

# Maximum dorsiflexion increases Achilles tendon force during exercise for midportion Achilles tendinopathy

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Rehabilitation is an important treatment for non-insertional Achilles tendinopathy. To date, eccentric loading exercises (ECC) have been the predominant choice; however, mechanical evidence underlying their use remains unclear. Other protocols, such as heavy slow resistance loading (HSR), have shown comparable outcomes, but with less training time. This study aims to identify the effect of external loading and other variables that influence Achilles tendon (AT) force in ECC and HSR. Ground reaction force and kinematic data during ECC and HSR were collected from 18 healthy participants for four loading conditions. The moment arms of the AT were estimated from MRIs of each participant. AT force then was calculated using the ankle torque obtained from inverse dynamics. In the eccentric phase, the AT force was not larger than in the concentric phase in both ECC and HSR. Under the same external load, the force through the AT was larger in ECC with the knee bent than in HSR with the knee straight due to increased dorsiflexion angle of the ankle. Multivariate regression analysis showed that external load and maximum dorsiflexion angle were significant predictors of peak AT force in both standing and seated positions. Therefore, to increase the effectiveness of loading the AT, exercises should apply adequate external load and reach maximum dorsiflexion during the movement. Peak dorsiflexion angle affected the AT force in a standing position at twice the rate of a seated position, suggesting standing could prove more effective for the same external loading and peak dorsiflexion angle.

## KEYWORDS

Achilles tendon, midportion Achilles tendinopathy, rehabilitation, tendon moment arm

## 1 | INTRODUCTION

Midportion Achilles tendinopathy is a debilitating condition affecting both professional athletes and recreationally active people; it is most common in those participating in running sports.<sup>1,2</sup> It has been reported that middle to long-distance runners have a lifetime risk of 50%,<sup>3</sup> and the annual incidence for

high-level club runners is 7–10%.<sup>4</sup> Rehabilitation exercises are the first line of treatment for midportion Achilles tendinopathy.<sup>5</sup>

Clinically, controlled stretching and strengthening exercises have been considered as beneficial in tendon management and eccentric loading exercises (ECC) have become the mainstream choice during rehabilitation.<sup>6</sup> Many systematic reviews and randomized controlled trials confirmed the

effectiveness of ECC for patients with midportion Achilles tendinopathy.<sup>7-9</sup> The exercise emphasizes performing eccentric heel drop with tolerable pain during the movement. The original rationale for ECC was that the tendon would be loaded more in the eccentric phase than the concentric phase due to the stretching of tendon and muscle.<sup>10,11</sup> Eccentric exercise resulted in a larger reduction in Achilles tendon (AT) volume, an indicator of a reduction in tendinopathy, than concentric exercise in healthy volunteers.<sup>12</sup> Although a randomized controlled study showed better outcomes for isolated eccentric than for isolated concentric rehabilitation,<sup>13</sup> the magnitude of loading was not controlled in the trial. A more recent systematic review concluded that there is no convincing mechanical evidence to support the eccentric component in preference to concentric loading.<sup>14</sup> Biomechanical studies addressing the hypothesis of greater loading in the eccentric phase used ultrasound<sup>15</sup> and/or motion capture<sup>16,17</sup> to estimate the AT moment arm and to quantify loading of the AT during eccentric and concentric motions. These biomechanical studies have shown that there is no significant difference in the force-displacement behavior of the tendon between the eccentric and concentric phases of motion during ECC.<sup>15</sup>

Other rehabilitation protocols including both eccentric and concentric components have been proposed and comparable outcomes to traditional ECC were observed.<sup>18-21</sup> Specifically, heavy slow resistance training (HSR)<sup>21</sup> was introduced based on a similar protocol for patellar tendinopathy.<sup>22</sup> One randomized control trial comparing ECC and HSR showed no difference in the clinical outcome, and HSR had the added advantage of taking one third of the time of that for the ECC regimen. Although HSR was more time efficient, it does require gym equipment, not needed for ECC, to perform the activity in three different postures. ECC emphasizes heel drop motions with the knee bent and/or the knee straight.<sup>13</sup> Participants are instructed to apply additional loading using a backpack and to avoid active heel rise during the activity. By contrast, HSR is performed with both heel drop and heel rise motions, in a standing or seated position, with the loading applied via equipment including a weighted bar, a seated calf-raise machine, and a seated leg press machine.<sup>21</sup>

Optical motion capture systems utilizing ground reaction force measurement have been widely used in kinetic analysis and biomechanical studies *in vivo*.<sup>16,17,23</sup> Models with AT moment arms measured from ultrasound, skin markers, or scaling pre-defined musculoskeletal geometry<sup>16</sup> have been used to estimate AT force. However, previous studies focused only on standing ECC exercises and did not include the tendon force in different postures and loading conditions.

The aim of this study was to quantify the loading of the AT during ECC and HSR exercises and investigate the variables that can increase the AT loading during these exercises. This information could be used to estimate the tendon loading during rehabilitation and provide biomechanical evidence to improve the efficiency of the protocol. Specifically, it aimed to

assess how AT force increases/decreases with external loading conditions and postures in these rehabilitation exercises. We hypothesized that the AT force would be different in the two types of exercises under the same loading conditions.

## 2 | METHODS

### 2.1 | Study design

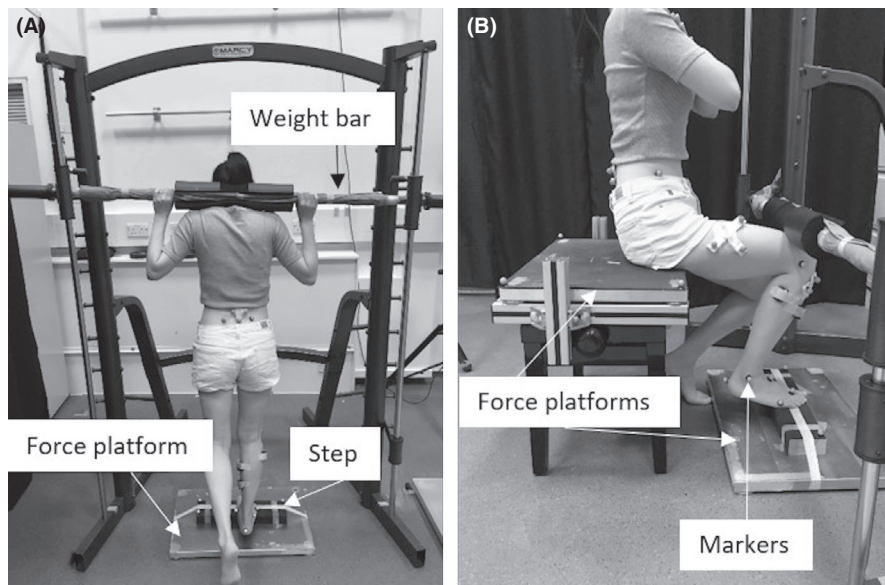
Healthy participants were asked to perform two types of rehabilitation exercises, ECC and HSR, under multiple controlled loads (loading magnitude was scaled to the weight of each participant). ECC was performed in a standing posture and HSR was performed in both standing and seated postures. The 3D kinematics were recorded, and a musculoskeletal model was used to estimate the ankle moment.<sup>24</sup> To obtain subject-specific moment arms, unloaded static MRI scans were performed for each participant, with their ankle fixed at six angles across the range of motion.

### 2.2 | Participants

Eighteen healthy volunteers (11 males, 7 females, age:  $29.6 \pm 3.8$  years, weight:  $70.7 \pm 12.4$  kg, height:  $171.8 \pm 7.5$  cm; mean  $\pm$  standard deviation) were recruited from the university environment. Participants with recent lower limb musculoskeletal injury or any other physical condition that could prevent them from performing the exercises described in the ECC and HSR program were excluded. Written informed consent of the study was collected after explaining the details of the experiment. This study was approved by the Imperial College Research Ethics Committee (18IC4371).

### 2.3 | Experimental protocol

To compare the loading conditions, ECC and HSR were modified slightly from the methods presented in the original studies. To control the loading method and trajectory, a commercially available Smith machine (Marcy) was used to apply additional weights to the participants. The bar of the machine was padded to ensure comfortable contact. To reduce overlap between the protocols and decrease the number of activities performed under high loading, the knee-straight ECC (active heel drop) was taken as the eccentric part of HSR. The seated leg press in HSR was not performed with the knee extended, as in the standing position; therefore, only two postures in HSR were tested (Figure 1). During the activity, different loads, scaled to the body weight (BW) of the participants, were applied. Four loading conditions (50% BW, 25% BW, bar weight, zero) were applied for each exercise, and each combination



**FIGURE 1** Experimental configurations for standing and seated positions. (A) Standing heel rise and (B) seated calf-raises were performed with a Smith machine to apply additional load

No.	Loading exercise	Posture	Loading condition as percent of body weight (% BW)
1	HSR	Standing knee-straight heel drop and rise	125
2	HSR	Standing knee-straight heel drop and rise	150
3	HSR & ECC	Standing knee-straight heel drop and rise	108–115
4	HSR	Standing knee-straight heel drop and rise	100
5	ECC	Standing knee bent heel drop and rise	108–115
6	HSR	Seated heel drop and rise	38
7	HSR	Seated heel drop and rise	63
8	HSR	Seated heel drop and rise	21–28
9	HSR	Seated heel drop and rise	13

**TABLE 1** Posture and loading combinations for eccentric loading (ECC) exercises and heavy slow resistance (HSR) exercises

*Note:* Loading conditions are expressed as the total load at the ankle. Three repetitions of each activity were performed. In the seated position, estimated weights of 6% BW for the shank and foot and 7% BW for half of the thigh<sup>26</sup> were employed to estimate the loading at the ankle.

was performed three times. The different testing combinations of posture and loading are summarized in Table 1.

Participants were instructed to perform the activities with the right leg only. To control the speed of the motion, both the eccentric and concentric phases were timed to last three seconds. Participants were informed verbally of progression through and completion of each phase. For HSR, multiple loading conditions were tested in both standing and seated positions; these were the application of 0% BW, the bar weight (8%–15% BW), 25% BW, and 50% BW. ECC was performed both with the knee bent and the knee straight while holding the bar on the Smith machine (Figure 1). ECC with the knee bent was tested with only one loading condition to avoid potential injury, because pilot work showed some participants were unable to perform the activity with any more additional weight in the knee bent position. It is worth noting that this means that, because the added load was the same for standing

and seated postures, the ankle sustained a higher load in the standing posture (BW plus the added weight) than in the seated posture (leg weight plus the added weight). All the participants followed the same test protocol and were able to complete the experiment.

In the standing position, to standardize the bar position between participants, the bar was held on the shoulders with two hands, aligned vertically with respect to the posterior edge of the right heel. Participants were instructed to maintain the knee angle, either straight or bent, during the motion and to use only the ankle to perform heel rise and drop. In the seated position, the knee was in a flexed position with the bar resting on the thigh. Participants were asked to align the bar with their big toe. Participants also were instructed to place only their forefoot on the step to allow full range of ankle motion during eccentric and concentric motion phases.

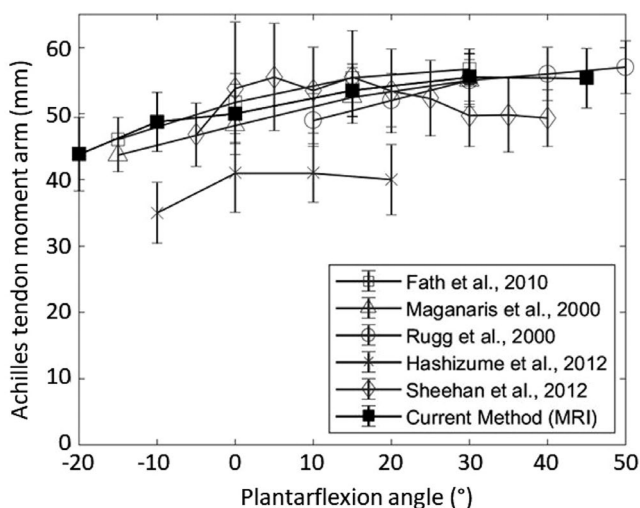
## 2.4 | Kinematic and kinetic data collection

Twenty-three reflective markers were attached to the skin of the participants during the trials to measure the right lower limb kinematics.<sup>24</sup> The locations of the markers are summarized in the Supplementary materials (Figure S1). A ten-camera optical motion capture system (Vicon Motion Systems) was used to record the positions of the markers.

Two force platforms were used to measure the external reaction forces. A step (length: 91 mm, width: 410 mm, height: 57 mm) was placed at the center of the anterior force platform. The vertical coordinate of the center of pressure in this coordinate system was set to the height of the wooden step.<sup>16</sup> This step enabled the ankle to reach its full range of motion. Participants were asked to perform heel rise and drop with the forefoot on the step to mimic the motion performed during the original ECC and HSR while the ground reaction force was recorded. The posterior force plate was mounted on an adjustable chair to record the seat reaction force. For each participant, the chair height was adjusted to ensure the thigh was horizontal.

## 2.5 | AT force estimation

Kinematics (Figures S2 and S3) and ground reaction force data were analyzed using inverse dynamics to calculate the ankle joint torque. Inverse dynamics were performed using FreeBody,<sup>24,25</sup> an open-source software package. The estimation of torque also required anthropometric parameters (e.g. the inertia, center of mass, and segmental mass) which were based on findings reported in the literature.<sup>26</sup>



**FIGURE 2** Achilles tendon moment arm measured from MRI scans, shown as black squares, compared with values from the literature

To obtain participant-specific AT moment arms in 3D, six static proton-density-weighted MRI scans for each participant were performed at six joint angles:  $-20^\circ$  and  $-10^\circ$ ,  $0^\circ$ ,  $15^\circ$ ,  $30^\circ$ , and  $45^\circ$  (negative for dorsiflexion), covering the foot and ankle to the distal third of the tibia. Six wedges were used to maintain the ankle joint angle (Figure S4). Only transverse sections were performed. The MRI sequences were characterized by a repetition time of 4220 ms, an echo time of 23 ms, a slice thickness of 1 mm, in-plane resolution of  $0.4 \times 0.4$  mm, field of view of 140 mm, and a 30% distance factor. A commercially available orthotic device was attached to the wedge to ensure foot and wedge alignment. The forefoot was secured by hook and loop fastener onto the wedge. A small four channel flexible body coil (4-channel Flex Coil, Siemens) was used to increase spatial magnetic field intensity around the ankle. After the MRI scans were obtained, the AT, calcaneus and talus bones were segmented using Mimics Innovation Suite (v.19, Materialise). The centerline for the AT was calculated in MATLAB (2017a, MathWorks). As the AT may be curved in plantar flexion, only the centerline of the first 2 cm from insertion was used to represent the line of action of the AT. The center of ankle joint rotation was estimated manually by fitting a sphere to the talar dome (Figure S5)<sup>25,27</sup> The moment arm was calculated based on the centerline of the AT and the center of the sphere. This differed from the work presented by Alexander et al.<sup>27</sup> in which a cylinder was fitted to the talus. While we did not perform an independent validation of this approach, it was derived from the same principle of fitting idealized geometry to anatomical structures. The FreeBody model calculates the ankle moment with six independent degrees of freedom; such a design preserves the anatomical variation and the deviation of rotation axes during the range of motion. To be consistent with the kinematic calculation from FreeBody, we fitted a sphere, instead of a cylinder, to the talus. If ankle rotation was performed beyond the range of motion of the MRI scans, the moment arm of the AT was linearly extrapolated, which is supported by findings in the literature.<sup>28</sup> During dorsiflexion, the forefoot experiences a larger range of motion than the hindfoot, though the movement of the two compartments is proportionate.<sup>28</sup> Finally, the AT force was calculated by dividing the ankle torque by the participant-specific effective moment arm estimated from the MRI.

## 2.6 | Statistical analysis

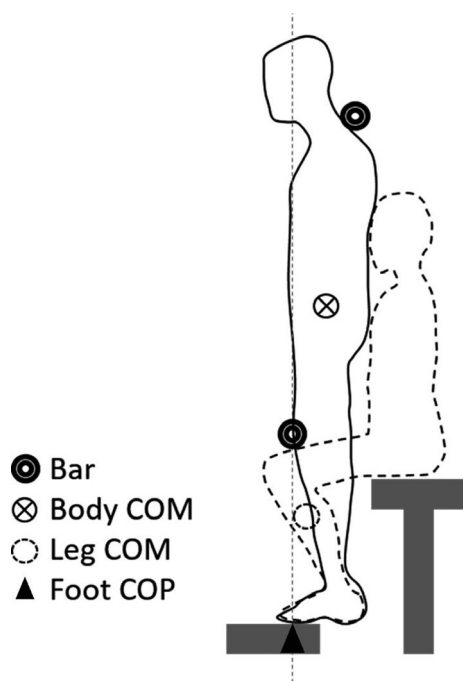
Participant-specific AT moment arms were compared with those reported in the literature using descriptive statistics. Since similar values have been reported in the literature,<sup>29–32</sup> we did not scale the moment arms to height or tibial length. The torque and AT force were normalized

by the participant's BW to enable comparison between participants. The normalized torque and force were compared using Wilcoxon signed rank tests for pair-wise comparison and a significance level of 0.05 was selected. To investigate the effect of loading on the normalized peak AT force, a stepwise multivariate regression analysis was performed with five parameters. The ankle loading, weight (in % BW), height, sex, and peak dorsiflexion angle were analyzed. The rate ratio (beta value), adjusted beta value, *p*-value, and 95% confidence interval were reported. The changes of moment arm, torque, and force during different ankle joint angles were evaluated with repeated measure ANOVA. All statistical analyses were performed using IBM SPSS software (v23.0, IBM Corp.).

### 3 | RESULTS

#### 3.1 | AT moment arm

The moment arms of the AT decreased as the ankle dorsiflexed (dorsiflexion 20°: 43.9 ± 5.6 mm; 10°: 48.8 ± 4.4 mm; 0°: 50.0 ± 4.4 mm; plantar flexion 15°: 53.5 ± 4.0 mm; 30°: 55.6 ± 4.1 mm; 45°: 55.3 ± 4.5 mm; *p* < 0.001). The moment arms estimated from MRI were consistent with moment arms reported in the literature (Figure 2).<sup>29–32</sup>



**FIGURE 3** The centers of mass (COM) during the exercises. The combined center of mass (body COM and bar) in standing is further away from the foot center of pressure (COP) than in the seated position. This could lead to a larger effect of dorsiflexion in the standing position

#### 3.2 | Ankle torque and AT force in ECC and HSR

In both ECC and HSR, the normalized ankle torque and AT force increased as the ankle dorsiflexed (Figure S6; *p* < 0.001). The ankle reached the largest dorsiflexion angle in the ECC with the knee bent. In ECC, although the peak normalized ankle torque with the knee straight was significantly higher than with the knee bent ( $0.174 \pm 0.013$  Nm/kg and  $0.169 \pm 0.013$  Nm/kg, respectively, *p* = 0.015), the peak normalized AT force with the knee bent was higher than for a straight knee (*p* = 0.004; Table 2). In HSR, the ankle was able to move through a larger range of motion in sitting than in standing. For the same weight lifted, there was no difference in the peak AT force between the eccentric and concentric phases in the standing position, whereas in the seated position the AT loading was found to be higher in the concentric phase (*p* = 0.0019, 0.0002, and 0.0003, for 21%–28%BW, 38%, and 63% BW, respectively; Table 2).

The multivariate analysis of normalized peak AT force for the whole cycle showed that external loading, peak dorsiflexion angle, and weight (in standing position only) were significant predictors of peak AT force. The rate ratios for external loading in standing and seated positions were similar, but the rate ratio for peak dorsiflexion angle in the standing position was twice that in the seated position (Table 3).

**TABLE 2** The peak Achilles tendon force predicted by the participant-specific MRI model in eccentric loading (ECC) and heavy slow resistance (HSR) exercise. \**p* < 0.01, \*\**p* < 0.001, Wilcoxon signed rank test

Loading applied: Position (%BW)	Peak Achilles force during ECC (%)	
	Knee straight	Knee bent
Standing (108–115)	3.775 ± 0.573	4.378 ± 1.158*
	Peak Achilles force during HSR (% BW)	
	Concentric	Eccentric
Standing (150)	5.05 ± 0.83	5.12 ± 1.03
Standing (125)	4.32 ± 0.79	4.34 ± 0.91
Standing (108–115) <sup>a</sup>	3.79 ± 0.54	3.78 ± 0.57
Standing (100)	3.56 ± 0.66	3.61 ± 0.71
Seated (63) <sup>b</sup>	2.29 ± 0.64	2.04 ± 0.63**
Seated (38) <sup>c</sup>	1.39 ± 0.42	1.21 ± 0.46**
Seated (21–28) <sup>a</sup>	0.89 ± 0.28	0.75 ± 0.29*
Seated (13)	0.41 ± 0.17	0.43 ± 0.18

<sup>a</sup>The weight of the bar.

<sup>b</sup>50% BW plus 13% BW as an estimation of the leg and half the thigh.

<sup>c</sup>25% BW plus 13% BW as an estimation of the leg and half the thigh.

\**p* < 0.01; \*\**p* < 0.001.

## 4 | DISCUSSION

This study investigated the AT force during two different rehabilitation regimens with ankle torque calculated using inverse dynamics and moment arms estimated from participant-specific MRIs. The key findings include (i) the AT force is not larger in the eccentric phase; (ii) external loading affects the force transmitted through the AT similarly whether seated or standing, but the effect of peak ankle dorsiflexion on this force is increased when standing; and (iii) after adjusting for the weight of the trunk, standing may still be more effective in increasing the AT loading than when seated.

This study estimates the AT moment arms using a series of participant-specific MRIs that could account for the deformation and curving of the tendon during the ankle range of motion.<sup>27,29–32</sup> An effort was made to control the loading factors (e.g. trajectory of the loading, loading conditions, and movement speed) during the motion to investigate the AT loading in different postures.

### 4.1 | AT moment arm in extreme dorsiflexion angles

As the moment arm drops in dorsiflexion, the moment arm at extreme ankle dorsiflexion ( $>20^\circ$ ) is critical for estimating the peak AT force. In the literature, the AT moment arm was measured previously up to only  $15^\circ$  dorsiflexion.<sup>29,30</sup> The MRI measurements in this study reached  $20^\circ$ . During the loading exercises, the ankle was further pushed beyond this, as a larger ankle dorsiflexion angle was observed in the knee bent and seated position. This finding is consistent with the literature and is possibly due to slackening of the gastrocnemius when the knee is bent.<sup>33,34</sup> The increase in dorsiflexion angle decreased the moment arm of AT. Therefore, despite the peak ankle torque being smaller than that in the straight knee condition, the force through the AT was larger.

In this study, the AT moment arm was estimated by extrapolation for extreme angles ( $>20^\circ$  dorsiflexion). The limitation of such extrapolation is acknowledged, but to the authors' knowledge, there is no literature reporting the moment arm in such an extreme range. This estimation was based on the relative forefoot and hindfoot kinematics reported during ankle dorsiflexion,<sup>28</sup> where the forefoot has a larger range of motion than the hindfoot, but the degree of rotation is proportional. More studies are needed to measure the moment arm in such ranges. Applying extra loading in an MRI scanner may be challenging and therefore ultrasound measurement techniques<sup>15</sup> could be an alternative method to measure the moment arm change during exercise.

### 4.2 | ECC vs. HSR

There was no evidence in our results that eccentric motion leads to larger AT loading compared with concentric motion in either ECC or HSR while standing. In the literature, no significant difference in AT force has been found in eccentric and concentric motions in the standing position with the knee straight.<sup>15–17</sup> In addition, ultrasound measurements of AT displacement during a standing heel rise showed no difference between the eccentric and concentric phases.<sup>15</sup> In the current study, different loading conditions were performed with a greater number of participants; these confirmed that similar forces result from both eccentric and concentric loading. In the seated position, larger tendon forces were seen in the concentric phase compared with the eccentric phase. However, as the forces were low, this is unlikely to be clinically significant.

The AT loading results suggest that ECC with the knee bent could be an efficient loading activity for rehabilitation. Because of the increased ankle range of motion, with a 79 N (0.08–0.16 BW, for our participants) bar, the peak AT force could reach a level similar to someone bearing 0.50 BW in a

**TABLE 3** Multivariate regression analysis of the normalized peak Achilles tendon force during standing and seated motion in heavy slow resistance training

Standing (adj. $R^2 = 0.617$ )					
Predictor	Rate ratio	Standardized beta	Lower	Upper	<i>p</i>
Loading (BW)	3.533	0.649	2.725	4.342	<0.001
Peak dorsiflexion angle ( $^\circ$ )	-0.072	0.428	-0.098	-0.047	<0.001
Body mass (kg)	-0.021	-0.254	-0.033	-0.008	<0.01
Seated (adj. $R^2 = 0.843$ )					
Predictor	Rate ratio	Standardized beta	Lower	Upper	<i>p</i>
Loading (BW)	3.394	0.787	2.98	3.808	<0.001
Peak dorsiflexion angle ( $^\circ$ )	-0.034	0.339	-0.043	-0.024	<0.001

straight knee position. This finding emphasizes the importance of bent knee rehabilitation in a home setting.

Increased dorsiflexion angle is key to load an AT effectively. In a clinical setting, dorsiflexion angle depends largely on the individual; it is often difficult to control when comparing the effectiveness of the different rehabilitation protocols, especially when the joint range of motion is restricted by pain. As joint angles are frequently not reported in published randomized controlled trials, this may account for the variation in effectiveness of exercise regimens. It may also be concluded that if a patient is to perform ECC rehabilitation activities at home, where they likely have a single mass with which to exercise, a larger load is needed when the knee is straight.

Peak ankle dorsiflexion angle significantly affected loading through the AT. Each 5° increase in peak ankle dorsiflexion equated to an increase of 0.35 BW on the AT force in the standing position and 0.17 BW in the seated position. This twofold effect when standing is possibly due to the longer moment arm of the center of mass (additional weight and whole body) in the standing position (Figure 3). In the seated position, the moment arm of the center of mass (additional weight, leg and part of the thigh) became shorter. Therefore, under the same loading, a standing position might be more effective in increasing AT load than a seated position.

Although our results suggest standing might be more effective in loading the tendon, a seated position still has an important role in training only the soleus muscle, and thus part of the AT. Loading sharing between gastrocnemius and soleus in different postures may play a role in rehabilitation. Previous studies have shown that the Achilles tendon fiber has a rotational structure and soleus makes up the anterior or anterior-medial aspect of the Achilles tendon.<sup>35,36</sup> Loading only part of the tendon may affect the efficacy of rehabilitation. More studies are required to investigate the differential loading between each component of the triceps surae in AT rehabilitation to clarify the effectiveness of training in standing and seated positions.

In HSR, the seated leg press was not tested as the knee joint was also extended. This exercise has the same ankle and knee angle as in the standing position, while the hip flexion angle differs. Biomechanical studies often consider the hip and ankle joints independently because no common muscle crosses both joints.<sup>37,38</sup> However, there is emerging evidence which shows that the flexed hip with extended knee could reduce the ankle dorsiflexion angle,<sup>39,40</sup> although the mechanism is not fully understood. It is possible that in the seated leg press position, ankle dorsiflexion could be further restricted, decreasing the peak force through the AT.

There are several limitations to this study. First, the testing protocol was not randomized. This was because that the protocol was designed to reach high load and simulated the rehabilitation of athletes, who were at high risk

for midportion Achilles tendinopathy. In order to avoid any effects of fatigue, the heavier protocols were performed first. To further avoid any chance of fatigue, a specific maximum voluntary contraction (MVC) trial was not performed; the heaviest loading condition (single leg standing heel rise and drop with the addition of 50% BW) was used as a surrogate trial to obtain MVC. In addition, this study was conducted on a healthy cohort rather than patients with tendinopathy. The kinematics of symptomatic patients performing these exercises with pain may be different from those in healthy volunteers. In addition, the moment arms were estimated from MRI scans and then this was compared with the position of the ankle during the rehabilitation exercises, measured using an optical motion capture system. We did not perform a comparison between the MRI and motion capture measures of ankle pose, thus theoretically introducing a potential error in the definition of ankle angles. The findings presented in this study are based on the mechanical performance of the AT estimated in a controlled laboratory scenario and can be used to facilitate the development of future rehabilitation protocols. It is acknowledged that the estimation of AT force was based on the assumption that ankle dorsiflexion torque was provided solely by triceps surae through the AT and did not consider the effect of other muscles. This was based on the estimation that triceps surae accounts for 90% of ankle torque during normal walking. The activity of other agonistic/antagonistic muscle forces may have affected the results presented here. Measuring muscle excitation could provide some additional information; however, accurate conversion of electric signal to force is still controversial. More research is needed to investigate the force distribution within triceps surae. Finally, direct measurement of the actual tendon force in our participants was not feasible; therefore, the values serve as theoretical estimation of tendon force during rehabilitation exercises.

## 5 | CONCLUSION

In the ECC and HSR, the AT force increased during ankle dorsiflexion. The peak force in the eccentric phase was not greater than that in the concentric phase while standing. External loading and peak ankle dorsiflexion angle were the major factors affecting peak AT force. Based on these biomechanical findings, it may be beneficial to reach full ankle dorsiflexion during these rehabilitation activities, as this provides the highest peak force through the tendon. This may help to improve the effectiveness of rehabilitation protocols and facilitate the biomechanical understanding of the AT force during rehabilitation. Further studies are required to clarify the change in AT moment arm in extreme dorsiflexion angles and the effects of other biomechanical aspects, such as shear force between the adjacent structures.

## 6 | PERSPECTIVE

This study has provided biomechanical evidence to increase the effectiveness of AT loading during rehabilitation. Based on the results, we highlighted the importance of maximum dorsiflexion of the ankle, especially in the standing position. The quantified effect of the loading and joint angle on the force through the AT can be used to compare the efficacy of different protocols. In ECC and HSR, the overall loading of the AT may be increased if exercises reach full ankle dorsiflexion. A standing position could be more effective than a seated one in increasing the tendon loading. From a clinical perspective, this has the potential to increase the effectiveness of the rehabilitation protocol.

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### DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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### REFERENCES

- Kingma JJ, de Knikker R, Wittink HM, Takken T. Eccentric overload training in patients with chronic Achilles tendinopathy: a systematic review. *Br J Sports Med.* 2007;41(6):e3.
- Jarvinen TA, Kannus P, Maffulli N, Khan KM. Achilles tendon disorders: etiology and epidemiology. *Foot Ankle Clin.* 2005;10(2):255-266.
- Kujala UM, Sarna S, Kaprio J. Cumulative incidence of achilles tendon rupture and tendinopathy in male former elite athletes. *Clin J Sport Med.* 2005;15(3):133-135.
- Lysholm J, Wiklander J. Injuries in runners. *Am J Sports Med.* 1987;15(2):168-171.
- Alfredson H, Cook J. A treatment algorithm for managing Achilles tendinopathy: new treatment options. *Br J Sports Med.* 2007;41(4):211-216.
- Allison GT, Purdam C. Eccentric loading for Achilles tendinopathy—strengthening or stretching? *Br J Sports Med.* 2009;43(4):276-279.
- Magnussen RA, Dunn WR, Thomson AB. Nonoperative treatment of midportion Achilles tendinopathy: a systematic review. *Clin J Sport Med.* 2009;19(1):54-64.
- Meyer A, Tumilty S, Baxter GD. Eccentric exercise protocols for chronic non-insertional Achilles tendinopathy: how much is enough? *Scand J Med Sci Sports.* 2009;19(5):609-615.
- Rowe V, Hemmings S, Barton C, Malliaras P, Maffulli N, Morrissey D. Conservative management of midportion Achilles tendinopathy: a mixed methods study, integrating systematic review and clinical reasoning. *Sports Med.* 2012;42(11):941-967.
- Stanish WD, Rubinovich RM, Curwin S. Eccentric exercise in chronic tendinitis. *Clin Orthop Relat Res.* 1986;208:65-68.
- Fyfe I, Stanish WD. The use of eccentric training and stretching in the treatment and prevention of tendon injuries. *Clin Sports Med.* 1992;11(3):601-624.
- Grigg NL, Wearing SC, Smeathers JE. Eccentric calf muscle exercise produces a greater acute reduction in Achilles tendon thickness than concentric exercise. *Br J Sports Med.* 2009;43(4):280-283.
- Mafi N, Lorentzon R, Alfredson H. Superior short-term results with eccentric calf muscle training compared to concentric training in a randomized prospective multicenter study on patients with chronic Achilles tendinosis. *Knee Surg Sports Traumatol Arthrosc.* 2001;9(1):42-47.
- Malliaras P, Barton CJ, Reeves ND, Langberg H. Achilles and patellar tendinopathy loading programmes: a systematic review comparing clinical outcomes and identifying potential mechanisms for effectiveness. *Sports Med.* 2013;43(4):267-286.
- Chaudhry S, Morrissey D, Woledge RC, Bader DL, Screen HR. Eccentric and concentric loading of the triceps surae: an in vivo study of dynamic muscle and tendon biomechanical parameters. *J Appl Biomech.* 2015;31(2):69-78.
- Weinert-Aplin RA, Bull AM, McGregor AH. Investigating the effects of knee flexion during the eccentric heel-drop exercise. *J Sports Sci Med.* 2015;14(2):459-465.
- Rees JD, Lichtwark GA, Wolman RL, Wilson AM. The mechanism for efficacy of eccentric loading in Achilles tendon injury; an in vivo study in humans. *Rheumatology (Oxford).* 2008;47(10):1493-1497.
- Kongsgaard M, Qvortrup K, Larsen J, et al. Fibril morphology and tendon mechanical properties in patellar tendinopathy: effects of heavy slow resistance training. *Am J Sports Med.* 2010;38(4):749-756.
- Silbernagel KG, Thomee R, Eriksson BI, Karlsson J. Continued sports activity, using a pain-monitoring model, during rehabilitation in patients with Achilles tendinopathy: a randomized controlled study. *Am J Sports Med.* 2007;35(6):897-906.
- Silbernagel KG, Thomee R, Thomee P, Karlsson J. Eccentric overload training for patients with chronic Achilles tendon pain—a randomised controlled study with reliability testing of the evaluation methods. *Scand J Med Sci Sports.* 2001;11(4):197-206.
- Beyer R, Kongsgaard M, Hougs Kjaer B, Ohlenschlaeger T, Kjaer M, Magnusson SP. Heavy slow resistance versus eccentric training as treatment for Achilles tendinopathy: a randomized controlled trial. *Am J Sports Med.* 2015;43(7):1704-1711.
- Kongsgaard M, Kovanen V, Aagaard P, et al. Corticosteroid injections, eccentric decline squat training and heavy slow resistance training in patellar tendinopathy. *Scand J Med Sci Sports.* 2009;19(6):790-802.
- Bohm S, Mersmann F, Tettke M, Kraft M, Arampatzis A. Human Achilles tendon plasticity in response to cyclic strain: effect of rate and duration. *J Exp Biol.* 2014;217(22):4010-4017.
- Cleather DJ, Bull AM. The development of a segment-based musculoskeletal model of the lower limb: introducing FreeBody. *R Soc Open Sci.* 2015;2(6):140449.
- Ding Z, Nolte D, Kit Tsang C, Cleather DJ, Kedgley AE, Bull AM. In vivo knee contact force prediction using patient-specific musculoskeletal geometry in a segment-based computational model. *J Biomech Eng.* 2016;138(2):021018.



26. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech.* 1996;29(9):1223-1230.
27. Alexander CF, Lum I, Reid S, et al. A simple but reliable method for measuring 3D Achilles tendon moment arm geometry from a single, static magnetic resonance scan. *J Biomech.* 2017;55:134-138.
28. Gatt A, Chockalingam N, Chevalier TL. Sagittal plane kinematics of the foot during passive ankle dorsiflexion. *Prosthet Orthot Int.* 2011;35(4):425-431.
29. Fath F, Blazeovich AJ, Waugh CM, Miller SC, Korff T. Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans. *J Appl Physiol (1985).* 2010;109(6):1644-1652.
30. Maganaris CN, Baltzopoulos V, Sargeant AJ. In vivo measurement-based estimations of the human Achilles tendon moment arm. *Eur J Appl Physiol.* 2000;83(4-5):363-369.
31. Hashizume S, Iwanuma S, Akagi R, Kanehisa H, Kawakami Y, Yanai T. In vivo determination of the Achilles tendon moment arm in three-dimensions. *J Biomech.* 2012;45(2):409-413.
32. Sheehan FT. The 3D in vivo Achilles' tendon moment arm, quantified during active muscle control and compared across sexes. *J Biomech.* 2012;45(2):225-230.
33. DiGiovanni CW, Kuo R, Tejwani N, et al. Isolated gastrocnemius tightness. *J Bone Joint Surg Am.* 2002;84(6):962-970.
34. Henriksen M, Aaboe J, Bliddal H, Langberg H. Biomechanical characteristics of the eccentric Achilles tendon exercise. *J Biomech.* 2009;42(16):2702-2707.
35. Klein Horsman MD, Koopman HF, van der Helm FC, Prose LP, Veeger HE. Morphological muscle and joint parameters for musculoskeletal modelling of the lower extremity. *Clin Biomech (Bristol, Avon).* 2007;22(2):239-247.
36. Standring S, Borley NR, Gray H. *Gray's anatomy: the anatomical basis of clinical practice.* 40. Edinburgh: Churchill Livingstone/Elsevier. 2008.
37. Andrade RJ, Lacourpaille L, Freitas SR, McNair PJ, Nordez A. Effects of hip and head position on ankle range of motion, ankle passive torque, and passive gastrocnemius tension. *Scand J Med Sci Sports.* 2016;26(1):41-47.
38. Mitchell B, Bressel E, McNair PJ, Bressel ME. Effect of pelvic, hip, and knee position on ankle joint range of motion. *Phys Ther Sport.* 2008;9(4):202-208.
39. Szaro P, Witkowski G, Smigielski R, Krajewski P, Ciszek B. Fascicles of the adult human Achilles tendon - an anatomical study. *Ann Anat.* 2009;191(6):586-593.
40. Edama M, Kubo M, Onishi H, et al. The twisted structure of the human Achilles tendon. *Scand J Med Sci Sports.* 2015;25(5):e497-e503.

## SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section.

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