Sensitivity of Plantar Pressure and Talonavicular Alignment to Lateral Column Lengthening in Flatfoot Reconstruction

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Background: Lateral column lengthening (LCL) of the calcaneus is commonly performed as part of correction of the adult acquired flatfoot deformity. Increases in postoperative lateral plantar pressure associated with pain in the lateral aspect of the foot have been reported. The aim of this study was to investigate changes in pressures in the lateral aspect of the forefoot with increments of 6, 8, and 10 mm of LCL in a cadaveric flatfoot model. The hypothesis was that increasing the LCL incrementally by 2 mm will linearly increase the plantar pressures in the lateral aspect of the forefoot.

Methods: Eight fresh-frozen cadaveric foot specimens were used. A robot compressively loaded the foot to 400 N with a 310-N tensile load applied to the Achilles tendon. A flatfoot model was created by resecting the medial and inferior soft tissues of the midfoot, followed by axial load of 800 N for 100 cycles. Kinematic and plantar pressure data were gathered after the different amounts of LCL (6, 8, and 10 mm) were achieved.

Results: The talonavicular joint demonstrated a median abduction angle of 4.4° in the axial plane and -2.6° in the sagittal plane in the flatfoot condition as compared with the intact condition. The 6, 8, and 10-mm LCLs showed axial correction of talonavicular alignment by -1.4°, -4.9°, and -9.2° beyond that of the intact foot, and sagittal correction of -0.1°, 1.3°, and 2.9°, respectively. LCL of 6, 8, and 10 mm showed consistently increasing lateral forefoot average mean pressure, peak pressure, and contact area.

Conclusions: LCL in 2-mm increments consistently reduced talonavicular abduction and consistently increased plantar pressure in the lateral aspect of the forefoot.

Clinical Relevance: The lateral column should be lengthened judiciously, as a 2-mm difference leads to significant difference not only in angular correction of the talonavicular joint but also with regard to pressure in the lateral aspect of the forefoot.

The lateral column lengthening (LCL) of the calcaneus, as described by Evans, is commonly performed as part of the surgical correction of Stage-II adult acquired flatfoot deformity¹⁻⁵. This procedure corrects forefoot abduction deformity, increases talonavicular coverage, and helps restore the medial longitudinal arch⁵⁻⁷. Although LCL offers powerful correction of talonavicular subluxation, postoperative increases in lateral plantar pressure associated with pain in the lateral aspect of the foot, stiffness of the foot, and stress fracture of the fifth metatarsal have been reported⁸⁻¹¹. It has been suggested that such compli-

cations are consequences of varus or supination of the forefoot^{3,7,8}. To prevent this problem, Davitt et al.⁸ recommended that the lengthening should be limited to <1 cm or combined with first-ray stabilization procedures such as a first tarsometatarsal fusion (i.e., the Lapidus procedure) or medial cuneiform osteotomy (i.e., the Cotton osteotomy)^{12,13}. Attempts have been made to avoid these complications by measuring intraoperative changes in plantar pressure associated with LCL^{14,15}. However, quantitative assessment of the effect of different amounts of LCL on plantar pressure and on foot kinematics is lacking.

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Surgeons estimate the appropriate amount of LCL in part on the basis of a preoperative clinical and radiographic assessment of forefoot abduction. The final determination is often made intraoperatively with the use of trial wedges by assessing the amount of lengthening that reduces abduction deformity of the talonavicular joint without causing excessive stiffness of the lateral aspect of the forefoot⁴. There is no objective information to suggest how much lengthening will yield an adequate correction of the deformity while not excessively increasing pressure in the lateral aspect of the forefoot. Past clinical experience has suggested that pressures in the lateral aspect of the forefoot are highly sensitive to the amount of LCL⁴. Specifically, a small change in LCL can substantially alter pressure in the lateral aspect of the forefoot.

Therefore, we chose the following research question for this study: What is the effect of clinically relevant levels of lateral column lengthening (6, 8, and 10 mm) on lateral plantar pressure of the forefoot and on talonavicular alignment? The hypothesis was that an incremental LCL by 2 mm would significantly increase lateral plantar pressure of the forefoot in a linear fashion while significantly reducing midfoot abduction at the talonavicular joint.

Materials and Methods

Eight fresh-frozen cadaveric human feet from four male and four female donors with a mean age of 51.0 years (range, thirty-five to seventy-three years) were used. The specimens had been stored at -20°C and thawed at room temperature for twelve hours prior to use. The tibia and fibula were truncated 20 cm from the ankle joint and prepared by affixing the fibula to the tibia with use of two 3.5-mm cortical screws¹⁶. The tibial shaft was embedded in a cylinder of epoxy cement (Bondo/3M, Atlanta, Georgia)¹⁶. The potted tibial shaft was fixed to a pedestal on the floor; thus, the plantar aspect of the foot faced upwards (Fig. 1). To facilitate loading of the Achilles tendon, a strap of polypropylene mesh fabric was sutured to the tendon with number-2 FiberWire (Arthrex, Naples, Florida). Subsequently, a wire cable attached to the strap was routed through a low-friction bushing to a pneumatic cylinder. A load cell (model MLP-200, 0.1% linearity; Transducer Techniques, Temecula, California) was attached in series with the cable to monitor the applied load, which was regulated through a valve on the pressure cylinder.

Each foot was loaded axially through an acrylic plate ($34.3\text{ cm} \times 34.3\text{ cm} \times 4.45\text{ cm}$) attached to the end of a robotic arm that had six degrees of freedom (ZX165U; Kawasaki Robotics, Wixom, Michigan) and a positioning repeatability of $\pm 0.3\text{ mm}$ ¹⁷. A universal force-moment sensor (Theta; ATI Industrial Automation, Apex, North Carolina) (capacity: $F_x = F_y = 1\text{ kN}$, $F_z = 2.5\text{ kN}$, $T_x = T_y = T_z = 120\text{ Nm}$; resolution: $F_x = F_y = F_z = 0.25\text{ N}$; $T_x = T_y = 0.025\text{ Nm}$, $T_z = 0.0125\text{ Nm}$) was mounted to the end of the arm to measure loads acting across the foot. The x, y, and z loading directions were based on the anatomy of the tibia and were identified with use of a digitizer (MicroScribe; Immersion, San Jose, California). The axes corresponded to anatomical directions and followed the conventions established by the International Society of Biomechanics¹⁸. The robot allowed abduction-adduction of the foot, medial-lateral and anterior-posterior translation, and compression-distraction; however, the robot did not allow rotation in inversion-eversion. For each condition tested, the foot was loaded to 400 N of axial force, which is half the force that would be applied ordinarily during standing by a person of average mean body weight (80.5 kg [86.6 kg for men, and 74.4 kg for women]) in the United States¹⁹. With 400 N axial force applied, the Achilles tendon was loaded to 310 N with use of a pneumatic cylinder to transfer the center of pressure from the heel to the midfoot,

Fig. 1

Optoelectronic sensors were attached to the tibia, talus, navicular, cuboid, and calcaneus. Each sensor was anchored. The specimen was mounted on a table in an upside-down position beneath the Novel Pliance sensor array, which was affixed to an acrylic plate connected to the robotic arm. For all test conditions, an axial load of 400 N was applied while the Achilles tendon was pulled with a force of 310 N. The arrow indicates the direction of loading by the robotic arm.

thereby replicating the midstance phase of walking²⁰. The 310-N load was determined from a pilot test, where load on the Achilles tendon was gradually increased under the prescribed axial load until the center of pressure moved to the midfoot.

Flatfoot Model

To create a mild flatfoot model, the tibialis posterior tendon, the medial and inferior talonavicular interosseous ligament, and the spring ligaments were resected as previously suggested²¹. After the medial soft-tissue resection, the foot was loaded to 800 N for one hundred cycles for preconditioning. While others have combined higher loads and thousands of cycles—without ligament resection—to create severe flatfoot deformity²², the goal of this study was to assess the effect of LCL in a milder deformity. Therefore, high numbers of cyclic loadings were avoided.

Lateral Column Lengthening

An osteotomy was made approximately 1.5 cm proximal to the calcaneocuboid joint. The cut was aimed between the middle and anterior facets of the calcaneus. A 10 mm $\times$ 30 mm standard-tooth oscillating saw blade with a cutting thickness of 0.4 mm was employed for the osteotomy (Synthes, West Chester, Pennsylvania). Therefore, the actual amounts of lengthening were 0.4 mm smaller than the actual size of standard wedges. Specially designed, custom-made stainless steel wedges (6, 8, and 10 mm) were employed for this study. The wedges were tapered plantarly as well as medially to fit the shape of the Evans osteotomy upon distraction (see Appendix). The order of lengthening was determined at random. These wedges were fixed to the calcaneus with a four-hole 3.5-mm plate with use of two 3.5-mm cortical screws (Synthes) (Fig. 2). For each wedge, the same distal hole was utilized for distal fixation but the holes used for proximal fixation varied according to the length of the wedge. For instances in which the proximal screw hole partially overlapped the previously made screw hole, the plate was rotated superiorly or inferiorly and another hole was drilled. Compressive loading across the construct also helped achieve stable fixation during static loading.

Fig. 2

For the LCL, an osteotomy was made in the anterior part of the calcaneus, 1.5 cm proximal to the calcaneocuboid joint. Wedges were fixed with a 3.5-mm plate and two 3.5-mm cortical screws. For each wedge, the same distal hole was used to fix the distal osteotomy site, whereas the hole used for proximal fixation varied according to the length of the wedge.

Angulation Data

Tarsal rotations were measured with use of a three-dimensional optoelectronic motion capture system (ProReflex MCU; Qualisys, Gothenburg, Sweden). Rigid bodies were embedded in the tarsal bones with use of a specially made interface plate²³. The tracking system was accurate to within 50 μ m root mean square corresponding to 2.6% error with a coefficient of variation of 3.2%, for displacements of 2.00 mm as measured with use of a digital caliper (Fowler, Newton, Massachusetts) as the reference standard. The caliper had a resolution of 10 μ m, an accuracy of 20 μ m, and a repeatability of 10 μ m. A wand was used to register specific anatomical points to create an anatomical coordinate system.

The differences in the talus and navicular rotation about the long axis of the tibia were used to define talus-navicular orientation in the axial plane (Fig. 3). Dorsiflexion-plantar flexion of the talus was defined with use of the transmalleolar axis. Both reference axes were defined according to standard conventions¹⁸.

Plantar Pressures

A custom Pliance 32 sensor array (Novel, Munich, Germany) was attached to the acrylic footplate. The sensor array had a total area of 320 $\times$ 160 mm², and

involved a total of 812 individual sensors, thus providing a resolution of 1.85 sensors/cm². Data were analyzed with standard software (model Pliance-Expert, version 11.3.12, Novel, Munich, Germany and St. Paul, Minnesota) (Fig. 4). The foot plantar surface was divided into twelve regions with use of a standard automasking algorithm. The average mean pressure, peak pressure, and contact area of each region were measured. Summary data were reported for the medial and lateral sides of the forefoot. The medial side of the forefoot contained the first and second metatarsals, while the lateral side of the forefoot contained the third, fourth, and fifth metatarsals. The average mean pressure, peak pressure, and contact area of the medial and lateral plantar pressures for each testing condition (flatfoot; 6-mm, 8-mm, and 10-mm LCL) were normalized to those for the intact condition. Since we hypothesized that plantar pressure shifts from the medial to the lateral part of the forefoot with progressive increments in LCL, the lateral-to-medial ratio of each outcome was calculated to characterize the relative shift in each outcome between these two regions of the forefoot.

Statistical Methods

Differences between each condition of the foot for all measures of talonavicular alignment and of plantar pressure were assessed with use of generalized estimating equations²⁴. This statistical method is appropriate for assessment of repeated measurements with small sample sizes in instances in which the assumption of normality may not hold. It is suitable for data sets with missing values²⁴. Significance was set at $p \leq 0.05$.

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Results

Angulation Analysis

The flatfoot condition showed a mean talonavicular abduction angle of 4.4° as compared with the initial condition. The 6, 8, and 10-mm LCL conditions showed mean abduction angles of -1.4°, -4.9°, and -9.2°, respectively, indicating a progressive increase in adduction at the talonavicular joint. The differences in the mean values were significant for the intact foot compared with the flatfoot condition,

Fig. 3

Fig. 3 Schematic of the reconstructed image of the talonavicular joint in the axial and sagittal planes. The talonavicular angle (shaded area) in the axial plane (A) was calculated on the basis of the relative rotation of each bone about the long axis of the tibia. The talonavicular angle (shaded area) in the sagittal plane (B) was calculated on the basis of the rotation about the transmalleolar axis. **Fig. 4** Representative plantar pressure recordings under different experimental conditions are shown. A qualitative increase in plantar pressure in the medial part of the forefoot was seen in the flatfoot condition. A qualitative increase in pressure in the lateral part of the forefoot was observed with increasing LCL.

Fig. 4

TABLE I Summary of Angulation Changes of Talonavicular Joint in Axial and Sagittal Planes

	Intact	Flatfoot	6-mm LCL	8-mm LCL	10-mm LCL
Axial plane*					
Mean	0.0	+4.4	-1.4	-4.9	-9.2
Standard deviation	0.0	+3.2	+3.1	+3.7	+4.8
Sagittal plane†					
Mean	0.0	-2.6	-0.1	+1.3	+2.9
Standard deviation	0.0	+1.2	+1.2	+2.2	+4.6

*In the axial plane, talonavicular abduction is indicated by a plus sign (+), and talonavicular adduction is indicated by a minus sign (-). All data are given as angles. †In the sagittal plane, talonavicular dorsiflexion is indicated by a plus sign (+) and talonavicular plantar flexion is indicated by a minus sign (-). All data are given as angles.

and for the flatfoot condition compared with the 6-mm LCL condition, the 6-mm condition compared with the 8-mm LCL condition, and the 8-mm condition compared with the 10-mm LCL condition ($p \leq 0.05$).

The flatfoot condition showed a mean talonavicular sagittal angle of -2.6° (plantar flexion) relative to the intact foot. The 6, 8, and 10-mm LCL showed mean dorsiflexion angles of -0.1° , 1.3° , and 2.9° , respectively, indicating an improvement or reduction of the talonavicular joint. The dif-

ferences in the mean values were significant for the intact versus flatfoot conditions, for the flatfoot condition versus the 6-mm LCL, and for the 6-mm LCL versus the 8-mm LCL ($p \leq 0.05$) (Table I). However, the difference was not significant between the 8-mm LCL versus the 10-mm LCL ($p = 0.067$). The angulation changes in the sagittal plane were relatively smaller than the changes in the axial plane with increment of LCL.

Plantar Pressure Analysis

Changes of the lateral-to-medial forefoot ratios for average mean pressure, peak pressure, and contact area were analyzed for each testing condition. The lateral-to-medial forefoot ratio for the average mean pressure decreased from 0 to -0.21 after creation of the flatfoot condition, indicating that 21% of lateral forefoot plantar pressure shifted to the medial side of the forefoot. The ratio increased to $+0.06$, $+0.26$ and $+0.46$ with respect to LCL of 6 mm, 8 mm, and 10 mm. Similarly, the lateral-to-medial forefoot ratio for peak pressure decreased from 0 to -0.23 after creation of the flatfoot condition, then increased to $+0.21$, $+0.39$, and $+0.78$ with LCL of 6 mm, 8 mm, and 10 mm, respectively. All of the ratio changes for average mean pressure and peak pressure were significant ($p \leq 0.05$). The lateral-to-medial forefoot ratio for the contact area increased to $+0.08$, $+0.22$, and $+0.41$ with LCL of 6, 8, and 10 mm, respectively. All of these changes were significant ($p \leq 0.05$). The average mean pressure (Fig. 5), peak pressure, and contact area changed approximately linearly with LCL.

Fig. 5

Data regarding the average mean pressure were collected and analyzed for the medial (first and second metatarsal heads) and lateral (third, fourth, and fifth metatarsal heads) parts of the forefoot. The pressures for the medial and lateral parts of the forefoot under each testing condition were normalized to the intact condition. Then, lateral-to-medial ratio changes were calculated to characterize the shifting of pressure between the medial and lateral parts of the forefoot. The average mean pressure shows a linear increase in the lateral-to-medial pressure ratios with each 2-mm increment in LCL. y axis = ratio for lateral-to-medial parts of the forefoot. x axis = maximum width of wedge.

Discussion

The results of this study indicate that significant differences in talonavicular angulation and plantar forefoot pressures occur with differential increments of LCL as small as 2 mm. A mild flatfoot condition was successfully created, as evidenced by an increased abduction and plantar flexion at the talonavicular joint along with increased pressure parameters on the medial side of the forefoot. The 6-mm LCL condition most closely replicated the angulation and plantar pressure of the intact condition in most specimens. Progressive increments in LCL progressively reduced the talonavicular joint and led to a

progressive increase in pressure on the lateral side of the forefoot.

Several limitations are inherent to the design of this experiment. First, in this model, load was only applied through the Achilles tendon to accentuate forefoot pressure, the major area of interest. Other muscle or tendon units were not loaded. We tried to create the midstance phase of walking gait by loading the Achilles tendon until the center of pressure was located in the midfoot region. Although this experimental condition may have emulated the plantar pressure environment for the midstance phase of walking gait, the geometric and plantar pressure data may not be representative of the physiologic condition. Second, a medializing calcaneal osteotomy or medial stabilization procedures were not combined with the LCL. The medializing calcaneal osteotomy has been shown to increase pressure in the lateral part of the forefoot with LCL^{25,26}. Hadfield et al. observed increased plantar pressures in the lateral part of the forefoot after a 1-cm medializing calcaneal osteotomy^{25,26}. Since the goal of the present study was to analyze changes in forefoot pressure occurring after different increments of LCL, medializing calcaneal osteotomy or medial stabilization procedures were not added. Finally, interpretation of the current study is limited due to the special shape and size of the metal wedges that were utilized. The metal wedges were tapered plantarly as well as medially to fit the shape of the Evans osteotomy upon distraction. Although similar results are likely to be seen for other wedge shapes, no comparisons were done between different shapes of wedges, such as the commercially available quadrangle-shaped wedges.

Evans first described how the LCL corrected calcaneal valgus deformities in children and adolescents^{1,5}. Others have applied this technique to the correction of the idiopathic flatfoot in juvenile pes planus^{27,28}. LCL has also been recommended for correcting forefoot abduction deformity associated with Stage-II posterior tibial tendon insufficiency (i.e., Stage-IIB adult acquired flatfoot deformity)^{4,13,29-31}. However, increased calcaneocuboid joint pressure and subsequent degenerative changes have been reported secondary to LCL^{13,32,33}. Furthermore, relative supination of the forefoot leads to increased loading of the lateral part of the forefoot^{12,13,29}. In this study, the calcaneal osteotomy was made 15 cm proximal to the calcaneocuboid joint and was aimed to cut between the anterior and middle facets. This technique is different from that previously described by Hintermann et al., who recommended osteotomy of the calcaneus along the anterior aspect of the posterior facet³⁴. Hintermann's technique is likely to increase the angle of Gissane at the anterior aspect of the posterior facet, thereby leading to correction in the sagittal plane. In the present experiment, there was a tendency for the distal fragment to rise superiorly with LCL, which could subsequently lead to an increase in the angle of Gissane.

Davitt et al. reported that plantar pressure measurement demonstrated a significant load shift to the lateral part of the forefoot in eleven pediatric feet after the Evans calcaneal lengthening procedure was performed⁸. A biomechanical experiment by Benthien et al. demonstrated increased pressures

in the lateral part of the forefoot after LCL along with a flexor digitorum longus transfer in a severe flatfoot model. They also found that adding a Cotton osteotomy of 15 mm decreased the increased pressure in the lateral part of the forefoot after the LCL³⁵. Scott et al. showed that LCL by either an Evans or a calcaneocuboid distraction arthrodesis increased pressures in the lateral part of the forefoot¹³. However, the increased pressure in the lateral part of the forefoot was not significantly reduced by 6 mm of Cotton osteotomy. Although the above studies showed conflicting results on the effect of a medial cuneiform osteotomy on lateral plantar pressure, they demonstrated increased pressure in the lateral part of the forefoot as the consequence of the LCL. The difference in observed effect of the medial cuneiform osteotomy may be due to the different amounts of correction performed under different testing conditions.

All of the previous clinical and biomechanical studies utilized a 10-mm bone graft or metal wedge for the LCL. The current investigation is, to our knowledge, the first study to evaluate the effect of different amounts of LCL on lateral plantar pressures of the forefoot. The senior author (J.D.) uses custom trial metal wedges of different lengths in increments of 2 mm to intraoperatively assess the amount of LCL that should be performed. This is to achieve reduction of the talonavicular joint on anteroposterior fluoroscopic images without creating excessive resistance to manual eversion. These metal wedges were utilized to obtain consistent lengthening (6, 8, and 10 mm) in each of the specimens. The average mean pressure, contact area, and peak pressure showed an approximate linear increase in pressure in the lateral part of the forefoot with respect to increments in LCL between 6 and 10 mm (Fig. 5). Since the study did not include LCL beyond 10 mm, it is unclear if the pressure in the lateral part of the forefoot would have continued to rise beyond 10 mm of LCL and, if so, in what manner.

Previous investigations have measured radiographic parameters to assess arch changes under different experimental conditions^{35,36}. Benthien et al. performed a 10-mm LCL in a flatfoot model and found improvements in the lateral talo-first metatarsal angle (-17° to -7°) and talonavicular angle (46° to 24°), along with increased pressure (from 24.6 to 33.9 kPa) in the lateral part of the forefoot³⁵. In a different cadaveric model (involving three specimens), on the basis of computed tomographic scan analysis, Dumontier et al. reported an increase of 18.6° of adduction at the talonavicular joint after 1 cm of LCL³⁷. Although relatively large degrees of corrections were observed, the flatfoot models varied in each study. During the preliminary trials that were used to design the current study, radiographic evaluation was difficult when the relative angular and translational motions of the tarsal bones were small. Therefore, optoelectronic three-dimensional motion capture was used to evaluate the angulations under the various test conditions. Although that instrumentation allowed determination of the three-dimensional spatial relationship of the talus and the navicular, lateral-view radiographs could have provided additional clinically relevant information on changes in adjacent joints, such as the calcaneocuboid joint.

Although various flatfoot models have previously been described^{18,38-40}, the present study targeted a mild flatfoot deformity. The goal of the experiment was to test the effect of different amounts of LCL. The study did not address severe flatfoot deformity, which would have required additional procedures, such as first ray stabilization and/or medializing calcaneal osteotomy, to restore the ability of the foot to withstand a dynamic axial load of 400 N. Creation of relatively mild flatfoot deformity was confirmed by relatively small changes in osseous angulation. The increase in pressures in the medial part of the forefoot with simulated deformity supported the validity of this model. Although the geometric analysis showed increased adduction of the navicular relative to the talus with increasing amounts of LCL, the talonavicular changes in the sagittal plane were small. Relatively small angular changes in the sagittal plane can perhaps be explained by the aim of creating only a mild deformity.

Despite its potential complications, including nonunion, lateral foot pain, and stress fracture, the LCL procedure is widely used for correction of Stage-IIB adult acquired flatfoot deformity⁴¹. Since flatfoot deformities vary in terms of the degree, location, and flexibility of deformity, an ideal length of LCL cannot be generalized. The current study demonstrated, however, that significant changes in plantar pressure in the lateral part of the forefoot occur secondary to increments of LCL as small as 2 mm. Increases in pressure in the lateral part of the forefoot were particularly evident with the larger wedge sizes. Geometric and plantar pressure parameters were restored most closely to the intact condition with 6 mm of lengthening, suggesting that this amount of LCL may

be appropriate in patients with mild to moderate flatfoot deformity.

The results of this study suggest that LCL that is performed within the clinically relevant range of 6 to 10 mm has the potential to be associated with a corresponding progressive increase in pressure in the lateral part of the forefoot, which can potentially lead to overload.

Appendix

A table showing a summary of dimensions of custom-designed metal wedges utilized for lateral column lengthening is available with the online version of this article as a data supplement at jbjs.org. ■

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