Abstract

Lapidus arthrodesis is used in the treatment of hallux valgus, first ray instability and midfoot arthritis. Despite being commonly performed, few studies have addressed the regional biomechanical implications of this procedure. Our objective was to analyse the stress concentrations caused by two commonly performed Lapidus arthrodesis on surrounding bone and soft tissue structures of the foot. A finite element model was used to simulate the normal intact foot and scenarios of tissues deficiencies that are often associated when a Lapidus arthrodesis is performed. Our model includes all the foot bones, cartilage and major tendons and ligaments that support the foot arch. Both tensile stress and compressive forces were measured in the midfoot bones, joints and tibialis posterior tendon. Results showed that the classical Lapidus arthrodesis is associated with an increase of about 76% in the compressive stress generated around the first and second cuneiform joint, while the isolated metatarsocuneiform

hallucis longus (FHL), flexor digitorum longus (FDL), both peroneal tendons (Brevis and Longus) and cartilages. Due to the difficulty of segmenting the spring ligament and both plantar ligaments (short and long) from CT images, these tissues were reconstructed based on their anatomy, using body atlas and surgeons' guidance. A statement on ethical approval by a committee is not required for this work, because neither intervention nor any contact was made with the volunteer whose foot was used for reconstruction and modelling. However, we have the informed consent signed by this person accepting the use of their images to be used for foot modelling. This model has been already used to analyse the adult acquired flatfoot development [13-15]. The complete FE model is shown in Figure 1.

Figure 1: Detailed description of the FE model.

Meshing of the model

The meshing of the model was performed using ICEM CFD V.15 (Canonsburg, Pennsylvania, United States), generating 28 cortical bone pieces, 24 trabecular bone pieces, 26 cartilage segments, 6 tendons, 3 ligaments and the plantar fascia. To optimize the mesh size of each segment and check the mesh quality, a trial error approach was employed, following the recommendations of Burkhart [16]. They stated that to obtain reliable results, the total number of inaccurate elements must be less than 5%. The following conditions were considered in order to achieve a reasonable mesh size without compromising the calculation time: a minimum mesh size sufficiently small to fit into the tightest segments, a maximum mesh size consistent with the minimum, avoidance of large differences in element size between regions, a mesh accuracy of more than 99% of the elements being better than 0.2 mesh quality (Jacobians) and checking that the poor elements were located away from the region of greatest interest (Hindfoot bones, metatarsals, PF and SL). The equilibrium was found with 265,547 linear tetrahedral elements (C3D4). All parameters were within good mesh quality ratios (Table 1). Both finite element analysis and simulations were conducted with Abaqus/CAE 6.14-1 (Dassault Systémes, Vélizy-Villacoublay, France) using the available nonlinear geometry solver.

Quality metric	(Assessment criteria)	Accurate elements	Inaccurate elements
Element Jacobians	>0.2	99.2%	0.8%
Aspect ratio	<3	95.5%	4.5%
Min. angles	>30	97.6%	2.4%
Max. angles	>120	98.7%	1.3%

Table 1: Mesh quality metrics based on Burkhart et al. (2013) recommendations.

Biomechanical properties of model tissues

Tissue properties (Young's modulus and Poisson's ratio) of cortical bone, trabecular bone, ligaments and plantar fascia were assigned in accordance with published data: Cortical bone (E=17000 MPa, v=0.3), trabecular bone (E=700 MPa, v=0.3), ligaments (E=250 MPa, v=0.28) and Plantar fascia (E=240 MPa, v=0.28) [13]. Tendons and cartilage were modelled based on the Ogden model (hyperelastic material), using parameters previously reported in the literature [17-19].

The stress and compressive forces in the midfoot joints for the classical Lapidus and the modified Lapidus were evaluated. Simulated fusions were performed with a complete union of the cuneiform metatarsal joint (CM) and union between the first and second cuneiforms (Figure 2) by replacing the joint cartilage with cortical bone tissue. Fixation elements such as plates or screws were not included, because a complete joint fusion was simulated.

Figure 2: Explanation of how the arthrodesis was simulated. The cartilage material was replaced by a cortical bone in each of the fused joints.

Loading and boundary conditions

A simulated load was applied in the vertical direction and at 10 degrees of inclination. The distribution was as follows: Tibiotalar load transmission was set at 90% and Fibula Talus load transmission at 10% [14,15]. Tendon traction forces were included as described by [20]. To simulate ground contact, all simulations were performed by maintaining fixed nodes in the lower part of the calcaneus and blocking vertical displacement (z-axis) of the lower nodes of the first and fifth metatarsals (Figure 3) [13].

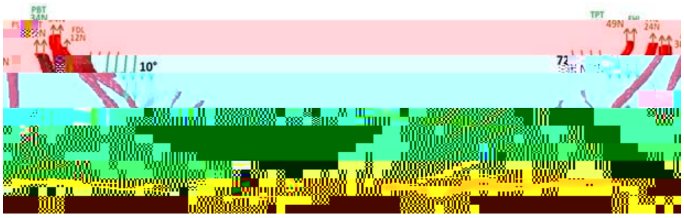

Figure 3: Boundary and loading settings applied to the foot model

Model validation

The model used in this analysis has been validated by other studies in relation to adult acquired flatfoot deformity [13]. These studies measured the vertical displacement of anatomical points in two different loading conditions: light loading and normal stance loading. Both were measured in the sagittal plane. This validation method compares the results obtained from the model against the average of values measured from real lateral RX images [21].

Model analysis and evaluation criteria

The biomechanical stress and compressive forces were quantified using the field output spectrum available in Abaqus/CAE. The parameters used for evaluation were both the Stress maximum principal (S. Max) and the Stress minimum principal (S. Min). These eigenvalues, which are generated in foot tissues during normal simulated stance, are closely related to the tensile stresses and compressive forces, respectively.

Results

Compressive forces in midfoot joints

This first analysis aimed to evaluate the differences in compressive forces within midfoot joints. Minimum principal stress was calculated in scenarios of intact and dysfunctional spring ligament. Simulations were performed in the reference case (with2 625.9 Tme (Lsim0BT8(r)-a

Figure 5: Compressive forces (Minimum principal stress in MPa) generated in the union cartilage of the first and second cuneiform and in the Navicular bone.

Bar charts were used to demonstrate results including the highest compressive forces obtained in all simulation scenarios (Figure 6).

Figure 6: Comparison of the highest values of compressive forces generated in all simulated cases.

Biomechanical stress comparison traction forces

The spring ligament and the plantar fascia are the static stabilizers of the plantar arch [14,15]. The tibialis posterior tendon is the main dynamic stabilizer of the arch. We evaluated traction forces generated at the insertion area on the first cuneiform bone of the tibialis posterior tendon for both Lapidus types. The results are shown in Figure 7. The maximum principal stresses were measured. The color scale was normalized to 20 MPa. In this diagram, the blue scale represents the lowest stress values, whilst the red represents the highest. Neither of the Lapidus procedures evaluated had a significant effect on the traction forces at the insertion region of the tibialis posterior tendon. However, in the presence of spring ligament failure there was a significant increase in traction forces in the tibialis posterior despite the presence of additional stabilization of the first ray from both types of Lapidus procedure. The modified Lapidus arthrodesis generated a 79% increase

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tendon is highlighted.

Maximum stress values generated for each scenario have been presented as bar graphs (Figure 8).

Figure 8: Comparison of the highest values of traction forces generated in all the cases simulated.

Discussion

Finite element analysis provides a useful means of understanding the effects of Lapidus arthrodesis on the surrounding tissues and how they behave as a result of these procedures. Few experimental studies have measured the comparative biomechanical effects of the Lapidus arthrodesis because of the difficulty of objectively measuring these parameters in cadavers. It is well known that a stable first ray prevents secondary issues such as metatarsophalangeal joint impingement and

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