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# A little bit faster: Lower extremity joint kinematics and kinetics as recreational runners achieve faster speeds

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## ABSTRACT

There appears a linear relationship between small increases in running speed and cardiovascular health benefits. Encouraging or coaching recreational runners to increase their running speed to derive these health benefits might be more effective if their joint level kinematic and kinetic strategy was understood. The aim of this investigation was to compare the peak sagittal plane motions, moments, and powers of the hip, knee and ankle at 85%, 100%, 115% and 130% of self-selected running speed. Overground running data were collected in 12 recreational runners (6 women, 6 men) with a full body marker set using a 12-camera Vicon MX system with an AMTI force plate. Kinematics and kinetics were analyzed with Vicon Nexus software. Participants chose to run at  $2.6 \pm 0.5$  m/s (85%);  $3.0 \pm 0.5$  m/s (100%);  $3.3 \pm 0.5$  m/s (115%); and  $3.7 \pm 0.5$  m/s (130%); these four speeds approximately correspond to 6:24-, 5:33-, 5:03-, and 4:30-min kilometer running paces. Running speed had a significant effect ( $P < 0.05$ ) on peak kinematic and kinetic variables of the hips, knees and ankles, with peak sagittal hip moments invariant ( $P > 0.54$ ) and the peak sagittal ankle power generation ( $P < 0.0001$ ) the most highly responsive variable. The timing of the peak sagittal extensor moments and powers at the hip, knee and ankle were distributed across stance in a sequential manner. This study shows that running speed affects lower limb joint kinematics and kinetics and suggests that specific intersegmental kinetic strategies might exist across the narrow range of running speeds.

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## 1. Introduction

Running is a primary locomotor mode for humans and is often used for sport, recreation or fitness training. Running for fitness is a popular activity, with more than 23 million individuals reporting that they run at least 50 days a year (Lamppa, 2009). The cardiopulmonary fitness needed to maintain a particular running speed is ultimately dictated by the energetic cost of producing the muscular contractions for each stride. The muscular effort can be shared among the hips, knees and ankles in different ways to achieve small increases in running speed.

Training is required to increase one's cardiovascular fitness to run the same distance at a higher running speed. Coaching recreational runners to increase their running pace to reap increased health related training effects might be more effective if the joint loading strategy was understood. Modulating the kinematic and kinetic contributions of the hips, knees and ankles to maintain a particular running speed may have a systematic strategy that relies on one joint over others as individuals maintain the faster running speed. This has been the case for walking at faster speeds, where the hip may contribute proportionally greater amounts of the work needed to maintain faster walking speeds (Farris and Sawicki, 2012).

There are several publications quantifying the kinetic contributions of the hips, knees and ankles as individuals go from maintaining jogging speeds ( $\sim 3$  m/s;  $\sim 6.7$  mph) to maintaining sprinting speeds ( $\sim 9$  m/s;  $\sim 20$  mph) (Chumanov et al., 2007; Dorn et al.,

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2012; Kivi et al., 2002; Mann and Hagy, 1980; Mero et al., 1992; Schache et al., 2011; Weyand et al., 2010), providing some information that coaches might use as the basis for their training strategies for elite athletes. However, little information is available on how recreational runners modulate their lower extremity joint kinematics and kinetics to maintain relatively small increases in running speed ( $\sim 2$  m/s to  $\sim 3$  m/s;  $\sim 4.5$  mph to  $\sim 6.7$  mph) that appear to have substantial effects on health (Lee et al., 2014). It might be that encouraging individuals to try longer strides and lower cadence, or shorter strides and higher cadence might increase running speed, or perhaps greater ankle or hip effort would aid them in maintaining slightly faster running speeds to reap increased health benefits.

The aim of this investigation was to compare the peak sagittal plane motions, moments, and powers of the hip, knee and ankle at 85%, 100%, 115% and 130% of self-selected running speed in individuals who are recreationally active. It was hypothesized that running speed would have a significant effect on peak values of the sagittal motions, moments and powers of the hip, knee and ankle.

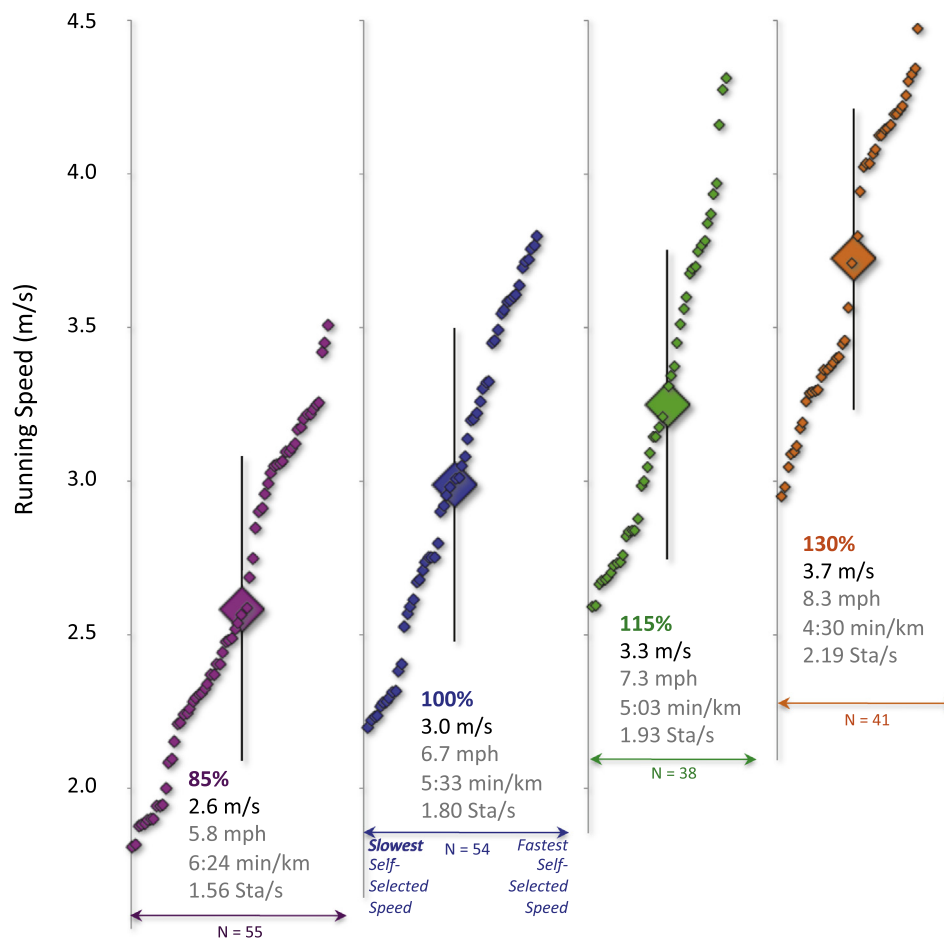
## 2. Methods

Twelve recreationally active runners (6 women, 6 men) gave their informed consent to participate in this IRB-approved protocol ( $36.6 \pm 12.4$  years of age; Height  $171.8 \pm 8.2$  cm; Weight  $67.4 \pm 14.0$  kg). Inclusion criteria were 18–60 years of age, running

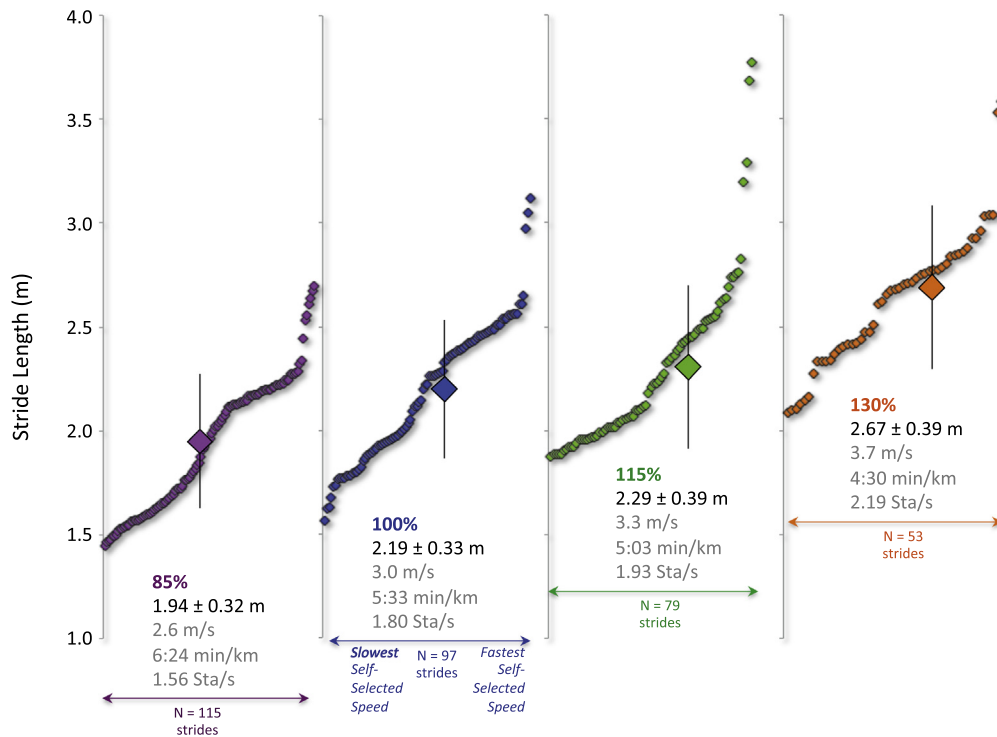
between 10 and 30 miles (16.09 and 48.28 km) per week run, with no lower extremity musculoskeletal injury that reduced running mileage for the past 6 months. Each runner came to the laboratory and changed into tight fitting shorts and top to improve marker placement accuracy. Thirty-eight spherical markers were placed on the participant's head, arms, trunk, legs and shoes according to the Vicon full-body Plug-In-Gait model (Oxford Metrics, Denver, CO). Markers were placed on each participant by a single researcher with more than ten years of marker placement experience to ensure the highest level of consistency across participants.

Participants were given an opportunity to warm up with light jogging and stretching. Participants' self-selected running speed was determined using a treadmill. Immediately following the treadmill running, kinematic and kinetic data were collected on the participants during overground running at four speeds ( $130 \pm 5\%$ ,  $115 \pm 5\%$ ,  $100 \pm 5\%$ , and  $85 \pm 5\%$  of self-selected running speed). A 12-camera Vicon MX system captured marker trajectories (250 Hz) and integrated ground reaction forces were collected (3000 Hz) using a flush-mounted AMTI OR6-6 force platform (AMTI, Watertown, Massachusetts, USA) in the center of the volume.

Runners performed each trial by running through a laboratory that was 20 m long and exiting into a hallway. Runners were instructed to achieve the desired running speed before entering the motion capture volume (8 m lead up), to maintain that speed throughout the motion capture volume (8.5 m long  $\times$  3 m wide  $\times$  2.2 m high), and strike a single force plate located in the center



**Fig. 1.** Running speeds chosen by the participants at 85%, 100%, 115% and 130% of preferred speed. The preferred speed is 100%; values for meters per second (m/s), miles per hour (mph), minutes per kilometer (min/km) and height-normalized strides per second (Sta/s) are also shown for comparison. Large diamonds represent the mean value within a running speed, vertical bars around the mean are standard deviations for each running speed; small diamonds represent individual subject's trials.

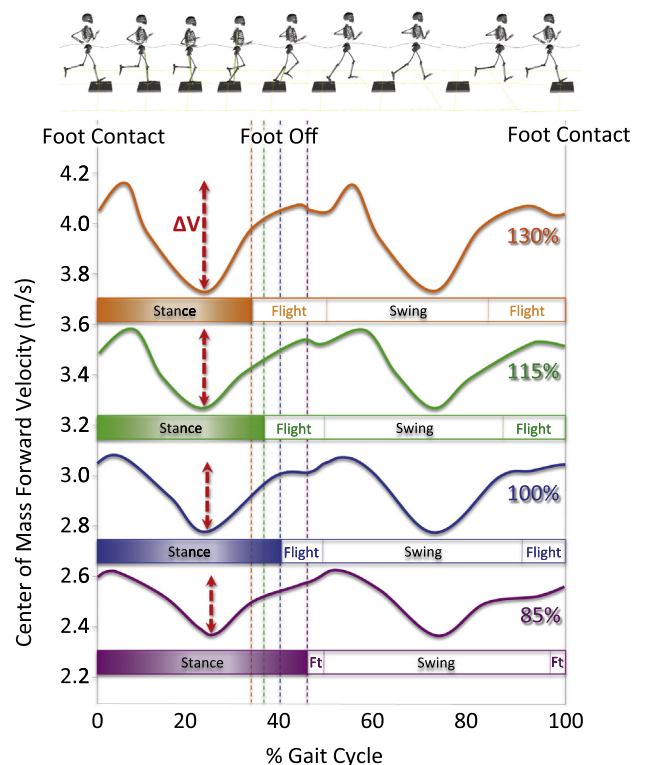


**Fig. 2.** Stride length in meters (m) under 85%, 100%, 115% and 130% of self-selected running speed. The preferred running speed is 100%; values for meters (m), meters per second (m/s), minutes per kilometer (min/km) and height-normalized statures per second (Sta/s) are also shown for comparison. Large diamonds represent the mean value within a running speed, vertical bars around the mean are standard deviations for each running speed; small diamonds represent individual subject's trials. A statistically significant main effect was observed for stride length across the four running speeds ( $P < 0.0001$ ).

of the volume without targeting. The participants maintained their running speed as they exited the laboratory. Runners began by attempting to run at 130% of self-selected running speed and then progressed to the goals of 115%, 100% and 85% of self-selected running speed. This non-random approach was chosen to prevent fatigue effects on joint kinematics and kinetics by minimizing the total amount of time spent in the laboratory ( $\leq 2$  h).

The runner was provided verbal feedback on the speed of each trial soon after it was completed based on sacral marker forward velocity. The forward velocity of the sacral marker follows a sinusoidal pattern and the mean value was calculated around this sinusoidal pattern; trials with sinusoids that trended upward (accelerating) or downward (decelerating) were discarded, and trials with three consistent sinusoids prior to foot contact on the force plate and two following the foot contact were retained. Trials were repeated until the participant had achieved five clean force plate strikes on one limb at each running speed or they had completed 30 attempts at that speed; after this the next speed was attempted. This continued until all speeds were completed or until 2 h had elapsed, whichever came first.

After each running trial through the motion capture volume, participants walked back to the starting point and were allowed rest as needed between trials; all runners were ready to begin the next trial within 5 s of reaching the starting location. Runners did not report being tired and did not appear winded due to the short running distance and the period of recovery walking back to the starting location. Often runners missed the target speed, but achieved a different target speed with clean force plate contact, and these trials were retained. Trials that did not have clean force plate contact, did not have constant speed based on sacral marker forward velocity, or did not fit into one of the four speed categories ( $\pm 5\%$ ) were discarded. Therefore each subject contributed a unique number of trials for each desired speed. Each runner had an estimated two-thirds of their total attempts deemed unacceptable



**Fig. 3.** Center of mass (COM) forward velocity across four running speeds for a representative individual. Change in forward velocity ( $\Delta V$ ) is indicated by vertical red dashed arrows. COM forward velocity is plotted over the gait cycle from foot contact to foot contact (% Gait Cycle). Stance and swing are separated by dashed vertical lines indicating foot-off; each line is color-coded to each speed (130% Orange; 115% Green; 100% Blue; 85% Purple). Note that stance duration decreases at faster running speeds, but flight time (Flight or Ft) increases.

due to changes in speed while in the volume, missing the force plate or being over 135% of self-selected running speed.

Data from each subject was processed in Vicon Nexus 1.5 software (Vicon, Centennial, Colorado, USA) with marker trajectories smoothed using a Woltring spline filter set at 20 MSE. Three-dimensional kinematic and kinetic data were calculated and plotted for each individual. Center of mass position was calculated within Nexus software using segment mass and inertial parameter data from Dempster (Dempster, 1955). Data were extracted from processed trials using the Parameter Calculator Plug-In (Vaquita Software, Zaragoza, Spain) in the Nexus pipeline to ensure transcription error-free data reduction. Kinetic variables were normalized to body weight. It was hypothesized that running speed would have a significant effect on parameters including: stride length (m); peak sagittal ankle angle (motion) (deg), moment (Nm/kg), and power (W/kg); peak sagittal knee angle (motion) (deg), moment (Nm/kg), and power (W/kg); peak sagittal hip angle (motion) (deg), moment (Nm/kg), and power (W/kg); the timing of the peak moments (% gait cycle) at the hip and ankle; and the timing of the peak powers (% gait cycle) at the hip and knee. Unbalanced repeated measures ANOVAs were used to determine statistically significant changes in these variables with running speed (Statview, SAS, Cary NC). A statistically significant level was set at  $P < 0.05$ .

### 3. Results

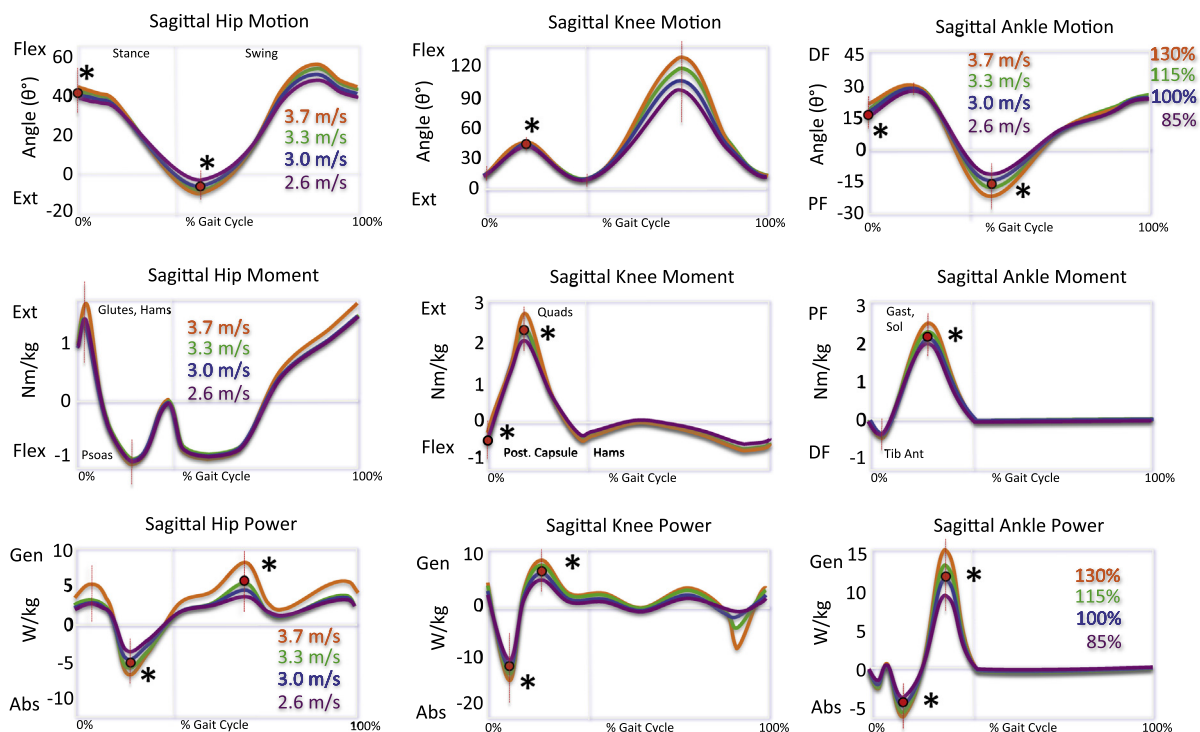
Participants chose to run at  $2.6 \pm 0.5$  m/s (85%);  $3.0 \pm 0.5$  m/s (100%);  $3.3 \pm 0.5$  m/s (115%); and  $3.7 \pm 0.5$  m/s (130%); these four speeds approximately correspond to 6:24-, 5:33-, 5:03-, and 4:30-min kilometer running paces (10:20-, 9:00-, 8:15-, and 7:30-min mile running paces). This translates to participant

height-normalized speeds of  $1.56 \pm 0.36$  statures/s,  $1.80 \pm 0.41$  statures/s,  $1.93 \pm 0.24$  statures/s, and  $2.19 \pm 0.26$  statures/s at 85%, 100%, 115% and 130% of self-selected running speed, respectively (Fig. 1).

All runners chose a rearfoot or midfoot contact pattern at these speeds based on the observation of negative value in the sagittal ankle moment at initial contact. Running speed had a significant effect on stride length ( $P < 0.0001$ ; Fig. 2). Forward velocity of the center of mass (COM) varied in a sinusoidal pattern during each stance phase with maxima observed at 5–10% of the gait cycle and minima observed at 22–27% of the gait cycle (Fig. 3). The change in forward velocity ( $\Delta V$ ) was substantial and represented 10–12% of running speed within each stride.

Statistically significant main effects were observed in the joint kinematics and kinetics at the hip, knee and ankle with running speed ( $P < 0.05$ ). See Fig. 4 for overall results, and Figs. 5, 6 and 7 for detailed kinematics and kinetics of the hip, knee and ankle, respectively. At the hip, running speed had a significant effect on peak sagittal hip flexion, though it increased only about  $1^\circ$  at each faster speed, in early stance ( $P < 0.0001$ ) and peak sagittal hip extension in early swing ( $P < 0.0001$ ), which increased about  $4.5^\circ$  between 85% and 130% of self selected running speed (Fig. 5). Contrary to the hypothesis, running speed did not have a significant effect on peak sagittal hip extensor moments (early stance) ( $P = 0.9413$ ) and peak sagittal hip flexor moments (mid-stance) ( $P = 0.5353$ ). Running speed had a significant effect on peak sagittal hip power absorption in mid-stance ( $P < 0.0001$ ) and peak sagittal hip power generation in swing phase ( $P < 0.0001$ ) (Fig. 5).

At the knee, running speed did not have a significant effect on peak sagittal knee flexion at initial contact ( $P = 0.3754$ ), but it had a significant effect on peak sagittal knee flexion in mid-stance ( $P = 0.0009$ ) (Fig. 6). Peak sagittal knee flexion in swing was highly variable between participants, and despite showing



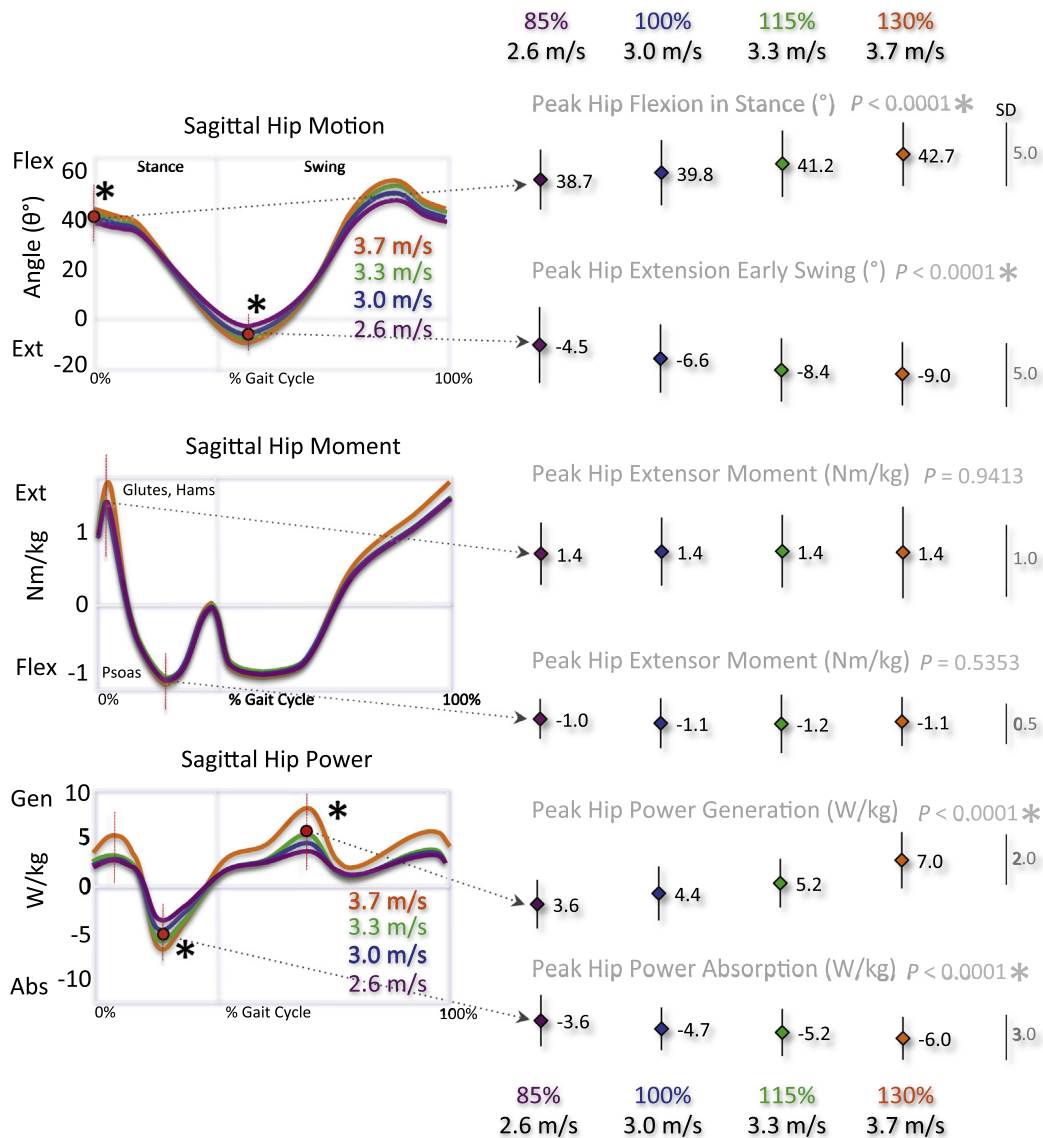
**Fig. 4.** Mean sagittal joint motions (degrees;  $^\circ$ ), moments (Newtonmeters per kilogram of body weight; Nm/kg) and powers (Watts per kilogram of body weight; W/kg) for the hip, knee and ankle plotted across the time-normalized running gait cycle (foot contact 0% to foot contact 100%). Maxima and minima are indicated by red vertical lines, and asterisks indicate a statistically significant main effect of running speed ( $P < 0.05$ ). The vertical line that separates stance from swing is presented only for the 100% of preferred running speed.

increases in peak sagittal flexion in the mean value across speeds, no statistically significant main effect was detected ( $P = 0.61$ ). Running speed had a significant effect on peak sagittal knee flexor moments, though magnitudes at each speed were very small at initial contact ( $P = 0.0462$ ), and a significant effect on peak sagittal knee extensor moments in stance phase ( $P = 0.0006$ ). Running speed also had a significant effect on peak sagittal knee power absorption in early stance ( $P = 0.0003$ ) and peak sagittal knee power generation in mid-stance ( $P = 0.0007$ ).

At the ankle, running speed had a significant effect on ankle dorsiflexion at initial contact ( $P = 0.0169$ ), but peak dorsiflexion in mid-stance was similar across running speeds (Fig. 7). Running speed also had a significant effect on peak plantarflexion in early swing ( $P < 0.0001$ ) with almost a  $5^\circ$  increase between the slowest and fastest speeds. No significant effect was detected in the peak sagittal ankle dorsiflexor moment in early stance with running speed ( $P = 0.5847$ ), but running speed had a significant effect on peak sagittal ankle plantarflexor moment ( $P < 0.0001$ ). Running

speed also had a significant effect on ankle power absorption ( $P < 0.0001$ ) and peak ankle power generation in late stance ( $P < 0.0001$ ).

The timing of minima and maxima sagittal moments appeared to show temporal alignment twice in stance phase at the hip and ankle (Fig. 8). At the hip, running speed did not have a significant effect on the timing of the early stance peak sagittal hip extensor moment ( $P = 0.8395$ ) with a mean value across all speeds  $5.3 \pm 2.9\%$  gait cycle. At the ankle, running speed did not have a significant effect on the timing of the early stance peak sagittal dorsiflexor moment ( $P = 0.3772$ ) with a mean value across all speeds of  $3.5 \pm 1.5\%$  gait cycle. For all speeds, the difference in timing between the maximum hip extensor moment and the minimum ankle dorsiflexor moment was less than  $1.9\%$  gait cycle. In mid-stance, running speed did not have a significant effect on the peak sagittal hip flexor moment timing ( $P = 0.2265$ ) with a mean value of  $20.2 \pm 4.8\%$  gait cycle and the timing of the peak sagittal plantarflexor moment ( $P = 0.0531$ ) with a mean value across running



**Fig. 5.** Mean sagittal hip motion (degrees; °), moment (Newtonmeters per kilogram body weight; Nm/kg) and power (watts per kilogram of body weight; W/kg) across four running speeds, in meters per second (m/s): 85%, 100%, 115% and 130% of preferred running speed (100%). Curves show the mean pattern plotted over the gait cycle from foot contact (0%) to foot contact (100%); point graphs show mean values and vertical bars are  $\pm$ standard deviation at identified minima and maxima. Standard deviation bars are shown for each point graph. Asterisks mark statistically significant differences in maxima and minima across running speeds ( $P < 0.05$ ).

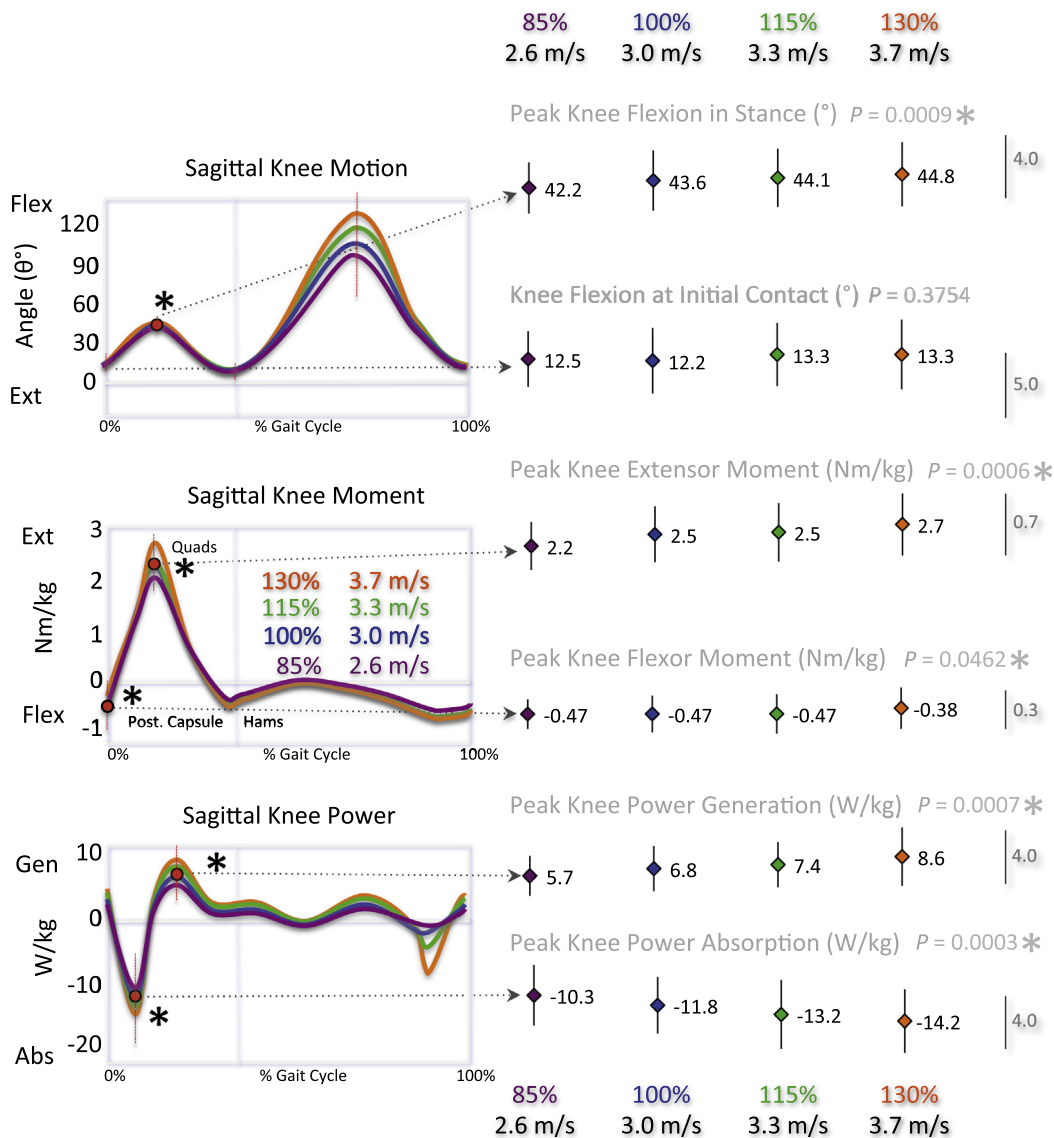
speeds of  $20.8 \pm 6.6\%$  gait cycle. For all speeds, the difference in timing between the peak sagittal hip flexor moment and the peak sagittal ankle plantarflexor moment was less than 0.8% gait cycle.

The timing of minima and maxima sagittal power appears to show temporal alignment twice in stance phase at the hip and knee (Fig. 9). At the hip, running speed did not have a significant effect on the timing of early stance peak sagittal hip power generation ( $P=0.0868$ ) with the mean value across all speeds  $8.76 \pm 5.09\%$  gait cycle. At the knee, running speed did not have a significant effect on the timing of peak sagittal knee power absorption in early stance ( $P=0.0792$ ) with the mean value at all speeds  $8.01 \pm 2.53\%$  gait cycle. For all running speeds, the timing difference in hip power generation maximum and knee power absorption minimum was less than 1.7% gait cycle. In mid-stance, running speed did not have a significant effect on the timing of peak sagittal hip power absorption ( $P=0.2227$ ) with the mean value across all speeds  $21.9 \pm 5.9\%$  gait cycle and the timing of peak sagittal knee power generation ( $P=0.1148$ ) with the mean value across all speeds  $19.1 \pm 5.0\%$  gait cycle. The maximum timing

difference across all speeds was 3.2% gait cycle between maximum hip power generation and minimum knee power absorption. (It should be noted that averaging all participant's data into a mean curve does not produce the same peak timing as averaging each individual's timing of the peaks (Dames et al., 2017). Therefore the peaks in Fig. 9 may not appear precisely aligned with these statistically determined timing values).

#### 4. Discussion

The aim of this study was to quantify the changes in the sagittal motion, moments and powers at the hip, knee and ankle in response to maintaining four different running speeds. The participants chose a somewhat slow preferred running speed,  $3.0 \pm 0.5$  m/s, about a 9:00-min mile (5:33-min kilometer) pace as the baseline. Even with the relatively small changes in speed at 85%, 115% and 130% of self selected speed, there were statistically significant changes that reveal certain kinematic and kinetic patterns at the hip, knee and ankle. Previous work has demon-



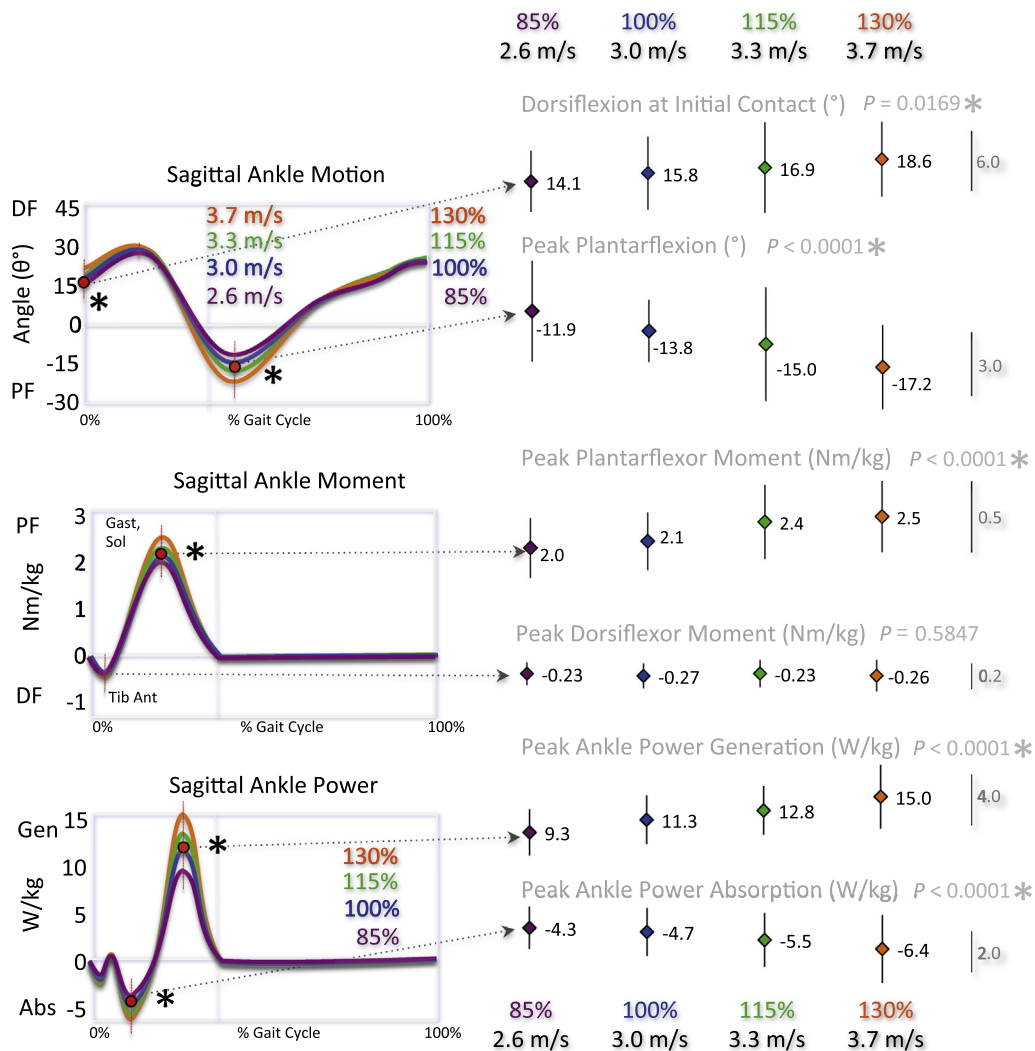
**Fig. 6.** Mean sagittal knee motion (degrees; °), moment (Newtonmeters per kilogram body weight; Nm/kg) and power (watts per kilogram of body weight; W/kg) across four running speeds, in meters per second (m/s): 85%, 100%, 115% and 130% of preferred running speed (100%). Curves show the mean pattern plotted over the gait cycle from foot contact (0%) to foot contact (100%); point graphs show mean values and vertical bars are  $\pm$ standard deviation at identified minima and maxima. Standard deviation bars are shown for each point graph. Asterisks mark statistically significant differences in maxima and minima across running speeds ( $P < 0.05$ ).

stated that the metabolic efficiency of the muscles of the lower extremity do not have a single running speed where all are at a minimum (Carrier et al., 2011), suggesting that a different balance of kinetic contributions at the hip, knee and ankle might be preferred to maintain faster running speeds. Ultimately, the metabolic cost of running at each speed is based on the energy cost of the muscle contractions that control the joint kinematics and kinetics. Several fundamental characteristics of running that are altered while running at faster speeds have been determined from this study. In addition, temporal relationships of several peak sagittal plane kinetic values did not appear to change at faster running speeds.

The changes in center of mass forward velocity at each running speed suggest that substantial mechanical and therefore metabolic energy is used decelerating and accelerating the body with each step during running ( $\Delta V$ ; Fig. 3). This may be an inescapable fact for any bipedal bouncing gait such as human running (Farley and Ferris, 1998). Simply reducing the  $\Delta V$  at any given running speed may theoretically improve running economy, but additional research effort is needed to support this conjecture. Modeling studies of human running (Hamner and Delp, 2013) suggest that the quadriceps is primarily responsible for the impulse that slows

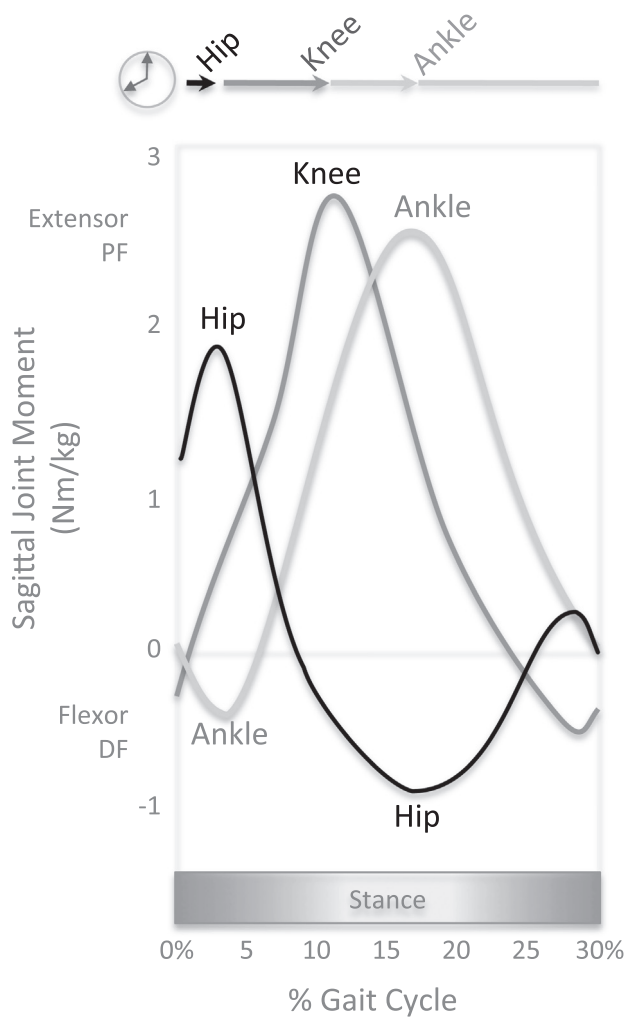
the center of mass in early stance, however other modeling work suggests that the quadriceps plays primarily a support role against gravity (Dorn et al., 2012); small differences in muscle activation algorithms and tendon slack length choices between these modeling approaches may account for these starkly different modeling results. Other possible explanations for the increase in center of mass  $\Delta V$  at faster running might simply be the cosine losses of the sinusoidal center of mass trajectory: larger vertical center of mass excursions (that are likely to be related to longer stride lengths) at faster running speeds result in larger changes in center of mass forward velocity. The joints of the lower extremity are assumed to interact to apply forces to the ground at faster running speeds that create the center of mass velocity changes observed in this study. The interactions of the peak kinematic and kinetic variables of the hip, knee and ankle also reveal timing aspects that suggest intersegmental synergy patterns that appear consistent at all running speeds (Figs. 8 and 9).

The timing of the peak sagittal extensor moments at the hip, knee and ankle were distributed across stance in a sequential fashion (Fig. 4). This pattern is consistent with work by others across a range of much faster running speeds (Dorn et al., 2012; Schache et al., 2011). The peak sagittal extensor moments at the hip



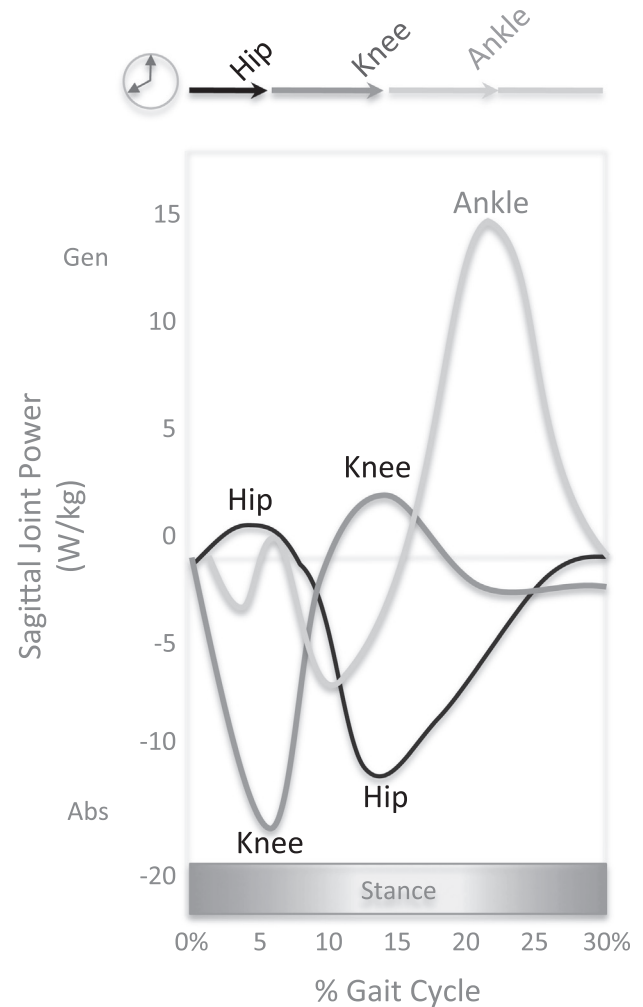
**Fig. 7.** Mean sagittal ankle motion (degrees; °), moment (Newtonmeters per kilogram body weight; Nm/kg) and power (watts per kilogram of body weight; W/kg) across four running speeds, in meters per second (m/s): 85%, 100%, 115% and 130% of preferred running speed (100%). Curves show the mean pattern plotted over the gait cycle from foot contact (0%) to foot contact (100%); point graphs show mean values and vertical bars are  $\pm$ standard deviation at identified minima and maxima. Standard deviation bars are shown for each point graph. Asterisks mark statistically significant differences in maxima and minima across running speeds ( $P < 0.05$ ).

Peak extensor moments are sequential from proximal to distal



**Fig. 8.** Group mean sagittal moments (Newton meter per kilogram body weight; Nm/kg) of the hip, knee and ankle plotted across stance phase during running at self-selected pace (3.0 m/s; 6.7 mph; 9:00 min/mile). Note the sequential pattern from proximal to distal joints – 1. Hip, 2. Knee, 3. Ankle. Maximum and minimum sagittal moments at the hip and ankle appear to be temporally linked, with the peak hip extensor moment corresponding with the peak ankle dorsiflexor moment (DF), and the peak hip flexor moment corresponding with the peak ankle plantarflexor moment (PF). There was no statistically significant main effect across speeds for the timing of the minima and maxima of moments at the hip or ankle ( $P > 0.05$ ).

Peak power generation is sequential from proximal to distal



**Fig. 9.** Group mean sagittal powers (Watts per kilogram body weight; W/kg) of the hip, knee and ankle plotted across stance phase during running at self-selected pace (3.0 m/s; 6.7 mph; 9:00 min/mile). Note the sequential pattern from proximal to distal joints – 1. Hip, 2. Knee, 3. Ankle. Sagittal power at the hip and knee also appear to be temporally linked, with the peak hip power generation (Gen) corresponding with the peak knee power absorption (Abs), and the peak hip power absorption (Abs) corresponding with the peak knee power generation (Gen). There was no statistically significant main effect across speeds for the timing of the minima and maxima of the hip and knee powers ( $P > 0.05$ ).

occurred first at about 4% of the gait cycle, the knee at about 11% of the gait cycle, and the ankle at about 17% of the gait cycle (Fig. 8). These moments represent sequential efforts by the hip extensors, knee extensors, and ankle plantarflexors to move each joint from flexion to extension during the stance phase of running. Similar proximal to distal peak moment patterns have been observed in human jumping (Bobbert and van Soest, 2001). During running, the data from the current study demonstrate that sagittal moments of the hip and ankle appear to be temporally linked: peak sagittal hip extensor moments occurred in close time proximity to peak sagittal ankle dorsiflexor moments in loading response (Fig. 8) consistent with previous investigations (Dorn et al., 2012; Hamner and Delp, 2013; Schache et al., 2011). In late stance, peak sagittal ankle plantarflexor moments occurred in close time proximity to peak sagittal hip flexor moments in mid-stance. Peak sagittal powers

of the hip, knee and ankle also appear to have a sequential proximal to distal order but a different temporal linkage (Fig. 9). Peak sagittal power generation of the hip occurred first at about 6% of the gait cycle, followed by the knee at about 15% and the ankle at about 23% (Fig. 9) consistent with previous modeling work on running (Hamner and Delp, 2013). The hip and knee appear to be temporally linked with peak sagittal power generation at the hip corresponding to peak sagittal knee power absorption in early stance; then, peak sagittal knee power generation corresponded with peak sagittal hip power absorption in midstance. These data suggest a concentric-eccentric hip and knee coupling during the first half of stance phase in running, consistent with operating the hip, knee and ankle in an integrated strategy. This temporal linkage may explain why spring-mass models of running have been so effective in explaining the movement of the center of mass



(Farley and Ferris, 1998; Farley and Gonzalez, 1996; Ferris et al., 1999). However, further statistical analyses are required to explore whether joint moments and powers are truly temporally linked.

Power generation at the ankle in late stance occurs while the knee and hip have low power output, as these more proximal joints reach their end range of extension just prior to foot-off. This suggests that push-off energy developed at the ankle joint is primarily applied to the center of mass although some energy may aid in flexing the hip and knee as swing begins after foot-off as in walking (Siegel et al., 2004). The kinetic timing interactions do not appear substantially altered to maintain faster running speeds ( $P > 0.05$ ) supporting the concept of similar intersegmental kinetic interactions across the narrow range of running speeds included in this study. Future work is needed to confirm this observation on this small cohort of recreational runners.

The most unexpected result was that the running speed did not have a significant effect on the peak sagittal moments of the hip (Fig. 5). This may be attributed to a movement strategy by the recreational runners in this study that results in increased co-contraction of the hip flexors (iliopsoas) and extensors (gluteals, hamstrings) at initial contact: the ground reaction force vector oscillates rapidly between flexing and extending the hip during loading response in running, requiring rapid oscillations of internal muscle moments that extend or flex the hip. Therefore a co-contraction to simply stiffen the joint at initial contact may be the “lowest cost” control strategy (Minetti et al., 2013; Schinkel-Ivy et al., 2014).

Running speed had a significant effect on hip flexion at initial contact, but the values were only about a  $1^\circ$  increase for each speed. Although statistical significance was detected, the change is unlikely to be of practical importance for running faster at these speeds. Running speed also had a significant effect on hip extension, but it increased only  $\sim 2^\circ$  at each faster speed. With greater hip range of motion and less time to accomplish it (shorter stance time), the statistically significant increase in peak joint power values with running speed are likely to be due to more substantial increases in hip joint angular velocity rather than large increases in hip joint moments.

In ankle kinematics, running speed had a significant effect on dorsiflexion at initial contact and plantarflexion in early swing phase, but maximum dorsiflexion was similar in mid-stance (Fig. 7). This suggests that the gastrocnemius and soleus generated the observed greater moments in mid-stance, halting dorsiflexion and initiating plantarflexion more rapidly across running speeds (None of these speeds were fast enough for the runners to select a forefoot contact pattern). Therefore, somewhat increased power absorption in early stance and much greater power generation in late stance during push-off just prior to flight phase was observed. There was nearly a 30% increase in peak power generation at the ankle during push-off from the slowest (2.6 m/s) to the fastest (3.7 m/s) running speed. This suggests that the ankle plays an important role in the lower limb joint kinematics and kinetics across running speeds. This may not be the optimal strategy for maintaining faster running speeds, but this strategy was demonstrated in this small cohort of recreational runners across this narrow range of running speeds.

The small cohort of recreational runners in this study may not be representative of all runners, making the conclusions here generalizable to only a small subset of runners across a relatively narrow range of running speeds. Elite runners may display very different kinematic and kinetic patterns at the hips, knees and ankles across a range of running speeds (Schache et al., 2011). Older versus younger runners may have different spring-mass characteristics (Pantoja et al., 2016) and the relatively wide range

of age in this study may influence the generalizability of these findings.

Even the construct of “running speed” may be artificial, as most trials in this study were discarded due to slow fluctuations in forward velocity that may be perfectly normal. Using timing gates to calculate average speed across the capture volume would have masked these common changes in instantaneous speed. Minetti et al. have shown that fluctuations in running speed have no substantial impact in the metabolic cost of running, and suggest a reexamination of how tightly controlled running at a particular speed might be for recreational runners (Minetti et al., 2013). It seems much more likely that running in real world settings might involve natural fluctuations in running speed over several seconds.

## 5. Conflict of interest statement

The authors declare that there is no conflict of interest regarding the content of this article.

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