INTRODUCTION

The gastrocnemius-soleus (G-S) complex associated with Achilles tendon is the most dominant extrinsic plantar flexor, which is essential for skeleton control and foot stability during gait. With the foot stabilized by muscles upon heel rise, substantial walking ground reaction forces (GRF) was solely imposed onto the forefoot, and, thereby, producing highly localized mechanical stresses underneath the metatarsal heads (MTHs) and underlying protective soft-tissue fat pad. Such ‘detrimental’ stresses, however, may only last for a short-dwelling time in normal foot, on condition that the coordinative muscles activate the right amount and at the exact right time during gait [1].

The altered muscle activity (e.g., equinus contracture due to forceful muscle contraction) may affect the normal foot mechanism and cause a premature heel-rise, and lead to prolonged and high-magnitude forefoot loading exposure throughout the whole stance phase of walking. This paper introduces a new three-dimensional finite element (FE) foot model, in which activations of major extrinsic plantar flexor muscles, including G-S complex, is taken into account. The effects of muscle force variations in G-S complex (i.e. Achilles tendon force (ATF)) on joint movements at ankle and metatarsophalangeal (MTP) joints, which produces necessary foot flexibility during the period of mid-stance to toe-off. The three-dimensional structures of foot skeleton, plantar fascia and Achilles tendon were then encapsulated into a homogenous mass of foot soft tissue, with interfascial elements sharing same nodes except at those joint space regions where contact conditions were defined.

METHODS

Finite element model: The model was refined from our previously developed soft-tissue-skeletal foot model to accommodate the muscle attachments [2]. The complete element mesh of the model with musculature simulated in the current study is shown in Fig 1, and it consists of 30 bones with joint cartilages, 134 ligaments (multiple element representation for major ligament bundles), and a fan-shaped plantar fascia structure.

Unidirectional cable elements were used to simulate the ligaments permitted these elements to resist tension producing forces. Six major extrinsic plantar flexors were considered. To model the gastrocnemius-soleus (G-S) complex, three-dimensional geometry of the Achilles tendon was reconstructed, and inserted into the posterior extreme of the calcaneus to mimic tendon-bone junction, which facilitates realistic application of the G-S muscle forces via the Achilles tendon. The long tendons of other five muscles were attached to the model using bar elements based on straight-line approximation, i.e. several bar elements stringed together to represent the actual tendon trajectory inside the foot. Surface-to-surface contact elements with no friction properly simulated the relative articulating movements between joint cartilages, as they allowed the bones to slide over one another with the only constraint governed by the congruent facets. These were located at the ankle joint, subtalar joint, and metatarsophalangeal (MTP) joints, which produces necessary foot flexibility during the period of mid-stance to toe-off. The three-dimensional structures of foot skeleton, plantar fascia and Achilles tendon were then encapsulated into a homogenous mass of foot soft tissue, with interfascial elements sharing same nodes except at those joint space regions where contact conditions were defined.

Figure 1: Cut through the element mesh of the finite element model of the soft-tissue-skeletal foot-ankle complex with internal ligamentous and muscular structures.

The material properties of different tissues of the finite element model, including cortical and cancellous bones, ligaments, and cartilages, are similar to those used in our previous study [2]. For Achilles tendon and other flexor tendons, they were represented as isotropic linear elastic materials. Young’s modulus (E) and the Poisson ratio (υ) for the Achilles tendon were defined as 816 MPa, and 0.3, respectively. The remaining flexor tendons were assigned a Young’s modulus of 450 MPa, a Poisson ratio of 0.3 and a cross-section area of 12.5mm². The plantar soft-tissue was modeled using an isotropic, nearly incompressible, hyperelastic Ogden formulation.

Simulated musculoskeletal loading: As the areas under the MTHs are of particularly clinical interest, the instant when forefoot force reached a peak (probably the sub-MTH peak pressure) was chosen for simulation. At the start of the simulation, forefoot kinematic data were used to assist in replicating foot pose relative to the ground. Specifically, the second metatarsal shaft has been oriented approximately at 25 degrees with the horizontal to reflect mean forefoot orientation in the sagittal plane at push-off. The tibia, fibula and the superior surfaces of the soft tissue were fully constrained. The applications of extrinsic muscle forces were simulated in the finite element model by force vectors in axial alignment with the tendons attached. The analysis started with a baseline model: a model driven by musculoskeletal loading. A maximum ground reaction force (GRF) approximating the second peak in walking GRF was produced solely by simulated contraction of the plantar flexors at the prescribed metatarsal shaft configuration. This loading protocol
mimicked the manipulations of the cadaver study by Sharkey [3] with the simulated activity of the flexor tendons with use of linear actuators. The initial orientation of metatarsals was slightly re-arranged due to the soft tissue’s large deformation, and movements of the bony joints. The model solution was finally converged with its shaft oriented at 26.8 degrees. The required muscle forces were computed. Their force magnitudes are directly in proportion to their physiological cross-sectional areas, resulting in a contractile stress of 0.2 MPa in all plantar flexor muscles. The model was solved in the general purpose FE analysis package ABAQUS (Hibbitt, Karlsson & Sorensen, Inc., RI).

**Verification of the analytic results:** Plantar pressures under the MTHs and phalanges of five foot rays were collected at 80 Hz with special resolution of 4 sensors per cm² using Pliance-X® pressure-sensitive array. Five trials were obtained when the subject taking their first step with an instrumented insole material with flexible sensors located underneath each bony prominence. Good agreement between analytically and experimentally determined pressure data were found at sub-MTH areas. This was also the case for toe pressures. The model was also internally verified by comparing the predicted metatarsal strain to the cadaveric experimental work from Sharkey et al [3]. Local coordinate systems were defined for the metatarsal elements to obtain axial surface strain components directly from finite element analyses.

**Parametric study on muscle force variations:** To study the effects of muscle force variations in gastrocnemius-soleus complex (i.e. ATF) on foot mechanism, a multi-step analysis procedure was adopted. With the maximum GRF maintained at the ground support, the loaded baseline model served as the starting condition for the subsequent analysis in which the full ATF (i.e. 100%) were adjusted in a step-wise manner with a 40% gradual decrease in a total of four steps (i.e. 90%, 80%, 70% and 60%) to simulate reduced muscle effectiveness at G-S complex. The state (stresses and strains) of the loaded model was updated throughout general analysis steps (i.e. the ending condition for the current step is the starting condition for the subsequent step), in response to incremental changes in muscle loads of the G-S complex. All data was reported in comparison with those of the baseline model.

**RESULTS AND DISCUSSION**

With fully muscular loads applied, the baseline foot model was solved in a typical geometry at push-off, with the ankle and metatarsophalangeal joints maintained at plantar-flexed and extended configurations, respectively. Subsequently parametrical analysis showed that the joint angles predicted by the model were significantly affected as a consequence of induced variations (i.e. reduction) in ATF. In response to the decreased muscle effectiveness, the ankle joint gradually moved by 8.8° from plantarflexion to dorsiflexion. This was coupled with metatarsophalangeal joint movements in extension.

The effect of decreased ATF on forefoot stress distributions was marginal (Fig. 2). For plantar pressure prediction, the highest pressure locations were shifted from the 2nd MTH to the 1st, 3rd, and 4th MTHs. This was accompanied by an extension of the contact area from the metatarsal region to its distal side (i.e. close to mid-foot region), and resulted in overall peak plantar pressure decrease by 22.7%. Similarly, in the metatarsal bones, VMS was initially peaked at the plantar aspect of the 2nd metatarsal shaft (19.9 MPa), and then shifted to the 3rd (15.8 MPa) and 4th metatarsal bones (13.5 MPa).

![Figure 2: Plantar pressure redistributions in response to muscle force variations in G-S complex](Image)

**CONCLUSIONS**

In summary, a musculoskeletal finite element foot model is presented, to investigate the mechanisms of how the gastrocnemius-soleus complex contributes to the weight-bearing function of the foot. The results obtained may have important clinical implications on surgical Tendon-Achilles Lengthening (TAL), i.e. a part of the clinical procedure to correct equinus deformity and treat certain foot pathologies associated with contracture of the G-S complex, such as plantar forefoot ulcerations [4]. It was concluded that reduced muscle strength in G-S complex increase the mobility of ankle joint, decreased the overall peak plantar pressure, and internal VMS stresses in metatarsal bones. These were mainly achieved by a more equal load sharing among the five foot rays.

**ACKNOWLEDGEMENTS**

This work was supported by a Grant from the Temasek Defence Systems Institute (Project no.: TDSI/09-009/1A), Singapore.

**REFERENCES**