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Increasing step rate reduces insole force and cumulative load in college runners

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Increasing Step Rate Reduces Insole Force and Cumulative Load in College Runners

By

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ABSTRACT

Competitive distance runners are at high risk of injuries, which result in financial burdens and impede performance. Running injuries are usually overuse injuries resulting from mechanical fatigue in repetitively loaded musculoskeletal structures. Mechanical fatigue refers to the accumulation of microstructural damage due to a combination of the number of loading cycles and load magnitude, potentially resulting in mechanical property degradation and ultimately failure. Increasing step rate (the number of steps per minute) has been proposed to reduce the risk of running injuries by reducing load magnitude during running. Based on fatigue failure behavior of musculoskeletal structures, the reduction in structure load magnitude in response to increased step rate should reduce the amount of damage accumulated for a given running speed and distance despite the increased number of loading cycles. Testing this hypothesis is difficult because of the difficulty in quantifying fatigue damage. However, in level running at submaximal speeds, muscle-tendon and bone forces increase with increasing vertical ground reaction force (vGRF). Therefore, changes in vGRF in response to increased step rate may reflect changes in structure-specific loads and cumulative damage. Further, peak vGRF can be approximated using force sensing insoles, allowing ecologically valid observations. To assess the efficacy of increasing step rate to reduce peak vGRF and potential cumulative damage in college runners, we examined changes in peak insole force and cumulative weighted peak force (CWPF), based on fatigue failure behavior of musculoskeletal structures, with increased step rate. We also evaluated the use of sacral acceleration, a correlate of vGRF, to detect these changes. 12 collegiate distance runners ran on an outdoor track at 3.83 m/s ± 5% for 1000m at their preferred step rate and at a 10% increased step rate while insole force and sacral acceleration were recorded. Average peak insole force and CWPF per kilometer decreased

significantly ($p < 0.001$) with increased step rate, suggesting increasing step rate can reduce peak vGRF and a general measure of cumulative damage in college runners. Changes in sacral acceleration measurements were consistent with changes in force measurements and peak acceleration correlated with peak insole force on an individual basis (mean $r = 0.62$), supporting the use of sacral acceleration to detect changes in vGRF and potential relative changes in structure loads and cumulative damage with increased step rate. These results suggest clinicians should consider interventions targeting an increase in step rate to help reduce the risk of injury in college runners and that the potential efficacy of those interventions can be evaluated using field-based accelerometry as a more accessible alternative to measuring forces.

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INTRODUCTION

Distance running is a popular sport associated with high rates of musculoskeletal injury. More than 18 million Americans registered for races in 2018 (Running USA, 2019). In the National Collegiate Athletics Association (NCAA), more than 27,000 distance runners compete on cross country teams (NCAA, 2021). Additionally, there are approximately half a million high school cross country runners (NFHS, 2019). Injury prevalence among runners of all levels ranges from 10-92% based on the definition of injury, with an overall prevalence of 42.7% (Francis et al., 2019). Depending on level, runners suffer 2.5 to 33.0 injuries per 1000 hours of training (Videbaek et al., 2015).

Injury prevalence is high specifically in elite and competitive runners. Between 56 and 77% of elite and competitive distance runners sustain time loss injuries each year (D'Souza, 1994; Jacobsson et al., 2013; Lysholm et al., 1987; Ristolainen et al., 2010) and 42-52% suffer multiple injuries per year (Jacobsson et al., 2013; Ristolainen et al., 2010). NCAA cross country runners experience 2.85 (men) to 3.44 (women) time loss injuries per 1000 athlete exposures, resulting in 2.81 time loss injuries per men's team and 3.56 time loss injuries per women's team during each cross country season (Kerr et al., 2016).

Running-related injuries result in financial burdens. Each injury incurs, on average, $194 in direct (health care) and indirect (loss of productivity) costs (Hespanhol Junior et al., 2016). Given recreational runners train on average 3 hours per week (Hespanhol Junior et al., 2013), a conservative estimate of the total annual cost of running-related injuries in America exceeds $1.3 billion.

Time loss injuries also limit performance. Raysmith and Drew (2016) found that elite track and field athletes who completed at least 80% of training weeks were seven times more likely to achieve performance objectives than those who completed less than 80% of training weeks. For elite and competitive athletes, injuries resulting in reduced training time negatively impact their athletic career.

Increasing step rate is often proposed as an intervention to reduce the risk of running-related injuries. Investigations of the association between step rate and injury risk in different populations of runners show contrasting results. A recent prospective study (Kliethermes et al., 2021) of 54 NCAA Division 1 cross country runners over three seasons investigated the associations between whole body biomechanical variables and bone mineral density and the risk of bone stress injury. A higher step rate at a self-selected moderate intensity speed was independently associated with bone stress injury risk, decreasing the risk of bone stress injury by 5% for every 1 step/min increase in step rate. Another prospective study in a smaller sample of 29 NCAA Division 3 cross country runners (Luedke & Rauh, 2021) found a nonsignificant difference in step rate at self-selected speed between injured runners and those who remained uninjured. Luedke et al. (2016) examined the relationship between step rate at both fixed and self-selected speeds and shin and knee injuries in high school runners. Individuals in the lowest tertial or half of measured step rate at fixed speed were 5.3 to 6.7 times more likely to incur a shin injury compared to runners in the upper tertial or half. Similar significant relationships were found for step rate at self-selected speed. No significant relationship was observed between step rate and anterior knee pain, potentially due to the low number of knee injuries. One study of adult recreational runners (Szymanek et al., 2020) showed no significant relationship between step rate during a 2mi or 5k physical fitness test and the risk of injuries resulting in time loss of at least seven days.

The differences in these results could be due to differences in running speed, sample size, injury definition, and population of runners. The average self-selected speed of the collegiate runners observed by Luedke and Rauh (2021), 4.6 m/s (5:50/mi), was faster than that of the collegiate runners studied by Kliethermes et al. (2021), mean = 3.87 m/s (6:56/mi), and may not represent the majority of training volume for collegiate runners. In contrast, Szymanek et al. (2020) studied recreational runners during a maximal effort fitness test. Since running speed correlates with step rate, differences in self-selected speeds may have influenced preferred step rates. However, Kliethermes et al. (2021) did not find self-selected speed to be associated with bone stress injury risk and Luedke et al. (2016) observed significant associations between step rate and anterior knee pain at both fixed and self-selected speeds. While different results have been reported for different populations of runners, the current literature suggests step rate may influence the risk of specific types of injuries in competitive runners.

Since running speed is the product of step length and step rate, for a given speed, an increase in step rate implies a decrease in step length and vice versa. Therefore, an increase in step rate can be cued by decreasing step length. Musculoskeletal and probabilistic modeling have been used to examine the relationship between step length and injury risk. Edwards et al. (2009) used experimental data and musculoskeletal modeling to estimate tibia contact force and strains in runners while running at their preferred running speed with their preferred stride length and with a 10% decreased stride length, where stride length is twice step length. Peak strains were then used with a probabilistic model based on empirical data of human bone to estimate the probability of failure (stress fracture injury) for different running distances at each step length. A 10% reduction in step length resulted in decreased tibia contact force and reduced probability of stress fracture by 3-6% at running distances of 3-7mi per day. These results are consistent with prospective data in collegiate runners (Kliethermes et al., 2021). In a similar approach, Firminger et al. (2017) estimated subject-specific metatarsal strains in response to a 10% reduction in stride length (10% reduction in step length) and found no significant effect of decreasing step length on metatarsal strains or probability of failure over 40km of running. Despite an increased number of steps per unit distance, the risk of metatarsal stress fracture was not influenced by decreasing step length. Given the reduction in stress fracture injury risk of the tibia in response to decreased step length (Edwards et al., 2009), decreasing step length or increasing step rate may reduce the overall risk of lower leg bone stress injury in runners.

Step rate has been shown to influence joint and structure-specific loads. A 10% increase in step rate at constant speed reduced peak hip abduction and internal rotation, as well as peak knee extension moments (Heiderscheit et al., 2011). Lenhart et al. (2014) used musculoskeletal modeling to estimate muscle-tendon forces and patellofemoral joint force at preferred and increased step rates. A 10% increase in step rate resulted in decreased gluteus medius, vastus lateralis, soleus, and patellar tendon forces during stance, as well as decreased patellofemoral force and impulse. A 10% increase in step rate was also found to decrease patella contact pressure and contact area (Lenhart et al., 2015). Bowersock et al. (2017) found a 10% decrease in step length at constant speed resulted in decreased peak tibiofemoral joint contact force and impulse per step and per kilometer run. Reduced patellofemoral and tibiofemoral joint contact forces and impulses have also been observed with a 7.5% increase in step rate (Willy et al., 2016b; Willy et al., 2016c). Willy et al. (2019) did not observe changes in patellofemoral or tibiofemoral contact forces or impulses per kilometer with a 7.5% decreased step length in ROTC cadets while carrying a 20kg load, though increasing step length by 7.5% resulted in increased contact forces and impulses per kilometer.

In female runners, a 5% increase in step rate resulted in decreased ankle plantarflexor moment and Achilles tendon load (Lyght et al., 2016). Load-bearing ROTC cadets showed decreased Achilles tendon force per step in response to a 7.5% decrease in step length, though this did not result in decreased impulse per kilometer (Willy et al., 2019). These results are consistent with the decreased soleus force estimated by Lenhart et al. (2014) with increased step rate. Increasing step rate provides a potential intervention to decrease the loads applied to commonly injured structures during running.

The reductions in structure loads with increased step rate are likely due to changes in whole body mechanics including the vertical ground reaction force (vGRF). The vGRF has been shown to decrease in response to increasing step rate by 8-10% from preferred at constant running speed (Adams et al., 2018; Heiderscheit et al., 2011; Lenhart et al., 2014). A targeted 10% increase in step rate (mean 7.1% increase) at a constant speed resulted in decreased vGRF force approximated by force sensing insoles (Musgjerd et al., 2021).

Peak vGRF may be related to injury risk in competitive runners. In a sample of 53 NCAA cross country runners, clusters of runners with greater peak vGRF at baseline were more likely to experience bone stress injuries in the following year than other clusters (Martin et al., 2022). In a smaller sample of 9 male NCAA runners followed over 60 days, the 3 runners who sustained injuries showed greater accelerometry-based estimates of average peak vGRF per step than those who remained uninjured (Kiernan et al., 2018). A greater vertical loading rate and rearfoot strike pattern have also been implicated in injury risk, yet there is limited evidence to support such associations (van der Worp et al., 2016; Hamill & Gruber, 2017; Anderson et al., 2020; Schmida et al., 2022). There is limited evidence to support peak vGRF at baseline is related to the risk of injury across all levels of runners (van der Worp et al., 2016), however changes in peak vGRF within individuals may reflect musculoskeletal structure loading and therefore may be related to injury risk.

In level running at submaximal speeds, muscle-tendon and bone forces increase with increasing vGRF (Dorn et al., 2012; Matijevich et al., 2019). Therefore, changes in vGRF in response to increased step rate may reflect changes in structure-specific loads. Structure-specific loads can be estimated using combinations of motion analysis, musculoskeletal modeling, and medical imaging. However, these measurements currently require sophisticated equipment and are limited to laboratory settings. Recent technology has allowed vGRF measurements during running using force sensing insoles, which measure the force normal to the plantar surface of the foot such that the peak insole force approximates the peak vGRF during running (Burns et al., 2019; Renner et al., 2019; Seiberl et al., 2018). As a practical alternative to lab-based assessments, peak insole force may be used to assess the efficacy of step rate interventions to reduce risk factors of injury in settings where estimating structure-specific loads is not feasible (e.g., during outdoor running).

To infer the potential effect of interventions targeting changes in step rate or length on the risk of injury, overuse injury can be modeled as a process of mechanical fatigue in musculoskeletal structures (Edwards, 2018; Gallagher & Schall, 2017). Mechanical fatigue refers to the accumulation of microstructural damage during cyclic loading, potentially resulting in mechanical property degradation and ultimately leading to failure (Edwards, 2018).

The simplest model of cumulative damage is the Palmgren-Miner rule (Miner, 1945), where cumulative damage (CD) is defined as the ratio of the number of loading cycles at a given peak cyclic stress to the number of cycles to failure at that stress. For a varied peak cyclic stress, the cumulative damage can be summed over the different stress cases:

$$CD = \sum_i \frac{n_i}{N_i} \qquad (1)$$

Where n_i is the number of cycles at the ith stress case and N_i is the number of cycles to failure if the structure were loaded at a constant cyclic stress equal to the stress of the ith case. This model assumes structural damage increases linearly with the number of cycles at a given stress whereas cumulative damage (Eq. 1) is associated with an exponential decrease in stiffness or modulus as structures are cyclically loaded to failure (Burr et al., 1998; Firminger & Edwards, 2021; Wren et al., 2003). However, an animal model of tendon (Fung et al., 2010) suggests cumulative damage is proportional to microstructural damage, which starts to occur before the progressive loss in stiffness or modulus that is observed as failure is approached.

Musculoskeletal structures loaded to fatigue failure show a stereotypical inverse power relationship described by the equation:

$$N = AS^{-b} \qquad (2)$$

where N is the number of cycles to failure, S is the peak cyclic stress or strain, and A and b are constants. This relationship, referred to as the S-N curve, is observed in different musculoskeletal structures including human tendons (Firminger & Edwards, 2021; Schechtman & Bader, 2002; Wren et al., 2003) and bones (Carter & Caler, 1985). Substituting Eq. 2 into Eq. 1, cumulative damage is proportional to the product of the number of steps and peak stress per step weighted by an experimentally-derived factor, summed over i steps of varying peak stress:

$$CD \propto \sum_i n_i S_i^b \qquad (3)$$

This metric therefore serves as a surrogate of cumulative damage for a specific musculoskeletal structure as defined here. For a given individual, the peak cyclic stress in Eq. 3 can be replaced with the peak cyclic force.

Using vGRF as a surrogate of structure-specific forces within individuals and substituting into Eq. 3 provides a surrogate of the general cumulative damage within an individual over a run. The constant b, which describes the slope of the S-N curve, varies by tissue: 4.3 for patellar tendon (Firminger & Edwards, 2021), 6.6 for cortical bone (Edwards et al., 2009; Carter & Caler, 1985), and 9.3 for the Achilles tendon (Wren et al., 2003). Given the association between step rate and the risk of bone stress injury in college runners (Kliethermes et al., 2021), a common (Kerr et al., 2016; Kliethermes et al., 2021) and debilitating (Miller et al., 2018) injury in this population, we chose to relate our findings to the risk of bone stress injury. Bone stress injuries occur at a rate of 0.27 and 0.22 injuries per 1000 athlete exposures for women and men, respectively, accounting for 10% (women) and 11% (men) of all time-loss injuries in college distance runners (Kerr et al., 2016). Among one team, bone stress injuries affect 30-32% of collegiate runners each year (Kliethermes et al., 2021). Therefore, we weighted the peak force by a factor of 6.6 and defined the cumulative weighted peak force (CWPF):

$$CWPF = \sum_{i=1}^{n} F_i^{6.6} \qquad (4)$$

where n is the total number of steps and F_i is the peak vGRF force of the ith step for each leg. This metric infers the relative change in cumulative damage with increased step rate compared to a preferred step rate. Previous work has shown a relationship between a similar measure of cumulative weighted vGRF, estimated using accelerometry, and injury (Kiernan et al., 2018).

Examining changes in CWPF to assess the efficacy of a step rate intervention in the field may be facilitated using inexpensive equipment such as accelerometers. Based on Newton's second law and treating the body as a point mass, the vGRF can be approximated by the product of body mass and center of mass vertical acceleration. Given the sacrum approximates the center of mass (Napier et al., 2020), vertical acceleration at the sacrum can be used with body mass to provide a simplified estimate of vGRF. This method produced a significant moderate ($r = 0.64$) correlation with force-instrumented treadmill measurements of vGRF in collegiate cross country runners (Day et al., 2021) and regression models using sacral acceleration (Alcantara et al., 2021) or hip acceleration (Neugebauer et al., 2014) have yielded more accurate predictions of vGRF. Therefore, changes in sacral accelerations may reflect changes in peak vGRF force per step and CWPF per unit distance. This would provide clinicians and coaches an accessible alternative to measuring forces when evaluating whether individuals could benefit from a gait retraining intervention to increase step rate.

Increasing step rate at constant speed has been shown to decrease peak vGRF in recreational runners (Adams et al., 2018; Heiderscheit et al., 2011; Lenhart et al., 2014), but few have evaluated the effect of the intervention on metrics of cumulative damage potentially related to injury risk (Edwards et al., 2009), and no study to date has investigated these effects in college runners specifically. Given the association between step rate and injury risk in this population (Kliethermes et al., 2021), investigating changes in vGRF and CWPF with increased step rate in college runners is highly relevant. In addition, it is unknown whether individual changes in peak vGRF and CWPF per unit distance with increased step rate can be detected using accelerometry.

The purpose of this study was to 1) examine the changes in peak insole force and CWPF per kilometer with increased step rate in collegiate distance runners and 2) evaluate the use of sacral acceleration to detect changes in peak insole force and CWPF per kilometer. We hypothesized that peak insole force and CWPF per kilometer would decrease with increased step rate. We also hypothesized that changes in peak insole force and CWPF per kilometer with increased step rate would be reflected by changes in peak vertical sacral acceleration and that peak vertical sacral acceleration multiplied by body mass would correlate with peak insole force.

METHODS

Participants

We recruited collegiate distance runners who compete in events of 1500m or longer and were not injured at the time of study. Injury was defined as a reduction in training due to pain. An a priori power analysis was performed using G*Power software (Dusseldorf, Germany) to determine the sample size needed to detect differences in peak insole force in response to increased step rate with sufficient power. The expected effect size was determined from preliminary data of five college or former college runners who ran at a constant speed of 3.83m/s at their preferred step rate as well as at step rates 7.5% and 10% greater than preferred while insole force was measured. This speed was chosen to represent the majority of training runs for this population. Differences between the preferred and 10% increased step rate conditions resulted in an effect size (mean divided by standard deviation of differences) of 1.07. With this expected effect size, power of 0.80, and significance level of 0.05, a minimum of nine participants was determined necessary to detect differences in peak insole force with sufficient power. To ensure sufficient statistical power, we recruited 12 distance runners from collegiate cross country and track and field teams at UC Davis and surrounding universities. We initially targeted NCAA runners, however the recruitment of NCAA athletes proved difficult. Therefore, we recruited runners from both NCAA and club teams. This study was approved by the UC Davis Institutional Review Board.

Procedure

Participants ran 1000m at 3.83m/s ± 5% around a track at their preferred step rate and at a 10% greater step rate while insole forces were recorded using wireless force sensing insoles (Fig. 1; sampling frequency 200Hz, range 2.5-5000N, resolution ±10N; Novel Loadsol, St. Paul, MN).

Each insole consists of a single capacitive sensor that measures the normal force under the plantar surface of the foot. Data were transmitted from the electronics housing, which was attached to the laces of the participants' shoes, via Bluetooth to an Android device. These sensors have been previously validated to accurately and reliably measure vGRF while running (Burns et al., 2019; Renner et al., 2019; Seiberl et al., 2018). The speed was chosen to represent the speed used during the majority of training runs for the NCAA distance runners at UC Davis.

Figure 1: The Loadsol force sensing insoles used in the study and insole force data recorded via the Loadsol app on an Android device. Insole sizes are marked with manufacturer labeled sizes and corresponding US women's (W) and men's (M) shoe sizes.

Participants reported their age, sex, height, years of running experience, and 5000m personal bests. Body mass was measured using a portable electronic scale. Three insole sizes were used in this study (Fig. 1). Participants were fit to the smallest insole size that covered the entire plantar area of the foot. The insoles were inserted into the shoes over the insoles of the participants' own shoes. The insoles were calibrated according to manufacturer instructions. The calibration for each sensor was verified with static single leg stance force within 5% of the participant's body weight.

An IMU (range ± 16g, resolution 62ug at ± 2g, non-linearity 0.5%, cross-axis ± 2%; ProMove Mini, Inertia Technology, Enschede, The Netherlands) was mounted on the sacrum using a neoprene belt secured over the waistband of the shorts. The vertical axis of the IMU coordinate system was aligned with the vertical axis of the trunk. The acceleration along this axis was taken as the vertical sacral acceleration and recorded at 200Hz. The belt was tightened as far as comfortable. To examine changes in heart rate with increased step rate, participants also wore a Polar chest strap heart rate monitor (H10, Kempele, Finland) that updated heart rate data at 1Hz via a Garmin device (Forerunner 230, Olathe, KS).

Participants first ran for 400m at their preferred step rate next to a pacer running at 3.83 m/s. Insole forces were recorded at 200Hz with Loadsol software via Bluetooth connection with an Android device. The preferred step rate was taken as the average step rate determined from the insole data during the final 30 seconds of this trial.

For the next two trials, participants matched their step rate to an audible metronome set at either the preferred step rate or a step rate 10% greater than preferred, in random counterbalanced order. To synchronize the force and acceleration data, participants performed a vertical countermovement jump before and after each trial. Participants were instructed to run next to the pacer at 3.83m/s. After acclimating to the new step rate for 200m, data were recorded for 1000m. Trials were accepted if the average speed, calculated from 1000m time, was within 5% of 3.83m/s and the average speeds of the two trials were within 2.5% of each other. Participants rested for a minimum of three minutes between trials. After each trial, participants reported their rating of perceived exertion (RPE) using a modified Borg CR10 scale (Foster et al., 2001; Fig. 2).

Rating of Perceived Exertion

Rating	Descriptor
1	Very, very easy
2	Easy
3	Moderate
4	Somewhat hard
5	Hard
6	
7	Very hard
8	
9	
10	Maximal

Figure 2: The RPE scale used in the current study.

Data processing

Raw force data were imported into MATLAB (R2021b, Natwick, MA) for data processing and analysis using custom MATLAB scripts. Raw force data were left unfiltered (Renner et al., 2019). For examples of raw force data, see Appendix. The raw vertical sacral acceleration data were low-pass filtered using a zero-lag 4th order Butterworth filter with a 5Hz cutoff frequency. A power spectral analysis of the acceleration signals revealed the majority of the signal was contained at 3Hz frequency (Fig. 3). The 5Hz cutoff frequency was chosen to filter the impacts represented by the larger frequency spikes. This cutoff frequency was previously found to result in the lowest absolute error between predicted peak vGRF using peak sacral acceleration times body mass and the observed peak vGRF in collegiate cross country runners (Day et al., 2021). With our data, we found that a 5Hz cutoff frequency resulted in stronger correlations between peak sacral acceleration and peak insole force than a 10Hz cutoff frequency that retained the impact acceleration peaks. The gravitational component of the vertical sacral acceleration was calculated as the mean static acceleration and subtracted from the acceleration signal. See Appendix for examples of raw and processed acceleration signals. Heart rate data were imported from the Garmin device as a .tcx file and converted to ASCII format for analysis in MATLAB.

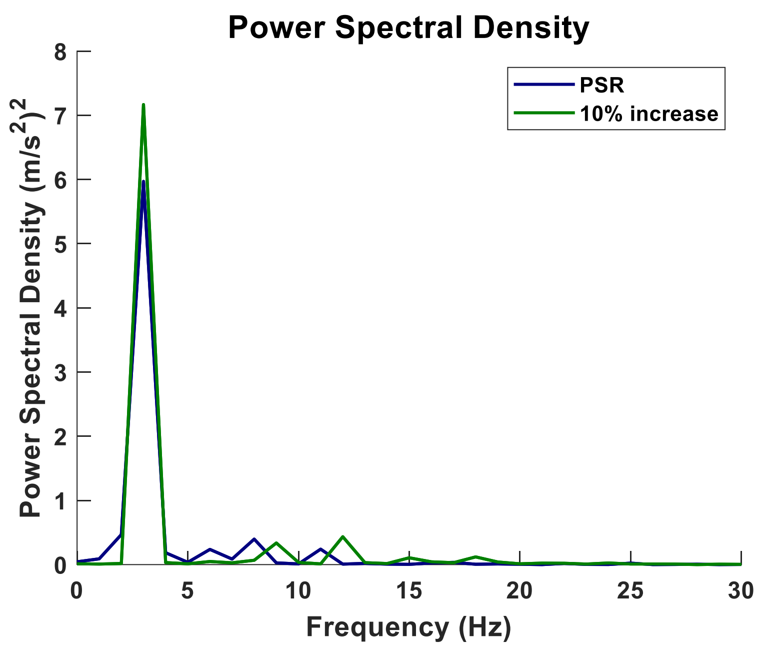

Figure 3: Power spectral densities of the acceleration signals for preferred (PSR) and 10% increased step rate conditions.

Force and acceleration data were synchronized (Fig. 4) via cross-correlation of the signals during the countermovement jumps. The total force and vertical sacral acceleration signals were cross-correlated at each synchronization event and interpolated between them to determine the time shift between the two signals for each sample. Synchronized force and acceleration signals at the synchronization events are shown in the Appendix.

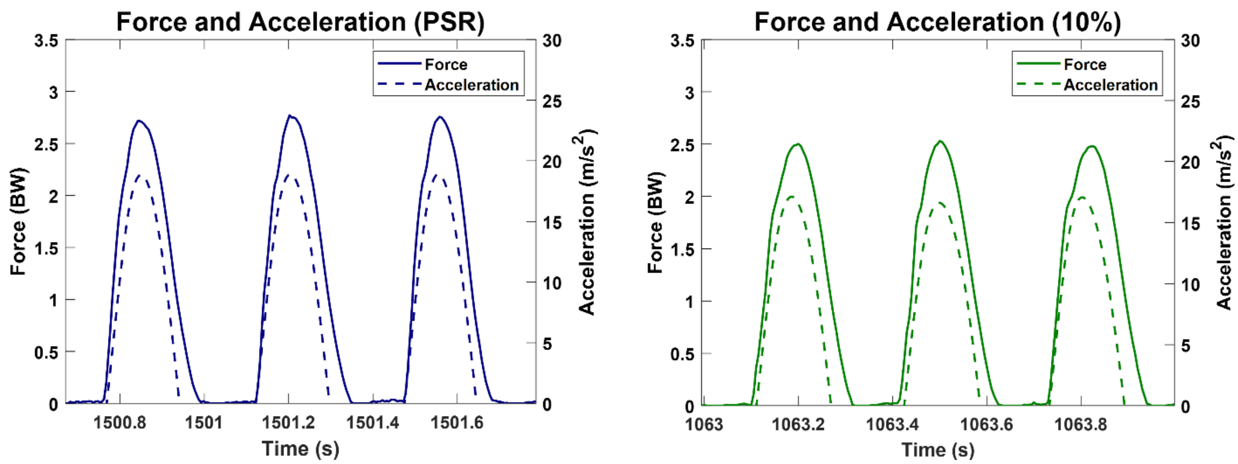

Figure 4: Synchronized force and acceleration data for preferred (PSR) and 10% increased step rate conditions for a representative subject.

Analysis

Peak insole force per step was averaged over each 1000m trial. Impulse per step was calculated as the integral of the force-time curve over stance, where stance start and end points were defined as the points at which the slope of force versus time exceeded 2500 N/s and fell below -2500 N/s, respectively (Fig. 5). This method was based on that of Seiberl et al. (2018), who used a ±1500 N/s threshold to identify stance start and end points with insole force data recorded at 100Hz in participants running at 2.2-3.3 m/s. For the greater sampling frequency and running speed in this study, we used a threshold of ±2500 N/s to identify stance start and end points. Contact time in seconds was calculated from the stance start and end times via the Loadsol sensors.. CWPF per kilometer was calculated for each 1000m trial according to Eq. 4 using the peak insole force. Values were averaged across right and left legs for each trial. Contact time was also calculated as a percent of time in ground contact. Percent of time in ground contact was calculated as the sum of the left and right ratios of contact time to stride time, which was determined using the stance start points of consecutive ipsilateral steps. Peak vertical sacral acceleration was averaged over steps for each trial. Cumulative weighted acceleration per kilometer was calculated for each 1000m trial according to Eq. 4, replacing peak force with peak acceleration. The heart rate for each step rate condition was averaged over the final 60 seconds of each trial.

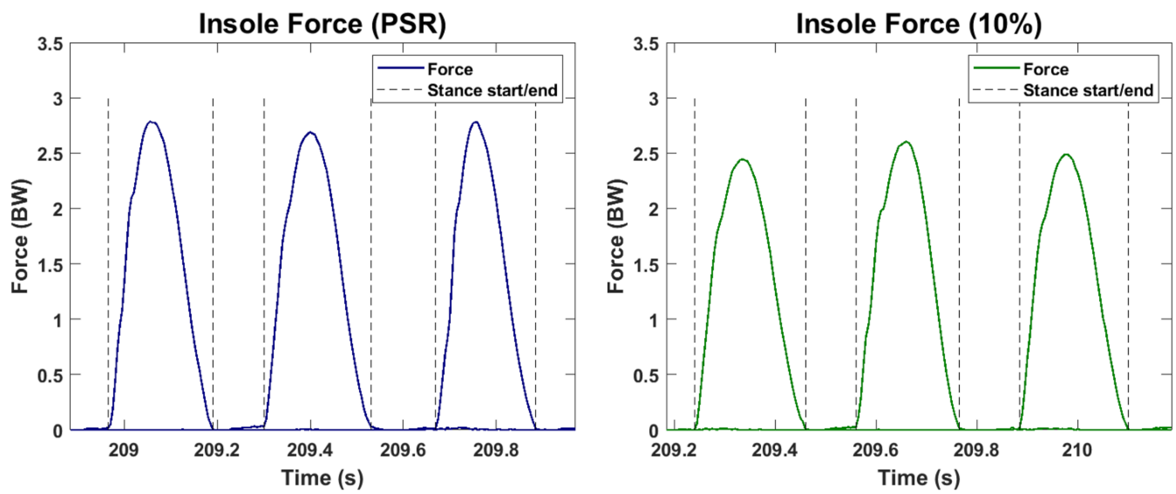

Figure 5: Identification of stance phase start and end points for preferred (PSR) and 10% increased step rate conditions for a representative subject.

Results are reported as mean ± standard deviation. Data were tested for normality using Shapiro-Wilk tests with a significance level of 0.05. For normally distributed data, the effect of step rate on the average peak insole force, CWPF per kilometer, average contact time, average impulse, average peak acceleration, cumulative weighted acceleration per kilometer, heart rate, and RPE was tested for significance using paired t-tests with a significance level of 0.05. For variables that violated the assumption of normality, the effect of step rate on those variables was tested for significance using Wilcoxon signed rank tests with significance level of 0.05. Acceleration data were analyzed for 11 participants due to lost signal for one participant. Heart rate data were analyzed for 11 participants due to a lost connection between devices.

The peak accelerations corresponding to the peak forces per step were extracted by identifying the peak accelerations nearest in time to the peak forces. Refer to the Appendix for temporal identification of peak forces and accelerations. Peak acceleration multiplied by body mass and peak force per step were pooled across step rate and a line was fit to right, left, and bilateral peak force vs. peak acceleration multiplied by body mass. Pearson correlation coefficients, r, were calculated for each participant and tested for significance. The average correlation coefficient was then calculated across participants. A strong correlation was defined as $r \geq 0.80$, a moderate correlation was defined as $0.50 \leq r < 0.80$, and a weak correlation was defined as $r < 0.50$.

RESULTS

Twelve college (club and NCAA) runners participated in the study (Table 1). Participants increased step rate by an average of 9.7 ± 0.9 % (mean ± SD) relative to the preferred step rate. Running speed showed a slight (< 1%) significant increase with increased step rate (Table 2). Average peak insole force and CWPF significantly ($p < 0.001$) decreased with increased step rate (Fig. 6). Average peak insole force decreased by 0.156 ± 0.068 BW, which represents a 5.94 ± 2.66 % decrease in peak force with increased step rate relative to preferred step rate. The decrease in average peak insole force translated to a 26.1 ± 12.6 % decrease in CWPF.

Table 1: Mean ± SD participant characteristics

	Female (n = 6)	**Male (n = 6)**	**Overall (n = 12)**
Age (y)	23.7 ± 2.9	23.0 ± 3.5	23.3 ± 3.1
Height (cm)	165.7 ± 6.8	179.5 ± 5.7	172.6 ± 9.4
Mass (kg)	58.8 ± 6.2	74.3 ± 10.1	66.5 ± 11.4
Running experience (yrs)	9.3 ± 3.7	10.3 ± 4.5	9.8 ± 4.0
5000m best (min)	19.4 ± 1.7	15.1 ± 0.8	17.2 ± 2.6

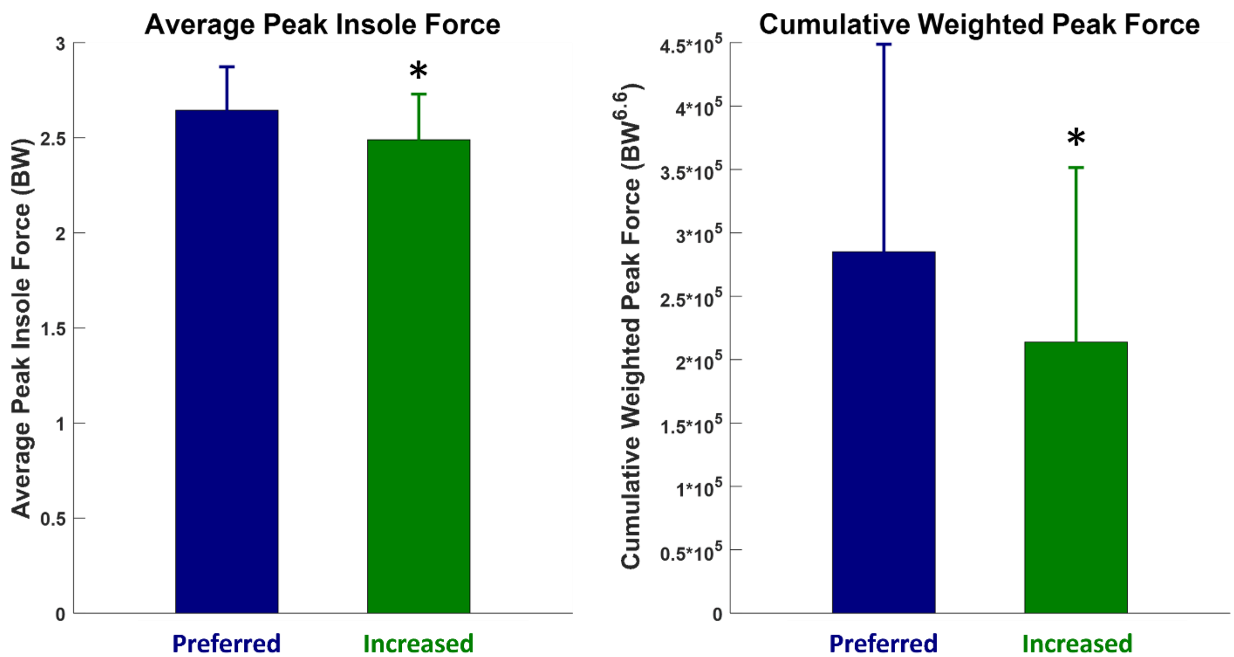

Figure 6: Average peak insole force and CWPF for preferred and 10% increased step rates. Error bars represent ± standard deviation. *Significant difference from preferred ($p < 0.001$).

Impulse per step also decreased with increased step rate (Table 2). Contact time per step decreased with increased step rate, however, percent of time in ground contact increased with increased step rate (Table 2). Consistent with the decreases in peak force and CWPF, peak acceleration and cumulative weighted peak acceleration significantly decreased with increased step rate (Table 2). RPE and heart rate both significantly increased with increased step rate (Table 2).

Table 2: Mean ± SD of variables for each step rate condition and changes between conditions

	Preferred	**Increased**	**Δ**	**p value**
Step rate (steps/min)	176 ± 8	193 ± 8	17.1 ± 1.4	< 0.001
Speed (m/s)	3.79 ± 0.07	3.83 ± 0.06	0.04 ± 0.04	0.010
Average peak insole force (BW)	2.64 ± 0.23	2.49 ± 0.24	-0.16 ± 0.07	< 0.001
CWPF ($BW^{6.6}$)	2.85×10^5 ± 1.64×10^5	2.14×10^5 ± 1.38×10^5	-7.14×10^4 ± 5.51×10^4	< 0.001
Impulse (BW*s)	0.352 ± 0.026	0.318 ± 0.023	-0.0343 ± 0.0075	< 0.001
Contact time (s)	0.228 ± 0.014	0.217 ± 0.014	-0.011 ± 0.004	< 0.001
Percent of time in ground contact	66.8 ± 2.9	69.7 ± 3.1	2.87 ± 0.94	< 0.001
Peak acceleration (m/s^2)	15.2 ± 1.8	14.3 ± 1.8	-0.889 ± 0.702	0.002
Cumulative weighted peak acceleration $(m/s^2)^{6.6}$	6.40×10^{10} ± 5.34×10^{10}	4.75×10^{10} ± 3.59×10^{10}	-1.64×10^{10} ± 2.34×10^{10}	0.019
RPE	3.3 ± 1.4	4.6 ± 1.6	1.3 ± 1.5	0.016
Heart rate (bpm)	161 ± 14	164 ± 12	3.0 ± 3.4	0.015

All individual correlations between peak force per step and peak acceleration per step multiplied by body mass were statistically significant ($p < 0.05$). Correlation coefficients varied by participant (Fig. 7). Bilateral correlations (pooled left and right data) were moderate or strong for 9 of 11 participants. The average of bilateral correlation coefficients across participants was moderate ($r = 0.62 \pm 0.37$, mean ± SD). The average correlation coefficient across participants improved when the data were separated into left ($r = 0.69 \pm 0.25$) and right ($r = 0.78 \pm 0.16$) sides. However, separating into right and left sides did not improve the strength of the correlations for all individuals.

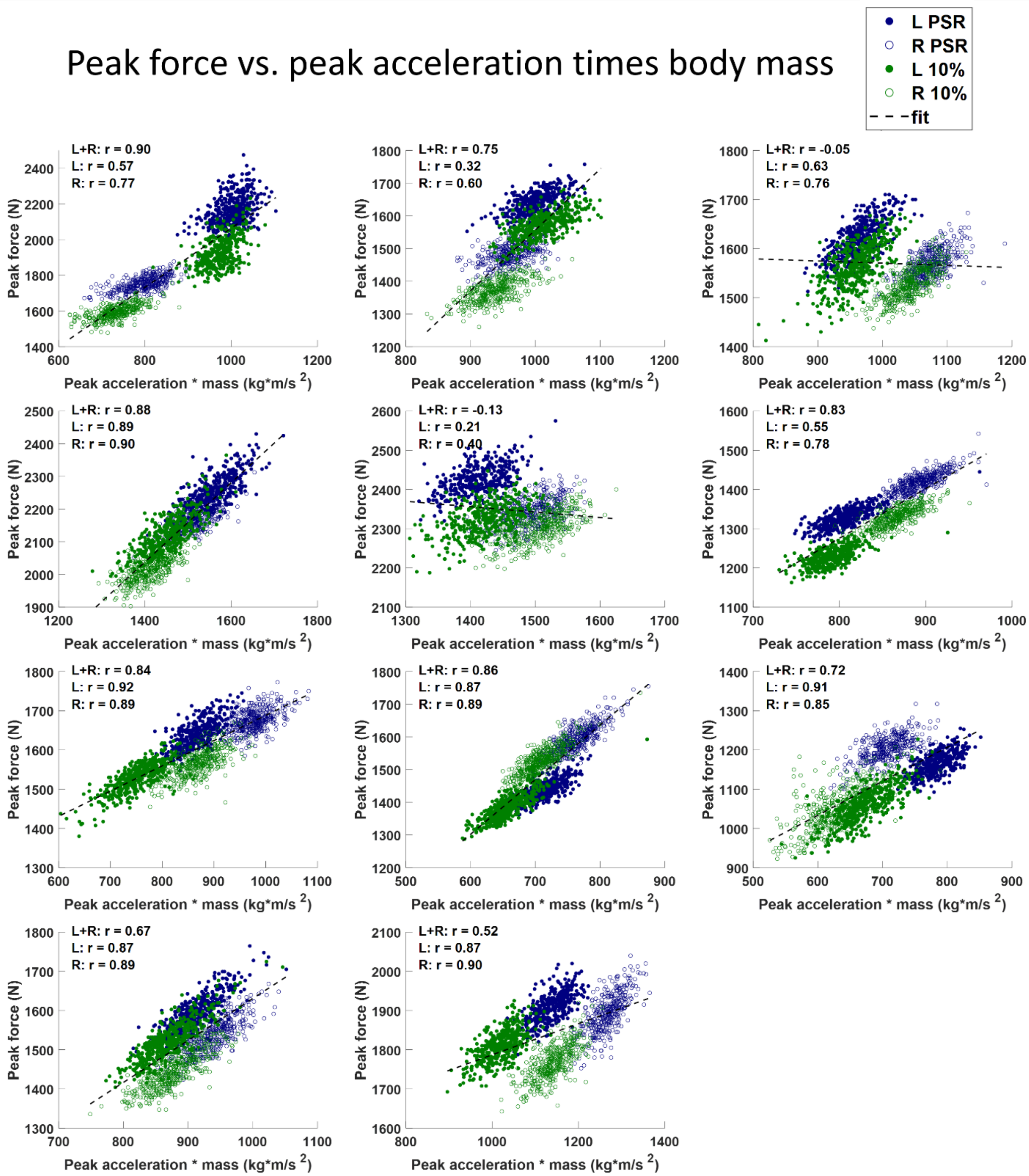

Figure 7: Individual participant correlations between peak force per step and peak acceleration per step multiplied by body mass. Left (L) and right (R) steps are plotted for preferred step rate (PSR) and 10% increased step rate conditions. The dashed line represents the line fit to the bilateral (L+R) data.

DISCUSSION

Our primary objective was to examine changes in peak insole force and CWPF per kilometer with increased step rate in college runners. We hypothesized that average peak insole force and CWPF per kilometer would decrease with increased step rate. In support of our hypothesis, average peak insole force and CWPF per kilometer significantly decreased while running at a 10% increased step rate relative to preferred, indicating that increasing step rate results in decreased peak vGRF and cumulative weighted peak vGRF per kilometer. The decrease in peak insole force with increased step rate is consistent with results reported for recreational runners (Musgjerd et al., 2021) and decreases in vGRF in recreational runners (Adams et al., 2018; Heiderscheit et al., 2011; Lenhart et al., 2014). A decrease in CWPF per kilometer suggests a decrease in cumulative damage in bone structures for which loads are reduced, such as the tibia (Edwards et al., 2009). Since other structures exhibit similar fatigue failure behavior (Firminger & Edwards, 2021; Schechtman & Bader, 2002; Wren et al., 2003), it may also indicate a decrease in cumulative damage of other musculoskeletal structures for which structure-specific loads are reduced and a relative decrease in injury risk in those structures (Edwards, 2018). Therefore, the observed decreases in peak insole force and CWPF provide ecologically valid evidence that a 10% increase in step rate may decrease the peak vGRF and a general measure of cumulative damage, reflecting decreases in bone and other structure-specific cumulative damage and relative decreases in structure-specific injury risk in college runners.

The peak and cumulative weighted force measurements of this study were limited to insole force used to approximate the vGRF. Therefore, we cannot infer relative changes in the potential risk of injury for specific musculoskeletal structures. However, changes in vGRF reflect changes in structure-specific forces (Dorn et al., 2012; Matijevich et al., 2019). Other studies have shown decreased hip (Boyer & Derrick, 2018; Heiderscheit et al., 2011), knee (Bowersock et al., 2017; Boyer & Derrick, 2018; Heiderscheit et al., 2011; Lenhart et al., 2014; Willson et al., 2014; Willy et al., 2016b; Willy et al., 2016c), ankle (Boyer & Derrick, 2018), tibia (Edwards et al., 2009), soleus (Lenhart et al., 2014), and Achilles tendon loads (Lyght et al., 2016; Willy et al., 2019) with increased step rate. Therefore, the decrease in CWPF may indicate a decrease in cumulative damage for those structures. However, increasing step rate may increase hip flexor muscle forces during early swing and hamstring muscle forces during late swing (Lenhart et al., 2014) and therefore may not be appropriate for reducing the risk of all injuries. Clinicians should consider the injury history of patients when considering an intervention to increase step rate. For some of the most common running injuries including Achilles tendinopathy, patellar tendinopathy, and tibial stress fracture (Lopes et al., 2012), increasing step rate may be an appropriate intervention. The CWPF metric in this study (Eq. 4) represents a surrogate of cumulative damage in bone (e.g., the tibia). When peak force was weighted by factors reflecting the fatigue failure behavior of Achilles tendon (9.3; Wren et al., 2003) or patellar tendon (4.3; Firminger & Edwards, 2021), CWPF per kilometer decreased by 36.2% ($p < 0.001$) and 16.8% ($p < 0.001$) on average, respectively, which is consistent with the 26.1% decrease specific to bone. Therefore, clinicians should consider interventions to increase step rate in runners susceptible to such injuries (e.g., those with a frequent injury history or those returning from injury) and implement interventions gradually to allow adaptation to potential shifts in load to different musculoskeletal structures.

CWPF is based on the Palmgren-Miner model of cumulative damage (Eq. 1; Miner, 1945), which does not account for the loading history of musculoskeletal structures. Because damage to a cyclically loaded structure results in decreased modulus (Burr et al., 1998;

Firminger & Edwards, 2021; Wren et al., 2003), subsequent loading bouts will decrease the number of cycles to failure at a given stress and lead to greater damage with increasing number of loading cycles than if the structure were undamaged to begin with. Therefore, differences in CWPF limit inferences about cumulative damage to single loading bouts. However, the purpose of this study was to compare relative changes in cumulative damage between step rate conditions per unit distance of running (i.e., for a single loading bout). Given a decrease in peak vGRF with increased step rate, corresponding decreases in structure-specific loads would result in reduced cumulative damage for a given training session. If the rate of damage formation exceeds the rate of repair and adaptation (increased number of cycles to failure), then these differences would only be amplified over multiple training sessions. However, more detailed and structure-specific models of damage repair and adaptation would need to be incorporated to infer differences in cumulative damage over multiple training sessions.

Slight changes in running speed and individual differences in responses may have contributed to the large variability in CWPF changes with increased step rate (SD = 12.6%). While speed was controlled to within 2.5% between trials, there was a small (< 1%) and significant increase in speed with increased step rate. Despite a slightly faster speed on average with increased step rate, all participants showed reduced peak insole force and CWPF.

Impulse per step also significantly decreased with increased step rate, likely due in part to decreased peak force and contact time. A decrease in vertical impulse suggests a lower vertical take-off velocity and thus a lower center of mass vertical excursion (Heiderscheit et al., 2011). These changes contribute to the decrease in stride time that is required to increase stride frequency (one-half step frequency). While contact time per step significantly decreased, the percent of time in ground contact significantly increased with increased step rate. On average, percent of time in ground contact was 2.87% longer when increasing step rate, which is consistent with Heiderscheit et al. (2011). This result indicates participants spent a slightly longer relative time in contact with the ground while running with an increased step rate. Interestingly, a shorter percent of time in ground contact was consistent with greater peak vGRF and associated with a higher risk of bone stress injury in collegiate runners (Martin et al., 2022). However, the difference in percent of time in ground contact observed here is smaller than the differences between clusters of different injury incidences (Martin et al., 2022).

This study also evaluated the use of sacral acceleration to detect changes in peak insole force and CWPF. We hypothesized that peak vertical sacral acceleration and cumulative weighted peak acceleration would decrease with increased step rate and that peak vertical sacral acceleration multiplied by body mass would correlate with peak insole force. Consistent with decreased peak force and CWPF, peak vertical sacral acceleration and cumulative weighted peak acceleration significantly decreased with increased step rate. Average peak vertical sacral acceleration decreased for all but one participant and cumulative weighted peak acceleration decreased for 9 of 11 participants. Individual correlations between peak insole force and peak acceleration were on average moderate ($r = 0.62$) with 9 of 11 participants showing significant and moderate or strong correlations using bilateral data. In one of the participants who showed a weak negative correlation ($r = -0.05$), the correlation coefficients were moderate when specific to the left ($r = 0.63$) or right ($r = 0.76$) sides. Together, these results suggest that changes in sacral acceleration reflect changes in peak insole force and that sacral acceleration can be used to detect changes in peak insole force and CWPF with increased step rate, supporting our hypotheses. This is consistent with previous studies showing vGRF can be estimated using sacral acceleration multiplied by body mass (Day et al., 2021) or acceleration-based prediction models in collegiate runners (Alcantara et al., 2021; Neugebauer et al., 2014).

For most participants, the strength of the correlations improved when the data were separated into left and right sides. Changes in insole sensitivity and/or insole movement relative to the foot may have contributed to the discrepancy. Additionally, while the IMU was secured over the sacrum, small movements of the IMU relative to the sacrum may have influenced our measurements of vertical sacral acceleration, which would translate to larger errors in predictions of peak force. The local coordinate system axis of the accelerometer aligned with the vertical axis of trunk does not perfectly represent vertical acceleration in the inertial reference frame. Estimates of peak force may be improved using additional sensor data from the IMU (gyroscope and magnetometer) to transform the vertical IMU axis to the vertical axis of the inertial reference frame or with additional variables to predict peak force (Alcantara et al., 2021; Neugebauer et al., 2014). Using a single controlled speed limited the individual variability in peak insole force, which may have limited the strength of the correlations. However, nearly all participants showed moderate or strong correlations, supporting the use of sacral acceleration to detect changes in vGRF with increased step rate.

RPE and heart rate significantly increased with increased step rate. The increase in RPE could be a result of increased mental focus to increase step rate. However, the concurrent increase in heart rate suggests that metabolic energy consumption may also increase in response to increasing step rate. The increase in heart rate contrasts with previous studies showing no significant increase in heart rate and metabolic cost with 10% increase in step rate (Hamill et al., 1995) and no significant increase in metabolic cost with 8% increased step rate (Swinnen et al., 2021). These differences may be due to differences in competition level of the samples. Hamill et al. (1995) studied healthy male adults who showed a U-shaped trend for heart rate and oxygen consumption with a minimum occurring at a step rate greater than preferred. The college runners in this study may have preferred step rates that are already at or near optimum for metabolic cost such that increasing step rate may increase the metabolic energy consumption at a given speed. In runners of similar fitness, Swinnen et al. (2021) found most runners minimized metabolic energy consumption at their preferred step rate.

Given the increase in RPE and potential increase in metabolic cost with increased step rate, increasing step rate should be prescribed only for “easy” runs in which the performance for that run is not the primary goal. Furthermore, the results of this study are specific to a speed of 3.8 m/s, which represents easy run speeds and the majority of runs in the population of NCAA runners. It is unknown whether increasing step rate at faster speeds would have the same effects on biomechanical and metabolic parameters. Practically, increasing step rate beyond the preferred step rate becomes more difficult with increasing speed. Therefore, increasing step rate for easy runs may be a viable intervention to reduce peak vGRF and CWPF for the majority of training runs. Ultimately, a decrease in the risk of injury would reduce the training time lost to injury and therefore positively affect performance (Raysmith & Drew, 2016).

Due to the limited sample available and limited variety of insole sizes, we did not control for habitual footstrike pattern among participants, which could have contributed to variability in individual responses. In addition, we did not control for footstrike pattern across the step rate conditions. Increasing step rate may result in decreased foot inclination angle at initial contact (Heiderscheit et al., 2011). However, footstrike pattern does not appear to confound the effects of step rate on biomechanical parameters. Increased step rate has been shown to reduce tibiofemoral joint and Achilles tendon loads independent of footstrike pattern (Bowersock et al., 2016; Lyght et al., 2016).

The results of this study are specific to the acute effects of increasing step rate in college runners; the longitudinal effects of increasing step rate in this population are unknown. Step rate can be increased through a gait retraining intervention over eight runs and maintained for at least one month later (Willy et al., 2016a; Willy et al., 2016b). With this protocol, changes in biomechanical parameters remained at the one month follow-up (Willy et al., 2016a; Willy et al., 2016b). Training to increase step rate can have longitudinal effects on metabolic cost. For example, well-trained female runners who completed a 10 day training program to increase their step rate to 180 steps/min showed reduced oxygen consumption at 3.4 and 3.8 m/s (Quinn et al., 2021). The varied responses in different populations suggests the effects of increasing step rate should be evaluated on an individual basis.

CONCLUSIONS

This study examined changes in peak insole force and CWPF per kilometer with increased step rate in collegiate distance runners and evaluated the use of sacral acceleration to detect changes in peak insole force and CWPF per kilometer. Decreases in average peak insole force and CWPF per kilometer suggest increasing step rate results in decreased peak vGRF and cumulative weighted peak vGRF per kilometer. The CWPF represents a potential surrogate of the general damage accumulated in musculoskeletal structures for a given training session. Since decreases in cumulative damage for specific structures may reflect relative changes in structure-specific injury risk, the reduced CWPF with increased step rate warrants further study to evaluate the effect of increasing step rate on injury risk in college runners. These results are consistent with previous findings in other populations of runners and provide ecologically valid evidence that increasing step rate is a viable intervention to reduce the vGRF and CWPF in college runners. Therefore, clinicians should consider interventions to increase step rate in runners susceptible to injury (e.g., those with a history of frequent injuries or those returning from injury). Increases in RPE and heart rate suggest a greater mental focus and/or metabolic cost with increased step rate in this population, therefore interventions to increase step rate should be constrained to easy running. Future work should investigate the longitudinal effects of a training protocol to increase step rate on vGRF and structure-specific loads in college runners. Prospective studies should be conducted to determine whether interventions targeting an increase in step rate reduce the risk of running injuries.

Decreases in peak vertical sacral acceleration and cumulative weighted peak acceleration were consistent with decreases in peak insole force and CWPF. These results, along with mostly moderate to strong correlations between peak insole force and peak vertical sacral acceleration multiplied by body mass, suggest that sacral acceleration may be used to detect changes in peak insole force and CWPF per kilometer. These results demonstrate that the previously investigated approach using sacral acceleration to estimate vGRF can be applied to evaluate potential relative changes in structure loads and cumulative damage in response to increasing step rate. This approach may be applied to other kinematic, training, or footwear interventions to evaluate potential changes in structure loads and cumulative damage in response to those interventions using accessible, inexpensive equipment in the field.

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APPENDIX

Shown below are examples of raw insole force data (Figs. A1-A2), insole force curves over a stride cycle (Fig. A3), raw and filtered acceleration signals (Fig. A4-A5), synchronized force and acceleration signals at the synchronization events (Fig. A6-A7), and synchronized force and acceleration data with temporal locations of peaks identified (Fig. A8).

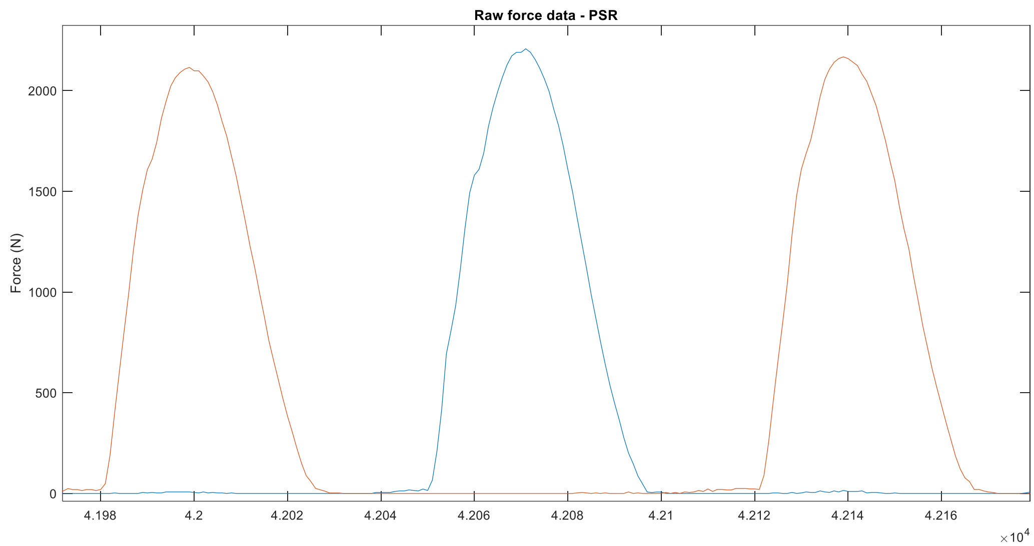

Figure A1: Raw insole force data for a representative subject for the preferred step rate condition (PSR). The horizontal axes are samples (200Hz) and the vertical axes are force in Newtons. Right steps are represented by the orange trace and left steps by the blue trace.

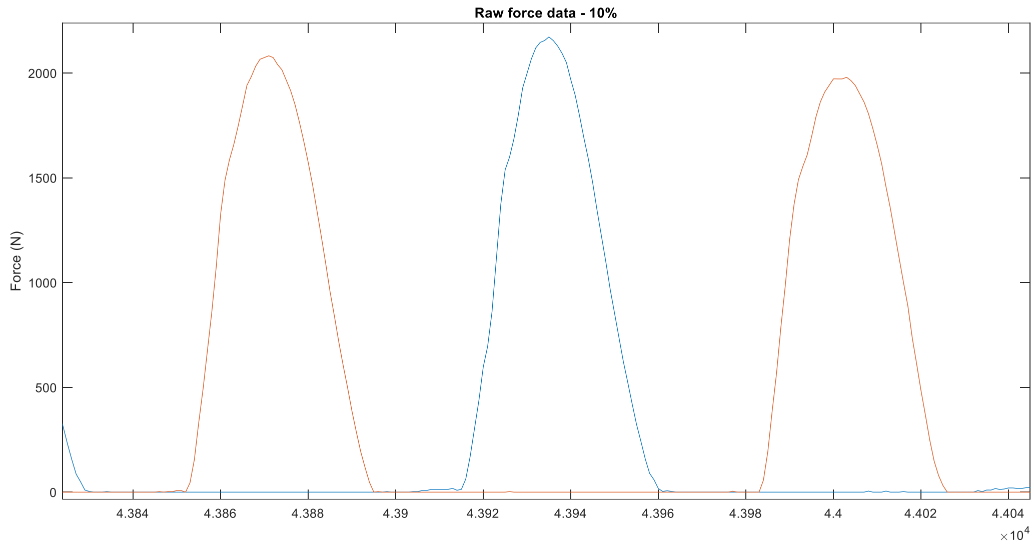

Figure A2: Raw insole force data for a representative subject for the 10% increased step rate condition. The horizontal axes are samples (200Hz) and the vertical axes are force in Newtons. Right steps are represented by the orange trace and left steps by the blue trace.

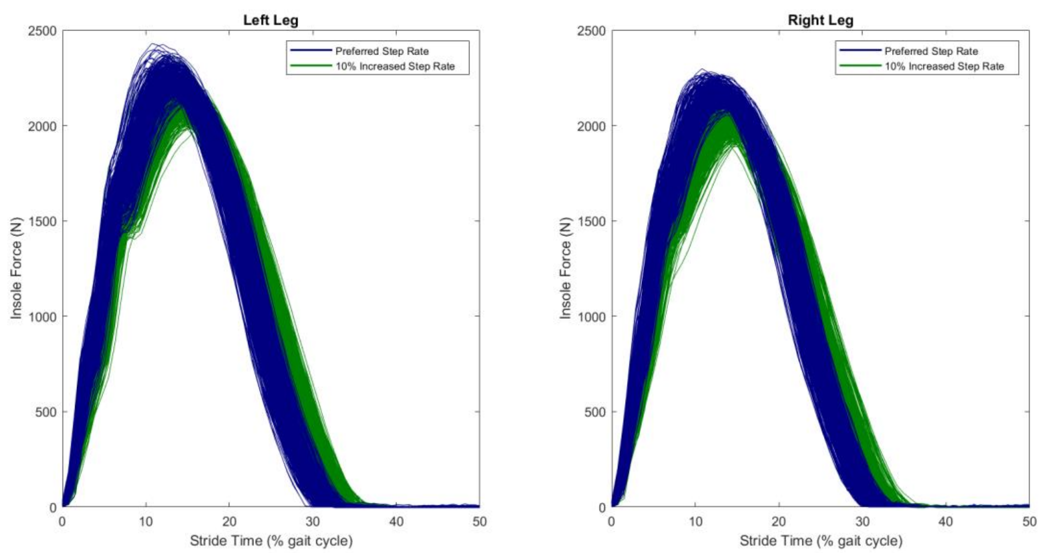

Figure A3: Left and right insole force curves for a representative subject over the stride (gait) cycle for both step rate conditions. Force curves from 0-50% of total stride time are displayed.

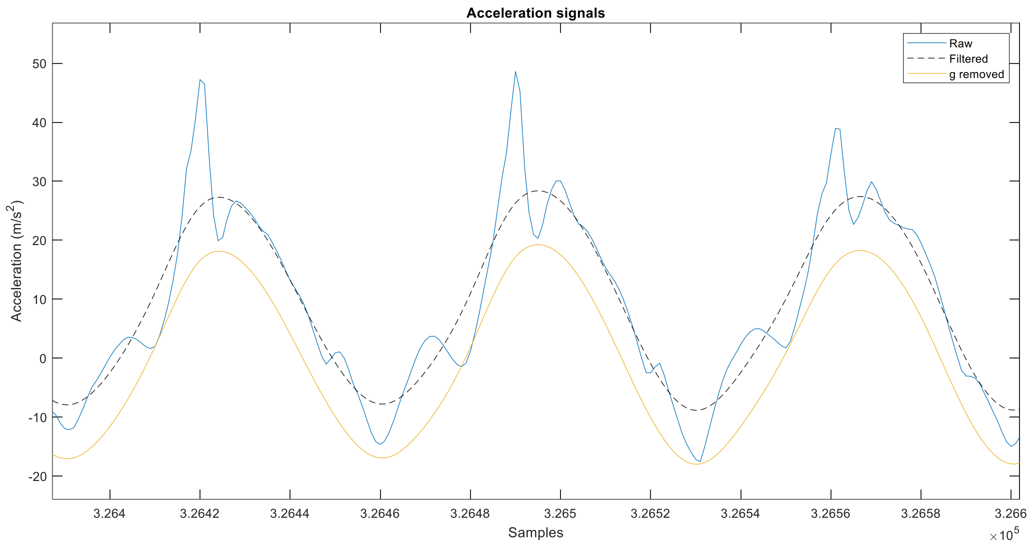

Figure A4: Raw acceleration, filtered acceleration, and filtered acceleration signals with gravity removed for the preferred step rate condition for a representative subject.

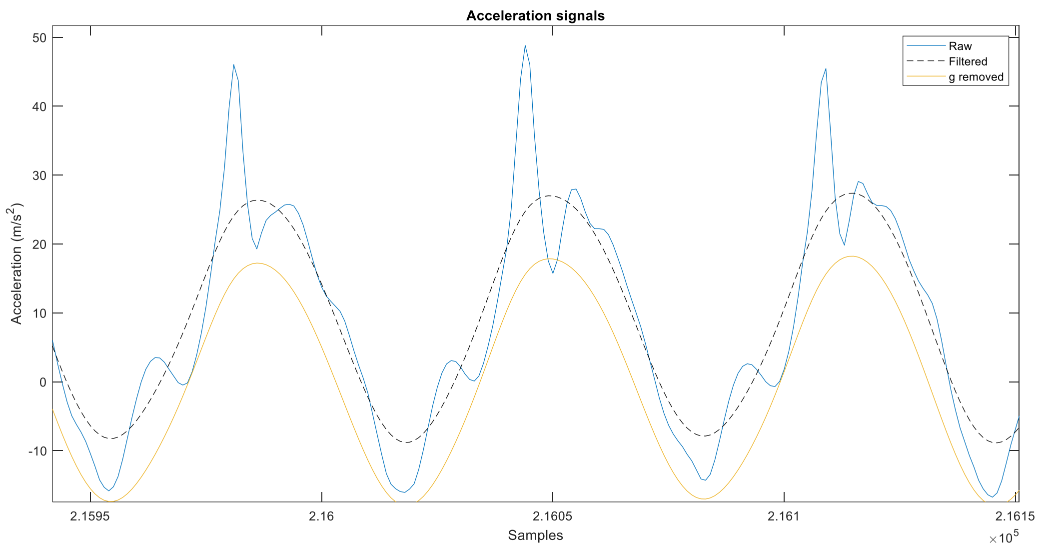

Figure A5: Raw acceleration, filtered acceleration, and filtered acceleration signals with gravity removed for 10% increased step rate condition for a representative subject.

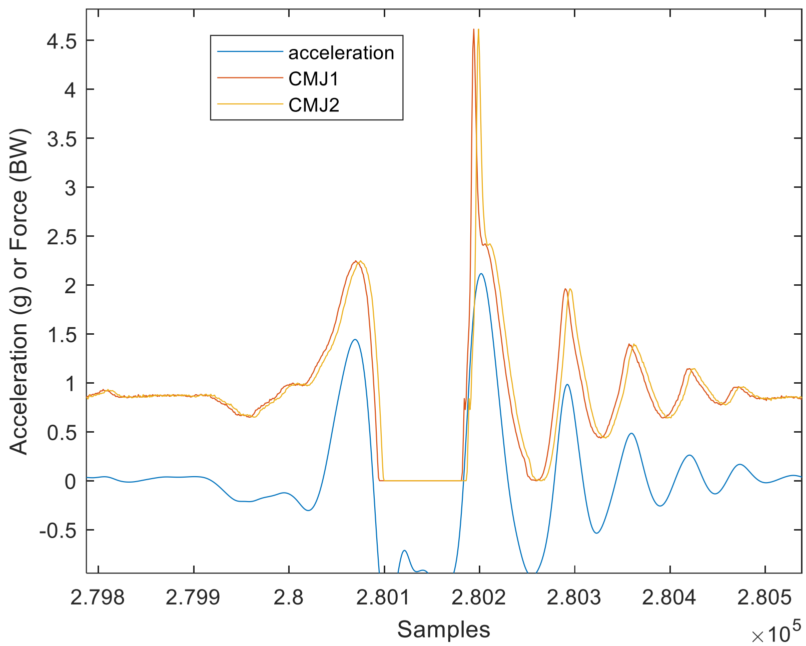

Figure A6: Representative example of synchronized force and acceleration signals at the second countermovement jump synchronization event (CMJ 2 force synchronized with acceleration).

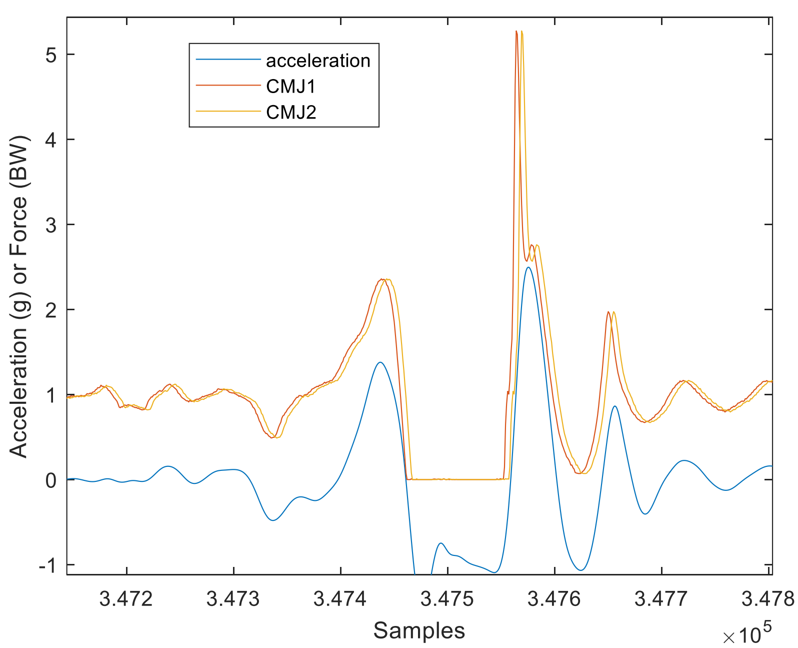

Figure A7: Representative example of synchronized force and acceleration signals at the first countermovement jump synchronization event (CMJ 1 force synchronized with acceleration).

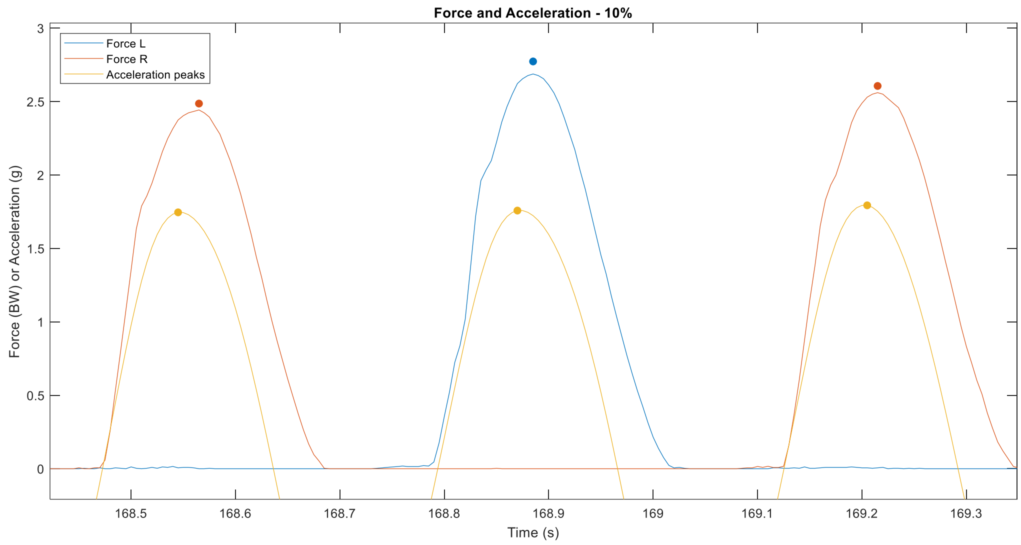

Figure A8: Synchronized force and acceleration data with temporal locations of peaks identified for a representative subject.