Tibio talar contact stress: An experimental and numerical study

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ARTICLE INFO

Keywords:

Stress on tibia
Tibio-talar contact
CAD
FE analysis

ABSTRACT

Aims: Tibio-talar contact stress has been evaluated and successively compared by performing an ankle contact finite element (FE) analysis and an experimental test carried on an assembled simple synthetic model of ankle equipped with a high-resolution (Tekscan) pressure sensor.

Methods: A numerical FEM analysis was carried out by simulating the ankle joint (foot, and tibia) in order to investigating the stress shielding on the contact surfaces. The foot was constrained at the base while a load of 980 N was applied on the top of the tibia. The same setup was experimentally reproduced by introducing a high-resolution (Tekscan) pressure sensor between tibia and foot.

Results: Results evidenced a good agreement between numerical and experimental data, a percentage difference of 15% was evaluated on the equivalent Von Mises contact stress.

Conclusion: The obtained results reveal interesting consequences deriving by taking into account how the stress shielding can influence the integrity and resistance of bones. The methods used for this validation enable formal comparison of computational and experimental results, and open the way for objective statistical measures of regional correlation between FE-computed contact stress distributions from comparison articular joint surfaces.

1. Introduction

The ankle joint is one of the three major joints of the lower limb; however, unlike the hip and knee joints, the ankle is rarely affected by primary osteoarthritis but is particularly sensitive to post traumatic osteoarthritis. Knowledge of the joint contact area, see Fig. 1, under physiological loads and throughout the range of motion is essential for understanding the biomechanics of the ankle joint; furthermore, it is beneficial for understanding the pathogenesis of joint degeneration as well as improving prosthetic design and ligament reconstruction surgery. Lesions occurring on the talus that are amenable to surgical treatment commonly occur over the anterior-medial or posterior-lateral talar shoulders; therefore, the joint contact characteristics in these regions under different loading conditions are of particular interest. There have been several reported experimental studies of ankle joint contact characteristics with wide variations in methodology and loading conditions resulting in varying reports of the extent and location of the contact area.^{1,2} The tibio-talar joint is rarely affected by primary osteoarthritis and the principal causes of cartilage degeneration are trauma and/or abnormal mechanics. Among others, the tibio-talar contact area (TTCA) has been widely investigated to monitor biomechanical changes due to articular incongruities or an altered loading. For an arthrodesis, the ankle is usually fused in a neutral sagittal position,^{3,4} i.e. the tibia is perpendicular to the long axis of the

foot. As a result, forward progression of the tibia at mid-stance can only be obtained by inducing early talar movement in the sagittal plane and dorsiflexion at the mid-tarsal joints. We hypothesized that the stress on the mid-tarsal joint during stance would be associated with an early forward displacement of the ground reaction force. Ankle arthrodesis is still the primary treatment for disabling arthritis of the ankle that does not respond to conservative treatment. This surgery may be followed by the development of degenerative osteoarthritis in the subtalar and mid-tarsal joints.^{5,6} The degenerative osteoarthritis in those joints is thought to be a consequence of the alteration of the foot dynamics after ankle arthrodesis. In normal gait, foot kinematics in stance are dependent on three rockers.⁷ At heel-off, the ground reaction force (GRF) is located close to the metatarsal heads. The contribution of the mid-tarsal joints to foot/tibia dorsiflexion in ankle arthrodesis has been reported to be either non-significant⁸ or significant,^{9,10} but lower than the contribution of the tibio-talar joint in normal gait. Thus, as the mid-tarsal joints do not completely compensate for the loss of ankle dorsiflexion, we hypothesized that early heel lift occurs during gait to allow further forward tilt of the tibia. At early heel-off, the GRF vector would be still posterior to the third rocker, exerting dorsiflexion stresses on the joints located between the ankle and the metatarsal heads. Integration of Geometric Morphometrics and FEA has yielded more sophisticated approaches. The talus is ideal for studying relationships between morphology and biomechanics as it transmits all forces encountered from

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<https://doi.org/10.1016/j.jor.2019.08.024>

Received 18 July 2019; Accepted 13 August 2019

Available online 14 August 2019

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Ankle Joint

Fig. 1. Ankle joint and its pathologies.

Fig. 2. Numerical and Experimental setup.

the foot to the leg^{11–16} and, unlike the diaphysis in long bones, it is predominantly stiffened by trabecular networks. This paper aims to investigate on the stress shielding occurring at the tibio talar contact surfaces. For this reason, a numerical FE model was developed and results compared with an experimental test carried on synthetic ankle joint.

2. Material and methods

A numerical model of the tibia and foot (28.854 elements and 8.754 nodes and 25.456 elements and 9.689 nodes, respectively) was obtained by matching nuclear magnetic resonance (MRI) for soft tissues, and a computerized tomography (CT) for bones, carried on a normal adult patient. Bony material properties were distinguished in cortical and trabecular bones, and were assumed to be linear elastic, isotropic

Fig. 3. Eq. Von Mises Stress [MPa] evaluated on the model.

Fig. 5. Eq. Elastic Strain [$\mu\text{mm}/\text{mm}$] evaluated on the model.

Fig. 4. Displacements [mm] evaluated on the model.

and homogeneous. A Young's modulus of 700 MPa was chosen for the trabecular bone, while for cortical bone a value of 17,000 MPa was selected, by imposing linear elastic, isotropic and homogeneous properties. Contact interfaces were imposed defining a penalty based method with a weight factor and a coefficient of friction of 0.4. The loading condition was simulated by imposing a vertical loads of 980 N and realizing a fixed rigid cube in contact with the foot,^{17,18} see Fig. 2. Non-linear finite element analyses were performed with Abaqus version 5.4 (Hibbitt, Karlsson and Sorensen, Inc., Pawtucket, RI) using the geometric nonlinearity and automatic time stepping options. The experimental setup was conceived in order to replace the numerical one as depicted in Fig. 2. A synthetic model of tibia and foot (model #1101 and #1131, Pacific Research Labs., Vashon Island, WA, USA) was subjected to a vertical load of 980 N by a material testing machine (Lloyd Instruments Inc., Fareham, UK), equipped with a 1000 N load cell ($\pm 0.5\%$ full scale accuracy). Between the two bony parts was inserted a Tekscan Model #5033 pressure sensor (Tekscan Inc.; Boston, MA). This sensor measures pressure at 1472 sensing elements (46,32 sensels), with a resolution of 144 sensels/ cm^2 .

3. Results

In Fig. 3 is depicted the contour map of Equivalent Von Mises stress evaluated on the FE model. As it is possible to see results evidenced a maximum value of about 23 MPa produced around the mid diaphyseal area. The foot appears in the picture subjected to a pressure of about 4 MPa. The maximum value of stress is reached internally on the contact zone with the tibia.

In Fig. 4 is reported the contour map of displacements which reached the value of 0,45 mm in the talus because of the effect of the force aging on the top of the tibia which reached the value of about 0,3 mm.

The equivalent elastic strain contour map, depicted in Fig. 5 follows

Table 1
Numerical results evaluated by FE analysis.

	Eq. Von Mises stress [MPa]	Displacements [mm]	Eq. Elastic strain [$\mu\text{mm}/\text{mm}$]
Complete model	22,95	0,45	0,0004
Contact surface in foot	7,15		
Experimental surface in foot	6,53

Fig. 6. Comparison between experimental a) and numerical b) results of the Eq. Von Mises Stress.

the same trend of the Eq. von mises stress depicted in Fig. 3, with a maximum value aging at about 4,3 mmm/mm. The numerical results are briefly reported in Table 1. The loading system can't really taking into account the interactions of the bone inside the leg with the other components (muscles, ligaments etc ...). The rigid constrain, obviously does not represent the real fixing configuration of the bone inside the bony chain. Finally in Fig. 6 is reported the comparison between the experimental and numerical results. As it can be evidenced results are in good agreements and the differences between the values register only a 15% of high stress for the numerical model. The maximum value reached by the experimental test id of about 6,5 MPa while the numerical one is about 7,15 MPa. The figure evidences almost the same zone interested by the contact highlighting the much more solicited areas.

4. Discussion

Knowledge about the stress distribution of the foot-ankle complex is a very important point for development of the prosthesis. Different studies investigated the stress strain distribution on long bones, femur or tibia, taking into account the presence of different kind of prosthesis.^{19–23} The problem in those cases is represented by the variability of results in a special way around the contact zone area. For this reason, a comparison between numerical and experimental results can be very useful.

The present direct comparisons between physical measurements and the FE model of ankle loading convincingly establish the

computational model's validity. This validation exercise not only showed reasonable comparison between the global metrics of contact stress and contact area, but also in the regional and spatial distributions of contact stress. Because OA usually initiates regionally, it seems especially important that contact FE models be validated regionally, rather than just in terms of global summary measures. Based upon these validation results, FE stresses could be reported over the whole articulating surface with a very reasonable degree of confidence (conservatively, within 10–15% of those measured experimentally).

Trauma to the ankle can disrupt the normal mechanics of the joint and result in overloading of the cartilage and subsequent degeneration. Lateral shift of the talus within the ankle mortise has been shown to reduce the contact area within the joint by up to 42%,^{24,25} increasing peak pressures in the articular cartilage by 50%.²⁶ These changes were measured using static models of loaded ankle joints in which the talus was fixed in a lateral position, and thus did not permit movement within the mortise. It has been suggested that during walking the talus may seek its own position beneath the tibia on loading, tending to reduce the forced lateral subluxation.²⁷ Excision of the lateral malleolar articular surfaces did not reduce stability, but division of the deltoid ligament caused a twofold increase in rotational instability. They concluded that a relatively small change in collateral ligament positioning may cause deranged ankle motion, even in the absence of a visible talar shift on static radiographs of the unloaded ankle. Anatomic studies of the medial and lateral ligaments have indicated that the origins of these ligamentous structures suggest that even small variations can result in abnormal ankle mechanics. The instability pattern consisted of external

rotation of the talus from beneath the tibial plafond. The extent of rotation was limited by an intact deltoid ligament that was under tension in the loaded and plantar flexed ankle. Section of the ligament allowed greater external rotation, resulting in significantly greater reductions in contact area with each fibular deformity. Rotational instability has been previously noted in studies of ankle ligaments and has been likened to the anterolateral rotatory instability seen in the anterior cruciate deficient knee. There are several plausible explanations for the modest focal discrepancies that were observed between measured and computed contact stress distributions at the periphery of contact. Inserting the Tekscan sensor into the articular joint could potentially disturb the natural contact pattern. Imperfect segmentation of the bones and/or inadequate replication of non-uniform cartilage thickness would influence the computed contact stresses. A final possible explanation would be spatial registration errors between the Tekscan and FE data. With the articulating surface of the tibio-talar joint being nearly cylindrical in nature, there was no evidence of sensor crinkling in the experimental testing. Since the sensor was very thin and easily conformed to the joint, it was deemed unlikely to interfere substantially with normal joint loading. The sensor was, therefore, not explicitly included in the FE model. Another limitation was that the articular surface geometry was based on CT source data, where cartilage cannot easily be visualized. For this reason, a constant uniform thickness cartilage layer was extruded from the bony subchondral surfaces along local surface normals. Given the agreement observed between computed and measured contact stress values, this simplification seems reasonable.

5. Conclusions

The computational investigation of altered contact mechanics associated with clinically observed residual incongruity following treatment and healing of intraarticular fractures depends upon faithful regional reproduction of the prevailing contact stress distributions. Simply establishing general agreement between computed and measured mean and maximum contact stresses and contact areas would not have been sufficient in this regard. In other studies similarly reliant upon accurate local prediction of contact stress, it will likewise be critically important that these “source” mechanical data be validated in an absolute and rigorous sense.

References

1. Calhoun JH, Li F, Ledbetter BR, Viegas SF. A comprehensive study of pressure

- distribution in the ankle joint with inversion and eversion. *Foot Ankle Int.* 1994;15:125–133.
2. Driscoll HL, Christensen JC, Tencer AF. Contact characteristics of the ankle joint. Part 1. The normal joint. *J Am Podiatr Med Assoc.* 1994;84:491–498.
3. Mazur JM, Schwartz E, SimonAnkle arthrodesis SR. Long-term follow-up with gait analysis. *J Bone Joint Surg Am Vol.* 1979;61:964–975.
4. Katcherian DA. Treatment of ankle arthrosis. *Clin Orthop.* 1998;349:48–57.
5. Broquin J, Emami A, Maurer P, Tomeno B. Arthrodesis of the ankle A study of the complications and long term results. *Rev Chir Orthopédique.* 1979;65:393–401.
6. Jackson A, Glasgow M. Tarsal hypermobility after ankle fusion—fact or fiction? *J Bone Joint Surg Br.* 1979;61-B:470–473.
7. Perry J. Kinesiology of lower extremity bracing. *Clin Orthop.* 1974;1:18–31.
8. King HA, Watkins Jr TB, Samuelson KM. *Analysis of Foot Position in Ankle Arthrodesis and its Influence on Gait Foot Ankle1.* 1980; 1980:44–49.
9. Wu WL, Su FC, Cheng YM, Huang PJ, Chou YL, Chou CK. *Gait Analysis after Ankle Arthrodesis Gait Posture.* vol. 11. 2000; 2000:54–61.
10. Hunt AE, Smith RM, Torode M, Keenan A-M. Inter-segment foot motion and ground reaction forces over the stance phase of walking. *Clin Biomech Bristol Avon.* 2001;16:592–600.
11. Filardi V. FE analysis of stress and displacements occurring in the bony chain of leg. *J Orthop.* 2014;11(4):157–165.
12. Filardi V, Milardi D. Experimental strain analysis on the entire bony leg compared with FE analysis. *J Orthop.* 2017;14(1):115–122.
13. Filardi V. Stress shielding in the bony chain of leg in presence of varus or valgus knee. *J Orthop.* 2015;12(2):102–110.
14. Filardi V. Flatfoot and normal foot a comparative analysis of the stress shielding. *J Orthop.* 2018;15(3):820–825.
15. Filardi V. Finite element analysis of the foot: stress and displacement shielding. *J Orthop.* 2018;15(4):974–979.
16. Filardi V, Simona P, Cacciola G, ... Milardi D, Alessia B. Finite element analysis of sagittal balance in different morphotype: forces and resulting strain in pelvis and spine. *J Orthop.* 2017;14(2):268–275.
17. Filardi V, Montanini R. Measurement of local strains induced into the femur by trochanteric Gamma nail implants with one or two distal screws. *Med Eng Phys.* 2007;29(1):38–47.
18. Montanini R, Filardi V. In vitro biomechanical evaluation of antegrade femoral nailing at early and late postoperative stages. *Med Eng Phys.* 2010;32(8):889–897.
19. Filardi V. The healing stages of an intramedullary implanted tibia: a stress strain comparative analysis of the calcification process. *J Orthop.* 2015;12:51–61.
20. Filardi V. Characterization of an innovative intramedullary nail for diaphyseal fractures of long bones. *Med Eng Phys.* 2017;49(1):94–102.
21. Filardi V. Healing of femoral fractures by the meaning of an innovative intramedullary nail. *J Orthop.* 2018;15(1):73–77.
22. Filardi V. Numerical comparison of two different tibial nails: expert tibial nail and innovative nail. *Int J Interact Des Manuf.* 2018;12(4):1435–1445.
23. Filardi V. Healing of tibial comminuted fractures by the meaning of an innovative intramedullary nail. *J Orthop.* 2019;16(2):145–150.
24. Kimizuka M, Kurosawa H, Fukubayashi T. Load-bearing pattern of the ankle joint. Contact area and pressure distribution. *Arch Orthop Trauma Surg.* 1980;96(1):45–49.
25. Ramsey PL, Hamilton W. Changes in tibiotalar area of contact caused by lateral talar shift. *J Bone Jt Surg.* 1976;58(3):356–357.
26. Zindrick MR, Hopkins GE, Knight GW, Hopkins CL, Patwardhan AG, Bunch WH. The effect of lateral talar shift upon the biomechanics of the ankle joint. *Orthop Trans.* 1985;9:332–333.
27. Vrahas M, Beenis B, Rudert ML, Fu FH. Intra-articular contact stresses with simulated ankle malunions. *Trans Orthop Res Soc.* 1990;36:502.