The Effects of Orthotic Heel Lifts on Achilles Tendon Force and Strain During Running

Dominic James Farris,1,2 Erica Buckeridge,1,3 Grant Trewartha,1 and Miranda Polly McGuigan1

1University of Bath; 2University of North Carolina-Chapel Hill and North Carolina State University; 3Imperial College London

This study assessed the effects of orthotic heel lifts on Achilles tendon (AT) force and strain during running. Ten females ran barefoot over a force plate in three conditions: no heel lifts (NHL), with 12 mm heel lifts (12HL) and with 18 mm heel lifts (18HL). Kinematics for the right lower limb were collected (200 Hz). AT force was calculated from inverse dynamics. AT strain was determined from kinematics and ultrasound images of medial gastrocnemius (50 Hz). Peak AT strain was less for 18HL (5.5 ± 4.4%) than for NHL (7.4 ± 4.2%) (p = .029, effect size [ES] = 0.44) but not for 12HL (5.8 ± 4.8%) versus NHL (ES = 0.35). Peak AT force was significantly (p = .024, ES = 0.42) less for 18HL (2382 ± 717 N) than for NHL (2710 ± 830 N) but not for 12HL (2538 ± 823 N, ES = 0.21). The 18HL reduced ankle dorsiflexion but not flexion-extension ankle moments and increased the AT moment arm compared with NHL. Thus, 18HL reduced force and strain on the AT during running via a reduction in dorsiflexion, which lengthened the AT moment arm. Therefore, heel lifts could be used to reduce AT loading and strain during the rehabilitation of AT injuries.

Keywords: injury, muscle mechanics, ultrasound, rehabilitation

The Achilles tendon (AT) is the most commonly injured tendon in the human body (Lesic & Bumbasirevic, 2004), particularly in athletic populations (Jozsa & Kannus, 1997). Acute injuries of the AT include partial or complete ruptures and the chronic condition tendinosis represents a gradual degeneration of the tendon, presenting as lesions in the tendon core (Aspenberg, 2007). Surgical procedures and primary treatment for AT ruptures are reasonably well established. However, the efficacies of current conservative and rehabilitation interventions for managing chronic AT conditions are less well established.

There is evidence to suggest that moderate loading of injured tendon may be beneficial to the healing process (Alfredson, 2003; Magnusson et al., 2003; Wang, 2006). This could be due to the increased collagen type I synthesis that has been demonstrated to occur with loading in healthy and injured AT (Langberg et al., 1999; Langberg et al., 2001; Magnusson et al., 2003). However, care must be taken when loading injured tendons as experimental models have shown a decrease in the load at which injured tendons rupture (Aspenberg, 2007). Thus, injured tendons are more susceptible to failure under loading and this should be considered in any rehabilitation program. Arampatzis et al. (2007) showed that exercises that elicited higher AT strains (≈4.5%) were necessary for adaptational responses to occur in AT mechanical properties as part of a training program. Such strains are slightly less than those typically experienced by the AT during slow running which are in excess of 5% (Lichtwark et al., 2007). Therefore, it might be desirable during recovery from AT injuries to allow individuals to run with an intervention that reduces AT loads and subsequent strains from normal but, not so much as to negate the adaptational responses to loading.

An intervention with the potential to achieve this is the use of orthotic heel lifts. Heel lifts have been suggested as an effective treatment for AT disorders based on clinical outcomes (Maclellan & Vyvyan, 1981; Grisogono, 1989) but evidence for the mechanism by which they work is incomplete. It could be that they simply alleviate pain or it could be that their effects on ankle mechanics alter tendon loading as described above and thus promote tendon healing. Dixon and Kerwin (1999) showed that heel lifts reduced peak ankle dorsiflexion during the stance phase of running and that this resulted in reduced peak gastrocnemius muscle-tendon...
unit (MTU) length. They interpreted this to imply a reduction in AT strain. However, tendon length changes can be independent of whole MTU length changes (Roberts, 2002) and are primarily determined by the forces applied to the tendon. Dixon and Kerwin (1998, 2002) presented AT force estimates from inverse dynamics that suggested peak AT force was not significantly reduced when runners ran with heel lifts. This is supported by data showing that peak sagittal ankle joint moments during running were not reduced by wearing shoes with raised heels (Reinschmidt & Nigg, 1995). However, these results appear somewhat contradictory to the reduction in AT strain presented by Dixon and Kerwin (1999). This may be because Dixon and Kerwin (1999) inferred reduced tendon strain from reduced lengthening of the MTU. It is unclear exactly how AT force and strain during running are influenced by the use of heel lifts. More direct measurements of AT length may help to clarify this and provide insight into how heel lifts may be useful in rehabilitation from AT injuries.

Tendinous movements within human muscle have been measured in-vivo using real time B-mode ultrasound imaging (Fukunaga et al., 1996). By combining the ultrasound system with motion analysis, the elastic behavior of the AT has been measured during running (Fukunaga et al., 2001; Lichtwark et al., 2007). The present study applied these methods to measure tendon strain in-vivo during running with heel lifts. The purpose of this study was to assess the effects of orthotic heel lifts on AT strain and force during the stance phase of running. It was hypothesized that heel lifts would reduce ankle dorsi-flexion and that this would result in a reduction in AT strain and force during the stance phase of running. A reduction in AT strain would be considered evidence to support the use of heel lifts in treating AT injuries.

### Methods

#### Participants

Ten healthy, active females [mean (SD), height = 1.68 (0.10) m, mass = 63.9 (10.8) kg, age 20 (1) years] without any current musculoskeletal injuries and whose normal running gait used a heel-striking ground contact, gave written informed consent to participate in the study. Ethical approval for the study was granted by an institutional ethics committee at the University of Bath and the procedures complied with the regulations set out in the Declaration of Helsinki (2008).

#### Experimental Protocol

To determine participant specific "normal" gait speeds, participants were asked to run across the width (10 m) of the laboratory at a natural cadence. A photo cell system (Newtest, Finland) consisting of two sets of light gates at a separation of 4 m was used to measure running speed. This was repeated four times and the mean and standard deviation were calculated. A range of mean plus or minus one standard deviation for each subject was set as acceptable speed for the subsequent experimental trials. This was to ensure consistent running speed and trials were rejected if these criteria were not met. These initial trials were used to visually confirm that all participants used a heel strike running style.

Four trials were completed in each of three conditions (total of 12 trials per subject). The three conditions were: without heel lifts (NHL); with 12 mm heel lifts (12HL); and with 18 mm heel lifts (18HL). All conditions were performed barefoot so as to negate the effects of different shoe designs and isolate the effects of the lifts. The heel lifts were commercially available (A. Algeo Ltd., UK) and made from high-density ethyl vinyl acetate (EVA) with a constant height along their length (not tapered). They were attached below the heel using minimal amounts of Transpore (3M, USA) tape (so as not to affect ankle mechanics) and similar taping was applied to the foot during the NHL condition to control for any possible effects of taping. In each condition the subjects were required to run across the laboratory, striking a single force plate with only their right foot during one stride. The order of experimental conditions was randomized for each subject. Subjects were allowed to familiarize themselves with the different heel lift conditions by running across the laboratory in the same manner as in experimental trials as many times as necessary.

#### Measurement of Muscle Activity

Surface electromyography (EMG) data were collected from medial and lateral gastrocnemius, soleus and tibialis anterior muscles of the right leg using a wireless system (Telemyo 2400T, Noraxon inc., USA). Pairs of electrodes (separation 20 mm) were placed on cleaned and abraded sites on the muscle belly. Signals were logged at 3000 Hz through the CODA system (see below) to allow for time synchronization with other data streams. Onset and ending of muscular activity were visually identified from raw EMG traces by the same experimenter as substantial departures from the background signal (Hodges and Bui, 1996). Raw EMG signals were rectified, high-pass filtered at 20 Hz and average filtered over 50 ms windows. The processed signals were then integrated from 100 ms before ground contact to toe off. To compute the average rectified EMG (AREMG), the integrated values were divided by time.

#### Kinematic Measurements

Three-dimensional (3D) marker locations were recorded at 200 Hz using a bilateral (two scanner), active marker, motion analysis system (CODA, Charnwood Dynamics Ltd, UK) to build a four segment model of the pelvis and right leg (pelvis, right thigh, right shank and right foot) using Visual3D software (C-Motion Inc., USA). Marker placements were 1st and 5th metatarsal-phalangeal joints; lateral and medial malleoli; calcaneus; lateral and medial epicondyles; greater trochanter; left and right anterior superior iliac spines; left and right iliac crests, to define segment endpoints and at least three ‘tracking’ markers attached to each segment to track segment orientation.
and position during trials. Raw 3D coordinate data were interpolated using cubic splines to fill gaps (maximum width = 20 samples) and filtered at 10 Hz using a fourth order, low-pass Butterworth digital filter. 3D Cardan joint angles were calculated using Visual3D as the orientation of the distal segment in the proximal segment’s coordinate system. The flexion-extension angles reported adhered to the convention that 180° represented full extension (or plantar flexion) of the joint. Data from Dempster (1955) were used to generate inertia parameters for segments which were modeled as rigid geometric cones and cylinders.

**Kinetic Analyses**

Ground reaction force (GRF) data were sampled at 1000 Hz by a single force plate (Kistler, 9287BA, Kistler Instruments Ltd., Switzerland) and logged by the CODA software. GRF data were filtered with a low-pass fourth order Butterworth filter using the same cut-off frequency (10 Hz) as for kinematic data as has been recommended for inverse dynamic analysis (Bisseling & Hof, 2006). Filtered GRF (down-sampled to 200 Hz) and kinematic data were used to compute 3D ankle and knee joint moments in the same coordinate system as joint angles, during the stance phase of the stride using Visual3D. The flexion-extension component of ankle joint moments was used in the subsequent calculation of AT force. All kinetic and kinematic variables were normalized to 101 points over stance time.

**Calculation of Achilles Tendon Force**

AT force was calculated using the following formula:

\[
\text{ATF} = \frac{M_{\text{ANK}}}{M_{\text{AT}}}
\]

where, ATF is AT force, \(M_{\text{ANK}}\) is the flexion-extension ankle moment and \(M_{\text{AT}}\) is the instantaneous moment arm of the AT.

The moment arm of the AT was the perpendicular distance from the line of action of the tendon to the center of rotation of the ankle joint. In the calculation of tendon length (see subsequent section), a position vector representing the line of action of the AT was calculated. The perpendicular distance between this vector and a virtual point midway between the two malleoli (representing the center of rotation of the ankle) was calculated for each frame of motion data and used as an instantaneous moment arm. The moment arm of the GRF vector (\(M_{\text{GRF}}\)) at the ankle joint was calculated as the perpendicular distance from the GRF vector to the virtual marker between the malleoli.

**Calculation of Achilles Tendon Strain**

Tendon length was defined as the distance between the insertion of the AT on the calcaneous and the muscle-tendon junction (MTJ) of the medial gastrocnemius and the AT. To calculate tendon length, a combination of synchronously collected real-time B-mode ultrasound images and motion analysis was used (Lichtwark and Wilson, 2005). The MTJ of the AT and gastrocnemius was imaged longitudinally using a 128 element, linear array ultrasound probe (LV7.5/60/96Z, TELEMED, Lithuania) operating at 8.0 MHz. The probe was secured to the subject’s leg using COBAN (3M, USA) bandaging tape. Images of the MTJ were sampled at 50 Hz and the position of the MTJ in the images was subsequently digitized using a custom MATLAB (Mathworks Inc, USA) interface. To obtain the position of the MTJ in the global coordinate system, the position and orientation of the ultrasound probe were tracked using three motion analysis markers that were attached to define a “probe coordinate system.” The orientation and position of the probe coordinate system relative to the ultrasound image was constant in the form of a rotation matrix and a 3D translation and known from a prior calibration [see Lichtwark and Wilson, 2005] for a description of this calibration process. A second rotation matrix describing the orientation of the probe coordinate system in the global (laboratory) coordinate system was calculated for each time point using the global position of the markers on the probe. Digitized coordinates of the MTJ within the ultrasound image were first rotated and translated into the probe coordinate system and then rotated and translated into the global coordinate system. Once the position of the MTJ was known in the global coordinate system, a vector between it and a CODA marker on the posterior-superior aspect of the calcaneous (representing the position of the insertion of the tendon) could be computed. The magnitude of this vector was taken as the length of the AT. Figure 1 illustrates that the marker for the AT insertion is displaced from the actual insertion in the sagittal plane by half the width of the marker (2.5 mm) and the thickness of tissues overlying the insertion (assessed on one individual with ultrasound imaging to be 6 mm). Taking this displacement (8.5 mm) to be one side of a right angled triangle and a typical distance from the marker to the MTJ (208 mm) as the hypotenuse, simple trigonometry gives a corrected AT length (the adjacent side of the triangle) value of 207.83 mm. Thus, the result- ing error was below the resolution of the tendon length calculation and this correction was deemed unnecessary for experimental trials. Another potential source of error is rotation of the probe relative to the leg which would change the plane of the image. Lichtwark and Wilson (2005) showed that longitudinal rotations of 22.9° and medio-lateral rotations of 0.8° as measured during hopping would result in maximum errors of 2.6 mm in tendon length (1.1% strain). Maximum probe rotation relative to the shank in the current study during running was less than 8.1° about the longitudinal axis and 1.2° about the medio-lateral axis. Thus, it was expected that errors would be generally less than 2.6 mm and would not affect the findings of this study. Errors due to movement of the MTJ out of the image plane could not be quantified but it was assumed that the majority of tendon length change would occur in the longitudinal direction and thus, in the image plane.
Tendon length values were calculated during the stance phase of running trials and also during three standing trials in which the subjects were barefoot (no heel lifts) and were instructed to stand in a natural posture. Tendon strain was then calculated during the stance phase of running as:

$$\varepsilon = \frac{\text{ATL} - \text{ATL}_{\text{ST}}}{\text{ATL}_{\text{ST}}} \times 100\%$$

where $\varepsilon$ = AT strain, ATL = instantaneous AT length, ATL$_{ST}$ = average AT length during standing trials. This equation does not provide a measure of true strain as this would require ATL$_{ST}$ to be replaced with the AT length when tendon force is equal to zero. However, this length is unknown using the present methods and so the above calculation provides a strain relative to standing which is comparable between experimental conditions. The length of the gastrocnemius MTU was computed using ankle and knee joint kinematics with the equations of Hawkins and Hull (1990) which return a muscle length normalized to shank length.

### Statistical Analyses

Values throughout stance and peak values for AT strain; AT force; ankle dorsi-flexion angle; knee flexion angle; knee and ankle joint moments; ankle and knee net, positive and negative joint work; AT moment arm; MTU length; AREMG and GRF moment arm were taken from each trial and averaged over all four trials in each condition for each subject. Group means were computed for these main outcome measures for each condition. After testing outcome measures for normality, statistical differences between heel lift conditions were tested for using a one-way repeated measures analysis of variance (ANOVA). Where the ANOVA $F$ ratio was shown to be significant ($p < 0.05$), Tukey’s honestly significant difference was used to assess between which conditions the differences existed. Effect sizes were also computed for outcome measures as the mean difference divided by the pooled standard deviation (Cohen, 1988). Time series data were cropped to periods of stance on the analyzed limb and normalized by interpolation (cubic spline) to 101 points over stance time and were plotted against percentage of stance.

### Results

Running speeds for the group [mean (SD)] were 3.1 (0.2), 3.0 (0.1) and 3.2 (0.2) m·s$^{-1}$ for NHL, 12HL and 18HL, respectively. Figure 2 shows group mean ankle and knee angle plots over the stance phase of a single running stride. For both 12HL ($p = .016$, ES = 1.76) and 18HL ($p = .016$, effect size (ES) = 2.27), the ankle was significantly less dorsiflexed at time of peak dorsiflexion than for NHL. Group mean peak dorsiflexion angles for each condition were: NHL: 81.1 (3.9)$^\circ$, 12HL: 86.4 (1.9)$^\circ$, 18HL: 87.1 (4.7)$^\circ$. There were no significant differences for knee angle between any conditions during stance ($p > .05$). Peak gastrocnemius MTU length was significantly ($p = .017$, ES = 2.2) less in 18HL than NHL (Table 1).

Group mean peak values of AT force were NHL = 2710 (830) N, 12HL = 2538 (823) N and 18HL = 2382 (717) N (Figure 3). Peak AT force was significantly less for 18HL than for NHL ($p = .024$, ES = 0.42) but not different between 12HL and NHL ($p > .05$, ES = 0.21). At the time of peak ankle moment, the AT moment arm was significantly ($p = .03$, ES = 0.37) greater for 18HL [65 (13) mm] than for NHL [60 (14) mm] (Figure 3). AT moment arms for 12HL were not significantly different from NHL ($p > .05$, ES = 0.23). Peak AT strain was significantly lower for 18HL [5.5 (4.4)%] compared with NHL [7.4 (4.2)%], $p = .029$, ES = 0.44]. The difference was not significant between 12HL [5.8 (4.8)%] and NHL ($p > .05$, ES = 0.35). Neither the plantar flexors nor tibialis anterior showed any significant differences in AREMG, on and off timings or timing of peak EMG amplitude between conditions ($p > .05$).

### Discussion

This study evaluated the effects of orthotic heel lifts on AT strain and force during the stance phase of running and supported the hypothesis that heel lifts reduce ankle dorsi-flexion and also reduce AT strain and forces.
Figure 2 — Group mean (a) MTU length (normalized to shank length), (b) ankle angle and (c) knee angle time histories over stance for NHL (solid line), 12HL (dashed line) and 18HL (dotted line). 180° represents full extension (or plantar-flexion) of the joint. Standard errors were omitted for clarity.

Figure 2 shows that the ankle joint was in a less dorsi-flexed position throughout stance in the 18HL condition compared with NHL. The effect of this change in angular displacement was a reduced length of the gastrocnemius MTU in 18HL (Figure 2, Table 1). The values for gastrocnemius MTU lengths are normalized to shank length and the reduction in peak length of 0.01 would equate to approximately 4 mm. This result is in
Table 1  Group mean (SD): Peak AT strain, MTU length (normalized to shank length), MA_GRF and CoP Position at the time of peak AT force

<table>
<thead>
<tr>
<th></th>
<th>NHL</th>
<th>12 HL</th>
<th>18 HL</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak AT strain (%)</td>
<td>7.7 (4.2)</td>
<td>5.8 (4.8)</td>
<td>5.5 (4.4)*</td>
</tr>
<tr>
<td>MTU length</td>
<td>1.14 (0.01)</td>
<td>1.13 (0.01)*</td>
<td>1.13 (0.01)*</td>
</tr>
<tr>
<td>MA_GRF (m)</td>
<td>0.17 (0.07)</td>
<td>0.18 (0.10)</td>
<td>0.17 (0.08)</td>
</tr>
<tr>
<td>COP relative to ankle joint (m)</td>
<td>0.13 (0.01)</td>
<td>0.13 (0.01)</td>
<td>0.13 (0.02)</td>
</tr>
</tbody>
</table>

*Statistically significant difference from NHL ($p < .05$).

Figure 3 — Group mean (a) flexion-extension ankle moment, (b) AT moment arm and (c) AT force time histories over stance for NHL (solid line), 12HL (dashed line) and 18HL (dotted line). Standard errors were omitted for clarity.
agreement with Dixon and Kerwin (1999) who also saw reduced dorsi-flexion and MTU length with the use of 15 mm heel lifts.

The values for peak AT strain were similar to those reported previously for running when measured using ultrasound techniques (Lichtwark et al., 2007). The present peak strain of 7.4% for NHL equated to an elongation of 15 mm which is less than that previously reported for the series elastic element of the gastrocnemius [20–25 mm (Ishikawa et al., 2007; Lichtwark et al., 2007)]. This is probably due to the fact that these previous studies included the aponeurosis length change whereas the current study did not. In fact, the present estimate of length change is in better agreement with other values reported for AT length change during running that did not include aponeurosis length change [=10 mm (Lichtwark and Wilson, 2006)]. Peak AT strain was reduced in 18HL compared with NHL. In fact, the mean difference in peak AT strain between 18HL and NHL was 1.9%. This result supports previous findings from less direct estimates of AT strain while running with and without heel lifts (Dixon and Kerwin, 1999). Given that average AT length during standing was 208 mm, this means that the average reduction in peak tendon length was 4 mm. This accounted for all of the reduction in total MTU length. Thus, the contractile part of the muscle did not change length between conditions.

The observed reduction in strain of the AT in 18HL suggests a reduction in force acting on the AT (assuming the material properties of the tendon are the same across conditions). There was a reduction in peak AT force for 18HL, coinciding with the reduction in AT strain. This suggested that there may have been some change in the kinetics at the ankle. However, the current study showed no changes in joint moment (Figure 3), GRFMA or CoP position (Table 1) at the ankle during stance. The AT force was calculated as the flexion-extension ankle moment in the AT force calculation. Based on the relative physiological cross-sectional areas of plantar flexors other than the triceps surae, the contribution of those other than the triceps surae to the net ankle moment in the AT force calculation is probably due to the fact that these previous studies did not account for any opposing contribution of the tibialis anterior or any assisting contribution of the plantar flexors. However, the values for AT force still seem low when compared with in-vivo measurements made using tendon buckle transducers during running which produce values of up to 4 kN at similar speeds (Komi, 1990). Komi’s (1990) data suggest the individual in their study was ~10 kg above the average mass of subjects in the current study which may explain the discrepancy. It was also the case that the current study did not account for any opposing contribution of the tibialis anterior or any assisting contribution of the plantar flexors other than the triceps surae made to the net ankle moment in the AT force calculation. Based on the relative physiological cross-sectional areas of plantar flexors, the contribution of those other than the triceps surae is likely to be trivial. Not accounting for tibialis anterior co-contraction might have resulted in an under-estimation of true tendon force values that might also help to explain why the reported AT forces were lower than those measured directly with buckle transducers (Komi, 1990). However, there were no significant differences in tibialis anterior AREMG or EMG onset, peak value and ending timings between heel lift and NHL conditions, suggesting that the activation of the tibialis anterior did not vary between conditions. Therefore, any changes in AT force calculation due to not accounting for tibialis anterior co-contraction could only be due to changes in tibialis anterior muscle length. Running speed was kept consistent for all conditions for each subject and so should not have affected any comparisons.
Our results suggest that heel lifts reduce force and strain on the AT during running. These effects were due to reduced dorsiflexion which lengthened the moment arm of the AT. These findings showed that the use of heel lifts during running was an effective means of loading the AT while not straining it to the normal levels. Loading of human AT via running has been reported to increase collagen synthesis within the tendon both as an acute response to a single bout of running (Langberg et al., 1999) and as a long term effect in response to 12 weeks running training (Langberg et al., 2001). Furthermore, resistance training that loads the tendon has been shown to increase the mechanical stiffness of the human AT (Kubo et al., 2004; Arampatzis et al., 2007) and eccentric loading of the AT has been shown to be effective in aiding symptomatic recovery from tendinopathies (Alfredson, 2003). These can be considered positive adaptations that would be desirable in the rehabilitation of AT injuries. Conversely, bed rest or paralysis which prevent AT loading tend to result in decreased stiffness of the AT (Narici and Maganaris, 2007) and stress-shielded rabbit tendons exhibited losses in tensile strength and Young’s modulus (Yamamoto, 1993). Therefore, loading of the AT may be crucial to successful recovery of tendon properties following injury. However, excessive straining of the AT may be the cause of the injury in the first instance (Smith et al., 2002) and may present a risk of reinjury until the AT is fully recovered. Rehabilitation of AT injuries can be lengthy and as such, adjusted AT loading may be required over a long time. The present study only assessed the effects of heel lifts after a short familiarization. Thus, the longer term effects of heel lifts on AT mechanics must be studied to confirm that their positive effects on AT loading are maintained throughout the course of their use.

In this study it was shown using in-vivo measurements that heel lifts adjust ankle mechanics to reduce AT force and strain. This has provided evidence for how heel lifts produce the previously observed clinical improvements (Macellman and Vyvyan, 1981). The peak strains when the heel lifts were worn were less than without heel lifts but still above the 4.5% that has been shown to be necessary for adaptive responses to occur in the AT (Arampatzis et al., 2007). The reduction in strain was only significant with the 18 mm heel lifts and so it is recommended that heel lifts of similar magnitude are used. Longitudinal studies of the long-term effects of heel lifts on running mechanics are needed along with studies examining how they interact with different types of footwear. Other future studies of interest may be to test if more proximal joint mechanics are affected. In addition, similar studies into the effects of heel lifts involving participants suffering or recovering from AT injuries may be important as injury may affect the properties of the AT or the mechanics of running.

Acknowledgments

The authors would like to acknowledge Glen Lichtwark (University of Queensland, Australia) for assistance with data analysis and MATLAB programming. The work undertaken for this study was funded by an internal institutional grant.

References


