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Biomechanical Properties of the Plantar Flexor Muscle–Tendon Complex 6 Months Post-Rupture of the Achilles Tendon

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ABSTRACT: We compared the effects of a non-weight bearing protocol (NWB) and a weight bearing (WB) protocol on energy stored, stiffness, and shock absorption in the plantar flexor muscle–tendon unit of patients managed non-operatively following an Achilles tendon rupture. Thirty-eight subjects were randomized to a WB cast fitted with a Bohler iron or a traditional non-weight-bearing cast. At a 6-month follow-up, a biomechanical assessment utilizing an isokinetic dynamometer allowed measurement of peak passive torque, energy stored, shock absorption, and stiffness. The WB group had greater peak passive torque (~20%). Irrespective of group, peak passive torque in unaffected legs was greater (~26%) than affected legs. Across the groups, energy stored in the NWB group was 74% of the WB group. The energy stored in affected legs was 80% of that in unaffected legs. Shock absorption was not significantly different across legs or groups. Irrespective of group, affected legs had significantly less stiffness (20–40%). While the augmentation of plaster with a Bohler iron to allow increased weight bearing had positive effects, deficits in affected compared to unaffected legs irrespective of group were notable, and should be addressed prior to participation in vigorous physical activities.

Keywords: Achilles; tendon; biomechanics; forces; rupture

Rupture of the Achilles tendon causes damage to collagen and elastin components of the tendon. The healing process leaves the tendon with altered architecture including disorganized fiber orientation, changes in cell and fiber density,[1] decreased type one collagen, and an increase in type 3 collagen.[2] Such changes likely affect the biomechanical properties, which in turn can affect maximal performance and efficiency of motion, rate of re-injury, shock absorption, and coordination. Normally, the anatomical structures provide the elasticity required to absorb and release energy, ultimately increasing the efficiency of motion in activities like walking, running, and jumping.[3] Muraoka et al.[4] showed that the length of the Achilles tendon and its aponeuroses are seven times longer than the muscle fascicles, while Herbert et al.[5] found significantly greater passive lengthening in the gastrocnemius tendon compared to the muscle fascicles. More recently Lichtwark and Wilson[6] showed that during gait, the active muscle fascicles remain relatively constant in length. Such findings suggest that the ability of the tendon to lengthen is of great importance in storing elastic energy, an attribute that might be affected by damage associated with its rupture and subsequent remodeling during the healing process.

Changes in tendon structure may also affect the likelihood of re-injury or strain injury to the musculotendinous complex of the plantar flexors. Stiffness (the degree of resistance offered by tissues in response to lengthening) may be important. Interest in stiffness as a factor influencing injury and re-injury arose from the observation that many muscle and tendon injuries occur when the joint is in the mid range and not in an overextended position. When tissues are too stiff, they cannot absorb sufficient energy during loading, and subsequently are more likely to be injured.[7,8] Given an incidence of re-injury to the tendon as high as 20% in conservatively managed patients,[9] it is of considerable interest to assess the stiffness of the plantar flexor muscle–tendon unit following rupture.

Shock absorption is an important role of muscle and tendons. During gait, a loss of shock absorption can lead to increased impulse forces at foot strike being transmitted towards the head.[10] These larger forces can place additional stress upon joint and bone structures, and have been implicated in the progression of osteoarthritis.[11] The plantar flexor muscles are important in limiting shock transmission,[10] and changes in their structure following rupture might influence their ability to undertake this role effectively.

Don et al.[12] suggested that alterations in the mechanical parameters of the Achilles tendon may lead to changes in muscle activation and hence the coordination of motion during gait. They observed that subjects with higher passive stiffness exhibited altered muscle activation patterns that influenced ankle range of motion and increased co-contraction.

The above-mentioned parameters therefore can negatively influence patient function following Achilles tendon rupture. During treatment, initial weight bearing status may influence tendon healing[13] and subsequently biomechanical parameters. Recently researchers[14] utilizing more aggressive non-operative protocols with early weight-bearing noted a far lower incidence of re-rupture, with rates approaching those of operative management. To date, no studies have explored the effects of the immobilization protocol on

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the mechanical properties of the plantar flexor muscle–tendon unit.

Our primary objectives were to compare the effects of a non-weight bearing protocol and a weight bearing protocol on energy stored, stiffness, and shock absorption in the plantar flexor muscles of a group of patients managed conservatively following an Achilles tendon rupture. Secondary measures that were assessed included strength and range of motion.

METHODS

Subjects

Subjects were recruited from the accident and emergency departments of three hospitals in the Auckland region between 2008 and 2010. Inclusion criteria were that subjects: had a functionally complete, isolated Achilles tendon rupture confirmed clinically by a palpable gap in the tendon and a positive Thompson test; presented for treatment < 72 h after injury; and could understand instructions and read in English. Exclusion criteria were: presence of an open injury; injury to the ipsilateral limb; neurological, rheumatic, and metabolic diseases that might affect the measurements undertaken; and previous history of injury or tendonitis affecting plantar flexor muscles. Written and verbal explanations of the procedures and an opportunity to ask questions about the study were provided to all subjects. Ethical approval was received from the Northern Regional Ethics Committee, and all subjects signed a document of informed consent.

Intervention

A third party with no affiliations to the investigators enrolled the patients using a computer generated software program and a sealed envelope protocol. Subjects were randomized to either a weight bearing cast fitted with a Bohler iron or a traditional non-weight bearing cast. Both groups had a total cast time of 8 weeks and were treated with a set protocol. The first cast was set to "full" equinus without manual pressure. Subjects fitted with a Bohler iron were instructed to begin weight bearing immediately. Subjects in the non-weight bearing group were given crutches and instructed not to place weight through the injured limb. After 4 weeks, both groups had their casts changed, and the amount of equinus was reduced to a "resting" position of 10–20° of plantar flexion. Instructions for the weight bearing group were to continue to weight bear as tolerated; the non-weight bearing group were instructed to continue to use crutches and not put weight through the injured limb. At 8 weeks the casts were removed, range of motion exercises were begun, as was weight bearing as tolerated in normal shoes fitted with a 2 cm heel raise. After 4 weeks the raises were removed. Graduated strength training was instigated at 6 weeks after cast removal with return to sports a further 3 months thereafter.

Procedures

A clinical and biomechanical assessment was undertaken 6 months post-injury. The clinical assessment included range of motion and strength testing. Questions concerning function (Lower Limb Task Questionnaire[15]) were also completed.

The biomechanical assessment involved an examination of passive motion on an isokinetic dynamometer. In accordance with manufacturer's instructions, subjects were seated with their ankle joint strapped to a Biodex® dynamometer (Biodex Corp, Shirley, NY). Initially, the ankle joint was positioned at 80% of maximal plantar flexion. The subject's knee was supported in extension. Subjects were asked to relax and not activate their lower limb muscles during the subsequent passive motion. The Biodex was programmed to move the ankle joint to 80% of the subject's maximum range of dorsiflexion at an angular velocity of 5°/s. Previous studies[16–18] showed that EMG responses of the plantar flexor muscles to stretch were minimal in these conditions. Each subject performed a familiarization set of ankle movements (10 reps) during which they watched their signals on the computer monitor and were given verbal feedback to ensure that muscle activity was not affecting their torque curves. Thereafter, they performed five reps that were saved for data analyses. Legs were tested in a random order.

Next, maximal effort strength testing was performed. Subjects were again seated in the dynamometer. An isokinetic evaluation that involved concentric muscle work was conducted at an angular velocity of 240°/s from a position of maximal plantar flexion to maximal dorsiflexion. Five submaximal reps were done as a "warm up" prior to testing. Thereafter, subjects performed three maximal effort repetitions of dorsi and plantar flexion on each leg. Standardized verbal encouragement was provided to ensure maximal effort.[19] The maximal torque recorded across the three trials was used in subsequent analyses.

Data Analyses and Processing

All data were processed using a standardized Matlab® script (The Mathworks, Natick, MA). Torque, angle, and velocity signals were filtered using a zero-phase low-pass (10 Hz), 2nd order Butterworth filter. To correct torque for gravity effects, the mass of the Biodex level arm was determined during cycles performed with no subject on the dynamometer.[20] The mass of the foot was neglected. Data without a smooth torque-angle relationship were excluded. All processing was performed on the 5th cycle to account for conditioning.[21] Thereafter, the torque and angle values were calculated each 5% of the total range of motion for loading and unloading phases. The energy stored during loading (E, the area under the loading curve), the energy restituted (ER, the area under the unloading curve), and the energy dissipated by the musculo-articular complex (ED, the hysteresis area) were calculated. ED was divided by E to calculate the dissipation coefficient (DC) representative of the shock absorbing quality of the plantar flexor muscles[21,22]:

$$\mathrm{DC} = \frac{\mathrm{ED}}{\mathrm{E}} = \frac{\mathrm{E} - \mathrm{ER}}{\mathrm{E}} \tag{1}$$

The torque-angle curves were fitted using a Sten–Knudsen model[23]:

$$\mathrm{T} = \frac{\mathrm{E}_0}{\alpha}\left(e^{\alpha\theta} - \mathrm{a}\right) \tag{2}$$

where T is the passive torque, θ the ankle angle, and a, E_0, and α are experimental constants determined using optimization. Both parameters (α, E_0, and α) were determined by minimizing the squared difference between the experimental and the modeled (Equation 2) responses using Matlab and the optimization toolbox (Levenberg–Marquard algorithm); R^2 was calculated to assess the quality of fit. Using this

exponential model (Equation 2), the stiffness (slope of the torque-angle relationship) was calculated at 0°, the mid range of dorsiflexion, and at the maximal range of motion of dorsiflexion.

Statistical Analyses

Statistical analyses were undertaken using the Statistical Package for Social Sciences (SPSS) Version 17.0 (SPSS, Inc., Chicago, IL). An alpha level of 0.05 was set across all tests. Descriptive data were assessed to determine the appropriateness of parametric analysis. The two treatment groups were compared for baseline characteristics of height, weight, and BMI using an independent *t*-test. Two factor ANOVA was used to assess differences in energy stored, peak passive torque, the dissipation coefficient, strength and ROM. The factors were groups (WB and NWB) and limbs (affected and unaffected). Departures from sphericity in the repeated measures were accounted for with the Huynh–Fedlt epsilon value. For stiffness, a three factor ANOVA was used; the 3rd factor was dorsiflexion position (0°, mid range, and maximal range). To assess where differences might lie across these three points, the Bonferroni post hoc test was utilized.[24]

RESULTS

Thirty-eight subjects volunteered to participate; eight were excluded due to technical issues related to data collection or irregular torque-angle traces. Their demographic characteristics were not significantly different from those subjects who completed data collection (Table 1).

Significant main effects for limbs and groups, but no interaction effect, were found for maximal passive peak torque at the end of dorsiflexion. Unaffected legs had larger torque (~26%) than affected legs, and the WB group had greater maximum torque (~20%) compared to the NWB group (Fig. 1).

Significant main effects for limbs and groups, but no interaction effect, were found for the amount of energy stored. Irrespective of group, the mean energy stored in affected limbs was 20% less than that in the unaffected limbs. Across the groups, energy stored in the NWB group was 26% less than the WB group (Fig. 2). The DC was not significantly different across limbs or groups.

Table 1. Subject Characteristics in the Weight Bearing and Non-Weight Bearing Groups

Characteristic	Weight bearing group ($N = 16$)	Non-weight bearing group ($N = 14$)
Sex	8 male	7 male
Age (years)	38 (5)	41 (13)
Height (cm)	171 (8)	172 (9)
Weight (kg)	93 (13)	87 (15)
LLTQ-ADL (/40)	34 (6)	35 (5)
LLTQ-Rec (/40)	21 (9)	23 (11)

LLTQ-ADL, lower limb Task Questionnaire-activities of daily living; LLTQ-Rec, lower limb Task Questionnaire-activities of recreation.

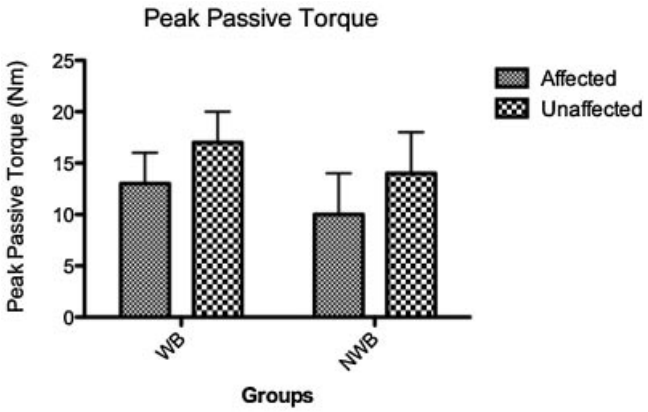

Figure 1. Peak passive torque across limbs and groups (WB: weight bearing; NWB: non-weight bearing). Significant main effects ($p < 0.05$) were observed across limbs and groups with no interaction effect.

For stiffness, there were significant main effects for limbs and position in range of motion, but no main effect for groups. There were no significant interaction effects. The affected limb had significantly less stiffness, which ranged from 20% to 40% at the start, mid range, and the end of range of motion (Fig. 3).

For range of motion for dorsiflexion, there was no significant main effect for limbs and group, and no interaction effect. However, for plantar flexion, there was a significant main effect for limbs, but no group or interaction effect. For the latter limb effect, a 3.5° mean deficit was observed in the affected limbs.

For strength testing, there were significant main effects across limbs but not across groups. There were no significant interaction effects. Peak plantar flexion torque was significantly less (~20%) in the affected limbs.

DISCUSSION

Our findings show that the group with a Bohler iron, which allowed increased weight bearing early in rehabilitation, had greater values of peak passive

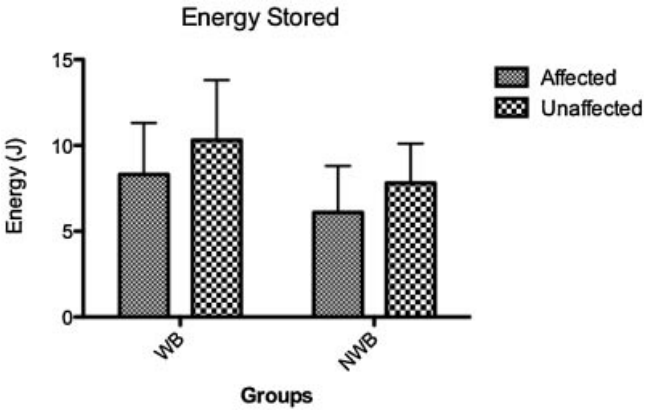

Figure 2. Energy stored across limbs and groups (WB: weight bearing; NWB: non-weight bearing). Significant main effects ($p < 0.05$) were observed across limbs and groups with no interaction effect.

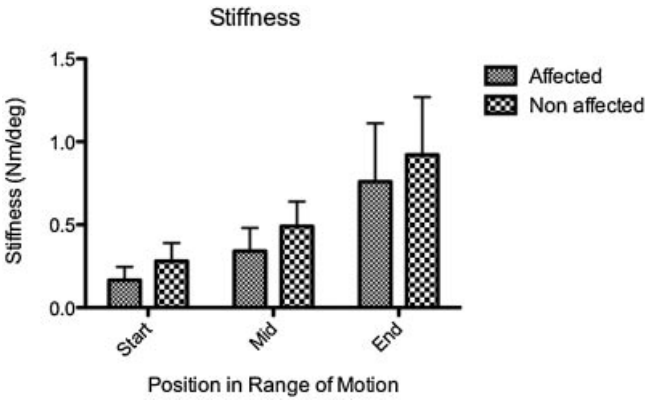

Figure 3. Stiffness across limbs at the start, mid range and end of range of motion tested. Significant main effects ($p < 0.05$) for limbs and position in the range of motion were observed.

torque and energy stored compared to those who were allocated to the non-weight bearing protocol. Irrespective of group allocation, differences across limbs were considerable in a number of variables, indicating that some mechanical parameters associated with muscle and tendon structure remain reduced at 6 months. Performance in physical tasks associated with recreational pursuits as measured by the Lower Limb Task Questionnaire indicated a need for continued rehabilitation. Performance in a number of these tasks could be influenced by the deficits in the biomechanical variables.

The reduced energy storage in the affected limbs of both groups may affect physical performance, particularly in activities that involve the stretch-shorten cycle.[25] Kubo demonstrated that the Achilles tendon contributed between 20% and 45% to the total amount of work in ankle joint exercises performed at different speeds, while Voigt et al.[26] showed that up to 60% of work associated with performing repetitive vertical jumps was generated from the tendon. Thus exercises should be instituted to remedy the deficit; previous research[27,28] showed that eccentric muscle work and plyometric exercises are beneficial. Furthermore, the reduced energy storage capacity of the affected limbs has ramifications for re-injury. Authors[8] have argued that a reduced capacity to store energy is a factor influencing the likelihood of a musculo-tendinous strain injury.

Stiffness values were similar to those in the literature.[29,30] The reduction in stiffness in the affected limb was notable, 60–80% of the unaffected limb. In patients treated surgically, Don et al. noted that the affected plantar flexors had increased stiffness at 3 and 6 months post-op, but this had reduced to values similar to the unaffected limb by 12 months. Similarly, in a longer follow up (1–5 years) by Bressel and McNair,[29] no difference in stiffness across limbs was observed in patients managed non-operatively. In our study, the difference across limbs was highest at the early and mid range of dorsiflexion. This part of the range of motion is often used in gait, and the findings suggest less ability to resist strain in the muscle and tendon (i.e., the musculo-tendinous complex is more compliant). This might reduce the likelihood of a strain induced injury[31]; however, in other instances this change may not be beneficial. For instance, in many gait activities contractile muscle elements would have to activate at higher levels to compensate for the stiffness loss in the overall musculo-tendinous unit. Don et al.[12] showed that changes in stiffness within the plantar flexors influenced the amount of dorsiflexion occurring in the latter part of the stance phase. In our study, the observed increased compliance would require increased activation in the plantar flexor muscles to maintain symmetrical motion with the opposite limb in the late stance phase. Such a strategy is inefficient metabolically, particularly as the passive mechanical properties significantly contribute to the torque production during walking.[32] It might also require an adjustment to the "central pattern generator's"[33,34] activation of the alpha motor neurons, as different signals would need to be transmitted from the neuron to the plantar flexors of each limb due to the stiffness asymmetry.

The reduced stiffness is likely closely linked to the decreased strength observed in the plantar flexors. While muscle cross-sectional area (CSA) was not measured in our study, muscle atrophy is a common feature related to strength deficits, and higher levels of muscle CSA are associated with higher levels of passive muscle stiffness.[35,36] Thus implementing a strength training program focused upon hypertrophy of these muscles would be valuable. Fry[37] suggested that loading levels of high intensity are most effective.

The observed strength deficit of ~18% was slightly less than the 23–26% reported by Moller et al.[38] and Keating and Will[39] at 6 months post-injury. Such deficits reflect processes associated with muscle atrophy and a reduction in neural activation. That the deficits were similar across both groups suggests that the early weight bearing protocol was insufficient to counter these processes. Previous work in animals showed that muscle atrophy can be reduced with earlier activity[40]; however, the increased activity experienced by animals is usually notably more than in humans where patients are still quite constrained in their ability to exercise, and hence loading levels sufficiently high for maintaining strength may not be achieved.

A positive finding was the lack of differences across limbs or across groups in shock absorption abilities of the passive plantar flexor muscles. Therefore, the intrinsic ability of these muscles to limit the joint and bone stresses associated with the transient impact forces that range between two and four bodyweights at foot strike during gait activities such as jogging seems unaffected.

Dorsiflexion range of motion had been restored to that of the unaffected side at 6 months, a finding

similar to Keating and Will.[39] While there was a significant difference across limbs with respect to plantar flexion range of motion in both groups, the mean difference was 3.5° (< 10%) which was similar again to Keating and Will[39] and is unlikely to be of clinical significance.

In conclusion, despite some positive effects as a result of the Bohler iron allowing earlier weight bearing and mobility for patients, we provide evidence that a number of biomechanical parameters important to function remain well below levels of those in the unaffected limb. These should be addressed through appropriate exercise regimes if patients are to return to sport safely and at pre-injury levels of performance.

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