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Predictors of pelvic acceleration during treadmill running across various stride frequency conditions

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Abstract

Pelvic running injuries often require extensive rehabilitation and pelvic girdle pain is a barrier to running engagement in population sub-groups, such as perinatal women. However, exploration into how external pelvic loading may be altered during running is limited. This study assessed which biomechanical variables influence changes in external peak pelvic acceleration during treadmill running, across various stride frequency conditions. Twelve participants (7 female, 5 male) ran (9 $km \cdot h^{-1}$) at their preferred stride frequency, and at ±5% and ±10% of their preferred stride frequency. Coordinate and acceleration data were collected using a motion capture system and inertial measurement units. Linear mixed models assessed peak tibial acceleration, displacement from hip to knee and ankle, contact time, and stride frequency as predictors of peak pelvic acceleration. Stride frequency and contact time interacted to predict peak vertical (p = .006) and resultant (p = .009) pelvic acceleration. When modelled, short contact times and low stride frequencies produced higher peak vertical (p = .007) and resultant (p = .016) pelvic accelerations than short contact times and average, or high stride frequencies. Increasing contact time, or increasing stride frequency at shorter contact times, may therefore be useful in reducing pelvic acceleration during treadmill running.

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Introduction

Given the high prevalence of lower-extremity injuries (Kakouris et al., 2021; Taunton, 2002), investigations of injury risk during running have predominantly focussed on the lower limb (Barton et al., 2016; Crowell & Davis, 2011; Milner et al., 2006). However, for every running stride, the pelvis is loaded twice as frequently as either leg. One study found that the sacrum and innominate bones were among the most common sites for bone stress injuries in runners (Kliethermes et al., 2021), and such injuries often lead to extensive rehabilitation and recovery time (Browning, 2001). Pelvic girdle pain is also particularly prevalent in certain population sub-groups, such as perinatal women (Norén et al., 2002), where running-related pelvic girdle pain has a reported prevalence of 53% among postpartum runners (Moore et al., 2021). Further, pelvic girdle pain has been cited as a barrier to engagement in running during pregnancy and postpartum (James et al., 2022).

For the lower limb, external measures of loading previously associated with risk of injury include high vertical ground reaction forces and horizontal braking forces (Davis et al., 2016; Napier et al., 2018). Meanwhile, surrogate measures, such as high vertical tibial accelerations, have also been associated with increased tibial stress fracture risk (Milner et al., 2006). Using wearable devices to measure triaxial segment acceleration allows large amounts of data to be collected, where force plates may be unavailable (Busa et al., 2016; Reenalda et al., 2016; Sheerin et al., 2019). Given that force is proportional to acceleration (Newton's Second Law) and there is a link between tibial acceleration and tibial stress fractures (Milner et al., 2006), pelvic acceleration provides a useful surrogate measure of pelvic loading, and may link to risk of pelvic stress fractures and pelvic girdle pain. Wearable devices are particularly useful when retraining running gait, to modify the risks for developing lower limb injuries, such as altering stride frequency (Bramah et al., 2019). An increase in stride frequency has been associated with reductions in tibial acceleration, vertical ground reaction and braking forces (Busa et al., 2016; Heiderscheit et al., 2011; Napier et al., 2019), as well as improved clinical outcomes in patellofemoral pain (Bramah et al., 2019). It is also an easily applied, self-regulated strategy that can be maintained beyond the initial intervention (Bramah et al., 2019). However, it is not yet known whether increases in stride frequency, and the associated reductions in distally measured biomechanical lower limb injury risk factors, translate into reductions in external measures of acceleration at the pelvis.

In order to alter stride frequency, stride time; comprised of contact time and aerial time (Morin et al., 2007), must change. Whereas increases in stride frequency from baseline have typically been associated with a reduction in contact time (Heiderscheit et al., 2011), reductions in stride frequency are sometimes achieved by maintaining contact time, but increasing aerial time (i.e., more time in the air between steps) (Morin et al., 2007). Manipulating contact time and stride frequency alters leg stiffness, which is the ratio of maximal vertical ground reaction force to maximal leg compression (Morin et al., 2007). Compared to normal running, increased stride frequency and shorter contact time was associated with increased leg stiffness, yet, only a longer contact time was associated with a reduced leg stiffness (Morin et al., 2007). Increasing leg stiffness, potentially by increasing joint stiffness, may lead to less dissipation of ground reaction forces proximally through the body and subsequently greater pelvic acceleration. High joint stiffness has been shown to increase the odds of sustaining overuse injuries (Messier et al., 2018) and is able to differentiate between runners with and without low back pain (Hamill et al., 2009). Therefore, greater leg stiffness when manipulating stride frequency could also negatively impact pelvic loading. However, effects of stride frequency and contact times on pelvic acceleration are yet to be explored.

Alternatively, pelvic acceleration could be affected by the magnitude of the initial impact shock. At a constant running velocity, changes in stride length accompany changes in stride frequency (Bailey et al., 2017). Runners may achieve the same stride length through landing with various degrees and combinations of hip or knee flexion, as indicated by the horizontal anteroposterior displacement from the ankle to both the knee and hip at landing (Lieberman et al., 2015). Landing with a reduced anteroposterior displacement from the knee to the ankle (i.e., greater knee flexion) is associated with reduced vertical peak impact forces, whereas a greater displacement from the hip to ankle has been associated with increased peak braking forces (Lieberman et al., 2015). Additionally, increased stride frequency is associated with reduced tibial acceleration, whereas head acceleration remains unchanged due to adapted levels of shock attenuation (Busa et al., 2016). However, relatively little is known about pelvic acceleration. Clarity is needed to confirm whether the level of dissipation or the magnitude of the initial shock (e.g., tibial acceleration) influences pelvic acceleration.

Considering the variety of factors that change when stride frequency is manipulated, it would be beneficial to determine whether any changes in stride frequency directly influence changes in pelvic acceleration, or whether any changes in pelvic acceleration are achieved indirectly through intermediary variables. Greater understanding of the association between tibial acceleration and pelvic acceleration may also provide insights into what extent initial tibial acceleration is dissipated from the tibia to the pelvis. Prior work has assessed shock attenuation from the tibia to the head (Busa et al., 2016; Dufek et al., 2009), but attenuation from the tibia to the pelvis during running is less understood. It may also be useful to explore whether contact time and stride frequency interact to predict pelvic acceleration, due to the typically inverse relationship observed between the two (Morin et al., 2007). These insights may aid in attribution of the correct predictor variable to any reductions found in pelvic acceleration, allowing the design of gait retraining strategies to be appropriately targeted.

The aim of this study was to assess which biomechanical variables influence changes in external peak pelvic acceleration during treadmill running, across various stride frequency conditions. It was hypothesised that i) increased stride frequency would be associated with decreased vertical, anteroposterior and resultant pelvic acceleration, ii) decreased contact time would be associated with increased vertical, anteroposterior and resultant pelvic acceleration, iii) increased vertical, anteroposterior and resultant tibial acceleration would be associated with increased vertical, anteroposterior and resultant pelvic acceleration, respectively, and iv) decreased anteroposterior displacement from the knee to the ankle would be associated with reduced vertical and resultant pelvic acceleration, whereas decreased anteroposterior displacement from and hip to ankle would reduce anteroposterior pelvic acceleration.

Materials and methods

Fourteen healthy runners took part in the study, providing written, informed consent. Recruitment for this study commenced on 15th September 2020 and ended on 16th December 2020. Two participants were excluded due to data loss. Therefore, data from twelve healthy runners (7 female, 5 male, mean (SD): 28.3 (5.9) years, 67.1 (12.0) kg, 1.70 (0.09) m) were analysed. Inclusion criteria required participants to run at least twice per week for a minimum of 30 minutes per run. Participants were not eligible to participate in the study if they had history of anterior knee pain, current lower-limb injuries, neurological impairments, cardiovascular pathologies, or were pregnant. Participants completed Physical Activity Readiness questionnaires and demographics forms to ensure their suitability to participate. Participants self-reported running a mean (SD) of 5.3 (3.0) times and 61.8 (51.8) km per week and had been running for a mean time of 9.2 (6.1) years. Seven participants self-classified as recreational runners and five self-classified as competitive runners. Ethical approval was gained from Cardiff Metropolitan University's Ethics Committee (project reference number: sta-2663).

The study followed an experimental, repeated measures design. Participants completed a warm-up at a self-selected speed, up to a maximum of 9 $km \cdot h^{-1}$ and familiarised themselves with running on the laboratory treadmill (Sprintex Ortho Treadmill, SPRINTEX Trainingsgeräte GmbH, Kleines Wiesental, Germany) in their normal running shoes for six minutes. All subsequent trials were undertaken at 9 $km \cdot h^{-1}$ for every participant, so that comparisons could be made between conditions, and the velocity was low enough to accommodate the adoption of a range of stride frequency conditions, as seen in previous research (Farley & González, 1996). Initially, participants performed a control trial, where they ran for one-minute. During the last 20 seconds of this trial, the Runmatic iPad application (Balsalobre-Fernández et al., 2017) was used to establish participants' preferred stride frequency (Hz) via attainment of video data (240 Hz) and digitisation of initial contact and toe-off events. Calculation of a two-way mixed intraclass correlation coefficient (ICC(3,1)) showed that the reliability (absolute agreement) of each participant's preferred stride frequency across the three consecutive 10-second intervals in the last 30 seconds of the trial was excellent ($r = 0.91$, 95% CI = 0.778 – 0.969, $p < 0.001$). The preferred stride frequency value was then used to calculate a pulse period, producing a metronome beat that equated to the preferred stride frequency. A pulse period for ± 5% and ± 10% of this preferred stride frequency was also calculated. In a randomised order, five one-minute trials were then performed where participants ran at their preferred stride frequency, and at ± 5% and ± 10% of this preferred stride frequency, dictated by an audible metronome. Participants were asked to synchronize foot strike frequencies with the metronome beat.

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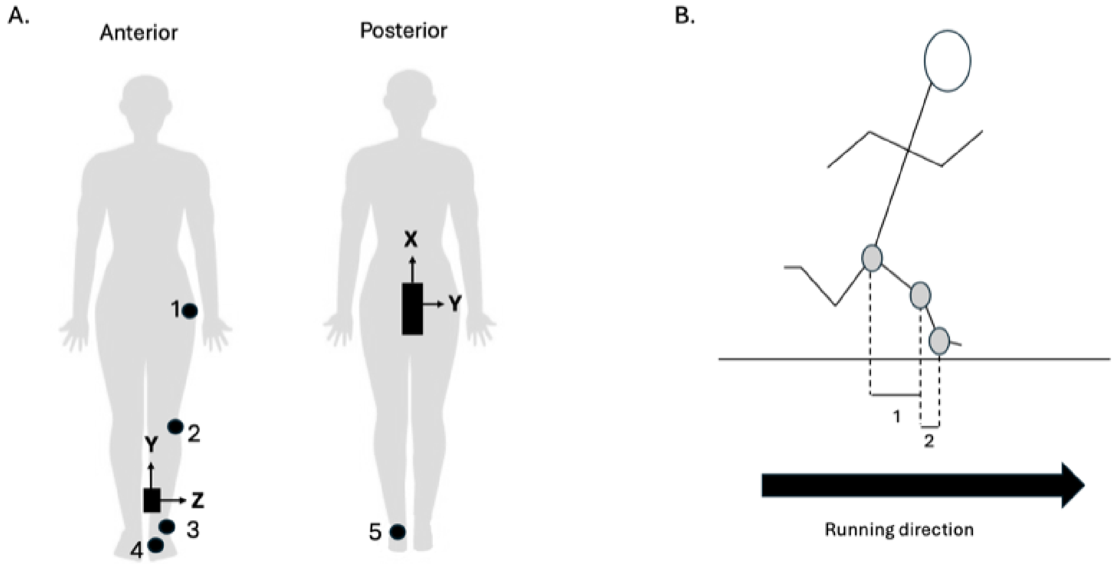

Figure 1. (A) Placement of markers (black circles) and inertial measurement units (IMU; black rectangles) on the anterior (left) and posterior (right). Markers were placed at the greater trochanter (1), lateral epicondyle (2), lateral malleolus (3), head of the 2nd metatarsal (4), and calcaneus (5). The anterior view shoes the tibial IMU, where the x axis is in the anteroposterior direction, the posterior view shows the sacrum IMU, where the z axis is in the anteroposterior direction. (B) Runner in the sagittal plane showing the anteroposterior displacements from hip (greater trochanter marker) to knee (lateral epicondyle marker; 1) and from knee (lateral epicondyle marker) to ankle (lateral malleolus marker; 2).

Kinematic data were collected using a motion capture system (200 Hz; Nexus 2.11, Vicon, Oxford, UK). Reflective markers (radius 14 mm) were placed on the left lower limb on the greater trochanter (hip), lateral epicondyle (knee), lateral malleolus (ankle), head of the second metatarsal (toe) and calcaneus (heel; Figure 1). Inertial measurement units (IMU; 225 Hz; Blue Trident, Vicon Motion Systems Ltd., Oxford, UK; mass: 12 g; dimensions 40 mm x 30 mm x 15 mm) were placed on the pelvis and left distal tibia. The pelvic IMU was placed specifically on the sacrum, as seen in previous research (Reenalda et al., 2016), and tibial IMU on the antero-medial surface to more closely resemble acceleration of the bone than the proximal tibia (Sheerin et al., 2019) and minimise movement artefact due to wobbling mass (Figure 1). Sampling frequencies of 200 Hz have been reported to be acceptable when measuring peak tibial acceleration during running (Mitschke et al., 2017), indicating that our sampling rate was appropriate (225 Hz). The pelvic IMU was attached to the skin using double-sided tape and overlayed with kinesiology tape (Reenalda et al., 2016), and the tibial IMU was attached using a Velcro strap. The IMUs were positioned so that acceleration posterior and upwards from the pelvis was positive and acceleration upwards from the tibia was positive and posterior from the tibia was negative (providing data as shown in Figure 2). Video data (100 Hz), synced with the motion capture system (including the IMUs), were also captured in the sagittal plane, allowing initial contact and toe-off events to be digitised and identified across all devices. For each left foot stride, the first visible frame of left foot contact with the treadmill was identified as initial contact, and the first frame where the left foot subsequently left the treadmill was identified as the corresponding toe-off.

Kinematic data were labelled, and acceleration data were automatically up sampled to 400 Hz via linear interpolation in Nexus (2.11, Vicon, Oxford, UK) in order to synchronise with the other devices (optical and video cameras). Coordinate and acceleration data were filtered with a low-pass, fourth order, recursive Butterworth filter. Cut-off frequencies for the coordinate, pelvic and tibial acceleration data were 13 Hz, 10 Hz and 70 Hz respectively, determined via residual analysis (Winter, 2009) and visual inspection. A custom Matlab code (MATLAB, Mathworks Inc., Natick, MA, USA) based on previous methods (Moe-Nilssen, 1998) aligned the acceleration data to the global axes (vertical, anteroposterior and mediolateral) and subtracted gravity from the vertical acceleration, so that the acceleration reported was purely due to motion. The resultant acceleration was determined from the three raw, unaligned acceleration components and subsequently filtered.

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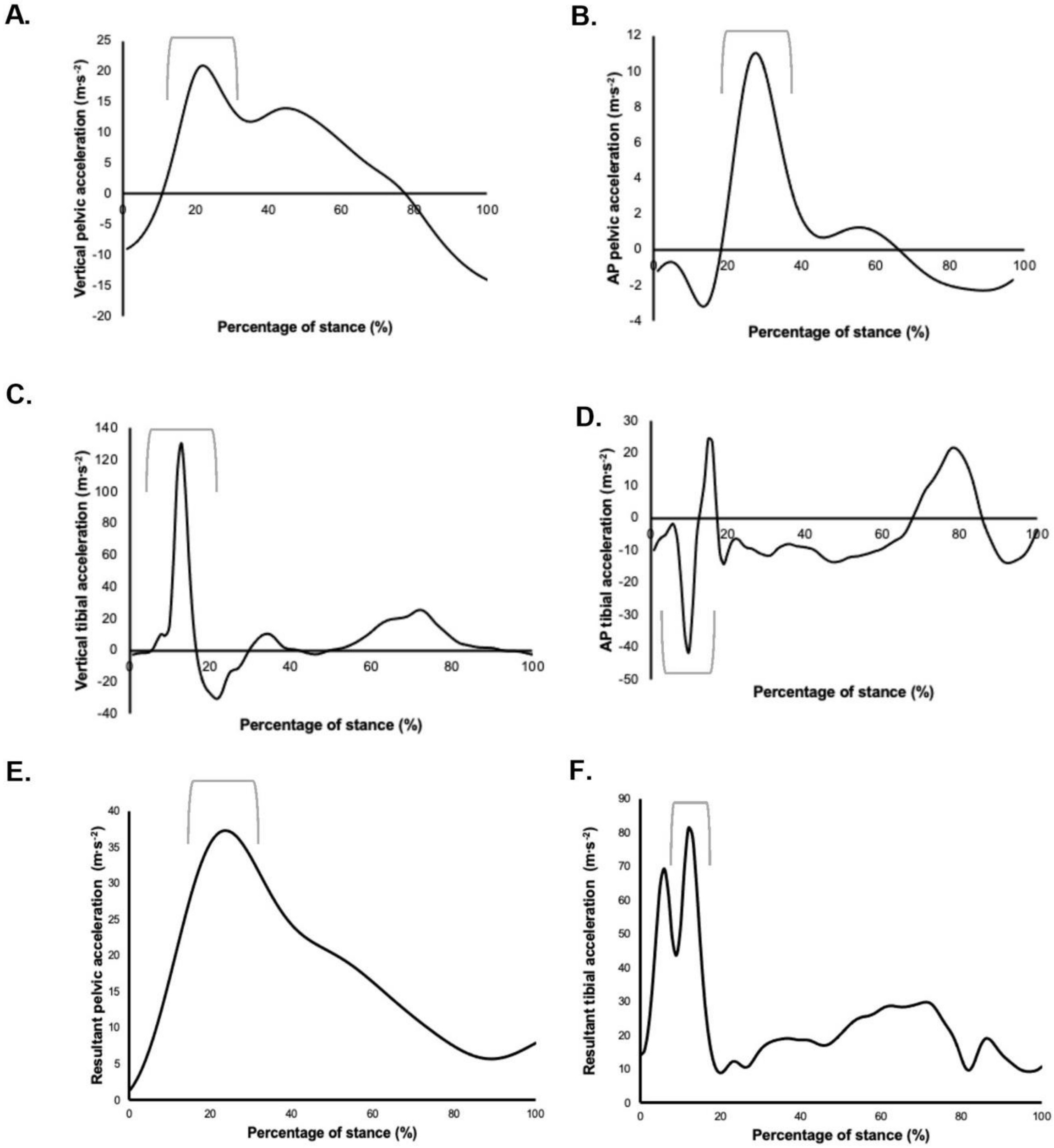

Figure 2. An example of pelvic and tibial acceleration for the stance phase of a step. (A) Vertical pelvic acceleration. (B) Anteroposterior (AP) pelvic acceleration. (C) Vertical tibial acceleration. (D) Anteroposterior (AP) tibial acceleration. (E) Resultant pelvic acceleration. (F) Resultant tibial acceleration. The peak taken from the stance phase is indicated by the grey bracket.

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Peak positive vertical, anteroposterior and resultant pelvic accelerations were identified for each stance phase (Figure 2 A, B, E). For tibial acceleration, corresponding positive vertical and resultant acceleration peaks, and negative anteroposterior acceleration peaks were identified (Figure 2 C, D, F). In the case of a double resultant tibial acceleration peak (Figure 2F), the largest peak was always selected (Garcia et al., 2021). The anteroposterior horizontal displacements (cm) from the knee and hip to the ankle, at the corresponding initial contacts were also attained from motion capture data (Figure 1) and contact time was determined by the time between initial contact and subsequent toe-off events, analysed in Nexus (2.11, Vicon, Oxford, UK). The stride frequency (Hz) achieved by participants was verified, using the digitised initial contact events. Shock attenuation was calculated for each step using the following equation: *Shock attenuation (%)* = [1 – (*peak pelvic acceleration* / *peak tibial acceleration*)] *100. This equation was adapted from previous research (Dufek et al., 2009), to calculate shock attenuation between the tibia and pelvis, rather than the tibia and head. Variables were averaged (mean) over the last ten (Riley et al., 2008) left stance phases of each trial.

Three linear mixed models (LMM) assessed predictors of peak vertical (LMM1), anteroposterior (LMM2) and resultant (LMM3) pelvic acceleration. Predictors included the corresponding peak tibial acceleration component, displacement from knee to ankle (LMM1&3) and/or hip to ankle (LMM2&3), stride frequency and contact time. To address the issue of independence of observations, "Participant" was used as a random grouping effect to account for repeated measures, and predictor variables were entered as fixed effects. Models used maximum likelihood estimation and statistical significance was accepted at alpha level .05. A contact time and stride frequency interaction was then added into each LMM, checked with a likelihood ratio test, to assess whether the inclusion of the interaction term significantly improved the models. If inclusion of the interaction improved the model (p < .05), the interaction was kept. In the case of a significant interaction, an estimated marginal means analysis was conducted. Values of contact time and stride frequency were fixed at the group mean and two standard deviations above and below the group mean to model and understand the interaction effect on peak pelvic acceleration. Predictors were standardised to z-scores in all models to allow simpler interpretation of dependent variable coefficients (magnitude of change for one standard deviation change in the predictor variables) and to aid estimation of the interaction terms. All statistical analyses were undertaken in R.

Results

Descriptive data showed that stride time decreased by 0.10 s. from the lowest stride frequency condition to the highest (Table 1). Further, contact time generally decreased as stride frequency increased, but the change between the lowest and highest stride frequency was small (0.03 s; Table 1). Vertical and resultant peak pelvic acceleration tended to increase as stride frequency increased, whereas anteroposterior pelvic acceleration decreased (Table 1). In contrast, vertical and resultant peak tibial acceleration decreased as stride frequency increased, with more variation evident for anteroposterior tibial acceleration (Table 1). These fluctuations in peak pelvic and tibial acceleration impacted on the corresponding shock attenuation observed for each condition (Table 1).

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Table 1. Group means (SD) for variables of interest, calculated from the last ten stance phases from each trial, for each stride frequency condition.

Variable	Stride frequency condition				
	–10%	–5%	Preferred	+5%	+10%
Stride frequency achieved (Hz)	1.27 (0.09)	1.29 (0.08)	1.35 (0.08)	1.42 (0.08)	1.46 (0.10)
Stride frequency change (%)	–6.41 (2.86)	–4.31 (1.72)	-	4.89 (2.46)	8.36 (4.53)
Stride time (s)	0.79 (0.05)	0.78 (0.05)	0.74 (0.04)	0.71 (0.04)	0.69 (0.05)
Contact time (s)	0.31 (0.03)	0.30 (0.03)	0.30 (0.03)	0.29 (0.03)	0.28 (0.03)
Vertical tibial acceleration ($m \cdot s^{-2}$)	63.92 (32.09)	63.02 (25.63)	61.41 (30.91)	57.07 (26.64)	51.58 (24.37)
AP tibial acceleration ($m \cdot s^{-2}$)	–72.11 (31.24)	–62.93 (24.24)	–64.26 (18.73)	–70.34 (24.87)	–75.66 (23.31)
Resultant tibial acceleration ($m \cdot s^{-2}$)	105.42 (27.82)	96.11 (22.40)	93.24 (26.84)	93.19 (25.34)	89.58 (24.67)
Vertical pelvic acceleration ($m \cdot s^{-2}$)	21.85 (4.71)	22.73 (4.75)	22.96 (4.69)	23.17 (4.84)	23.56 (5.15)
AP pelvic acceleration ($m \cdot s^{-2}$)	7.76 (3.11)	7.63 (3.25)	7.19 (3.24)	6.86 (3.53)	6.59 (3.53)
Resultant pelvic acceleration ($m \cdot s^{-2}$)	34.03 (5.32)	35.16 (4.72)	35.39 (4.55)	35.69 (5.31)	35.57 (5.51)
Vertical shock attenuation (%)	58.96 (19.76)	58.95 (18.04)	54.69 (22.88)	51.28 (22.99)	46.45 (21.17)
AP shock attenuation (%)	86.22 (9.36)	84.47 (12.97)	87.48 (7.58)	88.47 (8.08)	90.56 (5.51)
Resultant shock attenuation (%)	65.70 (9.91)	62.15 (7.72)	59.76 (10.74)	58.69 (14.75)	57.50 (14.01)
Displacement from knee to ankle (cm)	–0.53 (2.22)	–0.15 (2.61)	–0.61 (2.77)	–1.28 (2.46)	–1.34 (2.34)
Displacement from hip to ankle (cm)	15.92 (2.86)	16.25 (3.01)	15.46 (3.15)	14.60 (3.12)	14.60 (2.79)

Note: A positive displacement from the knee to ankle indicates the ankle is anterior to the knee. AP denotes anteroposterior. Shock attenuation is displayed as a percentage of corresponding tibial acceleration. Stride frequency change (%) describes the actual change in stride frequency achieved, relative to the preferred stride frequency condition. A negative value represents a reduced stride frequency compared to preferred stride frequency.

In the statistical testing of predictors of pelvic acceleration, inclusion of the interaction term significantly improved both the vertical and resultant models (vertical: p = .003, resultant: p = .005), and therefore the interaction was included (Table 2; LMM4&5). However, the inclusion of the interaction term did not significantly improve the anteroposterior model (p = .890) and therefore it was omitted. The anteroposterior model also showed no predictors of peak pelvic acceleration (p = .886; Table 2; LMM2). One standard deviation decrease in resultant peak tibial acceleration predicted a 1.19 $m \cdot s^{-2}$ increase in resultant peak pelvic acceleration (p = .010; Table 2; LMM5).

Additionally, stride frequency and contact time significantly interacted to predict peak pelvic acceleration in both the vertical (p = .006) and resultant (p = .009) model (Table 2; LMM4&5). Analysis of the interaction showed that participants with a contact time and stride frequency that was below average for this cohort, had significantly higher peak vertical and resultant pelvic accelerations than those with below-average contact times, but average (vertical: p = .007, predicted mean difference = 4.607; resultant: p = .016, predicted mean difference 4.437; Figure 3) or high stride frequencies (p = .007; predicted mean difference = 9.213; resultant: p = .016, predicted mean difference = 8.875; Figure 3). Additionally, by the linear nature of the analysis, the same predicted mean difference between average and high stride frequencies was seen between low and average (vertical: p = .007, predicted mean difference = 4.607; resultant: p = .016, predicted mean difference 4.437; Figure 3). For those with contact times that were average or above average for this cohort, stride frequency did not affect peak vertical or resultant pelvic acceleration (Vertical - average contact time: p = .903; above average contact time: p = .169 Resultant -average contact time p = .986, above average contact time p = .180; Figure 3).

Table 2. Linear mixed model outcomes for predicting peak vertical, anteroposterior and resultant pelvic acceleration.

LMM	Dependent Variable	Fixed Factors	Coefficient (SE)	*p*
1	Vertical	Vertical tibial acceleration	–0.46 (0.59)	.446
	pelvic	Displacement from knee to ankle	–0.46 (0.83)	.581
	acceleration	Stride frequency	–0.72 (0.49)	.147
		Contact time	–2.20 (0.90)	.017*
2	AP pelvic	AP tibial acceleration	–0.12 (0.17)	.510
	acceleration	Displacement from hip to ankle	0.07 (0.49)	.892
		Stride frequency	–0.36 (0.22)	.115
		Contact time	0.44 (0.44)	.322
3	Resultant	Resultant tibial acceleration	–1.13 (0.47)	.020*
	pelvic	Displacement from knee to ankle	–0.95 (0.91)	.297
	acceleration	Displacement from hip to ankle	1.20 (1.13)	.291
		Stride frequency	–0.68 (0.55)	.218
		Contact time	–2.27 (1.01)	.029*
4	Vertical	Vertical tibial acceleration	–0.55 (0.56)	.328
	pelvic	Displacement from knee to ankle	–0.30 (0.78)	.697
	acceleration	Stride frequency	–0.22 (0.49)	.661
		Contact time	–1.50 (0.88)	.093
		Stride frequency * Contact time	1.04 (0.36)	.006*
5	Resultant	Resultant tibial acceleration	–1.19 (0.44)	.010*
	pelvic	Displacement from knee to ankle	–1.08 (0.86)	.213
	acceleration	Displacement from hip to ankle	1.91 (1.10)	.087
		Stride frequency	–0.09 (0.56)	.873
		Contact time	–1.75 (0.97)	.077
		Stride frequency * Contact time	1.06 (0.39)	.009*

Note: A positive displacement from the knee to ankle indicates the ankle is anterior to the knee. Inclusion of the interaction term significantly improved model 1 and 3 ($p < .05$) but not model 2 ($p > .05$). In text results therefore relate to models 2, 4 and 5. *Significant at .05 level. SE = standard error. LMM = Linear mixed model. AP = Anteroposterior.

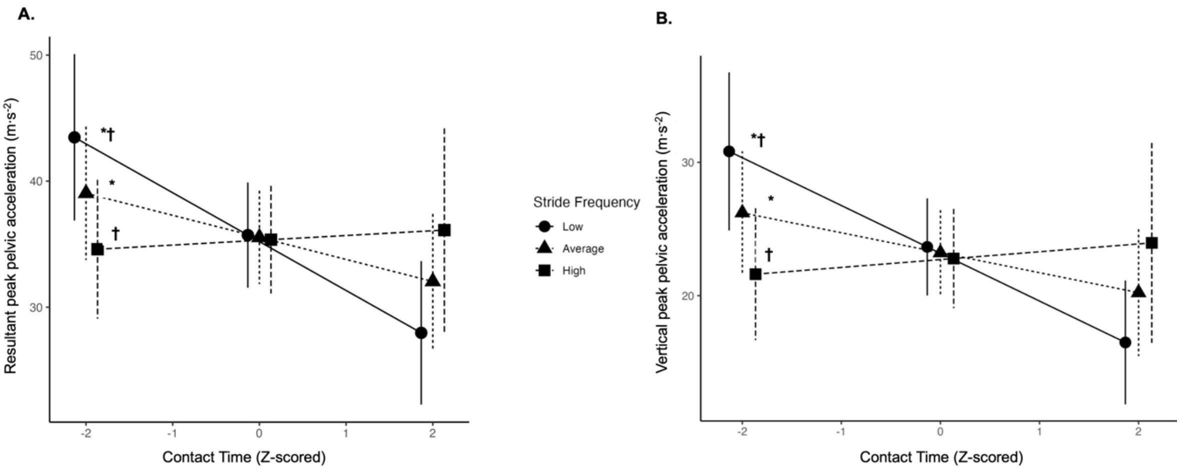

Figure 3. The modelled interaction of stride frequency and contact time. Effect of the interaction on (A) resultant and (B) vertical peak pelvic acceleration. Values of contact time and stride frequency are fixed at: Sample mean (contact time = 0; stride frequency = Average); two standard deviations below the mean (contact time = –2; stride frequency = Low); two standard deviations above the mean (Contact time = 2; stride frequency = High) to demonstrate the effect on vertical and resultant peak pelvic acceleration. * Denotes significantly different from high stride frequency. † Denotes significantly different from average stride frequency.

Discussion and implications

This study investigated which biomechanical variables influence changes in external peak pelvic acceleration during treadmill running, across various stride frequency conditions. Stride frequency and contact time interacted to predict vertical and resultant peak pelvic acceleration. When modelled, the interaction showed that short contact times and low stride frequencies produced higher vertical and resultant peak pelvic accelerations than short contact times and average or high stride frequencies. A decrease in resultant tibial acceleration also predicted an increase in resultant pelvic acceleration, however vertical or anteroposterior tibial acceleration did not predict vertical or anteroposterior pelvic acceleration. These findings suggest that an accelerometer placed on the pelvis is necessary if clinicians are interested in altering or assessing pelvic accelerations and that caution is warranted extrapolating tibial accelerations to pelvic accelerations.

Stride frequency, independently, did not predict any component of pelvic acceleration, with hypothesis one unsupported. Increased stride frequency has previously been associated with increased leg stiffness (Morin et al., 2007) which may reduce attenuation of ground reaction forces. The descriptive data supports this, with lower vertical and resultant shock attenuation in the higher stride frequency conditions (Table 1). Interestingly, anteroposterior shock attenuation generally increased under the same conditions and produced a much higher level of shock attenuation than the vertical direction across all stride frequencies (Table 1). Therefore, the lower limb appears able to attenuate a greater proportion of horizontal acceleration than vertical acceleration during treadmill running. Notably, trends for peak pelvic acceleration were the opposite to trends in peak tibial acceleration. Specifically, descriptive data showed that vertical and resultant peak tibial acceleration increased, whilst anteroposterior peak tibial acceleration generally decreased, as stride frequency decreased (Table 1), in line with previous research (Busa et al., 2016; Giandolini et al., 2015). Despite a relationship being found between stride frequency and tibial acceleration previously (Giandolini et al., 2015), the pelvis is more proximal in the kinetic chain. Therefore, there are more degrees of freedom within the musculoskeletal system that may mediate the relationship between peak pelvic acceleration and stride frequency than there are for peak tibial acceleration. This is likely to explain why stride frequency, independently, did not predict any component of pelvic acceleration.

Despite our second hypothesis, that a decreased contact time would independently predict increased pelvic acceleration being unsupported, stride frequency and contact time interacted to predict vertical and resultant peak pelvic acceleration. When values were modelled, short contact times and low stride frequencies produced higher vertical and resultant pelvic acceleration than short contact times but high stride frequencies. For longer modelled contact times, pelvic acceleration was generally lower than shorter contact times, however stride frequency did not influence peak vertical or resultant pelvic acceleration (Figure 3). A longer contact time has been associated with reduced leg stiffness (Morin et al., 2007), potentially leading to greater shock attenuation and lower pelvic acceleration. Additionally, an increased contact time allows a longer time period to transfer and attenuate load during running, potentially leading to more gradual production of a later but smaller acceleration peak.

The findings indicate that at short contact times, increasing stride frequency, commonly used as a strategy to reduce lower limb loading, may translate into reductions in peak vertical and resultant acceleration at the pelvis. In this cohort, there were smaller changes in contact time (0.03 s) compared to stride time (0.1 s), across stride frequency conditions (Table 1), indicating that greater changes occurred within swing time. The ratio of stride time to contact time is known as duty factor (Bonnaerens et al., 2021). Increasing stride frequency, through a decrease in swing time when contact time is maintained, leads to an increase in duty factor. An increased duty factor can be achieved by employing a grounded running technique, that is, running without a flight phase, where duty factor is above 50% (Bonnaerens et al., 2019). Higher duty factors have been associated with lower peak vertical ground reaction forces and peak braking forces, to a greater extent than stride frequency (Bonnaerens et al., 2021). This may suggest that a gait retraining strategy associated with reduced lower limb loading, such as increasing duty factor (an increased contact time to stride time ratio), may also translate into changes at the pelvis. However, this requires further examination. Grounded running is often accompanied by an increased stride frequency or reduced speed (Bonnaerens et al., 2019). Further investigations should therefore also consider the effect of this on cumulative load at the pelvis, as although per step metrics may be lower, cumulative loads have been shown to increase at the knee for increased steps at slower speeds (Petersen et al., 2015).

Peak vertical and anteroposterior tibial acceleration did not predict peak vertical and anteroposterior pelvic acceleration, nor did the anteroposterior displacement variables included in the models. Therefore, hypotheses three and four were unsupported. In contrast, resultant tibial acceleration was negatively associated with resultant pelvic acceleration, suggesting that initial tibial shock was more influential for the resultant rather than individual acceleration vectors. The negative association means lower resultant pelvic accelerations corresponded with higher resultant tibial accelerations. This may appear counter-intuitive, if the magnitude of tibial acceleration was the driving mechanism for pelvic acceleration, as one might expect a greater tibial acceleration, and therefore shock that needs attenuating, to produce a greater pelvic acceleration.

We hypothesise that resultant peak pelvic acceleration is influenced to a greater extent by spatiotemporal characteristics, such as contact time, and, proximal active and passive attenuating mechanisms (Pratt, 1989) of peak resultant tibial acceleration rather than the magnitude of peak tibial acceleration per se and segment geometry. Passive mechanisms include ligament and muscle oscillations, whilst active mechanisms include joint stiffness and muscle activation. Specifically, active mechanisms proximal to the tibia are hip and knee joint stiffness and thigh muscle activations. Thigh muscle activations increase at faster stride frequencies as the lower limb muscles pre-activate prior to initial contact (Chumanov et al., 2012) and are effective at attenuating high frequency shocks (Boyer & Nigg, 2007). Additionally, knee joint stiffness was found to be greater in runners with low back pain, indicating decreased attenuation compared to those without low back pain (Hamill et al., 2009). Further to this, increased knee joint stiffness increased the odds of injury in high level runners, potentially indicating the clinical impact that this decreased attenuation poses (Messier et al., 2018). It is possible that lower resultant tibial acceleration being associated with higher resultant pelvic acceleration could be due to increased lower limb stiffness, and therefore higher transmission of shock, when tibial acceleration is low. Reduced tibial accelerations typically occur at increased stride frequencies (Busa et al., 2016; Schubert et al., 2014), which have been associated with increased leg stiffness (Morin et al., 2007) and demonstrated decreased resultant shock attenuation in this study (Table 1). The potential for variation in shock transmission, and lack of predictors identified by this study, suggests that placing an IMU on the pelvis is required to estimate pelvic acceleration, rather than one placed further down the kinetic chain on the tibia or undertaking a visual gait assessment of anteroposterior displacement variables.

This study explores predictors of externally measured pelvic acceleration only and should be interpreted as such. That is, in this manuscript we do not make claim to any findings regarding internally measured bone loading at the pelvis or related to injury occurrence or pain reductions, with theoretical links between pelvic acceleration and injury. While there is potential influence of wobbling mass affecting externally measured pelvic acceleration, the measure that we have provided is an accessible, non-invasive and useful way of gaining information pertaining to the loading of the pelvis. It also allows the development of future ecologically valid, field-based studies that allow data collection in real-life sporting environments (e.g., for outdoor running), in addition to more traditional laboratory-based studies. Participants were constrained to a set speed for this study, which may have altered their running style, however, originally, prior to COVID, this study was part of a larger project where it was important to control for speed, due to group comparisons, therefore this was a necessary constraint. Not all participants achieved the desired change in stride frequency for the more extreme conditions (±10%; Table 1). While this should be acknowledged when interpreting the descriptive results (Table 1), the LMMs used the achieved stride frequency as a predictor of pelvic acceleration, rather than the desired change or assessing differences between conditions. The difficulty in achieving these changes in stride frequency should therefore be taken into account when considering practicality of these strategies, however, do not affect the interpretation of the LMM results, specifically.

The findings of this study suggest that for runners with short contact times, increasing contact time or stride frequency may reduce pelvic acceleration. Future prospective studies are needed to assess the theoretical link between pelvic acceleration, injury and pain. These should incorporate participants with pelvic pain and assess whether these proposed strategies, and any changes in pelvic acceleration, also translate into changes in pain and/or pelvic injury incidence. Future investigations should also monitor any effects of these changes on cumulative and lower limb loading, to verify that there are no unintended adverse effects when adopting these strategies.

Conclusion

A stride frequency and contact time interaction was evident when predicting peak pelvic acceleration during treadmill running. When modelled, short contact times and high stride frequencies produced lower vertical and resultant peak pelvic acceleration than those with short contact times and lower stride frequencies. For longer contact times, stride frequency did not significantly affect vertical or resultant peak pelvic acceleration. Increasing contact time, or increasing stride frequency at shorter contact times, may therefore be useful in reducing pelvic acceleration during treadmill running. Future research should investigate this further, as well as the potential of these strategies to reduce pelvic pain in runners. Peak tibial acceleration and anteroposterior displacement variables did not predict peak vertical or anteroposterior pelvic accelerations. Thus, spatiotemporal variables and lower limb shock attenuation mechanisms appear more important for pelvic acceleration than the magnitude of tibial acceleration, and caution is warranted extrapolating tibial accelerations to pelvic accelerations.

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