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ORIGINAL RESEARCH

Biomechanics of the First Ray Part I. The Effects of Peroneus Longus Function: A Three-Dimensional Kinematic Study on a Cadaver Model

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The closed kinetic chain effects of peroneus longus (PL) activity on the medial column of the foot were investigated in seven fresh-frozen cadaver specimens using a three-dimensional radiowave tracking system. Specimens, consisting of the distal half of the leg and the intact ankle and foot, were mounted on a nonmetallic loading frame which allowed positioning of the foot to simulate midstance position of gait. The tibia and fibula were axially loaded to 400 N. Receiving transducers were attached to the first metatarsal, medial cuneiform, navicular, and talus. Tarsal movements were measured as specimens were axially loaded and midstance motor function was simulated using pneumatic actuators. Tensile loads of 0–150% of predicted maximum force were incrementally applied to the PL tendon. Three-dimensional data sets recording osseous positions and orientations were collected and analyzed. Significant frontal plane rotation of the medial column in the direction of eversion occurred when PL strength was increased ($p = .0001$). Increasing PL loads produced significant but less pronounced angular changes in the sagittal and transverse planes of the medial column. The patterns of motion suggest that PL creates an eversion "locking" effect on the first ray of the foot, stabilizing the medial column. (The Journal of Foot & Ankle Surgery 38(5):313–321, 1999)

Key words: first ray hypermobility, foot biomechanics, medial column, peroneus longus, tarsal movements

Introduction

The First Ray

Anatomically, the first ray has been discussed by many authors (1–6). It consists of the first metatarsal and the medial cuneiform and has distinct ligamentous attachments (1). The plantar first metatarsocuneiform ligament has been described as a dense structure which acts as a major component in stabilizing the first metatarsal during weightbearing (4, 6). The shape of the medial cuneiform has also received attention in the literature (4, 7–10, 11). The oblique setting of its distal articular facet, often seen in metatarsus primus varus, has been referred to as an atavistic cuneiform and implicated in the pathogenesis of hallux abducto valgus deformity (7–9).

Morton, from 1924 to 1935, demonstrated the importance of the first metatarsal segment in the maintenance of the longitudinal arch (7, 12–15). He also advanced the concept of first ray hypermobility. He stated: "Hypermobility of the first metatarsal segment is responsible for the widest range of foot trouble" (15). His criteria for hypermobility include clinically demonstrable hyperextension (increased first metatarsal dorsal excursion), widening

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of the space between the first and second cuneiforms, and a thickened second metatarsal shaft (7, 12–15). To combat hypermobility, Rutherford advocated metatarsocuneiform arthrodesis, while Courriades proposed fusion of the first and second cuneiforms (16–18). Klaue and associates suggested a direct relationship between painful hallux valgus deformity and hypermobility of the first metatarsocuneiform joint (19).

Clinically, normal motion of the first ray was described by Root and associates to be roughly 5 mm above and 5 mm below the plane of the lesser metatarsals (20). This approximation of medial column mobility was confirmed by a clinical investigation by Klaue et al. They found dorsiflexion of the first ray to be 5.3 ± 1.4 mm in normal feet and 9.3 ± 1.9 mm in hallux valgus feet (19). While the quantification of first ray movement is important clinically, the measured excursion is a summation of movement contributed from each osseous segment of the medial column.

First ray motion has been studied by numerous investigators with variable results. Not only is there inherent difficulty in measuring the segmental source of the small movements, but it is further complicated by differing measurement methods and experimental protocol (1–3, 21–23). In a classic investigation, Hicks studied 15 normal, freshly prepared, undissected limbs (1). He was able to control axial loading of the specimens as well as loading of all extrinsic tendons to simulate muscle function. He described the axis of first ray motion passing through the dorsum of the foot from the third metatarsal base to the navicular tuberosity. The movement about this axis is that of dorsiflexion inversion and plantarflexion eversion. This motion was later confirmed by other investigators using varying methods on cadaveric specimens (2, 21).

Alternatively, other investigations report contrasting first ray movements monitored while the entire foot was manipulated via internal and external rotation of the leg (3, 22). At first, this appears to conflict with Hick's findings. However, with closer examination of the data, it is evident that tarsal movements induced by rotating the tibia alter the net movements observed at the first ray. This is likely achieved through a coupling effect from more proximal joints.

The amount of first ray motion was described by Hicks as $22° \pm 8°$ about his described axis in open kinetic chain. Other authors have focused on range of motion of the first metatarsocuneiform joint in the sagittal plane with measurements ranging from 4.37° to 12.38° in open kinetic chain (21, 24, 25). Furthermore, frontal plane motion of the first metatarsal relative to the medial cuneiform was seen by Kelso to be $8.23° \pm 4.12°$ (21). Similarly, Wanivenhaus saw a total of 6.2° of frontal plane motion at the first metatarsocuneiform joint (26). In a more sophisticated study, Kitaoka (1995) utilized the Fastrak® tracking system[3] and found, with loading, most of the movement of the first ray occurred at the talonavicular joint. Motion of the first metatarsal relative to the navicular averaged $7.2° \pm 1.5°$ and was greatest in dorsiflexion and inversion. Unfortunately, the motion at the naviculocuneiform and the first metatarsocuneiform joints was not evaluated separately in this study (27).

Mechanical Influences of Peroneus Longus

Peroneus longus (PL) arises from the head and lateral shaft of the fibula and becomes tendinous as it enters the middle one-third of the lateral compartment of the leg. The tendon courses around three pulleys including the lateral malleolus, the peroneal tubercle, and the cuboid. PL then crosses the sole of the foot to insert on the plantar lateral base of the first metatarsal (Figs. 1 and 2). The tendon of PL also sends a slip to insert on the plantar lateral medial cuneiform (28, 29).

Even as far back as 1867, PL has been cited as having an effect on the joints of the medial border of the forefoot (30). Duchenne's in-depth descriptions regarding PL activity were from extensive clinical observations. He saw the great importance of PL action in maintaining the configuration of the foot. Upon loss of PL function, tibialis anterior would gradually elevate the first metatarsal resulting in a flat foot (30). This has also been confirmed by other authors (31–35). Conversely, contracture of PL produced an exaggeration of the plantar arch and a decrease in transverse diameter of the forefoot (30). Likewise, other authors have described such observations,

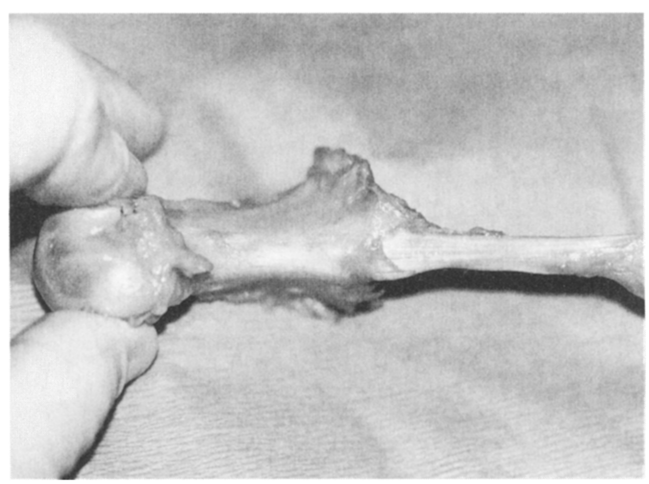

FIGURE 1 Cadaver specimen of first metatarsal with intact peroneus longus insertion at its base.

[3] Polhemus, Inc., Colchester, VT.

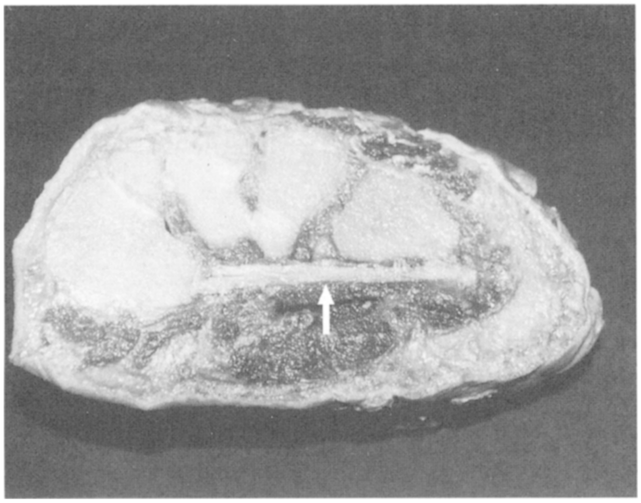

FIGURE 2 Oblique cross section of midfoot illustrating the course of peroneus longus (*arrow*) in the plantar foot to its insertion on the first metatarsal.

many cases in relation to Charcot-Marie-Tooth (CMT) patients (33, 36–39).

Functional information has also been gained from electromyographic (EMG) investigations of PL. In a relaxed standing position, PL as well as tibialis anterior and tibialis posterior are electrically silent. Yet, dynamically, they play a significant role in supporting the longitudinal arch during ambulation (36, 40). The most elaborate EMG study of PL was performed by Gray and Basmajian in conjunction with cinematography during walking. PL was found to be active in midstance and heel-off and generally more active in flat-footed persons (41). A similar gait phase pattern has been noted by other investigators (42, 43).

The effect of foot type on PL function has been theorized. Root and his associates discuss that PL in a normal foot, as a result of the inferior position of the cuboid, exerts a stabilizing influence on the first ray against ground reaction forces (44). As the foot pronates, the insertion of the PL moves closer to the ground plane and the level of the cuboid, thus lessening the stabilizing influence of PL (44, 45). This theory has not been proven in the laboratory nor have the effects of PL in other cardinal planes been reviewed.

In an early study of PL, Jones applied 100 pounds of hand-operated tension to the PL tendon with bathroom scales under the forefoot. Tension on the PL tendon produced 73.5 pounds of pressure under the first metatarsal and 26.5 pounds under the lateral four metatarsals (46). More recently, Bohne studied the action of PL against forces that increase the metatarsal angle. With a medially directed force on the first metatarsal of cadaver feet, sequential sectioning of the soft tissues between the first and second metatarsals was performed. A statistically significant medial displacement of the first metatarsal was observed after transection of the PL tendon. They concluded that PL plays a role in opposing medial deviation of the first metatarsal (47).

A limited amount of kinematic investigations associated with PL have been done. Thordarson et al. in 1995 determined through use of a three-dimensional tracking system that the unopposed pull of PL consistently abducted the forefoot in the transverse plane. The movements occurring at the first metatarsal–first cuneiform joint as well as the navicular–first cuneiform joint were not evaluated. Their study also did not elaborate on the effects of varying strengths of PL (48).

Thus far, study concerning the precise direct effect of PL on the first ray osseous motion has been limited. No study to date describes first ray motion with weightbearing load and simulated intact muscle function. As a result, our comprehension of medial column function and the condition of first ray hypermobility is not complete. The purpose of Part I in this series of investigations is to more precisely define first ray motion as a function of PL activity in a closed kinetic chain environment.

Materials and Methods

Specimen Acquisition and Preparation

Seven fresh cadaver lower limb specimens (all lefts to simplify load frame configuration) were obtained from the Department of Biological Services at the University of Washington. Each was transsected 15–20 cm above the tibial plafond and deep frozen to −20°C. They were wrapped in saline-soaked towels, sealed in plastic bags, and frozen. Screening was performed by radiography and by visual inspection for abnormal joint space narrowing, significant alignment abnormalities of the medial column and rearfoot, and poor bone stock. Specimens without gross arch height abnormalities were accepted for investigation.

The mean age at time of death was 84.1 years, ranging from 76 to 94 years (3 male, 4 female). Before testing, each specimen was thawed for 6–8 hours at room temperature. The skin, muscle, and soft tissue were removed from around the leg and dorsal foot, carefully preserving the posterior leg tendons, ligaments, interosseous membrane, plantar soft tissues, and the joints of the foot.

The Load Frame

Each specimen was mounted on a custom loading frame (Fig. 3).[4] This custom load frame, made of acrylic and aluminum, permitted actuator function on the upper portion and specimen mounting and loading on the lower

[4] Designed by BioConcepts, Inc., Seattle, WA and fabricated by Advanced Biomedical, Inc., Oakland, CA.

FIGURE 3 Testing apparatus consisting of custom load frame and actuator controls with specimen in place.

part. The lower portion of the frame was constructed of nonmetallic materials to prevent signal interference of the radiowave tracking system. The elevated top surface of the frames' upper portion permitted mounting of pneumatic load cylinders five feet above the load platform. A downward load was applied through a polycarbonate rod onto the tibia by means of a center actuator. The center actuator was surrounded on three sides by smaller actuators designed to apply load to tendinous structures via nonmetallic cables.

A ½-inch, threaded, nonmetallic rod was placed into the tibial medullary canal to act as vertical load transfer device to the specimen. A load distributing plate for the tibia and fibula was utilized as described in a previous investigation (49). This arrangement permitted unconstrained transverse rotation and vertical displacement of the tibia. The plantar aspect of the foot was placed on nonskid material on the load frame platform. The plantar soft tissues were left intact which permitted the foot to seek equilibrium of arch height under load.

Three-Dimensional Tracking System

Osseous segments of the medial column were tracked with a radiowave tracking system.[5] This 3-D motion tracking system used radio frequency waves generated from a transmitter (a black box mounted posterior and lateral to the foot) and receiving transducers which were attached to the osseous segments. The radio signals were collected, and the positions of the sensors were determined by computer algorithms that convert the signals to data points which were stored on computer files. The accuracy was tested and verified on a calibrated platform and determined to be within 0.10 mm and 0.15° for all receivers. The unit tracked motion in a global coordinate system using 6 degrees of freedom (linear displacement along X, Y, Z coordinates and rotational displacement around each axis).

Attaching and Aligning the 3-D Motion Tracking System

Nylon threaded pins were anchored to the center of the dorsal first metatarsal, dorsal medial cuneiform, dorsal navicular, and lateral talus. A carbon fiber rod was drilled down the center of each nylon threaded pin. The carbon fiber rods were small and did not interfere with joint motion. The signal receivers of the three-dimensional tracking system were attached to the carbon fiber rods. Thus, each receiver moved with the movement of the bone to which it was attached. The nylon threaded pin was cinched down tightly with a nut on the other side of the bone to prevent rotation of the pin.

Tendon Attachment Clamps

Serrated, nonmetallic tendon clamps were custom fabricated and attached to proximal tendon stumps (Fig. 4). The clamping devices were connected to separate tendon pneumatic actuators via nonmetallic cables made of braided Dacron® cord.[6] This system permitted a constant load to be applied to the tendons independent of alterations in foot position or tendon displacement.

Determining Tendon Actuator Forces

Relative forces placed on each tendon were determined using Brand's calculation of physiological cross-sectional area of muscles of midstance (50). Tendons loaded included PL, peroneus brevis, Achilles, tibialis posterior, flexor digitorum longus, and flexor hallucis longus. Preliminary testing was performed to evaluate the proper Achilles load necessary to maintain consistent balance throughout the testing sequence. This was found to be 30% of predicted maximum Achilles load. Collectively, tendon load was kept constant throughout testing with the exception of the PL tendon, which was incrementally increased.

Experimental Sequence

The testing procedure was undertaken at room temperature, keeping the specimen moist with intermittent misting. Each specimen was taken through a conditioning sequence

[5] Fastrak®, Polhemus, Inc., Colchester, VT.

[6] Western Filament, Inc., Grand Junction, Co.

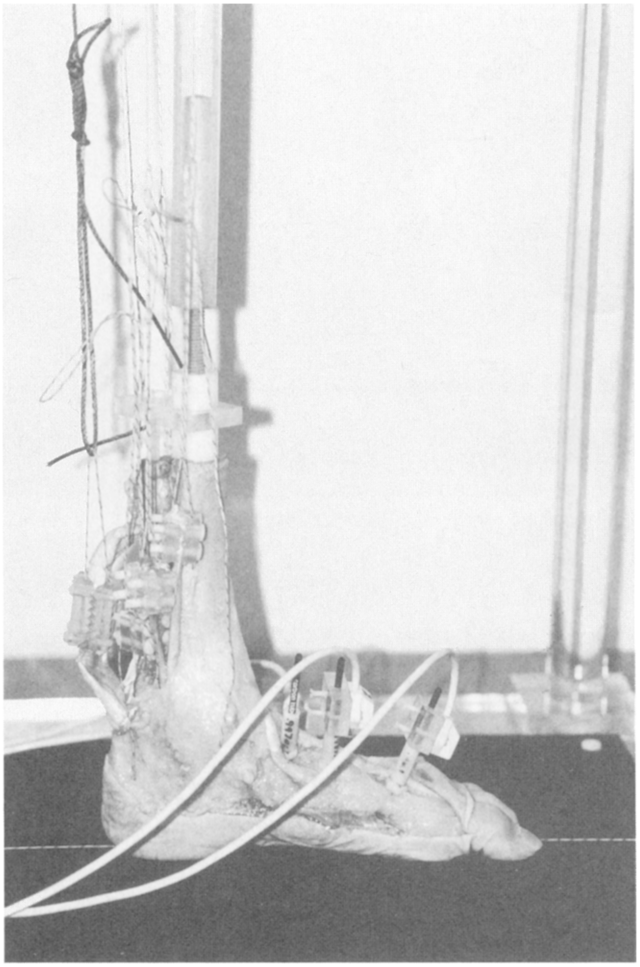

FIGURE 4 Test specimen with load applied on tibia, cables attached to the tendons of midstance, and Polhemus Fastrak® sensors attached to osseous segments.

which consisted of 100 N of weightbearing load to the specimen and submaximal tendon load to the tendons for three sets of 15 seconds each. The specimen was then loaded with 400 N, while the tendons of midstance were loaded, except for PL. Then PL strength was increased: 0%, 50%, 100%, then 150% of the predicted maximum PL load. Three-dimensional data were recorded for the position and orientation of the bone segments at each load increment. Two trials were run for each sequence to ensure data reproducibility.

Arch Height Documentation

Utilizing a custom loading x-ray apparatus, a position-controlled lateral radiograph of each test specimen was obtained with 222 N of axial load applied. The x-ray beam was accurately aimed at the dorsal aspect of the naviculocuneiform joint for each specimen. The lateral radiograph was used to establish a *tarsal index* as described

$$i_t = 100 \times P_C / L_T \times t_g \varepsilon TC$$

i_t = tarsal index

P_C = talocalcaneal overlap in mm

L_T = length of talus in mm

t_g = tangent

εTC = talocalcaneal inclination angle in the longitudinal direction of the foot

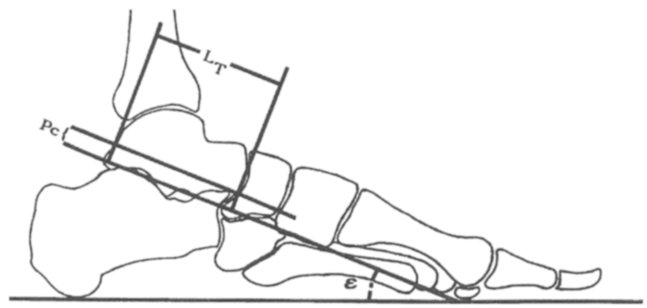

FIGURE 5 Breakdown of equation describing Benink's tarsal index.

by Benink (51) (Fig. 5). This classified the foot type for each of the specimens tested.

Statistical Analysis

Postprocessing of the kinematic data was performed on a personal computer using a custom software program and a statistical software program.[7] Kinematic data was analyzed using a one-way analysis of variance (ANOVA) with post hoc testing using Scheffe's Multiple Comparisons Test for significant values at $p < .05$.

Results

First Metatarsal Motion

First metatarsal rotational orientations of loaded foot specimens with variable PL strength were recorded in all three cardinal planes. With increased strength, PL effectively everted and slightly plantarflexed the first metatarsal from its normal resting position. At 150% PL load, mean frontal plane rotational differences (eversion around the X axis) measured $8.06° \pm 3.07°$ ($p = .0001$) (Fig. 6). Differences in sagittal plane rotation (plantarflexion around the Y axis) measured $3.8° \pm 0.54°$ at 150% PL load ($p = .0001$) (Fig. 7). Differences in

[7] Statview 4.0, Abacus Systems, Berkeley CA.

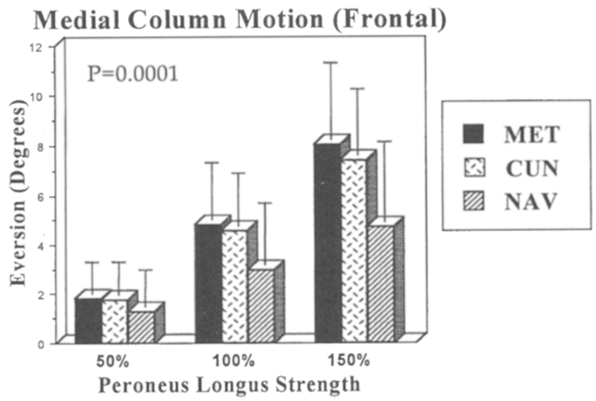

FIGURE 6 Bar graph of medial column motion in the frontal plane with increasing peroneus longus pull ($p = .0001$).

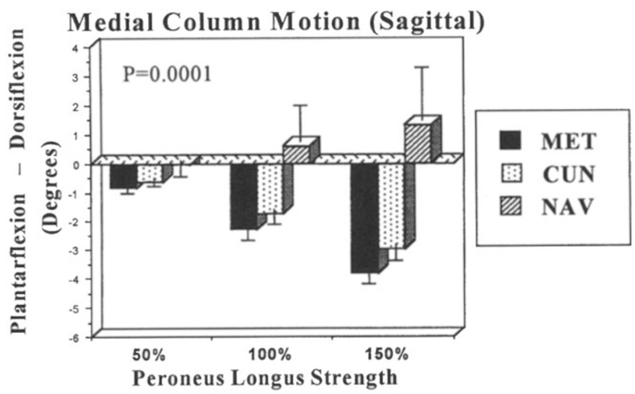

FIGURE 7 Bar graph of medial column motion in the sagittal plane with increasing peroneus longus pull ($p = .0001$).

transverse plane (Z axis) and linear displacement of the first metatarsal with increasing PL strength were not significant.

Medial Cuneiform Motion

Medial cuneiform rotational orientations of loaded foot specimens with variable PL strength were recorded in all three cardinal planes. With increased strength, PL effectively everted and slightly plantarflexed the medial cuneiform from its normal resting position. At 150% PL load, mean frontal plane rotational differences (eversion around the X axis) measured $7.44° \pm 2.64°$ ($p = .0001$) (Fig. 6). Differences in sagittal plane rotation (plantarflexion around the Y axis) measured $2.97° \pm 0.57°$ at 150% PL load ($p = .0001$) (Fig. 7). Mean transverse plane rotational differences (abduction around the Z axis) measured $2.12° \pm 1.79°$ at 150% PL load ($p = .0036$) (Fig. 8). Differences in linear displacement of the medial cuneiform with increasing PL strength were not significant.

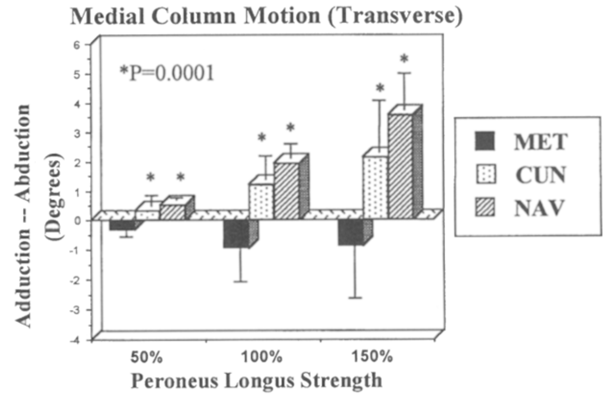

FIGURE 8 Bar graph of medial column motion in the transverse plane with increasing peroneus longus pull ($p = .0001$).

Navicular Motion

Navicular rotational orientations of loaded foot specimens with variable PL strength were recorded in all three cardinal planes. With increased strength, PL effectively everted and slightly dorsiflexed the navicular from its normal resting position. At 150% PL load, mean frontal plane rotational differences (eversion around the X axis) measured $4.73° \pm 3.22°$ ($p = .0001$) (Fig. 6). Differences in sagittal plane rotation (dorsiflexion around the Y axis) measured $1.34° \pm 1.8°$ at 150% PL load ($p = .0085$) (Fig. 7). Mean transverse plane rotational differences (abduction around the Z axis) measured $3.58° \pm 1.22°$ at 150% PL load ($p = .0001$) (Fig. 8). Differences in linear displacement of the navicular with increasing PL strength were not significant.

Talar Motion

Talar rotational orientations of loaded foot specimens with variable PL strength were recorded in all three cardinal planes. With increased strength, PL effectively abducted, dorsiflexed, and slightly everted the talus from its normal resting position. At 150% PL load, mean transverse plane rotational differences (abduction around the Z axis) measured $4.47° \pm 0.66°$ ($p = .0001$). Sagittal plane rotation (dorsiflexion around the Y axis) measured $3.47° \pm 0.75°$ at 150% PL load ($p = .0001$). Frontal plane rotational differences (eversion around the X axis) measured $0.9° \pm 0.38°$ at 150% PL load ($p = .0143$). Differences in linear displacement of the talus with increasing PL strength were not significant.

Arch Height Calculations

Benink's *tarsal index* formula was utilized to calculate the arch height of each specimen (51) (Fig. 5). The *tarsal* *index* for the seven specimens ranged from 2.1 to 10.7, with a mean of 5.80 ± 2.9. This is an above-average arch height. There was no significant correlation between arch height and the effect of PL action. However, the effect of arch height on PL function could not be determined due to the small sample size.

Discussion

The concept of hypermobility of the first ray, as initially proposed by Morton, has been supported by subsequent investigation (7, 12–15, 19, 25, 26, 52). It appears to have a clear relationship with hallux abducto valgus (2, 19, 53). While the ranges of motion of the normal and abnormal first ray have been determined, a clear etiology of hallux abductovalgus and hypermobility still remains elusive. Prior to this investigation, our comprehension of closed kinetic chain function of PL was largely circumstantial. First, it was clear from Duchenne's classic observations, which have been confirmed by others, that the functional position of the first ray depends upon the agonist and antagonist balance of PL, tibialis anterior, and tibialis posterior (30, 31, 33–39, 54). Second, PL, when unopposed, functions as an abductor of the forefoot (48). Third, dynamic EMG confirms PL activity during midstance into propulsion with enhanced electrical activity in flat-footed individuals (41).

Our functional simulation of peroneus longus provided significant insights into the subtle mechanical relationships of the medial column. With increasing tensile load on the PL tendon, the first metatarsal was observed primarily everting and, to a lesser extent, plantarflexing. This varies considerably from previous theoretical descriptions (44, 45). During the testing, the frontal plane movement was easily observed on all specimens without the need for instrumentation. Interestingly, this correlates well with the early findings of Duchenne. He saw PL as playing an important role in the configuration of the foot. Aside from lowering the first ray, it produced torsion of the medial column on the lateral column enhancing the transverse arch of the foot (30).

From Duchenne's work, it is obvious that PL has the capacity to work in the sagittal plane. However, since PL is possibly not able to overcome weight-bearing load, it is likely working to prepare for the next functional event in the gait cycle, namely the windlass effect. In the normal foot, the windlass mechanism and PL work in tandem to take the foot smoothly into propulsion. If there is a defect in the dynamics of either the windlass or PL function, a pathologic situation may result, such as first ray hypermobility and resultant hallux valgus or functional hallux limitus (7, 12–15, 19, 23, 44, 55, 56). This important aspect of foot mechanics needs further research to delineate these relationships.

Our data is in accordance with the first ray axis described by Hicks and Kelso as that of dorsiflexion inversion and plantarflexion eversion (1, 21). With respect to the segments of the medial column, while the first metatarsal adducted toward the midline of the body, the medial cuneiform, navicular, and talus abducted with PL action (Fig. 8). Conceptually, as PL tension increased, all segments resupinated, and the foot assumed a more cavus appearance. This conflicts with Thordarson's report of PL activity causing an abducted forefoot. However, his test model lacked an opposing tibialis posterior muscle during simulated PL activity. This would naturally skew his conclusions as to the true function of PL (48).

In their biomechanical work, Root and his colleagues stressed the importance of foot type on PL function (44). In their theory, arch height affects the mechanical advantage of PL in controlling sagittal plane motion of the first ray. Based on this theory, it is often felt that a high-arched foot would predispose PL to have a mechanical advantage. However, the authors feel that this may be true in local cavus foot deformities and less likely in global cavus deformities. The arch height indices in this investigation were determined to be above average. Unfortunately, no conclusions could be drawn with our limited number of specimens. A research study focused on the variable arch height and type of cavus would help to resolve this issue.

A difference in magnitude of frontal plane rotation between the bones of the medial column was evident in this investigation. The first metatarsal everted more than the medial cuneiform with increasing PL pull. Similarly, the medial cuneiform everted more than the navicular. Collectively, under the influence of PL, the medial column functions as a series of links. As motion is taken up at each of the tarsal joints, this imparts torsional load to more proximal segments. PL is essentially creating an eversion "locking" of the first ray against the rest of the foot. Along the same lines as this locking mechanism, Berndt and Harty discussed torsional impaction of diarthrodial joints (57). With eversion torsion of the medial column, the ligaments at the first metatarsocuneiform and the naviculocuneiform joints invariably become taut, causing tightening of the joints. Subsequently, stabilization of the medial column results. To conceptualize further, a locked medial column, as it is drawn into plantarflexion by peroneus longus, can cause reciprocal supination of the rearfoot. It can be inferred that, through this "locking mechanism," PL plays a role in counteracting first ray hypermobility. In this way, PL may limit the development of hallux abducto valgus deformity as well.

All experimental studies using cadaveric specimens have inherent limitations. The simulation of contraction of the muscles of midstance by applying tension to the tendons may not be entirely physiologic. However, in this study, tensions were based on available physiologic muscle data (50). In addition, only midstance function of PL was evaluated, but according to Gray and Basmajian's work, PL is active in midstance (41). The role that the intrinsic muscles play in first ray function, especially flexor hallucis brevis, is unknown and warrants further investigation. Despite these limitations, such cadaveric studies have the capacity to reveal subtle biomechanical relationships that go unrecognized with in vivo investigations.

Further evaluation in propulsion phase of gait would be beneficial. Additionally, future research should evaluate the effect of arthrodesis at the first metatarsocuneiform joint, and more clearly define the clinical examination for hypermobility of the first ray. In addition, the effect of equinus deformity on first ray mechanics warrants investigation.

As was stressed by Hammond, tendon transfers should not be approached lightly. He saw dramatic alterations in forefoot alignment following PL transposition to tibialis anterior for a plantarflexed first ray. This tendon transfer lead to an elevated first ray and was similarly described by Lapidus as an etiology for dorsal bunion formation (32, 33). Essentially, the agonist (PL) was transferred to the antagonist's location. For a pathologically plantarflexed first ray with a dynamic component, a PL lengthening procedure may make more sense (58). Or even better, PL can be combined with peroneus brevis to affect balancing of the foot. Hansen has used such a transfer in situations of instability from damage to peroneus brevis, recalcitrant sesamoiditis from a plantarflexed first ray, and neuropathic ulcerations under the first metatarsal head (59). Following transfer of PL to peroneus brevis, the distal PL stump should be attached to the lateral foot for static stabilization of the medial column.

The predisposing factors to the genesis of hallux abducto valgus still remain a great topic of controversy. Many authors have advocated that the main deformity arises more proximally at the first metatarsocuneiform joint (7–10, 12–15, 60–62). Hypermobility has also been implicated as a causative factor of first ray structural malalignment (7, 12–15, 19, 24, 45, 53, 60, 63, 64). In this light, PL through its locking effect, may impede the development of hallux abducto valgus deformity. Arthrodesis of the metatarsocuneiform joint has been advocated by many authors for the treatment of metatarsus primus varus with hypermobility. (8–10, 16, 17, 19, 53, 60, 62, 65–73). Since arthrodesis of this joint acts to provide plantarward stability of the first metatarsal, it may functionally mimic PL activity.

Conclusion

PL is primarily an evertor of the medial column in closed kinetic chain midstance. A lesser magnitude of plantarflexion motion occurs as a result of PL activity when the foot is loaded and midstance muscles are active. Through a torsional mechanism, PL creates an eversion "locking effect" on the first ray of the foot, essentially stabilizing the medial column.

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