Journal of Biomechanics 46 (2013) 2578–2585

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

# Segmental motion of forefoot and hindfoot as a diagnostic tool

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ARTICLE INFO

*Article history:*
Accepted 25 August 2013

*Keywords:*
Foot and ankle modeling
Gait simulation
Motion Analysis
Neuromuscular dysfunction

ABSTRACT

Segmental motions derived from non-invasive motion analysis are being used to investigate the intrinsic functional behavior of the foot and ankle in health and disease. The goal of this research was to examine the ability of a generic segmented model of the foot to capture and differentiate changes in internal skeletal kinematics due to neuromuscular disease and/or trauma. A robotic apparatus that reproduces the kinematics and kinetics of gait in cadaver lower extremities was employed to produce motion under normal and aberrant neuromuscular activation patterns of tibialis posterior and/or tibialis anterior. Stance phase simulations were conducted on 10 donor limbs while recording three-dimensional kinematic trajectories of (1) skin-mounted markers used clinically to construct segmented foot models, and (2) bone-mounted marker clusters to capture actual internal bone motion as the gold standard for comparison. The models constructed from external marker data were able to differentiate the kinematic behaviors elicited by different neuromuscular conditions in a manner similar to that using the bone-derived data. Measurable differences between internal and externally measured kinematics were small, variable and random across the three axes of rotation and neuromuscular conditions, with a tendency toward more differences noted during early and late stance. Albeit slightly different, three-dimensional motion profiles of the hindfoot and forefoot segments correlated well with internal skeletal motion under all neuromuscular conditions, thereby confirming the utility of measuring segmental motions as a valid means of clinical assessment.



## 1. Introduction

Quantitative data from gait analysis have proven useful in clinical and research studies to assess the behavior of the hip, knee and ankle under various conditions. Traditional gait analysis treated the foot as a single rigid body (Kadaba et al., 1990; Davis et al., 1991), but unlike other segments of the lower extremity, the foot is composed of multiple bones and joints with relative motions between them. Advancements in non-invasive three-dimensional photogrammetry have enabled finer division of the foot into multiple segments with each one treated as a single and separate rigid body (e.g. Carson et al., 2001; Leardini et al., 2007; Bruening et al., 2012). These newer methods are coming into more widespread use (Theologis et al., 2003; Ness et al., 2008; Houck et al., 2009) and hold promise as a means of providing clinically useful information given that many common neuromuscular conditions (e.g. cerebral palsy, stroke, posterior tibial tendon dysfunction, and diabetes) produce aberrant motions within the foot. Objective assessments of the segmented model in terms of its ability to accurately reflect the internal functional consequences of pathology are therefore warranted.

Several prior studies have evaluated the performance of the multi-segmented foot model under normal conditions. Two *in vivo* studies of normal walking found reasonable agreement between externally measured segment kinematics and directly measured bone kinematics (Westblad et al., 2002; Nester et al., 2007). Another *in vivo* study by Shultz et al. (2011) examined the soft-tissue artifact inherent in external markers attached to the skin to define foot segments and noted maximal, consequential artifact at toe-off. Prior *in vitro* work in our laboratory examining the rigid body assumption and marker artifact found that the segmented foot model performed well under conditions representative of normal gait (Okita et al., 2009). Taken in total, these studies are encouraging but none provide appreciable insight concerning the diagnostic abilities of the segmented model approach, i.e. its ability to reliably differentiate internal kinematic behaviors induced by common neuromuscular conditions. The goal of this investigation was to examine the ability of a segmented foot model to capture internal functional deficits due to aberrant contractile activity of the tibialis anterior (TA) and tibialis posterior (TP). We examined two null-hypotheses: (1) forefoot and hindfoot

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http://dx.doi.org/10.1016/j.jbiomech.2013.08.014

segment kinematics can differentiate across different neuromuscular conditions as well as directly measured bone kinematics; and (2) there are no differences between the kinematic behavior of a segment and that of its underlying bone(s) irrespective of neuromuscular condition.

## 2. Methods

An established robotic apparatus that reproduces the kinematics and kinetics of gait in cadaver lower extremities (Sharkey and Hamel, 1998; Hoskins, 2006) was used to model gait under simulated normal muscle activity and then under abnormal phasic activity of either TA, TP, or both muscles. The experiment employed two sets of marker data: the first data set was compiled from skin mounted markers such as would be used in clinical applications to calculate motions of the forefoot and hindfoot segments; the second data set was derived from bone mounted marker clusters to calculate motions of the bones within each segment.

### 2.1. Cadaver modeling

Ten donated normal fresh frozen cadaver lower extremities (5 M/5 F, 42–84 years old) were evaluated. The tibia and fibula were transected approximately 23 cm superior to the sole of the foot, and all soft tissues 5 cm superior to the malleoli were removed. Special attention was paid to preserve skin and retinaculi about the ankle, and the entire tendon lengths from six muscle groups: (1) gastrocnemius and soleus complex, or triceps surae (TS); (2) tibialis anterior and extensor digitorum longus (TA); (3) tibialis posterior (TP); (4) flexor digitorum longus (FDL); (5) flexor hallucis longus (FHL); and (6) peroneus longus and peroneus brevis muscles (PER). An intramedullary tibial rod and polymethylmethacrylate were used to couple the shank to the kinematic actuators of the device. Tendons from each muscle group were connected to independent linear actuators through load cells, cables and freeze clamps (Sharkey et al., 1995).

The robotic dynamic activity simulator (Sharkey and Hamel, 1998; Hoskins, 2006) was used to conduct simulations of the stance phase of gait at 1/20th of normal walking speed. Simulations were conducted using an in-house library of shank kinematics and corresponding ground reaction force profiles taken from normal subjects; enabling selection of kinematic control using data sets from subjects with anthropometric characteristics matching those of each donated limb. Temporally-based muscle force profiles were constructed from rectified EMG profiles measured in normal subjects (Perry, 1992) with adjustments for force–length and force–velocity properties (Gallucci and Challis, 2002). Sequential set-up trials were conducted to match the shape of the vertical ground reaction force profile measured in the simulations to the target profile produced by the subject whose kinematic data were input to drive tibial kinematics, but with amplitudes scaled according to the estimated body mass of each donor (35–50Kg). During these trials, the height of the specimen carriage was iteratively lowered until the first peak of the vertical ground reaction force profile closely matched the target, after which plantar flexor forces were iteratively increased, while maintaining their proper relational magnitudes, until the second propulsive peak of the vertical ground reaction force also matched the target profile. Once established, all input parameters were held constant for all trials, except for the TA and TP forces that were experimentally manipulated.

Five sets of muscle force control profiles (*Normal*/*ExtTA*/*ExtTP*/*ExtTATP*/*NoTP* conditions) were used to simulate normal and aberrant function of TA and/or TP (Fig. 1c). The *Normal* baseline condition reproduced the normal temporal activities of all muscle groups as reported by Perry (1992). Three extended phase hyperactive conditions of TA and/or TP, representing the continuous abnormal firing patterns often found as components of cerebral palsy patients (Hoffer and Perry, 1983; Barto et al., 1984; Renders et al., 1997) and stroke victims (Perry et al., 1978) were investigated. In the *ExtTA* condition the TA was held constant at approximately 40% of the peak force used in the simulated *Normal* condition, while maintaining normal function in all other muscles. Similarly, during the *ExtTP* condition the TP was held constant at approximately 65% of its peak force in the *Normal* condition, while simulating normal contractile activity in the remaining muscles. In the *ExtTATP* condition both TA and TP were continually activated at approximately 40% and 65% of their peak forces in the *Normal* profiles. In the final *NoTP* condition, temporal activity of TP was completely absent.

### 2.2. Motion analysis

Three-dimensional photogrammetry data were captured at 100 Hz using a seven-camera Eagle system (Motion Analysis Corporation, Santa Rosa, CA) focused on a working volume of approximately 1 $m^3$, with typical reconstruction residuals of approximately 0.3 mm. Data from the force platform (AMTI, Newton, MA) and muscle force transducers (A.L. Design, Buffalo, NY) were collected in synchronization with the marker data.

Two separate but identically executed sets of experimental trials were conducted to avoid cross-talk and interference between skin and bone mounted marker sets and to avoid disrupting the skin to better reproduce the *in vivo* condition. In the first set of experimental trials, external markers made from 9.5 mm diameter polyethylene spheres were glued to the skin overlying nine strategic landmarks (Fig. 2). Skin marker data were collected over three simulations under each of the five neuromuscular conditions. External markers were then removed without removing the specimen from the simulator and marker clusters, consisting of four 6.0 mm diameter polyethelylene spheres connected through 1.6 mm diameter carbon graphite rods, were rigidly fixed through self-tapping screws into the 1st, 3rd and 5th metatarsals and calcaneus (Fig. 3). A second set of experimental trials was then conducted using the exact same sequence of simulation settings used in the trials tracking skin marker motions.

### 2.3. Data analysis

Marker trajectories recorded during the experimental sessions with skin and bone mounted marker sets were post-processed with EVaRT software (Motion Analysis Corporation, Santa Rosa, CA), and exported to custom written Matlab programs (The Mathworks, Natick, MA). All marker trajectories and ground reactions forces were lowpass filtered with a 4th order, dual-pass Butterworth filter at a 2 Hz cut-off frequency. Data from 0% to 90% of the stance phase were used for subsequent analyses. The last 10% of the stance phase was excluded due to marker dropout as a consequence of obstructed camera views.

A multi-segment foot model consisting of forefoot, hindfoot, and shank segments (Okita et al., 2009) was created using the skin-mounted marker data (Fig. 2). The model was intended to be representative of the clinical models in the literature (Davis et al., 2003; Humm et al., 1999; Kaufman et al., 2003; Kidder et al., 1996). A right-handed, orthogonal coordinate system was defined for each segment, with *u*, *v*, and *w* representing unit vectors in approximately posterior (−)/anterior (+), inferior (−)/superior (+), and lateral (−)/medial (+) directions for the left foot. Rotations about the *Z*-axis were considered plantarflexion (PF; −)/dorsiflexion (DF; +), about the *Y*-axis as internal (INT; −)/external (EXT; +) rotation and about the *X*-axis as inversion (INV; −)/eversion (EVR; +). For each specimen the neutral reference pose for all kinematic calculations was assigned to the instant when the tibia was vertically oriented during the midstance of a representative trial under the *Normal* condition.

The homogeneous coordinate transformation matrix of each segment at every frame of the stance phase with respect to the reference pose was calculated by using a least squared method (Challis, 1995). The rotations of the forefoot and hindfoot segments about global axes were obtained by taking *ZYX* Cardan decompositions of the rotation matrices with respect to the global frame. Intersegment rotations of the forefoot with respect to the hindfoot were obtained by taking *ZYX* Cardan decompositions of the relative coordinate transformation matrix between these segments. The coordinate system for each bone was defined by using three of its attached markers (Fig. 3). Similar to the segmental kinematic calculations, the three-dimensional rotations of the bones (1st, 3rd, and 5th metatarsals and calcaneus) were determined by taking *ZYX* Cardan decompositions of the rotation matrices with respect to the global frame. The neutral pose was determined using the same data frame used in the segmental calculation.

### 2.4. Statistics

A critical requirement for the experimental design was consistent and reproducible loading of the extremity between the skin and bone mounted marker trials. This was assessed with repeated-measures analyses of variance (ANOVA; SPSS, IBM Corp., Armonk, NY) run for each neuromuscular condition with ground reaction forces, tibia trajectories and muscle forces as separate dependent variables with the significance threshold at $p \leq 0.05$. Specimen was treated as a random factor and time, in 10% increments between 0% and 90% of the stance, was treated as the repeated variable.

The independent ability of either the segment or bone kinematic analyses to detect functional changes elicited by aberrant TA or TP activity was examined using Bonferroni corrected pair-wise comparisons of the *NoTP*, *ExtTA*, *ExtTP*, and *ExtTATP* conditions against the *Normal* condition at every 10% of the stance phase. The capability of the segmented foot model to accurately reflect internal skeletal behavior under each neuromuscular condition was examined using pair-wise *t*-tests comparing the global orientations of the clinical segments to their underlying bones at every 10% of the stance phase. The significance threshold was set at $p < 0.005$ to correct for 10 multiple comparisons (at every 10% of stance) made on each kinematic variable (e.g. PF/DF of the forefoot segment). Thus, the corrected significance threshold at each time point was set at $p \leq 0.05/10 = 0.005$.

## 3. Results

### 3.1. Experimental repeatability

All 10 specimens displayed consistent loading behavior across trials employing the skin and bone mounted markers, as determined