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Biomechanics of first ray hypermobility: An investigation on joint force during walking using finite element analysis

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ABSTRACT

Hypermobility of the first ray is suggested to contribute to hallux valgus. The investigation of first ray hypermobility focused on the mobility and range of motion that based on manual examination. The load transfer mechanism of the first ray is important to understand the development and pathomechanism of hallux valgus. In this study, we investigated the immediate effect of the joint hypermobility on the metatarsocuneiform and metatarsophalangeal joint loading through a reduction of the stiffness of the foot ligaments.

A three-dimensional foot model was constructed from a female aged 28 via MRI. All foot and ankle bones, including two sesamoids and the encapsulated bulk tissue were modeled as 3D solid parts, linking with ligaments of shell elements and muscles connectors. The stance phase of walking was simulated by the boundary and loading conditions obtained from gait analysis of the same subject.

Compared with the normal foot, the hypermobile foot had higher resultant metatarsocuneiform and metatarsophalangeal joint forces. The increases accounted for 18.6% and 3.9% body weight. There was also an abrupt change of metatarsocuneiform joint force in the medial–lateral direction. The predicted results represented possible risk of joint problems and metatarsus primus varus.

1. Introduction

The first ray is important in stability and maintains structural integrity of the foot [1]. Anatomically, the first ray conjuncts the transverse and the medial longitudinal arches. Some literatures resembled these arches to the beam and truss in architecture to explicate their mechanism on stability [2]. In the biomechanics of walking gait, the first ray exhibits dichotomous action in shock dissipation during heel impact and stabilization during propulsion [3,4]. The two sesamoid bones under the first metatarsal further enhance the first ray's capability in load-bearing and improve the mechanical advantage of the associated muscles [3].

The first ray failure could lead to foot disorders [5]. The concept of first ray insufficiency was firstly introduced by Morton [6]. Lapidus [7] further suggested the association between the first ray hypermobility and hallux valgus and advocated arthrodesis in treating the deformity. Although assessments of first ray hypermobility have become one of the common procedures in the physical examination of the foot, debates and controversies have not stopped [8–11]. Pathogenesis and influences of first ray hypermobility were also investigated [12,13].

Research on first ray hypermobility focused on the mobility and range of motion of the first ray. The normal dorsoplantar range of first ray was examined manually [14–16] or with mechanical devices [15,17,18]. Advanced technology enabled dynamic measurement of the stability and motion of the first ray [8,19]. On the other hand, the orientation of the first metatarsal axis and subtalar joint pronation were also believed to be associated with the pathomechanics of first ray and hallux valgus [18,20,21]. Plantar pressure study has been used to investigate the mechanism of first ray hypermobility and associated deformities [22,23]. Excessive dorsal excursion of the first ray impairs its load-carrying function, the subsequent lateralization of forefoot pressure could manifest metatarsalgia [8,22,23].

Besides range of motion and plantar pressure measurements, load transfer characteristics could be important information for the understanding of the biomechanical changes of hypermobility and the subsequent deformities. Computer simulation provides a versatile platform in assessing the internal features of the geometrically complicated foot structure. In this study, finite element models of a normal foot and a foot with hypermobile first ray were constructed and simulated. We hypothesized that first

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ray hypermobility would alter the metatarsocuneiform (MC) and metatarsophalangeal (MTP) joint load during gait.

2. Methods

The geometry of the right foot was acquired from a healthy female via coronal magnetic resonance (MR) images (3.0 T Seimens, Eriangen, Germany). The MR images were scanned with 1-mm interval at 0.625-mm resolution. The participant was aged 28, 165-cm tall and weighed 54-kg, with no reported musculoskeletal pathology and pain. A custom made ankle–foot–orthosis was applied to the tested limb to keep the foot and ankle in the predefined neutral position [24]. The study and the protocol were approved by The Human Subjects Ethics Sub-committee of The Hong Kong Polytechnic University (HSEARS20130911001).

The MR images were segmented and processed in the software, Mimics v10 (Materialise, Leuven, Belgium) and Rapidform XOR2 (3D Systems Korea Inc., Seoul, Korea). Three-dimensional solid parts were constructed on 30 bones, including two sesamoid bones under the first metatarsal, and the encapsulated bulk tissue. Two-dimensional shell surfaces were constructed on ligaments based on the constructed geometry, the MR images, an anatomy atlas [25] and confirmed with orthopedic surgeon. Muscles and plantar fascia were modeled as one-dimensional connector. Theses muscles included tibialis anterior, extensor hallucis longus, extensor digitorum longus, tibialis posterior, flexor hallucis longus, flexor digitorum longus, peroneus longus, peroneus brevis and triceps surae (Achilles tendon).

The bone was assumed isotropic and homogeneous, with assigned elastic modulus of 7300 MPa and Poisson's ratio of 0.3 [26]. The encapsulated bulk tissue was assigned with hyperelastic properties experimented by Lemmon et al. [27]. The supporting ground included two layers. The bottom layer was defined as rigid, while the upper layer was assigned with 40 GPa, representing concrete ground. The elastic modulus of all ligaments was assumed with 264.8 MPa, which was the reported average elasticity of rearfoot ligaments [28]. The thickness of ligaments was assigned with 1.5 mm as reported by Milz et al. [29] that the ligaments were normally 1–2 mm thick.

Stance phase was simulated by applying ground reaction force (GRF) and tibial inclination in consecutive quasi-static steps. The input data were originated from the gait experiment of the same participant. Five instants were extracted in the following percentage stance phase and named according to the features of GRF in stance: 0% (heelstrike), 27% (GRF first peak), 45% (GRF valley), 60% (Initial push-off), and 75% (GRF second peak). The percentage maximum voluntary contraction of muscles corresponding to these instants were adopted [30] and multiplied by their maximum force [31] for the calculation of muscle forces at the selected instants, except the Achilles tendon. The magnitude of Achilles tendon force was adopted from Fröberg et al. [32]. Frictionless contact was assumed in all joint facets as the friction between these joints was negligible [33]. The coefficient of friction of ground contact was 0.6 [34]. The constructed geometrical of the bony components were illustrated in Fig. 1, with a predicted MTP dorsiflexion angle of 27.3° at 75% stance. The axial direction was defined as the axis along the tibia and fibula segment. The MC joint force was defined as the force acting by the medial cuneiform on the first metatarsal and the MTP joint force was defined as the force acting by the first metatarsal on the first phalanx.

The hypermobility of the first ray was mimicked by reducing the tensile strength [35] of the deep transverse metatarsal ligament (DTML) between the first and second metatarsals by arbitrarily 90%. The MC and MTP joint force were extracted from the simulation. Plantar pressure study of the same participant was conducted for model validation. The participant was asked to walk at their self-comfortable speed with flat wooden sandals. The plantar pressure data were measured by a dynamic in-sole plantar pressure measuring system (F-Scan, Tekscan, USA).

3. Results

Fig. 2 presents the MC joint force of the normal foot and the foot with hypermobile first ray in anterior–posterior (AP), axial and medio-lateral (ML) directions during stance. The joint forces in all directions were relatively small before the GRF first peak (27% stance). The magnitudes were less than 5 N. The increasing trend of joint force became more prominent after the GRF valley (45% stance). The axial force and AP force continued to increase until the simulated end of stance phase, while the ML force (medial) declined and reached another maximal in the opposite direction (lateral).

The difference of MC joint force between the normal foot and the foot with hypermobile became more apparent starting on the instant of GRF valley (45% stance). The hypermobile first ray had

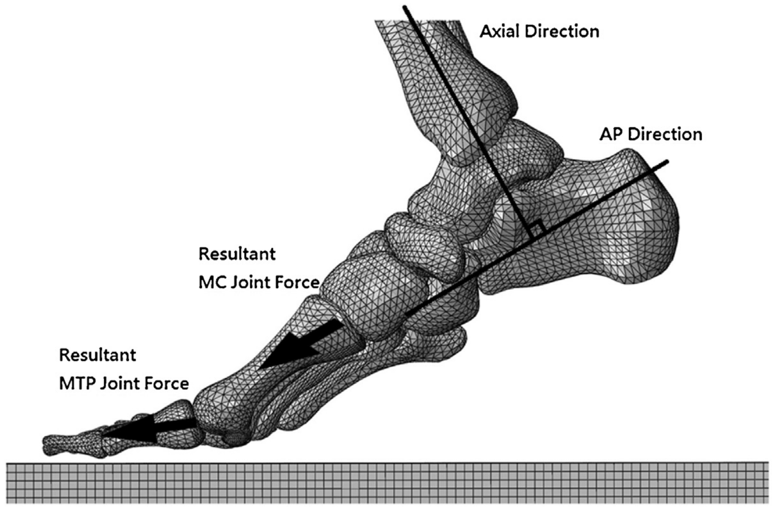

Fig. 1. Illustration of the joint forces and defined axes directions. The axial direction was along the axis of the tibia and fibula and the anterior–posterior (AP) direction was right angle to the axial direction in the sagittal plane. The MC joint force was acting by the medial cuneiform on the first metatarsal. The MTP joint force was acting by the first metatarsal on the first phalanx.

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