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ORIGINAL RESEARCH

The Effect of Step Frequency and Running Speed on the Coordination of the Pelvis and Thigh Segments During Running

Raphael M. Mesquita,[1] Jean-Benoit Morin,[2] and Arthur H. Dewolf[1]

[1]Laboratory of biomechanics and Physiology of Locomotion, Institute of NeuroScience, Université catholique de Louvain, Louvain-la-Neuve, Belgium; [2]Laboratoire Inter-universitaire de Biologie de la Motricité, Université Jean Monnet Saint-Etienne, Université Savoie Mont Blanc, Saint-Etienne, France

This study investigates the specific influence of step frequency (SF) and speed on the coordination between pelvic and thigh movements. Eight recreational male runners ran at different SFs and speeds on an instrumented treadmill. The coordination between the pelvis and thigh segments was analyzed using modified vector coding in the sagittal and frontal planes (FPs). Our findings show that hip range of motion increases as a function of SF in the sagittal plane. Pelvic tilt plays a compensatory role in hip extension, particularly at lower SFs. In the FP, pelvic roll increased at lower SFs, whereas the thigh abduction angle was participant dependant. Coordination analysis showed that thigh movements dominated the sagittal plane motion, which was simplified at higher SF. At low SF, the pelvic movements were increased and anticipated, playing a more dominant role in explaining motion. In the FP, pelvic movements dominated the motion. The increase in pelvic motion at low SFs stretches the hip flexors further and for a longer period. The link between SF, pelvic motion, and the risks of running-related injuries in the sagittal and FP is considered. Understanding these could help athletes and sports professionals optimize performance and reduce injury risk.

Keywords: running step frequency, coordination analysis, anterior pelvic tilt

Runners have been shown to self-optimize their preferred step length (SL),[1,2] which affects both the kinematics and kinetics of running.[3] The choice of one's SL is one that minimizes both the mechanical and metabolic power of the runner.[1,2] For example, to take short steps at a fixed speed, one must increase their step frequency (SF) at the expense of the increased internal power done to reset the limbs for the following step.[2] On the contrary, taking larger steps increases the muscular power done during the propulsion phase of stance.[2] This is because the distance a runner can cover over the contact phase is limited due to anatomical constraints,[4] such as the stretching of the hip flexors and coxofemoral joint capsule.[5] These constraints impact the peak extension of the hip over a stride.

Hip motion in the sagittal plane is determined by the amplitude and coordination between the thigh and pelvis motion. During a stride, the hip undergoes a slight flexion at touchdown (TD) before rotating backward. Maximal extension is reached after foot takeoff, the hip then reaches maximal flexion during the late swing phase prior to extending before TD.[5,6] Due to the structural limitations in hip extension mobility, to increase SL during the contact phase, one compensates by increasing anterior pelvic tilt angle.[7] Therefore, it is likely that the pelvic tilt angle in the sagittal plane and its coordination with the motion of the thigh will be affected by a modification of SL (or SF at the same given speed).

In a preliminary study, Dewolf and De Jaeger[8] showed that modifying SL by adjusting either running speed at a fixed SF or by altering SF at a given speed have different consequences on the sagittal plane pelvic movements and running pattern. Indeed, reducing SF at a given speed tilts the pelvis further than when changing SL by adapting the speed at a given SF. This is important since the magnitude of the anterior pelvic tilt during running is related to injuries in the lumbopelvic region, stemming from excessive stretch and thus strain induced on the muscle tendons attached to the pelvis.[9] Dewolf and De Jaeger[8] hypothesized that the observed differences were due to alterations in the lower limb stiffness, which, due to changes in the center of mass displacement, affect the range of motion and peak hip extension angle. In a recent study, we showed that the lower limb stiffness increases with SF but not with running speed.[3]

Therefore, here we hypothesized that the varied movements in the thigh and pelvis would change the coordination pattern between these segments, which would be discernible by varying SF but not speed. In addition, the movements of the center of mass in the frontal plane (FP) account for less than 3% of the work done to sustain the center of mass motion,[10] and hip motion in this plane serves as a "shock absorber" during the braking phase of stance.[11] Thus, varying the braking load by altering SF should also change this shock absorption mechanism and the coordination between pelvis and thigh in the FP.

To pursue this, we investigated the changes induced by SF and running speed on the hip joint, thigh, and pelvis movements at 4 different running speeds and 5 imposed SFs. We examined the coordination between the pelvis and thigh segments via modified vector coding.[12–15] This technique refers to the vector angle orientation between 2 adjacent time points on an angle–angle diagram.[16] It has been used to analyze the coordination patterns between different segments,[14–17] explore the variability of overuse injuries,[13] or investigate sprint coordination from a starting block.[12] It classifies the coordination of 2 segments by dominance phases (the segment with the most angular displacement) and provides an understandable approach to analyzing the interaction between segments.

Mesquita https://orcid.org/0000-0003-0781-0789
Morin https://orcid.org/0000-0003-3808-6762
Dewolf (arthur.dewolf@uclouvain.be) is corresponding author, https://orcid.org/0000-0003-1498-3987

Methods

Experimental Protocol

The protocol for this paper was the same as Mesquita et al[3]; therefore, similar sections were only explained briefly. Eight recreational healthy male runners (age = 22.2 [1.03] y, height = 1.84 [0.03] m, mass = 73.7 [5.65] kg; mean [SD]) with a weekly running distance of ~15 km and with no reported injuries in the last 6 months participated in the study. The sample size was based on an a priori statistical power analysis (power: $1-\beta=0.8$, minimal sample size = 8), where β is the probability of making type II error, from the results of lower limb stiffness from a pilot study.[18] Then, as the sample size was predetermined in the present study, to confirm sufficient statistical power of our results, an a posteriori power analysis using GLIMMPSE (version 3.1.2, University of Colorado Denver)[19] was done on the maximal anterior pelvic tilt angle data (frequency main effect: $1-\beta=0.846$, for a sample size of 8). Informed written consent was obtained and the study followed the Declaration of Helsinki guidelines. All procedures were accepted by the UCLouvain ethical committee (B403201838331).

Participants ran on a treadmill at 4 different constant speeds: 8, 11, 14, and 17 km·h^{-1}. At each speed, participants were asked to run at 6 different SFs: their preferred SF (PSF), and 2.0, 2.4, 2.8, 3.2, and 3.6 Hz conveyed via an electronic metronome and amplified on loudspeakers. Running at the PSF and running at 2.8 Hz showed similar results for mechanical work and lower-limb stiffness across all speeds; therefore, in the present study, only the results for 2.8 Hz were considered.

Experimental Setup and Data Collection

An instrumented treadmill (h/p/Cosmos—Arsalis) with a belt surface of 1.6 × 0.65 m was mounted onto 4 strain-gauge force transducers that measured the 3 components of the ground reaction force exerted by the treadmill under the foot.[20] The signals were amplified, low-pass filtered (4-pole Bessel filter with a −3-dB cutoff frequency at 200 Hz), and digitized by a 16-bit analog-to-digital converter at 1000 Hz.

Bilateral, full-body 3-dimensional kinematics were recorded at 200 Hz using a Qualisys system with 13 cameras (12 Mocap OQUS 6 + cameras and 1 video MIQUS M1 camera, Qualisys) placed around the treadmill. Participants were equipped with 29 retroreflective markers on specific anatomical locations. The markers used for this study are the anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), greater trochanter (GT), and the lateral condyle of femur (LE) on both sides of the body. Kinematic data were then oversampled at 1 kHz using the spline routine in MATLAB and filtered similarly to the kinetic data with a low-pass cutoff frequency at 30 Hz.

Data Processing

Data processing was performed via custom-made programs written with LABVIEW (National Instruments Corporation) (2021) and MATLAB (R2022a).

Gait Separation

The stride period was defined as the time between a right foot contact and the following right foot contact. Each foot contact was detected using the vertical component of the ground reaction force (F_v), filtered by a 2-way eighth-order Bessel filter with a low-pass cutoff frequency of 30 Hz, and corresponds to the moment at which $F_v > 10\%$ of body weight.

Elevation and Joint Angles

For each segment, the elevation angles (θ) in the sagittal plane were computed as: $\theta = \arctan\left(\frac{y_p - y_d}{z_p - z_d}\right)$, where ($y_p$, z_p) and (y_d, z_d) are, respectively, the y- and z-axis coordinates of the proximal (p) and distal (d) segment markers. The thigh segment angle was defined as the angle between the lateral condyle of femur, greater trochanter, and the vertical. The anteroposterior pelvic tilt angle was defined as the angle between the posterior superior iliac spine, ASIS, and the vertical (Figure 1A). The hip joint angle in the sagittal plane was computed from the elevation angle of the thigh and pelvis segments. Figure 1B shows the average traces of the hip, thigh, and pelvic tilt angles in the sagittal plane. The range of motion over the stride was computed as the difference between the maximal and minimal angles observed (Figure 1C).

In the FP, the elevation angles were computed as: $\theta = \arctan\left(\frac{x_p - x_d}{z_p - z_d}\right)$, where x_p and x_d are, respectively, the proximal and distal x-axis coordinates. Figure 4A shows the left–right pelvic roll angle (formed by the left ASIS, right ASIS, and vertical) and the abduction–adduction angle (lateral condyle of femur, greater trochanter, and vertical) of the right thigh in the FP.

Coordination Analysis

The coordination pattern between 2 segments was analyzed using a coupling angle (CA) mapping technique, which has been described thoroughly in Donaldson et al[12] and Needham et al.[14] In brief, the anterior pelvic tilt and thigh flexion elevation angles in the sagittal plane were time-normalized over 400 points then coupled based on a proximal–distal naming convention (Figure 2B). In the FP, the same was done between the pelvic roll and thigh abduction–adduction angle. The CA were computed from these proximal–distal angle plots, which express the relationship between 2 segments, using a modified vector coding method.[14,21] The CA represented the vector angle between adjacent points in the angle–angle figure relative to the right horizontal of the first quadrant, ranging between 0° and 360° (Figure 2A).

The CA position on a circular plane describes the relative rotation of any 2 segments. This rotation can be either in-phase (moving in a similar direction) or anti-phase (moving in opposite directions). Furthermore, segments can either rotate in a clockwise or counterclockwise manner. As a result, 4 possible relationships between segments exist, corresponding to the circular plane's 4 quadrants. Moreover, these quadrants are divided into two 45° bins based on the dominant segment, that is, the segment undergoes the greatest rotation in a given time period. Thus, 8 distinct coordination bins are used to explain the relationship between segment rotations (in-phase or anti-phase), distal or proximal segment dominance, and the direction traveled (Figure 2A).[12,17]

Figures 2C and 5B express the time-normalized average coordination profiles in the sagittal and FP, respectively. Segment dominance was expressed as a percentage by converting CA into gradians (90° = 100 gradians).[17] For example, a 90° CA equals 100 gradians, indicating 100% distal segment dominance, where the distal segment rotates and the proximal segment is fixed over that instant, whereas 50% dominance reflects equal rotation of both segments (Figures 2C and 5B). Dominance was limited between 50% and 100%, as the defined bins switch every 50%. Primary

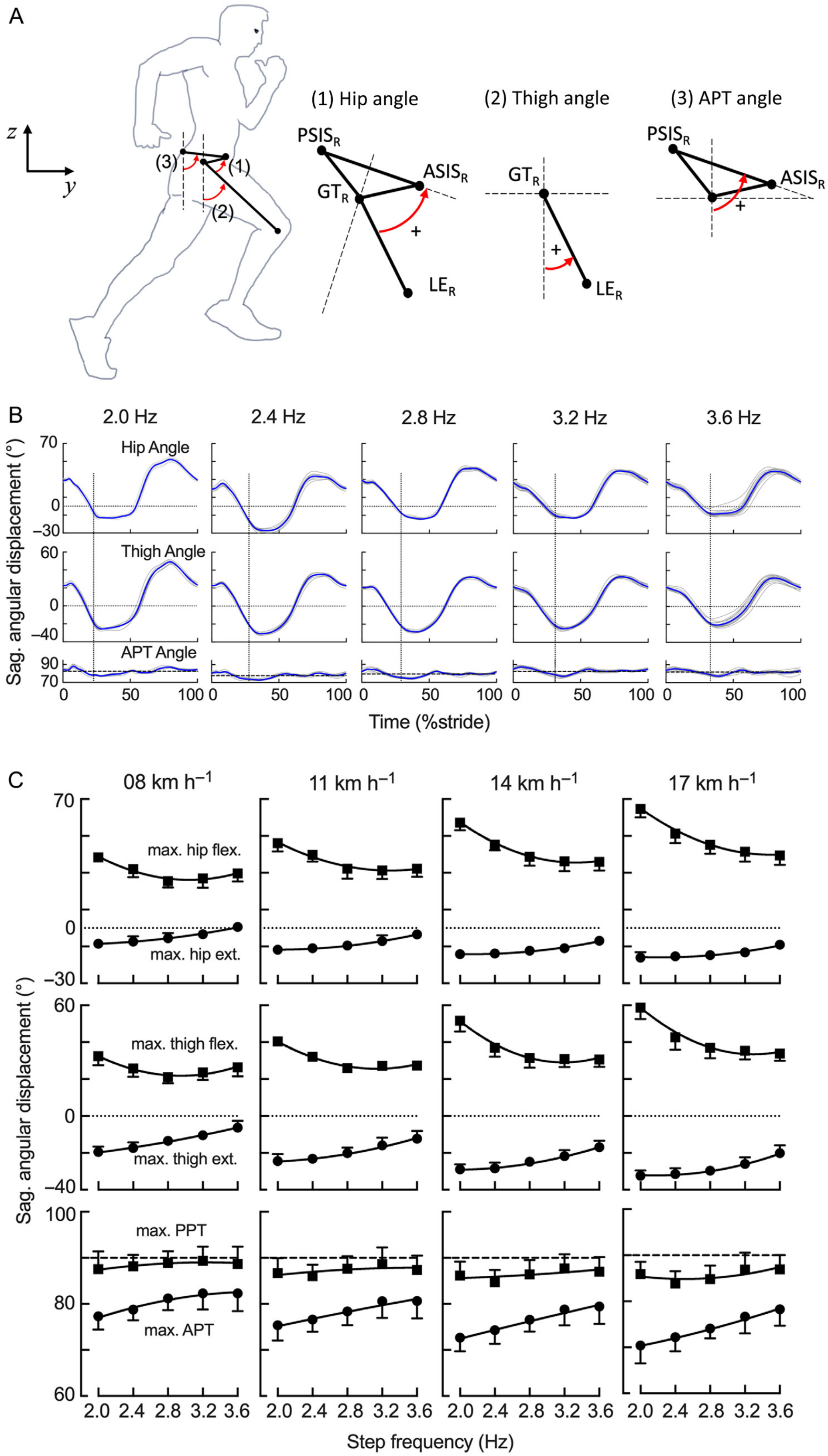

Figure 1 — (A) Diagrams of the hip (1), thigh (2), and anteroposterior pelvic tilt (3) elevation angles. The markers that form the right hip angle are the GT_R, the $ASIS_R$, and LER. The thigh elevation angle is formed by the GT, LE, and the vertical, whereas the APT is formed by the posterior superior iliac spine, the ASIS, and the vertical. (B) Typical time-normalized average traces of one participant (dark blue) and individual strides (light gray) of the hip, thigh, and anteroposterior pelvic angles of the right side beginning on a right foot contact at 14 km·h^{-1} for all step frequencies. The vertical dotted line represents the

coordination patterns were classified by color, and distal or proximal segment dominancy illustrated by light or dark shades of each color, respectively. Thus, changes between colors represented overall coordination changes and color tones within a given color represented change in segment dominance.[12,17]

The mean bin frequencies (Figures 3 and 5C) express the distribution of each segment rotational relationship between all participants. For each speed class, the sum of the 8 bins equals 100%.

Statistics

For each participant, the maximum angles were averaged across all strides of the one participant. First, a Shapiro–Wilk test was performed to verify normality. When the data did not meet the normal distribution (Shapiro–Wilk W test, $P < .05$), nonparametric statistics were used. A linear mixed effect model was used to assess the individual interaction effects of frequency and speed on the angle maxima. For the frequency bins, a Friedman test (Q test) with a Wilcoxon signed-rank comparison was used to analyze the difference of each bin per frequency class. For both parametric and nonparametric tests, a Bonferroni post hoc correction was used to counteract the multiple comparisons performed. Statistical tests were run on IBM SPSS Statistics. The results of the statistical tests were considered significant for a P value $< .05$. Figures portraying the maximum angles show average frequency classes per speed. These averages are only for presentation and were not used in the statistical analyses.

Results

In the sagittal plane, hip flexion decreased as a function of SF ($F = 25.1$, $p < .001$; Figure 1C), principally due to the difference between 2.0 Hz and the other frequencies (post hoc $P < .010$; between 2.0 and all other SFs). Hip extension lessened as SF increased ($F = 29.8$, $P < .001$). This effect was mainly due to the higher frequencies (post hoc $P < .001$ for 3.6 Hz compared with lower SFs). At the lower SFs, hip extension reached a maximal angle (post hoc $P = 1$; between 2.0 and 2.4). As a function of speed, both hip flexion and extension increased (flexion: $F = 149.2$, $P < .001$; extension: $F = 199.1$, $P < .001$).

The hip angle depends on thigh and pelvis movements. The thigh had similar speed and frequency effects as those observed in hip flexion/extension. In other words, thigh flexion and extension increased as a function speed (flexion: $F = 204.6$, $P < .001$; extension: $F = 199.1$, $P < .001$) and lessened as a function of increasing SF (flexion: $F = 48.5$, $P < .001$; extension: $F = 29.8$, $P < .001$). Regarding the pelvis, the maximal posterior tilt (PPT) slightly increased as a function of SF ($F = 8.7$, $P < .001$), whereas the maximal anterior tilt (APT) lessened when SF increased ($F = 26.0$, $P < .001$). As a function of speed, PPT decreased, whereas APT leaned more anteriorly compared with its average position (PPT: $F = 25.3$, $P < .001$; APT: $F = 348.7$, $P < .001$). The average orientation of the pelvis was tilted anteriorly over a full stride, but this average tilt increased at lower SFs ($F = 24.2$, $P < .001$) and with speed ($F = 124.2$, $P < .001$).

Most of the movement between the thigh and pelvis occurred in the distal (thigh) segment and was mostly in-phase, shown by the higher frequencies observed in the downward (negative) in-phase thigh movements and in the upward (positive) in-phase thigh movements ($P < .001$ for all Q tests done at each frequency and both variables; Figures 2B, 2C, and 3). As SF increased, the negative in-phase thigh bin along with the positive in-phase thigh bin increased their dominance over the stride (at 14 km·h^{-1}: $P < .010$ for both). Contrastingly, at the lower SFs, the motion of the pelvis did more movement than at higher frequencies. This was the case both for the anti-phase negative pelvic dominant motion (roughly occurring at TD; at 14 km h^{-1}: $Q = 12$, $P < .010$) and for the in-phase positive pelvic dominant motion (roughly during the TD of the contralateral limb; at 14 km·h^{-1}: $Q = 12.25$, $P = .010$).

As a function of speed, the in-phase positive movements of the thigh did not vary ($P > .050$ at all frequencies), whereas the in-phase negative thigh movements increased with speed at the lowest frequencies (at 2.0 Hz: $Q = 18.2$, $P < .001$) but not at the highest frequency (at 3.6 Hz: $Q = 3.1$, $P = .300$). This was most likely due to the increased role of the pelvis at the lower speeds and SFs.

In the FP, the average pelvic roll angle, over the entire stride was at 90° and did not vary with speed ($P = .70$) nor with SF ($P = .80$). The ROM of the pelvis increased with speed ($F = 95.1$, $P < .001$) and decreased with SF ($F = 43.0$, $P < .001$). Regarding the thigh abduction–adduction angles, both maximal angles were relatively small at SFs above 2.8 Hz. Thigh adduction during contact was not affected by speed ($F = 0.6$, $P = .600$) and decreased slightly with SF ($F = 5.9$, $P < .001$). Where maximal thigh adduction followed a normal distribution across all SF classes, its counterpart, maximum thigh abduction did not (2.0 Hz: $W = 0.9$, $P < .001$). Indeed, at the lower frequencies (2.0 and 2.4 Hz), the thigh abduction (during swing phase) was about 5 times greater than at the higher SF classes. At 2.0 Hz, an increase with speed was also observed ($Q = 25.5$, $P < .001$, post hoc between 8 and 17 km·h^{-1}, $P < .001$; Figure 4B and 4C).

Motion was mostly accounted by the positive and negative in-phase pelvic movements (at 2.0 Hz: $Q = 187.7$, $P < .001$), which did not differ from one another (post hoc $P = 1$) but were both different from other bins ($P < .001$ for all post hoc tests; Figure 5). With higher SFs, the ROM of the thigh segment in the FP was reduced. The segment dominance showed a simplification in the bin distribution, for example, at 3.6 Hz, motion was mainly done by the positive and negative in-phase pelvis movements and the anti-phase negative pelvic movements, whereas the other bins tended to further resemble each other and remain significantly lower than the ones cited above. With speed, the distribution of dominant motion did not vary in either the thigh or pelvis segments ($P > .05$ at all frequencies).

Figure 1 *(continued)* moment of take-off for the right foot, and the horizontal dotted line represents the instant at which the angle crosses 0°. In the APT elevation angle trace, the horizontal dashed line represents the average pelvic tilt angle over the stride. (C) The maximum and minimum mean (SD) angles for the hip, thigh, and pelvic tilt angles as a function of step frequencies at the different speeds. Dotted lines represent 0° and the dashed line represents 90° direction in the APT–PPT plots. In each panel, the points are the average of the data measured on all steps of all subjects. The vertical bars indicate the SD of the mean and are drawn when they exceed the size of the symbol. Filled and dashed lines are spline functions with 4 notches from all data points. These data are plotted via GraphPad Prism (version 10, GraphPad Software). APT indicates anterior tilt; ASIS, anterior superior iliac spine; $ASIS_R$, right ASIS; GT, greater trochanter; GT_R, right greater trochanter; LE, lateral condyle of the femur; LE_R, right lateral condyle of the femur; PPT, posterior tilt; $PSIS_R$, right posterior superior iliac spine. (Color figure online)

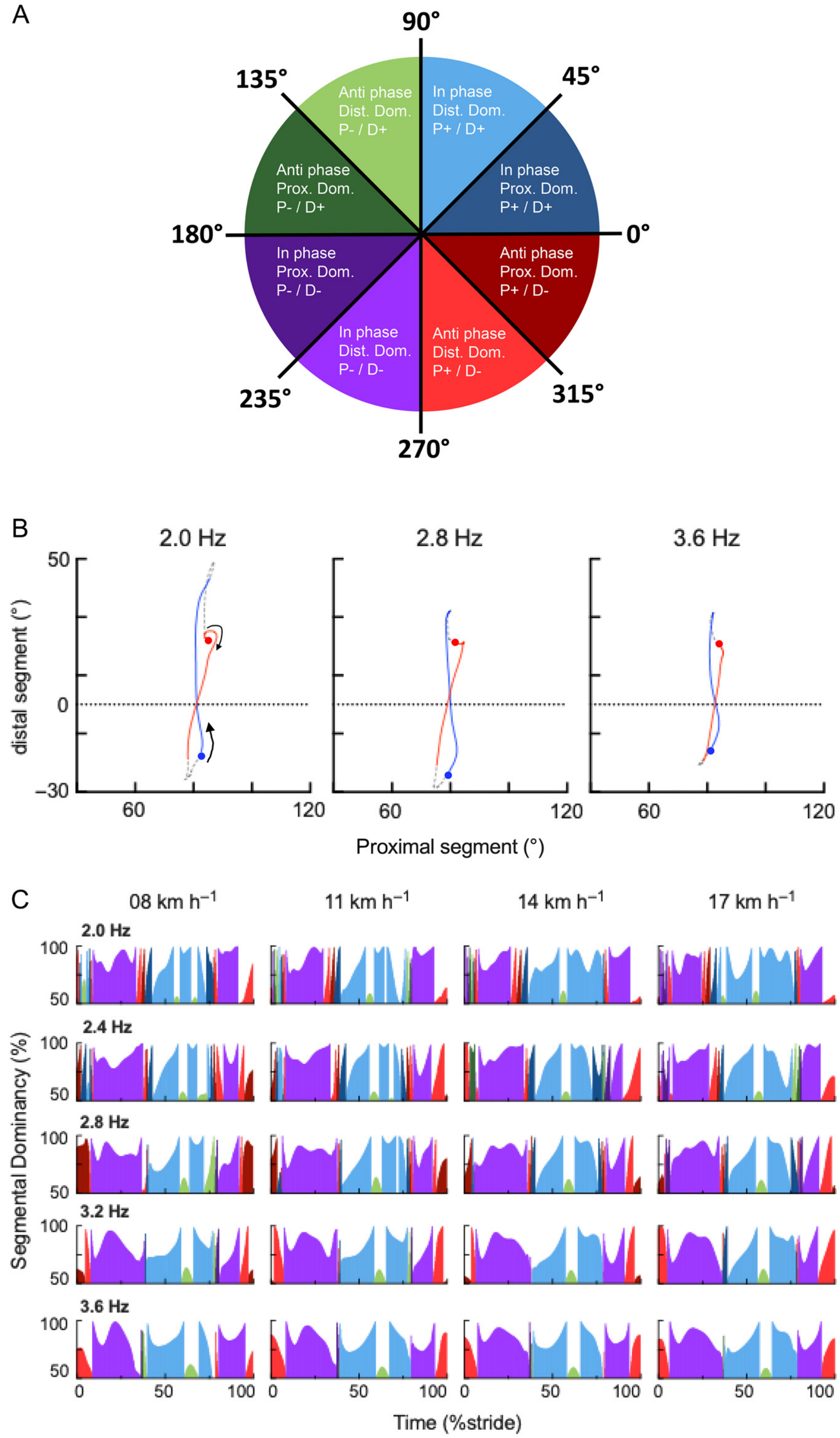

Figure 2 — (A) Coordination bins showing the proximal or distal segment dominancy and whether movements are in or out of phase adapted from Donaldson et al.[12] (B) Angle–angle average sagittal plane traces at 14 km·h^{-1} and plotted at 3 different step frequency classes (2.0, 2.8, and 3.6 Hz) for the same subject as in Figure 1. The distal segment corresponds to the thigh elevation angle, and the proximal segment is the pelvic tilt in the sagittal plane. The red and blue lines correspond, respectively, to the right and left feet during contact. The dots of corresponding colors represent the beginning of the contact phase. The dashed gray lines represent the aerial phases and the arrows indicate the direction of movement. (C) Time-normalized average coordination profiles between the anteroposterior pelvic tilt and thigh flexion–extension angle expressed between 50% and 100% for all subjects at the different speeds and frequency classes. The colors are the same as in panel (A) and represent the dominant segment and its angle.

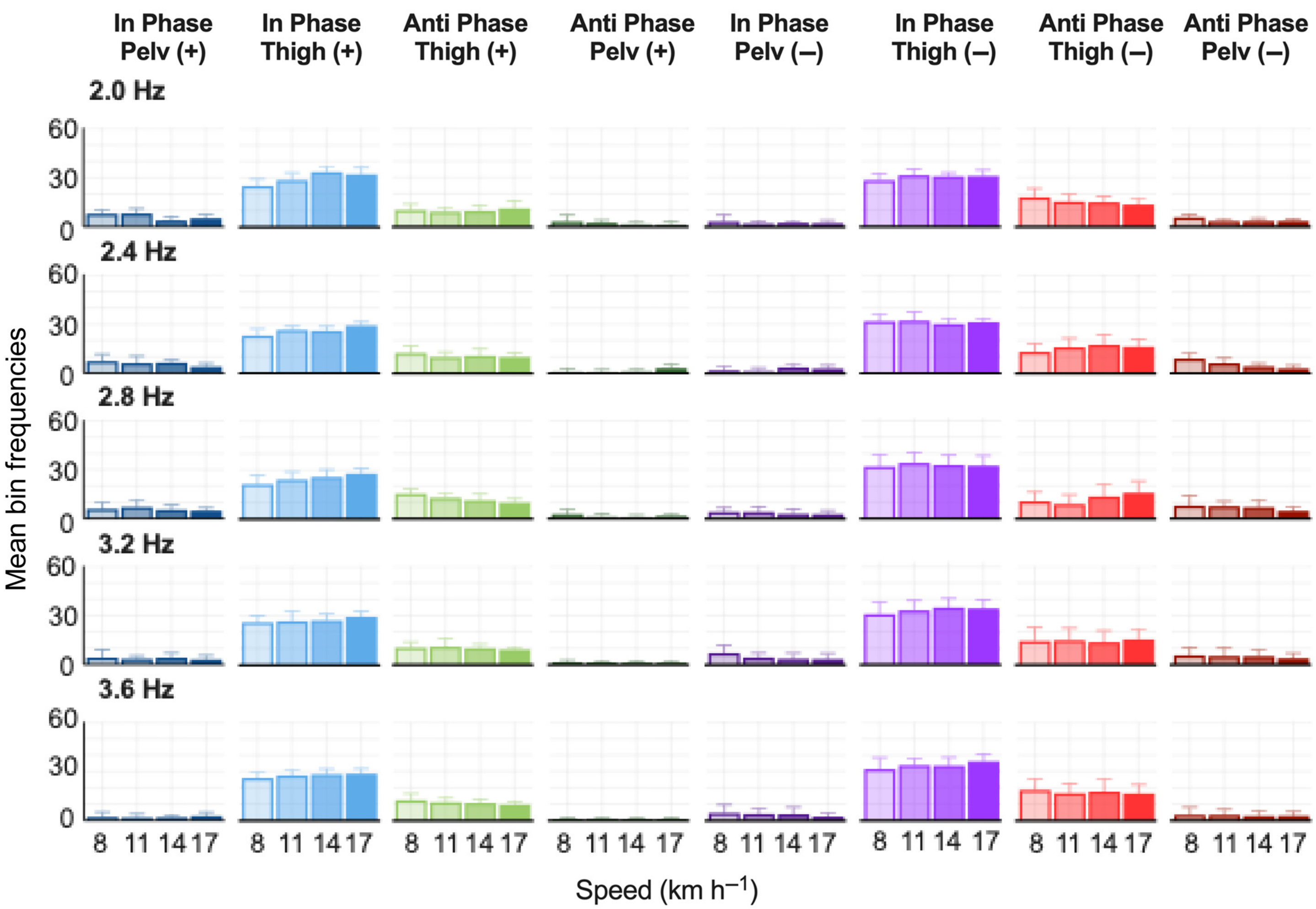

Figure 3 — Mean bin frequencies for all subjects at all speeds and step frequencies. Colors are the same as in Figure 2. (Color figure online)

Discussion

The purpose of this study was to understand how a change in SF and running speed affected the motion of the pelvis and thigh segments, and in turn the hip joint. We hypothesized that the coordination pattern between the thigh and pelvis motion would change by varying SF but not running speed in both sagittal and FPs. The following section discusses this, and its potential relationship to tissue strain and potential hamstring relate running injuries. It should be noted that although the statistical power is above 0.8, a sample size of 8 participants remains rather small and could be a limitation to our results. This is further discussed below.

The Effect of SF and Running Speed on the Hip Joint Motion

When running at lower SFs, one must decrease their lower limb stiffness to achieve the longer step required to match the imposed SF class.[3,4,22] Furthermore, the pelvis has been shown to tilt anteriorly to compensate for the stretching of the hip flexor musculature and hip joint capsule,[5] which limits hip extension when running speeds increase.[9,23] This compensatory role of the pelvis, most visible at lower SFs, allowed for a more extended thigh and thus larger ROM (Figure 1). When observing the maximal angle instances during running at low SFs, the timing of the maximal APT came earlier than that of peak hip extension. From a coordination point of view, running motion at low SF was first dictated by the pelvis followed by thigh movement (Figures 1–3), meaning that the proximal segment moved faster than the distal segment to prepare for the swing phase [5] stretching the hip musculature further and longer than at higher SFs.

At higher SFs, the pelvis was less anteriorly tilted compared with lower frequencies. Pelvic displacement is minimized in order to conserve energy when running at a PSF[11] and has been shown not to increase the energy consumption when SF rises by 10% from the PSF.[24,25] Moreover, the hip ROM is at its lowest as lower limb stiffens and SL is reduced.[2,3] In this case, hip extension was not maximized and the hip flexor muscles not overstretched. In terms of their coordination, the peak angle instances of both segments were about synchronized, motion was “simplified,” and the thigh accounted for almost all of the segmental dominancy.

Regarding running speed, it was hypothesized that the segmental coordination would not be affected. This was indeed the case for all SFs but 2.0 Hz, where the thigh dominancy was lessened and the movements of the pelvis accounted for a greater portion of dominancy over the stride. Thus, when SL is pushed to

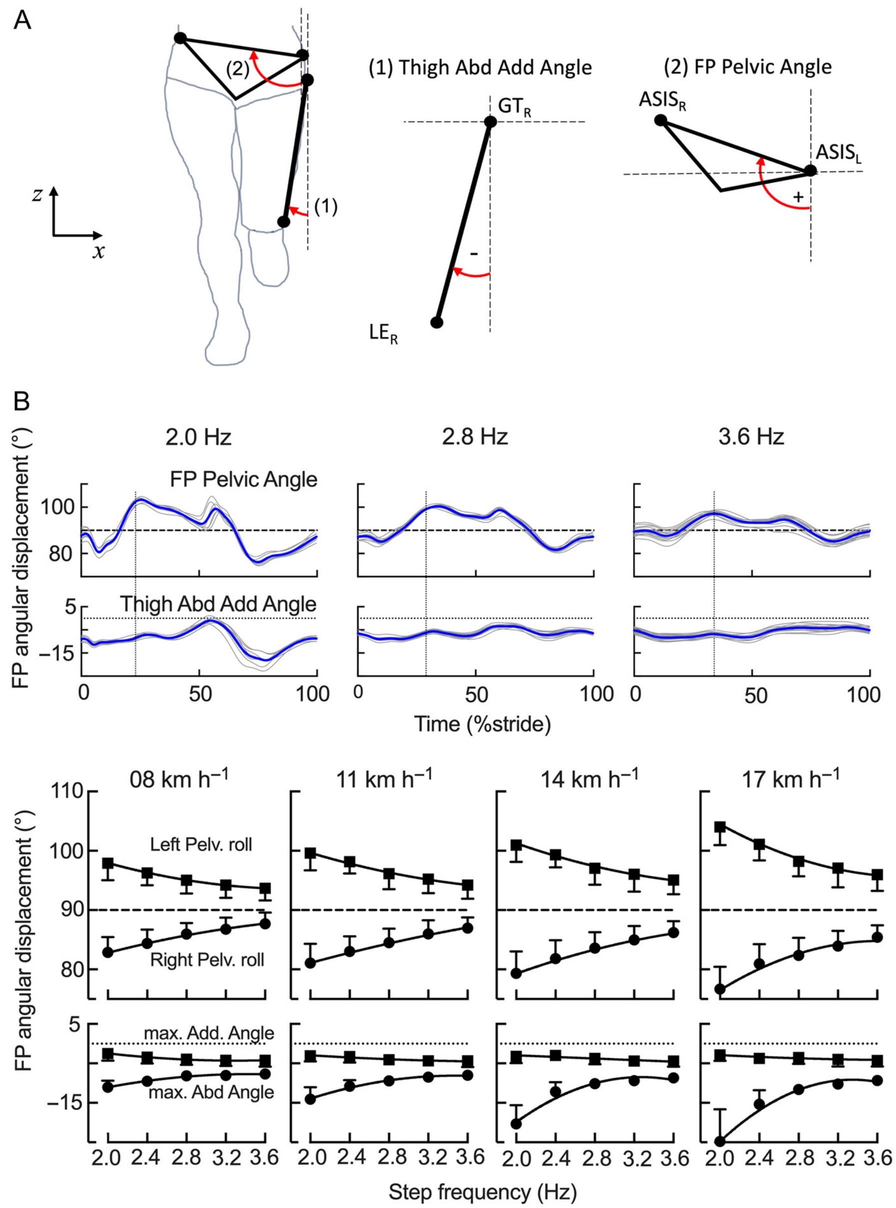

Figure 4 — (A) Diagrams of the thigh abduction (1) and pelvic roll (2) elevation angles in the FP. The thigh abduction–adduction elevation angle is formed by the GT, LE, and the vertical, whereas the pelvic roll angle is formed by the $ASIS_L$ and $ASIS_R$ and the vertical. (B) Typical time-normalized average traces of one participant (dark blue) and individual strides (light gray) of the hip, thigh, and anteroposterior pelvic angles of the right side beginning on a right foot contact at 14 km·h^{-1} for all step frequencies. The vertical and horizontal dotted and dashed lines are the same as Figure 1B. The maximum and minimum mean (SD) angles for the pelvic roll and thigh abduction–adduction angles for all subjects. The dashed line corresponds to 90°. ASIS indicates anterior superior iliac spine; $ASIS_R$ and $ASIS_L$, left and right ASIS; GT, greater trochanter; GT_R, right greater trochanter; FP, frontal plane. (Color figure online)

such extremes, the thigh is incapable of maintaining dominance throughout the stride and the increased anterior tilt is necessary to match the required distance.

In the FP, the values obtained for the pelvic roll were similar to Schache et al,[26] when running with an SF closest to the PSF of their participants (±2.8 Hz). At 3.2 and 3.6 Hz, the thigh movements kept to an axis and the ROM around the FP was rather small. Therefore, at these frequencies, the pelvic movements explained most of the motion (Figures 4 and 5), whereas, at lower SFs, and particularly at faster speeds, the pelvic roll motion doubled and the thigh abduction angle increased 5-fold. The latter varied largely within participants, as evidenced by the individual traces in Figure 4B and the high SD in Figure 4C. This drastic increase in pelvic roll and thigh abduction is likely to be related to the increased lower limb momentum during the swing phase described at low SFs,[3] which would increase thigh abduction and prevent the foot from catching the ground during the swing phase.

Thigh–Pelvis Coordination Analysis in the Context of Running-Related Injuries

Several studies have linked the APT position and biceps femoris stretch, and thus strain and injury risk with running.[7,27–33] Indeed, recent histological studies have shown that the mechanical forces

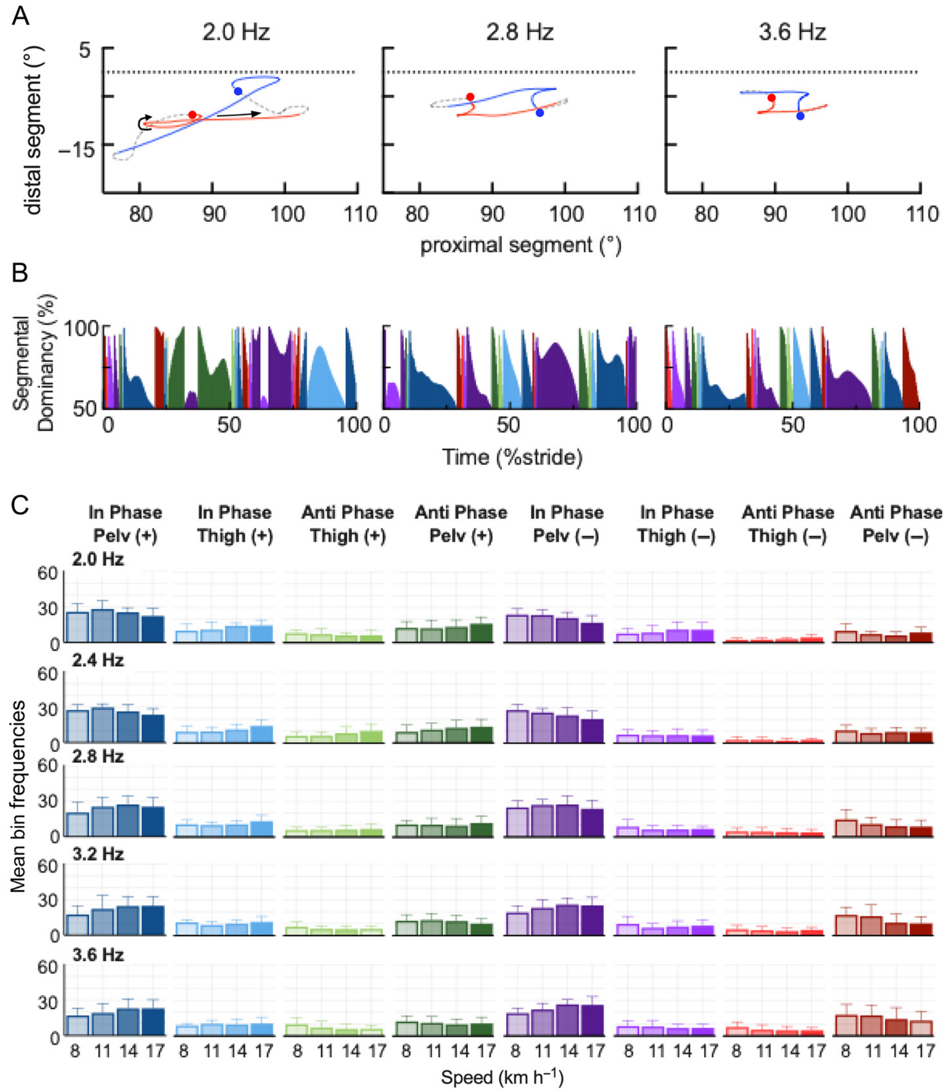

Figure 5 — (A) Angle–angle FP figures at 14 km·h^{-1} and plotted at 3 different step frequency classes (2.0, 2.8, and 3.6 Hz) for the same subject as in Figure 1. The distal segment corresponds to the thigh abduction–adduction angle and the proximal segment is the pelvic roll in the FP. The colors and lines are as in Figure 2B. (B) Time-normalized average coordination profiles for the FP relationship between the pelvic roll and thigh abduction–adduction in the same step frequency classes as in panel (A) for all subjects. All other indications are as in Figure 2C. (C) Mean bin frequencies for all subjects, speeds, and step frequencies. Colors are the same as in Figure 2. FP indicates frontal plane. (Color figure online)

within the hamstrings, which are evenly distributed between the BF and semitendinosus muscles,[30] may explain the prevalence for BF injury partly due to the muscle's higher vulnerability to repetitive loads[34–36] compared the medial hamstring muscles and by its elastic fiber disposition.[30] As seen in the present study, at the lower SF classes, the peak ATP and peak hip extension were further apart than at 2.8 Hz. This increased lag between peak values stretches the hip flexor muscles further and for a longer period, which is prevalent for BF injuries. Furthermore, the increased thigh abduction movements at the lower SFs may result in an increased muscle-tendon tissue stress as thigh abductor weakness has been linked to several different running-related injuries such as iliotibial tract syndrome.[37]

Recent evidence suggests that enhancing core stability may lead to improved pelvic movement stabilization, thus reducing the APT and lowering the risk of hamstring injuries.[38] Our previous results[3] indicate that an increase in lower limb stiffness could be another avenue to limit BF stretch by increasing SF and decreasing the ROM of the hip in both planes. This would reduce the solicitation of the pelvis motion and decrease thigh abduction–adduction movements potentially mitigating this injury risk. Increasing speed does not seem to change the segment coordination significantly, except at the lowest SF class. It should be reminded that this finding does not consider fatigue, which could enhance the change in coordination with speed at a less extreme SF.

Limitations

It should be noted that although the statistical power was met for one of our key variables, our sample remains relatively small and composed of only male participants. Thus, generalizing such results to female runners, for example, whose morphological hip characteristics are different, should be done with care. Furthermore, this study did not directly measure hip strain and running injuries. It would be of interest to analyze the segmental coordination of the pelvis and thigh in runners with a history of hamstring injuries.

In conclusion, this study shows the importance of SF in modulating hip ROM, and thus segmental coordination of thigh and pelvis during running. As expected, the SF (and with it, the lower-limb stiffness) played an important role in determining the thigh–pelvis coordination strategy chosen by the runners. On the other hand, running speed had little effect on the segment coordination. While the PSF is self-optimized by elite runners to reduce cost and improve gait stability,[1,2] a lower SF is both more costly but also potentially more related to tissue strain and in turn running-related injuries. Instead, our results suggest that increasing lower-limb stiffness by imposing higher SF changes the pelvis–thigh segmental coordination by reducing pelvic motion. This may be helpful in reducing hip (and lumbar) stress. Understanding these intricacies can help athletes and sports professionals to optimize performance and reduce the risk of injuries.

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