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Short communication

Measured and estimated ground reaction forces for multi-segment foot models

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ABSTRACT

Accurate measurement of ground reaction forces under discrete areas of the foot is important in the development of more advanced foot models, which can improve our understanding of foot and ankle function. To overcome current equipment limitations, a few investigators have proposed combining a pressure mat with a single force platform and using a proportionality assumption to estimate subarea shear forces and free moments. In this study, two adjacent force platforms were used to evaluate the accuracy of the proportionality assumption on a three segment foot model during normal gait. Seventeen right feet were tested using a targeted walking approach, isolating two separate joints: transverse tarsal and metatarsophalangeal. Root mean square (RMS) errors in shear forces up to 6% body weight (BW) were found using the proportionality assumption, with the highest errors (peak absolute errors up to 12% BW) occurring between the forefoot and toes in terminal stance. The hallux exerted a small braking force in opposition to the propulsive force of the forefoot, which was unaccounted for by the proportionality assumption. While the assumption may be suitable for specific applications (e.g. gait analysis models), it is important to understand that some information on foot function can be lost. The results help highlight possible limitations of the assumption. Measured ensemble average subarea shear forces during normal gait are also presented for the first time.

1. Introduction

The traditional single segment foot model used in clinical gait analysis and human movement research is being replaced by more complex multiple segment models (Baker and Robb, 2006); however, the addition of kinetics to these models has been hampered in large part by equipment limitations. In particular, complete ground reaction forces (GRFs comprising normal forces, shear forces, and a free moment) applied to discrete subareas of the foot have been difficult to obtain due to insufficient spatial and temporal resolution (Davis et al., 1998; Scott and Winter, 1993).

To overcome current measurement limitations, a few investigators have proposed combining a pressure mat with a force platform and estimating subarea shear forces by assuming that they are distributed in proportion to normal forces (Abuzzahab et al., 1997; Giacomozzi and Macellari, 1997; MacWilliams et al., 2003). Yavuz et al. (2007) performed the only previously published analysis of this proportionality assumption, testing its ability to predict peak shear stresses under the forefoot in normal and diabetic gait. Using a small, custom-built array of tri-axial force sensors, they reported errors as high as 98 kPa, suggesting that these were due to local differences in plantar frictional properties and/or intrinsic muscle activities. The proportionality assumption has not been tested across the entire foot (i.e. as applied to a multi-segment foot model), where interactions between foot segments may introduce additional errors. For example, when a vertical load is applied to an arch, shear reaction forces at the bases are directed toward the center of the arch (i.e. in opposite directions) and cancel if measured by a single force platform. Similar situations may exist in the loaded foot at the medial longitudinal arch (Sarrafian, 1987) or between the metatarsals and phalanges. The purpose of the present study was to evaluate the ability of the proportionality assumption to predict shear forces and free moments under a three segment foot model during normal gait. It was hypothesized that opposing shear forces would occur between segments for which the proportionality assumption could not account. Accurate measurement of subarea GRFs will improve multi-segment and finite element foot models, and lead to increased understanding of foot and ankle pathologies.

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2. Methods

2.1. Data collection

Seventeen normal, healthy pediatric subjects (9 males, 8 females), representing a range of ages (7–18 years, mean 12.6 ± 3.4) and associated foot sizes, volunteered and signed approved consent/assent forms. The right foot of each subject was evaluated. Subjects first walked at a self-selected speed across a floor containing two adjacent AMTI OR6-7-1000 force platforms (Advanced Mechanical Technology Inc., Watertown, MA, USA). Three trials were collected during which the entire foot made contact with a single platform. Next, subjects walked using a three-step targeting approach so that specified portions of the foot contacted the adjacent platforms. This was done for two different joints: (1) transverse tarsal (TT) and (2) metatarsophalangeal (MTP). For the TT trials, the segments were divided approximately at the transverse tarsal joints, so that the rearfoot contacted the first platform while the rest of the foot contacted the second platform. For the MTP trials, the division was just distal to the metatarsal heads, with the hallux and toes isolated on the second platform. For both conditions, a foot outline was drawn on the force platforms for visual guidance, but subjects were instructed to walk as normally as possible and the starting position was adjusted until the appropriate foot placement was achieved. Numerous trials were collected for each condition until at least three were identified with accurate foot placement, which was verified by two video cameras located on either side of the platform division (Fig. 1).

2.2. Data analysis

Six-channel GRF data were analog low-pass filtered at 1050 Hz and collected at 1560 Hz. All data processing was performed using a Visual 3D software (C-Motion Inc. Germantown, MD, USA). For the targeted trials, the channels from the two force platforms were combined to create a single virtual platform. Both single and virtual platform GRF data were then filtered using a low pass Butterworth filter (100 Hz cutoff frequency), and a threshold cutoff of 5 N was applied. The virtual platform was used in conjunction with the normal force components from the individual platforms (i.e. similar to subarea forces created by integrating over a pressure mat) to estimate the shear forces and free moment on each foot segment using the proportionality assumption as per Cowley (2001) and MacWilliams et al. (2003). Curves were time-normalized to stance phase for comparisons between trials. A representative trial for each subject and each joint condition was chosen by first calculating mean GRFs across the three self-selected speed trials. Root mean square (RMS) differences were then calculated between virtual GRF components from the TT and MTP trials and the mean GRF components from the self-selected speed trials. The trial with the smallest RMS differences across all GRF components was chosen as representative for that subject and joint. Using the representative trials, RMS errors were computed between the estimated and measured GRFs over time periods in which both segments were in contact with the platform. All forces were normalized by body weight (% BW) for comparisons between subjects.

3. Results

Ensemble averages of measured shear and normal GRF components demonstrated reasonably low variability across subjects (Fig. 2). With few exceptions, errors in estimated shear forces were similar in magnitude and direction across subjects, and mean curves for both measured and estimated forces show typical errors (Fig. 3A-B). RMS errors ranged from 1% BW to 6% BW, and were greater in the MTP trials than in the TT trials (Table 1). Peak absolute errors up to 12% BW were observed in the MTP trials, anteroposterior (A/P) direction, during terminal stance. Mean measured and estimated free moments were more variable than shear and normal forces, and were not representative of all individual moment patterns (Fig. 3C). Free moment RMS errors ranged from 0.001 to 0.027 Nm/kg, with maximum errors up to 0.040 Nm/kg.

4. Discussion

Combining a pressure mat and a force platform to estimate subarea GRFs has appealed to several investigators because of its simplicity and practicality (Abuzzahab et al., 1997; Giacomozzi et al., 2006; Giacomozzi and Macellari, 1997; Giacomozzi et al., 2000; MacWilliams et al., 2003; Saraswat et al., 2010) but the underlying proportionality assumption has not been rigorously evaluated. In this study, we used two adjacent force platforms to test the accuracy of the approach as it applies to a three segment foot model.

In the A/P direction, in early to mid-stance, the proportionality assumption slightly underestimated the braking force on the rearfoot, and attributed it instead to the forefoot (Fig. 3A, left

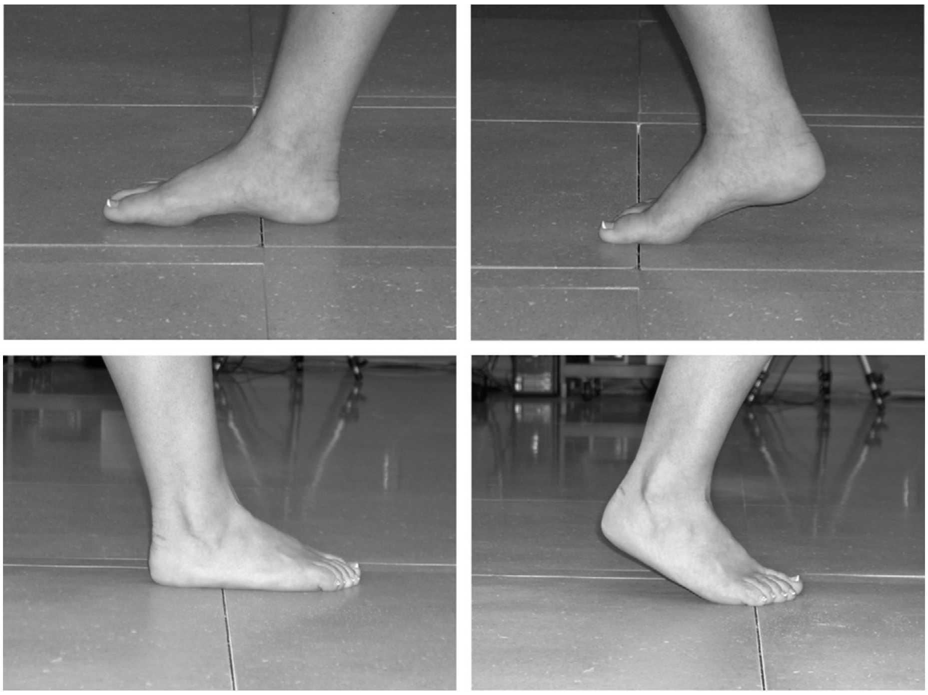

Fig. 1. Example photographs showing foot placement spanning two adjacent force platforms. Left panels: Transverse tarsal (TT) trial with foot divided approximately at the transverse tarsal joints. Right panels: Metatarsal–phalangeal (MTP) trial with foot divided between metatarsals and phalanges.

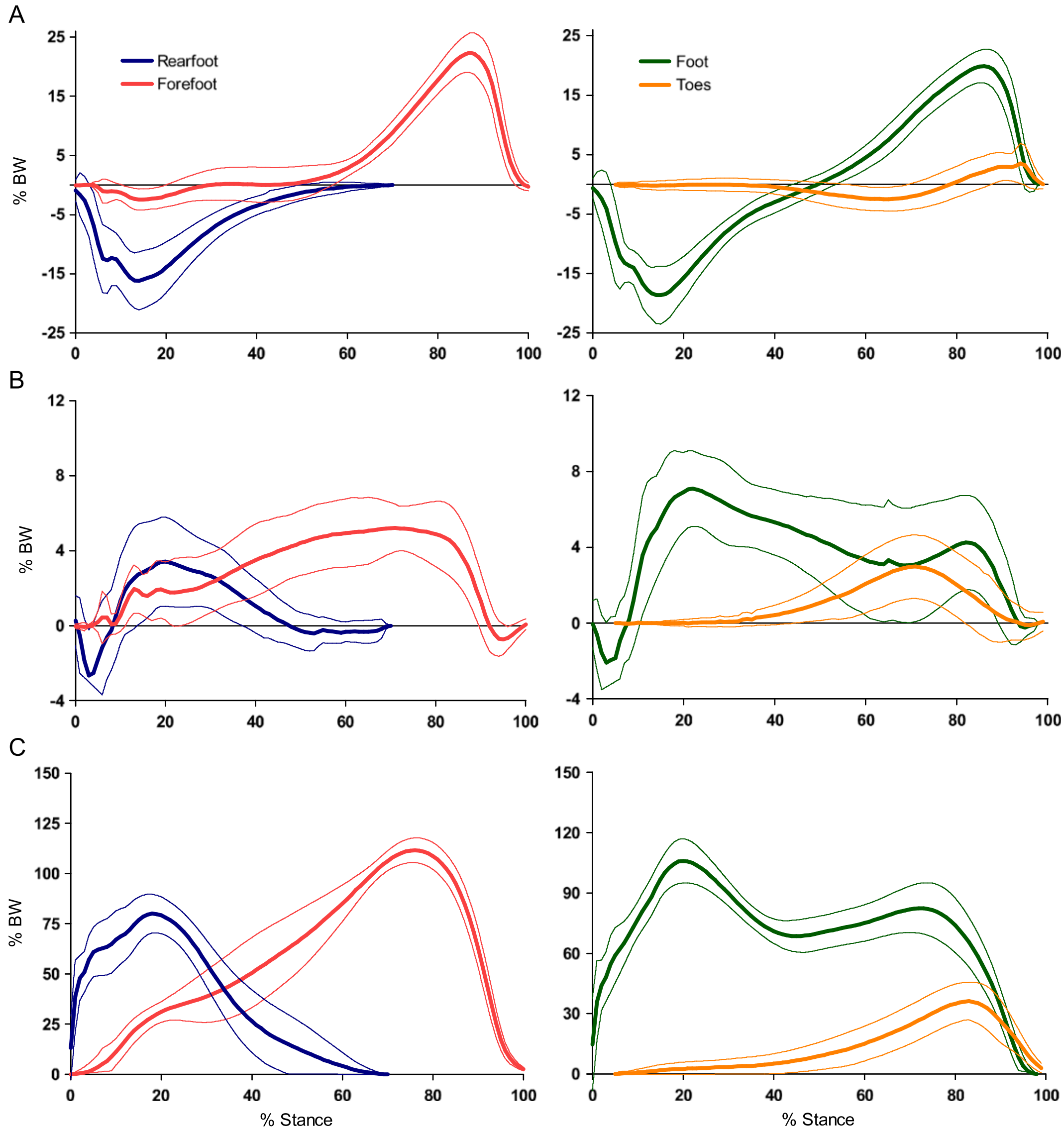

Fig. 2. Ensemble average (mean ± standard deviation) measured GRFs for each joint condition during normal, targeted gait (n=17). Graphs in the left column are from the TT trials, while graphs in the right column are from the MTP trials.

panel). Small A/P opposing shear forces were noted in mid-stance, presumably due to collapse of the medial longitudinal arch, but the effect was smaller than hypothesized and did not occur in all subjects (Fig. 2A, left panel). More pronounced A/P opposing forces were seen in late stance, as the toes exerted a small braking force during the initiation of the forefoot roll-over (third rocker). The propulsive contribution of the hallux and toes was consequently overestimated by the proportionality assumption (Fig. 3A, right panel), and peak errors up to 150% of the measured value were observed in late terminal stance.

In the mediolateral (M/L) direction, many subjects showed minor, opposing forces during mid-stance likely due to transverse plane foot rotation (Fig. 3B, left panel). Larger M/L errors were seen between the forefoot and toes throughout terminal stance, but the measured and estimated values were both directed medially (Fig. 3B, right panel), suggesting distinct differences in loading between the segments. Both measured and estimated free moment patterns (Fig. 3C) were more variable than shear forces, and estimation errors varied considerably across subjects and gait cycle. Maximum measured free moments were generally produced by the forefoot, in terminal stance (Fig. 3), and errors up to 250% of the measured values were observed.

Giaccomozi and Macellari (1997) suggested accounting for the effects of foot rotation by partitioning the total free moment among the various segments, adjusting the shear forces, and recalculating subarea free moments. This method requires that the sensors be small enough that they can be treated as points (containing only negligible free moments), and therefore cannot be applied or evaluated in this study. It is possible that Giaccomozi's method could reduce some of the errors seen in the M/L direction due to foot rotation. However, the largest free moments were generally confined to the forefoot, after the rearfoot was unloaded (Fig. 3C), and Yavuz et al. (2007) actually found greater M/L peak forefoot errors using Giaccomozi's method compared with the simple version favored by MacWilliams et al. (2003). The additional assumptions required in Giaccomozi's method may result in shear errors that are compounded, rather than reduced. No statements can be made regarding the accuracy of free moments calculated using Giaccomozi's method.

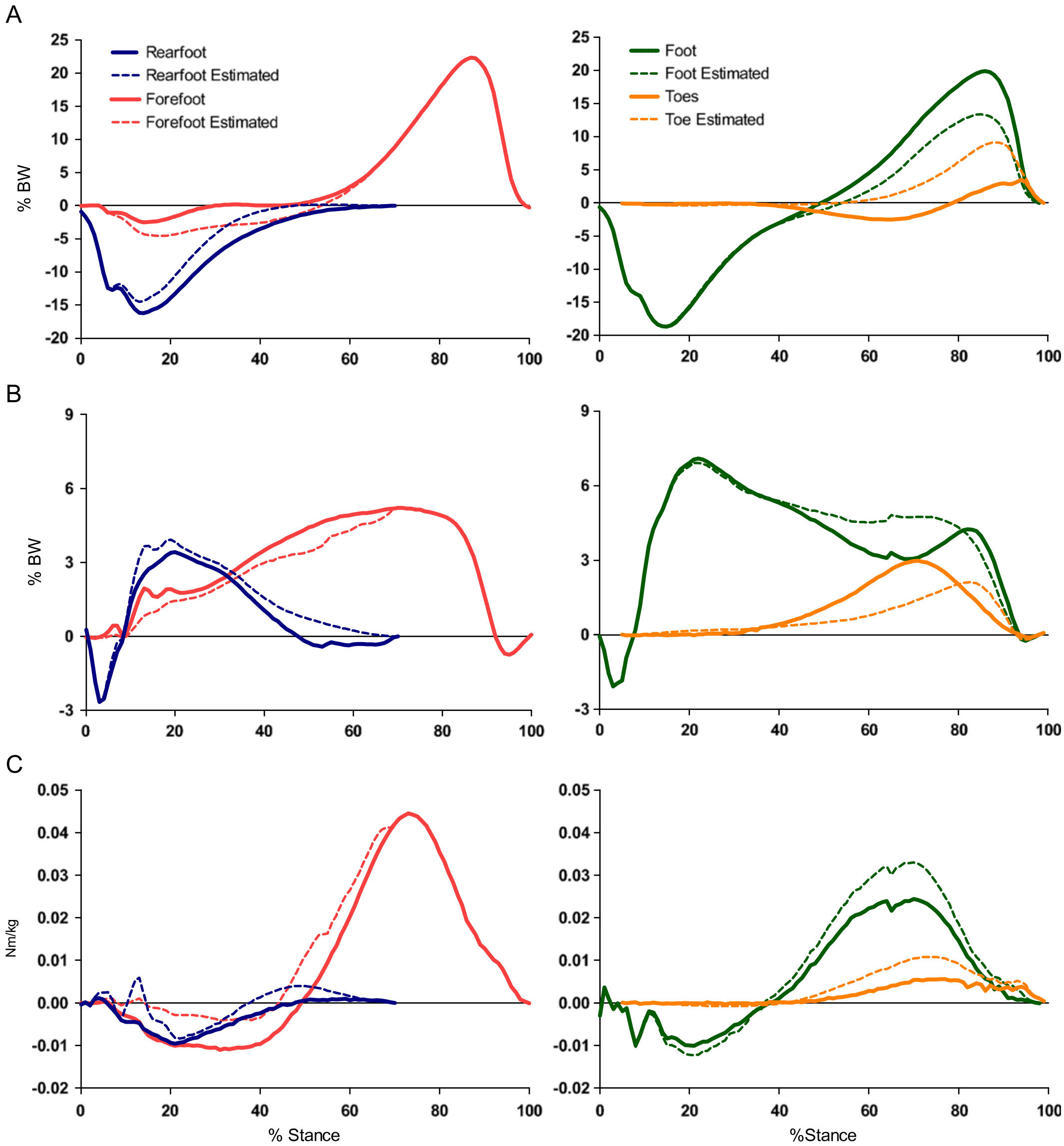

Fig. 3. Mean ($n=17$) measured (solided lines) and estimated (dashed lines) shear forces and free moments during normal, targeted gait. Graphs in the left column are from the TT trials while graphs in the right column are from the MTP trials.

The split force platform methodology limited force analysis to two of the three segments in a single trial. This allowed the rearfoot and toes to be completely isolated (although separately), but not the forefoot. However, the most important interactions between the forefoot and rearfoot occurred while the force on the toes was small, and similarly, forces on the rearfoot were small or non-existent for the important interactions between the forefoot and toes.

Targeted, or visually guided, walking was used to achieve practical subarea GRF measurements from adjacent force platforms. Previous studies have shown good agreement between targeted and non-targeted ground reaction forces (Grabiner et al., 1995; Patla et al., 1989) as well as plantar pressures (Bryant et al., 1999; McPoil et al., 1999; Meyers-Rice et al., 1994). The representative trials were also chosen to minimize differences between the targeted GRFs and the self-selected speed GRFs. Some small differences were noted, with mean vertical RMS differences less than 3% BW for the shear forces and less than 10% for the normal forces. However, all GRF patterns were of the same shape and, with very few exceptions, fell within two standard deviations of the group ensemble averages from the self-selected speed trials, suggesting that the presented forces were reasonably representative of normal gait.

Other limitations are also inherent in the use of adjacent force platforms. Foot positioning was verified visually, but not quantified. There may be some increased inter-subject variability in force measurements due to the A/P positioning of the foot relative to the force platform division. However, the foot structure is somewhat arch-like at both the TT and MTP joints, and forces under each joint are generally small (Wearing et al., 2001). The MTP joint line runs slightly diagonal to the force platforms, so the fourth and fifth toes contacted both force platforms during some trials. Again, however, the forces on the lesser toes are small in comparison to the force on the hallux (Wearing et al., 2001) and should have minimal effect on the results. There is also a small gap between platforms, but this is less than 3 mm.

Extending the error analysis from shear forces to joint moments using inverse dynamics is straightforward, but due to the need for a full model description and validation, is saved for future work. However, a few comments can be made in this regard. Because

Table 1
Estimated shear force and free moment RMS and maximum errors (mean ± standard deviation, $n=17$). It should be noted that shear force errors on adjacent segments are equal and opposite, while no such relationship exists for the free moments.

Segment	M/L shear (% BW)		A/P shear (% BW)		Free moment (Nm/kg)	
	RMS	Max	RMS	Max	RMSE	Max
Rearfoot	1.3 ± 0.6	2.3 ± 0.8	2.6 ± 1.1	3.6 ± 1.3	.009 ± .004	.021 ± .008
Forefoot	1.3 ± 0.6	2.3 ± 0.8	2.6 ± 1.1	3.6 ± 1.3	.012 ± .006	.020 ± .009
Foot	1.2 ± 0.6	2.5 ± 1.1	4.0 ± 1.6	7.8 ± 3.0	.007 ± .003	.013 ± .005
Toes	1.2 ± 0.6	2.5 ± 1.1	4.0 ± 1.6	7.8 ± 3.0	.004 ± .002	.009 ± .005

the proportionality assumption simply re-distributes the total measured GRF, joint moments proximal to the foot will generally be unaffected by distribution errors. The normal GRF component and center of pressure dominate foot joint moment calculations, and most foot joint moments will likely be only minimally affected by small errors in shear force distribution. While the method may therefore be suitable for applications such as clinical gait analysis, more dynamic activities may exacerbate the errors. Analysis of shear forces alone, particularly at the metatarsophalangeal joint(s), reveals that some information on foot function can be lost when using the proportionality assumption. In all applications, it is important to understand the associated limitations, while for more accurate measurements, a device capable of directly measuring subarea GRFs may be needed (Davis et al., 1998; Mackey and Davis, 2006).

Conflict of interest statement

None of the authors have any conflicts of interest or external funding sources to disclose.

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