DYNAMIC LOADING OF THE PLANTAR APONEUROSIS IN WALKING
Erdemir, Ahmet;Hamel, Andrew J;Fauth, Andrew R;Piazza, Stephen J;Sharkey, Neil A
Journal of Bone and Joint Surgery; Mar 2004; 86, 3; ProQuest
pg. 546

COPYRIGHT © 2004 BY THE JOURNAL OF BONE AND JOINT SURGERY, INCORPORATED

DYNAMIC LOADING OF THE PLANTAR APONEUROSIS IN WALKING

BY AHMET ERDEMIR, PHD, ANDREW J. HAMEL, PHD,
ANDREW R. FAUTH, MSC, STEPHEN J. PIAZZA, PHD, AND NEIL A. SHARKEY, PHD

Investigation performed at the Center for Locomotion Studies, Pennsylvania State University, University Park, Pennsylvania

Background: The plantar aponeurosis is known to be a major contributor to arch support, but its role in transferring Achilles tendon loads to the forefoot remains poorly understood. The goal of this study was to increase our understanding of the function of the plantar aponeurosis during gait. We specifically examined the plantar aponeurosis force pattern and its relationship to Achilles tendon forces during simulations of the stance phase of gait in a cadaver model.

Methods: Walking simulations were performed with seven cadaver feet. The movements of the foot and the ground reaction forces during the stance phase were reproduced by prescribing the kinematics of the proximal part of the tibia and applying forces to the tendons of extrinsic foot muscles. A fiberoptic cable was passed through the plantar aponeurosis perpendicular to its loading axis, and raw fiberoptic transducer output, tendon forces applied by the experimental setup, and ground reaction forces were simultaneously recorded during each simulation. A post-experiment calibration related fiberoptic output to plantar aponeurosis force, and linear regression analysis was used to characterize the relationship between Achilles tendon force and plantar aponeurosis tension.

Results: Plantar aponeurosis forces gradually increased during stance and peaked in late stance. Maximum tension averaged 96% ± 36% of body weight. There was a good correlation between plantar aponeurosis tension and Achilles tendon force ($r = 0.76$).

Conclusions: The plantar aponeurosis transmits large forces between the hindfoot and forefoot during the stance phase of gait. The varying pattern of plantar aponeurosis force and its relationship to Achilles tendon force demonstrates the importance of analyzing the function of the plantar aponeurosis throughout the stance phase of the gait cycle rather than in a static standing position.

Clinical Relevance: The plantar aponeurosis plays an important role in transmitting Achilles tendon forces to the forefoot in the latter part of the stance phase of walking. Surgical procedures that require the release of this structure may disturb this mechanism and thus compromise efficient propulsion.

The plantar aponeurosis originates from the medial tubercle of the calcaneus and inserts into the phalanges through a complex network of fibrous tissue. Hicks[1] described the function of the plantar aponeurosis as being analogous to a windlass mechanism, in which the arch of the foot elevates by winding of the plantar aponeurosis around the heads of the metatarsals during toe extension. Various studies of cadavera have revealed that release of the plantar aponeurosis decreases arch height, confirming the arch-supporting function of the plantar aponeurosis[2-4]. In a 4.5 to fifteen-year follow-up study of plantar fasciotomy, patients had postoperative flattening of the longitudinal arch and exhibited peak ground reaction forces that were lower than those of matched controls, particularly during push-off, indicating less efficient propulsion during walking[5]. Ker et al.[6] identified the plantar aponeurosis as a mechanism of energy storage in the foot, and the results of their in vitro experiment have been supported by subsequent computer models[7,8]. Bojsen-Moller and Lamoreux[9] suggested that another function of the plantar aponeurosis was to provide better cushioning against the high ground reaction forces occurring in late stance phase by tightening of the soft-tissue framework beneath the metatarsal heads.

The weight-bearing forces underneath the forefoot are well distributed across the metatarsal heads and phalanges when the plantar aponeurosis is intact[10-12]. Additionally, loading of the plantar aponeurosis by toe extension during the late stance phase stabilizes the transverse and longitudinal arches and transforms the foot into a rigid lever for efficient propulsion[1,13,14]. Positive correlations between plantar aponeurosis strain and Achilles tendon force and toe extension have been established under static loading conditions with the foot in a fixed position[15]. However, it is necessary to evaluate plantar aponeurosis loading during normal movements produced by muscle action in order to fully characterize its function and therefore help to explain the etiology of plantar fasciitis and the outcome of plantar fasciotomy.

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

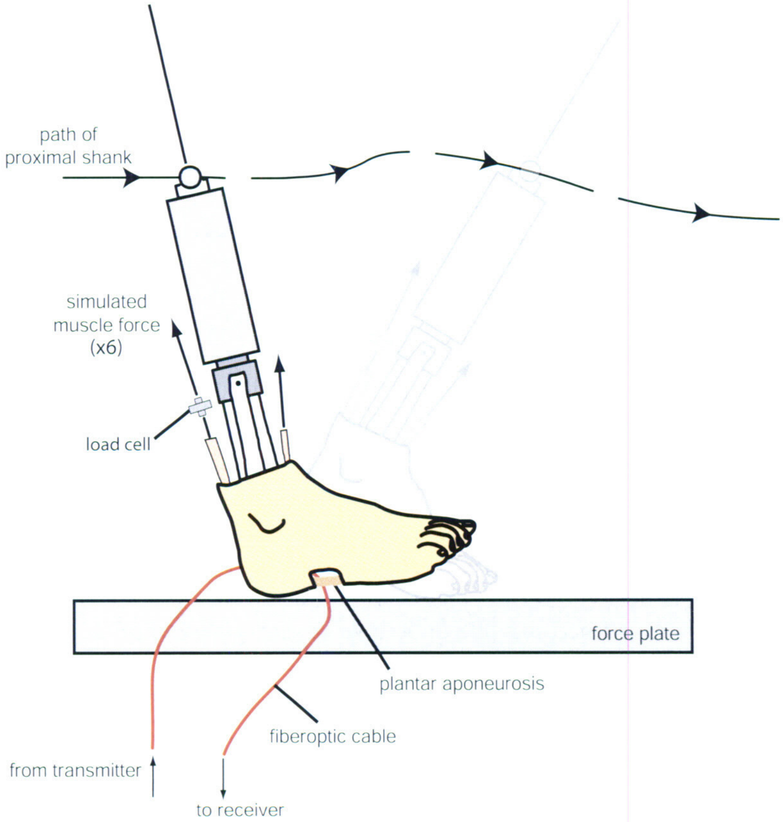

Fig. 1
The stance phase of walking was simulated by reproducing the normal movement of the proximal part of the tibia and body-weight-specific ground reaction forces. The dynamic gait simulator prescribed the path of the proximal part of the shank and the forces applied to the extrinsic tendons of six muscle groups: the triceps surae, tibialis posterior, flexor hallucis longus, flexor digitorum longus, peronei, and dorsiflexors (extensor hallucis longus, extensor digitorum longus, and tibialis anterior). In addition to fiberoptic light intensity, ground reaction forces were measured by a force plate and tendon forces were recorded with load cells attached in series.

The objectives of this study were to measure plantar aponeurosis force during simulations of the stance phase of walking in cadavera and to determine its relationship to Achilles tendon force under dynamic conditions. Our investigation differs from previous studies[2-4,11,12] in that the plantar aponeurosis was left intact and its function was investigated in situ under movements and forces approximating those that occur during normal walking.

Materials and Methods

A dynamic gait simulator prescribed translation of the proximal part of the tibia and applied dynamic forces to six tendons of extrinsic foot muscles to reproduce time-scaled (fifteen times slower) movements of the foot and ground reaction forces scaled to body weight that are characteristic of the stance phase of walking (Fig. 1)[16]. With use of linear actuators, physiological forces were applied to the tendons of six muscle

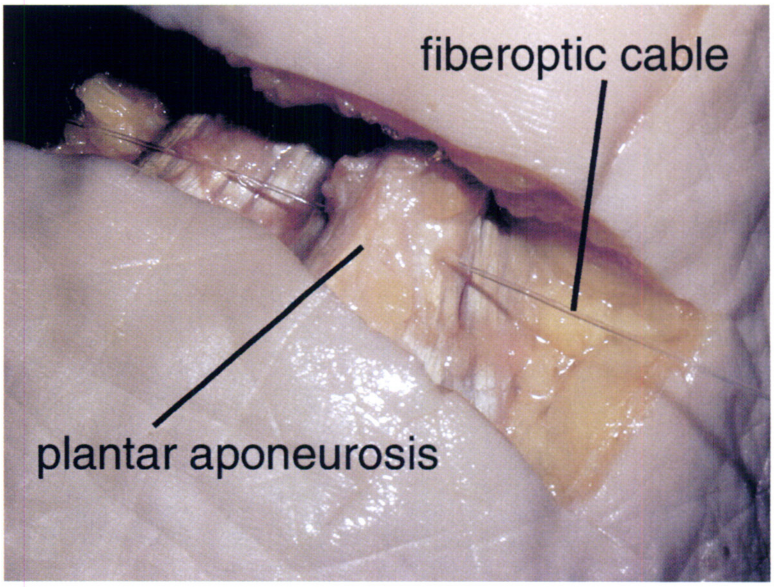

Fig. 2
An oblique plantar view of the exposure shows the insertion of the fiberoptic cable transverse to the loading axis of the proximal part of the plantar aponeurosis.

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

groups: the triceps surae, tibialis posterior, flexor hallucis longus, flexor digitorum longus, peronei, and dorsiflexors (extensor hallucis longus, extensor digitorum longus, and tibialis anterior). The loading profile of each muscle group was based on normalized electromyographic data from the stance phase of walking[17] and was constructed by assuming a linear relationship between electromyographic amplitude and muscle tension. Maximum voluntary contraction, which was used to normalize electromyographic data, was assumed to correspond to peak contractile tension within the muscle. Muscle tensions in the gait simulator were adjusted in order to obtain vertical ground reaction force profiles similar to those previously measured during walking. The operation of the gait simulator was described in detail by Sharkey and Hamel[16].

Seven nonpaired unembalmed cadaver feet from four male and three female donors (average age [and standard deviation] at time of death, 68.4 ± 22.5 years; range, twenty-three to ninety-three years) were tested. The body masses were unknown; therefore, each specimen was assigned a body mass based on age, gender, and foot size for simulation purposes (average and standard deviation, 57 ± 10 kg). A 2-cm-wide strip of superficial plantar tissue located approximately 2 cm anterior to the heel was removed to expose the proximal part of the plantar aponeurosis near its origin. A 0.5-mm-diameter fiberoptic cable (PGR500; Toray Industries, Tokyo, Japan) was passed through the plantar aponeurosis perpendicular to its longitudinal axis for the purpose of measuring the force it carried (Fig. 2). The ends of the cable were then attached to a transmitter-receiver unit (HFBR 1414T and 2416T; Agilent Technologies, Palo Alto, California). Tension in the plantar aponeurosis compressed the fiberoptic cable, thereby decreasing the intensity of transmitted light[18-20]. During each trial, the Achilles tendon forces as well as the remaining muscle forces applied by the gait simulator were recorded with use of uniaxial load cells (Omega Engineering, Stamford, Connecticut) and ground reaction forces were recorded with use of a triaxial force plate (Advanced Mechanical Technology, Watertown, Massachusetts). After at least ten setup and conditioning trials, three experimental trials were performed, during which data were captured at 30 Hz.

Following gait simulations, the fiberoptic transducer was calibrated in situ by directly loading the extracted plantar aponeurosis complex with use of a materials testing device (858 Mini Bionix, MTS Systems, Eden Prairie, Minnesota) in order to determine the relationship between plantar aponeurosis force and fiberoptic light intensity. Calibration was accomplished by first placing each intact specimen in a fixture that aligned the longitudinal axis of the plantar aponeurosis with the loading axis of the testing device. The calcaneus and

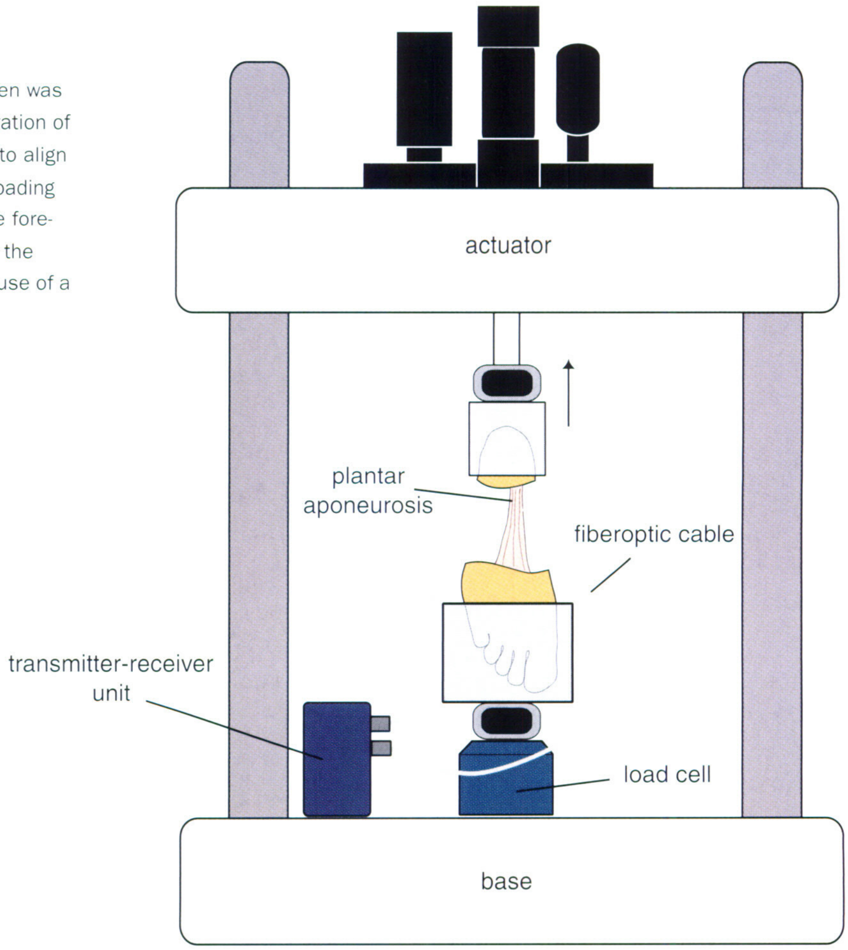

Fig. 3
Following division of the midfoot structures, the specimen was mounted in the materials testing device for in situ calibration of the fiberoptic measurement system. Guides were used to align the longitudinal axis of the plantar aponeurosis to the loading axis of the materials testing device during fixation of the forefoot and hindfoot into aluminum fixtures. Axial forces in the plantar aponeurosis were simultaneously recorded with use of a load cell.

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

the forefoot were individually potted with polymethylmethacrylate (GC America, Alsip, Illinois) into either end of the fixture, and all other soft-tissue structures except the plantar aponeurosis were severed at the midfoot-hindfoot articulations. Use of this fixture allowed anatomical orientation of the plantar aponeurosis with respect to the calcaneus and forefoot during calibration. The specimen was then placed in the materials testing device (Fig. 3), and the plantar aponeurosis was ramp loaded and unloaded for ten cycles to a peak force of 1000 N at 100 N/s. Raw fiberoptic output and tension in the plantar aponeurosis were recorded simultaneously. The peak force was then increased by 250 N, and cyclic loading was repeated. This process was iterated until plantar aponeurosis or fixture failure was observed. The procedure provided a range of calibration data that included the range of fiberoptic output experienced in the walking simulations. The calibration loading rate was set at 100 N/s to approximate the loading rates of the in situ trials. Calibration constants for each specimen were calculated by fitting a polynomial to the plantar aponeurosis tension versus raw fiberoptic output of calibration trials. A first-order polynomial denotes a linear fit, whereas higher-order polynomials can be used to model nonlinear data. As a result of nonlinearities inherent in the fiberoptic measurement technique[18], an iterative process was employed to obtain calibration constants. First, a linear regression (first-order polynomial) was completed and the quality was determined by evaluating the root-mean-squared error of the fit. Subsequently, polynomials of increasing order were fitted to the data until the decrease in root-mean-squared error between sequential polynomial fits became lower than 5 N.

Specimen-specific calibration constants were used to calculate plantar aponeurosis force for each of the three experimental trials for each specimen. The average of these three trials provided the force profile of the plantar aponeurosis for each specimen. Specimen data were later pooled, and a linear regression analysis was used to quantify the relationship between Achilles tendon force and plantar aponeurosis tension. The significance level (α) was set at 0.05. Achilles tendon forces applied by the gait simulator, ground reaction forces, and plantar aponeurosis loads were normalized to estimated body weight for the analysis.

Results

The mean peak vertical ground reaction force (and standard deviation) during the gait simulations was 590 ± 108 N (1.05 ± 0.04 times estimated body weight; Fig. 4). The mean peak Achilles tendon force applied by the gait simulator was 1041 ± 213 N (1.86 ± 0.18 times estimated body weight; Fig. 4). The mean peak load was 94 ± 21 N for the tibialis posterior, 86 ± 16 N for the flexor hallucis longus, 28 ± 4 N for the

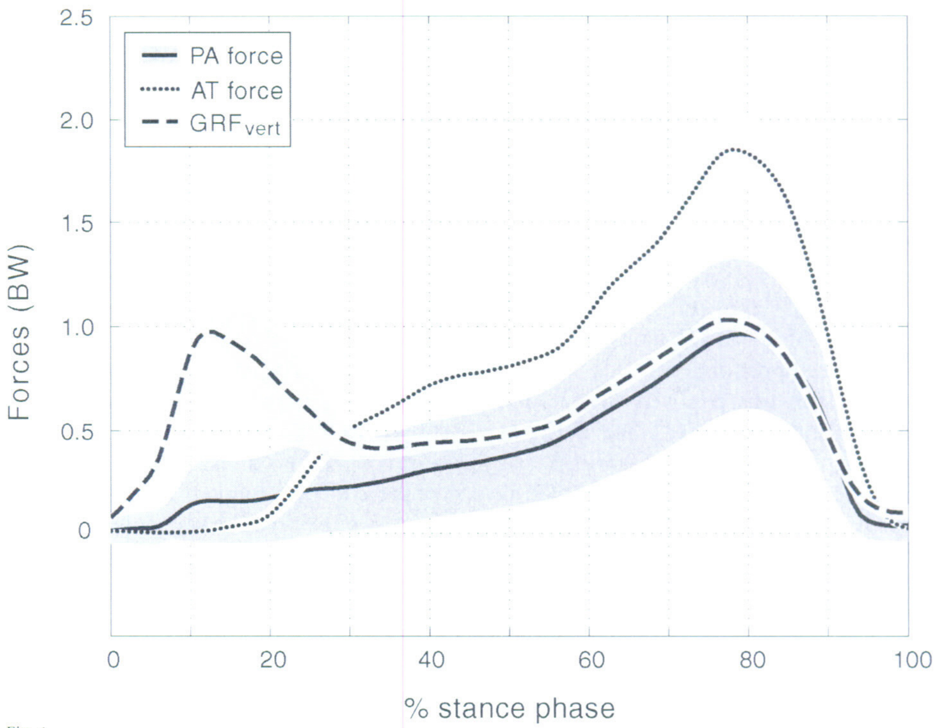

Fig. 4

Mean plantar aponeurosis (PA) forces recorded by the fiberoptic transducer, Achilles tendon (AT) forces applied by the gait simulator, and vertical ground reaction forces (GRF_{vert}) recorded by the force plate, all normalized to estimated body weight (BW). Shading represents the standard deviation. Plantar aponeurosis tension gradually increased during stance phase, reaching peak values at the start of push-off.

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

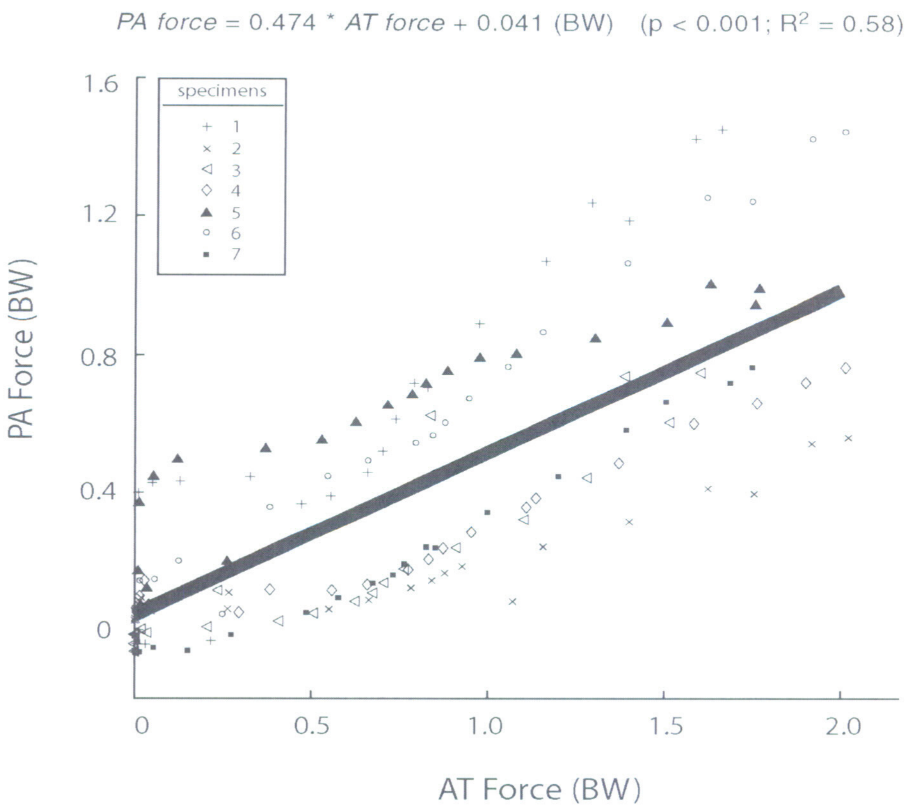

Fig. 5

Achilles tendon (AT) force was found to be an effective predictor of plantar aponeurosis (PA) tension ($r = 0.76$). Linear regression analysis revealed plantar aponeurosis force as a function of Achilles tendon force (see equation). Both Achilles tendon and plantar aponeurosis forces were normalized to estimated body weight (BW).

flexor digitorum longus, 69 ± 14 N for the peronei, and 115 ± 19 N for the dorsiflexors.

Nonlinear calibrations were required to model specimen-specific calibration data as denoted by polynomials of orders two to four (see Appendix). Plantar aponeurosis failure loads were observed to be in the range of 916 to 1743 N (2.15 to 2.80 times estimated body weight; see Appendix).

Plantar aponeurosis tension was relatively low at heel strike; gradually increased during midstance, when both the heel and the forefoot were in contact with the ground; and peaked at approximately 80% of stance phase (Fig. 4). Peak plantar aponeurosis forces during simulated walking were 538 ± 193 N (0.96 ± 0.36 times estimated body weight).

Linear regression analysis revealed that Achilles tendon force and plantar aponeurosis tension were well correlated (Fig. 5, $r = 0.76$). Plantar aponeurosis force as a function of Achilles tendon force is given by the following equation, with the coefficient of Achilles tendon force having a standard deviation of 0.015: plantar aponeurosis force = 0.474 * Achilles tendon force + 0.041 (body weight) ($p < 0.001$).

Discussion

Our investigation differs from previous studies of the mechanics of the plantar aponeurosis in that the forces in this structure were measured in situ during lifelike movement and loading conditions. Previous studies of cadavera have been mostly quasi-static experiments performed with the foot in a standing posture[2,4,21,22] or at specific stages of the stance phase of gait[3,12], although some plantar fasciotomy studies have involved simulation of walking[11,23] or running[6]. The load-bearing role of the plantar aponeurosis has also been confirmed by computer models used to investigate arch support[7,24], energy storage[8], and stress distribution[25] after partial or full release of the plantar aponeurosis and by a finite element model of calcaneal loading[26]. In the present study, we used a cadaver model of the stance phase of gait that prescribed the knee trajectory and muscle forces to provide a reasonable estimate of the magnitude of plantar aponeurosis forces in relation to Achilles tendon loading.

Giddings et al.[26] created a finite element model of the foot to investigate calcaneal loading and predicted plantar aponeurosis forces of 1.8 times body weight during walking and 3.7 times body weight during running. Their predictions seem to overestimate the forces in the plantar aponeurosis when compared with the peak loads measured in our study (0.96 times body weight) and with the failure loads observed during our calibrations (see Appendix) and those reported by Kitaoka et al.[27]. We believe that the discrepancy in the magnitude of the plantar aponeurosis load between our study and

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

theirs was caused by the linear spring model of the plantar aponeurosis in their analysis and a possible overestimation of its elastic modulus. These differences aside, the temporal loading patterns and the ratio of plantar aponeurosis tension to Achilles force were similar between the studies.

Several in vitro studies have shown the strain within the plantar aponeurosis to be up to 7% with the foot loaded in a standing posture[28,29] and when combined with Achilles loading and metatarsophalangeal extension[15]. The loading profile of the plantar aponeurosis during midstance, with the foot axially loaded, is consistent with the strain measurements reported by Kogler et al.[28,29]. In addition, our finding that Achilles tendon force is a reasonable predictor of plantar aponeurosis tension (Fig. 5) agrees with the findings of Carlson et al.[15] concerning the relationship between Achilles tendon force and plantar aponeurosis strain as measured with a strain-gauged extensometer.

Certain limitations are associated with the gait simulator and with the calibration of the fiberoptic system used in our study. The body mass of each specimen was estimated in order to complete simulations in the gait simulator[16]. Vertical ground reaction forces simulated by the gait simulator during midstance were approximately 50% of the estimated body weight (Fig. 4), which was less than the 80% of body weight that is typically measured for normal subjects[30]. This limitation in the reproduction of ground reaction forces may have resulted in plantar aponeurosis forces being underestimated during the midstance phase of the cadaver simulations as a result of reduced foot loading. Measured peak tensions in the plantar aponeurosis, however, would not be affected by this underestimation because these recordings occurred at approximately 80% of the stance phase, during which time the ground reaction forces were accurately reproduced.

Plantar aponeurosis forces were measured with use of a fiberoptic technique[18,20]. Other possible ways to measure these forces include the use of transducers in which the tissue of interest is passed through a buckle-shaped device[31-33] and the use of implantable transducers that can be placed inside the tissue[34-36]. Each of these transducers operates in a similar fashion: axial loading of the tissue produces deformation of the implanted structure, which is measured with strain gauges, and a pre-experiment or post-experiment calibration test relates the signal to axial force. Shortening of the tissue and impingement of the transducer on the neighboring tissues are potential disadvantages of buckle transducers[37]. Differences in orientation and alignment between the calibration and the experiment may adversely affect the implantable transducers[34,38,39]. The fiberoptic transducer, however, is easy to place, is inexpensive, and only minimally disrupts the integrity of the tissue (fiberoptic cable diameter <1 mm). The response of the fiberoptic transducer is nonlinear, possibly as a result of the nonlinearities in plantar aponeurosis-fiberoptic cable interaction, and loading rate discrepancies between walking simulations and calibration may influence predictions of plantar aponeurosis force[18]. Second to fourth-order polynomials described the calibration data to overcome the nonlinearities. To avoid errors in the fiberoptic data due to extrapolation, calibration forces encompassed the full range of plantar aponeurosis loading observed in the gait simulator. Loading rates of experimental trials and calibration were also matched to reduce errors. Nonuniform loading of the plantar aponeurosis during walking simulations or calibration might also induce additional measurement errors that would remain unaccounted for in our approach. To minimize these errors, a custom-made fixture was used to mount the forefoot and calcaneus to orient the plantar aponeurosis along the loading axis of the materials testing device, which prevented off-axis loading during calibration and also provided anatomical orientation of the plantar aponeurosis relative to the calcaneus and forefoot. The fiberoptic cable was passed through the plantar aponeurosis at a level closer to its calcaneal insertion to minimize the effects of twisting of this soft-tissue structure.

Peak plantar aponeurosis force during walking was found to occur in late stance (Fig. 4), and the patterns of plantar aponeurosis correlated well with Achilles tendon loading (Fig. 5). These measurements suggest that this structure is critical for transmission of forces to the forefoot. Our findings reveal the cause of imbalance in the distribution of ground reaction forces underneath the metatarsal heads and toes after complete plantar fasciotomy[11,12]. Loss of the pull of the plantar aponeurosis on the phalanges, particularly in late stance, unloads the toes and shifts the ground reaction forces to beneath the metatarsal heads[11,12].

We believe that simulations of normal walking and other dynamic activities are necessary for a complete understanding of plantar aponeurosis function because the plantar aponeurosis is substantially loaded late in stance phase by contraction of the plantar flexors. Studies that lack lifelike movements and muscle forces may result in definition of the plantar aponeurosis solely as a passive arch-supporting element during standing rather than as a structure that dynamically transfers loads during movement.

Appendix

eA A table showing the calibration test data is available with the electronic versions of this article, on our web site at www.jbjs.org (go to the article citation and click on "Supplementary Material") and on our quarterly CD-ROM (call our subscription department, at 781-449-9780, to order the CD-ROM). ■

Ahmet Erdemir, PhD
Department of Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, OH 44195

Andrew J. Hamel, PhD
Stryker Endoscopy, 5900 Optical Court, San Jose, CA 95138
Andrew R. Fauth, MSc
Stephen J. Piazza, PhD
Neil A. Sharkey, PhD
Center for Locomotion Studies (A.R.F., S.J.P., and N.A.S.), Department of Kinesiology (A.R.F., S.J.P., and N.A.S.), Department of Mechanical and Nuclear Engineering (S.J.P.), and Department of Orthopaedics and Rehabilitation (S.J.P. and N.A.S.), Pennsylvania State University, 29 Recreation Building, University Park, PA 16802-5702. E-mail address for S.J. Piazza: steve-piazza@psu.edu

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.

In support of their research or preparation of this manuscript, one or more of the authors received grants or outside funding from the International Society of Biomechanics and the Irma and Harold Zipser Fellowship. None of the authors received payments or other benefits or a commitment or agreement to provide such benefits from a commercial entity. No commercial entity paid or directed, or agreed to pay or direct, any benefits to any research fund, foundation, educational institution, or other charitable or nonprofit organization with which the authors are affiliated or associated.

References

1. **Hicks JH.** The mechanics of the foot. II. The plantar aponeurosis and the arch. *J Anat.* 1954;88:25-30.
2. **Kitaoka HB, Luo ZP, An KN.** Mechanical behavior of the foot and ankle after plantar fascia release in the unstable foot. *Foot Ankle Int.* 1997; 18:8-15.
3. **Sharkey NA, Ferris L, Donahue SW.** Biomechanical consequences of plantar fascial release or rupture during gait: part I—disruptions in longitudinal arch conformation. *Foot Ankle Int.* 1998;19:812-20.
4. **Thordarson DB, Kumar PJ, Hedman TP, Ebramzadeh E.** Effect of partial versus complete plantar fasciotomy on the windlass mechanism. *Foot Ankle Int.* 1997;18:16-20.
5. **Daly PJ, Kitaoka HB, Chao EY.** Plantar fasciotomy for intractable plantar fasciitis: clinical results and biomechanical evaluation. *Foot Ankle.* 1992; 13:188-95.
6. **Ker RF, Bennett MB, Bibby SR, Kester RC, Alexander RM.** The spring in the arch of the human foot. *Nature.* 1987;325:147-9.
7. **Kim W, Voloshin AS.** Role of plantar fascia in the load bearing capacity of the human foot. *J Biomech.* 1995;28:1025-33.
8. **Simkin A, Leichter I.** Role of the calcaneal inclination in the energy storage capacity of the human foot—a biomechanical model. *Med Biol Eng Comput.* 1990;28:149-52.
9. **Bojsen-Moller F, Lamoreux L.** Significance of free-dorsiflexion of the toes in walking. *Acta Orthop Scand.* 1979;50:471-9.
10. **Hedrick MR.** The plantar aponeurosis. *Foot Ankle Int.* 1996;17:646-9.
11. **Hamel AJ, Donahue SW, Sharkey NA.** Contributions of active and passive toe flexion to forefoot loading. *Clin Orthop.* 2001;393:326-34.
12. **Sharkey NA, Donahue SW, Ferris L.** Biomechanical consequences of plantar fascial release or rupture during gait. Part II: alterations in forefoot loading. *Foot Ankle Int.* 1999;20:86-96.
13. **Sarrafian SK.** Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading. *Foot Ankle.* 1987;8:4-18.
14. **Mann RA, Hagy JL.** The function of the toes in walking, jogging and running. *Clin Orthop.* 1979;142:24-9.
15. **Carlson RE, Fleming LL, Hutton WC.** The biomechanical relationship between the tendoachilles, plantar fascia and metatarsophalangeal joint dorsiflexion angle. *Foot Ankle Int.* 2000;21:18-25.
16. **Sharkey NA, Hamel AJ.** A dynamic cadaver model of the stance phase of gait: performance characteristics and kinetic validation. *Clin Biomech (Bristol, Avon).* 1998;13:420-33.
17. **Perry J.** *Gait analysis: normal and pathological function.* Thorofare, NJ: Slack; 1992.
18. **Erdemir A, Piazza SJ, Sharkey NA.** Influence of loading rate and cable migration on fiberoptic measurement of tendon force. *J Biomech.* 2002; 35:857-62.
19. **Erdemir A, Hamel AJ, Piazza SJ, Sharkey NA.** Fiberoptic measurement of tendon forces is influenced by skin movement artifact. *J Biomech.* 2003;36:449-55.
20. **Komi PV, Belli A, Huttunen V, Bonnefoy R, Geyssant A, Lacour JR.** Optic fibre as a transducer of tendomuscular forces. *Eur J Appl Physiol Occup Physiol.* 1996;72:278-80.
21. **Murphy GA, Pneumaticos SG, Kamaric E, Noble PC, Trevino SG, Baxter DE.** Biomechanical consequences of sequential plantar fascia release. *Foot Ankle Int.* 1998;19:149-52.
22. **Thordarson DB, Schmotzer H, Chon J, Peters J.** Dynamic support of the human longitudinal arch. A biomechanical evaluation. *Clin Orthop.* 1995; 316:165-72.
23. **Donahue SW, Sharkey NA.** Strains in the metatarsals during the stance phase of gait: implications for stress fractures. *J Bone Joint Surg Am.* 1999; 81:1236-44.
24. **Arangio GA, Chen C, Salathe EP.** Effect of varying arch height with and without the plantar fascia on the mechanical properties of the foot. *Foot Ankle Int.* 1998;19:705-9.
25. **Gefen A.** Stress analysis of the standing foot following surgical plantar fascia release. *J Biomech.* 2002;35:629-37.
26. **Giddings VL, Beaupre GS, Whalen RT, Carter DR.** Calcaneal loading during walking and running. *Med Sci Sports Exerc.* 2000;32:627-34.
27. **Kitaoka HB, Luo ZP, Growney ES, Berglund LJ, An KN.** Material properties of the plantar aponeurosis. *Foot Ankle Int.* 1994;15:557-60.
28. **Kogler GF, Veer FB, Solomonidis SE, Paul JP.** The influence of medial and lateral placement of orthotic wedges on loading of the plantar aponeurosis. *J Bone Joint Surg Am.* 1999;81:1403-13.
29. **Kogler GF, Veer FB, Verhulst SJ, Solomonidis SE, Paul JP.** The effect of heel elevation on strain within the plantar aponeurosis: in vitro study. *Foot Ankle Int.* 2001;22:433-9.
30. **Chao EY, Laughman RK, Schneider E, Stauffer RN.** Normative data of knee joint motion and ground reaction forces in adult level walking. *J Biomech.* 1983;16:219-33.
31. **Dennerlein JT, Miller JM, Mote CD Jr, Rempel DM.** A low profile human tendon force transducer: the influence of tendon thickness on calibration. *J Biomech.* 1997;30:395-7.
32. **Komi PV, Salonen M, Jarvinen M, Kokko O.** In vivo registration of Achilles tendon forces in man. I. Methodological development. *Int J Sports Med.* 1987;8(Suppl 1):3-8.
33. **An KN, Berglund L, Cooney WP, Chao EY, Kovacevic N.** Direct in vivo tendon force measurement system. *J Biomech.* 1990;23:1269-71.
34. **Hall GW, Crandall JR, Carmines DV, Hale JE.** Rate-independent characteristics of an arthroscopically implantable force probe in the human achilles tendon. *J Biomech.* 1999;32:203-7.
35. **Fleming BC, Peura GD, Beynnon BD.** Factors influencing the output of an implantable force transducer. *J Biomech.* 2000;33:889-93.
36. **Xu WS, Butler DL, Stouffer DC, Grood ES, Glos DL.** Theoretical analysis of an implantable force transducer for tendon and ligament structures. *J Biomech Eng.* 1992;114:170-7.
37. **Lewis JL, Lew WD, Schmidt J.** A note on the application and evaluation of the buckle transducer for the knee ligament force measurement. *J Biomech Eng.* 1982;104:125-8.
38. **Herzog W, Archambault JM, Leonard TR, Nguyen HK.** Evaluation of the implantable force transducer for chronic tendon-force recordings. *J Biomech.* 1996;29:103-9.
39. **Fleming BC, Good L, Peura GD, Beynnon BD.** Calibration and application of an intra-articular force transducer for the measurement of patellar tendon graft forces: an in situ evaluation. *J Biomech Eng.* 1999;121:393-8.

Reproduced with permission of the copyright owner. Further reproduction prohibited without permission.