Biomechanical Consequences of Foot and Ankle Injury and Deformity: Kinematics and Muscle Function

By

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Biomechanical consequences of foot and ankle injury and deformity: kinematics and muscle function

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ABSTRACT

The overall aim of this thesis was to discuss kinematics and muscle function changes due to foot and ankle injury or deformity. The first study aims to characterize gait patterns of subjects with a common lower limb injury, ankle fractures. Using three-dimensional movement analysis with a modified multi-segment foot model, the inter-segment foot kinematics was determined during gait in 18 subjects one year after surgically treated ankle fractures. Gait data were compared to an age- and gender-matched control group and the correlations between functional ankle score and gait parameters were determined. It was observed that even with fairly good clinical results, restricted range of motion at and around the injured area, and less adducted forefoot were found in the injured limb. The second study aims to quantify the effect of subtalar inversion/eversion on the dynamic function of the main ankle dorsiflexors/plantarflexors: gastrocnemius, soleus and tibialis anterior. Induced acceleration analysis was used to compute muscle-induced joint angular and body center of mass accelerations. A three-dimensional subject specific linkage model was configured by gait data and driven by 1 Newton of individual muscle force. The excessive subtalar inversion or eversion was modified by offsetting up to ±20° from the normal subtalar angle while other configurations remain unaltered. We confirmed that in the normal gait, muscles generally acted as their anatomical definitions and muscles can create motion in joints, even not spanned by the muscles. The plantarflexors play important roles in body support and forward progression. Excessive subtalar eversion had negative effect on ankle plantarflexion, which may induce a less plantarflexed ankle, less extended knee and more flexed hip after initial contact. This thesis focused on gait kinematics and muscle functions in the foot and ankle area employing both experimental gait and computational simulations. The findings can be regarded as references for evaluating of future patients and for dynamic muscle functions during gait.
PREFACE

This thesis is based on the following publications that will be referred to by their roman numerals:

I Ruoli Wang, Charlotte K.Thur, Elena M. Gutierrez-Farewik, Per Wretenberg, Eva Broström

One year follow-up after operative ankle fractures: a prospective gait analysis study with multi-segment foot model. Submitted.

II Ruoli Wang, Elena M. Gutierrez-Farewik

The effect of subtalar inversion/eversion on the dynamic function of the tibialis anterior, soleus, and gastrocnemius during the stance phase. Submitted
Division of work between authors

The research project was initiated by Dr. Elena Gutierrez-Farewik (EGF) and Dr. Eva Broström (EB), where EGF was the main supervisor and co-author in Paper I and II. EB acted as co-supervisor and was advisor of the work resulting in Paper I. Dr. Charlotte K. Thur (CKT) and Dr. Per Wretenberg (PW) were clinical advisors and co-authors in Paper I. Ruoli Wang (RW) continuously discussed the progress throughout the work with EGF and EB.

Paper I

The experimental data was collected by EB and RW. The data processing and statistical analysis were done by RW. The 90% of the paper was written by RW and 10% by CKT with input from EB, EGF and PW.

Paper II

The simulations were done by RW and the experimental data was collected by RW and EGF. The paper was written by RW with input from EGF.
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<td>MTP</td>
<td>Metatarsophalangeal</td>
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<tr>
<td>RMS</td>
<td>Root mean square</td>
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<td>EMG</td>
<td>Electromyography</td>
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<td>IC</td>
<td>Initial contact</td>
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<td>GRF</td>
<td>Ground reaction force</td>
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<td>IAA</td>
<td>Induced acceleration analysis</td>
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<td>OMAS</td>
<td>Olerud/Molander ankle score</td>
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<td>COM</td>
<td>Center of mass</td>
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<tr>
<td>ORIF</td>
<td>Open reduction internal fixation</td>
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<td>3D</td>
<td>Three-dimensions/dimensional</td>
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<td>ANOVA</td>
<td>Analysis of Variance</td>
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INTRODUCTION

The study of human motion can be traced back thousands of years ago[1]. Gait mechanics of foot temporal and stride parameters was documented as early as 1836 by the Weber brothers[2]. Foot motion study has been for over a century by captivated clinicians and researchers[1].

The human foot, the only part of the body that acts on an external surface in upright, unsupported positions, supports and balances the body during gait. With muscle coordination, the foot can be compliant to cope with uneven ground surface to achieve a smooth motion and maintain dynamic stability. Ankle injuries, foot pain and muscle dysfunctions are common and stem from the large impact forces and rotational moments during weight-bearing activities[3]. As the distal end of the lower extremity, its position or movement can influence the position, movement or loading at the knee or hip of either limb[4]. This thesis includes two parts: Study I is the experimental gait analysis of the foot kinematics in patients with ankle fractures. Study II is the muscle-driven dynamic simulation studying the influence of abnormal foot kinematics on the individual muscle functions during walking.

RELATED FUNCTIONAL ANATOMY OF THE ANKLE AND FOOT

The foot and ankle make up a complex anatomical structure consisting of 26 irregularly shaped bones, 30 synovial joints, and more than 100 ligaments, tendons, and muscles acting on the segments[4]. The foot is considered to have four subdivisions: the hindfoot, midfoot, forefoot, and the phalanges (Fig 1). Other than the talocrural joint (ankle), most of the motion in walking occurs at three of the synovial joints: the subtalar, midtarsal, and metatarsophalangeal joints (MTP)[5].

Talocrural joint

The ankle or talocrural joint is comprised of 3 bones: tibia, fibula and talus (Fig 2). The articulations of this joint complex are between the dome of the talus and the tibia plafond, medial facet of the talus and the medial malleolus, and lateral facet of the talus and the lateral malleolus respectively. Although the ankle joint was considered as a hinged synovial joint allowing only dorsiflexion/plantarflexion movement, the anatomical axis of the joint has been demonstrated horizontal and oblique to the frontal plane of the foot due to outward rotation of the lower end of the tibia[7]. Moreover, movement of the foot at the ankle joint is rarely alone; it is invariably combined with motion about the subtalar and midtarsal joints[8]. The lateral and deltoid ligaments have important roles in maintaining stability in the articular motions.
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**Figure 1:** Bones in the foot, modified from Abboud[6]

**Figure 2:** The joints in the foot (subtalar, midtarsal, MTP) with major functional significance during walking, modified from Perry[5]
**Subtalar joint**

The subtalar joint is situated between the talus and calcaneus (Fig 2). With the ankle joint, the oblique orientation of the subtalar joint axis (from the posterior lateral plantar surface to the anterior dorsal medial surface of the talus, Fig 3) allows the foot to move relative to the tibia in a complex manner[9], which is usually defined as pronation and supination. The prime function of the subtalar joint is to absorb the rotation of the lower extremity during the support phase of gait[4].

**Midtarsal and metatarsophalangeal joints**

The midtarsal joint is the junction of the hindfoot and forefoot which contributes to the shock absorption of forefoot contact (Fig 2). The MTP joint is the toe break, which allows the foot to roll over the metatarsal heads rather than the tips of the toes. The five metatarsal heads provide a broad area of support across the forefoot[5].

![Figure 3: The axis of the subtalar joint, modified from Hamill and Knutzen[4]](image)

**Ankle dorsiflexor and plantarflexor**

Twenty-three muscles act on the ankle and the foot, and play important roles in sustaining impacts of very high magnitude, and in generating and absorbing energy during movement[4]. Ankle plantarflexors refers to the muscles which can extend the ankle resulting in the forefoot moving away from the body, while ankle dorsiflexors can flex the ankle resulting in the forefoot moving toward the body. The gastrocnemius, together with the soleus, are the chief plantarflexors of the ankle joint. The gastrocnemius spans the knee joint, so it is also a powerful flexor of that joint. The other plantarflexor muscles produce only 7% of the remaining plantarflexor force[10]. The most medial dorsiflexor is the tibialis anterior, whose tendon is farthest from the joint, thus giving it a significant mechanical advantage as a powerful dorsiflexor[10]. Previous studies reported that the gastrocnemius, soleus and tibialis anterior also have inversion leverage of the subtalar joint[9,4].
Contemporary gait analyses primarily focus on the measurement of joint kinematics and kinetics, electromyography (EMG), oxygen consumption and foot plantar pressures. Gait analysis used in this thesis involves markers placed on specific anatomic landmarks. The markers are covered in a retro-reflective material which can reflect the light from infrared cameras to sensors mounted on the camera. The marker positions are used to describe the three-dimensional positions and movements of body segments and joints. The assumption of this method is that the surface-mounted markers reflect the motion of the underlying bones or structures. Measurement errors introduced with soft tissue deformations have been estimated in studies comparing surface-mounted marker movement to intracortical pin-mounted markers. The least error has been reported in the sagittal plane and the larger error in the frontal and transverse planes of the knee motion[11]. Westblad et al.[12] reported skin movement artifact for movement of the calcaneus relative to the tibia during stance phase, where root mean square (RMS) was small at 2.5° (inversion/eversion), 1.7° (plantarflexion/dorsiflexion) and 2.8° (adduction/abduction). Nest et al.[13] compared kinematic data from a four-segment foot model to the kinematics of the foot bones comprising four segments. They found differences were greatest for motion of the combined navicular/cuboid relative to calcaneus and the medial forefoot segment relative to the navicular/cuboid. RMS error of commercially-available capture systems in calculating the distance of two markers in a volume with a length of 2.0-4.6 m was reported between 0.6 mm and 1.7 mm[14]. Dynamic motion capture with more cameras resulted in higher error, and error in calculating a known angle between markers on a rotating plate were between 1.4° and 4.2°[15].

Normal gait can be divided into stance and swing phases. The stance phase is approximately the first 60% of the gait cycle and starts with initial contact (IC) when foot just touches the floor. Loading response (0-10% of gait cycle) is the initial double stance which ends when the contralateral foot is lifted for swing[5]. Following loading response is mid-stance (10-30% of gait cycle) and terminal stance (30-50% of gait cycle), which are the single-limb support interval. The final phase (50-60% of gait cycle) of stance is the pre-swing, the second double support in the stance phase.
BIOMECHANICS OF THE FOOT AND ANKLE IN NORMAL GAIT

Foot motion definition

Although several specialized movement names are assigned to the foot movement, they are still generally regarded to three basic planes (sagittal, frontal and transverse planes). Plantarflexion is the movement when the distal aspect of the foot is angled downwards in the sagittal plane away from the tibia, and dorsiflexion is the movement when the distal aspect is angled towards the tibia in the sagittal plane. Hindfoot inversion takes place in the frontal plane when the medial border of the foot lifts so that the sole of the foot faces medially towards the other foot. Hindfoot eversion is the opposite movement of the hindfoot. Forefoot adduction is the movement when the distal aspect of the forefoot is angled towards the midline of the body in the transverse plane. Forefoot abduction is the movement when the distal aspect is angled away from the midline of the body. In orthopedics, a varus deformity is a term for the inward angulation of the distal segment of a bone or joint. The opposite of varus is called valgus. Common confusion exists over the use of the terms inversion and eversion with pronation and supination. Foot pronation consists of a combination of ankle dorsiflexion, calcaneal evasion, and forefoot abduction. Foot supination is the opposite of pronation, with ankle plantarflexion, calcaneal inversion, and forefoot adduction[4].

Figure 5: Foot motion definition: (A) plantarflexion-dorsiflexion (B) inversion-eversion (C) forefoot adduction-abduction (D) supination-pronation[4,6]
The foot's movement during the stance phase

At IC, the ankle is almost neutral or slightly plantarflexed and the subtalar joint is inverted. In a short period afterwards, the foot is passively plantarflexed in a smooth, regulated manner such that the ankle joint plantarflexion is stopped synchronously with the forefoot making contact with the ground[16]. During the loading-response, only the lateral side of the foot makes contact with the ground so to transfer weight to the forefoot. The effect of the ground reaction force (GRF) on the lateral side of the forefoot tends to evert the forefoot[17]. The ankle changes its direction towards dorsiflexion after foot-flat and the tibia becomes the moving segment. Ankle dorsiflexion continues throughout mid-stance and reaches its maximum in terminal stance. At the same time, the forefoot gradually moves towards inversion. The subtalar joint slowly reverses eversion toward inversion throughout the terminal stance, particularly during toe-rise and reaches its peak in pre-swing[7]. There is a rapid ankle plantarflexion following terminal double support which reaches the maximum at the end of the stance phase[5].

Brief muscle roles at ankle joint in stance

As described by Perry[5], after IC, in response to the large plantarflexion moment generated by the GRF, ankle dorsiflexors decelerate the ankle plantarflexion. This dynamic response also contributes to limb progression, when the tibia actively advances while the foot lowers down. Following the forefoot floor contact, the GRF advances forward along the foot so to create a large external dorsiflexion moment. The soleus and gastrocnemius react eccentrically to restrain the rate of ankle dorsiflexion. In pre-swing, the soleus and gastrocnemius act concentrically during the toe-rise and reduce the intensity of their action since the body weight transfers to the other limb. The tibialis anterior and toe extensors begin to activate at the end of the pre-swing to decelerate the ankle plantarflexion.

Figure 6: Muscle roles and body weight vector in stance phase. Arrow indicates the direction of motion, modified from Perry[5].
A COMMON ANKLE INJURY: ANKLE FRACTURE

Definition and incidence

Ankle fracture in this thesis refers to the malleolar fractures. It is one of most common lower limb fractures, and the frequency has been increasing over the past few decades, especially in elderly women[18,19]. According to previous epidemiological studies, the incidence of ankle fractures is between 107 and 184 per 100,000 persons per year[18,20,21,22]. Another study had shown that this rise has continued during the entire 1980s and 1990s[19]. In the United States, ankle fractures have been reported to occur in as many as 8.3 per 1000 medical-care recipients, a figure that appears to be rising steadily[23].

Classification

In order to describe fractures and help physicians to determine appropriate treatment, two classification schemes based on radiographic presentation, called Danis-Weber and Lauge-Hansen, are widely used. Lauge-Hansen’s classification, first reported in 1950[24], takes the posture of the foot at the moment of injury and the direction of deforming force into consideration, and subsequently divides ankle fractures into five types. While it certainly provides better understanding of injury mechanisms, resulting in improved technique in closed treatment of unstable fractures[25], it is complicated and difficult to apply. Weber’s classification divides ankle fractures into 3 types (A, B, C) on the basis of the anatomy of the fracture of the lateral malleolus[26]. It is easy to use and requires few clinical details, but its weakness of ignoring the biomechanical aspect of the medial injury makes the evaluation of results difficult. Another commonly used classification scheme for ankle fractures is the simple anatomic division into uni-, bi- and trimalleolar fractures. Some authors have advocated modifications to the existing schemes to achieve more biomechanical and clinical relevance[27].

Management

An ankle fracture can be treated with or without surgery followed by immediate mobilization or a period of immobilization. A simple lateral malleolus fracture (Weber A) is reducible and usually well responds nonsurgical with a closed treatment. In bimalleolar fractures and fractures involving deltoid disruption, these fractures are considered unstable and generally require surgical intervention to restore the alignment of the ankle joint[28]. Immobilization can result in decreased range of motion, muscle atrophy and decreased peak muscle torque at the ankle[29]. Prolonged non-weight bearing was recommended for ankle fractures in diabetic patients[30]. Rehabilitation intervention timing and methods after fractures are still being debated. Rehabilitation can begin during immobilization with removable devices, where active and passive exercise[31] or early weight-bearing[32] can conducted. Rehabilitation can also start following the period of immobilization, where interventions may include exercise and manual therapy. Plaster casts, ankle braces, and orthoses are common immobilization tools which may also affect the results of rehabilitation[29].

COMPUTATIONAL METHODS IN MUSCULOSKELETAL COORDINATION

In biomechanics, we can either input the muscle forces to predict the displacement of the body segments or compute joint moments and forces from a combination of measured external forces, segment kinematics, and anthropometric data. The first technique is referred to as a forward dynamics approach, whereas the latter is an inverse dynamics approach.
The inverse dynamics method is commonly employed in clinical gait analysis to compute the net joint moments, and net joint powers[33]. The foot, shank and thigh are considered to be rigid segments connected by joint articulations. The measured ground reaction force and estimated segmental accelerations are inserted into the Newton-Euler equations of motion, starting at the most distal segment (e.g. foot) and solving for the proximal joint force and moments (e.g. ankle)[34]. One limitation of the traditional Newton-Euler inverse dynamics method is its inability to identify the role of individual muscles in coordinating the body segments[34]. In order to understand the individual muscle contributions to the movement, additional methodologies are needed to decompose the net joint moments or joint forces, which can be estimated directly from the inverse dynamics, into individual muscle moment or muscle forces. Static optimization is one method, but is not entirely reliable to study muscle coordination because of the uncertainty in the optimization criterion inherent in this approach[35,36]. EMG activity is often recorded in gait studies, but the relationship with certain muscle force is still debatable[37,38]. Various methods can be used to find muscle or joint moment contributions with forward dynamics. One method is to use the net joint moments computed from traditional inverse dynamics as input to a forward dynamical model[39]. One of the most difficult aspects of generating muscle-driven dynamical simulations compatible with experimental observation is finding an appropriate muscle activation pattern. Optimization theory and a dynamical model to iteratively find the muscle excitations are usually applied[40].

Induced acceleration analysis (IAA) is an approach which lies at the intersection (conceptually) of the field of forward dynamics and inverse dynamics, and which may serve as an enhancement to the conventional inverse dynamical approach. The basis of the analysis is the identification of the instantaneous contribution of a particular muscle (e.g. gastrocnemius) or muscle group (e.g. ankle plantarflexor) to an outcome measurement (e.g. acceleration of the center-of-mass of the body). Zajac et al.[41] first introduced IAA as a tool to demonstrate that the gastrocnemius, anatomically a knee flexor and ankle plantarflexor, in certain circumstances can act as a knee extensor. The mechanical analysis of the whole musculoskeletal system revealed that muscle groups crossing a joint would generally act to accelerate all joints of the body[41]. In recent years, researchers using this approach have expanded our understanding of how individual muscles or muscle groups control body motion, e.g. contribution to the vertical GRF[42] and the energetics of the body segment during the normal gait[43]. Clinical IAA studies have demonstrated that excessive external tibial rotation, a transverse plane misalignment of the lower leg, can reduce the lower limb muscles’ capacity to extend the hip and knee during single-limb stance, which may be a significant contributor to crouch gait[44,45]. IAA analysis of stiff-legged gait studies indicated that variable causes of the stiff-legged gait were highly related to patients’ specific impairments[46]. However, it is noteworthy that IAA is a snapshot in time of contributions of individual forces acting on the body segments without regard to the cumulative effects of past muscle and gravity force trajectories on the system behavior[34].

**Musculoskeletal model**

Whether in forward or inverse dynamics, one has to employ an anatomical model of the musculoskeletal system. In gait analysis using surface-mounted markers, a model is required to infer the position of the body segments from the measurement positions of the markers. In forward simulations, a musculoskeletal model containing accurate three-dimensional (3D) geometry of each muscle is often used to comprehend the dynamic function of the individual muscles.

**Kinematic model in gait analysis**

The most widely used whole-body gait models consist of 15 rigid body segments (head, torso, upper arms, lower arms, hands, pelvis, thighs, shanks, and feet). However, representing the foot as a single rigid body with a revolute ankle joint is inadequate to demonstrate the true 3D foot motion. During the
last few years, many noteworthy biomechanical foot models which include multiple segments have been developed[47,48,49].

There are some consenses in these models. Since the number of segments that can be tracked is limited when using the typical camera configuration for a full body motion analysis, most foot models contain three or four segments and express angular relationship as Euler angles[50,51]. Most models reference their dynamic angles to a standard zero position, where static joint angles are defined to be zero[52,53], which contributed to the reduction of the possible variations from marker placement. However, there are some inconsistencies in segment definitions. For instance, the group from Marquette University defined the forefoot segment with cuneiform, cuboid and metatarsal bones[51]. The Oxford foot model defined forefoot segments rigorously only with metatarsal bones, and the midfoot segment was considered as a mechanism transmitting joint between the forefoot and hindfoot segment[47].

**Muscle models and foot constraint**

Early muscle model studies have led to databases of origins and insertions of lower extremity muscles based on cadaver studies[54,55]. However, these databases had limitations of small sample sizes, lack of gender and racial variety, and wrapping points allowing muscle lines-of-action to pass through bones[56]. For most muscles, the origin and insertion points are enough, however, for muscles such as the quadriceps, addition landmarks wrapping around bones are needed while the body is in many postures. Kepple et al. created a new musculoskeletal database using a large number of specimens and allowing for comparisons of gender and racial variation, but still faced problems of software implementation[56]. In order to remedy the limitations associated with the earlier databases, Delp et al. [57] created a standard implementation musculoskeletal database with the muscle-tendon actuator model proposed by Zajac [41]. This is a generic model which scales a specific musculotendon actuator, so that its force-length relationship can be evaluated.

An appropriate ground contact model determines how the interaction of the foot and ground will be defined during the stance phase, which has often been a challenge in the computational simulation. In IAA, this was especially substantial for decomposing GRF arising from certain muscles. If assuming the biomechanical system to be in rigid contact with the environment, performing the decomposition is a relatively straightforward procedure. For example, one can simulate foot-flat phase by fixing the foot to the ground, and the corresponding GRF made by an individual muscle force equals the enforcement of the kinematics constraints of the fixing joint. When foot contact with the environment was modeled using spring-damping units under the sole, the decomposition was more complex. Anderson et al. employed five spring-damping units on each foot, whose forces were always on but varied exponentially with displacement[58]. In a study by Neptune et al.[59], the contact between the foot and the ground was modeled as 30 independent visco-elastic elements with Coulomb friction in order to include the mechanical properties of a shoe and underlying soft tissues.

**SPECIFIC AIMS**

The scope of the thesis focused on the biomechanics consequence of the injury and deformity in the foot and ankle joint. Study I aimed to quantify foot motion changes in subjects with post-operative ankle injury (i.e. ankle fractures) and Study II aimed to identify the lower limb muscle function in the presence of foot deformity (i.e. hindfoot inversion or eversion). The specific aims were:

**Study I**

1. To determine whether ankle fractures resulted in kinematic deviations at or around the injured area.
2. To identify the secondary effects caused by unilateral ankle fractures, i.e. motion between other segments in bilateral limbs.
3. To explore whether the clinical ankle function score Olerud/Molander Ankle Score (OMAS) was associated with kinematics parameters.

**Study II**

1. To study the effect of subtalar inversion/eversion on the dynamic function of the tibialis anterior, gastrocnemius and soleus to accelerate the subtalar, ankle, knee and hip joints.
2. To compute the forward (propulsion) and vertical (support) acceleration of the body center of mass (COM) and study the effect of the subtalar angle on the propulsion and support accelerations of COM.
MATERIALS AND METHODS

Detailed description of all the materials and methods used in this thesis are given in the original studies. A summary of these methods are presented here. Subject participation was voluntary. Ethical approval for this study was obtained from Karolinska Institutet Ethics Committee.

SUBJECTS

Study I

Eighteen patients with ankle fractures who were treated with open reduction and internal fixation (ORIF) at Karolinska University Hospital, participated in a follow-up study using clinical gait analysis including a multi-segment foot model. Twelve patients had a lateral malleolar fracture and 6 patients had a trimalleolar fracture. An age- and gender-matched control group was gathered from a cohort of healthy adults without musculoskeletal disease or history of lower-extremity injury (Table 1).

Study II

Eight healthy adult controls without musculoskeletal disease or history of lower-extremity injury participated in the study (Table 1).

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<tr>
<td>Characteristics</td>
</tr>
<tr>
<td>Number of subject</td>
</tr>
<tr>
<td>Age (yrs)(^1)</td>
</tr>
<tr>
<td>Male/Female</td>
</tr>
<tr>
<td>Height (cm)(^2)</td>
</tr>
<tr>
<td>Body weight (Kg)(^2)</td>
</tr>
</tbody>
</table>

\(^1\) median(range)  
\(^2\) Mean (S.D.)

GAIT ANALYSIS

Procedure (Studies I and II)

Subjects were tested in 3D gait analysis along a 10m walkway using an 8-camera motion analysis system (Vicon MX 40, Oxford, UK). Retro-reflective markers were placed on bony landmarks or specific anatomical positions as required by the kinematics model. The subjects walked barefoot at a self-selected pace. A series of walking trials were collected to achieve three left and three right trials yielding complete data sets in Study I and 1 representative trial used as normal input configuration in Study II.
Model

Study I

All subjects were tested with a modified version of the Oxford foot model[47]. The model simplified the foot structure to three rigid segments (tibia, hindfoot, and forefoot) and one vector (hallux). The midfoot was regarded as a mechanism transmitting motion between the hindfoot and forefoot. All inter-segment motions except hallux were free of constraints, i.e. six degrees of freedom. A set of 18 markers (9mm) was placed on body landmarks on each side in a static trial and 4 of them were then removed in the dynamical trials (Fig 7 and Table 2).

Table 2: Names and positions of markers

<table>
<thead>
<tr>
<th>Marker Name</th>
<th>Position</th>
<th>Segment</th>
</tr>
</thead>
<tbody>
<tr>
<td>L/RMKN</td>
<td>Left/Right medial femoral condyle</td>
<td>Femur</td>
</tr>
<tr>
<td>L/RLKN</td>
<td>Left/Right lateral femoral condyle</td>
<td>Femur</td>
</tr>
<tr>
<td>L/RHFB</td>
<td>Left/Right head of fibular</td>
<td>Tibia</td>
</tr>
<tr>
<td>L/RTUB</td>
<td>Left/Right tibial tuberosity</td>
<td>Tibia</td>
</tr>
<tr>
<td>L/RSHN</td>
<td>Left/Right anterior aspect of shin</td>
<td>Tibia</td>
</tr>
<tr>
<td>L/RMMA</td>
<td>Left/Right medial malleolus</td>
<td>Tibia</td>
</tr>
<tr>
<td>L/RANK</td>
<td>Left/Right lateral malleolus</td>
<td>Tibia</td>
</tr>
<tr>
<td>L/RPCA</td>
<td>Left/Right posterior medial aspect of heel</td>
<td>Hindfoot</td>
</tr>
<tr>
<td>L/RCPG</td>
<td>Left/Right wand marker on posterior calcaneus</td>
<td>Hindfoot</td>
</tr>
<tr>
<td>L/RHEE</td>
<td>Left/Right posterior distal aspect of heel</td>
<td>Hindfoot</td>
</tr>
<tr>
<td>L/RLCA</td>
<td>Left/Right lateral calcaneus</td>
<td>Hindfoot</td>
</tr>
<tr>
<td>L/RSTL</td>
<td>Left/Right sustentaculum tali</td>
<td>Hindfoot</td>
</tr>
<tr>
<td>L/RP1M</td>
<td>Left/Right base of first metatarsal</td>
<td>Forefoot</td>
</tr>
<tr>
<td>L/RP5M</td>
<td>Left/Right base of fifth metatarsal</td>
<td>Forefoot</td>
</tr>
<tr>
<td>L/R1DM</td>
<td>Left/Right head of first metatarsal</td>
<td>Forefoot</td>
</tr>
<tr>
<td>L/R5DM</td>
<td>Left/Right head of fifth metatarsal</td>
<td>Forefoot</td>
</tr>
<tr>
<td>L/RTOE</td>
<td>Left/Right marker between second and third metatarsal heads</td>
<td>Forefoot</td>
</tr>
<tr>
<td>L/RHLX</td>
<td>Left/Right base of hallux</td>
<td>Hallux</td>
</tr>
</tbody>
</table>
A modified method based on a spherical rotation coordinate system [60] was adopted to obtain frontal hallux joint rotation (varus/valgus) relative to forefoot. A unit vector was used to represent the long axis of the hallux segment and the rotation was determined in a reference coordinate XYZ, which was assumed to be fixed and aligned to the forefoot segment. Thus hallux/forefoot varus/valgus can be measured as an angle (\(\theta\)) between the unit vector (\(r\)) of the hallux and its projection on the sagittal plane of the forefoot (XZ plane, Fig 8).
**Study II**

All subjects were tested with a conventional full-body marker set (Fig 9), plus a modified Oxford foot model marker set which was described in the previous section.

![Figure 9: Marker placement for the whole body model set.](image-url)
INDUCED ACCELERATION ANALYSIS (STUDY II)

Mathematical model

The generalized equations-of-motion of a multi-articulated body system[41] can be written as:

\[ \ddot{\mathbf{q}} = \mathbf{I}^{-1}(\ddot{\mathbf{q}}) = \mathbf{R}(\mathbf{q}) \mathbf{f}_M + \mathbf{G}(\mathbf{q}) + \mathbf{C}(\dot{\mathbf{q}}, \dot{\mathbf{q}}^2) + \mathbf{\Gamma}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{F}_E \]  \hspace{1cm} (1)

Where \( \mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}} \) are the vectors of generalized coordinates, velocities and accelerations; \( \mathbf{I}(\mathbf{q}) \) the system mass matrix; \( \mathbf{f}_M \) the vector of muscle forces; \( \mathbf{R}(\mathbf{q}) \) the matrix of muscle moment arms; \( \mathbf{G}(\mathbf{q}) \) the vector of gravitational force; \( \mathbf{C}(\dot{\mathbf{q}}, \dot{\mathbf{q}}^2) \) the vector of Centripetal and Coriolis forces; \( \mathbf{\Gamma}(\mathbf{q}, \dot{\mathbf{q}}) \) the vector of ligament torques; \( \mathbf{F}_E \), the vector of external force (e.g. GRF).

Thus the accelerations \( \ddot{\mathbf{q}} \) are:

\[ \ddot{\mathbf{q}} = \mathbf{I}(\mathbf{q})^{-1} \{ \mathbf{C}(\dot{\mathbf{q}}, \dot{\mathbf{q}}^2) + \mathbf{G}(\dot{\mathbf{q}}) + \mathbf{\Gamma}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{R}(\mathbf{q}) \mathbf{f}_M + \mathbf{F}_E \} \]  \hspace{1cm} (2)

Since \( \mathbf{I}(\mathbf{q})^{-1} \) is non-diagonal, any one muscle force \( \mathbf{f}_M^i \), contributes instantaneously to any acceleration \( \ddot{\mathbf{q}}_k \) in \( \ddot{\mathbf{q}} \), and thus to all segmental and joint linear and angular accelerations[34].

The contribution of an individual muscle force \( \mathbf{f}_M^i \) to the instantaneous accelerations of the segments \( \ddot{\mathbf{q}} \) at a certain instant is presumed to be the summed contribution arising from \( \mathbf{f}_M^i \) at that instant, and the GRF due to the immediate past trajectory of \( \mathbf{f}_M^i \) [61]. Eq (2) thus can be reformulated as Eq (3):

\[ \ddot{\mathbf{q}} = \mathbf{I}(\mathbf{q})^{-1} \{ \mathbf{R}(\mathbf{q}) \mathbf{f}_M^i + \mathbf{F}_E^{M_i} \} \]  \hspace{1cm} (3)

All other muscle forces, gravitational forces, and forces terms arising from angular velocities were set to zero. The portion of the GRF caused by muscle \( M_i \) activation was calculated using a ground-foot contact model, which is described in section Ground-foot contact.

Musculoskeletal model

In the current study, a subject 3D linkage model configured by gait data and driven by one unit muscle force was used. This model was developed based on Delp model[57] using SIMM (MusculoGraphics, Inc, Chicago, IL) and consistent with the conventional gait model with modifications made for additional degrees of freedom at foot/ankle.

Ground-foot contact

Three joints were added to model the rigid ground/foot contact. Because these explicit joints were used to constrain the foot, the measured GRF from the walking trials were not used, instead, the joint reaction force calculated by the dynamic simulation acted to constrain the foot.
**Configuration data**

The excessive subtalar inversion or eversion was modeled by offsetting up to ±20° from the normal subtalar angle while other configurations remain unaltered.

**DATA ANALYSIS**

**Kinematics (Study I)**

Discrete kinematics and temporal-spatial parameters were calculated for each gait cycle, and the average from the three left and three right gait cycles were used for further statistical analysis. The kinematics were represented as relative angles and are summarized in Table 3.

Statistical analyses were performed using the SPSS software package. Kinematics and temporal-spatial parameters were analyzed using a two-way repeated analysis of variance (ANOVA) with side (injured side and non-injured side) as within group factor and group (ankle fractures and control group) as the between-group factor[62]. If a significant interaction (p ≤ 0.05) was found between factors, Bonferroni simple main effects tests were performed. The Spearman’s rank correlation coefficient was used to identify associations between OMAS and the inter-segment foot kinematics parameters[62].

**Table 3: Kinematic parameters in Study I**

<table>
<thead>
<tr>
<th></th>
<th>Stance and Swing phase</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Hindfoot/Tibia angle</td>
</tr>
<tr>
<td>Sagittal plane</td>
<td>Max Dorsi, Max Plan, ROM</td>
</tr>
<tr>
<td>Frontal plane</td>
<td>Max Inv, Max Ever, Ave</td>
</tr>
<tr>
<td>Transverse plane</td>
<td>Max Int, Max Ext, ROM</td>
</tr>
</tbody>
</table>

Max: maximum  
Min: minimum  
Ave: average  
Dorsi: dorsiflexion  
Plan: plantarflexion  
ROM: range of motion  
Inv: inversion  
Ever: eversion  

**Induced joint angular and body center of mass accelerations (Study II)**

Induced acceleration analysis was used to calculate effects of the excessive subtalar inversion/eversion on the potential dynamic function of the tibialis anterior, gastrocnemius, and soleus during the stance...
phase in five subtalar configurations (Inversion 20°, Inversion 10°, Normal, Eversion 10°, Eversion 20°). Six parameters were used as an IAA profile for each muscle: hip flexion/extension angular acceleration, knee flexion/extension angular acceleration, ankle dorsi/plantarflexion angular acceleration, subtalar inversion/eversion angular acceleration, COM in the global anterior (propulsion) and vertical (support) directions.
RESULTS AND DISCUSSION

MULTI-SEGMENT FOOT KINEMATICS (STUDY I)

The main contribution from this study is that it describes characteristic multi-segmental foot motions in patients one year post-operatively (Table 4 and see Fig 1 in Paper I), which was difficult to evaluate clinically. Still, very few gait studies have focused on the ankle joint (see Paper I) and the first study we know of evaluating post-operative ankle fractures with a multi-segment foot model.

Table 4: Results summery in Study I: inter-segmental kinematics.

<table>
<thead>
<tr>
<th>Inter-segmental foot kinematics</th>
<th>Ankle fracture group Vs. Control group (Injured side)</th>
<th>Injured side Vs. Non-injured side (Ankle fracture Group)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hindfoot/Tibia</td>
<td>Max plantarflexion (Swing)↓</td>
<td>Max plantarflexion (Swing)↓</td>
</tr>
<tr>
<td></td>
<td>Sagittal ROM (Swing)↓</td>
<td>Sagittal ROM (Swing)↓</td>
</tr>
<tr>
<td>forefoot/Hindfoot</td>
<td>Transverse ROM↓</td>
<td>Transverse ROM↓</td>
</tr>
<tr>
<td>forefoot/Tibia</td>
<td>Max Plan (Swing)↓</td>
<td>Max plantarflexion (Swing)↓</td>
</tr>
<tr>
<td></td>
<td>Sagittal ROM (Swing)↓</td>
<td>Sagittal ROM (Swing)↓</td>
</tr>
<tr>
<td></td>
<td>Max adduction (Swing)↓</td>
<td>Max adduction (Swing)↓</td>
</tr>
<tr>
<td></td>
<td>Transverse ROM↓</td>
<td>Transverse ROM↓</td>
</tr>
<tr>
<td>Hallux/Forefoot</td>
<td>Max dorsiflexion↓</td>
<td>Max dorsiflexion↓</td>
</tr>
<tr>
<td></td>
<td>ROM (Swing)↓</td>
<td>ROM (Swing)↓</td>
</tr>
</tbody>
</table>

Swing: swing phase only
Stance: stance phase only
Max: maximum
ROM: range of motion
Ave: average

The findings in this thesis of a smaller ROM in the injured talocrural joint corresponded to the previous report and were believed as a result of stiffness, pain and swelling[63]. Our findings of smaller transverse ROM in the forefoot and sagittal ROM in the hallux of the injured side could also be a sign of residual joint stiffness following surgery and immobilization.

The observed reduction of less hindfoot and forefoot plantarflexion and hallux dorsiflexion during pre-swing could be a compensation strategy for the restricted motion of the injured ankle joint, which indicated that patients tended to lift rather than push off the foot, prolonging the double-support phase.

Although no direction comparison can be made between our study and the study by Becker[64], our observations of less adducted forefoot in the injured side indicated that the forefoot may be the compensation area of the injured ankle. We also found, that compared to the controls, the hallux of the non-injured foot was more varus during the stance phase. Further investigation was needed to identify whether it was also an influence of the injured ankle.

In our study, Olerud/Molander ankle score was found to fair-moderately correlate with Hindfoot/Tibia peak dorsiflexion and sagittal ROM in the swing phase, which contradicted the study by Losch et al.[65], who did not find significant correlations between gait and clinical parameters examined by a
different functional score. However, temporal-spatial parameters indicated weak correlations with the clinical score both in our and their study.

**INDUCED JOINT ANGULAR ACCELERATIONS AND BODY CENTER OF MASS ACCELERATIONS (STUDY II)**

The main contribution of this study was to identify how gait deviations in one plane (i.e. excessive subtalar inversion or eversion) can affect the dynamic function of the tibialis anterior, gastrocnemius and soleus to accelerate joints in other planes (e.g. sagittal plane) and body COM (see Fig 1-4 in Paper II). The findings of the current study attempted to shed some light on the relationship between the pathological gait and individual muscle function.

In accordance with a previous study[66], the unaltered gait, the muscles generally acted as expected, i.e. tibialis anterior dorsiflexed the ankle, and soleus and gastrocnemius plantarflexed the ankle. We also found that the gastrocnemius can extend knee in the 1st and 3rd rockers, i.e. contrary to its anatomical description as a knee flexor, which corresponded to the previous observations of the bi-articular muscle’s counterintuitive function[67].

Our findings suggest that less effective ankle dorsiplantarflexors may result from excessive subtalar eversion. This can diminish the gastrocnemius’ ability to plantarflex the ankle, and the soleus’ ability to extend the knee, and increase the tibialis anterior’s ability to flex the hip during the 1st rocker, which may lead to a less plantarflexed ankle, less extended knee and more flexed hip after IC.

It is worth noting that, we found that in normal gait, the soleus and gastrocnemius had potentials to evert the subtalar joint, which was in contrast with their anatomical function as invertors[5]. This can be interpreted using inertial couplings, where the large plantarflexion acceleration generated by the soleus and gastrocnemius at the ankle also caused eversion acceleration at the subtalar joint. It overwhelmed the inversion accelerations caused by the muscles’ and ground foot joint reaction force’s smaller inversion leverage.

Our findings of vertical support and forward progression accelerations generated by plantarflexors during the late-stance in normal gait corresponded to previously reported findings[39,68]. The findings of the tibialis anterior’s ability to support body and decelerating progression after initial contact was consistent with its established action to resist foot fall in the 1st rocker. In our study, the soleus was also found to have greater decelerating potential in the 2nd rocker. Furthermore, excessive subtalar inversion had a negative effect on the ankle dorsiflexor’s supporting function, but generated larger support in excessive subtalar eversion.
CONCLUSIONS AND FUTURE WORK

The objective of the thesis has been to discuss gait changes and muscle roles due to foot and ankle injury or deformity. Gait analysis and computational simulation are two independent but integrated methods, which comprised two individual studies here.

**Study I** presented new data of gait and foot motions in patients one year after ankle fracture surgery. Although the clinical functional score showed fairly good postoperative results, some kinematic deviations were still observed, even in the non-injured area, e.g. the forefoot. Restricted ROM at and around the injured ankle was believed to be a sign of residual stiffness due to the surgery and immobilization, which also possibly led to the secondary motion restriction and deviations found in the forefoot and hallux segment. Gait analysis can be considered as an additional dynamic post-treatment evaluation for patients with foot injury. The strategy adopted to compensate ankle injury can be used as a reference for evaluation of future patients.

Suggestion of future studies includes a 3D multi-segment foot kinetic model and plantar pressure analysis. It will help to relate foot motion with kinetics and loading patterns which may lead to a better understanding of gait strategy and help in the specific rehabilitation decision-making. Additionally, collecting data from a larger cohort has the potential to characterize specific subgroups based on fracture classification, gender or age.

**Study II** identified how one plane gait deviation (subtalar inversion or eversion) can alter the dynamic functions of individual ankle dorsi/plantarflexors. Joint accelerations and body COM accelerations generated by one unit muscle force were calculated in five subtalar configurations. It was confirmed that, in normal gait, muscles generally act as their anatomical definitions and can also create motion in joints they do not span. We also found that excessive subtalar eversion had a negative effect on the ankle plantarflexors and tibialis anterior. Induced acceleration analysis demonstrated its ability to isolate the contributions of individual muscles to a given factor and provided a means to analyze how muscles can create motion in joints. Although gait deviations here were manipulated from normal configurations, IAA can shed some light on the interaction between pathological gait and individual muscle functions.

Future improvement considering more accurate foot-ground constraints with underfoot spring elements, and real pathological gait data and muscle excitation pattern (EMG) input will help to create a more realistic computational model and provide a better solution to quantify muscle roles in pathological gait. In addition, the analysis involving kinematics, kinetics and individual muscle function can give a whole picture of the biomechanical consequences arising from certain foot deformity or injury.
SUMMARY OF PAPERS

PAPER I

The study aimed to quantify foot kinematics and tempo-spatial changes in patients one year after surgical treated ankle fractures. A validated multi-segment foot model was used in the 3D gait analysis. The gait parameters from 18 subjects were compared to age and gender matched controls. We found that patients with ankle fractures experienced motion restriction not only in the injured talocrural joint, but also in forefoot and hallux segment, which indicated a sign of the stiffness after surgery and mobilization. Less adducted forefoot and a slightly varus hallux in the injured foot could be the compensation strategy for the injury joint. Findings of this study showed that unilateral talocrural fractures can still affect other areas in the foot one year after the surgery. Moreover, gait analysis with multi-segment foot model provided a quantitative and objective way to dynamically evaluate postoperative foot and ankle injury.

PAPER II

The study aimed to determine how one plane deviation (i.e. subtalar inversion/eversion) can alter the capacity of muscles to generate joint angular and body COM accelerations in other planes (e.g. sagittal plane). IAA was used to compute the accelerations produced by the gastrocnemius, soleus and tibialis anterior in 5 subtalar inversion or eversion configurations. A subject specific 3D linkage musculoskeletal model configured by the gait data was driven by 1 N of muscle force. The main findings were that, in normal gait, muscles generally function as their anatomical definitions. It was confirmed that muscles can create motion in joints which they do not span, due to inertial couplings. The gastrocnemius and soleus had contributions to the body vertical support and forward progression. Furthermore, excessive subtalar eversion was found to diminish ankle dorsi/plantarflexor’s function. In conclusion, IAA demonstrated its ability to isolate the contributions of individual muscle to a given factor. Further analysis with accurate foot/ground constraints, muscle excitation pattern and real gait input data could improve the understanding of interaction of pathological gait and muscle roles.
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