A COMPUTATIONAL BIOMECHANICS STUDY TO INVESTIGATE THE EFFECT OF MYOELECTRIC STIMULATION ON PERONEAL MUSCLES IN PREVENTING INVERSION-TYPE ANKLE LIGAMENTOUS SPRAIN INJURY

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A three-dimensional multi-body lower limb model with 16 bones and 22 ligaments was developed to study ankle ligamentous inversion sprain. A male athlete who was diagnosed with a grade I anterior talofibular ligament (ATaFL) sprain during an accidental injury in laboratory in a published report. His ankle kinematics injury data profile was computed. The effect of delivering myoelectric stimulation on peroneal muscles was simulated as torques during ankle inversion. Largest strain in the ATaFL was 8.3%, 9.0% and 11.4%, respectively, at different inversion velocity thresholds of 300 deg/s, 400 deg/s and 500 deg/s. A ligament strain/sprain more than 10-15% would lead to a ligament tear suggesting that applied muscle moments could successfully prevent ankle inversion sprain when an injury identification threshold does not reach 400 deg/s.

KEY WORDS: ankle supination sprain, injury biomechanics, computational biomechanics, computational modeling.

INTRODUCTION: Ankle inversion ligamentous sprain is one of the most common injuries in sports (Fong et al., 2007). It accounts for approximately 85% of ankle injuries. The suggested injury mechanism is excessive ankle supination with inversion and plantarflexion. To our knowledge, the first biomechanics quantitative investigation on a lab accidental ankle supination injury was reported (Fong et al., 2009). Besides inversion and plantarflexion, the report revealed that internal tibial rotation also occurs during the injury.
A patient was diagnosed with a grade I mild anterior talofibular ligamentous ankle sprain injury by an orthopaedics specialists. During the reported accident, high-speed videos were captured by three video cameras and kinematics data were presented afterwards. Information on the ankle ligament strain during injury may help better understanding the injury mechanism. Computational models of musculoskeletal joints and limbs provide useful information about joint mechanics, while validated models can be used to predict ligament strains in order to understand normal joint function and to serve as clinical tools to predict and help preventing sports injuries (Liacouras & Wayne, 2007). The purpose of this study was to develop a three-dimensional multi-body lower limb model, and to evaluate the efficacy of myoelectric stimulation in preventing ankle inversion sprain by using this model. Scenarios simulations of ankle inversion twisting motion from the previous sub-injury laboratory trials were conducted to observe the strains in the anterior talofibular ligament (ATaFL).

METHODS: The study was approved by the University Clinical Research Ethics Committee. The procedures included three parts: model development, model validation and model simulation.

I. Model development

One male subject (age, 29 years; height, 1.75m; body mass, 62.6 kg) who had accidentally sprained his right ankle during the previous laboratory test (Fong, et al., 2009) was recruited to participate in this study. A computer tomography (CT) scan was carried out. The scanning provided detailed three-dimensional (3D) anatomical features of 16 bones, namely, half of the femur (distal), tibia, fibula, talus, calcaneus, cuboid, navicular, medial cuneiform, intermediate cuneiform, lateral cuneiform, first metatarsal, second metatarsal, third metatarsal, fourth metatarsal, fifth metatarsal and phalanges in a whole. The images were transferred from Digital Imaging and Communications in Medicine (DICOM) files into Materialise's Interactive Medical Imaging Control System (MIMICS) (Materialise, Ann Arbor, MI). This software yielded 3D surface model of each bone as Stereolithography (STL) files for export. To reduce the size of surface files and subsequent models, the STL files were remeshed in MIMICS to smooth the surface of each bone. The remeshed STL files were then imported in a 3D solid modeling program (SolidWorks, TriMech Solutions, LLC, Columbia, MD). In SolidWorks, the ScanTo3D package was used to reconstruct each bone and simplify the bone surfaces. The individual bones were then assembled into a lower extremity and registered with respect to one another in the anatomical position. Since toe involvement is minimal in this study, the phalanges were excluded in the model. All other joints of the foot and ankle were constrained only by joint geometry and ligament stability. Ligaments were represented as linear, elastic spring elements, with stiffness values from the literature. The origin and insertion locations of the ligaments were determined from dissection and anatomical atlases. Ligament preloads were induced by reducing the lengths by 2%. Each bone was allowed to move in all six degrees of freedom, leaving body motion as a function of ligament behavior, surface contact and external constraints. During simulation, tibia was fixed in all simulation scenarios, while fibula, talus, and calcaneus were allowed to move freely in all six degrees of freedom. Tarsal and metatarsal bones were fused by 'lock' mates and moved as a unit for simplification purposes, leaving body motion to be a function of ligament behavior, surface contact, and external constraints (Wei et al., 2011). 3D contacts were implemented between adjacent bones to inhibit intersection during the simulation (Iaquinto and Wayne, 2010).

II. Model validation

The model was validated against two documented cadaveric studies (Colville et al., 1990; Wei et al., 2010). Colville and colleagues conducted an experimental cadaver study by investigating the strains on ankle ligaments while moving the ankle joint from 20 degrees dorsiflexion to 30 degrees of plantarflexion and stress in different directions were applied. Subtalar joint motion was removed in this cadaver study by placing two screws through the calcaneus into the talus. In this simulation, tibia was fixed while fibular was moved freely. Tarsals and metatarsals were fused to move as a unit. Continuous dorsiflexion and plantarflexion was applied to the ankle, 3Nm of internal or external rotation torque was then applied to the talus for different dorsi-plantarflexion angles. Subsequently, ligament strains of anterior tibiofibular ligament (ATiFL) and posterior tibiofibular ligament (PTiFL) were calculated in the model and compared to the cadaver data. The second validation was aimed to examine ankle injury caused by excessive external ankle rotation (Wei et al., 2010). Subtalar joint motion was allowed in this experiment by applying a relatively rigid foot constraint. External ankle rotation was progressively increased by increments of 5 degrees in the testing until ligament injury or bone fracture occurred. Failure ankle rotation and failure torque were documented. Talus was set to be free in this experiment in order to fully imitate situation. The ankle was initially positioned at 20 degrees dorsiflexion and 10 degrees of eversion. Compressive load of 2000N was applied to the proximal end of the model with one-sixth of loading to fibula. 40 degrees of external ankle rotation was input according to the average failure rotation degree recorded in the cadaver study (Wei et al., 2010).

III. Model simulation

By using the validated model, 3-fold body weight of the injured athlete (Fong et al., 2009) was input to simulate weight-bearing load because of the unavailability of the ground reaction, and it was an average value during the heel impact in landing and ground contact (Cavanagh and Lafortune, 1980). 1530N and 310N vertical load was carried by tibia and fibula respectively (Lambert, 1971; Wang et al., 1996). In the simulation, a simulated myoelectric stimulation at a delay time of 25ms (Ginz et al., 2004) was given when a risky ankle inversion velocity reached 300 degrees/s. This velocity was identified as the threshold when the anti-sprain system activated (Chan et al., 2006) and provides stimulation to the peroneal muscles, ankle evertors (Fong, 2012).

RESULTS:

I. Model validation

In the first experimental cadaver study (Colville el al., 1990), ligament strains in the ATiFL and PTiFL at different dorsi-plantarflexion angles were recorded either with or without an additional rotational torque. In this 3D computational model, strain in the ATiFL increased 0.6% when the ankle was moved from the neutral position to 20 degrees of dorsiflexion. In addition, strain in this ligament was increased by applying external toque, and decreased by applying internal torque on the talus at all joint angles. This finding was consistent with the cadaver study. Meanwhile, the PTiFL experienced a maximal elongation of 0.4% when the ankle was moved from the neutral position to 20 degrees of dorsiflexion. Besides, external torque on the talus decreased strain, and internal torque on the talus increased strain. The trends were consistent with previous study.

In the second validation test, the results indicated that there was a similar rotation-torque relationship during external ankle rotation in the model when comparing with the cadaver study by Wei and colleagues (2010). In the cadaver study, an average failure torque of 69.5Nm and an average failure external rotation of 40.7 degrees were documented. While in the computational model, a close maximum torque of 60.3Nm at 40 degrees of external rotation was found for the average ligament stiffness taken in the literature.

II. Computational biomechanics simulation

The initial length of ATiFL in situ level was about 28.8mm. After a series of simulations, the new length at the end angle of different calculated range of inversion was obtained in the spring of the model. When the injury identification threshold reached 300 deg/s, the ATaFL experienced a maximum strain of 8.3% under the influence of simulated myoelectric stimulation. Similarly, once the threshold of 400 deg/s and 500 deg/s was identified, the maximum strain of 9.0% and 11.4% was produced respectively. Therefore, the higher the injury identification threshold identified, the larger the maximum strain produced. A successful simulation scenario should avoid the ligament approaching 10-15% elongation (Beumer et al., 2003; Funk et al., 2000; Yahia et al., 1990). Applying stimulation with a delaty time of 25ms could successfully prevent the ankle inversion sprain once lower threshold of 300 deg/s or 400 deg/s was identified, whereas it failed once when the threshold identified at 500 deg/s.

Table 1

The parameters for calculating maximum strain in the ATaFL and the results

New range of inversion (Degrees)	Length (mm)	Elongation (%)
31.7	31.2	8.3
34	31.4	9.0
36.5	32.1	11.4

DISCUSSION: This study has developed a 3D multi-body lower limb model whereby the kinematics for joint motion were dictated by 3D articular contacts, ligamentous restraints, force loading, and external perturbations. The model was validated against two experimental studies, and subsequently used to evaluate the effect of myoelectric stimulation in preventing ankle inversion sprain.

The model has several limitations needed to be improved in the future. The ligaments were simplified as linear tension-only spring but in reality some of them are nonlinear or viscoelastic (Spratley and Wayne, 2011; Wei et al., 2011). Ground reaction force data was neglected in which the magnitude and duration of the load were unknown in this study.

CONCLUSION: The current study developed a good 3D computational lower-limb model and myoelectric stimulation was demonstrated to be effective in preventing ankle inversion sprain under the condition that a lower inversion velocity threshold could be identified. This study would contribute to the research on the intelligent anti-sprain system.

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