

The Effects of Running Cadence Manipulation on Plantar Loading in Healthy Runners

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Key words

- pressure
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Abstract

Our purpose was to evaluate effects of cadence manipulation on plantar loading during running. Participants ($n=38$) ran on a treadmill at their preferred speed in 3 conditions: preferred, 5% increased, and 5% decreased while measured using in-shoe sensors. Data (contact time [CT], peak force [PF], force time integral [FTI], pressure time integral [PTI] and peak pressure [PP]) were recorded for 30 right footfalls. Multivariate analysis was performed to detect differences in loading between cadences in the total foot and 4 plantar regions. Differences in plantar loading occurred between cadence conditions. Total foot

CT and PF were lower with a faster cadence, but no total foot PP differences were observed. Faster cadence reduced CT, pressure and force variables in both the heel and metatarsal regions. Increasing cadence did not elevate metatarsal loads; rather, total foot and all regions were reduced when healthy runners increased their cadence. If a 5% increase in cadence from preferred were maintained over each mile run the impulse at the heel would be reduced by an estimated 565 body weights*s (BW*s) and the metatarsals 140–170 BW*s per mile run despite the increased steps taken. Increasing cadence may benefit overuse injuries associated with elevated plantar loading.

Introduction

Over the past decade, it is estimated that the total number of marathon finishers increased by 37% in the United States (299,018 in 2000 & 473,354 in 2009) [22]. One consequence of the rise in popularity of running is an increased occurrence of running related injuries [11]. According to one study, it is estimated that approximately 90% of those training for, or participating in a marathon, will experience an injury [33], and as many as 76% of recreational runners will become injured each year [11]. The majority of these injuries occur in the distal lower extremity, with between 5.7–39.3% occurring in the foot [11]. Despite the acknowledgement that the etiology of overuse related running are likely multifactorial [2,17], impact loading has been implicated by many researchers since the late 1970's as having a role in lower extremity overuse injury [4,18]. Specific to the foot, the cyclic loading of the metatarsals in distance running has been reflected in higher bone strain based on cadaveric and mechanical models of stress fracture [8,13]. More recently, elevated forefoot loads

were observed in running a marathon, suggesting a shift in loading occurs over the course of a run, which may help explain the occurrence of metatarsal pain and stress fracture in long-distance runners [26]. In a recent review of Achilles tendinopathy, altered plantar pressures and elevated external loads were implicated in the development of injury [24]. Similarly, applied loads have also been implicated in running-related plantar fasciitis [29], which has been shown to be another commonly reported running injury [34].

The developments of in-shoe loading systems have enabled plantar loading variables to be quantitatively measured within footwear during walking and running in a reliable and relatively accessible manner [20,21]. These systems allow for the collection of consecutive foot strikes both within and beyond the laboratory setting and are readily available for use in health care settings. In-shoe loading systems allow for the assessment of total foot loading as well as regional foot loading to be quantified. High plantar loads applied to the foot during stance (elevated pressure time integrals) during running and walking have been

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positively related to foot pain and malalignment [3]. In addition, altered plantar force variables (center of force displacement, regional peak force, time to peak force, force-time integral, peak pressure and center of pressure deviation) have been identified as risk factors for running-related Achilles tendinopathy [1,12,19], lower leg pain [5] and patellofemoral pain [35]. Remarkably, plantar loading variables (position of center of pressure during foot flat and toe off) have also been found to be more discriminative than kinematic variables when prospectively predicting lower leg injury risk [37].

In the interest of altering gait mechanics and thereby reducing loads experienced by the ankle, knee, and hip, alterations in running form have been suggested [5, 7, 15]. Recent studies examining the influence of cadence manipulation on joint kinematics, kinetics and lower extremity muscle activation patterns [5, 7, 15] suggest that when speed is held constant, small increases in cadence (5–10%) result in decreases in step length, hip adduction angle, frontal plane joint moment at the hip, braking impulse and foot inclination angle. Furthermore, decreased peak vertical ground reaction force occurrence and reduced knee flexion angles while running at faster cadences contributed to a decrease in energy absorption requirements at the knee by as much as 34% [15]. Faster cadences also show increased hip and lower extremity muscle activity, primarily during late swing phase [5]. Research suggests that these faster cadences decrease stride length and vertical oscillation during treadmill running [5, 7]. When the body center of mass oscillation is reduced due to a shortened stride length coupled with a faster cadence, a reduction in loading variables and motions linked to injury should theoretically be improved. Therefore, recent studies have suggested that increasing cadence may cause favorable changes for reducing injury potential as well as facilitating recovery from some existing injuries [9, 10, 16, 28, 38].

However, it is unknown how plantar loading changes when cadence is systematically manipulated for a given running speed. Simply increasing a runner's cadence has the potential to alter foot strike patterns and plantar loading distribution. Increasing running cadence typically minimizes foot inclination angle [15], likely resulting in a mid- to forefoot strike pattern. This change in running style potentially reduces plantar loading at the heel and shifts plantar loads anterior toward the metatarsal regions. Evaluation of plantar loading during running while manipulating cadence will provide insight into alterations in regional foot loads that cannot be captured through traditional force platform measurements. Thus, the objective of this study was to evaluate the effects of cadence manipulation on in-shoe plantar loading (PF, FTI, PP and PTI) during running at a constant speed under various step rate conditions. It was hypothesized that a 5% increase of cadence from preferred would decrease total foot and heel plantar loading and increase plantar pressure loading in the metatarsal regions. We expect that decreasing cadence by 5% will increase total foot, heel and metatarsal loading. Therefore overall loading changes are expected due to the changes in total foot loading, whereas the peak total foot loads would be increased with a lower cadence and reduced with a lower cadence.

Materials and Methods

Thirty-eight healthy recreational runners (19 males and 19 females; mean $\pm$ SD: age = 23.0 $\pm$ 3.5 years, height = 174.2 $\pm$ 7.6 cm,

mass = 30.6 $\pm$ 4.7 kg) agreed to participate in this study. All subjects ran a minimum of 12.87 km per week, had been running for at least 6 months before study enrollment, and had prior experience with running on a treadmill. Subjects were excluded if they had any known musculoskeletal disorder that would affect running, or had experienced a lower extremity injury within the past 6 months. The university Institutional Review Board approved all study procedures, and all subjects provided consent prior to participation. Research was conducted in accordance with international ethical standards [14].

The Pedar-X (Novel GmbH, Munich, Germany/St. Paul, MN) insole shoe system, consisting of analogue to digital (A/D) conversion electronics and a battery pack, was used to collect kinetic plantar loading data. Each in-shoe sensor was connected to the analog to digital conversion box by cables secured by straps connected to the subject's lower extremities. The sensor insoles were 2 mm thick and consisted of 99 individual sensors that sampled at a rate of 150 Hz. Each capacitive sensor in the matrix of each sensor insole was calibrated at discrete pressure values from 0–600 KPa in an air pressure chamber as described by Kernozek, et al. [20]. Bluetooth technology was used to transmit the data to a laptop computer, and Pedar software (Novel GmbH, Munich, Germany/St. Paul, MN) was used to store and analyze data. Each running condition was performed on a SensorMedics series 2000 treadmill (Anaheim, CA). QWiK TIME Quartz metronome (QT3, China) was used as an auditory cue and to monitor cadence.

Prior to data collection body measurements were collected from each subject. Height and leg length (greater trochanter to medial malleolus) were measured using a retractable tape measurer. A calibrated large dial Health O Meter scale (12" x 19" base dimensions; 400 lb capacity) was used to measure the subject's body weight in pounds. Participants were sized and fitted to a New Balance running shoe (model 625, Boston, MA), and a calibrated sensor (0–60 N/cm²) was placed into both shoes. Finally, the sensor cabling was secured to the body to minimize the risk of falling (• Fig. 1). Only the sensor insole under the dominant foot, the preferred foot used to kick a ball [25], was sampled.

Participants were allowed 5 min to warm-up on the treadmill at their preferred running pace. Each participant ran 4 min at his/her preferred speed (2.75 $\pm$ 0.44 m/s) for 3 cadence conditions: preferred cadence (preferred), 5% increased cadence from preferred (+5%), and 5% decreased cadence from preferred (–5%). For each participant, treadmill speed was consistent across all conditions. In a manner similar to Heiderscheit et al. [15], preferred cadence was determined by visual inspection by counting foot strikes over 60 s at normal running speed; a metronome was used to visually confirm that cadence and provided an auditory cue to adjust cadence $\pm$ 5%. Visual inspection was used to confirm accuracy of the conditions. Data collection in the $\pm$ 5% cadence conditions began after a 1 min adjustment period when participants had fully adapted to the new running parameter. Our goal was to determine if there were immediate changes in plantar loading with rather small changes in cadence ($\pm$ 5%) and all participants were able to match the adjusted cadence $\pm$ 1 step per minute within the practice period. The order of the conditions was randomly assigned in order to minimize fatigue effects. Data were collected for 30 consecutive strides and averaged at each cadence.

Plantar loading variables analyzed for each participant included: contact time (CT), peak force (PF), force time integral (FTI), peak pressure (PP) and pressure time integral (PTI) for 4 anatomical



Fig. 1 Experimental setup.

areas of the plantar foot and the total foot. Multimask software was used to regionalize 4 anatomical areas of the plantar foot: the heel (H), medial metatarsal (MMT), central metatarsal (CMT) and lateral metatarsal (LMT) (◻ Fig. 2). These regions were selected because they have been shown to incur the most loading during running [30,36]. Plantar variables for each subject were averaged over each condition for the total foot and 4 plantar regions of the foot. Both PF and FTI were normalized by body weight (BW). Repeated measures multivariate analysis of variance was performed for each of the 4 plantar regions evaluating the effects of cadence manipulation on the variables of interest. Post hoc pairwise comparisons ($\alpha=0.05$) were performed, and Cohen's D effect sizes were calculated to evaluate condition differences. An effect size ≥ 0.2 was considered to be small, ≥ 0.5 medium, and ≥ 0.8 large [6].

Results

Overall, total foot plantar loading variables differed when cadence was increased and decreased by 5% from preferred ($p<0.001$). Plantar loading variables in 4 specific regions of the foot also differed with cadence manipulation ($p<0.001$). ◻ Table 1 provides complete results for all comparisons including statistical results and effect sizes for all univariate tests. *Total foot* CT and PF were lower with increased cadence as compared to preferred cadence ($p<0.001$ and 0.004, respectively). CT was greater with decreased cadence as compared to preferred cadence ($p=0.023$). *Heel* FTI, PP and PTI were lower when cadence was increased ($p<0.001$, 0.020 and <0.001). PF and PP were greater when cadence was decreased ($p=0.001$ and 0.020). *Medial metatarsal* CT, FTI and

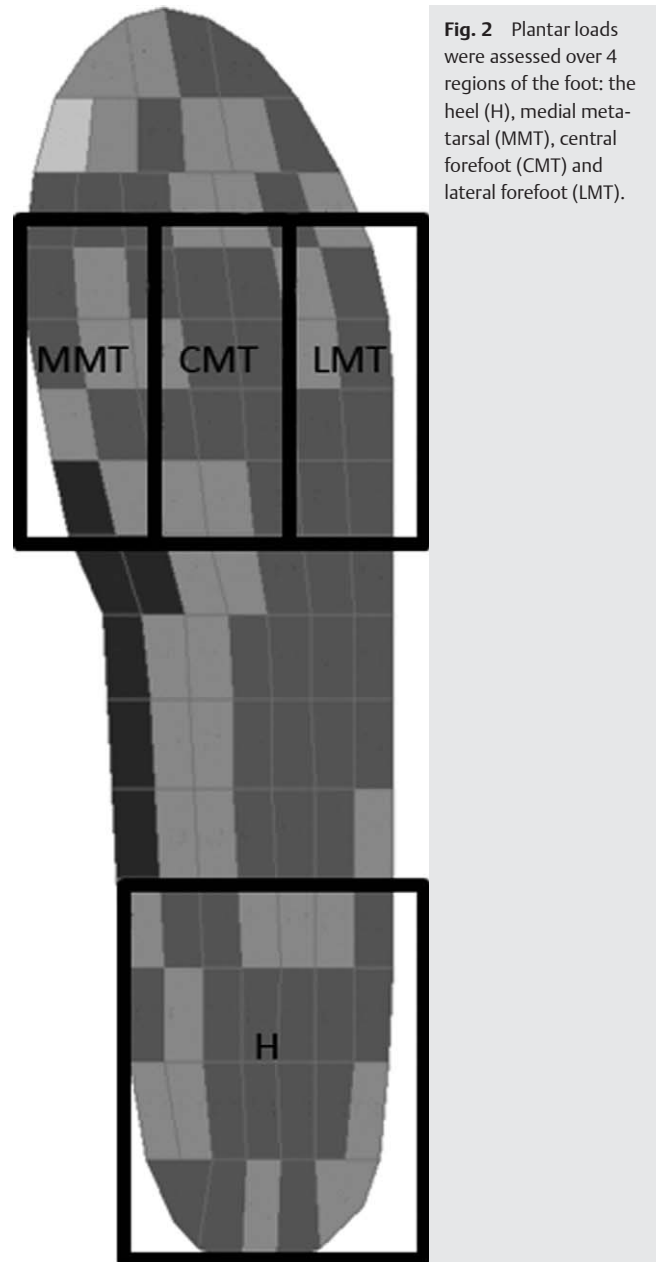


Fig. 2 Plantar loads were assessed over 4 regions of the foot: the heel (H), medial metatarsal (MMT), central forefoot (CMT) and lateral forefoot (LMT).

PTI were lower when cadence was increased ($p=0.001$, 0.046, and 0.006). CT was greater when cadence was decreased ($p=0.016$). *Central metatarsal* CT, FTI, and PTI were lower when cadence was increased ($p=0.001$, 0.048, and 0.006) and greater when cadence was decreased ($p=0.006$, 0.038, and 0.016). *Lateral metatarsal* CT, PP and PTI decreased when cadence was increased ($p<0.001$, <0.001 , and <0.001). PP increased when cadence was decreased ($p=0.002$).

Discussion

Our findings suggest that slight modifications in cadence ($\pm 5\%$ from preferred) have small but statistically significant effects on total and regional plantar loading during running. Faster cadences decreased heel loading and did not increase metatarsal loads as expected; rather, metatarsal loads were generally reduced with a faster cadence. Furthermore, running at increased cadences appeared to be inversely related to total foot plantar

Table 1 Mean (SD) plantar loading variables, p-values and effect sizes during running at preferred, decreased and increased cadences.

	Preferred cadence	–5% cadence	+5% cadence	–5% vs. preferred cadence		+5% vs. preferred cadence	
				p-value	Effect size	p-value	Effect size
Contact time (ms)							
Total foot	285.7 (31.4)	292.5 (36.8) (+)	276.0 (32.5) (–)	0.02***	–0.2	<0.01***	0.3
Heel	228.2 (58.3)	241.0 (67.2)	223.2 (64.6)	0.06	–0.2	0.86	0.1
Medial forefoot	259.5 (29.4)	266.1 (33.7) (+)	253.3 (29.5) (–)	0.02***	–0.2	<0.01***	0.2
Central forefoot	264.7 (29.2)	272.8 (35.4) (+)	259.0 (29.1) (–)	0.01***	–0.2	<0.01***	0.2
Lateral forefoot	269.4 (29.5)	274.3 (35.8)	261.5 (29.7) (–)	0.15	–0.2	<0.01***	0.3
Force time integral (%BW*s)							
Heel	7.1 (2.6)	7.2 (2.8)	6.4 (2.7) (–)	1.00	–0.0	<0.01***	0.3
Medial forefoot	5.3 (1.1)	5.5 (1.3)	5.1 (1.2) (–)	0.11	–0.2	0.04***	0.2
Central forefoot	5.9 (0.9)	6.1 (1.0) (+)	5.7 (1.0) (–)	0.04***	–0.2	0.04***	0.2
Lateral forefoot	3.3 (0.9)	3.4 (1.0)	3.2 (1.0)	0.30	–0.1	0.13	0.1
Peak force (%BW)							
Total foot	198.6 (39.3)	196.5 (38.7)	194.0 (39.2) (–)	0.38	0.05	<0.01***	0.1
Heel	87.8 (27.9)	92.5 (29.7) (+)	83.3 (28.3)	<0.01***	–0.2	0.14	0.2
Medial forefoot	35.0 (6.3)	35.7 (6.6)	34.6 (6.6)	0.54	–0.1	1.00	0.1
Central forefoot	38.9 (6.4)	39.2 (6.7)	38.6 (6.7)	1.00	–0.1	0.94	0.1
Lateral forefoot	20.8 (5.2)	21.2 (5.7)	20.8 (5.5)	0.88	–0.1	1.00	0.0
Peak pressure (kPa)							
Total foot	284.8 (62.6)	297.2 (61.0)	279.5 (65.6)	0.16	–0.2	1.00	0.1
Heel	201.6 (57.4)	210.3 (63.7) (+)	190.2 (60.5) (–)	0.02***	–0.1	0.02***	0.2
Medial forefoot	261.4 (67.0)	256.0 (68.2)	248.6 (71.1)	0.07	0.1	0.07	0.2
Central forefoot	232.6 (35.9)	234.7 (37.5)	229.7 (36.8)	0.97	–0.1	0.71	0.1
Lateral forefoot	198.0 (37.0)	190.2 (38.4) (+)	187.3 (36.3) (–)	<0.01***	0.2	<0.01***	0.3
Pressure time integral (kPa*s)							
Heel	20.9 (5.9)	21.3 (6.5)	19.2 (6.2) (–)	0.49	–0.06	<0.01***	0.3
Medial forefoot	40.4 (10.5)	40.2 (10.5)	37.4 (10.8) (–)	1.00	0.0	<0.01***	0.3
Central forefoot	36.3 (6.7)	37.4 (7.4) (+)	35.1 (6.7) (–)	0.02***	–0.2	<0.01***	0.2
Lateral forefoot	31.7 (6.1)	31.5 (6.3)	29.7 (5.9) (–)	1.00	0.04	<0.01***	0.4

*** indicates $p < 0.05$, (+) indicates a positive change, (–) indicates a negative change

loading variables including contact time, total foot peak force, and total foot peak pressure. Thus, our hypotheses related to total foot and heel load was supported while our hypotheses related to metatarsal loads were rejected.

Our findings may be partly explained by a decreased foot inclination angle and decreased horizontal distance from the COM to the heel at initial contact, which was observed by Heiderscheit et al. [15] when cadence was increased 5 and 10% from preferred. Additionally, such a decrease in COM vertical excursion observed when running at a faster cadence [7] may reduce total foot loading by minimizing the magnitude of the vertical ground reaction force and therefore the plantar loading experienced throughout the foot in running. On the contrary, an increase in COM vertical excursion with a reduced cadence may be related to the slightly greater loads observed in the study with a 5% cadence reduction.

The plantar loading changes observed in this study were relatively small with differences ranging from 2.4–8.0%. However, if a 5% increase in cadence from preferred was maintained over a run at that runner's chosen speed, these findings would equate to decreased impulsive loads in the heel by 565 BW*s and the metatarsal region by 140–170 BW*s for each mile run. This suggests that despite the greater number of steps taken during the course of the run, running at a faster cadence has the potential to reduce the overall magnitude of loads as well as the regional loads experienced by the runner. Greater reductions in loads may be observed with greater increases in cadence, with previous work having examined cadence adjustments ranging from 5–30% [5, 15, 16]. Hobara et al. [16] report that cadence increases of 17–18% may be needed to minimize loading related to tibial

stress fracture risk. Thus, the evaluation of plantar loads with greater step rate adjustments would similarly help identify an optimal cadence to minimize plantar loading variables.

Plantar loading variables appear to be important to running-related injury. Burns et al. [3] evaluated plantar loading in individuals with pain and a pes cavus foot structure. They reported a greater peak pressure time integral from the entire foot, metatarsal regions and a greater peak pressure under the heel in those with foot pain and malalignment. Runners who developed Achilles tendinopathy demonstrated elevated peak force and force time integrals in the lateral metatarsal region prior to injury [12]. Greater peak force throughout the foot, especially under the second and third metatarsal and at the lateral heel, was also found in runners who developed patellofemoral pain. Because the slight 5% increase in cadence in our study decreased loads in the total foot as well as loading throughout heel and metatarsal regions, this increase in cadence may be a useful adjunct to rehabilitation strategies for running injuries such as foot pain, Achilles tendinopathy and patellofemoral pain [3, 12, 35]. It should be mentioned, however, that our differences were small and this study did not examine a cause and effect relationship between cadence and running related injuries. Additionally, in our investigation, only the acute effects of cadence manipulation were examined. Therefore, until more research is performed in this area, cadence advice should be considered within the context of clinical presentation.

Several studies support a relationship between injury and altered loading between the medial and lateral aspects of the foot during different phases of the gait cycle [12, 37]. Altered

medial/lateral loading was observed in runners who experience Achilles tendinopathy [12]. Willems et al. [37] reported a more centralized heel strike, greater pressure on the medial aspect of the foot and a more significant lateral roll-off in runners who experienced patellofemoral pain. While not a specific outcome in our study, visual inspection of our study's mean data suggest that a change in cadence did not influence medial/lateral loading. Rather, similar reductions in medial/lateral loading were observed with increasing cadence, suggesting little influence on directional loading. This finding would appear to warrant further study.

A limitation of this study is that the changes observed in plantar loading variables were measured immediately after cadence manipulation, making it therefore unknown if these will be maintained at long-term follow-up. In addition, participants were provided a small window of time for adaptation to the new cadence. All participants were able to adapt to the new cadence within the time constraints with an observed variation of ± 1 BPM. However, the time requirements needed to adequately re-train a runner's cadence is in need of further investigation. Researchers have shown that runners who received audiovisual feedback for 15 sessions over 3 weeks were able to achieve significant reductions in step length [23]. In their study, participants were able to shorten step length to target levels on the first day and maintain these target levels throughout the testing; given that treadmill speed was held constant, alterations in cadence would have been necessary for a reduction in step length. Thus the research by Morgan et al. [23] suggests that both short-term and long-term alterations in cadence were shown to reduce step length with training. Nonetheless, our study focused on immediate changes in loading with auditory cueing from a metronome. It is uncertain if these plantar loading-related changes would persist without cueing or if these changes would remain long term. A second limitation is related to our use of a treadmill during testing. Studies have shown that some kinematic and kinetic parameters in running on a treadmill are not exactly the same as in running over ground [27,31,32]. While anterior/posterior and medial/lateral ground reaction forces are lower than during treadmill running when compared to running over ground, the overall pattern of the ground reaction force profile across the stance phase is similar [32]. To adequately control experimental parameters such as running speed and cadence the researchers chose to utilize a treadmill, which may limit the generalization to other running environments. A third limitation in our study may be related to the comprehensive nature of our analysis of plantar loading data, which inflated the type I error rate due to the statistical comparisons of 5 dependent variables for each plantar region. Given to the complexity of the description of human movement and plantar loading, the authors chose to use a multivariate analysis with follow-up univariate tests using a Bonferroni correction. Type I error is a frequent concern within our field since it is very difficult to adequately choose a small set of dependent variables that adequately describe our biomechanical constructs of interest.

In conclusion, our findings suggest that slight cadence manipulations alter total and regional plantar loading during running. A 5% increase in cadence resulted in runners experiencing, on average, 3.4% less time in contact with the ground during each foot strike. This resulted in 2.2% lower heel loading and 2.2–2.4% lower forefoot loading over time. These results suggest that healthy runners who increase cadence by 5% should experience

decreases in plantar loading that may be associated with lower extremity injuries. Therefore, cadence should be considered in technique training for runners and during the rehabilitation of running injuries associated with elevated plantar loading.

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