Cycling Cleat Positioning Influences Achilles Tendon Strains, but at What Energetic Cost?

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ABSTRACT

Introduction: Patellar and Achilles tendinopathy are overuse injuries associated with repetitive strain and often arise following sudden increases in exercise intensity or volume. As such, interventions that reduce tendon strain may represent effective methods for reducing overuse injury risk in this at-risk population of novice cyclists. One potential intervention is through the anterior/posterior positioning of the cycling cleat.

Methods: Ten recreational athletes cycled on a stationary cycle ergometer at three cleat positions (neutral, 20 mm anterior, 20 mm posterior), four power outputs (150 W, 200 W, 250 W, 300 W) and two rider positions (seated, standing) for a total of 24 conditions. Motion capture and plantar pressure data were collected, and peak Achilles and patellar tendon strains were obtained using musculoskeletal modelling. Metabolic output for each condition was also modelled using a combination of musculoskeletal modelling and a previously published metabolic model.

Results: Peak Achilles tendon strain was significantly reduced with a posterior cleat position compared to a neutral (seated: $p = 0.047$; standing: $p = 0.019$) and anterior (seated: $p < 0.001$; standing: $p < 0.001$) position during both standing and seated cycling. However, peak patellar tendon strain ($p = 0.358$) and modelled metabolic output ($p = 0.778$) were not influenced by cleat position.

Conclusion: Cycling with a 20 mm posterior cleat position represents an effective intervention for reducing the risk of developing Achilles tendinopathy without concurrently increasing patellar tendon strain and sacrificing performance.

KEYWORDS

Biomechanics, Overuse Injury, Cycling, Tendinopathy, Metabolic Cost

INTRODUCTION

Cyclists are at a high risk of developing an overuse injury, with overall incidences of 47.4% (Dannenberg et al., 1997) and 51.5% (de Bernardo et al., 2012) amongst recreational and elite cyclists, respectively. Achilles and patellar tendinopathy are two of the most common overuse injuries amongst elite cyclists, with a combined incidence of 15.4% of all observed injuries (de Bernardo et al., 2012). Overuse injuries generally have an insidious onset, often following rapid increases in training load or abrupt changes in exercise modality (Kannus, 1997; Milgrom et al., 1985; Salzler et al., 2012), and are characterized by debilitating pain that may persist for years (Cardoso et al., 2019; Kettunen et al., 2002; Rutland et al., 2010). Such a 'perfect storm' for the development of lower-limb tendinopathy currently exists within the cycling community, as there has been a 1200% increase in at-home stationary cycling within the past 5 years likely due to the increased popularity of interactive at-home stationary cycle ergometers coupled with changes in exercise preferences from the COVID-19 pandemic (Peloton, 2024). These at-home stationary cycle ergometers have recently been combined with "clipped in" (i.e., clipless) pedals with users having minimal to no knowledge of how to fit a cleat prior to cycling.

While the exact pathophysiology of tendinopathy is unknown, the progressive degeneration of tendon material properties in response to repetitive loading is analogous to the principle of mechanical fatigue, in which a material is weakened and eventually fails from damage accumulation caused by repetitive sub-ultimate loading. Mechanical fatigue principles have previously been utilized to model overuse injury risk in several biological tissues, including the tibia (Edwards et al., 2009), metatarsal bone (Firminger et al., 2015), and the patellar/Achilles tendons (Firminger et al., 2020). Biological tissues display non-linear relationships between tissue-level strain and fatigue life (i.e., the number of repetitive loading cycles to failure), indicating that minor changes to tendon strain may elicit an order of magnitude change in fatigue life (Firminger and Edwards, 2021; Wren et al., 2003). As such, interventions that offer even modest decreases to tissue-level strains are likely to greatly reduce tendinopathy risk.

Many aspects of bicycle fit have been examined with respect to lower-limb joint biomechanics when cycling, including vertical/horizontal saddle position, handlebar height/reach, and crank length (Wanich et al., 2007). A low saddle height has been shown to increase internal knee adduction moment (Wang et al., 2020), while no changes in knee joint contact forces were observed due to anterior/posterior shifts in saddle position (Bini et al., 2013). An opposite effect was observed at the ankle, as greater seat height increased the ankle joint's contribution to total net lower-limb mechanical work (Bini et al., 2010). Changes in power output and rider position have also been studied during cycling. More specifically, muscle activation at the medial gastrocnemius was relatively unchanged as cycling intensity (i.e., power output) increased, while significant increases in muscle activity were observed at the rectus femoris, vastus lateralis, biceps femoris, and gluteus maximus (Holliday et al., 2023), suggesting larger changes in more proximal muscle activation with increasing mechanical work during cycling. Rider position has also been shown to influence muscle activation, as cycling in a standing position increased muscle activation at the rectus femoris and vastus medialis; however, once again, no changes in activity of distal muscles such as the medial gastrocnemius (Berkemeier et al., 2020).

One additional area of adjustment is the positioning of the cleat that attaches the cycling shoe to the pedal. Most commercially available clipless road cycling shoes now contain adjustable threaded inserts that allow the user to adjust the anterior/posterior position of the cleat by upwards of 40 mm. It is generally recommended to align the cleat to the first metatarsal head (Silberman et al., 2005); however, there is an absence of scientific evidence to back this suggestion. Prior research has illustrated that compared to a preferred (i.e., more anterior) cleat position, cycling with the cleat mounted posteriorly produced no changes to steady-state oxygen consumption or energy cost (Millour et al., 2020; Paton, 2009; Viker and Richardson, 2013). However, decreases in soleus and medial/lateral gastrocnemius activation were observed with a posterior cleat position (Litzenberger et al., 2008; Mcdaniel et al., 2013; Millour et al., 2020). It could be that the oxygen consumption measure may not be sensitive enough to detect metabolic changes associated with the previously observed muscle activation changes, and a modelled metabolic procedure may be suitable to detect these changes. Cycling shoe cleat position represents a potential mechanism to alter lower-limb tendon loading, yet no studies have examined the effect of cleat positioning on patellar and Achilles tendon loads.

The primary purpose of this study was to examine the effect of cycling cleat anterior/posterior positioning on lower-limb tendinopathy risk during seated and standing cycling. Further, the secondary purpose of this study was to determine the effects of the same cycling cleat and rider positioning on modelled energetics of cycling. For the primary purpose, our first hypothesis was that a posterior cleat position would reduce Achilles tendon strain due to a decreased moment arm between the ankle joint centre and the point of force application (i.e., centre of pressure) and reductions in muscle activity from previous literature, while an anterior position would produce the opposite effects resulting in increased Achilles tendon strain. For the patellar tendon, our second hypothesis was that there would be opposite trend of increasing and decreasing patellar tendon strain with a more posterior and anterior cleat position, respectively. Further, for our secondary purpose, our third hypothesis was that a posterior position would reduce modelled energetic cost of cycling because of a reduction in plantarflexor muscle activation, similar to the results of prior research (Litzenberger et al., 2008; Mcdaniel et al., 2013; Millour et al., 2020). Finally, our fourth hypothesis was that cycling in a standing posture would increase both patellar and Achilles tendon strains due to increased weightbearing requirements of supporting the full body weight on the pedal, also increasing modelled energetics across cycling cleat conditions.

METHODS

Data Collection and Analysis

Ten recreational cyclists (3 female, 7 male) were recruited from the greater Calgary area to participate in this study. Participants were free of any lower limb injury at the time of data collection and had no history of lower-limb injury within 6 months prior to participating in the experiment. This sample size was selected using previous stationary cycling data with modelled anterior/posterior changes in cleat position, assuming a power of 0.8 and an alpha criterion of 0.05 (Asmussen et al., 2024). Ethics approval was obtained from the University of Calgary's Conjoint Health Research Ethics Board (REB21-1480), and all participants provided written informed consent.

All testing was performed on an electromagnetically braked cycle ergometer (Velotron; SRAM, Chicago, IL, USA). Prior to cycling, participants were allowed to adjust the seat height, seat fore/aft position, handlebar height, and handlebar fore/aft position of the cycle ergometer to their personal preference and kept constant throughout all conditions. Participants cycled at four power outputs (150W, 200W, 250W, 300W), three cleat positions (neutral, anterior, posterior), and two rider positions (seated, standing) for a total of 24 conditions in a commercial cycling shoe (Shimano; RC100, Sakai, Osaka, Japan). In the factory neutral position, the cleat was centered approximately under the first metatarsal head, consistent with previous literature recommendations (Silberman et al., 2005). For the anterior and posterior cleat positions, the cleat was moved 20 mm anteriorly and 20 mm posteriorly from the factory neutral position, respectively. The condition order was quasi-counterbalanced across cleat position (6 order combinations) and rider position (2 order combinations) for all participants to minimize an order bias. For each cleat position, participants cycled at all power outputs and rider positions to reduce the length of the data collection session. The power output order was randomized within a cleat position and kept consistent within an individual. Data for each trial was collected for 10 seconds (15 crank revolutions) once the participant was able to consistently cycle at the required cadence of 90 RPM at the desired power output. Cadence was visually inspected by both the participant and experimenter. If a participant had a cadence that was too high or too low, the trial was repeated. Motion capture data was collected at 250 Hz using an 8-camera system (Motion Analysis, Santa Rosa, CA, USA), and a total of twenty-one markers were placed on the pelvis and right lower limb (Fletcher et al., 2019). Pressure insoles (Pedar; Novel, Munich, DE) were placed in both shoes; however, data was only collected from the right insole to allow for a higher sampling rate of 200 Hz versus 100 Hz if data was collected from both insoles.

Plantar pressure data was used to calculate centre of pressure and pedal reaction force, which were subsequently combined with the three-dimensional marker data for the right leg only. The pedal reaction force from the plantar pressure data was transformed from the pressure sensor's coordinate system to the global coordinate system defined by the motion capture system using the orientation of the foot from the markerset. Given the relatively planar nature of the cycling motion and the use of rigid cycling shoes, we used a modification of the gait10dof23musc model in OpenSim (Delp et al., 2007). Although the number of muscles is reduced in this model (23 total), it was deemed suitable for the purpose of this study and also reduced the computation time for the dynamic optimization procedure. Model scaling was performed with OpenSim's marker-based scaling procedure. Using the scaled model and the three-dimensional marker data, OpenSim's model-constrained inverse kinematics routine was performed to determine joint angles during the cycling task. OpenSim Moco was then utilized to solve the muscle redundancy problem with direct collocation and the MocoInverse tool to provide muscle-driven simulations of cycling that matched the kinematics and reaction forces from the pedals. The direction collocation procedure incorporated both muscle activation dynamics and muscle-tendon dynamics with compliant tendons. The muscles in the model were replaced with the DeGrooteFregly2016 muscle model to reduce simulation time and to help with convergence of the optimization procedure. We computed the average angle data and average pedal data across the 15 pedal strokes and performed the optimization routine on one pedal revolution from top dead centre (i.e., bicycle crank positioned vertically upwards). The optimization routine minimized a cost function with two terms. One term minimized the sum of muscle excitations, which are meant to represent the input from the motor neurons to the muscle fibers. The second term was to minimize the reliance on reserve actuators at each joint. The cost function was as follows:

$$J = (\sum_{m=1}^{m=n} \int_{t_0}^{t_f} e_n^2(t)dt + \sum_{j=1}^{j=r} \int_{t_0}^{t_f} u_j^2(t)dt) \qquad (1)$$

where *n* is the number of muscles, t_0 is the initial time, t_f is the final time, *e* is the excitations to the muscles, *r* is the number of reserve actuators, *u* is the actuator controls. For the rotational degrees of freedom, these reserve actuators were bound to not produce a torque greater than 5 Nm. The reserve actuators at the pelvis were essential for mimicking the reaction forces from the bike saddle. The outputs from the direct collocation routine provided the states and controls, with the controls being the excitation to muscles and inputs to the reserve actuators. As a result, we extracted excitations, forces, lengths, and velocities for the muscles, lengths and velocities for the muscle-tendon units, and forces, lengths, and velocities for the tendons.

Next, we used the Achilles and patellar tendon forces from the direct collocation procedure, and tendon strains were estimated using previously-published tendon resting lengths (Achilles tendon: 180.6 mm, patellar tendon: 49.6 mm) and stiffnesses (Achilles tendon: 397 N/mm, patellar tendon: 3641 N/mm) (Pearson et al., 2007; Rosso et al., 2012; Werkhausen et al., 2018; Yoo et al., 2007). Achilles and patellar tendon strains were normalized to and subsequently averaged across all recorded pedal strokes using the following equation:

$$\varepsilon = \frac{F}{k * l_0} \qquad (2)$$

where $\boldsymbol{\varepsilon}$ is tendon strain, *F* is tendon force, *k* equals tendon stiffness and l_0 equals initial tendon length. Peak strains were extracted from this analysis and used for statistical analysis. Using outputs from the direct collocation procedure along with a number of model-based estimates (e.g., participant mass, muscle mass, fiber-type distribution), we estimated the metabolic power associated with cycling in each condition using modifications of the UmbergerMetabolicMuscle model and similar to previous work (Asmussen et al., 2019; Uchida et al., 2016; Umberger, 2010). The model provided estimates of energy cost of muscle contraction for each muscle in our model.

Statistical Analysis

The effect of cleat position, power output, and rider position on (1) Achilles tendon strain, (2) patellar tendon strain, and (3) modelled metabolic output were analyzed with three 3 (cleat position: anterior, neutral, posterior) × 4 (power output: 150W, 200W, 250W, 300W) × 2 (rider position: seated, standing) repeated-measures ANOVAs with an alpha criterion of $\alpha = 0.05$. If the assumption of sphericity was violated, a Greenhouse-Geisser correction was applied and adjusted degrees of freedom were reported. Post hoc analyses were performed on any significant interaction or main effects and a Bonferroni confidence interval adjustment was used to correct for multiple comparisons. Eta-squared was used to report effect sizes and values greater than $\eta^2 = 0.01$, 0.06, and 0.14 denoted small, medium, and large effect sizes, respectively.

Results

Achilles Tendon Strain

A 3-way interaction was not observed for Achilles tendon strain ($F_{2.6,18.4} = 0.509$, $p = 0.659$, $\eta^2 = 0.0003$). A significant 2-way interaction was observed between cleat position and rider position ($F_{2,14} = 10.815$, $p = 0.001$, $\eta^2 = 0.0037$). That is, a greater difference in Achilles tendon strain was observed between a standing versus seated posture with an anterior cleat position (mean difference = 3.5% strain, $p < 0.001$) compared to a posterior cleat position (mean difference = 2.8% strain, $p < 0.001$) (see Figure 1, Table 1).

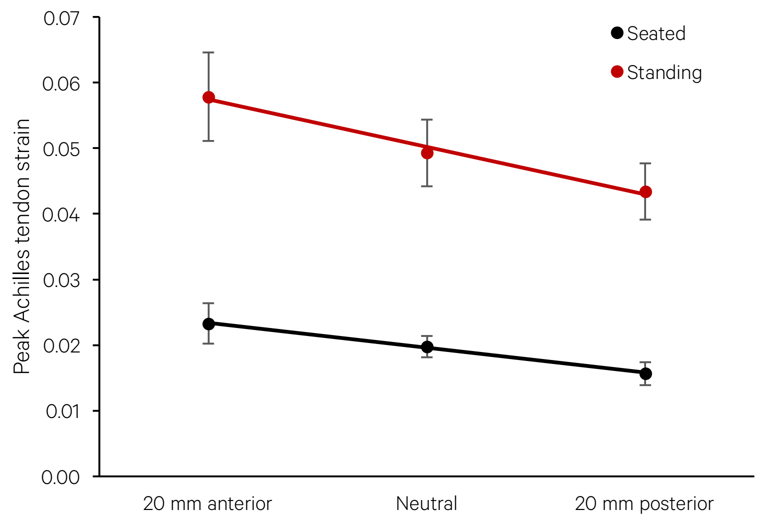

Figure 1: Peak Achilles tendon strain (estimated marginal mean for all power output conditions) during seated and standing cycling at each cleat position. Both rider positions illustrate a decrease in strain as cleat position is moved posteriorly; however, this effect is more pronounced when in a standing posture as indicated by the more negative slope.

Since significant, congruent changes in Achilles tendon strain were observed during both seated and standing cycling, combined with a small effect size for the interaction effects, the main effects of the 3-way ANOVA were still examined. Peak Achilles tendon strain was significantly affected by cleat position approaching a medium effect size ($F_{2,14} = 32.29$, $p < 0.01$, $\eta^2 = 0.0578$), rider position with a large effect size ($F_{1,7} = 253.792$, $p < 0.001$, $\eta^2 = 0.7771$; see Table 1, Figure 2), and power output with a small effect size ($F_{3,21} = 72.690$, $p < 0.001$, $\eta^2 = 0.0452$; see Figure 3). In terms of cleat position, post hoc tests revealed that peak Achilles tendon strain was significantly reduced during seated cycling in the posterior position versus both the anterior position ($p < 0.001$) and the neutral position ($p = 0.047$), but, although approaching significance, strains were not reduced in the neutral versus anterior position ($p = 0.056$). In a standing position, peak Achilles tendon strain was significantly reduced in the posterior position versus both the anterior position ($p < 0.001$) and the neutral position ($p = 0.019$), while the neutral position also significantly reduced strains relative to the anterior position ($p = 0.005$). The main effect for rider position also showed that Achilles tendon strain was greater when cycling in a standing compared to a seated position across all power outputs and cleat positions ($p < 0.001$).

Table 1: Estimated marginal means of peak Achilles tendon strains at different cleat positions and power outputs during seated and standing cycling.

Seated

Cleat Position	Power Output 150W	200W	250W	300W	Estimated Marginal Mean
20mm Anterior	1.8 (0.4)	2.2 (0.3)	2.6 (0.4)	3.0 (0.4)	2.4 (0.6)
Neutral	1.5 (0.4)	1.8 (0.2)	2.1 (0.2)	2.6 (0.3)	2.0 (0.6)
20mm Posterior	1.2 (0.2)	1.5 (0.3)	1.7 (0.3)	2.1 (0.4)	1.6 (0.6)[d,e]
Estimated Marginal Mean	1.5 (0.4)	1.8 (0.4)[a]	2.1 (0.4)[a,b]	2.5 (0.4)[a,b,c]	

Standing

Cleat Position	Power Output 150W	200W	250W	300W	Estimated Marginal Mean
20mm Anterior	5.0 (1.2)	5.6 (1.0)	5.8 (0.9)	6.3 (1.0)	5.7 (1.8)
Neutral	4.5 (0.9)	4.8 (0.8)	5.2 (0.7)	5.5 (0.8)	5.0 (1.2)[d]
20mm Posterior	4.0 (0.6)	4.1 (0.8)	4.7 (0.6)	4.8 (0.8)	4.4 (1.2)[d,e]
Estimated Marginal Mean	4.5 (1.0)	4.8 (1.6)	5.2 (1.0)[a]	5.5 (1.6)[a,b,c]	

Significantly different ($p \leq 0.05$) from: [a] = 150W; [b] = 200W; [c] = 250W; [d] = 20mm anterior; [e] = neutral

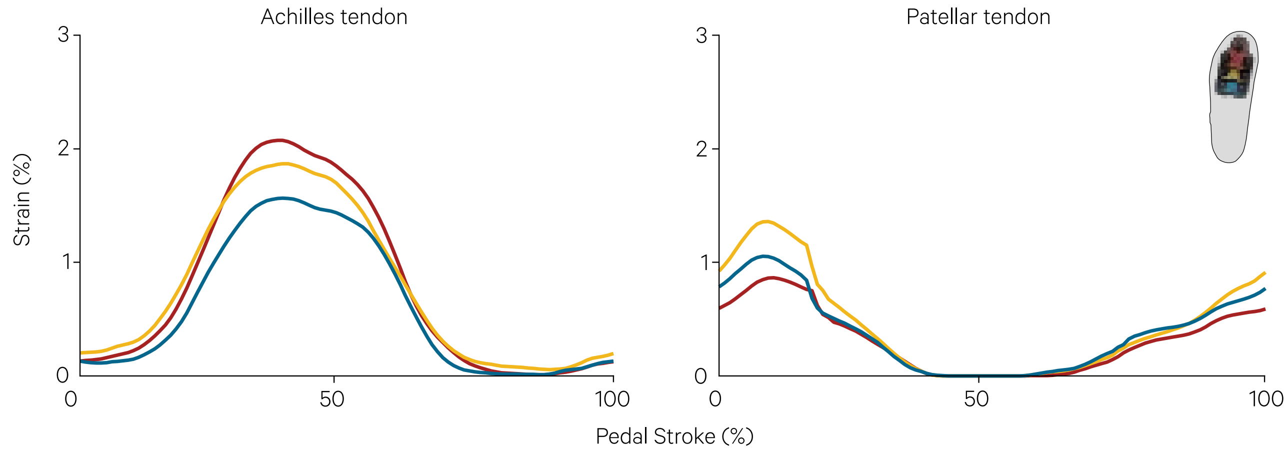

Figure 2: Stroke-averaged Achilles and patellar tendon strains for a representative participant cycling at 200W power output in a seated posture. Data has been normalized to a single pedal stroke (0% = top dead centre, 50% = bottom dead centre). Blue line = posterior cleat position, yellow line = neutral cleat position, red line = anterior cleat position. Significant effects of cleat position were observed for peak Achilles tendon strain only.

Patellar Tendon Strain

A 3-way interaction was not observed for peak patellar tendon strain ($F_{1.2,8.2}$ = 1.756, p = 0.225, η^2 = 0.0018). A significant interaction was observed between rider position and power output ($F_{1.2,8.7}$ = 6.678, p = 0.026, η^2 = 0.0011), as peak patellar tendon strain illustrated a greater decrease with power output when seated compared to standing. Once again, there was less than a small effect size for these interaction effects and therefore, we analyzed the main effects from the 3-way ANOVA. Peak patellar tendon strain was not significantly affected by cleat position ($F_{1.2,8.5}$ = 1.022, p = 0.358, η^2 = 0.0078; see Figure 2, Table 2) or power output ($F_{1.1,7.5}$ = 0.469, p = 0.527, η^2 = 0.0009); however, a significant main effect of rider position was observed, with a small effect size ($F_{1,7}$ = 6.208, p = 0.042, η^2 = 0.0437) such that patellar tendon strain was higher when standing versus seated cycling.

Table 2: Estimated marginal means of peak patellar tendon strains at different cleat positions and power outputs during seated and standing cycling.

Seated

Cleat Position	Power Output 150W	200W	250W	300W	Estimated Marginal Mean
20mm Anterior	1.0 (0.7)	1.0 (0.6)	1.0 (0.7)	1.1 (0.7)	1.0 (1.1)
Neutral	1.3 (1.6)	1.2 (1.0)	1.3 (1.3)	1.4 (1.4)	1.3 (2.8)
20mm Posterior	1.1 (0.8)	1.1 (0.7)	1.2 (0.8)	1.4 (1.2)	1.2 (1.7)
Estimated Marginal Mean	1.1 (2.0)	1.1 (1.5)	1.2 (1.5)	1.3 (2.0)	

Standing

Cleat Position	Power Output 150W	200W	250W	300W	Estimated Marginal Mean
20mm Anterior	1.6 (1.0)	1.5 (1.0)	1.5 (1.2)	1.4 (1.1)	1.5 (2.4)
Neutral	1.8 (1.7)	1.7 (1.4)	1.7 (1.5)	2.0 (2.2)	1.8 (3.6)
20mm Posterior	1.8 (1.3)	1.7 (1.3)	1.7 (1.3)	1.6 (1.3)	1.7 (2.4)
Estimated Marginal Mean	1.8 (2.1)	1.7 (2.1)	1.6 (2.1)	1.7 (2.6)	

Significantly different ($p \leq 0.05$) from: [a] = 150W; [b] = 200W; [c] = 250W; [d] = 20mm anterior; [e] = neutral

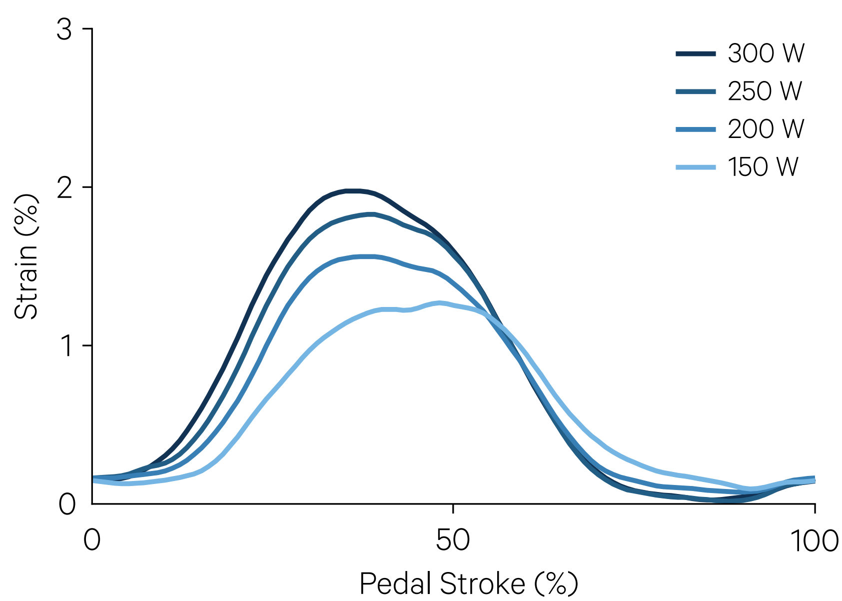

Figure 3: Representative stroke-averaged Achilles tendon strains for all power outputs during seated cycling with a neutral cleat position. Significant effects of power output on peak Achilles tendon strain were observed ($p < 0.001$).

Modelled Metabolic Output

A 3-way interaction was not observed between rider position, power output, and cleat position for modelled metabolic output ($F_{6,42}$ =1.544, p = 0.188, η^2 = 0.0071). However, a 2-way interaction was observed between rider position and power output with a small effect size ($F_{3,21}$ = 5.464, p = 0.006, η^2 = 0.0162). Inspection of this interaction revealed that within the seated condition, modelled metabolic output significantly increased with increasing power output (p = 0.024) until 250W, at which point it did not significantly increase between 250-300W (p = 0.637). However, within the standing cycling condition, there was a more muted effect as modelled metabolic output did not increase significantly between 150-200W (p > 0.999) and between 250-300W (p > 0.999) but was significantly increased between 150W vs 250W (p < 0.001), 150W vs 300W (p = 0.006) and between 200W vs 250W (p < 0.001). Main effects of rider position ($F_{1,7}$ = 11.258 p = 0.012, η^2 = 0.0629) and power output ($F_{3,21}$ = 84.292, p < 0.001, η^2 = 0.4151) were observed on modelled metabolic output, but a main effect of cleat position was not observed ($F_{2,14}$ = 0.256, p = 0.778, η^2 = 0.0011; see Figure 4). As expected, there was both a significant increase in metabolic power and a large effect size due to an increase in mechanical power output requirement, regardless of cycling posture, and metabolic power increased when cycling in a standing posture versus a seated posture with a medium effect size.

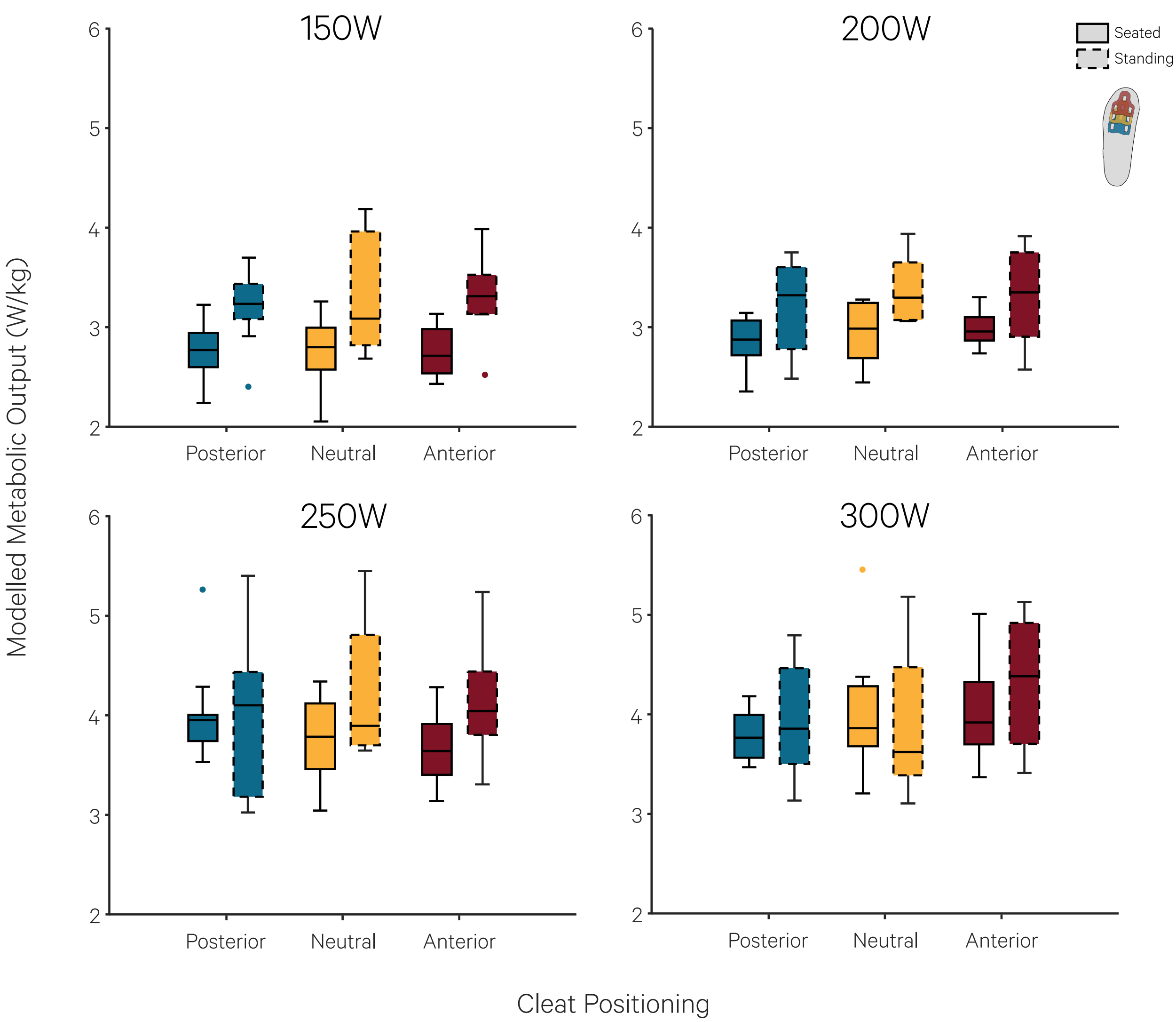

Figure 4: Body mass normalized modelled metabolic power during seated (solid lines) and standing (dashed lines) cycling for each power output. For each graph, blue = posterior cleat position, yellow = neutral cleat position, red = anterior cleat position. The line inside each box represents the median, while the top and bottom edges represent the upper and lower quartiles, respectively. Whiskers extend to the maximum and minimum non-outliers, while any dots represent outlier values. Metabolic power significantly increased when standing compared to seated ($p = 0.012$) and with power output for both seated and standing cycling ($p < 0.001$), but no effect of cleat position was observed ($p = 0.778$).

Discussion

This study represents the first of its kind to systematically examine the effect of cleat positioning on lower-limb tendon strains when cycling at different mechanical power outputs and rider positions. We illustrated that moving the cleat position posteriorly significantly decreased peak Achilles tendon strain, yet no changes in strain were observed at the patellar tendon. Furthermore, we observed no significant differences in modelled metabolic output from cleat position changes. As such, cycling with a 20 mm posterior cleat position may be beneficial for reducing Achilles tendinopathy risk without sacrificing performance or increasing strain at the patellar tendon.

Our first hypothesis was that a posterior cleat position would reduce Achilles tendon strain and our fourth hypothesis was that standing cycling posture would increase Achilles tendon strain. In line with these hypotheses, we observed a significant interaction for peak Achilles tendon strain between cleat positioning and rider position. That is, peak Achilles tendon strain decreased by 0.8% as the cleat was moved posteriorly during standing cycling compared to a decrease of 1.3% strain during seated cycling (see Figure 1). While the Achilles tendon must withstand a greater weightbearing load during standing cycling with the same amount of mechanical power output requirements to move the flywheel, this alone would not alter the relationship between strain and cleat position. During standing cycling, the rider has more degrees of freedom to move their torso and pelvis as they are not restricted to sitting on the saddle. We thus performed a post hoc kinematic analysis to examine potential kinematic changes arising from different rider and cleat positions, discovering that, compared to seated cycling, participants cycled with more anterior translation of the pelvis (standing vs. seated difference = 22.3 cm; $F_{1,8} = 578.628$, $p < 0.001$) and a greater sagittal knee range of motion (standing vs. seated difference = 8.6°; $F_{1,7} = 62.367$, $p < 0.001$). Further, we discovered that the timepoint of peak Achilles tendon strain occurred later in the pedal stroke during standing cycling (154° from top dead centre when seated vs. 191° from top dead centre when standing, $F_{1,7} = 41.089$, $p < 0.001$), allowing participants to adopt a more plantarflexed foot at the timepoint of peak Achilles tendon strain when standing, thereby reducing the external moment arm between the point of force application and the Achilles tendon (standing vs. seated difference = 10.2° more plantarflexed; $F_{1,7} = 49.168$, $p < 0.001$). Similarly, we also discovered that shifting the cleat posteriorly delayed the timing of peak Achilles tendon strain in both seated and standing cycling (169° vs 177° from top dead centre, anterior vs. posterior cleat positioning; $F_{2,14}$ = 4.315, p = 0.035). We speculate that this effect would act to reduce Achilles tendon strain as, from previous literature, the soleus and gastrocnemius muscles illustrate a lower activation as the crank angle approaches 180° (Ericson et al., 1985; Ryan and Gregor, 1992) since the movement of the pedal is predominantly in a posterior direction. As our results showed higher Achilles tendon strains during standing cycling, we therefore postulate that the aforementioned kinematic and temporal changes observed during standing cycling may allow the rider to find a more optimal position that reduces the demands on the ankle plantarflexors and the Achilles tendon. Overall, this finding illustrates that posterior cleat positioning may be even more beneficial in terms of Achilles tendon overuse injury risk reduction in situations necessitating standing, such as during sustained, lengthy climbs requiring high power outputs.

In agreement with our first hypothesis solely, we discovered a main effect of cleat position on peak Achilles tendon strains. The mechanism for this observed decrease in peak Achilles tendon strain as the cleat was moved posteriorly is primarily due to a decreased moment arm between the ankle joint and the point of force application (i.e., the contact point between the pedal and the shoe). We would therefore expect to see further reductions in Achilles tendon strain if the cleat position were moved further posteriorly. However, prior research has shown that different muscular activation strategies may be employed with posterior cleat positioning (Millour et al., 2020), potentially limiting the amount of Achilles tendon strain reduction and/or leading to the key "tipping point" of increased metabolic output. Future studies investigating the biomechanical and physiological effects of larger changes to cleat positioning are therefore warranted.

We can further frame these results related to the first hypothesis in context of overuse injury (i.e., Achilles tendinopathy) by using mechanical fatigue principles (Edwards, 2018) and integrating our peak Achilles tendon strains with previously published Achilles tendon fatigue data (Wren et al., 2003). This process accounts for the highly non-linear relationship observed during cycling loading of Achilles tendon between peak strain and the number of repetitive loading cycles until failure (i.e., fatigue life). We found that cycling with a 20 mm posterior cleat position resulted in peak Achilles tendon strain reductions of 14.5% (compared to neutral cleat position) and 27.2% (compared to 20 mm anterior cleat positions), representing respective 330% and 1800% increases to fatigue life. While this analysis does not account for the biological processes of repair and adaptation known to occur in tendon (Magnusson et al., 2008; Stauber et al., 2020), it remains useful to quantify the relative risk reduction in a more biologically-relevant scenario of overuse injury. As such, a 20 mm posterior shift in cleat position represents an exponential reduction in Achilles tendon overuse injury risk, which may allow cyclists to remain injury-free with increasing cycling volume or intensity.

Peak patellar tendon strains were not significantly affected by cleat positioning or power output, failing to support our second hypothesis. The main parameter in our patellar tendon strain estimation is the flexion/extension knee moment, which was also unaffected by shifts in cleat positioning. This lack of observed change in the patellar tendon was also observed in previous research in which six participants cycled at different saddle heights, pedalling rates, and power outputs with an anterior and a posterior foot position (separated by approximately 10 cm) (Ericson and Nisell, 1987). Results from this study indicated that patellar tendon forces, as measured by the flexion/extension moment, increased significantly with decreased saddle height and increased workload, but were unaffected by pedalling rate and foot position. When combined, the results of our study and prior research indicate that any changes to the cleat/foot position are not large enough to necessitate a different neuromuscular or biomechanical strategy to be adopted at the knee; however, alternative methods such as bicycle fit/rider position and the interaction with power output may be beneficial in terms of altering patellar tendon strain.

Given that our tendon strains were estimated largely through modelling procedures, it is pertinent to assess our results against those obtained using other methodologies. Prior work from Gregor and colleagues (Gregor et al., 1987) used a buckle transducer to directly measure Achilles tendon force during cycling at 265 W and 90 RPM in a single subject, illustrating a similar pattern of Achilles tendon force that increased from top dead centre until approximately 30% of the pedal stroke, then decreasing until nearly zero force around 70% of the pedal stroke (Gregor et al., 1987). This behaviour is similar to our normalized time-series results for seated cycling at 250 W and 90 RPM (see Figure 2), which illustrate an increase in strain from top dead centre until peak strain occurring on average at 41% of the pedal stroke. Comparing peak force values, we observed an average peak force of 0.59 bodyweights (430 N) while Gregor et al.(1987) illustrated a peak force of 0.83 bodyweights (661 N) (Gregor et al., 1987). This difference in peak force likely stems from our estimate for the Achilles tendon moment arm or different co-contraction strategies of non-triceps surae muscle not examined in this study; however, this parameter should not alter the main findings of our within-subject design experiment.

Similar to prior research and against our third hypothesis, we discovered that anterior/posterior cleat positioning did not significantly influence modelled metabolic output. It could be postulated that the metabolic modelling procedure we utilized lacked enough precision to identify subtle differences between the various conditions. Although plausible, we believe this is not the case. We are given confidence in our results for several reasons. Firstly, we observed significant increases in modelled metabolic output when standing and with increased power output, both of which were expected and in line with previous research (Berkemeier et al., 2020; Seabury et al., 1977). Secondly, we selected a modelling approach as it directly accounted for participant kinematics, and associated muscle excitations, within the objective function (Umberger, 2010); such parameters are not directly accounted for in other frequently-used methodologies such as indirect calorimetry through expired gas analysis. We hypothesized that kinematic and kinetic changes may occur with cleat position alterations, potentially leading to changes in modelled metabolic output. We did in fact observe significant effects of cleat position on sagittal plane knee range of motion ($F_{2,14} = 7.130$, $p = 0.007$); however, these effects did not elicit changes to modelled metabolic power. This result is consistent with previous literature that has reported limited to no changes to the rate of oxygen consumption, lactate, or heart rate when cycling with a traditional (i.e., under the metatarsal heads) cleat position versus a 5-50 mm posterior cleat position (Millour et al., 2020; Paton, 2009; Van Sickle and Hull, 2007; Viker and Richardson, 2013) even though a posterior position significantly reduced ankle plantarflexor muscle recruitment (Mcdaniel et al., 2013; van Sickle, 2003). One reason for the lack of metabolic differences in these previous studies at the whole body level might be the concomitant reductions in plantarflexor muscle activity with increases in tibialis anterior and rectus femoris muscle activation (Chartogne et al., 2016). As such, this finding indicates that posterior shifts to cleat positioning may represent a mechanism with which to reduce Achilles tendon overuse injury risk without sacrificing performance.

This work has several limitations to be addressed. Firstly, we utilized a pressure-sensing insole to obtain estimates of pedal reaction force. As this insole cannot provide shear forces, we created a representation of 3D pedal reaction force using the pedal angle throughout the pedal stroke. We did not observe significant changes to pedal angle between conditions, giving us confidence in this methodological approach for our within-subject design. The next limitation is that participants cycled at each condition for approximately 60 seconds, which potentially limited their ability to fully adapt. We would expect, however, that increasing the adaptation period for each condition would only act to amplify our results. Finally, we used a modelling procedure to estimate metabolic output at each condition and did not use inspired/expired gas analysis as for the higher power output, our participants would be cycling above anaerobic threshold making it difficult to assess measured metabolics. As previously discussed, we have confidence in our modelled metabolic cost results as we observed expected 1) significant increases in energetic cost when cycling in a standing position, 2) significant increases in energetic cost with increased power output, and 3) no changes in energetic cost based on cleat position, all finding that are consistent with previous research.

Conclusion

We discovered that regardless of power output and rider position, cycling with a 20 mm posterior cleat position significantly reduced Achilles tendon strain, without altering the modelled metabolic output or patellar tendon strain. As such, cycling with a posterior cleat position represents an easy, viable intervention to reduce the risk of Achilles tendinopathy without sacrificing cycling performance.

Contributions

Contributed to conception and design: CRF, MJA
Contributed to acquisition of data: CRF, MJA
Contributed to analysis and interpretation of data: CRF, MJA
Drafted and/or revised the article: CRF, MJA
Approved the submitted version for publication: CRF, MJA

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Data and Supplementary Material Accessibility

Data used within this manuscript is available upon request.

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