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Posterior Tibial Tendon Dysfunction and Flatfoot: Analysis with Simulated Walking

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Abstract

Many biomechanical studies investigated pathology of flatfoot and effects of operations on flatfoot. The majority of cadaveric studies are limited to the quasistatic response to static joint loads. This study examined the unconstrained joint motion of the foot and ankle during stance phase utilizing a dynamic foot-ankle simulator in simulated stage 2 posterior tibial tendon dysfunction (PTTD). Muscle forces were applied on the extrinsic tendons of the foot using six servo-pneumatic cylinders to simulate their action. Vertical and fore-aft shear forces were applied and tibial advancement was performed with the servomotors. Three-dimensional movements of multiple bones of the foot were monitored with a magnetic tracking system. Twenty-two fresh-frozen lower extremities were studied in the intact condition, then following sectioning peritalar constraints to create a flatfoot and unloading the posterior tibial muscle force. Kinematics in the intact condition were consistent with gait analysis data for normals. There were altered kinematics in the flatfoot condition, particularly in coronal and transverse planes. Calcaneal eversion relative to the tibia averaged $11.1 \pm 2.8^\circ$ compared to $5.8 \pm 2.3^\circ$ in the normal condition. Calcaneal-tibial external rotation was significantly increased in flatfeet from mean of $2.3 \pm 1.7^\circ$ to $8.1 \pm 4.0^\circ$. There were also significant changes in metatarsal-tibial eversion and external rotation in the flatfoot condition. The simulated PTTD with flatfoot was consistent with previous data obtained in patients with PTTD. The use of a flatfoot model will enable more detailed study on the flatfoot condition and/or effect of surgical treatment.

Keywords

Flatfoot; Kinematics; Simulator; Cadaver; Gait Simulation

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CONFLICT OF INTEREST STATEMENT

The authors have no conflicts to report.

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INTRODUCTION

In recent years, there have been multiple reports related to posterior tibial tendon dysfunction (PTTD). PTTD is a common cause of acquired flatfoot deformity in adults; the condition may reach a prevalence of 10 % in elderly women.⁽¹⁾ Most of the clinical investigations reported the effects of various operations to correct PTTD and flatfoot deformity, but *in vitro* and radiologic studies have also been published.^(2–5) The optimum management of stage 2 PTTD⁽²⁾ in which there is a mobile flatfoot has been a subject of debate for more than two decades. There are questions regarding the pathoanatomy of the deformity resulting from tendon dysfunction. A comprehensive understanding of the flatfoot malalignment will lead to more effective techniques for correcting it.

Previous investigators recognized the importance of defining the flatfoot and quantitating the degree of deformation.⁽⁶⁾ Clinical measurements such as arch height have been used, but were not consistent between examiners.⁽⁶⁾ Radiologic measurements of flatfoot have been performed in patients with PTTD.^(4,5) Analysis of foot prints and ground reaction data in flatfeet have been reported.⁽⁷⁾ Previous reports determined the contribution of various static elements in supporting the arch.⁽⁸⁾ Others indicated that the posterior tibial tendon (PTT) plays a role in the dynamic support of the arch.⁽³⁾ Most of these previous reports were *in vitro* studies with static loading of specimens. Recent gait analysis studies have revealed kinematics changes of the foot between normal and flatfoot.^(9–12) *In vitro*, dynamic joint simulators have been developed recently for foot/ankle as well as other joints and applied for biomechanical studies of pathological situations of foot/ankle problems.^(13–19) However, there have not been any publications, to our knowledge, detailing the three-dimensional kinematics of the flatfoot during simulated walking of the entire stance phase of gait.

We developed a dynamic foot-ankle simulator capable of recreating the stance phase of gait in cadaveric lower extremities.⁽¹⁵⁾ It allows simulation of functional activities by allowing unconstrained motion of the foot and ankle while simultaneously applying time-histories of forces that affect the foot. A dynamic simulator will be useful for improving our understanding of the mechanical behavior of the flatfoot compared to the normal foot. The purpose of the study was to examine unconstrained joint motion of the foot and ankle during stance phase in cadaveric lower extremities with simulated PTTD with flatfoot deformity utilizing a dynamic foot-ankle simulator.

METHODS

Twenty-two fresh-frozen lower extremities were evaluated after gross visual screening for preexisting abnormalities of the foot. The mean age of the specimens was 78 years old (range, 47 to 95). Five were female, 17 male. Fourteen specimens were left feet and eight were right. The institutional research ethics committee reviewed and approved the study.

The custom simulator was able to subject cadaveric foot and ankle specimens to specified time-histories of normative ground reaction forces (GRF), and tendon loading based on physiological cross-sectional area (PCSA) and electromyography (EMG) data. Adjustments were made the initial muscle force profiles to match normal patterns of calcaneal-tibial angles and gross forefoot motion, while allowing totally unconstrained joint motions. The inputs were the forward tibial motion, fore-aft shear loading, tibial loading, and muscle loading. The outputs were measured foot reaction (force and center-of-pressure advancement) and joint kinematics. Since few biomechanical models were available for this hybrid (load and motion) control system, the tibial loading was prescribed to the normal vertical GRF profile as an approximation using a closed-loop servomotor so as to match the normal vertical GRF despite the disturbances from the muscle loading actuators. Ankle joint

kinematics was controlled by muscle loading, which was initially targeted from the muscle gain, PCSA, and EMG data for each muscle group.

The leg specimen, amputated at the mid-tibial level, was potted in an acrylic plastic tube with polymethylmethacrylate cement, and fixed on the tendon loading unit (Figure 1).⁽¹⁵⁾ The tendon-loading unit, consisting of six pneumatic cylinders, was connected through a linear ball screw to the servomotor tibial angle control unit, and through a hinge joint to the vertical loading unit. All the above units created a four-bar rocker configuration to control the angle of the loading unit or tibia relative to the ankle joint during the simulation. The vertical-loading unit, mounted on the frame, was connected to a linear slide powered by a linear ball screw and servomotor. This servomotor axis provided loading along the tibial axis by moving both the vertical loading unit and tendon loading unit onto the force plate. The sole of the foot contacted the custom-made force plate, which recorded the simulated vertical GRF. The bearing plate beneath the force plate allowed the relative forward translation of the foot and ankle and accommodated the anteroposterior shear force⁽²⁰⁾ by coupling the plate to another closed-loop servo axis. Most structures were built using non-metallic tubing or acrylic plastics, to minimize any interference with the magnetic tracking system.

The tibial angle control unit translated the tibial motion between 20° anterior and 40° posterior to vertical.⁽²¹⁾ During testing, the tibial angle was monitored by a potentiometer. The tibial angle was used to determine the percentage of stance phase from the tibial angle history profile.⁽²¹⁾ The estimated percent stance phase was used to prescribe the tendon and tibial loading.^(20,21) The extrinsic muscles acting on or around the ankle were divided into six functional units: gastrocnemius-soleus, PTT, flexor hallucis longus-flexor digitorum longus, anterior tibial, extensor hallucis longus-extensor digitorum longus, and peroneus longus-peroneus brevis. A total of six pneumatic cylinders driven by servo-pneumatic valves were used to apply loads to the tendons. Sutures were attached to each tendon using a modified Krackow technique except for the Achilles tendon, which was sutured using a Leeds-Keio artificial ligament (Ellis Developments, Nottingham, UK) to sustain the comparable physiological loading. A uniaxial load cell was connected to each cylinder to monitor the load feedback signal to an IBM PC. A custom-written program (LabVIEW®, National Instruments, Austin, TX) was used for control and kinetic data acquisition. Tendon loads were estimated from PCSA and EMG data in the literature.^(20, 22) A linear EMG and force relationship was assumed.^(23,24) An unknown muscle gain K and the corresponding cross sectional area ($PCSA_i$) were multiplied by the relative EMG data (EMG_i) to provide absolute forces to each cylinder ($F_i = K * PCSA_i * EMG_i$). To determine the unknown muscle gain or muscle stress, simulations were repeated until the gross foot-ankle kinematics matched reasonably well to the stance phase center-of-pressure and motion patterns, while the gain of the muscles was simultaneously adjusted. The applied vertical and fore-aft loads were reduced for the cadaver feet due to the limited suture strength and age of the specimens.

Each specimen was cycled additional times through the entire stance phase to reduce the viscoelastic effect of soft tissues. The leg was then continuously moved from tibial flexion -20° (initial contact) to 40° flexion (pre-swing) while applying forces to the six muscle groups and subjecting it to the ground reaction force profiles. After testing the intact condition, a flatfoot condition was created by sectioning the peritalar soft tissue structures, including the spring ligament, long and short plantar ligaments, talocalcaneal interosseous ligament, medial talocalcaneal ligament, and tibionavicular portion of the superficial deltoid ligament. In order to simulate stage 2 PTTD and flatfoot, the posterior tibial muscle action was not simulated. A previous study⁽⁸⁾ showed that flatfoot deformity occurred after these procedures, similar to that caused by stage 2 PTTD clinically. The applied vertical and fore-

aft loads were not changed before and after the flatfoot model was created. Testing was performed 3 times for each condition and the average result was used for analysis.

Three-dimensional movements of the calcaneus and first metatarsus relative to the tibia were obtained with a magnetic tracking system (3Space Tracker System, Polhemus, Colchester, VT).^(25,26) The electromagnetic transmitter was fixed to the loading frame. Sensors were rigidly fixed to the anteromedial tibial diaphysis at the junction of the middle and distal thirds, posterolateral calcaneal body, and diaphysis of the dorsal-medial first metatarsal with acrylic mounting posts.⁽²⁶⁾ The relative angular motion between bones (calcaneal-tibial and first metatarsal-tibial) was expressed in terms of Eulerian angle description.^(25,26) Bone positions were described relative to the previously-described coordinate system.⁽²⁶⁾ The x axis was oriented proximal-distal, the y axis was oriented medial-lateral, and the z axis was oriented anterior-posterior. The x axis was along the tibial shaft through the ankle center, the z axis was parallel to the projection of a line connecting the center of the heel and the second metatarsal on a plane perpendicular to the x axis passing through the ankle center, and the y axis was the product of the x and z axes following the right-hand rule passing through the ankle center. The three axes were also used to define the perpendicular planes: coronal (x–y plane), sagittal (x–z plane), and transverse (y–z plane). Motion was defined as inversion and eversion in the coronal plane, dorsiflexion and plantarflexion in the sagittal plane, and internal rotation and external rotation in the transverse plane.

Statistical analysis was performed with a paired t-test, comparing the bony positions in the intact and flatfoot conditions. The level of significance was set at $p < 0.05$.

RESULTS

All the results of angular motions of the calcaneus and the first metatarsal relative to the tibia are presented over an entire stance phase that started with heel strike and ended with toe-off (Figure 2). Angular movement patterns were consistent across all specimens. In the sagittal plane, the forefoot contacted with the force plate at about 10% stance phase. Maximum dorsiflexion was observed at about 80% stance phase. Then rapid plantarflexion occurred until maximum plantarflexion at toe-off. In the coronal plane, the hindfoot (calcaneal-tibial) and forefoot (metatarsal-tibial) moved from an initial inversion at heel strike to eversion. Maximum eversion occurred at a late phase of stance, prior to heel rise. Then the foot progressively inverted until toe-off. In the transverse plane, the hindfoot and forefoot moved from an initial internal rotation to external rotation. At mid-stance the foot reversed and began to internal rotate until toe-off.

In the flatfoot, there was increased eversion and external rotation, but not dorsiflexion. The coronal plane motion curves demonstrated the characteristic inversion in late stance was absent in the flatfoot, in calcaneal-tibial and metatarsal-tibial motions. The transverse plane motions were also abnormal, as there was increased external rotation of the ankle-hindfoot (calcaneal-tibial) and forefoot (metatarsal-tibial). The typical internal rotation in late stance was not present in the flatfeet.

Calcaneal-tibial maximum dorsiflexion was $14.1 \pm 4.1^\circ$ (mean $\pm$ SD). In the flatfoot, maximum dorsiflexion was $14.4 \pm 4.4^\circ$ (not significant: NS). Calcaneal-tibial maximum eversion was $5.8 \pm 2.3^\circ$. In the flatfoot, maximum eversion was $11.1 \pm 2.8^\circ$ (significant). Calcaneal-tibial maximum external rotation was $2.3 \pm 1.7^\circ$. In the flatfoot, maximum external rotation was $8.1 \pm 4.0^\circ$ (significant) (Table 1).

Metatarsal-tibial maximum dorsiflexion was $24.8 \pm 4.3^\circ$. In the flatfoot, maximum dorsiflexion was $25.5 \pm 5.4^\circ$ (NS). Metatarsal-tibial maximum eversion was $7.7 \pm 4.2^\circ$. In the flatfoot, maximum eversion was $17.9 \pm 6.1^\circ$ (significant). Metatarsal-tibial maximum external

rotation was $6.9 \pm 3.5^\circ$. In the flatfoot, maximum external rotation was $15.3 \pm 5.1^\circ$ (significant) (Table 2).

DISCUSSION

Flatfoot is a complex structural deformity which includes a decrease in the arch height, heel valgus, and forefoot abduction. The dynamic testing in this kinematics study demonstrated an increase in eversion and external rotation throughout most of stance phase. It also demonstrated the effects of the loss of posterior tibial muscle function as the typical pronounced inversion and internal rotation in late stance phase did not occur in simulated flatfoot gait. However, there still was some degree of motion toward inversion and internal rotation in toe-off, due to preservation of flexor digitorum longus, flexor hallucis longus, and gastrocnemius-soleus muscles which acted as secondary invertor.⁽²⁰⁾ The kinematics created *in vitro* were consistent with that observed clinically and radiologically, with forefoot external rotation and hindfoot eversion.

Several gait analysis studies have recently compared participants with flat feet to those with normal foot posture.^(9–12) Levinger et al.⁽¹⁰⁾ reported differences in foot motion between the two groups using a three-dimensional motion analysis system. Participants with flat feet demonstrated greater peak rearfoot eversion relative to the tibia (mean peak angular value of 5.8° vs 2.5°) and forefoot abduction relative to the rearfoot (mean peak angular value of 12.9° vs 1.8°), while similar rearfoot dorsiflexion/plantarflexion motion relative to the tibia compared with those with normal feet. The angular motion patterns of each foot segment during stance phase were similar between normal and flatfeet in the three planes. Tome et al.⁽¹¹⁾ reported the similar trend of foot motion changes with loss of arch height in flatfeet compared to normal. These findings were consistent with our results. The present study demonstrated that the changes in bone position during simulated walking in the flatfoot in stance phase approximated the in-vivo condition.

The pathology and pathoanatomy associated with posterior tibial tendonitis and tendon dysfunction have been studied. Mosier et al.⁽²⁷⁾ examined the histologic changes in surgical specimens from patient who underwent operative procedures for PTTD and flatfeet. Radiologic studies of flatfeet have also been conducted.^(4, 5) Karasick⁽⁴⁾ described the preoperative radiographic appearance of the acquired asymmetric flatfoot caused by a tear of PTT. These radiographic changes reflected the loss of tendon function and the development of a flatfoot deformity.

The disorder was examined experimentally in previous cadaveric testing. Chu et al.⁽³⁾ performed a cadaver study to determine if the application of muscle forces simulating the midstance part of gait had an effect on flatfoot deformity measured evaluated radiographically. Changes in the talar-first metatarsal angle and the height of the medial cuneiform were noted. The authors concluded that, the medial structures (the spring ligament and possibly the plantar fascia) must be severed to create an effective flatfoot model and cyclical loading of the foot further increases flattening of the arch. In the present study, the planar fascia was not sectioned to create the flatfoot, as it was not felt to be ruptured in patients with PTTD.

Niki et al.⁽²⁸⁾ investigated the functional role of PTT in acquired flatfoot; intact (normal) specimens were loaded to simulate heel strike, stance, and heel rise, with and without PTT functioning. Subsequently, a procedure was performed on the specimens to create a simulated flatfoot deformity. The flattened feet were then used to determine the effect of restoring PTT function to a flatfoot model. Small, statistically significant changes in the angular positions of the hindfoot complex were observed between functional and

dysfunctional PTT conditions. They concluded that the position of the foot could not be corrected by PTT as it had little effect in overcoming the soft tissue laxity. The present study demonstrated the value of dynamic testing as the effect of the flatfoot malalignment and loss of posterior tibial muscle function caused marked changes in the hindfoot alignment throughout the stance phase of gait.

A biomechanical evaluation of the foot and ankle should include the dynamic effects that may not be detectable in static measurement. Gait analysis can allow the investigation of important dynamic foot and ankle joint mechanics, but it is possible with cadaveric specimens to investigate internal parameters such as individual tarsal bone kinematics or soft tissue properties. The dynamic foot/ankle simulator will serve to study these internal parameters in normal and pathological conditions, and to study not only conventional but new treatment methods. Such investigations are obviously not ethical or possible in live subjects or patients.

There are several limitations of this study. First, it is not possible to simulate contraction of the intrinsic muscles of the foot. Secondly, the cadaveric specimens were elderly, and natural changes of foot posture and motion may exist with increasing age. Thirdly, full-scale normal ground reaction force loading was not applied to the cadaveric lower extremities as high loads may potentially affect suture fixation and soft tissue constraints with repeated testing. Further studies are needed to examine kinematics of the midfoot relative to the tibia or the calcaneus, and the calcaneus relative to the talus.

From the results of this study, we concluded that a simulator for the stance phase of walking was developed that which provided valid motions of the foot and ankle. The flatfoot deformity was associated with significant malalignment of the hindfoot (calcaneal-tibial) and forefoot (metatarsal-tibial) in multiple planes, throughout simulated stance phase of gait.

Acknowledgments

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Figure 1.
Schematic drawing of the dynamic foot/ankle simulator.

Figure 2.

Angular motions of the calcaneus (left) and the first metatarsal (right) relative to the tibia during the stance phase of walking in normal and flatfoot conditions (mean $\pm$ SD). Positive and negative values indicate dorsiflexion and plantar flexion, eversion and inversion, and external rotation and internal rotation, respectively.

Table 1

Calcaneal-tibial motion (Mean±SD)

	Normal	Flatfoot
Sagittal plane		
Max. plantarflexion at foot flat	10.9±4.2	11.5±4.4
Max. dorsiflexion	14.1±4.1	14.4±4.4
Max. plantarflexion at preswing	13.9±8.0	12.6±7.8
Total motion, foot flat to max. dorsiflexion	25.0±5.1	25.9±5.5
Total motion, max. dorsiflexion to preswing	28.0±7.7	27.0±7.3
Coronal plane		
Max. eversion	5.8±2.3	11.1±2.8 *
Max. inversion	4.0±4.0	-1.9±2.4 *
Total motion	9.8±3.6	9.2±2.9
Transverse plane		
Max. external rotation	2.3±1.7	8.1±4.0 *
Max. internal rotation	2.8±3.4	1.5±3.7 *
Total motion	5.1±2.7	9.6±3.8 *

* Significantly different ($p < 0.05$)

Table 2

Metatar sal-tibial motion (Mean±SD)

	Normal	Flatfoot
Sagittal plane		
Max. plantarflexion at foot flat	14.2±4.0	14.7±3.7
Max. dorsiflexion	24.8±4.3	25.5±5.4
Max. plantarflexion at preswing	17.1±9.8	13.3±9.2 *
Total motion, foot flat to max. dorsiflexion	39.0±4.5	39.9±5.4
Total motion, max. dorsiflexion to preswing	41.9±8.6	38.5±8.5 *
Coronal plane		
Max. eversion	7.7±4.2	17.9±6.1 *
Max. inversion	6.1±5.9	0.3±5.1 *
Total motion	13.8±6.9	18.2±8.0 *
Transverse plane		
Max. external rotation	6.9±3.5	15.3±5.1 *
Max. internal rotation	7.7±6.4	2.1±6.3 *
Total motion	14.6±6.0	17.5±5.1 *

* Significantly different ($p < 0.05$)