SOURCE OF INCREASED DORSIFLEXION DURING GAIT OF PERSONS WITH PARTIAL FOOT AMPUTATIONS WHEN SHOD OR IN ‘BELOW-ANKLE’ DEVICES

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Introduction: Previous studies of barefoot walking in persons with partial foot amputation (PFA) suggest that ankle dorsiflexion range of motion is either reduced or similar to that observed in able-bodied persons [1-3]. In contrast, studies measuring ankle motion in shoe or shoe plus below-ankle prosthesis/orthosis conditions suggest that ankle dorsiflexion is increased [2], although a clear mechanical reason for this difference is not apparent. It seems logical that some of the differences observed in the ankle kinematic data between barefoot and shod/device conditions may arise from the effect of motion between the residual foot and the shoe/device (e.g., heel slippage, Figure 1A), as has been reported in one investigation [4], or bending between the end of the residuum and devices such as toe fillers (Figure 1B). Both of these scenarios will cause movement between markers defining the local coordinate system of the foot, which could result in an exaggeration of measured dorsiflexion. The ability of marker placement protocols to account for these unique issues has not been systematically investigated but is necessary to obtain a more accurate understanding of gait in persons with PFA. Therefore, the purpose of this investigation was to determine if standard placement of markers on the shoe would lead to exaggerated measurements of range of motion at the ankle for three PFA levels.

Figure 1 – Hypothetical sources of ankle dorsiflexion measurement errors that could occur in gait analysis of persons with PFA. (A) Heel slippage, (B) Bending of the shoe under the end of the residual foot.

Methods: An articulated mechanical model of the leg with three different length partial foot residua (Figure 2) was used to compare the ankle kinematic data derived from a
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potentiometer (embedded in the ankle of the mechanical model) and conventional marker-based gait analysis systems (using a Helen Hayes (HH) marker set). The partial foot model consisted of shank and foot components separated by a hinge joint representing the ankle. The shank piece was made of wood and was connected to the ankle joint through steel uprights on both sides that were screwed into the wooden piece. The foot sections of the device were made using a 90 durometer polyurethane rubber (Smooth-On Inc., Easton, PA, USA) and included several threaded inserts that allowed for the connection of additional pieces to extend the foot length and mimic different levels of PFA. In this way the different lengths of the metatarsophalangeal, transmetatarsal, and Lisfranc residua could be modeled based on anthropometric data from Dillon [5]. A rotational potentiometer was placed in line with the hinge joint and was used to measure ankle motion within the device. The housing of the potentiometer was mounted to the shank component and its shaft was connected to the ankle joint shaft. The ankle joint shaft was rigidly connected to the partial foot piece. Because of this arrangement, rotation of the shank with respect to the partial foot piece turned the shaft of the potentiometer, changing its electrical resistance. The potentiometer was connected to the Motion Analysis system, allowing the signal to be synchronized with the marker displacement data. After calibration (using markers on the unshod device), the potentiometer signal was considered the ‘gold standard’ measurement of ankle joint motion in this study. A strap was connected to the posterior-superior aspect of the partial foot model and the top of the shank piece. During experiments, the strap was pulled taught when the ankle was slightly plantarflexed and fixed over a screw on top of the shank piece. As the shank was rotated over the foot, simulating tibial rotation during stance phase, the strap became more taught, coupling movement of the shank and foot segments.

Ankle joint centers were estimated in both cases by the average of medial and lateral ankle marker positions and a knee center was estimated by the average of medial and lateral markers located near the top of the shank piece. Ankle markers were placed directly along the ankle shaft of the mechanical model. For the HH marker set, a heel marker and toe marker were placed on the shoe as described by Kadaba et al. [6]. A unit vector was created between the ankle and knee centers and a second unit vector was created between the heel and toe markers. The ankle angle for the HH marker set was determined by calculating the arcsine of the dot product of these two vectors.
For each PFA level, a nylon stocking was placed over the foot and the foot was placed into the shoe. The shoe used was a men’s size 8W, E.Z. Strider Walking Shoe with a synthetic leather upper, synthetic rubber sole and two Velcro® straps along the dorsal surface. The insole was removed during the experiments. In each case, the shoe contained a toe-filler necessary to create a snug fit of the partial foot model. Velcro® straps on the shoe were pulled to a force level of 12 lbs (~54 N) using a spring scale and were latched down [7].

Data collection with the partial foot model began with the model in a relatively plantarflexed angle. A technician pushed the shank downward and forward to simulate the dorsiflexion that would occur between mid- and late-stance phase of walking. Four trials were captured for each simulated amputation level. Data were analyzed for potentiometer-measured angles ranging from neutral (0°) to approximately 15-20° of dorsiflexion. Lastly, the forefoot marker was moved proximally, such that it was just proximal to the end of the residual foot inside the shoe, and the experiments were repeated. This change placed the toe marker proximal to the point of bending of the shoe and was done to examine the amount of error due to bending of the shoe (Figure 1B).

Results: Comparisons of the ankle angle measured using the potentiometer and the HH marker set are shown for the three simulated amputation levels in Figure 3. The ankle angles were closely approximated using the HH marker set for the metatarsophalangeal and transmetatarsal levels. However, measurement using the HH marker set at the Lisfranc level overestimated the actual ankle dorsiflexion as measured by the potentiometer. Moving the toe marker proximally greatly reduced errors in ankle dorsiflexion measurement at the Lisfranc level (Figure 4).

Figure 3 – Ankle angles measured using the Helen Hayes marker set and a potentiometer inside the mechanical model. Data are shown for each of the three simulated amputation levels, (A) Metatarsophalangeal, (B) Transmetatarsal, and (C) Lisfranc.
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**Discussion:** The results of this experiment strongly suggest that the source of ankle dorsiflexion measurement error, at least for our simulated case, was due to bending in the forefoot (Figure 1B) since movement of the toe marker would not appreciably affect errors due to “heel slippage” as shown in Figure 1A. The results shown in this abstract may vary for different partial foot devices. In particular, devices with poor suspension may lead to heel slippage and errors shown in Figure 1A. Investigators involved in collecting and reporting kinematic data of PFA should consider the appropriateness of the marker set given the type of device being studied, the level of amputation and the errors that are likely involved. In general, studies that report kinematics need to more explicitly describe the marker sets used and how markers were placed in the absence of forefoot landmarks and the presence of a prosthesis/orthosis [8]. When the device has good suspension, the errors associated with measurement of ankle dorsiflexion can probably be alleviated by moving the forefoot marker to a position just proximal to the end of the residual foot. Movement of the heel marker may also be needed to indicate an appropriate neutral alignment of the ankle. More information on this study can be found in Dillon et al. [7].

**Figure 4** – Comparison of ankle dorsiflexion measured using the potentiometer and the Helen Hayes marker set when the toe marker was moved just proximal to the end of the Lisfranc residual foot model.

**REFERENCES:**


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