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Effect of running cadence on tibial acceleration: implications for runners experiencing stress injuries

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Table of Contents

Abstract

I. Purpose

The purpose of this study was to determine the influence of cadence modification on tibial acceleration (TA) in the interest of exploring potential methods to reduce the incidence of stress-related running injuries.

II. Methods

Eleven injury-free distance runners were recruited to run at 6 different cadences with acoustic pacing. An inertial measurement unit (IMU) containing an onboard triaxial accelerometer was mounted on the right medial distal tibia to record acceleration at each of these paces. A repeated-measures ANOVA was conducted to determine differences in tibial acceleration over varying cadences. Effect sizes (η^2 and Cohen's *d*) were examined to determine meaningfulness of the results. Secondary analyses were conducted to determine potential covarying effects.

III. Results

The repeated measures ANOVA indicated that cadence significantly affected peak axial tibial acceleration ($F=7.59$, $p<0.001$, $\eta^2=0.43$). Post-hoc tests showed that peak axial TA was significantly lower at the fastest cadence compared to the slowest cadence ($p<0.001$, Cohen's $d=0.813$, mean difference=1.085 g). At lower cadences, females have exceptionally higher TA ($p=0.01$), and the difference in TA among females from their slowest to fastest cadence was significant ($p<0.001$). The difference in TA among males from slowest to fastest cadence did not change significantly ($p=0.78$).

IV. Conclusion

The hypothesis was fully supported: TA decreases as cadence increases. Upon further examination, these differences were driven primarily by females. A decrease in TA is indicative of decreased external force being applied to the tibia. Results of this study suggest that increasing

cadence while maintaining the same speed could result in potentially lowered risk of injury, and is especially important for females.

Introduction

Distance running recreationally and competitively is a common method of cardiovascular exercise with many health benefits; however, running-related injuries are a significant risk, with as many as 79% of distance runners experiencing an injury due to overtraining that necessitated changes in training (Lun et al., 2004). Stress fractures, also called fatigue fractures and hairline fractures, are among these. Over time, tiny cracks in the bone form as a result of repeated cumulative loading beyond the rate at which the bone can repair and remodel. Up to 49% of these stress fractures occur in the tibia (Matheson et al., 1987). Recovery times are long, and those who were previously injured and rehabilitated are far more likely to experience recurrent stress fractures. All factors leading to the development of stress fractures are not fully understood. However, it is understood that the load on the bone is determined by biomechanics–the mechanics of movement of the human body. These involve the impact forces transmitted through the musculoskeletal system during the process of running and jumping, which is often expressed as the ground reaction force (GRF).

Tibial acceleration (TA) is one proxy method to observe this impact shock. It has been demonstrated in two prospective studies that those who go on to develop tibial stress fractures have higher vertical loading and as well as greater heel pressure on impact than a healthy population prior to injury (Davis et al., 2004; Nunns et al., 2016). While these prospective studies do not measure TA and no data yet exist, it has been well demonstrated that TA peaks have significant correlation with the average loading rate ($r = 0.274–0.439$) and instantaneous loading rate ($r = 0.469$) of the vertical GRF and that those with tibial stress fractures have a higher axial TA than those without (Greenhalgh et al., 2012; Zifchock et al., 2008). These findings have led researchers to link axial TA with the potential likelihood of developing stress fracture injuries and to continue to measure it in the interest of hopefully decreasing injury rates. Faster cadence is one well-proven variable among many that can decrease TA and GRF, and therefore possibly decrease the risk of injury if changes to running patterns are made with this in mind.

Researchers have previously achieved increases in cadence, which can be done through the non invasive use of metronome pacing with no adverse effects (Nijs et al., 2021). While the overwhelming majority of these researchers utilize a combination of force plates and optical motion capture to obtain cadence and impact force data for gait modification experiments, a low mass skin mounted triaxial accelerometer can be used instead to collect TA data (Bassett et al., 2016; Norris et al., 2014). The benefit of using an accelerometer compared to the more commonly utilized force plates and optical motion capture system combination is that this method is highly portable, lower in cost, and requires less time to obtain data while still being highly accurate. This makes it a much more accessible option for trainers to practically implement accurate cadence modification in their runners to hopefully decrease rates of stress fracture injuries.

Triaxial accelerometers in inertial measurement units (IMUs) can measure forces in the axial (also referred to as vertical), anterior-posterior, and medio-lateral directions. Resultant TA then can be composed from these three. Peak axial TA is the only parameter among these that has been linked to risk of injury (Milner et al., 2006). The purpose of this study was to determine the relationship between cadence and TA in habitual runners, in the interest of potentially decreasing the risk of tibial stress fractures. It is expected that quicker cadence will be associated with lower resultant and axial TA.

Methods

I. Participants

Approval was obtained from the UTC Institutional Review Board prior to the beginning of the study. Written consent was also obtained from participants prior to the beginning of the study, detailing potential risks and benefits in a clearly explained form. Five male and six female participants were sampled for this study. All participants were injury and pain free at the time of data collection and for at least the preceding 3 months. All ran at least 10 miles per week for the last 3 months. Participants' range in miles per week was 13.5-50 miles, with an average of 23.05. Participants' age range was 19-47 years old, with an average of 34.09. One male participant was a current varsity athlete, and the rest were recreational runners. Participants range in weight was 95-180lbs, with an average of 139.64. Participants range in height was 61.75-75 inches, with an average of 67.11 inches.

II. Procedure

Participants wore their own running shoes. The accelerometer was mounted on the skin over the right distal medial tibia, with double sided tape and then affixed with a strap; to ensure accurate results with minimal artifact, the strap was as tight as the participant could tolerate (Johnson et al., 2020). Participants were instructed to run on a treadmill for 5 minutes to warm up, or less or more if they preferred, and then an additional 2 minutes to determine their preferred speed and cadence. Once the participant reached their desired running speed, cadence was counted for one minute. Cadence was accurately counted using a free mobile app on a smartphone, called Tap Counter by Svimph. Cadence was generated for 5% quicker, 10% quicker, and 15% quicker, as well as 5% slower, 10% slower, and 15% slower than the preferred cadence (Schubert et al., 2013). Metronome cadence was generated by a free mobile app called Metronome Beats by Stonekick, and played over a phone speaker at maximum volume. The metronome also had a visual cue in the form of a moving slider on the screen that participants could see at all times. TA was recorded running at a 5% quicker, 10% quicker, and 15% quicker cadence for 1 minute each, with 1 minute of no pacing in between each (Figure 1). During the no pacing period, participants were instructed to continue moving at whatever pace they preferred; participants chose to walk, run slower, or continue the same speed. Importantly, during the time that TA was recorded, each participant's speed remained at what was determined in the preferred condition.

Figure 1: Victoria Hilfiker running experiment with participant.

III. Data Handling

The inertial measurement devices utilized were triaxial Delsys Avanti EMG sensors with on-board IMUs (Delsys, Boston, Massachusetts. All experiments were conducted on treadmills to keep velocity constant while cadence was increased. Acceleration data were filtered with a 100Hz low-pass Butterworth filter (Figure 2). Stride frequency was determined by the number of tibial acceleration peaks in a minute, with each peak representing a step. The peak TA was extracted from twenty seconds of steps and averaged to determine the average TA for each cadence (Figure 3).

Figure 2 Raw axial accelerometer signal of the y axis (axial TA).

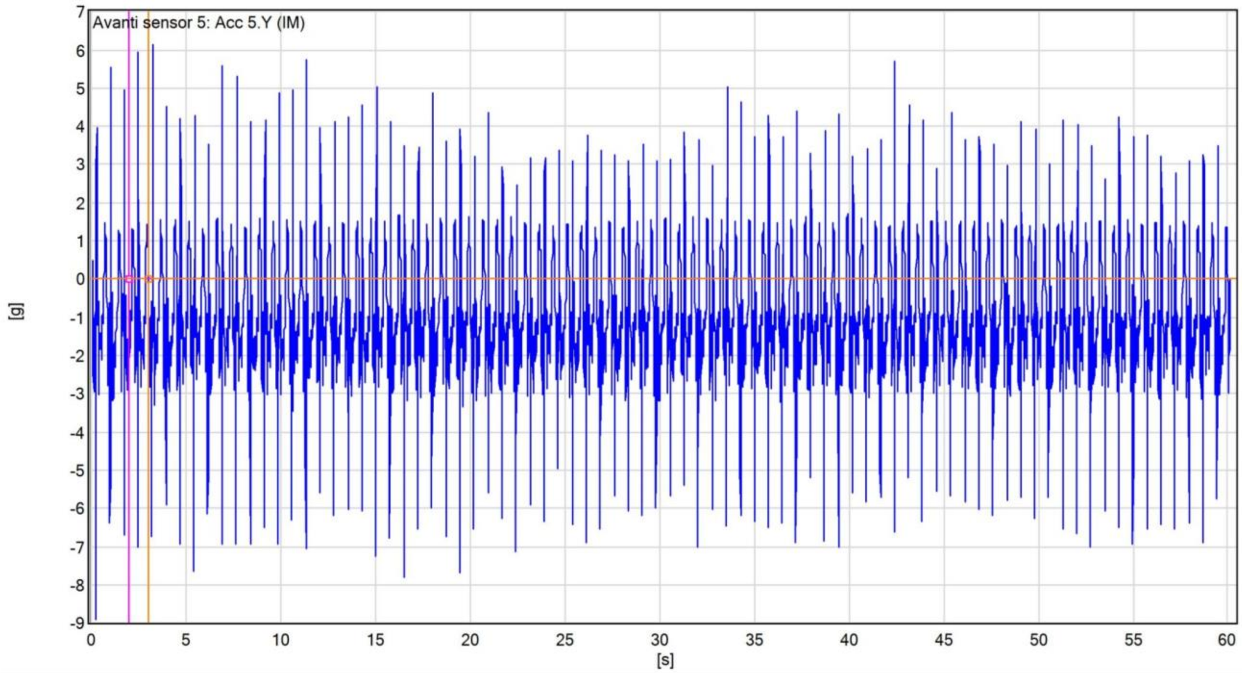
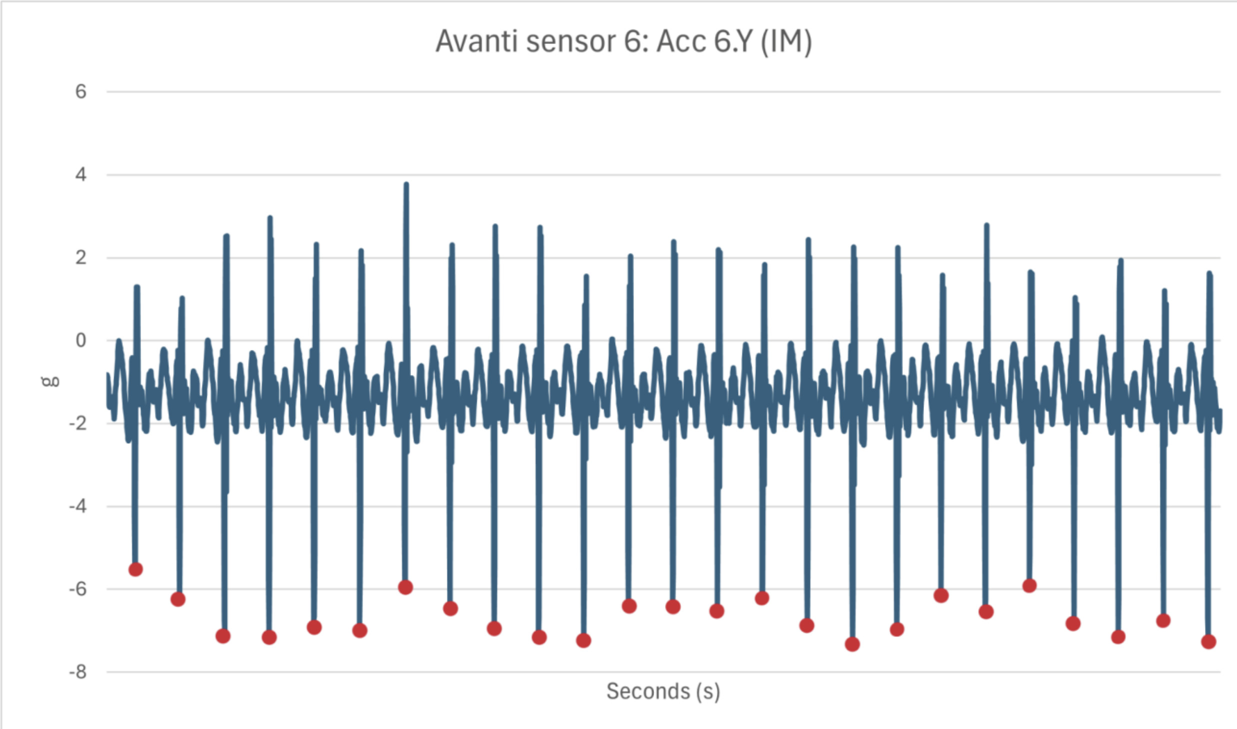


Figure 3 Twenty seconds of filtered and cleaned axial accelerometer signal of the y axis. The red dots indicate the maximum TA corresponding to each right footstep.



IV. Analyses

JASP (version 0.18.3.0, JASP Team, 2024) was used for the statistical analyses. Data were reduced manually in Excel sheets and inputted to JASP. Peak TA values were compared at each cadence using repeated measures ANOVA (Yong et al., 2018). Statistical significance was inspected; however, because this study was underpowered with a small sample size, we relied upon effect sizes to determine meaningfulness of our findings. I based my interpretation on effect sizes and interpreted a moderate effect shown by η^2 of 0.06 or Cohen's *d* of 0.5 or significance value of $p<0.05$.

Results

All data were normally distributed upon visual inspection of histograms. The average preferred speed was 6.3 ± 0.79 mph. The average cadence of the preferred condition was 168.36 ± 10.54 steps per minute prior to pacing. The repeated measures ANOVA indicated that cadence significantly affected peak axial tibial acceleration ($F=7.59$, $p<0.001$, $\eta^2=0.43$). Post-hoc tests showed that peak axial TA was significantly lower at the fastest cadence compared to the slowest cadence ($p<0.001$, Cohen's $d=0.813$, mean difference=1.085 g) (Table 2). There was also a significant difference between the 15% slower cadence and 5% faster cadence (Cohen's d=0.62, p=0.003, mean difference=0.82 g), the 15% slower and 10% faster (Cohen's d=0.60, p=0.004, mean difference=0.80 g), the 10% slower and the 15% faster (Cohen's d=0.61, p=0.003, mean difference=0.82 g), and the 5% slower and 15% faster (Cohen's d=0.47, p=0.04, mean difference=0.63 g) (Table 2).

When co-varying for sex, there is an interaction between sex and cadence for TA. At lower cadences, females have exceptionally higher TA (p=0.01), and the difference in females from their slowest to fastest cadence was significant (p<0.001) (Figure 5). The difference in males from slowest to fastest cadence did not change significantly (p=0.78) (Figure 5). Covarying for speed did not change the model (interaction p=0.88; main effect p=0.64). Covarying for mileage did not change the model (interaction p=0.18; main effect p=0.77). The average preferred cadence for females was 169 steps per minute. The average preferred cadence for males was 167.6 steps per minute. The average preferred speed for females was 6.1 mph. The average preferred speed for males was 6.6 mph.

Table 1 TA at different cadences.

Descriptives

Cadence	TA Mean (g)	SD
5% faster	5.456	1.003
10% faster	5.483	1.330

15% faster	5.195	1.091
5% slower	5.828	1.213
10% slower	6.014	1.503
15% slower	6.280	1.729

TA=tibial acceleration

Table 2 Post hoc comparisons between cadences and TA from repeated measures ANOVA.

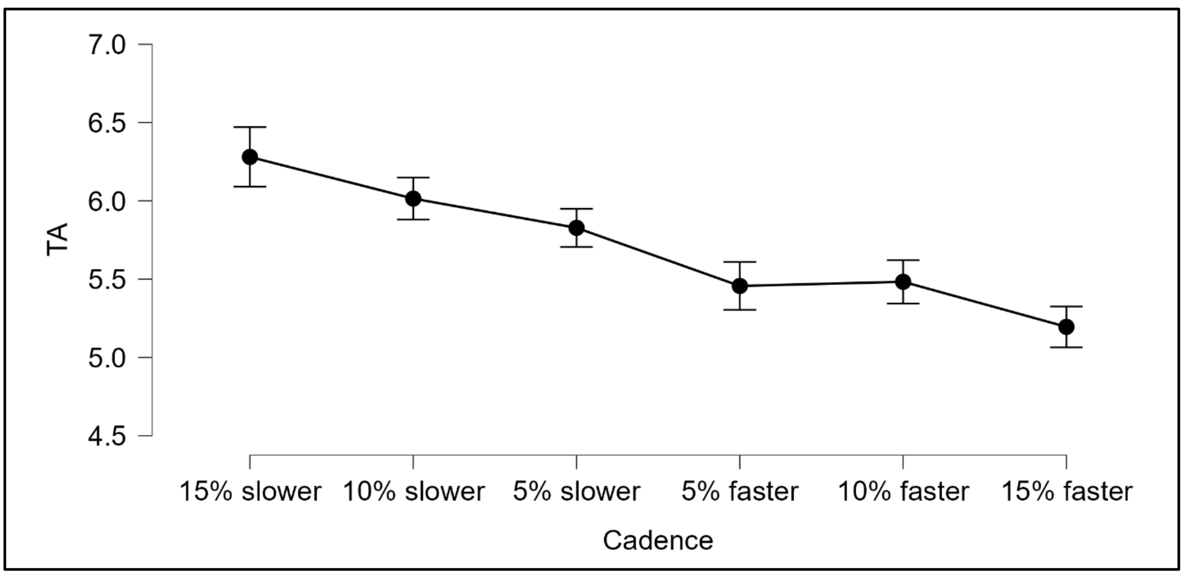
Post Hoc Comparisons - Cadence

		TA Mean Difference	t	Cohen's d	p_{holm}
15% slower	10% slower	0.266	1.285	0.199	0.851
	5% slower	0.453	2.188	0.339	0.267
	5% faster	0.824	3.981	0.618	0.003
	10% faster	0.797	3.852	0.598	0.004
	15% faster	1.085	5.243	0.813	< .001
10% slower	5% slower	0.187	0.903	0.140	0.851
	5% faster	0.558	2.696	0.418	0.095
	10% faster	0.531	2.567	0.398	0.120
	15% faster	0.819	3.958	0.614	0.003
5% slower	5% faster	0.371	1.793	0.278	0.553

	10% faster	0.344	1.664	0.258	0.614
	15% faster	0.632	3.056	0.474	0.040
5% faster	10% faster	-0.027	-0.129	-0.020	0.898
	15% faster	0.261	1.263	0.196	0.851
10% faster	15% faster	0.288	1.391	0.216	0.851

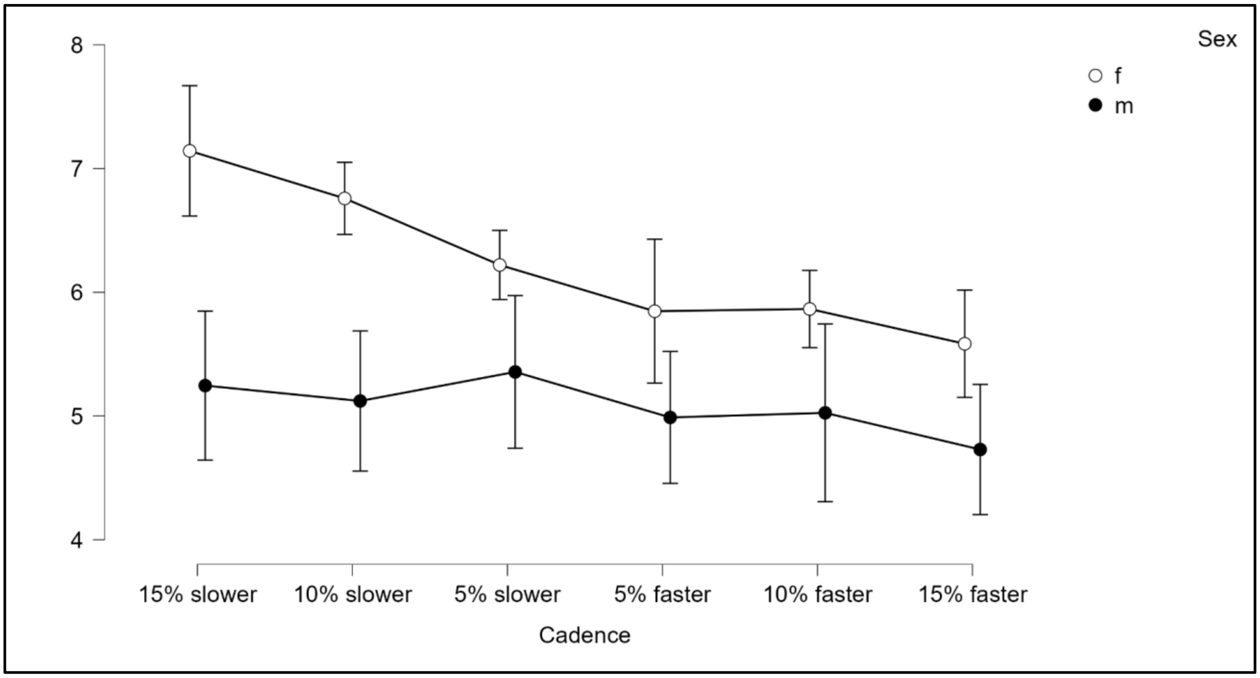
P-value adjusted for comparing a family of 15. TA=tibial acceleration

Figure 4 Depiction of relationship between cadences and TA across all runners.



TA=tibial acceleration

Figure 5 Depiction of relationship between cadences and TA across all runners by sex.



f=female; m=male; TA=tibial acceleration

Discussion

The hypothesis was fully supported: TA decreases as cadence increases (Table 1; Figure 4). Upon further examination, these differences were driven primarily by females (Figure 5). Females had significant changes in TA, and had much higher TA overall. This is in line with existing research indicating that females have higher loading rates and TA (Sinclair, 2016; Johnson et al., 2024; Schinkel-Ivy et al., 2012). Males had no significant changes in their TA, so all changes in the data set are driven by females. This suggests that increasing cadence is more important for females than males, and that any changes observed in the overall TA are driven by the females.

While existing research utilized more expensive and complex equipment, these results demonstrate that readily available tools like an accelerometer and phone apps can be used to set up reliable pacing. The lower TA at the faster cadences indicates that the GRF are lower and that there is less force being applied to the bone. This indicates that running with a faster cadence may make a person less likely to experience stress fractures. While a greater number of steps are taken at a quicker cadence and it could be thought that that would offset any benefits of the lessened impact force per step, existing research by Edwards et al (2009) shows that the strain on the bone impacts the likelihood of stress fractures more than the number of loading cycles.

As for how much faster of a cadence is best, the results show that there is little difference between the TA at 5% faster and 10% faster cadence (Cohen's d=-0.022), and that there is a small effect size between the 5% faster and 15% faster (Cohen's d=0.212) as well as the 10% faster and 15% faster (Cohen's d=0.235), indicating that running at 5% faster cadences is practically as useful in comparison. While I was unable to simultaneously capture TA during the preferred cadence condition, my data indicate that it would have fallen between 5% slower and 5% faster (Figure 1). This is in line with existing research indicating that only small modifications of 5% increases above preferred cadence are necessary to reap benefits (Heiderscheit et al., 2011). This is useful because running at a 15% faster cadence is not practical for runners; anecdotally, all of the participants found it unnatural, uncomfortable (though not painful), and quite resoundingly disliked it. A small change of 5% faster however is within reason, and with gait training, runners can make changes longer term (Clansey et al., 2014).

Pacing presents its own issues though, as in this study, some participants struggled to consistently match their footsteps to the metronome beat and visual cues. Those interested in utilizing pacing in further studies should take into account that some will not be able to achieve reliable pacing in one session, and that some people may learn better with auditory or visual cues. I hypothesize that multiple sessions spread out over a time span could remedy this, which is another route for further studies.

Also of note with regards to the reliability of results is that TA cannot account for the internal force on the tibia, and thus it cannot present the actual maximum stress placed on the tibia (Zanderbergen et al., 2023). EMG data could be used in conjunction with TA data to potentially give a better picture of the force on the bone, which could be a route for future research. However, it's important to note that EMG data, like TA, is an estimate; it measures electrical signals of the motor units, not the actual contractile force which can differ from the electrical signal (Hug et al., 2015). Because maximum force on the tibia is not at peak TA, reducing the amount of force at heel strike (which is what increasing cadence to decrease TA aims to do) may do little to decrease the likelihood of stress fracture; however running with a shorter stride may have a stronger effect since it would reduce knee flexion and therefore internal muscle forces (Sasimontonkul et al., 2007). This is not in disagreement with the current study, but rather further supports the use of TA data in cadence increases as a method of decreasing potential risk of stress fractures.

First, while TA does not present all forces, it accurately depicts an increase in force experienced by the body when compared to other methods of observing force on the body (Crowell & Davis, 2011; Nunns et al., 2016; Greenhalgh et al., 2012). Second, running with a quicker cadence results in a shorter stride length when speed remains constant. A study by Mercer et al (2003) found that shock absorption by the body increased when stride length increased ($p<0.05$), independent of stride rate. This is due to increased demand on the body to absorb shock through the mechanism of increased eccentric muscle activation, primarily at the knee (Derrick et al., 1998). It is suggested that decreases in stride length can therefore reduce energy absorption demands which would be useful for decreasing injury risk as well as the treatment of existing injuries (Heiderscheit et al., 2011). So, while TA is only a measure of external forces on the tibia, a lower TA can indicate that decreases in stride length are having the intended external effect which indicates that they are also having the beneficial internal effect of decreasing knee flexion and muscle strain on the tibia. A faster cadence is associated with decreases in injury risk to the lower extremities for this reason (Wellenkotter et al., 2014; Heiderscheit et al., 2012; Edwards et al., 2009; Yong et al., 2018, Chumanov et al., 2012).

I. Limitations

We did not control for cadence adherence, as we had no method to do so without adding more tools to the project such as motion capture cameras which are not within the scope of this project. We also did not examine whether participants were rearfoot strike or forefoot strike runners as wear pattern is not a reliable indicator and we had no other method available without adding more equipment that was not within the scope of this project. Also of importance is that all data collection was conducted on a treadmill, and the relationship between TA and running surface compliance is not straightforward (Sheerin et al., 2019). The reliability is also limited by the fact that the available accelerometers that were used in this experiment are heavier than recommended which means that the oscillations potentially do not accurately represent the oscillations of the bone; however, they were wrapped on tightly which helped the validity (Johnson et al., 2020). Another limitation is that the resultant TA was not calculated, since axial TA is the only metric associated with risk of stress fractures as the vertical motion produces the highest peaks in acceleration (figure 2). Nonetheless, the omission of resultant TA could possibly affect the reliability of the results if the sensor is not properly aligned along the vertical plane, and we did not ensure alignment with any other procedure besides estimating it. However, in existing research, resultant TA tends to show the same patterns as axial TA (Milner et al., 2020).

Conclusion

This study supported the hypothesis that increased cadence is associated with decreased TA, primarily for females. This decrease in TA is indicative of decreased external force being applied to the tibia through GRF. A faster cadence at the same speed also results in shorter stride lengths, which has many proven health benefits. The decrease in TA with increase in cadence suggests that the one session gait modification had the intended effect of decreasing external force on the bone, which means that necessarily in this case due to decreased stride length, the internal force on the bone was also decreased to some degree. Results of this study support my hypothesis that TA measured by an accelerometer is correlated with running cadence. Results of this study suggest that increasing cadence while maintaining the same speed could result in potentially lowered risk of injury, and is especially important for females.

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Conflict of Interest Statement

There was no conflict of interest.

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