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Short communication

Trailing limb angle is a surrogate for propulsive limb forces during walking post-stroke

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ABSTRACT

Background: Propulsive deficits following stroke have been attributed to reduced plantarflexion moments and a reduced trailing limb angle. We sought to determine the validity of the trailing limb angle as a surrogate measure of the anterior ground reaction force, as well as to determine the anatomical landmarks for the trailing limb angle that best approximate the orientation of the ground reaction force.

Methods: Forty-four participants with chronic stroke walked on a treadmill. At peak propulsion, we computed the trailing limb angle, the anterior orientation of the ground reaction force (the gold standard), and the hip extension angle for correlational analyses. Further, we compared the absolute error of the various trailing limb angle computations to determine which landmarks provided the most accurate representation of the anterior angle of the ground reaction force.

Findings: For the paretic and non-paretic limbs, the anterior angle of the ground reaction force was related to all measures of trailing limb angle as well as the peak propulsive force. The hip extension angle, however, was not related to the angle of the ground reaction force. Only the choice of distal landmarks significantly influenced the error between trailing limb angle and the anterior angle of the ground reaction force.

Interpretation: These data suggest that measuring the sagittal plane orientation of the entire limb serves as a suitable surrogate for measuring the anterior angle of the ground reaction force. Although greater errors were observed with kinematic measures of orientation, the magnitude of the error is likely within acceptable ranges.

1. Introduction

Paretic propulsion deficits during gait are common following stroke (Bowden et al., 2006; Mahon et al., 2015), and have been related to gait speed (Bowden et al., 2006), energy cost (Penke et al., 2018), and locomotor endurance (Awad et al., 2015). Reduced propulsion has been attributed to a diminished plantarflexor moment and a reduced trailing limb angle (Hsiao et al., 2015a). The plantarflexors provide a substantial portion of the anteriorly directed work to the center of mass (CoM) during late stance (Farris and Sawicki, 2012), but may be affected in individuals post-stroke due to weakness or activation deficits (Fimland et al., 2011; Ramsay et al., 2011). Even the provision of adequate plantarflexion moments can be ineffective, however, if the limb is not positioned appropriately (Peterson et al., 2010). Specifically, the limb must be posterior to the CoM during pushoff; otherwise, the plantarflexor moment will generate a ground reaction force (GRF) that tends to raise the CoM vertically, rather than accelerate it anteriorly.

Given the importance of limb posture for producing propulsive forces (Peterson et al., 2010), several methods of measuring limb positioning have been used. Reductions in hip extension angle are often noted, although this only accounts for a single joint. More comprehensively, the trailing limb angle (TLA) represents the orientation of the entire lower extremity. Prior investigators have used the 5th metatarsal (Genthe et al., 2018; Hsiao et al., 2015b) and the location of the center of pressure (Hsiao et al., 2015a) to represent the distal end of the limb. Likewise, the greater trochanter (Genthe et al., 2018; Hsiao et al., 2015b) and the center of the pelvis (Peterson et al., 2010) have been used to represent the proximal end of the limb. Identification of valid TLA landmarks that accurately represent the GRF angle may enable the implementation of clinical applications, (Genthe et al., 2018) without requiring force plates.

The underlying premise of the TLA is that it represents the anterior orientation of the GRF, although this has never been validated (Hsiao et al., 2015b). The purpose of this project was therefore to determine the validity of the TLA as a measure of the anterior component of the GRF in both the paretic and non-paretic limbs. A secondary purpose

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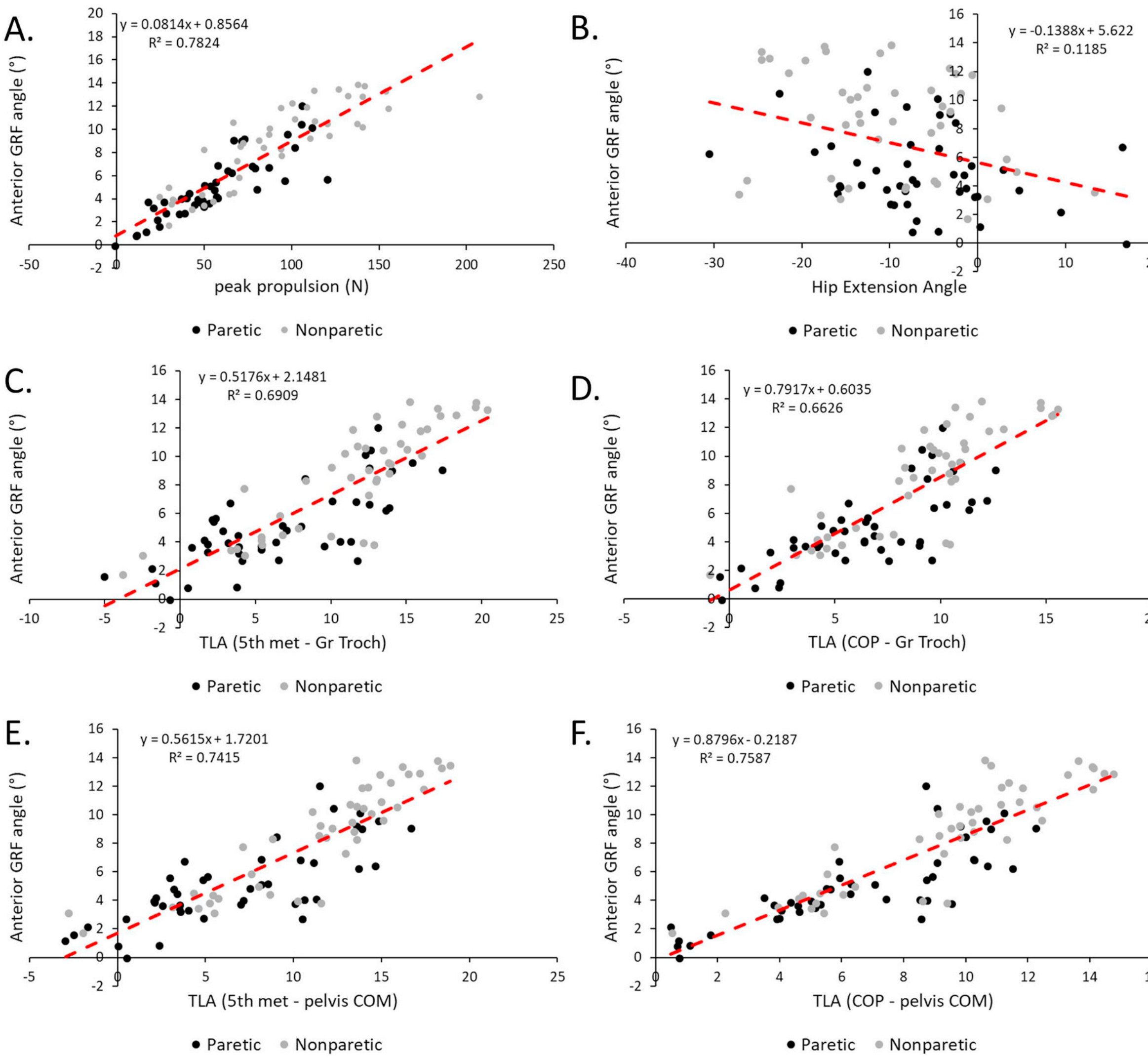

Fig. 1. Relationship between the anterior angle of the GRF and A) peak propulsive force, B) hip extension (note: hip extension here is negative), C) trailing limb angle as computed from the 5th metatarsal to the greater trochanter, D) trailing limb angle from the CoP to the greater trochanter, E) trailing limb angle from the 5th metatarsal to the pelvis CoM, and F) the trailing limb angle from the CoP to the pelvis CoM. Paretic limbs are marked with black circles and non-paretic limbs are grey circles.

was to determine the landmarks that produce the most accurate method of calculating the TLA with respect to the GRF.

2. Methods

We analyzed data from individuals with chronic hemiparesis (> 6 months) post-stroke. Each participant was able to walk independently on a treadmill and had no complaints of cardiorespiratory or musculoskeletal disorders that would preclude gait training. Additionally, there were no concomitant neurologic disorders beyond the stroke. All participants provided informed consent using a form approved by the institutional review board.

All participants walked on a dual-belt instrumented treadmill (Bertec Corp, Columbus, OH, USA). The target treadmill speed was based on each participant's overground comfortable gait speed; however, the selected speed was typically slower to allow participants to walk for up to 10 min without stopping. Only data from the second minute of walking was analyzed, to minimize effects of treadmill accommodation and fatigue. As participants walked, lower extremity kinematics were tracked with an 8-camera motion capture system (Vicon, Los Angeles, CA, USA) sampling at 120 Hz. The marker set has been described previously (Lewek et al., 2018). GRF data were sampled simultaneously from the treadmill's force plates at 1200 Hz. We used a phase-corrected, low-pass Butterworth filter to smooth marker trajectories at 6 Hz, and GRFs at 20 Hz.

As the gold standard, the sagittal angle of the GRF was computed relative to vertical throughout the stance phase (i.e., arctangent of anterior-posterior and vertical force components). The hip extension angle was computed using Visual3D (C-Motion; Bethesda, MD, USA). Finally, we modeled the limb as a vector in the sagittal plane, spanning from one of two distal points (i.e., 5th metatarsal and center of pressure [CoP]) to one of two proximal points (i.e., greater trochanter and pelvis CoM). Thus, four different TLAs were calculated relative to vertical. Finally, we identified the anterior GRF angle, hip extension angle, and each TLA at the point of peak propulsion for both the paretic and non-paretic limbs. We analyzed the time of peak propulsion to ensure a consistent event across subjects and to be consistent with prior literature (Hsiao et al., 2015b).

Statistical analyses were performed with SPSS (ver 24, IBM, Chicago, IL, USA). We used Pearson correlations to assess the relationship between peak propulsive force, the TLAs, hip extension angle, and the anterior angle of the GRF. Given the kinematic differences between limbs following stroke, analyses were performed for both limbs together as well as separately. Comparisons between limbs

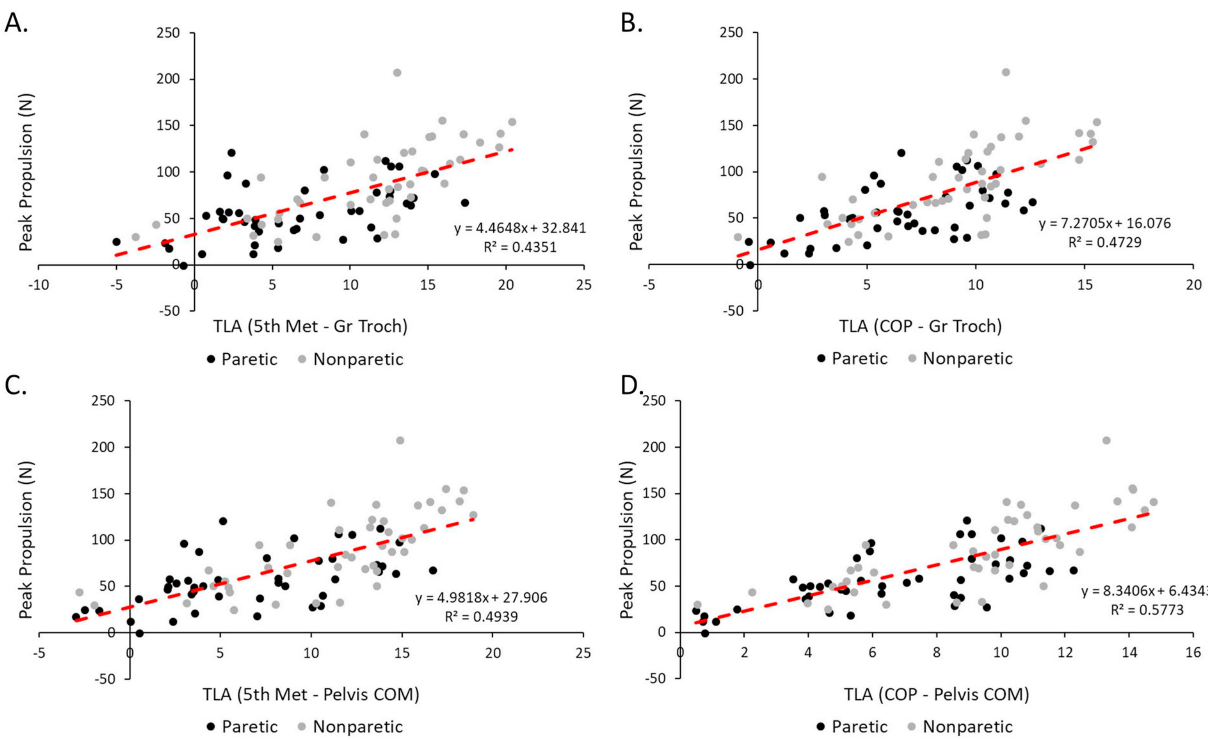

Fig. 2. Relationship between the peak propulsive force and A) the trailing limb angle as computed from the 5th metatarsal to the greater trochanter, B) trailing limb angle from the CoP to the greater trochanter, C) trailing limb angle from the 5th metatarsal to the pelvis CoM, and D) the trailing limb angle from the CoP to the pelvis CoM. Paretic limbs are marked with black circles and non-paretic limbs are grey circles.

were performed using paired samples *t*-tests, with effect sizes calculated as Cohen's d. Absolute errors were computed for each TLA computation as the absolute value of the difference between TLA and the anterior angle of the GRF. To determine which TLA computation exhibited the smallest error, we performed a 2-way repeated measures analysis of variance (repeating for two proximal and two distal landmarks).

3. Results

We analyzed data from 44 participants (21 female/23 male) of which 19 exhibited left hemiparesis and 25 exhibited right hemiparesis. Participants were a mean (standard deviation: SD) of 57 (11) years old; and a mean (SD) duration post-stroke of 50 (56) months. The comfortable overground walking speed was a mean (SD) of 0.71 (0.29) m/s, with the treadmill speed set to 0.67 (0.29) m/s ($P = 0.127$; d = 0.23).

Pooling data from both paretic and non-paretic limbs, the anterior angle of the GRF was significantly related to peak propulsive force ($P < 0.001$; $r = 0.885$), each of the calculated TLAs (all $P < 0.001$; all $r > 0.814$), and hip extension angle ($P = 0.001$; $r = 0.344$; Fig. 1). Likewise, peak propulsive forces were related to each of the TLA computations (all $P < 0.001$; all $r > 0.660$; Fig. 2) and the peak hip extension angle ($P = 0.001$; $r = 0.336$). When the paretic (all $P < 0.001$; all $r > 0.738$) and non-paretic (all $P < 0.001$; all $r > 0.827$) limbs were analyzed separately, the anterior GRF angle was related to each of the computed TLAs. Hip extension angle, however, was not related to the anterior GRF angle (paretic: $P = 0.054$; $r = 0.293$; non-paretic: $P = 0.060$; $r = 0.286$). All measures were significantly different between limbs ($P < 0.001$; Table 1).

Comparing the mean absolute error associated with the different TLA computations revealed statistically significant effects for the distal landmark ($P < 0.001$; $\eta_p^2 = 0.293$; Table 2), with calculations involving the CoP producing smaller errors (by ~1°) than calculations involving the 5th metatarsal. There was no effect, however, for the proximal landmark ($P = 0.088$; $\eta_p^2 = 0.066$).

4. Discussion

Our overall goal was to determine the validity of the TLA as a surrogate measure of the anterior orientation of the GRF during the propulsive phase of gait. Our data suggest that the TLA is highly related to the GRF orientation and that accurate estimates of propulsive force can be determined from a variety of proximal and distal landmarks.

Importantly, all TLAs were related to peak propulsion. Given the relationship between TLA and propulsive force, it is possible that propulsion can be altered by manipulating TLA (Lewek et al., 2018). Individuals post-stroke can significantly increase propulsion with visual feedback of propulsive forces (Genthe et al., 2018). Because this requires a force plate, however, we speculate that feedback of TLA may instead serve as a surrogate measure, and could help increase propulsion. This could be especially useful as a tool for integrating biofeedback to improve the effectiveness of assistive gait technologies (Takahashi et al., 2015). Toward this end, we are encouraged by our finding of high correlations within the paretic leg only.

We noted large differences in TLA between limbs. Although the TLA can be calculated from a number of landmarks, our data indicated that the mean absolute difference between these calculations was ~1°. Nevertheless, TLA calculations using the CoP produced less error and should be used when available. To compute TLA without kinetic measures, however, use of the 5th metatarsal in these calculations provides a reasonable determination of the anterior angle of the GRF.

Despite clinical emphasis placed on increasing hip extension (Ada et al., 2003) to improve propulsion, our data suggest that the hip extension angle was not a good measure of the anterior angle of the GRF. We speculate that this poor relationship is due to redundancy that allows multiple joint configurations to achieve a given limb configuration. Specifically, changes at the knee and ankle joint can influence the hip joint angle relative to the orientation of the entire limb. Although the hip extension angle may be an important source of neurophysiologic inputs (Lewek et al., 2007), we cannot recommend use of the hip extension angle as a surrogate for the anterior GRF orientation and we

Table 1
Between limb differences for outcome measures.

	Paretic limb Mean (SD)	Non-paretic limb Mean (SD)	Significance
Peak propulsion (N)	56 (30)	90 (42)	$P < 0.001$; d = 1.06
Anterior angle of GRF (°)	5.0 (2.8)	8.5 (3.7)	$P < 0.001$; d = 1.17
Hip Extension angle (°)	6.5 (8.9)	10.2 (9.1)	$P = 0.001$; d = 0.53
TLA (5th met – gr troch) (°)	6.6 (5.4)	11.3 (5.5)	$P < 0.001$; d = 1.10
TLA (5th met – pelvis CoM) (°)	6.7 (5.2)	11.4 (5.2)	$P < 0.001$; d = 1.22
TLA (CoP – gr troch) (°)	6.5 (3.5)	9.1 (3.7)	$P < 0.001$; d = 0.88
TLA (CoP – pelvis CoM) (°)	6.6 (3.3)	9.3 (3.5)	$P < 0.001$; d = 1.17

Values represent mean (SD). All values are taken from the point of peak propulsion. d = Cohen's d.

Table 2
Mean error for trailing limb angle computations.

<table>
<tr><td colspan="2"></td><td colspan="2">Proximal landmarks</td><td></td></tr>
<tr><td colspan="2"></td><td>Gr troch</td><td>Pelvic CoM</td><td></td></tr>
<tr><td rowspan="2">Distal landmarks</td><td>5^{th} met</td><td>3.4 (2.4)°</td><td>3.1 (2.4)°</td><td rowspan="2">Distal landmarks main effect: $P<0.001$; $\eta_p^2=0.293$</td></tr>
<tr><td>CoP</td><td>2.2 (1.8)°</td><td>2.0 (1.6)°</td></tr>
<tr><td colspan="2"></td><td colspan="2">Proximal landmarks main effect: $P=0.088$; $\eta_p^2=0.066$</td><td>Interaction effect: $P=0.485$; $\eta_p^2=0.011$</td></tr>
</table>

Values represent mean (SD).

encourage clinicians to consider the location of the foot relative to the pelvis during evaluation and treatment, rather than considering only the hip angle.

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