Journal of Advanced Sport Technology 1(1):44-51

Original Research

Finite Element Analysis of Anterior Talofibular Ligament under Different Strategies of Landing

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ABSTRACT

Ankle injury research is typically performed with cadavers or by the clinical measuring technique which both of them have several problems such as general degradation of tissues. Numerical methods help us to overcome these problems. Therefore, the goal of this study was to compute the stress in Anterior Talofibular (ATF) ligament under different strategies of landing using finite element modeling for males and females. These strategies included the variation of the ankle flexion angle and hindfoot angle at the instant of initial contact with the ground. For calculating stress in ATF ligament, ankle torques were inputted to finite element model. Ankle torques were obtained from a five-link dynamic model using inverse dynamic approach. Input data for dynamic model included motion kinematics during drop landing from 60 cm height during 0.198 seconds after initial contact of the foot with the ground. Finite element model was created by manipulating of CT and MR images using the appropriate software. Results showed that stress in ATF ligament increase when torques in the ankle joint increase. Results from finite element model showed that when initial ankle with the ground was 68.7 degrees peak ATF ligament stress for females was lower than when initial ankle with the ground was 88.7 degrees. Additionally, increased initial plantar flexion of ankle leads to increasing stress up to 5.27 times in ATF ligament. Comparing this model and previous investigations' results showed that this method is a useful tool to simulate landing condition for any person and to help us to predict whether those conditions probably lead to injuries or not.

Keywords: Landing Biomechanics, Anterior Talofibular Ligament, Finite Element Analysis, Dynamic Modeling

Introduction

Ideally, ankle joint testing would be performed on human volunteers, but volunteer studies are limited to non-injurious loads and flexion angles for obvious reasons. Ankle injury research is typically performed with cadavers. Cadaveric specimens may be compromised by the lack of blood pressure, muscle tension and general degradation of tissues [46]. Also, clinical measuring technique is invasive and might disrupt the integrity of the tissue, and the consistency of the results was greatly influenced by position and calibration of these devices. Furthermore, these techniques can only measure the strain in a particular region, but not the distribution and the direction of the stress. In most of modeling investigations, neutral position of standing was studied, whereas particularly sports involving running and jumping are known for the high incidence of inversion traumata [47]. Previous researchers studied biomechanics of lower extremity using experimental approach.

Lateral sprain injuries, particularly ATF ligament sprains are common in sport injuries [48]. Ankle sprain occurs more frequently in landing, when athlete's foot touches the ground [49]. A typical mechanism that contributes to increase ankle sprain is excessive plantar flexion at touch-down [49, 50]. This inappropriate foot positioning prior to touch-down is a potential cause of the increased sprain. It has been shown that stress and strain in ATF ligament increase as the ankle is moved progressively through plantar flexion [51, 52]. Previous studies on ankle torque during landing [53, 54] showed that more plantar flexion of the ankle at touchdown led to higher ankle torques [54].

Studying on ATF ligament during different strategies of landing haven't been studied yet. Hence, the aim of this study was to determine ATF ligament stress under different strategies of landing.

Material and Methods

The geometry of the finite element model was obtained from 3D reconstruction of CT and MR images from the right foot of a normal female subject of age 24.6, height 170 centimeters and mass 60 kilograms. Axial CT images with intervals of 0.7 mm for estimating bony structures of foot and coronal MR images were taken with intervals of 0.2 mm for estimating ATF ligament structure. Both CT and MR images were taken in the neutral unloaded position. The images were segmented using Mimics® (version 10/01, Materialise Software) to obtain the solid models for each bone and ATF ligament. This software also used for accessing to point cloud of each bone and ATF ligament. Point clouds imported to Geomagic Studio® (version 7, Raindrop Geomagic) for smoothing, reducing noise, creating shell, and making patch for all point clouds. Then files saved in a usable format which can be opened in Abaqus software. Stp format of structures then imported and assembled in CATIA® (v5R19). Assembled model of ankle and foot complex then imported to Abaqus® (version 6/8-1) software for estimating ATF ligament stress under different strategies of landing. The finite element model, as shown in Figure1 consisted of 20 bony segments, including the distal segments of the tibia and fibula and 18 foot bones: talus, calcaneus, cuboid, navicular, 3 cuneiforms, 5 metatarsals and 6 components of the phalanges. The second to fifth metatarso phalanges were fused together and assumed to be one part. In this model we consider bones as rigid part and ligaments as deformable parts [55]. All tissues were idealized as homogeneous, isotropic [55]. The Young's modulus and Poisson's ratio for different structures were selected from the literature (Table 1) [55, 56].

Table 1. Material properties and element type of finite element model.

Component	Element type	Young's modulus E(MPA)	Poisson's ratio	Density(Kg/m^3)
ATF ligament	Solid – Homogeneous	260.0	0.490	1200.0
Other ligaments	Truss	260.0	-	1200.0

The interaction among the head of ATF ligament and talus and fibula were defined as surface-to-surface contact with the friction coefficient of 0.9. Tie contact was used in both head of ATF ligament with talus and fibula. Except the ATF ligament a total number of 77 ligaments consider as tension-only trusses [55] which defined by connecting the corresponding attachment points on the bones (Gray's Anatomy). The bony structures were meshed with 4-node 3D bilinear rigid quadrilateral elements, ATF ligament was meshed with 4-node linear tetrahedron elements and other ligaments were meshed with 2-node linear 3D truss. We also consider inertia for each bone. Boundary conditions were defined as talus could only moves around medio-lateral axis and fibula was bounded in all six degree of freedom. Explicit dynamic step used for solving.

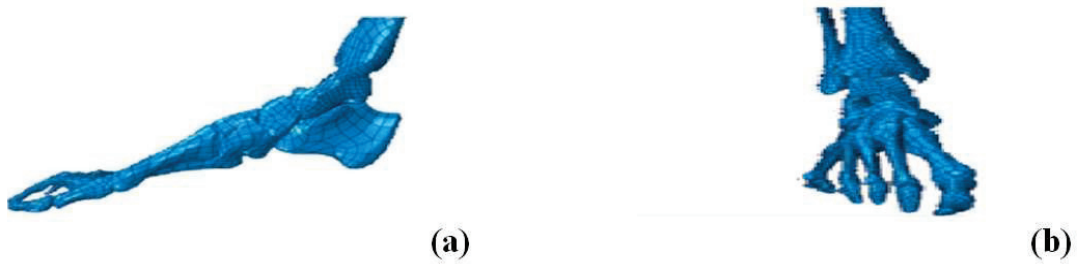
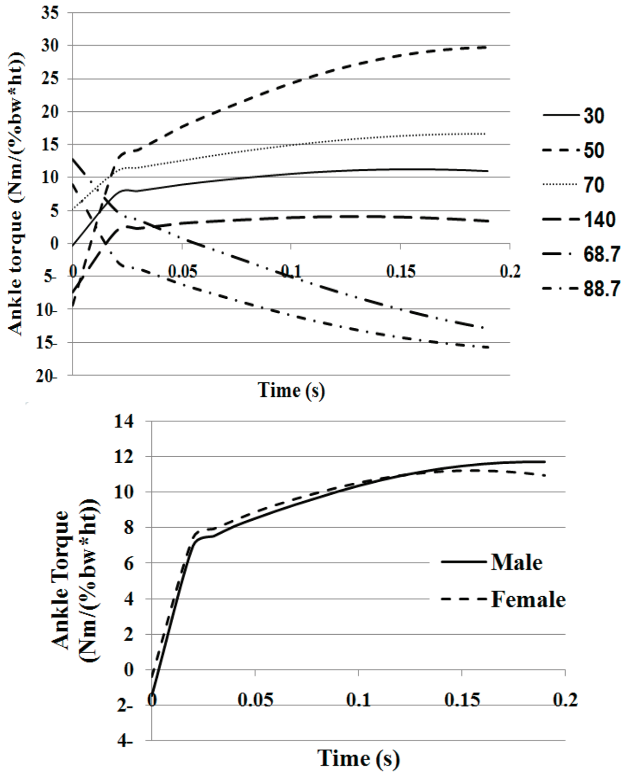


Figure 1. Finite element model of ankle and foot complex (a) Medial view; (b) Anterior view.

Ankle torques during different strategies of landing achieved from Mirtavoosi et al. study [54]. These strategies included the variation of the ankle flexion angle and hindfoot angle at the instant of initial contact with ground. Using anthropometric data, five-link dynamic models were built to analyze the ankle torque for male and female. Input data included motion kinematics during drop landing from 60 cm height during 0.198 seconds after initial contact of foot with the ground which was obtained from Decker et al. study [53]. During this time maximum flexion of knee occurred. These models were built for females with mean age of 24.6 years and mean height of 1.7 meters and mean mass of 60.1 kilograms and for males with mean age of 28.3 years and mean height of 1.8 meters and mean mass of 81.8 kilograms. We used two different dynamic models for males and females and one finite element model to estimate ATF ligament stress in males and females, because we consider that, their skeletal geometry is not different or slightly different to each other. We believe that because we analyze the stress in ATF ligament not in bones, and the bones were considered as rigid parts, this probable slight difference has no effect on stress in ATF ligament. Besides, input data of finite element model was different ankle torques which was obtained from different dynamic model. As mentioned before, input data consist of the kinematics of motion for three different strategies. First strategy was defined as landing on plantar surface of foot, while initial ankle position at contact was changed. Second strategy was defined as toe-landing in which the initial position of ankle was supposed to be unchanged, while the angle between hindfoot and ground was changed. Third strategy was defined as toe-landing in which the initial position of hindfoot was supposed to be unchanged, while the angle of ankle with ground was changed. Here we supposed the angle of hindfoot and ground 30º for females and males. Using inverse dynamic approach, ankle torque was obtained in all mentioned strategies. Normalized ankle torque from two conditions of each strategy (Figure 2a) and similar condition to Decker et al. study (Figure 2b), input to finite element model to achieve stress distribution in ATF ligament. Normalized peak ankle torques in all strategies are shown in Figure 3.



(a) (b)

Figure 2. Normalized ankle torque obtained from dynamic model that was input to the finite element model (a) Different condition of landing for females (70, 140, 68.7, 88.7 are the position of ankle joint with ground (degree) and 30, 50 are the position of hindfoot with ground (degree)); (b) Ankle torques which were obtained from dynamic model [54] in similar condition to Decker et al. study.

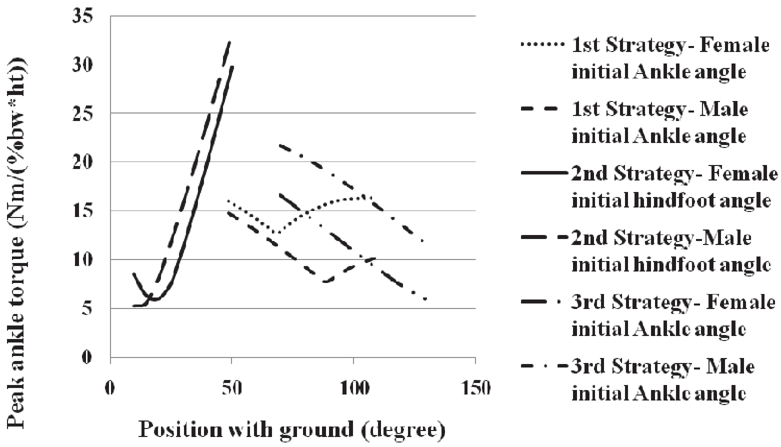


Figure 3. Normalized peak ankle torque in first, second and third strategy for males and females.

Results

Finite element model validation

Torques which were obtained from dynamic model [54] in similar condition to Decker's study [53] (Figure 2b) were given to finite element model. Peak ATF ligament stress for male was 30.54 MPa and for female was 30.01 MPa (Figure 4). These results showed that injury probability in males is more than females. This result was in agreement with Decker et al. experimental study [53].

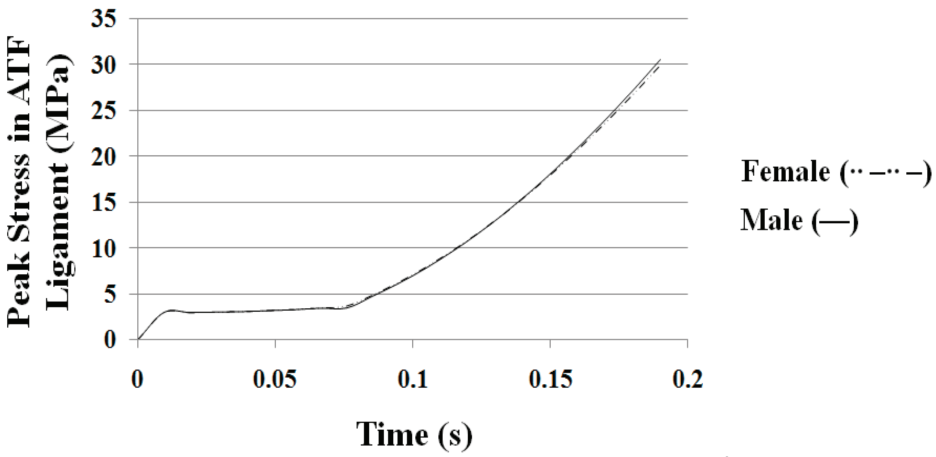


Figure 4. Peak stress in ATF ligament in similar condition to Decker et al.'s study for male and female [53].

Finite element model results

Results from finite element model showed that when initial ankle with ground was 68.7 degree peak ATF ligament stress for females was lower than when initial ankle with ground was 88.7 degree. Also for males we have the same results. In second strategy, in both males and females, when the angle of hindfoot with ground was 50 degree, peak ATF ligament stress was higher than when the angle of hindfoot with ground was 30 degree. In third strategy, in both males and females, when the angle of hindfoot with ground was 70 degree, peak ATF ligament stress was higher than when the angle of hindfoot with ground was 140 degree. Results are shown in Figure 5 and Table 2.

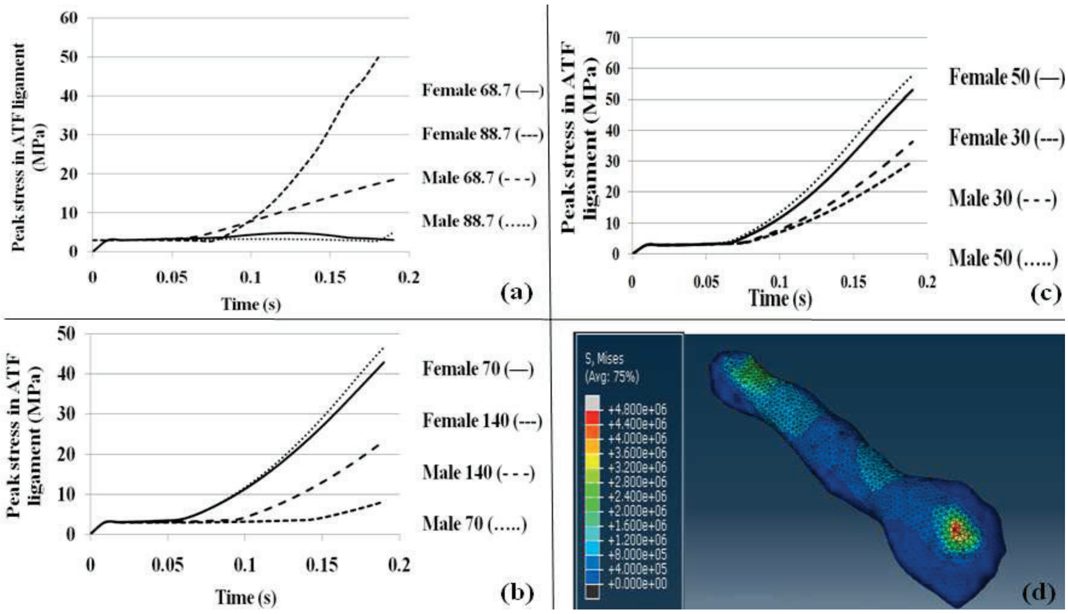


Figure 5. Diagram of peak stress in ATF ligament under 2 different conditions for males and females in : (a) First strategy (b) Second strategy (c) Third strategy; (d) Stress distribution pattern in ATF ligament in females, initial angle of ankle with ground = 68.7.

Table 2. Peak stress in ATF ligament in different conditions of landing for males and females.

Strategies	Peak stress in ATFL in females (MPa)	Peak stress in ATFL in males (MPa)
First strategy- Initial ankle angle with ground = 68.7	4.8	18.47
First strategy- Initial ankle angle with ground = 88.7	49.93	4.976
Second strategy- Initial hindfoot angle with ground = 50	53.03	57.93
Second strategy- Initial hindfoot angle with ground = 30	30.01	36.32
Third strategy- Initial ankle angle with ground = 70	42.87	46.61
Third strategy- Initial ankle angle with ground = 140	8.134	23.21

Discussion

Peak ATF ligament stress for male was 30.54 MPa and for female was 30.01 MPa due to different peak ankle torque input to finite element model in similar condition to Decker et al.'s study. This result showed that injury probability in males is more than females which was in agreement with previous studies [53, 57]. In this study peak ATF ligament stresses in males and females were close to each other. It might be because of peak ankle torques for males and females were close to each other. Peak ankle torque in females was 4% lower than males [54].

In first strategy dynamic model showed that when initial ankle with ground was 68.7 degree in females, peak ankle torque was minimum [54]. We have lower peak stress when ankle had lower peak torque and vice versa. In second strategy it has been shown that when the angle of hindfoot with ground were 50 and 30 degree, peak stress in ATF ligament in males was higher than females. Results showed that when the ankle is more plantarflexed, peak ankle torque increased [54] and peak stress in ATF ligament increased too. In third strategy it has been shown that when the angle of ankle with ground were 140 and 70 degree, peak stress in ATF ligament in males was higher than females. Results showed that when the ankle is more plantarflexed, peak ankle torque increased [54] and peak stress in ATF ligament increased too. Other studies showed that inversion sprains often occur when the foot was plantarflexed [48, 58]. Therefore, our model confirmed that susceptibility to sprains is increased by initial plantar flexion. Also biomechanics of ankle joint showed that because of alignment of joint's bones, plantarflexion caused ankle instability. In this situation surfaces of talus which joint to malleolus, are wider in front. So in dorsiflexion, talus was fixed tightly. But in plantarflexion, talus can move a little laterally [59]. In other words probability of injury increased by plantarflexion. In plantarflexion ATF ligament is parallel to the long axis of the foot and assumes a more parallel alignment with the long axis of the fibula, thus making it more susceptible to injury in this position [6, 15, 16].

Conclusion

By comparison of this model and previous investigations, it has been shown that this method is a useful tool to simulate any condition for each person and to identify whether those conditions probably lead to injuries or not.

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