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# Cartilage Stiffness Effect on Foot Biomechanics of Chinese Bound Foot: A Finite Element Analysis

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## ABSTRACT

The purpose of this study is to investigate the effect of cartilage stiffness on inner foot biomechanics of Chinese bound foot while balanced standing using finite element method. A three-dimensional FE model of bound foot involving 28 bones, 72 ligaments, 5 plantar fascia, cartilages, and encapsulated soft tissue was constructed and validated. To conduct the sensitivity analysis of cartilage stiffness, the incremental Young's modulus of 1MPa, 5MPa, 10MPa, and 15MPa were assigned to the cartilage. 25% of the body weight was applied to the Achilles tendon to adjust the anterior-posterior displacement of center of pressure agreeable with the measured result. As the Young's modulus of cartilage increased, the peak von Mises stress in the fifth metatarsal increased obviously, while that in the calcaneus remains unchanged. The plantar fascia experienced reduced total tension with stiffer cartilage. The cartilage stiffening also caused a general increase of contact pressure at mid- and forefoot joints. Cartilage stiffening due to foot binding gave rise to risks of foot pain and longitude arch damage. Knowledge of this study contributes to the understanding of bound foot biomechanical behavior and demonstrating the mechanism of long-term injury and function damage in terms of weight-bearing due to foot binding.

## Introduction

The human foot is a complex mechanical structure consisting of 28 bones, a number of ligaments, muscles, and other connective tissues. As the terminal portion of limb, it bears body weight and allows locomotion. Regardless of causes, the bone or joint deformity may cause functional impairments (Rao et al., 2012). Foot binding, an unique Chinese custom, had ever gained great popularity and enduring influence on Chinese history. For over 1000 years, young girls about four to seven years old curled (folded) their second to fifth phalanges under the sole wrapped tightly with bandage in order to compress the foot. In addition to phalange fracture, the metatarsals were rearranged into an extremely high arch (Zhang et al., 2014) and the calcaneus was reoriented sagittally towards the alignment of the long axis of the lower leg (Howard and Pillinger, 2010). Thus, a smaller and narrower foot size forms, which was considered to be a means of displaying social status and was correspondingly adopted as a symbol of beauty in ancient Chinese culture (Greenhalgh, 1977). The bound foot was also believed as the object of great sensuality from a masculine point of view (Blake, 1994) and intended to limit mobility of women, resulting in substantial disability in basic physical activity (Greenhalgh, 1977).

Previous research interest of foot binding has widely focused on the origin, social and historical background, cultural significance, and feminist perspective. A few studies pay attention to analysis on morphological characteristics of bound feet initially reported by Jackson (Jackson, 1990) with a systematic description. Further studies mainly concern imaging analysis based on footprints and computed tomography (CT) images (Munk and Poon, 1996; Reischl et al., 2008; Howard and Pillinger,

2010). Three-dimensional (3D) skeleton model of bound foot has also been reconstructed by segmenting X-ray images in image processing software. Apart from the deformity of the second to fifth phalanges, the calcaneal-first metatarsal angle, talo-first metatarsal angle and horizontal metatarsal angle illustrated a broken longitudinal arch which is extremely high compared with that of a normal foot (Zhang et al., 2014). With regard to biomechanical consequences, Zhang et al. (Zhang et al., 2015) investigated gait kinematics of bound feet women and results indicated reduced range of motion in the ankle. Gu et al. (Gu et al., 2015) compared plantar pressure between bound foot and normal foot, observing increased plantar loading on the rearfoot in bound foot.

The gross pressure distribution recorded from experimental measurement could unilaterally interpret the functional role of different anatomical components. The states of internal stress/strain in bones and soft tissues and contact pressure at joints remain unaddressed due to the difficulties and limitations of the experimental approach. Complementary to measurements, finite element (FE) analysis is capable of simulating the mechanical responses of biological systems via a numerical model, complex material properties and varying boundary and loading conditions. Many geometrically accurate 3D FE models have been developed to evaluate the biomechanics of foot-ankle complex. Using FE method, Cheung et al. (Cheung et al., 2006) reported that the increasing Achilles tendon force caused increased strain on plantar fascia. Sun et al. (Sun et al., 2012) predicted higher the stress and strain on the plantar fascia and metatarsal in the high-arched foot compared with low-arched foot. A non-linear finite element model was developed recently to assess the effect of tendon force on foot biomechanics by a force sensitivity study (Morales-Orcajo et al., 2017).

Despite of the severe deformity, some women with bound feet, especially those living in rural areas, working outdoor has been reported (Qin et al., 2015), indicating the certain remaining mechanical competence of the bound foot. An in-depth analysis on internal foot biomechanics will be of great clinic and historic significance in better understanding of injury mechanism and changes in foot function as a result of foot binding. Literature documented that cartilage is regularly subjected to high levels of stress is stiffer than that under low stress levels (Swann and Seedhom, 1993). Thus, the cartilage stiffness of bound foot is supposed to increase due to the chronic tight compression of the bandage. The purpose of this study is to investigate cartilage stiffness effect on the internal foot structure of Chinese bound feet while weight-bearing by predicting stress/strain within and between different structural components and contact pressure at joints using finite element analysis.

## Methods

**Finite element model construction.** The geometrically accurate FE model of the bound foot was reconstructed from CT images of a 92-year old female (height: 153cm; weight: 47.5kg). The coronal CT images were obtained with a space interval of 2 mm without weight-bearing. The images were segmented using MIMICS 16.0 (Materialise, Leuven, Belgium) to obtain the boundaries of the skeleton and the soft tissue. The uneven surfaces caused by the stacking of the medical images were processed using Geomagic Studio 2013 (Geomagic, Inc., Research Triangle Park, NC, USA). Each surface component was then imported into Solidworks 2016 (SolidWorks Corporation, Massachusetts, USA) individually to form solid parts. Except the first to fifth phalanges were fused, the other bony components were sagittally connected with cartilaginous structures created by Boolean operations allowing relative bone movements. The encapsulated soft tissue was subtracted from the whole foot volume by the bony and cartilaginous structures. The whole foot model consisted of 28 foot bony segments, including tibia, fibula, talus, calcaneus, cuboid, navicular, three cuneiforms, five metatarsals and 14 phalanges. Link elements (tension-only) were used to simulate ligaments bearing the tension load. A total number of 72 ligaments and five plantar fascia were included and defined by connecting corresponding anatomical locations on the bones by reference to an anatomy book [13].

**Figure 1.** Three-dimensional finite element model and application of boundary and loading conditions.

97 GRF: ground reaction force.

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99 Mesh sensitivity was performed to ensure the accuracy and validity of the model and optimum  
100 requirement on the computational resources. The tissues of bone, cartilage and encapsulated soft tissue  
101 were meshed tetrahedral solid elements (Gu et al., 2010a). The ANSYS Workbench 17.0 (ANSYS,  
102 Inc., Canonsburg, USA) was used for FE analysis. Automated surface-to-surface contact algorithm in  
103 ANSYS Workbench was used to simulate the interaction between the surfaces of the cartilaginous and  
104 bony structures. The contact algorithm considers both contact elements faces which in turn prevent  
105 any penetration of the nodes on the target surface into the master surface. All the bones and cartilages  
106 were bonded to the encapsulated soft tissue.

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108 **Material properties.** All the materials except for the soft tissue were considered isotropic and linearly  
109 elastic with material properties obtained from previous literature (Cheung et al., 2005). The material  
110 constants of Young's modulus  $E$  and Poisson's ratio  $\nu$  were given to describe the elasticity. To conduct  
111 the sensitivity analysis of cartilage stiffness, the incremental Young's modulus of 1MPa, 5MPa, 10MPa,  
112 and 15MPa were assigned to the cartilage. Young's modulus of 1MPa was chosen as a reference value  
113 to present the normal cartilage stiffness (Cheung et al., 2005). The encapsulated soft tissue was set as  
114 nonlinear hyperelastic material which was defined as Moonley-Rivlin model (Coefficients:  
115  $C10=0.08556$ ;  $C01= -0.05841$ ;  $C20=0.03900$ ;  $C11=-0.02319$ ;  $C02=0.00851$ ;  $D1=3.65273$ ;  
116  $D2=0.00000$ ) (Qiu et al., 2011). The element types and material properties used are listed in Table 3.

117

118 **Boundary and loading conditions.** A balanced standing condition was considered for the FE analysis.  
119 The superior surfaces of the encapsulated soft tissue, distal tibia and distal fibula were fixed. The foot-  
120 ground interaction was simulated as a foot-plate system (Fig. 6). The plate was assigned with an elastic  
121 property to simulate the concrete ground support (Table 3). The plate was allowed to move freely only  
122 in the vertical direction. A vertical ground reaction force (GRF) of a half-body weight (210 N) was  
123 applied at the inferior surface of the plate. The interaction between the foot plantar surface and the  
124 superior surface of the plate was simulated as contact with friction. The coefficient of friction was set  
125 to 0.6 (Cheung et al., 2005).

126 Only the Achilles tendon force (ATF) was considered, while other intrinsic and extrinsic muscle  
127 forces were neglected. Five equivalent force vectors representing the Achilles tendon force were  
128 applied at the insertion of the posterior calcaneus (Fig. 6). The normal ATF is commonly estimated as  
129 50% of the force applying on the foot while balanced standing in research (Chen et al., 2010). Cheung  
130 et al. (Cheung et al., 2006) reported that the increase of ATF resulted in anterior shifting of center of  
131 pressure (COP). In order to simulate the posterior shifting of COP due to foot binding (Gu et al., 2015),  
132 the magnitude of normal ATF multiplied by a factor of 0.8, 0.5, 0.3 was also applied to Achilles tendon  
133 respectively. Fig. 3a shows the definition of COP ( $\bar{x}$ ,  $\bar{y}$ ) location which was calculated as

134

$$135 \bar{x} = \sum p_i x_i / \sum p_i, \quad \bar{y} = \sum p_i y_i / \sum p_i \quad (1)$$

136

137 Where  $p_i$  is the nodal pressure values at each finite element node ( $x_i, y_i$ ).

138

139 **Model validation.** For the purpose of balancing reliability and time cost, the basic pressure-based  
140 validation approach is more advocated and used in the majority of studies (Behforootan et al., 2017).  
141 The FE model was validated by comparing plantar pressure distribution and peak pressure obtained  
142 from computational prediction in FE software and experimental measurement by a Novel emed  
143 pressure platform (Novel, Munich, Germany) in a standing position. The measurement was performed  
144 on the same subject who had volunteered for the medical image scanning. The participant was asked  
145 to stand still on the pressure platform for five seconds. Data of the middle three seconds were selected  
146 and averaged.

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## Results

**Finite element model validation.** Fig. 2a shows the peak pressure on the forefoot and the rearfoot measured by Novel system and predicted by simulation with different ATF and  $E=1\text{MPa}$  for Young's modulus of cartilage. The values of peak pressure predicted from different ATF showed to be nuanced and were in good agreement with that tested in the *in vivo* measurement. Comparison of the anterior-posterior displacement of COP between measurement and predictions are shown in Fig. 1b. It deviates largely with regard to 1 normal ATF and 0.3 normal ATF, while there is a better consistency between measurement and prediction for 0.5 normal ATF. Therefore, 52.5N was applied to Achilles tendon of the FE model of bound foot for further analysis. With a reference value,  $E=1\text{MPa}$  for Young's modulus of cartilage, and 0.5 normal ATF, the FE model was validated with plantar pressure measurement as shown in Fig. 3. It shows that the plantar pressure distribution from experimental measurement and computational prediction were comparable. The peak pressure of 0.148MPa concentrated on the heel region was predicted as compared to that of 0.137MPa measured.

**Figure 2.** Comparison of peak pressure (a) and anterior displacement (b) of center of pressure between measurement and predictions of different Achilles tendon forces, with Young's modulus of cartilage,  $E=1\text{MPa}$ .

**Figure 3.** Comparison of plantar pressure distribution between measurement (a) and prediction (b), with Young's modulus of cartilage,  $E=1\text{MPa}$ , and 50% of normal Achilles tendon force.

**2-dimensional COP displacement.** There was a trivial effect of cartilage stiffness on the COP displacement (Fig. 4b). In the anterior-posterior direction, it varied relatively larger than in the medial-lateral direction, but the difference is slight. The 2-dimensional displacement of ( $\text{COP}_x=29.03\text{ mm}$ ,  $\text{COP}_y=55.83\text{ mm}$ ), ( $\text{COP}_x=29.17\text{ mm}$ ,  $\text{COP}_y=55.48\text{ mm}$ ), ( $\text{COP}_x=29.24\text{ mm}$ ,  $\text{COP}_y=55.9\text{ mm}$ ), and ( $\text{COP}_x=29.29\text{ mm}$ ,  $\text{COP}_y=56.05\text{ mm}$ ) were predicted with regard to the simulated case of Young's modulus of cartilage,  $E=1\text{MPa}$ ,  $E=5\text{MPa}$ ,  $E=10\text{MPa}$ ,  $E=15\text{MPa}$ , respectively.

**Figure 4.** Definition of location (a) and comparison of 2-dimensional displacement (b) of center of pressure (COP) between different Young's modulus of cartilage.

**Peak von Mises stresses in bones.** Table 1 displays the peak von Mises stresses in selected bones with incremental cartilage stiffness. In all simulated cases, the highest Mises stresses were predicted at the mid-shaft of the third metatarsal, followed by the first metatarsal (Fig. 5). As comparing to the reference value,  $E=1\text{MPa}$ , a general increase in peak von Mises stress in bones was observed in the simulated cases of larger Young's modulus of cartilage. It remained unchanged or minimal changes in the peak stress in the second metatarsal and calcaneus with increased cartilage Young's modulus. Compared with  $E=1\text{MPa}$ , the peak von Mises stress increased 65.9% in navicular, 55.8% in the fifth metatarsal, 44.8% in cuboid, 16.5% in the first metatarsal, 15.9% in the third metatarsal, 11% in talus, and 2.8% in the fourth metatarsal in the simulated case of  $E=5\text{MPa}$ ; As the Young's modulus increased to 10MPa and 15MPa, sequential increase of stress only existed in the distal site of the fifth metatarsal and the mid-shaft of the first metatarsal, but the differences are very small (Fig. 4).

**Figure 5.** von Mises stress in five metatarsals, with Young's modulus of cartilage, (a)  $E=1\text{MPa}$ , (b)  $E=5\text{MPa}$ , (c)  $E=10\text{MPa}$ , (d)  $E=15\text{MPa}$ .

198 **Plantar fascia tension and strain.** Table 2 displays the predicted plantar fascia tension and strain with  
199 different cartilage stiffness. The plantar fascia experienced reduced total tension with the increased  
200 Young's modulus of cartilage. The first and fourth ray sustained major tension loading and strain in all  
201 simulated cases. They underwent about 60% overall tension. The fifth ray played a minimum role  
202 assuming less than 11% overall tension. There was a slight decrease of plantar pressure strain in the  
203 fourth ray as the Young's modulus of cartilage increased, while the changes in other rays were trivial.  
204

205 **Contact pressure of midfoot and forefoot joints.** Fig. 6 shows the effect of cartilage stiffness on the  
206 contact pressure of joints at the midfoot and forefoot. There was a general increase of contact pressure  
207 at mid- and forefoot joints with the increase Young's modulus of cartilage. The increased cartilage  
208 stiffness produced obvious increase in joint contact pressure when the Young's modulus increasing  
209 from 1MPa to 5MPa. At the midfoot, joint contact pressure on the medial side (talonavicular and  
210 medial cuneonavicular joints) was larger than that on the lateral side (calcaneocuboid, intermediate  
211 cuneonavicular, and lateral cuneonavicular joints). At the forefoot, joint contact pressure at the fourth  
212 and fifth tarsometatarsal joint and metatarsophalangeal joint (MTP) was larger than others. Reduced  
213 contact pressure due to the stiffer cartilage was found at the intermediate cuneonavicular joint.  
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215  
216 **Figure 6.** The effect of cartilage stiffness on the contact pressure of joints at midfoot and forefoot. C-  
217 N: cuneonavicular joint; T-M: tarsometatarsal joint; MTP: metatarsophalangeal joint.  
218

219 **Discussion**

220 The practice of foot binding spanned over 1000 years in Chinese history. It began in Southern  
221 Tang Dynasty, flourished in Song Dynasty, and finally was forbidden in the early 20th century (Qin et  
222 al., 2015). To date, there are still elder women with bound feet living in some rural and impoverished  
223 areas of China, but only few. Due to exposure to chronic intense pressure of the bandage, the property  
224 of articular cartilage may vary from normal state. This study explored the effect of cartilage stiffness  
225 on biomechanical consequences of bound foot during balanced standing.

226 Many studies have constructed FE models from normal feet or a pathological model modified  
227 from the normal one to investigate the internal biomechanical features of the foot with a great success  
228 (Wang et al., 2016). This study developed a complete foot FE model from an elder woman with bound  
229 foot. For a foot FE model, except for the bone tissue modeled in almost all relevant studies, there is a  
230 large percent studies included ligaments, tendons, and soft tissue (Behforootan et al., 2017). About half  
231 of the existed research involved cartilages in modeling FE foot model (Behforootan et al., 2017). For  
232 studies excluded cartilaginous components, they used frictionless contact elements allowing free  
233 movements between bones. Alternatively, as performed in the model of this study, it is possible to  
234 connect two adjacent bones with low stiffness elements, which provides more realistic simulation of  
235 articular behavior. As reported in previous study, the longitude displacement of COP deviated  
236 posteriorly for bound feet in comparison with normal feet. Consistently, the COP measured in the  
237 current study located at the heel region while balance standing. Base on the prediction from a FE foot  
238 model constructed by Cheung et al. (Cheung et al., 2006) that increasing Achilles tendon loading  
239 caused anterior shifting of COP in a standing position, this study attempted to simulate the posteriorly  
240 shifted COP of bound foot by decreasing the Achilles tendon force. A factor of 0.8, 0.5, and 0.3 of  
241 normal Achilles tendon force was included for simulation, and the predicted results showed that the  
242 factor of 0.5 achieved the most agreeable displacement of COP between prediction and measurement.  
243 The model was validated by plantar pressure measurement and the predicted plantar pressure  
244 distribution and COP location showed a good agreement with those measured by *in vivo* experiment.

245 From the FE predictions, the highest von Mises stress in bones was found concentrating at the  
246 mid-shaft of the third metatarsal. FE analysis of a normal foot model indicated similar location of the  
247 highest stress, however, the magnitude is about two-fold greater than that predicted in the bound foot

248 model (Cheung et al., 2005). Moreover, according to Cheung et al. (Cheung et al., 2005), the peak von  
249 Mises stress in calcaneus of the normal foot is more than three-fold greater than that predicted in the  
250 bound foot model. Normally, the calcaneus plays an important role in weight bearing, supporting about  
251 40% of the body weight (Rodgers, 1988). Due to the severe deformity at forefoot, the heel region of  
252 bound feet experienced increased plantar pressure. Higher plantar pressure in normal foot was regarded  
253 to be associated with increased von Mises stress in bones (Chen et al., 2001). Different from what was  
254 expected, the magnitude of stress in calcaneus of the bound foot showed to be lower than the stress in  
255 the first to fourth metatarsals. This may be caused by the thickening of the fat pad at the heel. With a  
256 five-fold increase of Young's modulus of cartilage, the stiffer cartilage affected the peak von Mises  
257 stress in most bones except for the second metatarsal and calcaneus, indicating higher risk of foot pain  
258 even after foot binding process complete. In fact, apart from the initial pain in the first year of foot  
259 binding, long-term pain was also complained (Stone, 2012). As the Young's modulus increased to  
260 10MPa and 15MPa, the first and fifth metatarsal showed minor increments of stress, indicating that  
261 these are possible areas are more vulnerable to pain as cartilages persistently stiffening during the  
262 lifetime.

263 The FE analysis showed reduced total tension force with the increase of Young's modulus of  
264 cartilage. For the incremental Young's modulus, the five rays of plantar fascia segments sustained  
265 14.82%, 14.58%, 14.35%, and 14.19% of the applied force on the plate, respectively, which is  
266 prominently less than the prediction (44% of applied force) from a normal foot model (Cheung et al.,  
267 2006). Regarding to the strain, all five segments decreased by one order of magnitude as to bound foot  
268 compared to that predicted from a normal foot (Yu et al., 2008). The reduced plantar fascia tension and  
269 strain demonstrated the broken structure of longitude foot arch. One FE study evidenced that the high-  
270 arched foot generated higher plantar fascia stress and strain than the normal foot (Sun et al., 2012).  
271 The current study predicted adverse result in bound foot, demonstrating that the extremely high arch  
272 is very likely to resulting in function damage in supporting the foot shape. Admittedly, the bound foot  
273 is associated with resembling the state of high-heeled shoes. Yu et al. (Yu et al., 2016) found decreased  
274 total tension and average strain of plantar fascia when the heel height increased from 0 to 5.08cm,  
275 however, this is still greater than the predictions from bound foot. Furthermore, the effect of cartilage  
276 stiffness on the plantar fascia strain was minimal. The change of strain in the fourth ray showed to be  
277 relatively noticeable with a gradual decrease as the Young's modulus of cartilage increased from the  
278 reference value to fifteen-fold. In a study evaluating the calcaneus functional adaptation in foot binding,  
279 the mechanical influence of plantar fascia was even disregarded (Reznikov et al., 2017).

280 The predicted contact pressure at midfoot and forefoot joints showed a general increase with the  
281 incremental Young's modulus of cartilage. Similar variation was observed in joint contact pressure  
282 with respect to the restricted foot due to ankle arthrodesis. Wang et al. (Wang et al., 2015) presented  
283 larger contact pressure at midfoot joint in the foot with ankle arthrodesis than a normal one. Despite  
284 the severe deformity of the fourth and fifth phalanges, the structure of MTP at these sites practically  
285 remains intact. The contact pressure at the fourth and fifth MTP exhibited higher than other MTP,  
286 which may result from the hyperflexed position of the phalanges. At the midfoot, the medial  
287 cuneonavicular joint showed prominent contact pressure. As demonstrated previously, the excessive  
288 contact stress on articular surface is thought to be a predominant association with osteoarthritis  
289 (Buckwalter and Martin, 2006). The medial side of midfoot is considered an area of high risk of  
290 arthritis. Effect of cartilage stiffness on contact pressure was most pronounced in the fifth MTP. It can  
291 be speculated that the increased contact pressure at the fifth MTP potentially aggravates the pain caused  
292 by foot binding.

293 It should be noted that the effect of cartilage stiffening considered in this study was simplified by  
294 a uniform increase of Young's modulus over all cartilages of the entire foot. In the real cases, the  
295 degree of cartilage stiffening may vary at different foot segments. Lacking specific information about  
296 mechanical properties of plantar fascia for bound foot, this study assumed the Young's Modulus,  
297 Poisson's Ratio, and cross-section area the same as those of the normal foot. Since the Achilles Tendon

298 plays a more important role while balanced standing, other intrinsic and extrinsic muscle forces were  
299 neglected. Based on validated plantar pressure distribution and COP location, the resulted predictions  
300 are able to represent the tendency of mechanical responses to stiffened cartilages. In addition, the foot  
301 binding was usually performed based on the experience of one's mother (Stone, 2012) with the main  
302 purpose of restricting foot size. An extremely high arch and folded phalanges would be formed  
303 commonly, while the relative rearrangement of different phalangeal segments may vary uniquely (e.g.  
304 fully folded phalanges and semi-folded phalanges). Knowledge of this study accounts for  
305 biomechanics of bound foot that is featured by severely damaged fourth and fifth phalanges.  
306

## 307 **Conclusion**

308 This study constructed a subject-specific FE model of Chinese bound foot and preliminarily  
309 showed its mechanical response to stiffened articular cartilages. Compared with previous  
310 biomechanical analysis that aimed to inform the external loading (i.e. plantar pressure) of bound feet,  
311 the numerical model developed an approach for predicting changes in internal tissue stress/strain due  
312 to the abnormal foot structure, which will therefore improve the comprehension of the theories of  
313 bound foot injury and function loss. More importantly, simulations on the effect of increased cartilage  
314 stiffness provided additional interpretation on the mechanism of aggravating foot pain and functional  
315 damage of longitude arch associated with foot binding. We found that the peak von Mises stress in the  
316 fifth metatarsal increased obviously as Young's modulus of cartilage increased. Cartilage stiffening  
317 also caused generally increased joint contact pressure at medial mid-foot and lateral forefoot. Larger  
318 bony stress and joint loading are the potentials for rising pain. The reduced total tension of plantar  
319 fascia with stiffer cartilage may indicate the progressively weakened weight-bearing ability of the  
320 longitude arch. Further work is supposed to analyzing higher loading condition of stance phase in gait.  
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383 **Table 1.** Peak von Mises stresses (Unit: MPa) in selected bones with different cartilage stiffness. M1:  
 384 the first metatarsal; M2: the second metatarsal; M3: the third metatarsal; M4: the fourth metatarsal;  
 385 M5: the fifth metatarsal; CAL: calcaneus; TAL: talus; NAV: navicular; CUB: cuboid.  
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Cartilage stiffness ( <i>E</i> )	M1	M2	M3	M4	M5	CAL	TAL	NAV	CUB
1-MPa	2.78	1.94	3.34	2.18	0.86	1.11	1.09	0.44	0.58
5-MPa	3.24	1.94	3.87	2.24	1.34	1.12	1.21	0.73	0.84
10-MPa	3.38	1.98	3.97	2.30	1.76	1.12	1.2	0.80	0.89
15-MPa	3.49	1.97	4.00	2.30	2.02	1.13	1.22	0.86	0.89

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390 **Table 2.** Plantar fascia tension and strain with different cartilage stiffness.

Plantar fascia	Tension (N)				Strain (%)			
	1-MPa	5-MPa	10-MPa	15-MPa	1-MPa	5-MPa	10-MPa	15-MPa
1 <sup>st</sup> ray	8.83	9.07	9.03	9.00	0.043	0.044	0.044	0.044
2 <sup>nd</sup> ray	4.71	4.58	4.58	4.63	0.023	0.022	0.022	0.023
3 <sup>rd</sup> ray	3.67	4.06	4.13	4.18	0.018	0.020	0.020	0.020
4 <sup>th</sup> ray	9.37	8.83	8.22	7.73	0.046	0.043	0.040	0.038
5 <sup>th</sup> ray	3.06	2.61	2.73	2.84	0.015	0.013	0.013	0.014

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394 **Table 3.** Material properties and mesh element types for the foot model components.

Component	Element Type	Young's Modulus <i>E</i> (MPa)	Poisson's Ratio $\nu$	Cross-section Area (mm <sup>2</sup> )
Bone	Quadratic Tetrahedron	7300	0.3	-
Cartilage	Quadratic Tetrahedron	1, 5, 10, 15	0.4	-
Ligaments	2-node linear 3-D spar	260	0.4	18.4
Plantar fascia	2-node linear 3D spar	350	0.4	58.6
Plate	Quadratic hexahedron	17000	0.1	-
Soft tissue	Quadratic Tetrahedron	-	-	-

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398 **Figure 1.** Three-dimensional finite element model and application of boundary and loading conditions.  
399 GRF: ground reaction force.

400 **Figure 2.** Comparison of peak pressure (a) and anterior displacement (b) of center of pressure between  
401 measurement and predictions of different Achilles tendon forces, with Young's modulus of cartilage,  
402  $E=1\text{MPa}$ .

403 **Figure 3.** Comparison of plantar pressure distribution between measurement (a) and prediction (b),  
404 with Young's modulus of cartilage,  $E=1\text{MPa}$ , and 50% of normal Achilles tendon force.

405 **Figure 4.** Definition of location (a) and comparison of 2-dimensional displacement (b) of center of  
406 pressure (COP) between different Young's modulus of cartilage.

407 **Figure 5.** von Mises stress in five metatarsals, with Young's modulus of cartilage, (a)  $E=1\text{MPa}$ , (b)  
408  $E=5\text{MPa}$ , (c)  $E=10\text{MPa}$ , (d)  $E=15\text{MPa}$ . Rectangle indicates location of peak stress in the fifth  
409 metatarsal.

410 **Figure 6.** The effect of cartilage stiffness on the contact pressure of joints at midfoot and forefoot. C-  
411 N: cuneonavicular joint; T-M: tarsometatarsal joint; MTP: metatarsophalangeal joint.

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