https://doi.org/10.1016/j.clinbiomech.2017.11.001

Finite Element Simulation on Posterior Tibial Tendinopathy: Load Transfer Alteration and Implications to the Onset of Pes Planus

Duo Wai-Chi Wong[1, 2], Yan Wang[1, 2], Aaron Kam-Lun Leung[1, 2], Ming Yang[1,3], Ming Zhang[1, 2,*]

[1]Interdisciplinary Division of Biomedical Engineering, Faculty of Engineering, The Hong Kong Polytechnic University, Hong Kong, China

[2]The Hong Kong Polytechnic University Shenzhen Research Institute, Shenzhen, China

[3]Department of Pediatric Orthopedics, The Third Affiliated Hospital, Southern Medical University, Guangzhou, China

Correspondence: Prof. Ming Zhang

Interdisciplinary Division of Biomedical Engineering, Faculty of Engineering, The Hong Kong Polytechnic University, Hung Hom, Kowloon, Hong Kong, China

Email: ming.zhang@polyu.edu.hk

Word Count: main text: 3779; abstract: 244

Abstract

Background: Posterior tibial tendinopathy is a challenging foot condition resulting in pes planus, which is difficult to diagnose in the early stage. Prior to the deformity, abnormal internal load transfer and soft tissue attenuation are anticipated. The objective of this study was to investigate the internal load transfer and strain of the ligaments with posterior tibial tendinopathy, and the implications to pes planus and other deformities.

Methods: A three-dimensional finite element model of the foot and ankle was reconstructed from magnetic resonance images of a 28-year-old normal female. Thirty bones, plantar fascia, ligaments and tendons were reconstructed. With the gait analysis data of the model subject, walking stance was simulated. The onset of posterior tibial tendinopathy was resembled by unloading the tibialis posterior and compared to the normal condition.

Findings: The load transfer of the joints at the proximal medial column was weaken by posterior tibial tendinopathy, which was compromised by the increase along the lateral column and the intercuneiforms during late stance. Besides, the plantar tarsometatarsal and cuboideonavicular ligaments were consistently over-stretched during stance. Particularly, the maximum tensile strain of the plantar tarsometatarsal ligament was about 3-fold higher than normal at initial push-off.

Interpretation: Posterior tibial tendinopathy altered load transfer of the medial column and unbalanced the load between the proximal and distal side of the medial longitudinal arch. Posterior tibial tendinopathy also stretched the midfoot plantar ligaments that jeopardized midfoot stability, and attenuated the transverse arch. All these factors potentially contributed to the progress of pes planus and other foot deformities.

Keywords: flatfoot; pes planus; posterior tibia tendon dysfunction; tenosynovitis; arch collapse

1. Introduction

Posterior tibial tendinopathy (PTT), or insufficiency is a common condition and is recognized as the major cause of acquired pes planus (flatfoot) in adults (Kohls-Gatzoulis et al., 2004; Lin et al., 2015). Although the prevalence was estimated to be 3%, the figure was believed to be underestimated because it was frequently undiagnosed (Neville et al., 2016). PTT is not necessarily symptomatic, otherwise associates with a board spectrum of generalized medical problems and pain that overlooks until secondary complaints emerged (Beeson, 2014; Bluman et al., 2007; Kohls-Gatzoulis et al., 2004). Deland et al. (2005) demonstrated that PTT is difficult to diagnose radiographically and could be noticed until ligaments are damaged.

PTT is a disabling condition and could lead to other foot deformities. Besides pes planus, patients complain of pain at the ankle and foot arch, and diminished endurance (Squires and Jeng, 2006). The intensity of the pain would increase, yet shifts capriciously as the condition exacerbates (Gluck et al., 2010). Ultimately, abnormal walking gait and difficulty in wearing shoes would be developed with deformity progression (Squires and Jeng, 2006). In fact, the importance of tibialis posterior is indisputable. It is the major dynamic stabilizer of the arch and midfoot and facilitates a rigid level at the midfoot for other structures to function effectively (Alvarez et al., 2006). Impairment of the posterior tibial tendon not only attenuates stability but may alter load transfer among joints and other stabilization structures (Imhauser et al., 2004).

It is difficult to investigate or determine the biomechanics of PTT onset. The majority of research pertained to clinical description and management (Beeson, 2014). The interaction between different intrinsic and extrinsic factors was identified as the risk factors (Beeson, 2014). However, the pathomechanism of PTT remains poorly understood (Beeson, 2014). Some scientists attempted to uncover the etiology of PTT by kinematics and plantar pressure studies. Levinger et al. (2010) found that patients with pes planus had a greater rearfoot eversion, which was also demonstrated in a cadaveric study that simulated pes planus by sectioning the peritalar constraints and unloading the posterior tibialis (Watanabe et al., 2013). Plantar pressure studies showed that plantar load failed to transmit anteriorly with tibialis posterior unloaded (Neville et al., 2013), in addition to the medial shift, which was found to be exaggerated by the pes planus deformity (Imhauser et al., 2004).

However, it remains infeasible to assess internal load transfer and soft tissue strain of walking gait by in vivo experiment and cadaveric study, and this information would be necessary to better identify the mechanism of deformity initialized by PTT. Computer simulation via finite element (FE) analysis can reveal the internal biomechanical information in a controlled environment and predefined condition, which becomes a versatile tool to predict the pathomechanism of the musculoskeletal system and to assist surgical planning (Ni et al., 2016; Wang et al., 2015; Wang et al., 2016; Wong et al., 2014b; Yu et al., 2016). The objective of this study is to assess the load transfer of the foot and ankle, so as the strain of the midfoot ligaments during walking stance by a theoretical FE foot model with onset PTT simulated. It was hypothesized that PTT disturbs normal load transfer, attenuate plantar midfoot ligaments, and therefore jeopardize midfoot stability and expose the foot to pes planus progression.

2. Methods

2.1 Geometry Reconstruction and Assembly

A healthy female subject, aged 28, 165 cm tall and weighed 54 kg, was recruited. She reported no musculoskeletal disorder, pain, and previous foot surgery. The participant signed an informed consent form prior to the start of the experiment.

The magnetic resonance images of the right foot were obtained from a 3.0-T scanner (Trio-Tim, Siemens Medical Solutions, Erlangen, Germany). The foot was scanned at neutral and non-weight bearing conditions facilitated by an ankle-foot-orthosis, such that the soft tissue was minimally compressed. The images were scanned at T1 sequence, 1 mm slice interval and had a pixel size of 0.625 mm. The images were segmented and reconstructed by the software Mimics v10 (Materialise, Leuven, Belgium) and Rapidform XOR2 (INUS Technology Ltd., Seoul, Korea).

As shown in Figure 1, the thirty foot bones including the distal portion of tibia and fibula, and the encapsulated soft tissue were reconstructed. Ligaments, fascia, and tendons were modeled based on the constructed bony structures and the clinical images, which was subsequently confirmed by an orthopaedic surgeon. Forefoot ligaments, including the collateral ligaments at the metatarsophalangeal joints, deep transverse metatarsal ligaments, and the sesamoid ligaments, while the other ligaments were modelled as shells. Muscles/tendons were simplified as uniaxial connectors. The interior surface of the encapsulated soft tissue was tied to the bony structures. Since the geometry of the cartilages was too small to be constructed, they were substituted by the contact algorithm assigning to the bone-to-bone interactions. The contact assumed that the interaction was frictionless with a non-linear contact stiffness (Athanasiou et al., 1998). The coefficient of friction between the encapsulated soft tissue and ground plate was assumed 0.6 (Zhang and Mak, 1999).

2.2 Mesh Creation

The mesh was created using the finite element software, Abaqus 6.11 (Dassault Systèmes, Vélizy-Villacoublay, France). Linear tetrahedral elements (C3D4) was constructed in the solid parts, such as the bones and the encapsulated soft tissue. Quadrilateral elements (S4R) were created on shell parts, while truss parts were assigned with two-node truss elements (T3D2).

The mesh size was approximately 4 mm for the encapsulated soft tissue and 2.5 mm for the other structures. Local refinement of mesh was carried out on small parts, contact regions, and abrupt geometry. There were 124,730 elements in the bone. The encapsulated soft tissue was meshed with 84,258 tetrahedral elements (C3D4) and covered by 9,356 triangular elements (S3) representing the skin layer. Mesh convergence test was previously conducted with an estimated error less than 5% (Wong et al., 2015).

2.3 Material Properties

All material properties of the model parts were determined from existing literature. The bones were linearly elastic with an elastic modulus of 10 GPa and Poisson's ratio of 0.3 (Chen et al., 2003). The encapsulated soft tissue was modeled as hyperelastic material with the second-order polynomial strain energy potential equation using the coefficients, $C_{10} = 0.08556$ Nmm^{-2}, $C_{01} = -0.05841$ Nmm^{-2}, $C_{20} = 0.03900$ Nmm^{-2}, $C_{11} = -0.02319$ Nmm^{-2}, $C_{02} = 0.00851$ Nmm^{-2}, $D_1 = 3.65273$ mm^{2}N^{-1} (Lemmon et al., 1997) while the skin was assigned hyperelasticity with the first-order Ogden model, using the coefficients, $\mu = 0.122$kPa and $\alpha = 18$ (Gu et al., 2010). The thickness of the skin was assumed 2.0 mm (Pailler-Mattei et al., 2008). Forefoot ligaments modelled with truss elements were assigned with 264.8 MPa elastic modulus (Siegler et al., 1988) and a cross-section area of 18.4 mm^2 (Milz et al., 1998). Other ligaments modelled with shell elements were assigned with an approximated thickness of 1.5 mm (Cheung et al., 2005). The plantar fascia was modeled as slip-ring components with specific stiffness on different columns from 182.2 N/mm to 232.5 N/mm (Kitaoka et al., 1994).

2.4 Boundary and Loading Conditions

The boundary and loading conditions were acquired from gait experiment of the model subject. Five instants during walking stance were identified by representative data of the vertical ground reaction force (GRF) and shank-to-ground angle profile. These instants, represented by percentage stance phase, were extracted and named: 15% (neutral stance), 25% (GRF first peak), 45% (GRF valley), 60% (initial push-off) and 75% (GRF second peak). As shown in Figure 1, the tibial and fibula ends were fixed. The GRF was applied under the floor plate and the floor plate was rotated by the shank-to-ground angle.

The muscle forces were estimated by the multiplication of the maximum muscle capacity (Arnold et al., 2010) and the percentage muscle activation from an electromyography study during walking gait (Perry and Burnfield, 1993). The Achilles tendon force was obtained from another study (Fröberg et al., 2009), since estimating tendon force from triceps surae (gastrocnemius and soleus) was difficult.

2.5 Model Output and Analysis

The simulation was conducted with Abaqus 6.11 (Dassault Systèmes, Vélizy-Villacoublay, France) using the standard quasi-static solver. The onset of PTT condition was mimicked by unloading the tibialis posterior which was then compared to that with normal tibialis posterior loading (Imhauser et al., 2004; Wong et al., 2017). The joint forces of the rearfoot, medial column and midfoot were analyzed. The joint force was represented by the contact force of the bone-to-bone interaction, which incorporated the non-linear contact stiffness of the cartilage. The tensile strains of the seven selected plantar ligaments, including the plantar first metatarsocuneiform, intermetatarsal, tarsometatarsal, intercuneiform, cuneocuboid, naviculocuneiform, cuboideonavicular ligaments, were investigated.

2.6 Sensitivity Analysis

The sensitivity test targeted on the variance of the insertion site of selected ligaments because the strain of the ligaments was one of the main outcome measures of this study. The insertion locations of the seven selected ligaments, as shown in Figure 4, were moved randomly and differently in the proximal-distal direction with a range of 3 mm, which was the reported maximum variance of insertion in a cadaveric study (Campbell et al., 2014). The randomized values were generated by spreadsheet software (Microsoft Excel, Microsoft Corporation, Washington, USA). Eleven sets of randomization were conducted such that 12 sets of data were produced including the reconstructed model. This approach of simple random sampling in sensitivity analysis was also reported in the literature (Clemson et al., 1995). The sensitivity was evaluated using the coefficient of variation (CV%) which was expressed as the ratio of the standard deviation to the mean value. Since the evaluation criteria for the CV% in this application was lacking, we assumed that a CV% over 18% was high, with reference to a relevant kinematic study (Yavuzer et al., 2008).

2.7 Experimental Validation

The model subject was invited to conduct walking trials with the plantar pressure measurement (F-scan® System, Tekscan, USA). The subject was instructed to walk barefoot at self-selected comfortable speed with the sensor adhered to the plantar surface. Six successive walking steps were completed and the maximum plantar pressures at different phases of walking were extracted and averaged over trials. The agreement between the plantar pressure measurement and finite element prediction was evaluated using Intraclass correlation (ICC), based on a mean-rating of the five time instants, consistency, two-way mixed model. The statistical analysis was carried out using SPSS 21 (SPSS Inc., Chicago, United States).

Besides, the FE model and platform was previously constructed and validated. The FE prediction was compared with the plantar pressure distribution of the model subject, cadaveric experiment and pendulum impact experiments (Wong et al., 2016; Wong et al., 2015; Wong et al., 2014b), despite there were some differences in configurations and applications among the previous work.

3. Results

3.1 Joint Forces

As shown in Figure 2, PTT reduced the load transfer across the ankle joint and the talonavicular joint remarkably. During initial push-off, the reduction both accounted for more than 10%. The reduction was then compromised by an apparent increase of 35% at the calcaneocuboid joint, particularly at the GRF second peak instant.

The subtalar joint and first metatarsophalangeal joint did not demonstrate substantial change (Figure 2 and Figure 3) in the magnitude of the joint force. Yet, the two consecutive joints at the medial column, the medial cuneonavicular joint, and the first tarsometatarsal joint, showed opposite change during initial push-off. The load at the proximal joint sharply decreased but increased at the distal joint. On the other hand, PTT reduced the intercuneiform joint forces during early stance but increased during late stance, compared to normal.

3.2 Ligament Strain

Figure 4 presents the maximum tensile strain of the midfoot plantar ligaments at GRF first peak and initial push-off. At GRF first peak, PTT reduced the tensile strain of the ligaments, except the plantar tarsometatarsal ligament and cuboideonavicular ligaments that increased from 6.5% to 7.0% and from 2.3% to 5.0% respectively.

Conversely, PTT increased tensile strain at initial push-off, except the plantar cuneocuboid ligament. Moreover, the strain of the plantar tarsometatarsal ligament was about 3-fold higher compared to the normal condition.

3.3 Sensitivity Analysis

The average CV% was 11.3% and 13.8% respectively for the joint forces and ligament strain and we considered the findings as insensitive in general (CV% < 18%). However, it should be noted that the first tarsometatarsal joint force and the strain of the plantar cuneocuboid ligament presented relative high level of variation, which were 37.6% and 24.3% respectively. The accuracy and interpretation on these outcomes with respect to external validity should be carefully noted.

3.4 Experimental Validation

The ICC of the maximum plantar pressure indicated moderate correlation with an average measure of 0.68, while that of the plantar contact area showed excellent correlation with an average measure of 0.93.

3.5 Comparison with Existing Literature

In Figure 5, the change of the arch height from normal to PTT condition is compared to existing literature. The arch height decreased by 0.6 mm with PTT, which was agreeable with existing finite element prediction and cadaveric experiments, given the variances among specimens/individuals and differences in configurations (Cheung and Zhang, 2006; Imhauser et al., 2004; Kitaoka et al., 1997).

Figure 6 illustrates the plantar pressure distribution during initial push-off. The predicted peak pressure shifted medially after PTT, which was consistent with other observations. A cadaveric study that unloaded the posterior tibialis found a medial shift of forefoot loading (Imhauser et al., 2004). Patients with PTT also showed decreased lateral forefoot loading during terminal stance (Neville et al., 2013).

4. Discussion

Posterior Tibial Tendinopathy (PTT) is described as a silent disabling condition; it is difficult to detect in the early stage and is, therefore, likely to progress into pes planus (Singh et al., 2012). Angular or kinematic changes of pes planus were widely studied (Haleem et al., 2014; Spratley et al., 2013; Zhang et al., 2015), but they could fail to fully address the pathomechanism since the onset does not present observational or radiographic changes (Deland et al., 2005). The significance of this research lies in its potential to reveal the internal load transfer during walking stance and to identify the etiology of pes planus, or PTT, at its onset stage, based on a theoretical model. It could also provide clues on the correction of biomechanical environment facilitated by surgical or orthotic treatments (Vulcano et al., 2013).

Tibialis posterior and triceps surae (Achilles tendon) should have been activated in the early stance. PTT inevitably reduced the load on the foot, and the impaired stability, inflicted by PTT, limited the force transmission efficiency of triceps surae (Neville et al., 2013). The joint forces were generally enervated by PTT during early stance, as demonstrated in our prediction. During late stance, the balance between foot invertors and evertors is crucial to the midfoot stability and adequate load transfer to the forefoot. Since the main branch of the tibialis posterior inserts in the medial column, PTT could slash the medial load transfer along talus, navicular and the medial cuneiform. The load transfer through the lateral column and intercuneiforms was then compromised and increased.

The tensile strain of the midfoot plantar ligaments was increased in PTT condition during late stance and reflected the loss and complement of midfoot stability. Our prediction indicated that the plantar tarsometatarsal ligaments were consistently over-stretched throughout stance. Midfoot splay, along with the deterioration of the transverse arch, was anticipated as these ligaments failed (Hicks, 1954). In addition, the large increase of strain on the plantar tarsometatarsal ligament would weaken the relationship between the forefoot and midfoot and may induce pes planus and other forefoot deformities (Johnson and James, 2005; Zhang et al., 2013).

In addition, the medial longitudinal arch structure was under-loaded proximally and over-loaded distally. The imbalance of the load-bearing between the proximal and distal side could be the cause of the instability of the keystone, the navicular bone. Despite the displacement of the navicular depression was minor, the change is accumulative that could give rise to subsequent arch collapse chronically (Kamiya et al., 2012). Similarly, a subtle change of soft tissue accumulates overtime and remarkably reduces stability (Lever and Hennessy, 2016). The combined effect of altered joint loading in the medial column and ligament strain not only lead to pes planus but also could predispose the risk of metatarsus primus varus and hallux valgus (Wong et al., 2014a; Wong et al., 2014b).

There were some limitations in this study. Besides simplifications and assumptions inherent to the geometry, material and the loading case, we assumed that the boundary and loading conditions between normal and PTT were similar, under the premise that the immediate effect (onset) was studied and no structural change was anticipated. Some studies demonstrated that Stage II PTT patients have different spatio-temporal gait parameters and foot joint angles (Ness et al., 2008). However, there was no substantial evidence that the boundary and loading conditions, including the shank-to-ground angles and the resultant ground reaction forces was significantly changed (Ledoux and Hillstrom, 2002; Levinger et al., 2010). While investigation of the foot angles could have implications to the evaluation of PTT, this study meanwhile focused on the evaluation of joint loading and ligament strain, otherwise difficult to be assessed thought experimental approach. Pre-stain of ligaments and plantar fascia was not considered in this model. To the extent of our knowledge, we believe that there is no finite element foot model taking pre-strain into account and this information is also lacking (Morales-Orcajo et al., 2016), despite that pre-strain has been commonly considered in knee models (Galbusera et al., 2014). Neglecting the influence of pre-strain may underestimate the strain of the ligaments and the joint loading.

On the other hand, the external validity of the study was hindered by the single subject design which was commonly faced by research using a theoretical approach, such as the finite element method (Ren et al., 2016). A sensitivity test was conducted and incorporated into the data analysis to account for some variances in population. There were some outcome measures demonstrating high variance during the sensitivity test, such that interpretation, particularly on the first tarsometatarsal joint force and the strain of the plantar cuneocuboid ligament should be treated carefully. In fact, the sensitivity test only considered a single factor (the insertion of some ligaments), whilst inter-subject variability includes many other variations, such as bone morphology, insertion, and size of ligaments, etc. Meanwhile, it remains difficult and impractical to reconstruct a few sophisticated foot models together with corresponding gait experiments and validations. Subjects with the required features were also difficult to be identified, for example, patients with an onset of PTT or pes planus.

To this end, we have selected a subject with typical physique and foot characteristics, which we viewed the model as representative. Validation was conducted on the model subject using plantar pressure measurement. While most of the previous validations of finite element foot models observed the plantar pressure distribution qualitatively or by comparing the peak pressure (Morales-Orcajo et al., 2016), recent studies attempted to quantify the degree of agreement using statistical approach, such as ICC or Bland-Altman analysis (Edwards and Troy, 2012). A qualitative comparison was reported in our previous study (Wong et al., 2014a) and, in this study, ICC was conducted. Although the maximum plantar pressure of the experiment was consistently lower than the prediction, the comparison using ICC demonstrated moderate to excellent correlation (0.68 to 0.93). Some deviations could be due to the fact that the sensors redistributed the plantar pressure slightly. We decided that the validation demonstrated sufficient agreement with the agreeable qualitative comparison on the plantar pressure pattern (Wong et al., 2014a), moderate-to-excellent correlation and uncertainty identified (Anderson et al., 2007). In addition, the sensitivity test showed that majority of the outcomes are not sensitive to the variance of ligament insertion. We assumed that the model should be adequately reliable based on our validation and verification measures as recommended (Viceconti et al., 2005).

PTT is difficult to detect in the early stage, and thus assuming the state of onset in the simulation was a challenging task in this study. Clinically, PTT was categorized into four stages (Bubra et al., 2015). Stage I presents no clinical deformity but a partial dysfunction of the tendon. Stage II shows non-functionality of the tendon and resulting deformity. Stage III PTT demonstrates irreversible foot deformity while tibiotalar degeneration would be found in Stage IV patients. We determined the stage of onset as the in-between of stage I and II, with the loss of tendon function, but before the start of deformity. A sensitivity analysis on the level of functional loss could help understand the gradual change of the biomechanical environment over the stages. Future study should challenge different stages and severities of PTT, and the biomechanics of surgical and orthotic treatments.

5. Conclusions

PTT changed the load transfer mechanism and the strain of the midfoot plantar ligaments. These changes disturbed the load balance of the medial longitudinal arch; splayed and collapsed the transverse arch, which potentially contributed to pes planus and other deformities.

6. Acknowledgement

The work was supported by the University Grants Committee of Hong Kong, China, under the General Research Fund (PolyU 152065/17E, PolyU 152002/15E, PolyU 152216/14E), and the Key Program of the National Natural Science Foundation of China (Grant No.: 11732015).

References

Alvarez, R.G., Marini, A., Schmitt, C., Saltzman, C.L., 2006. Stage I and II posterior tibial tendon dysfunction treated by a structured nonoperative management protocol: an orthosis and exercise program. Foot & ankle international 27, 2-8.

Anderson, A.E., Ellis, B.J., Weiss, J.A., 2007. Verification, validation and sensitivity studies in computational biomechanics. Computer methods in biomechanics and biomedical engineering 10, 171-184.

Arnold, E.M., Ward, S.R., Lieber, R.L., Delp, S.L., 2010. A model of the lower limb for analysis of human movement. Annals of biomedical engineering 38, 269-279.

Athanasiou, K., Liu, G., Lavery, L., Lanctot, D., Schenck Jr, R., 1998. Biomechanical topography of human articular cartilage in the first metatarsophalangeal joint. Clinical orthopaedics and related research 348, 269-281.

Beeson, P., 2014. Posterior tibial tendinopathy: what are the risk factors? Journal of the American Podiatric Medical Association 104, 455-467.

Bluman, E.M., Title, C.I., Myerson, M.S., 2007. Posterior tibial tendon rupture: a refined classification system. Foot and ankle clinics 12, 233-249.

Bubra, P.S., Keighley, G., Rateesh, S., Carmody, D., 2015. Posterior tibial tendon dysfunction: an overlooked cause of foot deformity. Journal of family medicine and primary care 4, 26.

Campbell, K.J., Michalski, M.P., Wilson, K.J., Goldsmith, M.T., Wijdicks, C.A., LaPrade, R.F., Clanton, T.O., 2014. The ligament anatomy of the deltoid complex of the ankle: a qualitative and quantitative anatomical study. J Bone Joint Surg Am 96, e62.

Chen, W.-P., Ju, C.-W., Tang, F.-T., 2003. Effects of total contact insoles on the plantar stress redistribution: a finite element analysis. Clinical Biomechanics 18, S17-S24.

Cheung, J.T.-M., Zhang, M., Leung, A.K.-L., Fan, Y.-B., 2005. Three-dimensional finite element analysis of the foot during standing—a material sensitivity study. Journal of Biomechanics 38, 1045-1054.

Cheung, J.T., Zhang, M., 2006. Finite Element and Cadaveric Simulations of theMuscular Dysfunction of Weightbearing Foot. HKIE Transactions 13, 8-15.

Clemson, B., Tang, Y., Pyne, J., Unal, R., 1995. Efficient methods for sensitivity analysis. System Dynamics Review 11, 31-49.

Deland, J.T., Richard, J., Sung, I.-H., Ernberg, L.A., Potter, H.G., 2005. Posterior tibial tendon insufficiency: which ligaments are involved? Foot & ankle international 26, 427-435.

Edwards, W.B., Troy, K.L., 2012. Finite element prediction of surface strain and fracture strength at the distal radius. Medical engineering & physics 34, 290-298.

Fröberg, Å., Komi, P., Ishikawa, M., Movin, T., Arndt, A., 2009. Force in the achilles tendon during walking with ankle foot orthosis. The American journal of sports medicine 37, 1200-1207.

Galbusera, F., Freutel, M., Dürselen, L., D'Aiuto, M., Croce, D., Villa, T., Sansone, V., Innocenti, B., 2014. Material models and properties in the finite element analysis of knee ligaments: a literature review. Frontiers in bioengineering and biotechnology 2.

Gluck, G.S., Heckman, D.S., Parekh, S.G., 2010. Tendon Disorders of the Foot and Ankle, Part 3 The Posterior Tibial Tendon. The American journal of sports medicine 38, 2133-2144.

Gu, Y., Li, J., Ren, X., Lake, M.J., Zeng, Y., 2010. Heel skin stiffness effect on the hind foot biomechanics during heel strike. Skin Research and Technology 16, 291-296.

Haleem, A.M., Pavlov, H., Bogner, E., Sofka, C., Deland, J.T., Ellis, S.J., 2014. Comparison of deformity with respect to the talus in patients with posterior tibial tendon dysfunction and controls using multiplanar weight-bearing imaging or conventional radiography. J Bone Joint Surg Am 96, e63.

Hicks, J., 1954. The mechanics of the foot: II. The plantar aponeurosis and the arch. Journal of anatomy 88, 25-30.

Imhauser, C.W., Siegler, S., Abidi, N.A., Frankel, D.Z., 2004. The effect of posterior tibialis tendon dysfunction on the plantar pressure characteristics and the kinematics of the arch and the hindfoot. Clinical Biomechanics 19, 161-169.

Johnson, J.E., James, R.Y., 2005. Arthrodesis techniques in the management of stage-II and III acquired adult flatfoot deformity. J Bone Joint Surg Am 87, 1865-1876.

Kamiya, T., Uchiyama, E., Watanabe, K., Suzuki, D., Fujimiya, M., Yamashita, T., 2012. Dynamic effect of the tibialis posterior muscle on the arch of the foot during cyclic axial loading. Clinical Biomechanics 27, 962-966.

Kitaoka, H.B., Luo, Z.P., An, K.-N., 1997. Effect of the posterior tibial tendon on the arch of the foot during simulated weightbearing: biomechanical analysis. Foot & ankle international 18, 43-46.

Kitaoka, H.B., Luo, Z.P., Growney, E.S., Berglund, L.J., An, K.-N., 1994. Material properties of the plantar aponeurosis. Foot & ankle international 15, 557-560.

Kohls-Gatzoulis, J., Angel, J.C., Singh, D., Haddad, F., Livingstone, J., Berry, G., 2004. Tibialis posterior dysfunction: a common and treatable cause of adult acquired flatfoot. Bmj 329, 1328-1333.

Ledoux, W.R., Hillstrom, H.J., 2002. The distributed plantar vertical force of neutrally aligned and pes planus feet. Gait & posture 15, 1-9.

Lemmon, D., Shiang, T., Hashmi, A., Ulbrecht, J.S., Cavanagh, P.R., 1997. The effect of insoles in therapeutic footwear--a finite element approach. Journal of Biomechanics 30, 615-620.

Lever, C.J., Hennessy, M.S., 2016. Adult flat foot deformity. Orthopaedics and Trauma 30, 41-50.

Levinger, P., Murley, G.S., Barton, C.J., Cotchett, M.P., McSweeney, S.R., Menz, H.B., 2010. A comparison of foot kinematics in people with normal-and flat-arched feet using the Oxford Foot Model. Gait & posture 32, 519-523.

Lin, Y.-C., Mhuircheartaigh, J.N., Lamb, J., Kung, J.W., Yablon, C.M., Wu, J.S., 2015. Imaging of Adult Flatfoot: Correlation of Radiographic Measurements With MRI. American Journal of Roentgenology 204, 354-359.

Milz, P., Mhz, S., Steinborn, M., Mittlmeier, T., Putz, R., Reiser, M., 1998. Lateral ankle ligaments and tibiofibular syndesmosis: 13-MHz high-frequency sonography and MRI compared in 20 patients. Acta Orthopaedica Scandinavica 69, 51-55.

Morales-Orcajo, E., Bayod, J., de Las Casas, E.B., 2016. Computational foot modeling: scope and applications. Archives of Computational Methods in Engineering 23, 389-416.

Ness, M.E., Long, J., Marks, R., Harris, G., 2008. Foot and ankle kinematics in patients with posterior tibial tendon dysfunction. Gait & posture 27, 331-339.

Neville, C., Bucklin, M., Ordway, N., Lemley, F., 2016. An Ankle-Foot Orthosis With a Lateral Extension Reduces Forefoot Abduction in Subjects With Stage II Posterior Tibial Tendon Dysfunction. journal of orthopaedic & sports physical therapy 46, 26-33.

Neville, C., Flemister, A.S., Houck, J., 2013. Total and distributed plantar loading in subjects with stage II tibialis posterior tendon dysfunction during terminal stance. Foot & ankle international 34, 131-139.

Ni, M., Wong, D.W.-C., Mei, J., Niu, W., Zhang, M., 2016. Biomechanical comparison of locking plate and crossing metallic and absorbable screws fixations for intra-articular calcaneal fractures. Science China Life Sciences 59, 958-964.

Pailler-Mattei, C., Bec, S., Zahouani, H., 2008. In vivo measurements of the elastic mechanical properties of human skin by indentation tests. Medical engineering & physics 30, 599-606.

Perry, J., Burnfield, J.M., 1993. Gait analysis: normal and pathological function. Slack.

Ren, S., Wong, D.W.-C., Yang, H., Zhou, Y., Lin, J., Zhang, M., 2016. Effect of Pillow Height on the Biomechanics of Head-neck Complex: Investigation on Crano-cervical Pressure and Cervical Spine Alignment. PeerJ 4.

Siegler, S., Block, J., Schneck, C.D., 1988. The mechanical characteristics of the collateral ligaments of the human ankle joint. Foot & ankle international 8, 234-242.

Singh, R., King, A., Perera, A., 2012. Posterior tibial tendon dysfunction: a silent but disabling condition. British journal of hospital medicine (London, England: 2005) 73, 441-445.

Spratley, E.M., Matheis, E.A., Hayes, C.W., Adelaar, R.S., Wayne, J.S., 2013. Validation of a population of patient-specific adult acquired flatfoot deformity models. Journal of Orthopaedic Research 31, 1861-1868.

Squires, N.A., Jeng, C.L., 2006. Posterior tibial tendon dysfunction. Operative Techniques in Orthopaedics 16, 44-52.

Viceconti, M., Olsen, S., Nolte, L.-P., Burton, K., 2005. Extracting clinically relevant data from finite element simulations. Clinical Biomechanics 20, 451-454.

Vulcano, E., Deland, J.T., Ellis, S.J., 2013. Approach and treatment of the adult acquired flatfoot deformity. Current reviews in musculoskeletal medicine 6, 294-303.

Wang, Y., Li, Z., Wong, D.W.-C., Zhang, M., 2015. Effects of ankle arthrodesis on biomechanical performance of the entire foot. PloS one 10, e0134340.

Wang, Y., Wong, D.W.-C., Zhang, M., 2016. Computational models of the foot and ankle for pathomechanics and clinical applications: a review. Annals of biomedical engineering 44, 213-221.

Watanabe, K., Kitaoka, H.B., Fujii, T., Crevoisier, X., Berglund, L.J., Zhao, K.D., Kaufman, K.R., An, K.-N., 2013. Posterior tibial tendon dysfunction and flatfoot: analysis with simulated walking. Gait & posture 37, 264-268.

Wong, D.W.-C., Niu, W., Wang, Y., Zhang, M., 2016. Finite element analysis of foot and ankle impact injury: risk evaluation of calcaneus and talus fracture. PloS one 11, e0154435.

Wong, D.W.-C., Wang, Y., Chen, T.L.-W., Leung, A.K.-L., Zhang, M., 2017. Biomechanical consequences of subtalar joint arthroereisis in treating posterior tibial tendon dysfunction: a theoretical analysis using finite element analysis. Computer methods in biomechanics and biomedical engineering.

Wong, D.W.-C., Wang, Y., Zhang, M., Leung, A.K.-L., 2015. Functional restoration and risk of non-union of the first metatarsocuneiform arthrodesis for hallux valgus: A finite element approach. Journal of Biomechanics 48, 3142-3148.

Wong, D.W.-C., Zhang, M., Leung, A.K.-L., 2014a. First ray model comparing normal and hallux valgus feet. Computational Biomechanics of the Musculoskeletal System, 49.

Wong, D.W.-C., Zhang, M., Yu, J., Leung, A.K.-L., 2014b. Biomechanics of first ray hypermobility: An investigation on joint force during walking using finite element analysis. Medical engineering & physics 36, 1388-1393.

Yavuzer, G., Öken, Ö., Elhan, A., Stam, H.J., 2008. Repeatability of lower limb three-dimensional kinematics in patients with stroke. Gait & posture 27, 31-35.

Yu, J., Wong, D.W.-C., Zhang, H., Luo, Z.-P., Zhang, M., 2016. The influence of high-heeled shoes on strain and tension force of the anterior talofibular ligament and plantar fascia during balanced standing and walking. Medical engineering & physics 38, 1152-1156.

Zhang, M., Mak, A., 1999. In vivo friction properties of human skin. Prosthetics and orthotics international 23, 135-141.

Zhang, Y.-J., Xu, J., Wang, Y., Lin, X.-J., Ma, X., 2015. Correlation between hindfoot joint three-dimensional kinematics and the changes of the medial arch angle in stage II posterior tibial tendon dysfunction flatfoot. Clinical Biomechanics 30, 153-158.

Zhang, Y., Xu, J., Wang, X., Huang, J., Zhang, C., Chen, L., Wang, C., Ma, X., 2013. An in vivo study of hindfoot 3D kinetics in stage II posterior tibial tendon dysfunction (PTTD) flatfoot based on weight-bearing CT scan. Bone and Joint Research 2, 255-263.

Figure Legends

Figure 1. Geometry, and the boundary and loading conditions of the finite element model. Note: This figure was reproduced and modified from (Wong et al., 2016) under the Creative Common Attribution.

Figure 2. Joint forces of the rearfoot comparing simulated PTT and normal condition during stance: (a) subtalar joint; (b) ankle (talocrural joint); (c) calcaneocuboid joint; (d) talonavicular joint.

Figure 3. Joint forces of the medial column and midfoot comparing simulated PTT and normal condition during stance: (a) medial cuneonavicular joint; (b) first tarsometatarsal joint; (c) first metatarsophalangeal joint; (d) intercuneiform (lateral-intermediate) joint; (e) intercuneiform (medial-intermediate) joint.

Figure 4. Maximum tensile strain of the plantar midfoot ligaments comparing the simulated PTT and normal condition at GRF first peak and initial push-off.

Figure 5. The change of arch height from normal to simulated PTT condition under initial push-off compared with existing published data.

Figure 6. Plantar pressure distribution of the normal and simulated PTT condition during initial push-off.