This item was submitted to [Loughborough's Research Repository](#) by the author.
Items in Figshare are protected by copyright, with all rights reserved, unless otherwise indicated.

Estimation of ligament strains and joint moments in the ankle during a supination sprain injury

PLEASE CITE THE PUBLISHED VERSION

<http://dx.doi.org/10.1080/10255842.2013.792809>

PUBLISHER

© Taylor & Francis

VERSION

AM (Accepted Manuscript)

PUBLISHER STATEMENT

This work is made available according to the conditions of the Creative Commons Attribution-NonCommercial-NoDerivatives 4.0 International (CC BY-NC-ND 4.0) licence. Full details of this licence are available at:
<https://creativecommons.org/licenses/by-nc-nd/4.0/>

LICENCE

CC BY-NC-ND 4.0

REPOSITORY RECORD

Wei, Feng, Daniel Tik-Pui Fong, Kai-Ming Chan, and Roger C. Haut. 2019. "Estimation of Ligament Strains and Joint Moments in the Ankle During a Supination Sprain Injury". figshare.
<https://hdl.handle.net/2134/21285>.

Estimation of ligament strains and joint moments in the ankle during a supination sprain injury

Journal:	<i>Computer Methods in Biomechanics and Biomedical Engineering</i>
Manuscript ID:	GCMB-2012-0287.R3
Manuscript Type:	Original Article
Date Submitted by the Author:	22-Jan-2013
Complete List of Authors:	Wei, Feng; Rehabilitation Institute of Chicago, Fong, Daniel; The Chinese University of Hong Kong, Chan, Kai-Ming; The Chinese University of Hong Kong, Haut, Roger; Michigan State University,
Keywords:	Anterior talofibular ligament, Ankle inversion sprain, Injury mechanism, Ankle biomechanics, Computational model, Dynamic simulation

SCHOLARONE™
Manuscripts

RESPONSES TO REVIEWERS' COMMENTS:

Manuscript ID GCMB-2012-0287.R2 (Computer Methods in Biomechanics and Biomedical Engineering): **Estimation of ligament strains and joint moments in the ankle during a supination sprain injury**

We have revised the manuscript to address the reviewer's last concern. In the revised manuscript we first accepted all the changes made to address the reviewers' previous comments. We then tracked the new changes that directly respond to the reviewer's last critical comment. We want to thank the reviewer for the insightful suggestions that have made this work better.

Reviewer(s)' Comments to Author:

Reviewer: 1

Comments to the Corresponding Author

Effect of variation on results of the value of the damping coefficient values must be added. Justification for not doing so, based on the existing study in the literature (lines 207-210) is not sufficient since that study looked at end range of motion where time effects (and thus damping) were not in play. This is not valid in your study.

We have varied the damping coefficient up and down one order of magnitude (similar to the study by Majors and Wayne, Ann Biomed Eng 39, 2807-2815, 2011) and found that joint moment calculated in the model changed by 0.45 Nm, approximately 2% of the maximum moment (23 Nm) generated in the simulation. In addition, we have also varied the stiffness and the exponent up and down one order of magnitude individually, just like Majors and Wayne (2011) in their study. The results showed that the joint moment was affected by approximately 1.5% and 4.3%, respectively. All these analyses, however, provided no change in the strain results calculated in the model. We therefore believe that ligament strains, generated in linear elastic spring elements, were primarily affected by the prescribed bone motions in the current study, not by the contact parameters. We have included these analyses and results in Discussion of the revised manuscript on page 9, lines 205-214. Again, we would like to thank the reviewer for the suggestions that have made this study more informative.

Estimation of ligament strains and joint moments in the ankle during a supination sprain injury

Feng Wei^{1,2}, Daniel Tik-Pui Fong^{3,4}, Kai-Ming Chan^{3,4}, Roger C. Haut⁵

¹Sensory Motor Performance Program
Rehabilitation Institute of Chicago
Chicago, IL 60611

²Department of Physical Medicine and Rehabilitation
Northwestern University Feinberg School of Medicine
Chicago, IL 60611

³Department of Orthopaedics and Traumatology
Prince of Wales Hospital
Faculty of Medicine, The Chinese University of Hong Kong
Hong Kong, China

⁴The Hong Kong Jockey Club Sports Medicine and Health Sciences Center
Faculty of Medicine, The Chinese University of Hong Kong
Hong Kong, China

⁵Orthopaedic Biomechanics Laboratories
Michigan State University
East Lansing, MI 48824

Corresponding Author:
Prof Daniel Tik-Pui Fong
Department of Orthopaedics and Traumatology
Prince of Wales Hospital
Faculty of Medicine, The Chinese University of Hong Kong
Hong Kong, China
Tel: (852) 2632 3535, Email: dfong@ort.cuhk.edu.hk

Word Count: 234~~20~~

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20
21
22
23
24
25
26
27
28
29
30
31
32
33
34
35
36
37
38
39
40
41
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

ABSTRACT

This study presents the ankle ligament strains and ankle joint moments during an accidental injury event diagnosed as a grade I anterior talofibular ligament (ATaFL) sprain. A male athlete accidentally sprained his ankle while performing a cutting motion in a laboratory setting. The kinematics data were input to a three-dimensional rigid-body foot model for simulation analyses. Maximum strains in 20 ligaments were evaluated in simulations that investigated various combinations of the reported ankle joint motions. Temporal strains in the ATaFL and the calcaneofibular ligament (CaFL) were then compared and the three-dimensional ankle joint moments were evaluated from the model. The ATaFL and CaFL were highly strained when the inversion motion was simulated (10% for ATaFL and 12% for CaFL). These ligament strains were increased significantly when either or both plantarflexion and internal rotation motions were added in a temporal fashion (up to 20% for ATaFL and 16% for CaFL). Interestingly, at the time strain peaked in the ATaFL, the plantarflexion angle was not large but apparently important. This computational simulation study suggested that an inversion moment of approximately 23 Nm plus an internal rotation moment of approximately 11 Nm and a small plantarflexion moment may have generated a strain of 15-20% in the ATaFL to produce a grade I ligament injury in the athlete's ankle. This injury simulation study exhibited the potentially important roles of plantarflexion and internal rotation, when combined with a large inversion motion, to produce a grade I ATaFL injury in the ankle of this athlete.

KEYWORDS: Anterior talofibular ligament, Ankle inversion sprain, Injury mechanism, Ankle biomechanics, Computational model, Dynamic simulation.

INTRODUCTION

Ankle sprain is a common sports trauma (Fong et al., 2007). Over 80% of these injuries are caused by excessive ankle joint supination which ruptures lateral ankle ligaments. Studies have been conducted to describe the kinematics of accidental ankle sprains using motion or video analysis (Kristianslund et al., 2011; Mok et al., 2011). These studies, however, do not provide data on ankle ligament strains. Additionally, it has been suggested that deviation of the ground reaction force away from the ankle joint center may result in an explosive ankle joint moment causing injury (Fong et al., 2009a). Such a moment would trigger a twisting motion to stretch ligaments vigorously (Delahunt et al., 2006).

An investigation by Fong et al. (2009b) of a grade I supination injury to the anterior talofibular ligament (ATaFL) has revealed that, in addition to plantarflexion and rearfoot inversion, internal rotation of the foot was noted to be greater than previously reported (Bahr et al., 1998). However, the roles of inversion, plantarflexion, and internal rotation, or their temporal combination on lateral ligament strains are still unclear. Recently, Wei et al. (2011b) has developed a computational model and utilized motion analysis-based kinematic data from laboratory tests to drive the model for estimations of dynamic ankle ligament strains and joint moments (Wei et al., 2011a). The model has shown its ability to help in the study of the motions that may predispose ankle ligaments to injury during external foot rotation. The purpose of the current study was to use this model to study the motions producing ankle ligament strains and joint moments during the accidental sprain event documented by Fong et al. (2009b). Since the injury was diagnosed as a grade I ATaFL sprain, and a combination of inversion, plantarflexion,

1
2
3 85 and internal rotation has been documented, a high strain in this ligament and high joint moments
4
5 86 in these three directions were expected.
6
7
8 87
9

10 88 **METHODS**

11
12
13 89 **Injury case**

14
15 90 A male athlete (23 years, 1.75 m, 62.6 kg) wore a pair of high-top basketball shoes and
16
17 91 performed a series of cutting motion trials in the laboratory. In one trial the athlete accidentally
18
19 92 sprained his right ankle, and the injury was immediately diagnosed as a grade I sprain of the
20
21 93 ATaFL by a well-trained orthopaedic specialist (K.M.C.). The injury occurred in the laboratory
22
23 94 that utilized marker-based motion and model-based image-matching video analysis systems
24
25 95 (Fong et al., 2009b).
26
27
28

29 96 **Computational modeling**

30
31 97 A 3D multi-body dynamic foot model (Wei et al., 2011b) was utilized for simulations of the
32
33 98 injury event. The model comprised five segments, namely the tibia, fibula, talus, calcaneus, and
34
35 99 tarsal and metatarsal bones. Details of model development and validation have been previously
36
37 100 described (Wei et al., 2011b). The model was constructed from a generic cadaver ankle (male, 19
38
39 101 years, 1.88 m, 86 kg) which was scanned using computed tomography (CT) in a neutral position
40
41 102 and with a separation of 0.6 mm between slices. CT images were converted into 3D models in
42
43 103 MIMICS (Materialise, Ann Arbor, MI) and imported into dynamic rigid-body motion simulation
44
45 104 software (SolidWorks, TriMech Solutions, LLC, Columbia, MD) (Iaquinto and Wayne, 2010;
46
47 105 Liacouras and Wayne, 2007). The neutral position of the ankle from the CT scan was maintained
48
49 106 in the model. The model includes 20 ligaments formulated as linear elastic, tension only springs
50
51 107 (Figure 1) with their stiffnesses (N/mm) adapted from the literature (Table 1). The slack length
52
53
54
55
56
57
58
59
60

of the ligaments was defined with the model positioned in neutral. An initial strain of 2%, implemented by inserting a spring element of length 2% shorter than the initial length (distance between insertion points), was assigned to each ankle ligament (Wei et al., 2011a), thereby inducing an *in situ* preload in the ligament. An initial strain of 0.5%, however, was applied to the interosseous ligaments (Liacouras and Wayne, 2007). The ground was simulated as a rigid platform in the software. The 3D contacts (Iaquinto and Wayne, 2010) were implemented between adjacent bones as well as between the bones and ground plate in order to prevent overlap during the simulation. This was done by calculating the interference at each time step and applying an outward force if any overlap was detected. The magnitude of the force F was a function of material stiffness k (10,000 N/mm), penetration depth g , exponent e (1.75), penetration velocity, and damping coefficient c (400 Ns/mm), with the penetration at maximum damping set to its lowest allowable value ($d = 0.001$ mm).

$$F = kg^e + \left(\frac{dg}{dt} \right) f(c, d)$$

Friction was neglected to simulate cartilage effects. In the model the tibia was only allowed to move vertically (one degree of freedom). The fibula, talus, and calcaneus were allowed to move in all six degrees of freedom, leaving bone motion to be a function of ligament behavior, surface contact, and external perturbations. For simplification purposes, the remaining bones of the foot (tarsal and metatarsal bones) were fused together and moved as a unit, with its motion primarily dependent on motions of the talus and calcaneus.

Injury simulation

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20
21
22
23
24
25
26
27
28
29
30
31
32
33
34
35
36
37
38
39
40
41
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

At the beginning of the simulation, three times body weight (1840 N) was applied to the proximal end of the model to simulate dynamic weight bearing (Cavanagh and LaFortune, 1980), distributing one-sixth of the load on the fibula and the rest on the tibia (Lambert, 1971; Wei et al., 2011b). The 3D, temporal kinematic data from the case report (Fong et al., 2009b), i.e. inversion-time data, plantarflexion-time data, and internal rotation-time data, were used as input for simulations (Wei et al., 2011a). These actual motions placed the neutral foot model in the starting position of the athlete's ankle in order to begin the simulation. Three motor elements were used to drive the motion of the bones, two on the talus for plantarflexion and internal rotation, respectively, and one on the calcaneus for inversion. The three axes of rotation were set as: (1) dorsi-plantarflexion – fixed to the talus through its estimated center and initially oriented medial-lateral; (2) internal-external rotation – along the tibial axis; and (3) inversion-eversion – fixed to the calcaneus through its estimated center and perpendicular to the previous two (initially oriented anteroposterior) (Figure 1). These axes of rotation were based on a joint coordinate system (Wu et al., 2002), as the same system was utilized to calculate the injurious kinematics (Fong et al., 2009b). To systematically investigate the contribution of each motion and their combinations, four motion simulations were performed: (1) pure inversion; (2) inversion plus plantarflexion; (3) inversion plus internal rotation; and (4) a combination of inversion, plantarflexion, and internal rotation. The SolidWorks Motion package from the simulation software was used to execute these simulations. Continuous rotation-time data were interpolated (using the Akima Spline method) into the SolidWorks Motion to drive the talus and/or the calcaneus movement. Ligament strains were estimated from the model (Iaquinto and Wayne, 2011; Wei et al., 2011b). Resistive moments, deduced by inverse dynamics from the motors along the three axes of rotation, were also obtained from the model analysis.

153

154 RESULTS

155 Under inversion alone, the CaFL was strained the most at 12% (Figure 2). The ATaFL and
156 LTaCL were strained to approximately 10%. When temporal plantarflexion or internal rotation
157 was added, both the CaFL and ATaFL were strained to approximately 14-16%. With all three
158 motions combined the ATaFL was strained the most at 20% followed by the CaFL at 16%.
159 Temporal strain profiles of the ATaFL and CaFL were compared, along with the foot motions
160 and ankle joint moments (Figure 3). There was a switch in magnitude of strains from the CaFL to
161 the ATaFL at 0.16 s after footstrike (heel strike). The highest moment was 23 Nm for inversion
162 followed by 11 Nm for internal rotation (Figure 3c).

163

164 DISCUSSION

165 A grade I ankle sprain is characterized as minimal tearing of ligament fibers (Noyes et al., 1989).
166 Rupture strain of ankle ligaments can be estimated in the range of 30-35% (Beumer et al., 2003;
167 Funk et al., 2000). Ligament collagen fibers are thought to tear when half of the rupture strain is
168 reached (Yahia et al., 1990). The ATaFL strain in the current study reached 15-20%, which is
169 approximately one-half of the suggested rupture strain. Previous studies also report that a 10 Nm
170 external moment applied to the foot will cause pain and discomfort, and an external moment of
171 41-45 Nm generates ankle fractures (Markolf et al., 1989). By investigating a supination injury
172 case, the current study indicated that an ankle joint moment up to 23 Nm may produce a grade I
173 ankle sprain.

174

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20
21
22
23
24
25
26
27
28
29
30
31
32
33
34
35
36
37
38
39
40
41
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

175 Clinically the mechanism of an ankle supination sprain has been associated with a combination
176 of inversion and plantarflexion. In contrast, Kristianslund et al. (2011) and Mok et al. (2011)
177 suggest the important role of internal rotation. Mok et al. (2011) also suggest that future designs
178 of injury prevention measures may only need to consider inversion and internal rotation. Yet, the
179 current study showed that either plantarflexion or internal rotation, when added to inversion,
180 increased the ATaFL strains from 10% to approximately 15-16%. And the temporal combination
181 of all three motions raised the ATaFL strain to over 20% (Figure 2), suggesting that, in addition
182 to inversion, both plantarflexion and internal rotation may be important in generating supination
183 ankle sprains, as estimated by this one case study.

184
185 While a few studies have presented constant stiffnesses for ankle ligaments (Attarian et al., 1985;
186 Siegler et al., 1988), nonlinear and viscoelastic behavior in those ligaments has been observed
187 (Attarian et al., 1985; Funk et al., 2000). Ankle ligament strains under a physiological range of
188 ankle motion are thought to be in the range of 5-10% (Colville et al., 1990; Ozeki et al., 2002).
189 Funk et al. (2000) reported the ankle ligament force-strain curves have a typical toe region up to
190 6% strain followed by a loading region (nearly linear) up to 20% strain. Although the linear
191 approximation used in the current study may have underestimated ankle ligament strains, the
192 error was likely minimal (within 3%) for a strain range of 16-20%, while larger errors (up to
193 50%) could be involved for strains less than 15% (Funk et al., 2000). Additionally, a previous
194 study by others has concluded that a viscoelastic assumption of the ankle ligaments could be
195 neglected for very slow ($< 0.0001/s$) or very fast ($> 1/s$) strain rates, but substantial effects may
196 exist on ligament behavior for intermediate strain rates (Funk et al., 2000). The modeled

197 ligaments in the current study experienced strain rates approximately in the range of 1-1.5/s,
198 implying that the viscoelasticity of those ligaments may be negligible.

199
200 Other assumptions of the model and the simulation should also be noted. Firstly, the tarsal and
201 metatarsal bones were fused together and moved as a unit in the model for simplification
202 purposes. Although the ligaments connecting the tarsal and metatarsal bones may be less
203 important than those listed in Table 1, this fusion could have caused an overestimation of the
204 joint moments calculated from the model, especially for the internal rotation moment. Secondly,

205 model sensitivity to the parameters chosen to define the contact was evaluated by individually
206 varying the stiffness, exponent, and damping coefficient up and down one order of magnitude,
207 similar to a previous study by others ~~the impact of damping coefficients, chosen to define the~~
208 ~~contact, on bone motion and ligament strain was not investigated in the current study. However a~~
209 ~~recent study, by others, using a similar modeling approach to study the wrist biomechanics varies~~
210 ~~the contact parameters up and down one order of magnitude and concludes little (<1%) to no~~
211 ~~change in range of motion of the wrist predicted by the model~~ (Majors and Wayne, 2011). Joint
212 moment was most affected by the exponent term (4.3%) followed by the damping parameter
213 (2.0%) and the stiffness value (1.5%). Varying these parameters provided no change in the
214 ligament strain results predicted by the model. Thirdly, a constant weight bearing load was used

215 in the current study due to the unavailability of ground reaction force (GRF) data from the case
216 report. In addition, this constant load was applied in the beginning of simulation, not at
217 heelstrike, because the time for the GRF to reach its maximum was unknown. Future
218 experimental studies with use of a force plate may provide detailed GRF data (both magnitude
219 and duration) so that a simulated weight bearing load could be applied at footstrike. Finally, the

1
2
3 220 effect of a lack of the proximal tibiofibular joint in the model on potential alterations of the ankle
4
5
6 221 stability is currently unknown.
7
8 222
9
10 223 While this computational ankle model has provided insights into motions responsible for ankle
11
12 224 injury by simulating both *in vitro* (Wei et al., 2012; Wei et al., 2010) and *in vivo* experiments
13
14 225 (Wei et al., 2011a), it has been based on only a generic model, without potential effects of the
15
16 226 musculature. Therefore joint moments may have been underestimated (Wei et al., 2011a).
17
18 227 Additionally, the effects of anatomical differences between the injured subject and the cadaver
19
20 228 ankle used to build the model are unknown. Future studies that investigate ankle injury
21
22 229 mechanisms and prevention strategies may want to build subject-specific models, incorporate
23
24 230 muscle effects, and potentially integrate the nonlinear and viscoelastic responses of ankle
25
26 231 ligaments.
27
28
29
30
31
32 232

33
34 233 **ACKNOWLEDGEMENTS**

35
36 234 The authors thank Dr. Seungik Baek from Michigan State University for providing the MIMICS
37
38 235 software.
39
40
41 236
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

REFERENCES

- Attarian DE, McCrackin HJ, DeVito DP, McElhaney JH, Garrett WE, Jr. 1985. Biomechanical characteristics of human ankle ligaments. *Foot Ankle* 6(2):54-58.
- Bahr R, Pena F, Shine J, Lew WD, Engebretsen L. 1998. Ligament force and joint motion in the intact ankle: a cadaveric study. *Knee Surg Sports Traumatol Arthrosc* 6(2):115-121.
- Beumer A, van Hemert WL, Swierstra BA, Jasper LE, Belkoff SM. 2003. A biomechanical evaluation of the tibiofibular and tibiotalar ligaments of the ankle. *Foot Ankle Int* 24(5):426-429.
- Cavanagh PR, LaFortune MA. 1980. Ground reaction forces in distance running. *J Biomech* 13(5):397-406.
- Colville MR, Marder RA, Boyle JJ, Zarins B. 1990. Strain measurement in lateral ankle ligaments. *Am J Sports Med* 18(2):196-200.
- Delahunt E, Monaghan K, Caulfield B. 2006. Altered neuromuscular control and ankle joint kinematics during walking in subjects with functional instability of the ankle joint. *Am J Sports Med* 34(12):1970-1976.
- Fong DT, Chan YY, Mok KM, Yung P, Chan KM. 2009a. Understanding acute ankle ligamentous sprain injury in sports. *Sports Med Arthrosc Rehabil Ther Technol* 1:14.
- Fong DT, Hong Y, Chan LK, Yung PS, Chan KM. 2007. A systematic review on ankle injury and ankle sprain in sports. *Sports Med* 37(1):73-94.
- Fong DT, Hong Y, Shima Y, Krosshaug T, Yung PS, Chan KM. 2009b. Biomechanics of supination ankle sprain: a case report of an accidental injury event in the laboratory. *Am J Sports Med* 37(4):822-827.

1
2
3 259 Funk JR, Hall GW, Crandall JR, Pilkey WD. 2000. Linear and quasi-linear viscoelastic
4
5
6 260 characterization of ankle ligaments. *J Biomech Eng* 122(1):15-22.
7
8 261 Iaquinto JM, Wayne JS. 2010. Computational model of the lower leg and foot/ankle complex:
9
10 262 application to arch stability. *J Biomech Eng* 132(2):021009.
11
12 263 Iaquinto JM, Wayne JS. 2011. Effects of surgical correction for the treatment of adult acquired
13
14 264 flatfoot deformity: a computational investigation. *J Orthop Res* 29(7):1047-1054.
15
16
17 265 Kristianslund E, Bahr R, Krosshaug T. 2011. Kinematics and kinetics of an accidental lateral
18
19 266 ankle sprain. *J Biomech* 44(14):2576-2578.
20
21
22 267 Lambert KL. 1971. The weight-bearing function of the fibula. A strain gauge study. *J Bone Joint*
23
24 268 *Surg Am* 53(3):507-513.
25
26
27 269 Liacouras PC, Wayne JS. 2007. Computational modeling to predict mechanical function of
28
29 270 joints: application to the lower leg with simulation of two cadaver studies. *J Biomech Eng*
30
31 271 129(6):811-817.
32
33
34 272 Majors BJ, Wayne JS. 2011. Development and validation of a computational model for
35
36 273 investigation of wrist biomechanics. *Ann Biomed Eng* 39(11):2807-2815.
37
38
39 274 Markolf KL, Schmalzried TP, Ferkel RD. 1989. Torsional strength of the ankle in vitro. The
40
41 275 supination-external-rotation injury. *Clin Orthop Relat Res* (246):266-272.
42
43
44 276 Mok KM, Fong DT, Krosshaug T, Engebretsen L, Hung AS, Yung PS, Chan KM. 2011.
45
46 277 Kinematics analysis of ankle inversion ligamentous sprain injuries in sports: 2 cases
47
48 278 during the 2008 Beijing Olympics. *Am J Sports Med* 39(7):1548-1552.
49
50
51 279 Noyes FR, Grood ES, Torzilli PA. 1989. Current concepts review. The definitions of terms for
52
53 280 motion and position of the knee and injuries of the ligaments. *J Bone Joint Surg Am*
54
55 281 71(3):465-472.
56
57
58
59
60

- 282 Ozeki S, Yasuda K, Kaneda K, Yamakoshi K, Yamanoi T. 2002. Simultaneous strain
283 measurement with determination of a zero strain reference for the medial and lateral
284 ligaments of the ankle. *Foot Ankle Int* 23(9):825-832.
- 285 Pfaeffle HJ, Tomaino MM, Grewal R, Xu J, Boardman ND, Woo SL, Herndon JH. 1996. Tensile
286 properties of the interosseous membrane of the human forearm. *J Orthop Res* 14(5):842-
287 845.
- 288 Siegler S, Block J, Schneck CD. 1988. The mechanical characteristics of the collateral ligaments
289 of the human ankle joint. *Foot Ankle* 8(5):234-242.
- 290 Wei F, Braman JE, Weaver BT, Haut RC. 2011a. Determination of dynamic ankle ligament
291 strains from a computational model driven by motion analysis based kinematic data. *J*
292 *Biomech* 44(15):2636-2641.
- 293 Wei F, Hunley SC, Powell JW, Haut RC. 2011b. Development and validation of a computational
294 model to study the effect of foot constraint on ankle injury due to external rotation. *Ann*
295 *Biomed Eng* 39(2):756-765.
- 296 Wei F, Meyer EG, Braman JE, Powell JW, Haut RC. 2012. Rotational stiffness of football shoes
297 influences talus motion during external rotation of the foot. *J Biomech Eng*
298 134(4):041002.
- 299 Wei F, Villwock MR, Meyer EG, Powell JW, Haut RC. 2010. A biomechanical investigation of
300 ankle injury under excessive external foot rotation in the human cadaver. *J Biomech Eng*
301 132(9):091001.
- 302 Wu G, Siegler S, Allard P, Kirtley C, Leardini A, Rosenbaum D, Whittle M, D'Lima DD,
303 Cristofolini L, Witte H, Schmid O, Stokes I. 2002. ISB recommendation on definitions of

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20
21
22
23
24
25
26
27
28
29
30
31
32
33
34
35
36
37
38
39
40
41
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

304 joint coordinate system of various joints for the reporting of human joint motion--part I:
305 ankle, hip, and spine. International Society of Biomechanics. *J Biomech* 35(4):543-548.
306 Yahia L, Brunet J, Labelle S, Rivard CH. 1990. A scanning electron microscopic study of rabbit
307 ligaments under strain. *Matrix* 10(1):58-64.

For Peer Review Only

FIGURE LEGENDS

Figure 1: Posterior (a), medial (b), and lateral (c) views of the ankle model showing the axes of rotation used in the simulations (dotted lines in a and b) and the locations of 16 simulated ligaments (c): the interosseous ligaments (IOL-I and IOL-II); the anterior and posterior tibiofibular (ATiFL and PTiFL); the calcaneofibular (CaFL); the anterior and posterior talofibular (ATaFL and PTaFL); the anterior and posterior tibiotalar (ATiTL and PTiTL); the tibionavicular (TiNL); the talonavicular (TaNL); the calcaneocuboid (CaCL); the calcaneonavicular (CaNL); the medial, central, and lateral plantar fascia (MPF, CPF, and LPF). For clarity, the lateral, interosseous, medial, and posterior talocalcaneal ligaments (LTaCL, ITaCL, MTaCL, and PTaCL) between the talus and the calcaneus were set invisible. Ligament initial length and stiffness were documented in Table 1.

Figure 2: Maximum strains estimated from the model in various selected ligaments ($> 2\%$ strain) for different ankle motions.

Figure 3: Normalized kinematic data from Fong et al. (2009b) were re-plotted (a). Values were normalized between their initial and peak measures, i.e. plantarflexion: $-23.6^{\circ} \sim 1.0^{\circ}$; inversion: $15.6^{\circ} \sim 47.6^{\circ}$; internal rotation: $-12.3^{\circ} \sim 10.0^{\circ}$. Temporal profiles of strains (b) in the ATaFL and CaFL, and ankle joint moments (c) during the injury incident were estimated from the model by applying a combination of inversion, plantarflexion, and internal rotation simultaneously.

1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
18
19
20
21
22
23
24
25
26
27
28
29
30
31
32
33
34
35
36
37
38
39
40
41
42
43
44
45
46
47
48
49
50
51
52
53
54
55
56
57
58
59
60

TABLE

Table 1: Twenty ligaments were simulated in the model. The initial length was the length prior to the application of an *in situ* strain. The stiffness was adapted from Wei et al. (2011b).

Ligament names	Abbreviation	Initial length (mm)	Stiffness (N/mm)
Interosseous I	IOL-I	18.0	400 (Pfaeffle et al., 1996)
Interosseous II	IOL-II	17.2	400 (Pfaeffle et al., 1996)
Anterior Tibiofibular	ATiFL	18.1	90 (Siegler et al., 1988)
Posterior Tibiofibular	PTiFL	22.7	90 (Beumer et al., 2003)
Calcaneofibular	CaFL	41.3	70 (Liacouras and Wayne, 2007)
Anterior Talofibular	ATaFL	32.2	90 (Siegler et al., 1988)
Posterior Talofibular	PTaFL	18.8	70 (Liacouras and Wayne, 2007)
Anterior Tibiotalar	ATiTL	18.0	70 (Liacouras and Wayne, 2007)
Posterior Tibiotalar	PTiTL	16.0	80 (Liacouras and Wayne, 2007)
Tibionavicular	TiNL	39.3	40 (Siegler et al., 1988)
Talonavicular	TaNL	11.0	70 (Liacouras and Wayne, 2007)
Interosseous Talocalcaneal	ITaCL	8.7	70 (Liacouras and Wayne, 2007)
Lateral Talocalcaneal	LTaCL	12.2	70 (Liacouras and Wayne, 2007)
Medial Talocalcaneal	MTaCL	8.5	70 (Liacouras and Wayne, 2007)
Posterior Talocalcaneal	PTaCL	29.9	70 (Liacouras and Wayne, 2007)
Calcaneocuboid	CaCL	26.2	70 (Liacouras and Wayne, 2007)
Calcaneonavicular	CaNL	26.7	70 (Liacouras and Wayne, 2007)
Medial Plantar Fascia	MPF	147.9	200 (Iaquinto and Wayne, 2010)
Central Plantar Fascia	CPF	146.5	230 (Iaquinto and Wayne, 2010)
Lateral Plantar Fascia	LPF	133.2	180 (Iaquinto and Wayne, 2010)

Posterior (a), medial (b), and lateral (c) views of the ankle model showing the axes of rotation used in the simulations (dotted lines in a and b) and the locations of 16 simulated ligaments (c): the interosseous ligaments (IOL-I and IOL-II); the anterior and posterior tibiofibular (ATiFL and PTiFL); the calcaneofibular (CaFL); the anterior and posterior talofibular (ATaFL and PTaFL); the anterior and posterior tibiotalar (ATiTL and PTiTL); the tibionavicular (TiNL); the talonavicular (TaNL); the calcaneocuboid (CaCL); the calcaneonavicular (CaNL); the medial, central, and lateral plantar fascia (MPF, CPF, and LPF). For clarity, the lateral, interosseous, medial, and posterior talocalcaneal ligaments (LTaCL, ITaCL, MTaCL, and PTaCL) between the talus and the calcaneus were set invisible. Ligament initial length and stiffness were documented in Table 1.

127x97mm (300 x 300 DPI)

Maximum strains estimated from the model in various selected ligaments (> 2% strain) for different ankle motions.

48x26mm (600 x 600 DPI)

Normalized kinematic data from Fong et al. (2009b) were re-plotted (a). Values were normalized between their initial and peak measures, i.e. plantarflexion: $-23.6^{\circ} \sim 1.0^{\circ}$; inversion: $15.6^{\circ} \sim 47.6^{\circ}$; internal rotation: $-12.3^{\circ} \sim 10.0^{\circ}$. Temporal profiles of strains (b) in the ATaFL and CaFL, and ankle joint moments (c) during the injury incident were estimated from the model by applying a combination of inversion, plantarflexion, and internal rotation simultaneously.

240x524mm (300 x 300 DPI)