EXOSKELETONS

Improving the energy economy of human running with powered and unpowered ankle exoskeleton assistance

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Exoskeletons that reduce energetic cost could make recreational running more enjoyable and improve running performance. Although there are many ways to assist runners, the best approaches remain unclear. In our study, we used a tethered ankle exoskeleton emulator to optimize both powered and spring-like exoskeleton characteristics while participants ran on a treadmill. We expected powered conditions to provide large improvements in energy economy and for spring-like patterns to provide smaller benefits achievable with simpler devices. We used human-in-the-loop optimization to attempt to identify the best exoskeleton characteristics for each device type and individual user, allowing for a well-controlled comparison. We found that optimized powered assistance improved energy economy by 24.7 ± 6.9% compared with zero torque and 14.6 ± 7.7% compared with running in normal shoes. Optimized powered torque patterns for individuals varied substantially, but all resulted in relatively high mechanical work input (0.36 ± 0.09 joule kilogram^{-1} per step) and late timing of peak torque (75.7 ± 5.0% stance). Unexpectedly, spring-like assistance was ineffective, improving energy economy by only 2.1 ± 2.4% compared with zero torque and increasing metabolic rate by 11.1 ± 2.8% compared with control shoes. The energy savings we observed imply that running velocity could be increased by as much as 10% with no added effort for the user and could influence the design of future products.

INTRODUCTION

Addressing potential obstacles to engaging in running for fitness may increase participation and improve general health. Running provides many benefits, from improved cardiovascular health (*1*) to alleviated symptoms of depression (*2*, *3*) and even markedly reduced risk of all-cause mortality (*4*, *5*). Although running has a low barrier to entry, many factors can prevent participation. In 2018, only about 25% of Americans between 18 and 29 reported going for a run or a jog at least once within a 12-month period, and participation dropped to 20% for adults 30 to 49 years of age (*6*). Time commitments and the internal belief that individuals are not athletic are the two greatest barriers to increasing physical activity (*7*). Perceived exertion and activity intensity also have a negative association with frequency of exercise, whereas perceived ability, enjoyment, and interest can increase activity levels and overall energy expenditure (*7*–*9*).

Exoskeletons that augment running performance could potentially lead to greater participation in physical activity by decreasing activity intensity and fatigue while simultaneously enhancing perceived ability, interest, and enjoyment. The effects could be similar to those of electric-assist bicycles, which have been shown to reduce barriers to physical activity for noncyclists (*10*). Exoskeletons could allow entry-level runners to match the pace of more-fit friends, which would be meaningful because motivation and adherence to exercise are improved when exercising with a partner (*11*, *12*, *8*). Exoskeletons may even be used to allow fair and enjoyable competition between runners of disparate skill levels, much like a sport handicap. Enabling challenge and competition in this way might further improve participation (*9*).

The potential impact of reducing the energetic cost of running is clear, and several devices have already demonstrated success. For example, Nike's Vaporfly marathon shoes can reduce the metabolic cost of running by 4% (*13*), whereas connecting an elastic band between the feet can reduce energy cost by 6% (*14*) and a passive hip exoskeleton can improve energy economy by 8% compared with running without the exoskeleton (*15*). These devices have been successful without the need for motors or electrical power. Powered devices have also succeeded, such as a portable hip exosuit that reduced the energy cost of running by 8% compared with an unpowered mode and 4% compared with running without the exoskeleton (*16*). The same soft exoskeleton was recently used to automatically transition from walking to running assistance (*17*). These exciting results have the potential to make significant differences for elite athletes, but larger metabolic reductions may be needed to noticeably improve performance for amateur runners.

Spring-like assistance at the ankle joint might be very effective at reducing the energy cost of running. Legs naturally exhibit spring-like behavior during running (*18*–*20*) with mechanics that appear to be relatively consistent. For example, engaging a spring in parallel with the knee joint during running does not seem to alter total knee stiffness (*21*). Under the right conditions, a spring-loaded exoskeleton might perform some of the spring-like functions of leg muscles and tendons, saving metabolic energy. This may explain the success of spring-like devices acting about the hips, which are associated with elastically accelerating leg swing. Greater benefits might be possible by assisting the ankles, which have been shown to be important in controlling axial leg stiffness (*22*). This may be why ankle exoskeletons with springs and clutches in parallel with the Achilles tendon and calf muscles can improve the energy cost of walking (*23*). Because the legs exhibit simpler spring-like

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behavior during running, it is possible that spring-like assistance at the ankle joint could be even more effective in this context.

Unpowered spring-like exoskeletons might provide excellent performance trade-offs for running. The net benefit of an exoskeleton is the benefit of the assistance it provides plus the penalty for wearing it. Penalties include increased energy cost for added mass or restriction of movement. For example, adding mass to the legs can substantially increase the energy cost of running (*24*). Exoskeletons with spring-like behavior could operate without the mass of batteries and motors needed to perform net positive work. Although they might provide a lower benefit from assistance, a smaller penalty might yet lead to a larger net benefit. This framework might also help explain the lack of benefit of some past spring-loaded exoskeletons (*25*–*27*). Some benefits of assistance may have been lost because of energy dissipation in soft tissues (*30*), especially problematic for an unpowered device, whereas whole-leg designs could have led to high penalties for added mass. Interfacing in stiffer regions of the body, such as the heel and shin, with a mass-efficient, single-joint device could improve performance.

Although specialized shoes and hip exoskeletons have assisted running, past results suggest that powered ankle exoskeletons could be even more effective. Powered ankle exoskeletons have been used extensively during walking (*29*–*33*) and have produced improvements in metabolic rate as large as 24% (*34*) compared to walking with the exoskeleton in a zero-torque mode and 11% compared to walking in normal shoes (*35*). Preliminary running tests with untethered devices developed in industry suggest improvements (*36*), and our own prior single-subject experiment with a tethered device resulted in a 27% reduction in energy cost compared with a zero-torque mode (*34*). In addition to promising experimental findings, the ankle provides more positive work than the hip or the knee during normal running (*37*), suggesting that it may be a more effective location for applying assistance. However, the benefits of ankle exoskeletons may not offset the high penalties associated with their mass and distal location on the user's legs during running. The cost of carrying mass increases as the mass is placed more distally (*38*), and the negative effect of added mass to the foot on running speed and metabolic rate is well documented (*24*, *39*, *40*). These factors make assisting running with an ankle exoskeleton more challenging than assisting walking.

Accurately comparing the relative performance of powered and spring-like ankle assistance is possible with an ankle exoskeleton emulator and human-in-the-loop optimization. Exoskeleton emulators (*41*) can be used to emulate the behavior of a wide variety of exoskeleton designs without changing baseline device characteristics or developing new hardware. For example, an emulator can apply torque in a manner consistent with a clutched-spring exoskeleton (*42*), allowing researchers to adjust spring stiffness or clutch timing in software without affecting other important features such as device mass or interface comfort. This approach is especially powerful when paired with human-in-the-loop optimization to systematically explore the vast space of possible device characteristics.

Human-in-the-loop optimization can find device characteristics that outperform approaches based on bioinspiration, simulation models, or hand tuning (*34*, *43*, *44*). Selecting effective exoskeleton characteristics is not trivial; people are difficult to model or predict (*45*), often have different responses to the same exoskeleton assistance (*33*), and require significant training to learn to use exoskeletons effectively (*46*). For a given type of exoskeleton, human-in-the-loop optimization can be used to identify customized exoskeleton characteristics that are ideal for an individual participant. Averaging these characteristics across individuals can then be used to estimate generic exoskeleton characteristics that are expected to provide good, although suboptimal, performance enhancements for most users. When developing a new exoskeleton or addressing a new task, such as running with ankle exoskeletons, optimization can thereby identify good generic characteristics that cannot be known in advance. Human-in-the-loop optimization can also allow better across-device comparisons because the best versions of each device type for each individual runner can be found and compared.

The goal of this study was to assess reductions in metabolic cost during running with spring-like and powered ankle exoskeleton assistance, compared with running with the exoskeleton in

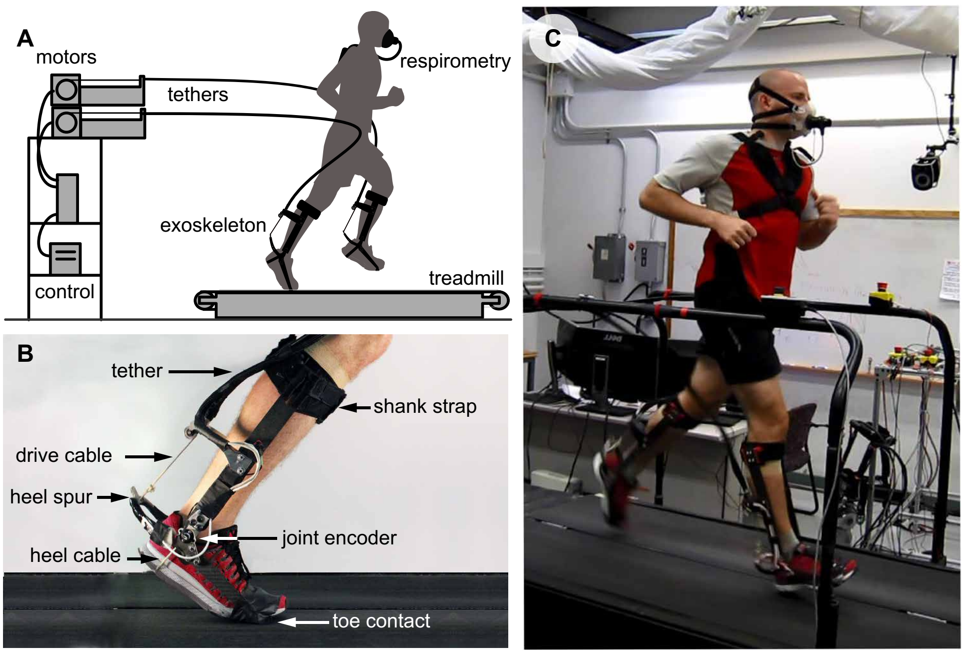

Fig. 1. Experimental setup. (**A**) Exoskeleton emulator testbed. A participant runs on a treadmill while wearing bilateral ankle exoskeletons actuated by motors located off-board with mechanical power transmitted through flexible Bowden cables. (**B**) Ankle exoskeleton. The ankle exoskeleton attaches to the user by a strap above the calf, a rope through the heel of the shoe, and a carbon fiber plate embedded in the toe of the shoe. The inner Bowden cable terminates on a 3D printed titanium heel spur that is instrumented with strain gauges for direct measurement of applied torque. A magnetic encoder measures ankle angle. (**C**) Participant running on the treadmill with bilateral ankle exoskeletons. Metabolic data are collected through a respiratory system by measuring the oxygen and carbon dioxide content of the participant's expired gasses.

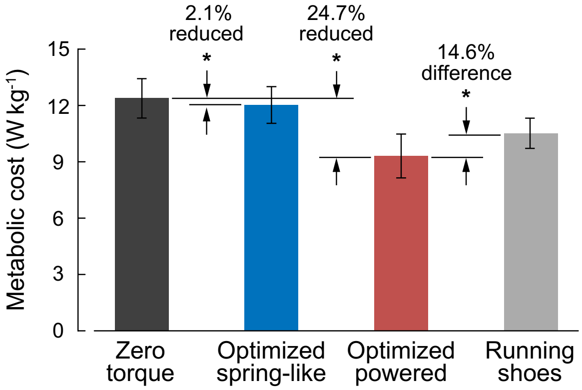

Fig. 2. Metabolic results. Optimized spring-like and Optimized powered assistance resulted in metabolic reductions of 2.2 and 24.7%, respectively, compared with zero-torque mode, while running at 2.7 m s^{-1}. Optimized powered assistance resulted in an improvement in running economy of 14.6% compared with running in normal shoes, whereas Optimized spring-like assistance resulted in an 11.1% increase in the energy cost of running. Error bars indicate SD. $^*P < 0.05$.

zero-torque mode and running in normal shoes. We expect the results to influence the design of future products by testing the feasibility of a passive spring-like approach and identifying broad design requirements for portable, powered ankle exoskeletons.

RESULTS

Human-in-the-loop optimization

We used human-in-the-loop optimization over a set number of generations to estimate optimal settings for each controller and each participant, referred to as the Optimized settings. This allowed for an accurate comparison of spring-like and powered strategies. The experiment was conducted using a highly adaptable exoskeleton emulator (Fig. 1A) (*41*) with bilateral ankle exoskeleton end effectors (Fig. 1B). The use of the emulator system enabled rapid alteration of device settings and ensured that the two assistance strategies were compared in isolation with constant inertial properties and interface stiffness. We conducted tests with 11 male participants running on a treadmill at a moderate pace of 2.7 m s^{-1} (Fig. 1C and movie S1).

Metabolic rate

Optimized powered assistance improved energy economy for the runners, whereas Optimized spring-like assistance was less effective (Fig. 2). Net metabolic rate under the Optimized powered condition was 24.7 ± 6.9% lower (mean ± SD) than running with the exoskeletons in zero-torque mode (t test, $P = 1 \times 10^{-6}$, $n = 11$) and 14.6 ± 7.7% lower than running with normal shoes (t test, $P = 1 \times 10^{-4}$, $n = 11$). Net metabolic rate under the Optimized spring-like condition was 2.1% ± 2.4 lower than running with the exoskeletons in zero-torque mode (t test, $P = 0.01$, $n = 11$) and 11.1 ± 2.8% higher than running with normal shoes (t test, $P = 7 \times 10^{-8}$, $n = 11$). Net metabolic rate under the Optimized powered, Optimized spring-like, zero-torque, and normal shoe conditions were 9.2 ± 1.2, 12.0 ± 0.9, 12.2 ± 1.1, and 10. 8 ± 0.9 W kg^{-1}, respectively.

Mechanics

The optimization process identified moderately high peak torques with the peak occurring late in stance for all participants under the powered condition (Fig. 3A). Optimized peak torque for the powered condition was 0.75 ± 0.14 N·m kg^{-1} on average, occurring at 76 ± 5.0% stance. Optimized onset of torque occurred at or just after foot flat (23 ± 8.7% stance). Optimized timing for turning off torque was near toe-off (97 ± 1.4% stance) for all participants. The average mechanical power produced during stance was 0.48 ± 0.12 W kg^{-1} corresponding to 0.36 ± 0.09 J kg^{-1} of work provided by the exoskeleton on each step. Work loops for each individual under the Optimized powered condition (Fig. 3B) have large areas, illustrating that a significant amount of energy was injected into the system. The peak mechanical power produced was 5.6 ± 1.0 W kg^{-1}.

Peak torque was lower and varied more in magnitude between participants for the Optimized spring-like condition compared with the powered condition (Fig. 3C). No clear trends in Optimized stiffness or engagement angle were identified, indicated by the wide range of slopes and x axis intercepts for individual torque-angle curves (Fig. 3D). Average power during spring-like assistance was negligible, as prescribed: −0.006 ± 0.012 W kg^{-1}, corresponding to −0.005 ± 0.009 J kg^{-1} of net work per step on average. Peak power experienced under the spring-like condition was 1.0 ± 0.7 W kg^{-1}. The small areas within the average work loops (Fig. 3D) demonstrate that the emulation of spring-like behavior was reasonable. The average mechanical work delivered by the exoskeletons per step in the zero-torque mode was negligible at −0.001 ± 0.003 J kg^{-1}.

DISCUSSION

Powered ankle exoskeleton assistance reduced the metabolic cost of treadmill running more than spring-like assistance in a controlled comparison. Powered ankle assistance produced greater improvements in energy economy than prior attempts using other devices. Embedding a similar powered strategy with user-specific customization into a mobile exoskeleton could substantially increase speed or endurance (*39*, *47*). This might make running easier and more enjoyable, potentially leading to increased participation and better health outcomes. Using the design specifications identified in this study, future experiments could test these ideas with product-like ankle exoskeletons.

A comparison of Optimized settings under the powered condition across participants suggests a role both for generic assistance and customization. Several features of the Optimized pattern of torque were similar across participants, including relatively late timing of peak torque and smoothly ramping torque to zero at the moment the foot left the ground. It is possible that an average of these parameters would deliver a substantial portion of the observed benefit, allowing for a generic design that does not require individualization. On the other hand, several features varied across participants, including the timing of torque onset and the magnitude of peak torque. This may indicate that customization would provide additional benefits beyond a generic design. Having now identified the features common to Optimized assistance, future research could compare generic and individually optimized assistance to determine the relative benefits of each in this context.

The powered strategy outperformed the spring-like approach, possibly due to differences in the timing and magnitude of peak torque and power. The powered strategy can assist when force production in the user's muscles is most limited: during late stance, when ankle velocity is high and muscle fascicle length and velocity are not optimal (*48*, *49*). This essentially augments human ability; the

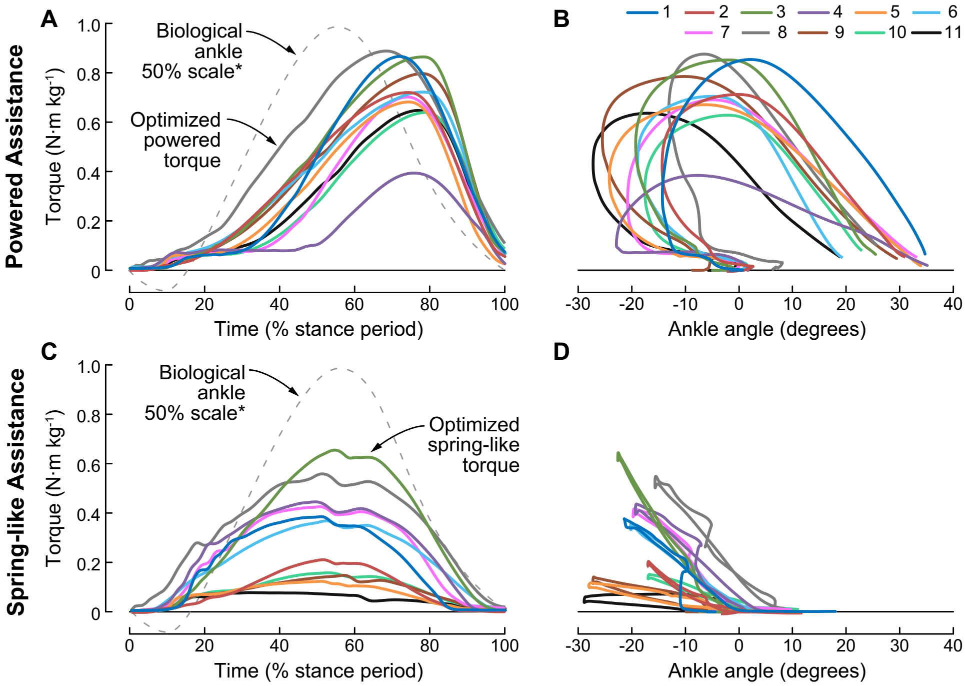

Fig. 3. Optimized assistance patterns applied to both ankles. (**A**) Measured exoskeleton torque patterns resulting from Optimized controller settings for the powered assistance strategy are shown as solid, colored lines for each participant. A 50% scale representation of the torque typically produced by the biological ankle during normal, unassisted running at 2.7 m s^{-1} is shown as a dashed line (*50*). (**B**) Torque versus angle curves show a large amount of work performed by the exoskeletons for all participants under the powered condition. (**C**) Optimized torque patterns for spring-like assistance are shown as colored, solid lines. A 50% scale representation of the torque typically produced by the biological ankle during normal, unassisted running is shown as a dashed line (*50*). (**D**) Torque versus angle curves show minimal work performed on the ankle for all participants under the passive spring-like condition.

powered controller can not only replace torque normally produced by muscles but also produce high torques when plantar flexors cannot. In contrast, the spring-like controller produced torque patterns similar to those normally produced by the biological ankle (*50*), with similar timing of peak torque but much lower torque overall (Fig. 3C).

Although the powered strategy applied larger amounts of net work and higher peak torques than the spring-like controller, Optimized settings did not maximize power input. This suggests that more powerful ankle exoskeletons will not necessarily produce better results, consistent with previous studies demonstrating that additional power is not always helpful (*34*, *51*).

It is unclear why the passive strategy was ineffective. Not only has spring-like assistance at the ankle led to larger improvements in energy economy during walking (*23*), but it also has led to larger improvements during running when applied to the hip (*15*). The lackluster performance of spring-like assistance at the ankle could relate to the fact that, unlike the ankle during slow walking or the hip during moderate-speed running, the ankle performs substantially more positive than negative work during moderate-speed running. This could limit the benefits of spring-like assistance, which must perform equal amounts of positive and negative work. If this hypothesis is true, athletes might benefit more from spring-like assistance at the ankle at higher running speeds, for which positive and negative ankle power are more balanced (*37*). Another possibility is that the spring-like assistance strategy implemented here cannot produce late-stance torques. More general energy-conserving torque profiles that include late peak torque are possible using mechanisms that store, hold, and return energy in more complex ways. Implementation in a real device would be more challenging but could potentially be enabled using electroadhesive clutches to individually engage and disengage multiple parallel springs (*52*). Our results indicate that it is unlikely that spring-like designs will outperform powered ones, but the limits of energetically passive assistance at the ankle during running merit further study.

It is likely that the benefits of both powered and spring-like assistance would be enhanced with additional time for training participants and optimizing exoskeleton characteristics. We based our protocol on a prior experiment in which ankle exoskeleton assistance was optimized during walking (*34*). On this basis, we set a stopping criterion of four generations for the optimization, or about 1 hour of running with each assistance type, before validation. This is similar to the amount of exposure in other experiments demonstrating improved running economy (*15*, *16*, *53*) and was sufficient to achieve large improvements in energy economy with Optimized powered assistance. However, some participants noted that they felt that they were still getting accustomed to the exoskeletons during the validation period. This suggests that more training may have been beneficial. There are also indications that the time-based stopping criterion may have been too brief. For example, variations in Optimized parameter values may have been due in part to the algorithm not yet converging, such that the Optimized parameters are not the globally optimal parameters. These results therefore provide a lower limit on what might be expected with additional training and tuning. Given the disparities in response, however, we would not expect a change in the outcome that powered assistance outperformed spring-like assistance.

The performance of these exoskeletons may be improved further by making the exoskeletons more form-fitting and lighter. A recent study revealed that the metabolic cost of running increases by about 1.11% per each 100 g of weight added per foot (*39*). By this estimate, we expect an increase in metabolic cost of about 9.7% between the zero-torque and control shoe conditions due to the mass of the exoskeletons (0.88 kg each). In our experiment, metabolic rate increased by 14.6% between the control shoe and zero-torque conditions. This higher-than-expected penalty is likely due to the fact that the exoskeletons not only add mass to the legs but also inhibit natural motion. For example, circumduction must be increased substantially to provide clearance between the exoskeletons during swing, which can be costly (*53*).

It remains to be seen whether assisting other joints could be more effective than assisting the ankle joint during running. We have demonstrated larger improvements with ankle assistance than prior studies addressing the hips or knees, but confounds, such as the presence or type of human-in-the-loop optimization, prevent a definitive conclusion about which is most beneficial. Future experiments could compare similarly optimized assistance strategies across joints. In particular, musculoskeletal simulations suggest hip assistance may be most effective at high running speeds (*55*). Although the passive strategy presented here was not effective in assisting the ankle, passive strategies may be more successful when assisting joints that produce near net zero work. For example, the hip provides large amounts of both positive and negative work, exhibiting more spring-like behavior. This may explain why passive hip assistance has proven more effective than this implementation of passive ankle assistance (*14*, *15*).

The results of this experiment have important practical implications, suggesting that portable ankle exoskeletons may be capable of improving running performance using a similar powered assistance strategy. Approximate estimates based on prior work suggest that a portable system with similar capabilities could increase running speed by as much as 10% with no increase in effort (*39*). Although a portable system with similar mass, adequate power, and slimmer fit would be a challenge to design, it is not outside the realm of possibility. Such devices might improve participation in running by decreasing the intensity of running and increasing enjoyment, interest, and perceived ability. Commercial devices could be used as handicaps to allow new joggers to keep up with more physically fit friends. Similar methods could be used to help the elderly maintain independent lifestyles or simply keep up with grandchildren. As designs and control strategies continue to improve, exoskeletons might make soldiers and first responders quicker and less likely to experience fatigue and decreased performance in the field. In the far future, performance-enhancing exoskeletons might even be used in bionic sports in which both engineers and athletes take center stage. It is easy to get carried away with the possibilities at a time when exoskeletons are just starting to hit their stride.

Table 1. Participant characteristics.

Participant	Sex	Mass (kg)	Age (years)	Height (m)
1	M	76.0	24	1.77
2	M	83.9	36	1.83
3	M	72.6	24	1.80
4	M	63.5	22	1.75
5	M	79.4	21	1.83
6	M	79.4	30	1.84
7	M	77.1	21	1.80
8	M	59.4	20	1.75
9	M	70.3	23	1.73
10	M	83.5	37	1.70
11	M	84.4	21	1.80

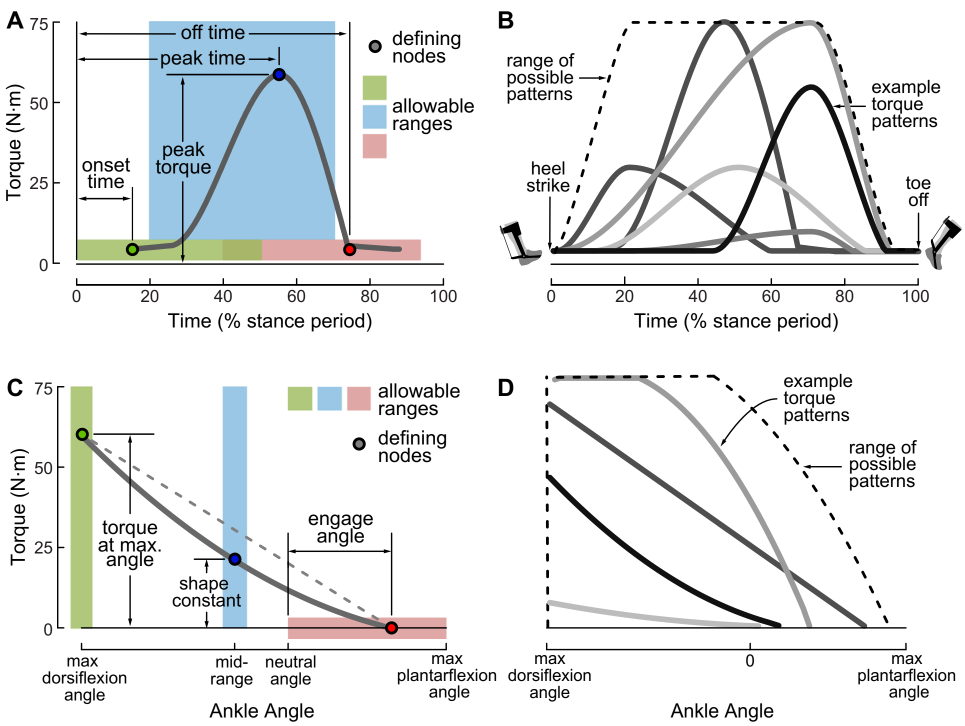

Fig. 4. Controller parameterizations. (**A**) Powered parameterization. Three nodes define the pattern of torque control under the powered condition. These nodes are connected by sinusoids and defined by four parameters: magnitude of peak torque, peak time, onset time, and off time. (**B**) Range and examples of possible powered torque patterns. A simple parameterization results in a wide range of possible torque patterns. (**C**) Spring-like parameterization. A quadratic function fitted to three nodes defines the relationship between ankle angle and applied torque. These nodes are defined by three parameters: the magnitude of torque applied at the maximum dorsiflexion angle; the ankle angle at which the spring engages; and a shape constant that determines the degree to which the spring is linear, stiffening, or softening. (**D**) Range and examples of possible spring-like torque patterns.

MATERIALS AND METHODS

Experimental protocol

Participants wore tethered bilateral torque-controlled ankle exoskeletons (*41*, *56*) with a mass of 0.9 kg each (Fig. 1B). Participants ran at a pace of 2.7 m s^{-1}. Twelve competitive runners were recruited for this study. Participants were required to be natural heel strike runners between the ages of 18 and 40. Eleven participants completed the protocol (age, 25.5 ± 6.1 years; mass, 75.4 ± 8.26 kg; height, 1.78 ± 0.05 m; all male), all of whom had completed at least one marathon and thus could complete the 1-hour optimization sessions without fatiguing or exceeding their aerobic threshold. Participant characteristics are detailed in Table 1. One participant did not converge during the first optimization day and was not interested in participating further; this participant's data were discarded. All participants provided written informed consent before participation, after the nature and possible consequences of the study were explained. The study protocol was approved and overseen by the Institutional Review Board of Carnegie Mellon University.

Participants experienced a day of habituation in which they were introduced to both controllers. Participants were allowed 5 min of running in zero-torque mode, followed by 5 min of running in a nonoptimized assistive mode. They then experienced one generation of trial controllers. The process was repeated for the second assistive mode. The order in which participants experienced the powered and spring-like assistive modes was randomized. Participants then experienced a separate day of optimization for each controller, described in the "Optimization strategy" section below.

A day of validation trials was experienced after optimization of both controllers. The following conditions were tested: zero-torque, Optimized spring-like, Optimized powered, and control shoes (Nike Zoom Pegasus 32). Validation trials were performed in randomized double-reversed order (ABCDDCBA). The trials using running shoes were predetermined to occur either as the first and last trials (A) or back to back as the middle conditions (D) to minimize the number of times the exoskeletons needed to be put on and taken off. Otherwise, the assignment of each condition to the A, B, C, or D position was randomized. The shoes used under the control shoe condition were identical to the shoes incorporated into the exoskeletons. Five minutes of rest were allowed between each validation trial, which was sufficient for metabolic rate to return to values close to quiet standing. Validation trials were 6 min in length, consistent with common practice, and the last 3 min of data were used to calculate measured outcomes.

Controller parameterizations

Powered assistance was inspired by a time-based approach used for walking assistance (*34*). Desired torque patterns for the powered strategy were defined by four parameters: peak torque magnitude (peak torque) and the timing of peak torque (peak time), onset of torque (onset time), and return to zero torque (off time). These four parameters define three nodes connected by sinusoids shown in Fig. 4A. Example patterns within the allowable parameter ranges are shown in Fig. 4B. Maximum torque was allowed to vary between 4 and 60 N·m, with 4 N·m being the minimum torque setting that could be accurately tracked by the emulator system and 60 N·m being the maximum comfortable torque setting found during pilot testing. Maximum torque was adjusted to 75 N·m for two participants who showed a preference for high torques during habituation. The timing of peak torque could be varied between 20 and 80% of average stance time. Onset time was constrained to be between 0 and 50% of stance time with no less than 20% of stance time between the onset of torque and peak torque. Off time ranged between 40 and 98% of stance time with no less than 20% of stance time between peak time and off time. Ninety-eight percent of stance time was selected as an upper bound to prevent plantar flexion torques being applied during swing. These ranges and constraints were set on the basis of comfort in pilot tests.

Average stride time was calculated as a running average of the time between heel strikes of the same foot, detected by pressure switches in the heel of the shoe. Left and right ankle exoskeletons were controlled in isolation of each other, although controller settings were shared across both.

Passive spring-like assistance was applied as a function of ankle angle and was defined by three parameters: (i) the torque experienced at the maximum dorsiflexion angle (maximum torque); (ii) the ankle angle at which the spring becomes engaged (engagement angle); and (iii) a shape constant that determines the degree to which the spring is linear, stiffening, or softening. These three parameters combined to define three nodes as shown in Fig. 4C. A quadratic function was fit to these three nodes to define torque as function of ankle angle. Example patterns within the allowable range of these parameters are shown in Fig. 4D. Maximum torque ranged from 4 to 60 N·m (or 75 N·m for two participants). Maximum torque was also used as a threshold in cases where spring-like torque would exceed this limit, for example on steps with greater than normal dorsiflexion. The shape constant set the magnitude of torque experienced at midrange. Midrange was defined as the angle halfway between the maximum plantarflexion angle and maximum dorsiflexion angle. For the purposes of the parameterization, maximum plantarflexion and maximum dorsiflexion were defined as the maximal angles measured during a 5-min trial of running in zero-torque mode before optimization. The last node, engagement angle, could be varied between the maximum plantar flexion angle and midrange.

Maximum expected dorsiflexion angle was calculated as the 75th percentile of the maximum dorsiflexion angles measured over 32 strides for each participant. Maximum expected plantar flexion was set to the average of the maximum plantar flexion angles measured over 32 strides. This ensured that the spring engaged and disengaged on every step despite small inter-step variations.

Low-level torque control for both assistance strategies was enabled through a combination of proportional control, damping injection, and iterative learning (*57*). Torque tracking was similar between the powered and spring-like control strategies. The root mean square error was 0.80 ± 0.55 N·m for the powered condition and 0.92 ± 0.50 N·m for the spring-like condition.

Optimization strategy

Human-in-the-loop optimization (*34*) was applied to both assistance strategies using a covariance matrix adaptation evolutionary strategy (CMA-ES) (*58*). Normal distributions for each parameter were created with means and variances defined as initial settings. Parameters were selected from these distributions to form a generation of trial controllers. The number of trial controllers included in a generation, λ, was calculated as a function of n, the number of parameters being optimized (Eq. 1). The powered controller was optimized using generations consisting of eight trial controllers. The passive controller was optimized using generations of seven trial controllers.

$$\lambda = 4 + \lfloor 3 \ln(n) \rfloor \tag{1}$$

Each of the trial controllers was applied over 2 min of running during which metabolic rate was measured breath by breath. Steady-state metabolic cost was estimated by fitting a first-order dynamical model (*59*) to the metabolic data collected using a respirometry system during each of the 2-min trials. A first-order model well approximates oxygen uptake dynamics during low to moderate intensity activities (*60*), such as moderate-speed running performed by the high fitness runners enrolled in this study. A retrospective analysis (fig. S1) suggests that the error in these estimations was about 12%, which is higher than in our prior study (*34*) but apparently sufficient for optimization based on outcome rank rather than absolute values. These data suggest that a faster time constant in the first-order model could reduce estimation error to about 6% in future studies.

A covariance matrix adaptation (CMA) was then performed to calculate new means and variances for each of the parameter distributions (*58*). These new distributions were then used to select the trial controllers for the next generation. On the basis of our prior study (*34*), we used number of objective function evaluations as the stopping criteria; each participant experienced four generations of trial controllers during the optimization period for both the powered and spring-like controllers. The mean parameter values calculated for the fifth generation were considered the Optimized controller settings. Refer to (*34*) for more details on this optimization approach.

Measured outcomes

For purposes of comparison, all measured outcomes were normalized to body mass.

Net metabolic rate

Net metabolic rate was the primary outcome of this study. Volumetric carbon dioxide expulsion, oxygen consumption, and breath duration were measured on a breath-by-breath basis using respirometry equipment (Quark CPET, COSMED). The respirometry system was calibrated according to the manufacturer specifications using gasses of known oxygen and carbon dioxide concentration and simulated breaths of known volume. Experiments were conducted in a large, well-ventilated room. Data from the last 3 min of each validation trial were substituted into a standard equation (*61*) to calculate average metabolic rate. Steady state was reached in the final 3 min of each 6-min trial, indicated by a negligible time rate of change in energy cost (0.09 ± 0.20 W s^{-1} across all trials). The net metabolic rate of running under each condition was calculated as the average metabolic rate minus the rate for quiet standing. The reported result for each condition is the average calculated from the double-reversed trials. These average values were used to determine the statistical significance of metabolic reductions by performing paired, two-tailed *t* tests.

Average exoskeleton work

Mechanical work delivered by the exoskeleton was calculated for each step by integrating the exoskeleton torque over ankle angle during the stance period.

Average exoskeleton power

Exoskeleton power was calculated for each step by dividing the work delivered during foot contact by stride time. Average power was calculated by summing the power delivered for all steps and dividing by the number of steps occurring in the last 3 min of data.

SUPPLEMENTARY MATERIALS

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Fig. S1. Accuracy of steady-state metabolic rate estimates during optimization.

Table S1. Values of Optimized parameters for each participant.

Table S2. Values of metabolic rate (watts per kilogram) for each participant under each condition.

Movie S1. Video showing a participant running with exoskeletons in Optimized powered, Optimized spring-like, and zero-torque modes and with running shoes.

Reference (*62*)

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