Incorporating Fibular Kinematics into Musculoskeletal Computer Models to Evaluate Ankle Syndesmosis Injury Mechanisms

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Introduction: High ankle sprains are a common injury that disrupts the ankle syndesmosis (i.e., distal tibiofibular joint) and often require surgical repair. During surgery, the tibia and fibula are realigned and fixed. Both fixation and inadvertent misalignment, which occurs in as many as 52% of patients [1], alter motion of the fibula. The characteristics and consequences of this altered motion are not well understood. Our goal is to use musculoskeletal models to characterize healthy and pathologic fibular motion. This will improve our understanding of the mechanisms of syndesmosis injury and inform injury repair. As a critical step toward this goal, this study aims to incorporate fibular motion into musculoskeletal models and validate those models using gait and ankle sprain data.

Materials and Methods: Fibular motion was added to two OpenSim models by incorporating a six degree-of-freedom (DOF) joint between the fibula and tibia [2, 3]. The models were adapted from the Gait2392 model (Model A), which includes the lower limbs and torso [2], and a full-body model (Model B), which also includes arms [3]. To examine gait, muscle-driven forward dynamics simulations were run by applying data from an experimental gait study [3-6]. To examine high ankle sprains, forward dynamic simulations were designed to replicate a cadaveric experiment by scaling the models to represent a 60 kg person, removing muscles to simulate the described muscle dissection, and applying a 600 N axial load to the tibia [7]. Rotations and translations of the fibula were examined for both simulations. Specifically, the magnitude and direction of fibular translation and rotation were compared to the values reported in the literature. Although the simulations describe 6 DOF fibular motion, reported results focus on motion directions for which direct comparisons to the literature could be made. Dorsiflexion (DF) and plantarflexion (PF) movements of the right and left limbs were examined separately for both simulations.

Results and Discussion: Both simulations successfully incorporated 6 DOF fibular motion. However, the fibula did not move as expected (Table 1). During the gait simulations, the fibula was expected to rotate internally and translate anteriorly and medially during PF, and to rotate externally and translate anteriorly and laterally during DF. However, in Model A, there was unexpected medial translation and internal rotation of the right fibula during DF. There was also unexpected external rotation and posterior translation of both fibulas during PF. Similarly, in Model B, unanticipated internal rotation and medial translation was observed for both fibulas and posterior translation of the left fibula during DF. In Model B, the direction of the fibula opposed the expected motion during PF. Importantly, rotation observed in the models was much smaller than expected, and AP translation in the models was generally larger than expected. Furthermore, the movement of the fibula would oscillate for each movement type in both legs of each model (e.g. AP translation for both fibulas in Model A were separated by a phase difference of ~0.7s). The differences observed between models could be attributed to the added constraints to Model B, where muscle contractions were not involved and the subtalar joints were locked. Differences from the experiment could result from inaccurate inertial properties of the modeled fibula and/or inclusion of the mediolateral center of mass displacement of the subject during walking, which was not present in the cadaver study.

Table 1. Mean (± st. dev) Fibular Motion During Gait Simulations

Model	Ankle Movement	Mean Fibular Displacement					
		Right Fibula			Left Fibula		
		ML Rotation[a] (°)	AP Translation[b] (mm)	ML Translation[c] (mm)	ML Rotation[d] (°)	AP Translation[b] (mm)	ML Translation[e] (mm)
Model A	Dorsiflexion	0.0622 ± 0.12	2.86 ± 0.64	-0.799 ± 1.2	-0.0183 ± 0.025	1.862 ± 0.21	-0.273 ± 0.18
	Plantar Flexion	-0.103 ± 0.069	-2.72 ± 0.41	-0.362 ± 0.39	-0.0693 ± 0.077	-3.16 ± 0.67	-1.26 ± 0.83
Model B	Dorsiflexion	0.0218 ± 0.0027	0.532 ± 0.015	-1.94 ± 0.26	0.0187 ± 0.0057	-0.0818 ± 0.55	-1.54 ± 0.26
	Plantar Flexion	-0.00352 ± 0.002	-0.398 ± 0.26	0.258 ± 0.20	-0.00233 ± 0.00072	-0.116 ± 0.017	0.282 ± 0.052
Cadaver Experiment	Dorsiflexion	-0.71 ± 0.45	-0.28 ± 0.21	0.67 ± 0.33	-0.71 ± 0.45	-0.28 ± 0.21	0.67 ± 0.33
	Plantar Flexion	0.43 ± 0.37	0.17 ± 0.31	-0.17 ± 0.16	0.43 ± 0.37	0.17 ± 0.31	-0.17 ± 0.16

Positive values correspond to [a]internal rotation, [b]anterior displacement, [c]lateral displacement.

Conclusion: Fibular motion was successfully achieved using computational models, but the 6 DOF joint did not fully replicate the experimental data. In the near future, the inertial properties of the fibula will be recalculated, and the high ankle sprain injury mechanism will be further explored. Future work will also examine the minimal clinically important change in fibular motion. By elucidating fibular movement in response to loading patterns, our simulations can inform design of fixation mechanisms that permit healthy movement at the distal tibiofibular joint as well as surgical methods that reduce the likelihood of inadvertent misalignment.

References: [1] Cosgrove CT et al. *J Orthop Trauma*. 2017;31(8):440-446. [2] Delp et al. *IEEE Trans Biomed Eng*. 2007 54(11):1940-1950. [3] Rajagopal, Apoorva, et al. *IEEE Trans Biom Eng*. 2016 63.10: 2068-2079. [4] John C. T., et al. *J Biomech.* 2012 45(14): 2438-2443. [5] Lindsjo U. *Clin Orthop Relat Res*. 1985 (199): 28-38. [6] Magan A.G. et al. *Br Med Bull.* 2014 111(1): 101-15. [7] Chen, D. et al. *J. Ortho Surg.* 2019 27(2):2309499019842879.