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Segmental Motion of Forefoot and Hindfoot as a Diagnostic Tool

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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 ten 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.

Keywords

Foot and ankle modeling; Gait simulation; Motion Analysis; Neuromuscular dysfunction

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CONFLICT OF INTEREST STATEMENT

There were no conflicts of interest in the performance of this research.

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INTRODUCTION

Quantitative data from gait analysis has 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, 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 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.

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.

Cadaver Modeling

Ten donated normal fresh frozen cadaver lower extremities (5M/5F, 42 to 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 (Figure 1-c). 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.

Motion Analysis

Three-dimensional photogrammetry data was captured at 100 Hz using a seven-camera Eagle system (Motion Analysis Corporation, Santa Rosa, CA) focused on a working volume of approximately 1 m³, 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 9 strategic landmarks (Figure 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 polyethylene 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 (Figure 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.

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 0–90% 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 (Figure 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 mid-stance 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 (Figure 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.

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 $p < 0.05$. 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 $p = 0.05/10 = 0.005$.

RESULTS

Experimental Repeatability

All ten specimens displayed consistent loading behavior across trials employing the skin and bone mounted markers, as determined from matching ground reaction force, muscle force, and tibial trajectory profiles (Figure 1). Repeated measure ANOVA tests verified that there were no differences in loading across the two marker sets with the majority (71%) of p -values exceeding 0.80 and the smallest at 0.173.

Methodological Sensitivity to Neuromuscular Dysfunction

Loss of TP function (*NoTP*) had no effect on the kinematic behaviors of any bone or segment relative to the *Normal* condition ($p > 0.010$ for all comparisons, and $p > 0.050$ for 94% of comparisons). Extended phase hyperactivity of the TA (*ExtTA*) precipitated up to 3.0 degrees more inversion of the 1st and 3rd metatarsals and forefoot segment than the normal condition, but only between 20 and 30% of the stance phase ($p < 0.0025$). Hyperactivity in TP (*ExtTP*) or TA and TP (*ExtTATP*) produced clearly evident changes in the INV/EVR and INT/EXT rotations of all bones (Figure 4) and segments (Figures 5) over large portions of the stance phase but had minimal effect on PF/DF. Greater inversion as a consequence of hyperactivity (*ExtTP* or *ExtTATP*) was most striking from 20 to 60% of stance, with the 1st and 3rd metatarsals, calcaneus, and forefoot segment inverting an additional 1.5 to 6.5 degrees ($p < 0.0014$ for all comparisons). Hyperactivity in TP or TA and TP increased the internal rotations of all bones and segments by 0.6 to 5.4 degrees between 10 and 90% of the stance phase ($p < 0.0025$ for all comparisons). Changes in plantar-dorsiflexion as a function of hyperactivity was only measureable in the 5th metatarsal during mid-stance (30–70% of stance) where dorsiflexion increased by a maximum of 1.8 degrees ($p < 0.0047$) and at 30% of stance when the 1st metatarsal was more plantarflexed by 1.1 degree ($p < 0.0021$). Forefoot-to-hindfoot inter-segmental motions were less affected, but nevertheless able to detect differences INV/EVR and INT/EXT rotations under hyperactive conditions.

Segmental versus Skeletal Kinematics

Pairwise comparisons of segment versus bone kinematics revealed small, albeit significant, differences in overall behaviors (Figures 6 and 7). In general, plots of segment rotations appeared similar but somewhat damped and were almost always closer to the neutral axis relative to those recorded for the corresponding bones. Measurable differences between internal and externally measured kinematics were variable and somewhat random across the three axes of rotation and neuromuscular condition, with a tendency toward more differences noted at the onset (weight acceptance) and end (toe off) of stance. The greatest difference measured for INV/EVR was 4.9 degrees and occurred between the 3rd metatarsal and forefoot segment at 90% of stance under the *ExtTATP* condition ($p < 0.0038$). The greatest difference in INT/EXT rotation was 3.8 degrees and occurred between the 5th

metatarsal and the forefoot segment at 0% of stance under the *NoTP* condition ($p < 0.0009$). The greatest difference in PF/DF was 5.6 degrees and occurred between the 3rd metatarsal and forefoot segment at 90% of stance under the *ExtTP* condition ($p < 0.0018$).

DISCUSSION

High fidelity cadaver simulations of foot and ankle function were conducted to quantitatively assess the ability of a generic segmented foot model to capture changes in the skeletal kinematics of the foot induced by aberrant function of TA and/or TP. The ability of the model to repeatedly produce the same loading conditions across experimental trials, an essential requirement for our experimental design, was confirmed. These highly repeatable gait simulations were conducted at 1/20th of normal walking speed, while producing realistic extrinsic muscle tensions and ground reaction forces. The tibial kinematics used to drive the model were taken from normal size-matched subjects and held constant across all five simulated conditions, as were normal contractile activities of TS, PER, FDL, and FHL. It should therefore be emphasized that these experiments were reductionist by design, intended to examine the kinematic consequences of specific dysfunctions of the TP and TA muscles in isolation, despite the fact that these deficits most often manifest as components of more extensive functional deficits involving other muscles and more proximal body segments.

Our modeling was intended to represent the aberrant actions of two muscles frequently affected by neuromuscular diseases of the lower extremity. Specific abnormal muscle actions were selected based predominantly on the reports of Perry and co-workers (Perry et al., 1978; Hoffer and Perry, 1983; Wills et al., 1988). These authors isolated three dominant patterns of abnormal muscle activity underlying cases of dynamic hindfoot varus: continuous activity of TP, continuous activity of TA, and phase reversal of TP to swing. The actual muscle forces produced in these conditions have not been measured and are undoubtedly variable across cases. In modeling hyperactivity we sought to use realistic forces well within the capability of the muscles in question but large enough to clearly reflect a departure from normal (40% of peak in TA, 65% of peak in TP). Attempts to use larger forces in pilot experiments severely overloaded the lateral aspect of the foot. The absence of TP contraction is seen in both neurological disorders and posterior tibial tendon dysfunction (PTTD; Johnson and Strom, 1989).

Methodological Sensitivity to Neuromuscular Dysfunction

The specific changes in neuromuscular function examined here caused readily measurable and statistically significant changes in segment and bone rotations within the coronal (INV/EVR) and transverse (INT/EXT) planes, with much less apparent changes to sagittal plane motions (PF/DF). Two patterns of behavior were clearly differentiated in Figures 4 and 5. The first pattern was produced under the simulated *Normal* and *NoTP* conditions. The second pattern occurred in the presence of a hyperactive TP and so included the *ExtTP* and *ExtTATP* conditions. The *ExtTA* condition produced intermediate profiles that were in most cases statistically similar to those seen in the *Normal* and *NoTP* conditions.

Both methods of kinematic assessment, whether internal or external were able to easily discern the effects of hyperactivity of TP during the stance phase of gait. As would be predicted, this abnormal neuromuscular firing pattern produced exaggerated metatarsal and calcaneal inversion and internal rotation and these shifts were similarly detected, but to a less exaggerated extent, in the segments constructed from external markers. We therefore conclude that the segmented foot model is quite good at detecting the presence of a hyperactive TP.

The kinematic consequences of hyperactivity of the TA were much less obvious, but nevertheless detectable as increased inversion of the 1st and 3rd metatarsals and forefoot segment. The diminished ability of the TA to change skeletal and segmental kinematics relative to TP is likely due to simple functional anatomy in relation to our experimental approach. In life, extended phase TA hyperactivity would tend to counter the plantarflexing force induced by the TS and produce altered shank kinematics and rotations at the ankle, thereby changing the foot's kinematic behavior in both the coronal and sagittal planes.

We observed only small changes in sagittal plane kinematics in these experiments; all neuromuscular conditions produced similar PF/DF rotations in all bones and segments. We interpret this finding to be a function of our experimental approach rather than a weakness in the sensitivity of the kinematic procedures. Only the contractile functions of TA and TP were altered while tibial kinematics and the plantarflexing force profile of TS were held identical across all conditions, whereas profiles in actual patients would vary depending upon degree of pathology. We conclude that the isolated neuromuscular manipulations examined could not appreciably alter the kinematics of the foot in the sagittal plane because of the artificial constraints imposed experimentally. Given the robust ability of both kinematic techniques to detect changes in the other two planes, changes in the sagittal plane kinematics of the foot would likely be highly manifest in actual patients.

The kinematic profiles generated in the absence of TP contraction (*NoTP*) were essentially the same as those generated under the simulated *Normal* condition. This is consistent with prior cadaver work examining PTTD and likely due to the load bearing capability of the intact osteo-ligamentous structure, as observed in acute early stage PTTD (Niki et al., 2001; Imhauser et al., 2004). Using segmented foot analyses similar to the methods employed in this work, Ness et al. (2008) and Houck et al. (2009) recorded motion profiles from PTTD patients that were similar in shape to the motions reported here, and similar in shape to those recorded in normal subjects but offset by several degrees. Our results reinforce prior cadaver work suggesting that the collapse of the longitudinal arch responsible for these offsets is acquired over time through attrition of soft tissue constraints and cannot be produced acutely by eliminating TP contraction.

Perhaps the most important and encouraging result from these sets of experiments was that the segmented model was able to successfully differentiate the neuromuscular deficits simulated and with only slightly diminished sensitivity relative to direct skeletal measurement. Inter-segmental forefoot-to-hindfoot motions were considerably less sensitive in the coronal plane, possibly due to diminished methodological capacity but more likely due to coupled motions between underlying bones and/or the continuous layer of skin and soft tissue bridging the two segments.

Segmental versus Skeletal Kinematics

The capability of the segmented foot model to accurately track actual bone motion was tested by comparing the motions of forefoot and hindfoot segments to those of their underlying bones. Although we found significant differences in measured kinematics in all three planes, these differences were small, tending to occur at the onset and/or later portions of the stance phase when the foot is at its extremes of motion and displacements between bone and soft-tissues would be greatest. Different loading patterns under each neuromuscular condition caused different relative bone motions with different variability, thus different segment deformations and associated errors, leading to the non-systematic differences noted across neuromuscular conditions (Figure 6 versus 7).

Despite small differences, the shapes of the curves generated from external markers were similar to those generated from bone markers but were in almost all cases somewhat

damped, less extreme and tending more toward the neutral position. These observations are likely attributable to displacements and slight lags in motion between bone and skin. Forefoot segment kinematics did not display any preferential agreement with the 1st, 3rd or 5th metatarsals, despite discernible differences in the motions of the bones themselves (Okita et al., 2009). The forefoot segment defined in this work using markers placed on the medial and lateral aspects of the foot captured the overall “average” behavior of forefoot and its underlying bones fairly well but one should use caution in extending these results to other models that define the forefoot segment using other rays as the primary axis (e.g. the 2nd metatarsal; Leardini et al., 2007) or that further divide it into medial and lateral segments (Nester et al., 2007).

The experimental apparatus used to conduct these simulations readily enables manipulation of muscle force timing and amplitude, making it an appropriate model for highly invasive studies of normal and pathologic muscle function assuming our input parameters properly represent in-vivo function, but it does have limitations. Notable deficits that may have affected the outcome of the present work include greatly reduced velocities relative to normal gait and the absence of coronal plane motions of the tibia due to technical constraints of the apparatus (e.g. force-velocity limitations of the kinematic and muscle actuators). The first limitation may have enhanced the performance of segmented model by reducing the amount of skin motion artifact as a consequence of inertial loading. Conversely, the slower speed may have caused greater rotational motion of the segments and bones as a consequence of the viscoelastic soft tissues that bind them, violating the rigid body assumption to a greater extent than would happen in practice. Although motions of the tibia in the coronal plane are minimal during normal ambulation, their absence in the simulation could conceivably produce segment and bone rotations that are somewhat different from those occurring in life.

All factors considered, we conclude that the external measurement technique used here is somewhat less robust than actual skeletal measurement, likely due to more excessive segment deformation and skin artifact. The high statistical power of our experimental design enabled us to detect significant differences between segment and skeletal motions but the differences were relatively small and of questionable clinical importance. Practically speaking, the three-dimensional motion profiles of the hindfoot and forefoot under all neuromuscular conditions were tightly coupled and highly correlated with internal skeletal behavior, leading us to conclude that segmented models of the type examined here are able to provide a reasonably accurate and certainly useful picture of internal skeletal motion under healthy and aberrant muscle conditions.

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Figure 1.

Repeatability plots of a) ground reaction forces, b) trajectories of tibia cluster markers, and c) muscle forces for a single representative specimen. Three trials were conducted under each neuromuscular condition while recording forces and the trajectories of external skin markers, followed by three trials under each condition while recording bone mounted markers.

All plots include the three trials recorded for each marker set.

1-a) Ground reaction forces under the *Normal* baseline condition. Vertical force profiles (upper left) were used to optimize the baseline simulations (*Normal* condition). 1-b) Tibia marker trajectories under the *Normal* condition. Tibial kinematics were based on normal gait

patterns and the same under all neuromuscular conditions. 1-c) Muscle force profiles. The *Normal* condition employed muscle profiles shown on the left 6 plots. The hyperactive TA and TP patterns shown on the right were used to simulate common components of neuromuscular pathology (*ExtTA*, *ExtTP*, *ExtTATP* conditions), while maintaining *Normal* contractile behavior in the remaining muscle groups. The temporal activity of TP was completely absent in *NoTP* condition.

Figure 2. Skin mounted marker placement and segment definitions. MM and MH markers are not shown. The tibial cluster was present in all trials and used to confirm repeatability.

Figure 3.

Bone mounted marker clusters. The tibial cluster was present in all trials and used to confirm repeatability. Marker clusters were rigidly mounted into the middle of the 1st, 3rd, and 5th metatarsals and the lateral aspect of the calcaneus after removing the external skin markers previously tracked to define the segment motion.

Figure 4.

Mean ($n=10$) three dimensional rotations of the 1st, 3rd and 5th metatarsal and calcaneus during the stance phase of gait under the five different neuromuscular conditions. The standard deviation band is shown for the *Normal* condition. Lighter shades indicate portions of the cycle demonstrating significant differences ($p<0.0049$) between *Normal* and *ExtTATP* conditions. Darker shades indicate portions with significant differences ($p<0.0036$) between *Normal* and *ExtTP* and between *Normal* and *ExtTATP* conditions. These shades extend $\pm 5\%$ of the time point being examined (e.g., 15~25% for 10%), except for 0% (0~5%) and 90% (85~90%) of stance.

Figure 5.

Mean ($n=10$) three dimensional rotations of the hindfoot and forefoot segments and forefoot-hindfoot rotations during the stance phase of gait under the five different neuromuscular activation patterns. The standard deviation band is shown for the *Normal* condition. Lighter shades indicate portions of the cycle demonstrating significant differences ($p < 0.0042$) between *Normal* and *ExtTATP* conditions. Darker shades indicate portions with significant differences ($p < 0.0040$) between *Normal* and *ExtTP* and between *Normal* and *ExtTATP* conditions. These shades extend $\pm 5\%$ of the time point being examined (e.g., 15~25% for 10%), except for 0% (0~5%) and 90% (85~90%) of stance.

Figure 6.

Three dimensional rotations (mean $\pm$ sd; $n=10$) of the foot segments (dash lines) and underlying bones (solid lines) during the stance phase of gait under the *Normal* neuromuscular condition. Shading indicates portions of the stance phase demonstrating significant differences ($p<0.0043$) between the segment and underlying bone profiles. These shades extend $\pm 5\%$ of the time point being examined (e.g., 15~25% for 10%), except for 0% (0~5%) and 90% (85~90%) of stance.

Figure 7.

Three dimensional rotations (mean $\pm$ sd; $n=10$) of the foot segments (dash lines) and underlying bones (solid lines) during the stance phase of gait under the *ExtTP* neuromuscular condition. Shading indicates portions of the stance phase demonstrating significant differences ($p < 0.0048$) between the segment and underlying bone profiles. These shades extend $\pm 5\%$ of the time point being examined (e.g., 15~25% for 10%), except for 0% (0~5%) and 90% (85~90%) of stance.