d_{Hop}, mm
No exoskeleton	2.2	22 ± 8	337 ± 41	2.2 ± 0.1	0.75 ± 0.09	18 ± 5
	2.5	20 ± 8	285 ± 35	2.5 ± 0.1	0.72 ± 0.08	20 ± 4
	2.8	13 ± 5	256 ± 29	2.8 ± 0.0	0.72 ± 0.07	17 ± 5
	3.2	9 ± 4	225 ± 19	3.2 ± 0.1	0.73 ± 0.06	17 ± 5
No spring	2.2	13 ± 6	352 ± 28	2.2 ± 0.1	0.78 ± 0.06	20 ± 5
	2.5	13 ± 6	300 ± 23	2.5 ± 0.0	0.75 ± 40.06	22 ± 10
	2.8	11 ± 4	264 ± 19	2.8 ± 0.0	0.74 ± 0.05	20 ± 7
	3.2	5 ± 2	264 ± 12	3.2 ± 0.1	0.79 ± 0.04	25 ± 7
Spring	2.2	17 ± 14	312 ± 35	2.3 ± 0.2	0.70 ± 0.05	25 ± 12
	2.5	15 ± 9	279 ± 27	2.5 ± 0.0	0.69 ± 0.07	22 ± 5
	2.8	13 ± 7	259 ± 23	2.8 ± 0.2	0.72 ± 0.04	28 ± 9
	3.2	9 ± 4	236 ± 22	3.1 ± 0.1	0.74 ± 0.05	29 ± 8
ANOVA P (S)		0.19	0.08	0.100	0.171	0.038
ANOVA P (ω)		<0.000*	<0.000*	<0.000*	0.014	0.424
ANOVA P (S × ω)		0.259	0.003*	0.263	0.071	0.432

Values are means ± SD. ANOVA *P* values are for main effects of the spring condition (S) and frequency (ω) and the interaction effect between the two (S × ω). d_{Hop}, distance moved by the center of pressure between consecutive hops. *Statistically significant effect.

these values were normalized to individual body mass (W/kg). Metabolic data were presented as the normalized net value, unless otherwise stated.

All of the kinematic and kinetic data for individual participants were reduced to the mean of at least 10 hops for each experimental condition. Unless otherwise stated, the values presented in this paper are the means ± SE for the whole participant group. To test for statistical differences in dependent variables between conditions, a two-way ANOVA with repeated measures was employed using SPSS software (IBM). The two independent variables for the ANOVA were as follows: *1*) spring condition (3 levels: NE, NS, S); and *2*) hopping frequency (4 levels: 2.2, 2.5, 2.8, 3.2 Hz). *F*-ratios for main and interaction effects were considered significant for $P < 0.05$. If a significant main effect was found, paired *t*-tests were used to make pairwise comparisons between levels of independent variables.

RESULTS

The S condition was described by similar hip and knee joint kinematics to the NS and NE conditions, but it did cause a shift in ankle angle to a generally more plantar-flexed position (Fig. 2). This shift was systematic, allowing the ankle to move through a similar range of motion (15–30°, dependent on frequency) in all conditions, but with a more plantar-flexed touchdown angle

for S. Similarly, joint kinetics (moments and powers) were similar across the NE, NS, and S conditions for the hip and knee (Figs. 3 and 4). The ankle joint moments showed that the exoskeleton provided between 30 and 50% of the total moment (Fig. 3), and this significant contribution was confirmed in the $\bar{P}_{mech}^{+}$ data, which showed that $\bar{P}_{exo}^{+}$ accounted for 36–38% of $\bar{P}_{tot}^{+}$, while $\bar{P}_{bio}^{+}$ accounted for 23–38% (Fig. 5). Total $\bar{P}_{ank}^{+}$ had generally greater peaks in the S condition (Fig. 4), indicating that the sum of $\bar{P}_{exo}^{+}$ and $\bar{P}_{bio}^{+}$ for S were greater than the biological power in NE and NS.

Net metabolic power typically followed a U-shaped relationship with frequency for NE, NS, and S (Fig. 6). This meant that net metabolic power requirements were least for the intermediate frequencies (2.5 and 2.8 Hz). Notably, there was a relatively large increase in net metabolic power from 2.8 to 3.2 Hz in the S condition (Fig. 6). This caused the mean value for net metabolic power for S to be greater than for NE at 3.2 Hz when it had been consistently less than NE and NS at all other frequencies. Net metabolic power was significantly less for S than NS at 2.2 (−16%, $P = 0.009$), 2.5 (−19%, $P = 0.006$,) and 2.8 Hz (−20%, $P = 0.047$), but only at 2.5 Hz (−13%, $P = 0.016$) compared with NE (Fig. 6).
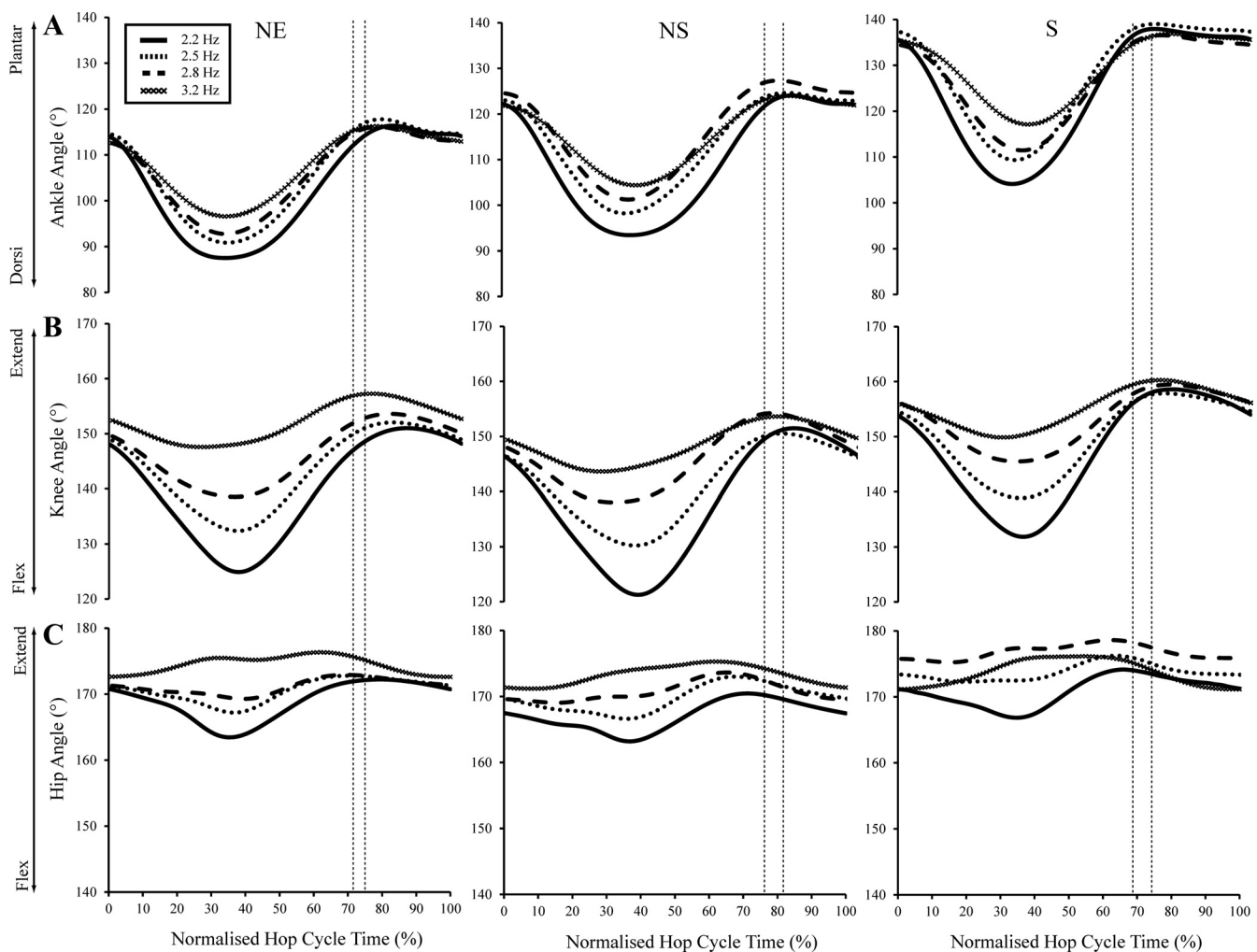

Fig. 2. Group mean ankle (*A*), knee (*B*), and hip (*C*) angles for each frequency, plotted over an average hop (normalized over 101 points). The *left* column is the no exoskeleton (NE) condition, *middle* is the no spring (NS) condition, and *right* is the spring-loaded (S) condition. Vertical lines represent the range of normalized take-off times across frequencies.
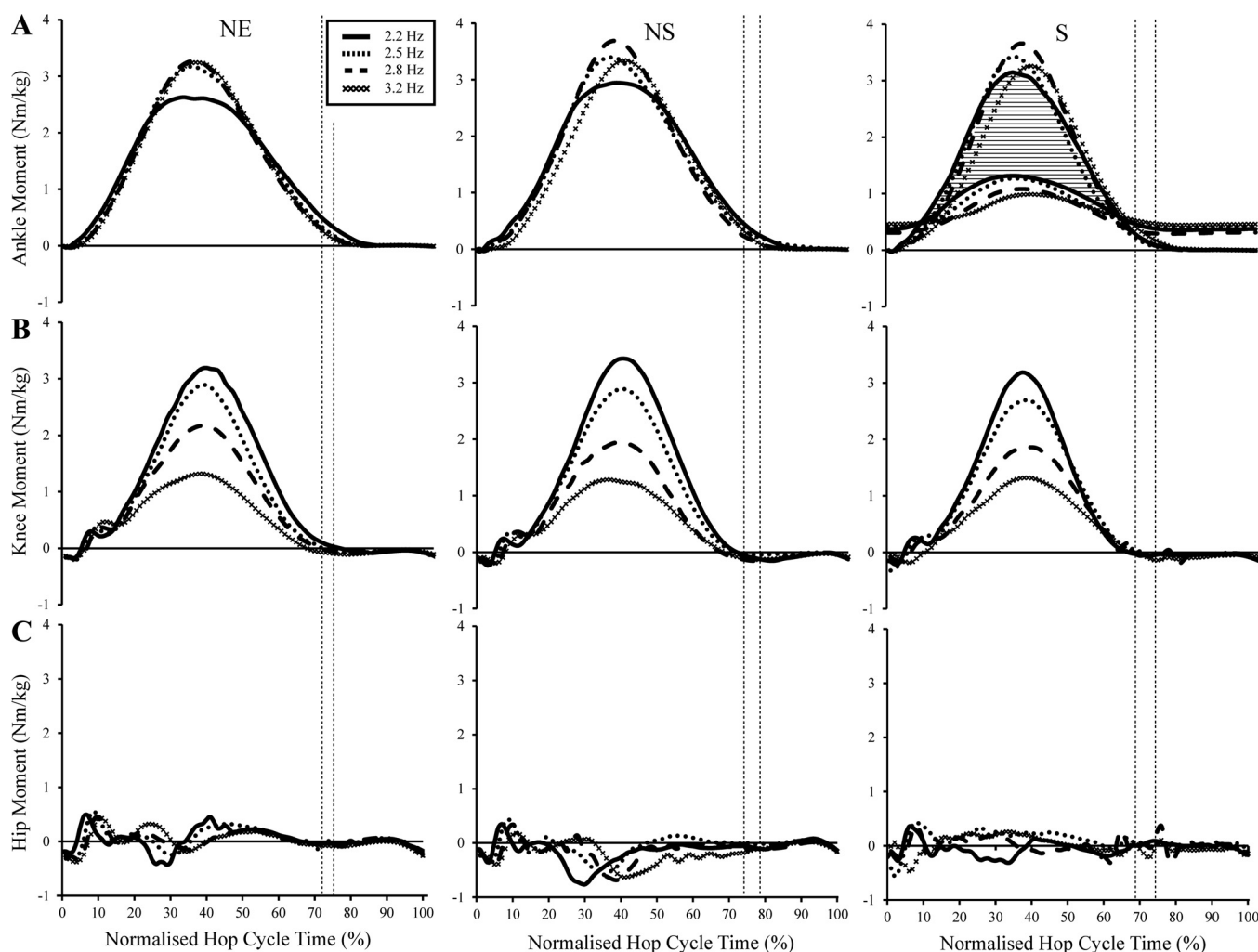
Fig. 3. Group mean ankle (*A*), knee (*B*), and hip (*C*) moments plotted over an average hop (normalized over 101 points). The additional data series of smaller magnitude on the S ankle plot are the exoskeleton moments for the corresponding frequencies, and the hatched area highlights the contribution from biological tissues to the total moment for 2.2 Hz. The *left* column is the NE condition, *middle* is the NS condition, and *right* is for the S condition. Vertical lines represent the range of normalized take-off times across frequencies.

The biological contribution to $\overline{P}_{tot}^{+}$ was significantly less for S than for NS ($\approx$30–40%) and NE ($\approx$20–30%) at all frequencies of hopping (Fig. 6*B*, ANOVA, $P < 0.001$). As can be observed in Fig. 6*B*, this difference was greater at low frequencies than at high frequencies, with the linear fits starting to converge at high frequencies. The biological contribution to $\overline{P}_{tot}^{+}$ was greater at low frequencies than at high frequencies for S, NS, and NE (Fig. 6*B*). This was in contrast to the biological contribution to $\overline{M}_{ank}^{+}$, which increased with hopping frequency (Fig. 6*C*). There was a significant difference in $\overline{M}_{ank}^{+}$ between the S and the NS conditions at 2.2 (21%, $P < 0.001$), 2.5 (24%, $P = 0.003$), and 2.8 Hz (20%, $P = 0.001$) (Fig. 6*C*). $\overline{M}_{ank}^{+}$ was only significantly less for S than NS at 2.2 (18%, $P = 0.007$) and 2.5 Hz (19%, $P = 0.05$) (Fig. 6*C*).

Time histories of the smoothed and normalized EMG signals are shown in Fig. 7. MG, LG, and SO all typically had lower activations toward the end of the aerial phase for S than for NE and NS. This was apparent in the RMS values for these muscles during the aerial phase, which were typically significantly less for S than for NS and NE (Fig. 8). During ground

contact, only SO had significantly lower activation for S (Fig. 8). Contrary to the plantar-flexor activations, TA exhibited significantly increased activation during ground contact and the aerial phase for all frequencies in the S condition compared with NE and NS (Fig. 8).

DISCUSSION

This study aimed to investigate the effects of spring-loaded ankle exoskeletons on the metabolic cost and mechanics of bilateral human hopping. First, it was hypothesized that the use of bilateral spring-loaded ankle exoskeletons during two-legged hopping would reduce ankle plantar-flexor muscle activity. This prediction was supported for all the plantar flexors tested during the aerial phase, but only SO during ground contact. It was also hypothesized that the reduced activity would reduce the contribution of plantar flexors to mechanical power output. This was found to be the case across all frequencies. The final prediction made was that using spring-loaded ankle exoskeletons would lower the metabolic power requirement
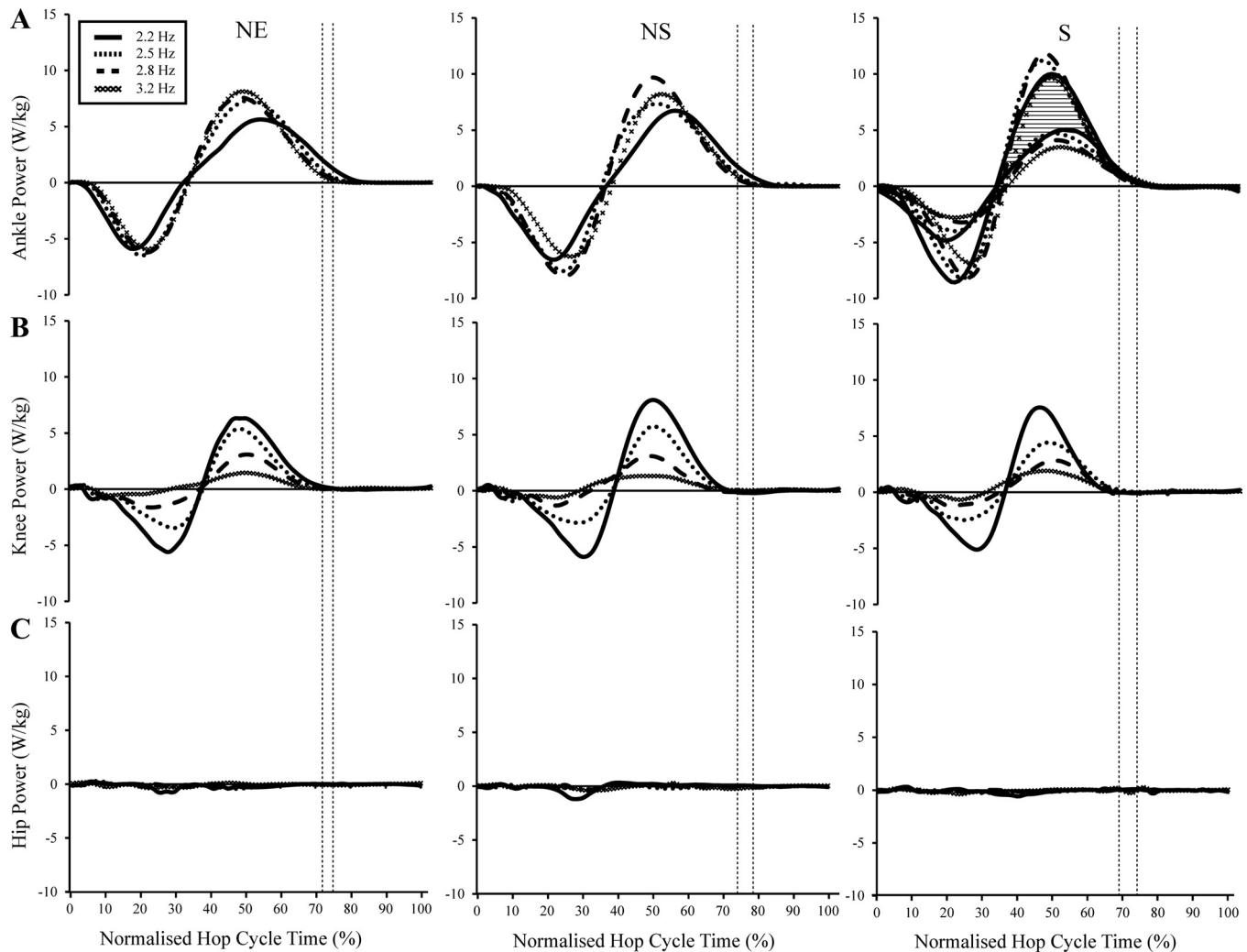
Fig. 4. Group mean ankle (*A*), knee (*B*), and hip (*C*) powers plotted over an average hop (normalized over 101 points). The additional data series of smaller magnitude on the S ankle plot are the exoskeleton powers for the corresponding frequencies, and the hatched area highlights the contribution from biological tissues to the total ankle power for 2.2 Hz. The *left* column is the NE condition, *middle* is the S condition, and *right* is for the NS condition. Vertical lines represent the range of normalized take-off times across frequencies.

of hopping compared with hopping without an exoskeleton or without a spring. This prediction was supported for some conditions at certain frequencies, but not all. Furthermore, where metabolic power reductions were achieved, they were less than previously observed with a full-leg exoskeleton (22). These findings raise interesting questions regarding the mechanisms by which spring-loaded ankle exoskeletons may be able to lower metabolic cost and also highlights some potential obstacles to this goal.

The results showed that the use of spring-loaded ankle exoskeletons resulted in significant reductions in the contribution of biological tissues (muscle-tendon units) to $\overline{P}_{tot}^{+}$ when hopping at all frequencies compared with the NE and NS conditions (Fig. 6*B*). This was due to a reduction in the biological plantar-flexion moment at the ankle (Fig. 3). This result was not surprising, as it has been previously shown that humans reduce biological ankle and limb stiffness to maintain overall system stiffness when hopping with spring-loaded ankle or full leg exoskeletons (8, 18, 22). The reduction in biological contribution to the plantar-flexion moment was

primarily achieved by a reduction in SO muscle activity during the ground contact phase of hops (Fig. 8). MG and LG RMS EMG values were only less during the aerial phase, and this was primarily due to a reduction in activity late in this phase (a similar trend was apparent in SO for the aerial phase). Therefore, it seems that preactivation of plantar flexors was also less when using spring-loaded exoskeletons.

The reduction in biological $\overline{P}_{mech}^{+}$ seen at all frequencies in the S condition was one indication that the mechanical demands on biological tissues were less with assistance. However, significant reductions in metabolic power only occurred at 2.5 Hz (−0.9 W/kg) compared with the NE condition and at 2.2 (−1.3 W/kg), 2.5 (−1.5 W/kg), and 2.8 Hz (−1.6 W/kg) compared with the NS condition. Furthermore, where there were reductions in metabolic power, they were not proportional with reductions in biological $\overline{P}_{mech}^{+}$ (Fig. 6). Therefore, while $\overline{P}_{mech}^{+}$ provides some indication of the mechanical demands of the task, it alone cannot fully explain the trends in metabolic cost that were observed. This is particularly notable at 3.2 Hz, where the metabolic power requirement was greater
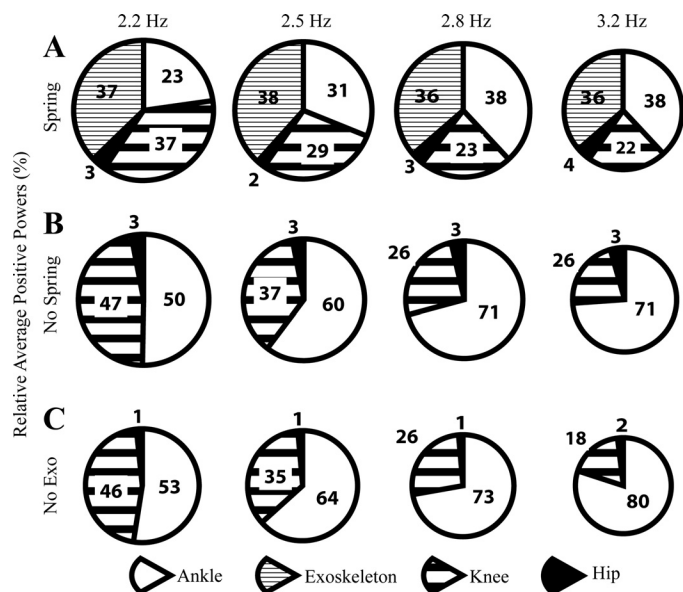
Fig. 5. Pie charts showing the percentage of total average positive power contributed by the biological tissues at the ankle (white), knee (thick hatching), hip (black), and by the exoskeleton (thin hatching) for the S (*A*), NS (*B*), and NE (*C*) conditions. The total area of the pies represents the total average positive power relative to all other pies.

for the S condition than for the NE condition, despite a reduction in $\overline{P}^{+}_{mech}$.

Another factor that may help to explain the observed trends in metabolic power is the metabolic cost associated with producing muscular force. An increasing number of studies of human (and animal) movement have indicated that a significant portion (up to 50%) of the metabolic cost of locomotion is related to the metabolic energy used by muscles to produce force (23, 24, 28, 29, 35). This cost has been proposed to increase proportionally with the rate at which force must be produced (29). The cost of producing force may be particularly important in muscle groups such as the plantar flexors that rely on the series compliance of tendons to cycle energy, while the muscle itself contracts relatively isometrically during stance (26, 30). In such contractions, the contractile elements of muscle do only small amounts of work, but must produce forces sufficient to "anchor" the tendon at the myotendinous junction. Several metrics have been used to relate muscle force to metabolic cost (9, 29, 32). McMahon et al. (32) used impulse over a gait cycle divided by time, giving an average force (N). Kram and Taylor (29) related metabolic cost of producing force to average rate of force (N/s). Dean and Kuo (9) related cost of force production to the rate of force cycling (N/s²). In the present study, $\overline{\dot{M}}^{+}_{ank}$ was considered as a metric for relating force production by the muscles acting at the ankle joint to metabolic cost, and this is analogous to Kram and Taylor's (29) rate of force metric. Figure 6 shows that $\overline{\dot{M}}^{+}_{ank}$ increased with frequency, lending some support to the notion that it becomes more influential at higher frequencies. In fact, at the two highest frequencies in this study, $\overline{\dot{M}}^{+}_{ank}$ could explain the trends in metabolic power. At 3.2 Hz, no significant differences in metabolic power were observed, and there were also no significant reductions in $\overline{\dot{M}}^{+}_{ank}$ (Fig. 6). For hopping at 2.8 Hz, there was a significant reduction in metabolic power for S
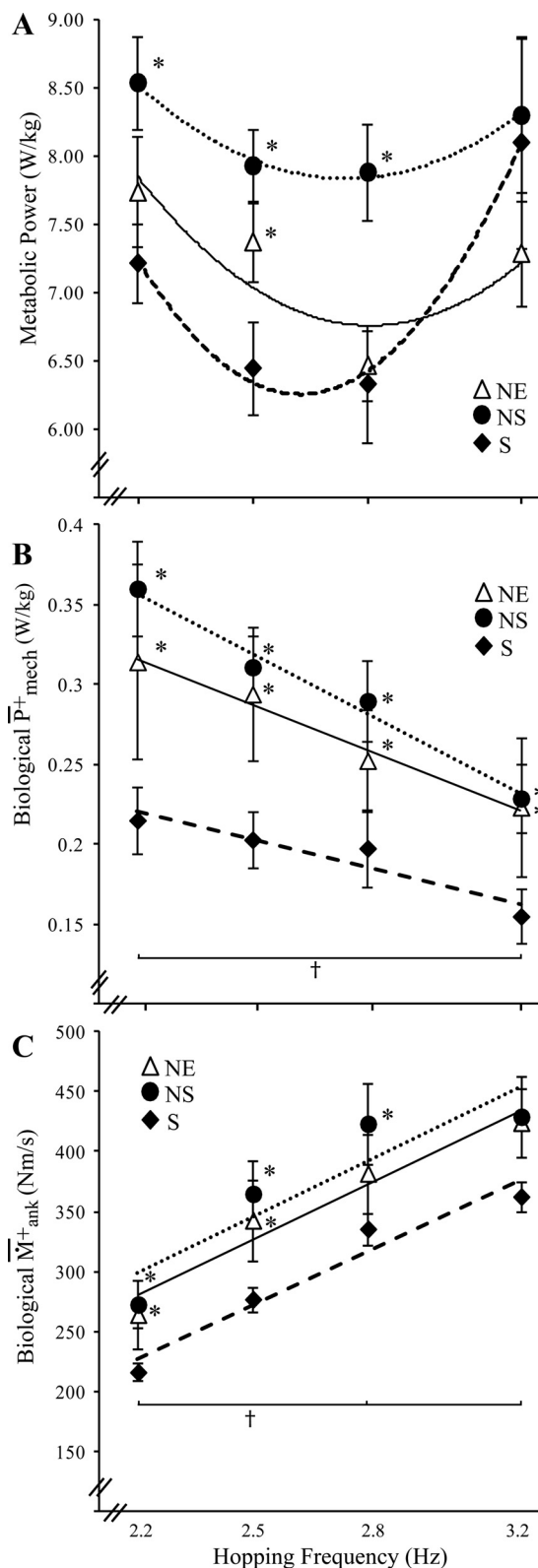

Fig. 6. Metabolic power (*A*), biological contribution to average positive mechanical power ($\overline{P}^{+}_{mech}$; *B*), and biological average rate of joint moment production at the ankle ($\overline{\dot{M}}^{+}_{ank}$; *C*) across hopping frequency for S (◆), NS (●), and NE (△). Values are means ± SE. *Significant difference from S at that frequency; †significant main effect of frequency.
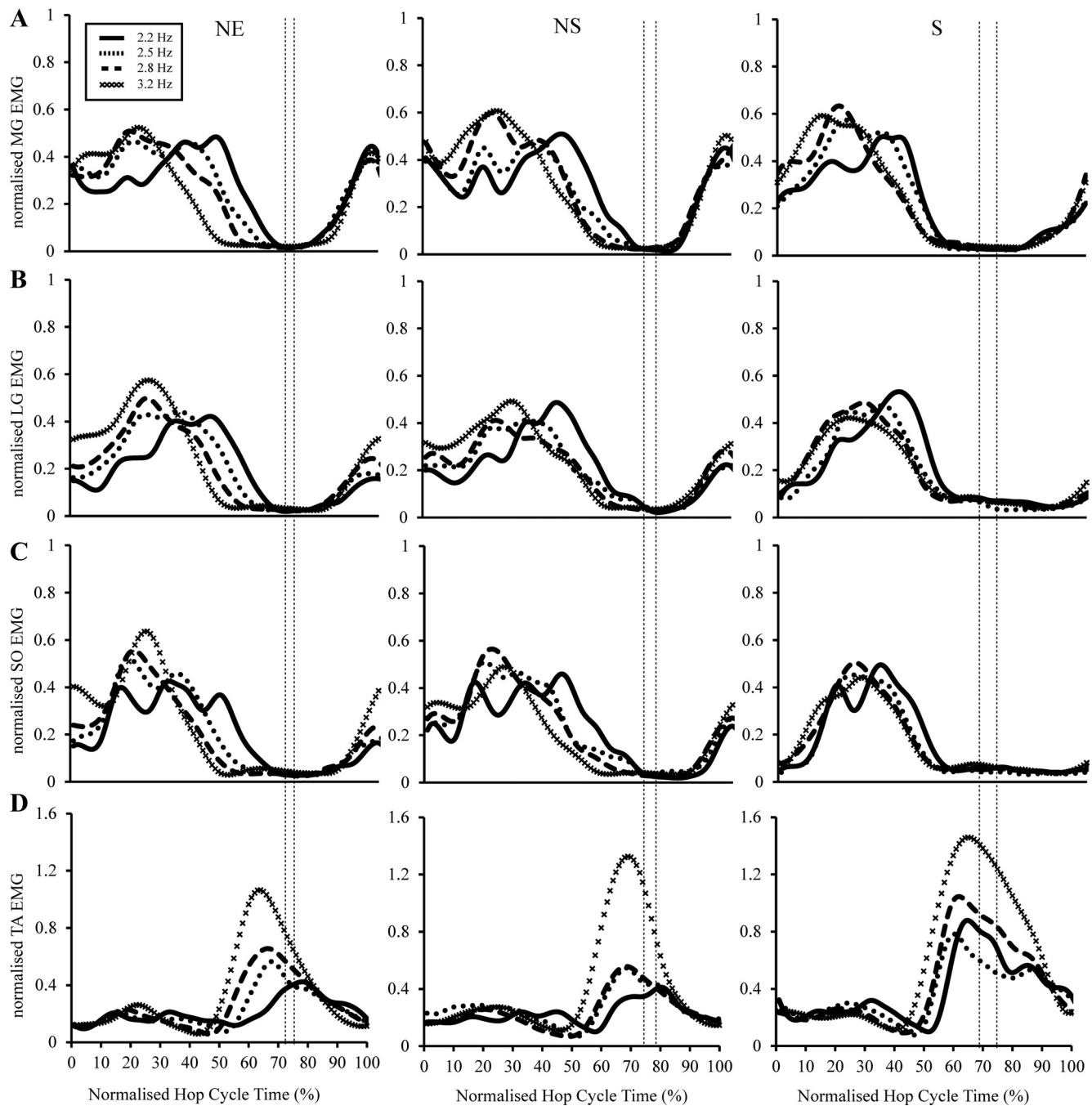
Fig. 7. Group mean normalized and smoothed time histories of electromyographic (EMG) activity for medial gastrocnemius (MG; *A*), lateral gastrocnemius (LG; *B*), soleus (SO; *C*), and tibialis anterior (TA; *D*) at each frequency for NE, NS, and S. Vertical lines indicate the range of normalized take-off times across frequencies.

compared with NS that coincided with a reduction in $\overline{\dot{M}}^+_{ank}$, but neither variable was reduced compared with NE (Fig. 6). Therefore, at higher frequencies, metabolic power seems more dependent on $\overline{\dot{M}}^+_{ank}$ than $\overline{P}^+_{mech}$, and, at 3.2 Hz, the exoskeletons did not reduce $\overline{\dot{M}}^+_{ank}$ sufficiently. Similar to the present finding, in a study of human bouncing tasks, Dean and Kuo (9) actually showed that the rate of force cycling in ankle plantar flexors increased with bouncing frequency and could explain why metabolic cost began to increase at frequencies above 3 Hz, despite muscular work decreasing. Dean and Kuo (9) related force mul-

tiplied by bounce frequency squared to metabolic rate. This is slightly different from the present definition, where $\overline{\dot{M}}^+_{ank}$ was related to metabolic rate, but both studies suggest that force-related metabolic costs dominate over work-related costs in determining metabolic cost at high frequencies.

However, at lower frequencies, this may not be the case. For both 2.2 Hz and 2.5 Hz, there were significant reductions in both $\overline{P}^+_{mech}$ and $\overline{\dot{M}}^+_{ank}$ for S compared with NE and NS, but metabolic cost was not reduced at 2.2 Hz compared with NE (Fig. 6). A possible explanation for this might be related to the
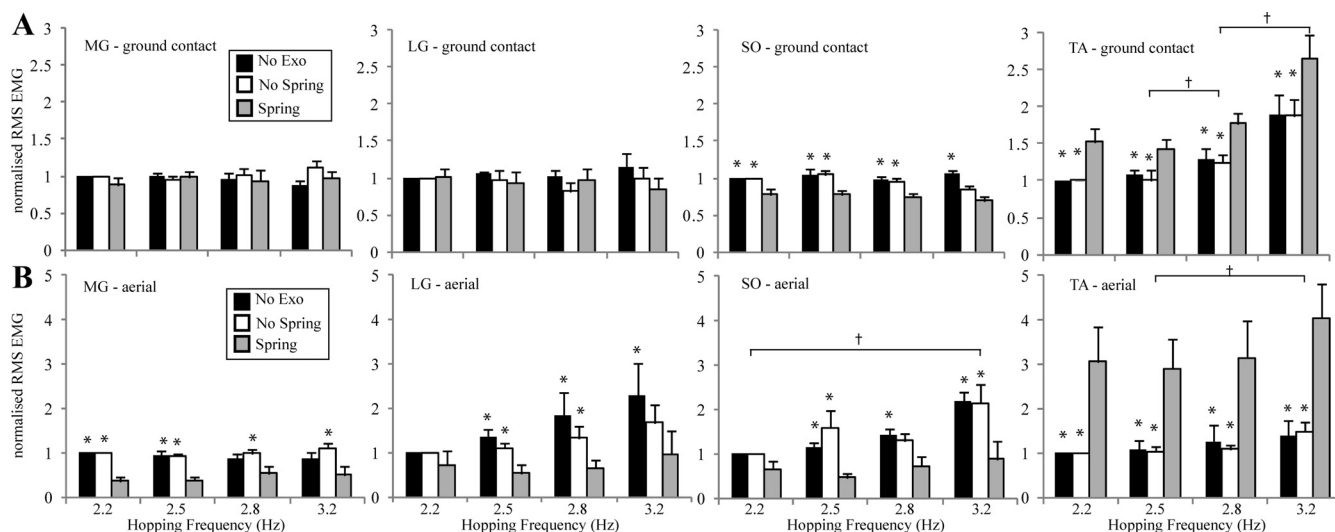
Fig. 8. Group mean ± SE root mean square (RMS) EMG values for MG, LG, SO, and TA during the ground contact (*A*) and aerial (*B*) phases of hops. *Statistical significantly different from the S condition. †Significant effect of frequency.

energy consumed by more proximal muscle groups. Although the muscles acting at the ankle joint were responsible for the majority of power output, a nontrivial contribution was provided by knee extensors (18–46%), and this was greatest at 2.2 Hz (Fig. 5). Work provided by more proximal muscle-tendon units has been postulated to be done less efficiently (37). This is owing to their lesser ability to make use of series elastic compliance than muscle groups such as the plantar flexors. Therefore, although the contribution of more proximal muscle groups to power output was less than at the ankle, they probably provided this power less efficiently. Thus they may be responsible for a larger proportion of the total energy consumed than mechanical power output suggests. Because the muscles acting at the knee were providing a greater proportion of the total positive power at 2.2 Hz, the effectiveness of the exoskeletons at reducing overall mechanical demands may have been diminished. This may have contributed to the lack of significant reductions in metabolic power compared with NE.

The influence of more proximal muscle groups was highlighted by the lesser reductions in metabolic power that were achieved in this study than by a full-leg exoskeleton in a previous study (22). With a full-leg exoskeleton, reductions were in the region of 2.0–2.5 W/kg (≈30%) (22). By comparison, in the present study, the greatest reductions observed were ~1.5 W/kg (≈20%). Therefore, it seems that it would be beneficial to assist at more proximal joints as well as the ankle. As can be seen from Fig. 5, the contribution of the hip to overall mechanical power output was minimal. Therefore, it might be that assisting at the ankle and knee joints would provide the desired reductions in metabolic cost without requiring a device that assists at the hip too. However, this may only apply to hopping in place, which requires no forward swing of the leg.

Another further explanation for the lesser metabolic power reductions observed here may be apparent in the RMS EMG data for TA. There was a significant increase in TA RMS EMG during both the aerial and ground contact phases of hopping across all frequencies for the S condition. Particularly notable was the two- to threefold increase in normalized EMG signal

magnitude during the aerial phase (Figs. 7 and 8). In the NE and NS conditions, there was apparent preactivation of the plantar flexors before ground contact that was opposed by some coactivation of TA (Fig. 7). In this case, TA may be serving to stiffen the ankle joint to allow preactivation of the plantar flexors without generating rotation at the ankle joint (there was <5° of rotation at the ankle during the aerial phase, despite considerable plantar-flexor activity). When hopping with spring-loaded exoskeletons, the preactivation of the plantar flexors was less (Figs. 7 and 8), and this resulted in significantly smaller RMS EMG values for the plantar flexors during the aerial phase (Fig. 8). This might lead one to think that TA should have exhibited lesser activation at the same time to maintain equilibrium of the force couple at the ankle joint and consistent kinematics. However, as shown in Fig. 2, the spring was actually exerting force to generate a plantar-flexion moment during the aerial phase. Therefore, TA had to be active to oppose this moment and was effectively coactivating antagonistically with the spring. It was inferred from the raised activation of TA in the spring-loaded condition that the moment generated by the spring during the aerial phase was greater than that generated by the plantar flexors in the NS and NE conditions. A consequence of raised TA coactivation could be increased metabolic energy consumption that partially canceled out the benefits of reduced plantar-flexor activation. Elevated levels of antagonistic coactivation have been cited as a factor contributing to increased energy consumption for locomotion in older populations (25, 33). Therefore, the elevated activation of TA to work against the spring during the aerial phase may have contributed to there being lesser metabolic power reductions than were expected.

The apparent engagement of the spring during the aerial phase of hops likely represents a design flaw of the exoskeletons. The spring was attached such that its slack length would be reached when the ankle was at 127° of flexion. This angle was chosen from a previous study using similar devices as the typical angle at touchdown (18). Thus it was intended that the spring would remain slack (i.e., exerting no force) until contact was made with the ground, and it could be used to store energy

transferred from kinetic and potential energy of the body. However, Fig. 2 shows that the touchdown angle for the NS and NE conditions was less (more dorsiflexed) than 127° ($\approx$116–125°). Extension of the spring would have been required to reach these angles in the S condition. As can be seen in Fig. 2, the touchdown angle of the ankle was greater (more plantar-flexed) in the S condition and at $\approx$132° should have placed the spring below its slack length. However, the spring was still exerting force before touchdown and thus must have been longer than its slack length. The results indicate that the angle at which the spring engages is an important factor and should be optimized (possibly on an individual basis) to maximize energy storage but minimize aerial coactivation of TA. Alternatively, a clutching mechanism that selectively engages the spring after landing and disengages it after take-off might solve this issue. Such mechanisms are useful for walking (39) and running (11) with spring-loaded assistive devices, to prevent wearers from working against springs during the swing-phases of gait.

Other aspects of the exoskeleton that should be considered as "tuneable" are its mass and spring stiffness. Mass is a relatively simple concept in that it is a matter of making the exoskeletons as light as possible. The exoskeletons used here were prototypes and thus were not fine-tuned to be as light as would be desirable. The added mass of the exoskeletons adds mechanical demands and associated metabolic cost to the task. In fact, it can be seen from the size of the pies in Fig. 5 that overall mechanical power output was greater in the NS and the S conditions than for NE. When drawing comparisons between the S and NS conditions, significant reductions in metabolic cost were observed at 2.2, 2.5, and 2.8 Hz. This indicated that the assistance provided by the spring was typically sufficient to counteract the added mass of the exoskeleton, but only improved upon energy consumption of normal hopping at 2.5 Hz. Reductions in metabolic cost relative to the NE condition would be expected to reach similar values to reductions compared with the NS condition as the mass of the device approaches zero.

With regard to spring stiffness, it may be necessary to design experiments to try and find an optimal stiffness for metabolic reductions. The spring chosen for this study was selected as the stiffest spring in which a pilot subject could complete the task. It was considered that this was the spring that would provide maximal unloading of biological tissues. However, maximal unloading does not necessarily correspond to the greatest reductions in metabolic cost (4). In fact, Grabowski and Herr (22) showed with their full-leg exoskeleton that a stiffer spring can increase the net metabolic power required for hopping compared with a less stiff spring. They proposed that this was because the stiffer exoskeleton may have compromised stability and/or sensory feedback. As a proxy for stability, Grabowski and Herr (22) calculated the distance moved by the center of pressure between consecutive hops and showed that this distance was typically larger for the stiffer exoskeleton. In this study, the same metric was calculated (d_{Hop}) and was typically greater for the S condition (Table 1), suggesting that participants were less stable in that condition. Therefore, although intuition might suggest that a greater stiffness (up to the normal ankle stiffness) would be more beneficial, it might actually impair the person's ability to hop. It seems plausible that there would be some optimal stiffness that maximizes energy storage and return without significantly impairing stability and sensory feedback. Furthermore, this optimal stiffness might vary with hopping frequency and between individuals, depending on their body mass and stature. Careful studies designed to explore these factors are still needed to fine-tune the exoskeleton design.

This study showed that spring-loaded ankle exoskeletons can be used to reduce the metabolic cost of bilateral human hopping at certain frequencies. It has highlighted the potential for such devices to assist locomotion by harnessing elastic energy and also some of the complex relations between mechanics and energetics that must be considered when attempting this. Hopping was studied here because the simple cyclic motion permitted a controlled experiment with an uncomplicated exoskeleton design. Elastic stretch and recoil of the Achilles tendon occurs in both walking and running (30), and thus more sophisticated elastic ankle exoskeletons that account for the more complex kinematics of these motions could assist ankle push-off and reduce the metabolic costs of these motions. This could benefit certain pathological populations who simultaneously lack ankle push-off and incur elevated metabolic costs during locomotion [e.g., poststroke or with cerebral palsy (21, 27, 38)]. They might also reduce musculoskeletal loading and offset increases in metabolic cost for individuals who must carry heavy additional loads, such as certain military personnel. Future studies should aim to investigate optimal spring stiffness and engagement angle. This is a large parameter space and simulations, such as that of Robertson and Sawicki (36) may prove a good initial approach to narrowing that space. They should also consider how the exoskeletons interact with the underlying muscle-tendon units and how they might be redesigned to be used in running or walking where the kinematics become more complex than for hopping.

DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: D.J.F. and G.S.S. conception and design of research; D.J.F. and G.S.S. performed experiments; D.J.F. and G.S.S. analyzed data; D.J.F. and G.S.S. interpreted results of experiments; D.J.F. and G.S.S. prepared figures; D.J.F. and G.S.S. drafted manuscript; D.J.F. and G.S.S. edited and revised manuscript; D.J.F. and G.S.S. approved final version of manuscript.

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