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The Relative Contribution of Ankle Moment and Trailing Limb Angle to Propulsive Force during Gait

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Abstract

A major factor for increasing walking speed is the ability to increase propulsive force. Although propulsive force has been shown to be related to ankle moment and trailing limb angle, the relative contribution of each factor to propulsive force has never been determined. The primary purpose of this study was to quantify the relative contribution of ankle moment and trailing limb angle to propulsive force for able-bodied individuals walking at different speeds. Twenty able-bodied individuals walked at their self-selected and 120% of self-selected walking speed on the treadmill. Kinematic data were collected using an 8-camera motion-capture system. A model describing the relationship between ankle moment, trailing limb angle and propulsive force was obtained through quasi-static analysis. Our main findings were that ankle moment and trailing limb angle each contributes linearly to propulsive force, and that the change in trailing limb angle contributes almost as twice as much as the change in ankle moment to the increase in propulsive force during speed modulation for able-bodied individuals. Able-bodied individuals preferentially modulate trailing limb angle more than ankle moment to increase propulsive force. Future work will determine if this control strategy can be applied to individuals poststroke.

Keywords

Gait; Propulsive force; Ankle moment; Trailing limb angle

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1. Introduction

Walking is a basic function that allows individuals to perform activities of daily living (Guralnik, Ferrucci, Simonsick, Salive, & Wallace, 1995; Guralnik et al., 1993). The ability to walk is strongly correlated with quality of life (Chiu, Chern, Shi, Chen, & Chang, 2000), and for individuals poststroke, walking ability is often mentioned as the most important goal for rehabilitation (Duncan et al., 2007). Because walking speed is a commonly used assessment of walking ability, gait rehabilitation for individuals poststroke often focuses on increasing walking speed.

It has been shown that the propulsive force, defined as the anterior component of ground reaction force (AGRF) during the latter half of stance, is correlated to walking speed in both healthy and stroke populations (Bowden, Balasubramanian, Neptune, & Kautz, 2006; Peterson, Cheng, Kautz, & Neptune, 2010). In fact, the change in paretic propulsive force is significantly correlated with changes in self-selected walking speed and fastest comfortable walking speed in stroke survivors following an intervention (Bowden, Behrman, Neptune, Gregory, & Kautz, 2013). This correlation indicates that paretic propulsive force can be modified through intervention and is related to the improvement in walking speed. Understanding the mechanism for increasing propulsive force would allow for the design of rehabilitation strategies targeting improved paretic propulsion and therefore increasing walking speed.

Previous work has shown that both lower extremity muscle strength and limb position were correlated with propulsive force in healthy and stroke populations (Peterson et al., 2010). In particular, propulsive force on the paretic side was strongly related to increased plantar flexor activity during gait in a study of individuals poststroke (Turns, Neptune, & Kautz, 2007). Additionally, the moment and power generated from the ankle plantar flexors during gait were highly associated with gait speed (Olney, Griffin, & McBride, 1994). Nadeau et al. suggested that weakness of the plantar flexors may be a critical factor that limits gait speed for individuals poststroke (Nadeau, Gravel, Arsenault, & Bourbonnais, 1999). These findings were further supported by several musculoskeletal simulation studies which showed that the force produced by the ankle plantar flexor muscles is a major contributor to walking propulsion (Knarr, Kesar, Reisman, Binder-Macleod, & Higginson, 2013; Liu, Anderson, Pandy, & Delp, 2006; McGowan, Kram, & Neptune, 2009; Nadeau et al., 1999; Neptune, Kautz, & Zajac, 2001; Turns et al., 2007).

Peterson et al. found that in addition to ankle plantar flexor force, the position of the foot relative to the body center of mass at terminal stance is an important predictor of propulsive force for able-bodied individuals and for the non-paretic leg of individuals poststroke (Peterson et al., 2010). Moreover, they suggested that the plantar flexor contribution to the propulsive force depends on the leg extension, defined as the angle between the laboratory's vertical axis and a vector from the foot's center of mass to the body's center of mass, for the paretic leg of individuals poststroke (Peterson et al., 2010). Similarly, Tyrell and colleagues found that trailing limb angle (TLA), the angle between the laboratory's vertical axis and the vector joining the greater trochanter with the fifth metatarsal head, increased significantly as

walking speed increased in stroke survivors (Tyrell, Roos, Rudolph, & Reisman, 2011). It was suggested that a better trailing limb angle at terminal stance allows a greater component of the ground reaction force to be directed anteriorly and, thus, facilitate forward propulsion (Tyrell et al., 2011). While previous studies demonstrated relationships between ankle strength and limb position with propulsive force, these relationships do not provide quantitative information about the potential impact of each of these factors on the propulsive force. That is, whether improving one variable would have greater impact on propulsive force than improving the other is unknown. More importantly, the relative contribution of increases in ankle moment and limb position to the increases in propulsive force when individuals increase their walking speed has not yet been studied.

The primary purpose of this study was to develop a mathematical model to quantify the relative contribution of ankle moment and TLA on peak propulsive force. Using a simplified model obtained through quasi-static analysis, we identified the potential impact of ankle moment and TLA on propulsive force. In addition, we determined the relative contribution of ankle moment and TLA to the change in propulsive force when walking at different speeds in able-bodied individuals. We hypothesized that: (1) ankle moment and TLA have similar impact on propulsive force and (2) when individuals walk with different speeds, the change in propulsive force is more due to the change in TLA than due to the change in ankle moment. Information from the biomechanical-based model would allow for the design of rehabilitation strategies for improving propulsive force in individuals poststroke.

2. Method

A total of 20 healthy individuals participated in this study (Table 1). Subjects were randomly divided into two equal size groups and walked at their self-selected (SS) and fast speed (FS, 120% of SS) on the treadmill. Data collected from the SS speed session of Group1 was used for model development. This model was then validated by using data from the FS session of Group1 and the SS and FS sessions of Group2. A t-test was performed to determine whether there was a significant difference between the contributions of TLA and ankle moment to increases in propulsive force. This study was approved by the Institutional Review Board of the University of Delaware and each subject provided written informed consent for participation in this study.

2.1 Experimental Procedure

Each subject walked at their SS and FS wearing a safety harness which provided no body weight support. Gait analysis was performed on an instrumented split-belt treadmill (Bertec Corp., Columbus, OH, USA) recording three dimensional forces with two embedded force plates capturing at 1080 Hz. Kinematic data was recorded with a 62 marker set and eight camera passive motion capture system that detects motion of the reflective markers at 60 Hz (Motion Analysis Corp., Santa Rosa, CA, USA). Data processing was completed using Cortex and Visual 3D (C-Motion Inc., Bethesda, MD, USA). Kinematic data were filtered using a bi-directional Butterworth low-pass filter at 6 Hz.

Heel strike (HS) was defined as the instant of the onset of the vertical ground reaction force (VGRF) and toe-off (TO) was defined as the instant as VGRF reaches zero. TLA was

defined as the angle between the laboratory's vertical axis and the vector joining the greater trochanter with the fifth metatarsal head (Figure 1). Peak AGRF was defined as the maximum AGRF during stance between the onset of the propulsion (anterior) phase of Anterior-Posterior GRFs and TO. Ankle moment was defined as the ankle plantar-flexion moment during stance. All data were averaged across strides with 120 seconds trial duration for a given speed.

2.2 Model Development

Typically the ankle moment is calculated based on GRFs and the body's kinematics and inertial properties through an inverse dynamics approach (Winter, 2005). However, to simplify the model, the effects of linear and angular acceleration were neglected as the foot is essentially static at the instant of peak AGRF (Shamaei, Sawicki, & Dollar, 2013). In addition, the weight of the foot was also neglected as it was small compared to that of the rest of the body. This quasi-static approach has been evaluated in previous studies (Wu & Ladin, 1996) and it was suggested that sufficient accuracy during stance phase can be obtained by ignoring the dynamic components (Wells, 1981). Applying this approach the ankle moment is described as (Shamaei et al., 2013):

$$M_a = F * d, \quad (1)$$

where M_a is the ankle joint moment about the medial-lateral axis of the ankle joint center, F is the ground reaction force, and d is the perpendicular distance from F to the ankle joint center (see Figure 2).

Because AGRF is the anteriorly directed component of the GRF, it can be obtained by multiplying GRF by a sine function of the angle θ between the GRF vector and the laboratory's vertical axis. Thus, F in eq. (1) can be replaced:

$$M_a = \frac{F_a}{\sin\theta} * d. \quad (2)$$

To use a more clinically meaningful variable, TLA was substituted for θ because during terminal stance TLA is closely aligned with the vector from the center of pressure (COP) to COM in the sagittal plane, as shown in Figure 1. Thus, solving for F_a , we get:

$$F_a = \frac{1}{d} M_a \sin(TLA), \quad (3)$$

Next, a coefficient C was introduced to account for $\frac{1}{d}$ and any small difference between θ and TLA; thus,

$$F_a = C M_a \sin(TLA). \quad (4)$$

The AGRF, ankle moment, and TLA at the instant of peak AGRF from Group 1's SS session were used to identify the value of C which best predicts F_a . We observed a linear

relationship between $M_a \sin(TLA)$ and F_a (Figure 3) with a slope of 7.013, which was used as C in our final model:

$$F_a = 7.013 M_a \sin(TLA). \quad (5)$$

To explain the relative contribution of TLA and ankle moment to the change in AGRF, an equation based on eq.(5) was derived as a function of the change in each parameter (see Appendix for details):

$$\frac{\Delta F_a}{F_{a1}} = \frac{\Delta \sin TLA}{\sin TLA_1} + \frac{\Delta M_a}{M_{a1}} + \frac{\Delta M_a \Delta \sin TLA}{M_{a1} \sin TLA_1}, \quad (6)$$

where “ Δ ” denotes the change from the SS to the FS session and subscript “ $_1$ ” denotes the value at the SS session. Equation (6) indicates that the percent change in propulsive force is the sum of the percent change in $\sin(TLA)$, ankle moment, and the product of the two. The last term in eq. (6) is relatively minor and negligible when the percent changes are small. Thus, the contribution of the change in $\sin(TLA)$ and ankle moment to the change in propulsive force was calculated as:

$$\% \text{ contribution } (\sin TLA) = \left(\frac{\Delta \sin TLA}{\sin TLA_1} / \frac{\Delta F_a}{F_{a1}} \right) * 100 \quad (7)$$

and

$$\% \text{ contribution } (\text{ankle moment}) = \left(\frac{\Delta M_a}{M_{a1}} / \frac{\Delta F_a}{F_{a1}} \right) * 100. \quad (8)$$

2.3 Model validation

The model was validated by using the FS data from group 1, and SS and FS data from group 2 to test the model's ability to generalize across different walking speeds and individuals. Pearson's correlation coefficients (r) were calculated by comparing the predicted to the measured peak AGRF to evaluate the ability of the model to predict propulsive force for all subjects at two different speeds. In addition, a paired T-test was used to compare the predicted to the experimental peak AGRF. Finally, the root-mean-square errors between the predicted and the measured peak AGRF were calculated. The significance level was set at an alpha of 0.05.

3. Results

The model predicted peak AGRF was the product of ankle moment, $\sin(TLA)$, and the coefficient 7.013 (see eq. (5)). Using the model obtained from Group1 SS session, we validated the model for Group1 FS session ($r^2 = 0.87$), Group2 SS session ($r^2 = 0.77$), and Group2 FS session ($r^2 = 0.86$) (see Figure 4). The root-mean-square errors were 15.77 N (11.39%) for the FS session of Group1 and 15.43 N (10.42%) for the combination of the SS

and FS sessions in Group2. No significant differences were found between the model predicted and experimental data.

Increases in peak AGRF (18.9%), TLA (12.44%), and ankle moment (6.37%) were observed when comparing data from the SS session to FS session (Table 2). The percent change in $\sin(TLA)$ is very similar to the percent change in TLA. A significant difference between the percent change in TLA and the percent change in ankle moment was observed ($p < 0.01$). Based on eq.(7) and eq.(8), 65% of the increase in peak AGRF was contributed from the increase in TLA and 33.7% of the increase in peak AGRF was contributed from the increase in ankle moment.

4. Discussion

In this study we developed a biomechanical-based model that quantifies the relative contribution of ankle moment and TLA to propulsive force. The results showed that the model using only the ankle moment and TLA had the ability to predict the peak AGRF with high accuracy. Based on the present model (eq.(5)), ankle moment and TLA each contributes linearly to AGRF. However, during speed modulation in able-bodied individuals, the increase in TLA contributed almost twice as much as the ankle moment to the increase in propulsive force.

It is worth noting that there are alternative approaches to determine the various contributors to propulsive forces. Previous studies have used forward dynamic simulation to quantify the various contributions to fore-aft center of mass accelerations (Anderson & Pandy, 2003; Hamner, Seth, & Delp, 2010; Klemetti, Steele, Moilanen, Avela, & Timonen, 2014; Steele, Seth, Hicks, Schwartz, & Delp, 2013). An advantage of the induced acceleration analysis is that it allows for predicting propulsive force by modifying only one variable at a time, which would isolate the impact of each variable to the increase in the predicted propulsive force. In addition, forward dynamic simulation may provide information about each individual muscle's contribution to the center of mass acceleration. However, for the purpose of the present study, we focused on two clinical relevant variables that are easy to interpret for clinicians: ankle moment and TLA. This study aimed to observe, rather than predict, the actual changes that occurred during speed modulation. A quasi-static approach provides a simpler solution and is capable of explaining the changes in propulsive force. Thus, we adopted the quasi-static approach to describe the relationships between ankle moment, TLA and propulsive force.

The variables included in this present biomechanical-based model, ankle moment and TLA, were similar to the variables that were previously identified as the best predictors of AGRF in regression models (Peterson et al., 2010; Peterson, Kautz, & Neptune, 2011; Roemmich, Stegemoller, & Hass, 2012). A previous study investigated ankle, knee, and hip moments during gait and found that only the ankle moment was positively related to propulsion during pre-swing at self-selected walking speed (Peterson et al., 2010). Similarly, Neptune et al. found that the contribution of ankle plantar flexors to support and forward progression dominated the contributions of the other lower extremity muscles during late stance (Neptune et al., 2001). In addition, Kepple and colleagues reported that forward progression

was produced primarily by the ankle plantar flexors with a significant assist from the knee extensors (Kepple, 1997). Interestingly, knee extension moment was also found to be related to propulsion at speeds considerably slower than the subjects' SS walking speed (0.3m/s and 0.6m/s) in able-bodied individuals (Peterson et al., 2010). Our model did not include knee moment but was able to predict the peak propulsive forces at terminal stance accurately. This could be explained by the inclusion of TLA in our model, which captures a combination of hip, knee, and ankle angle. A greater knee extension at terminal stance would yield a greater TLA and, therefore, increase propulsive force. As suggested by Siegel and colleagues in their study of mechanical energy flow, energy transfer from the ankle into the trunk is more effective with decreasing knee flexion angles (Siegel, Kepple, & Stanhope, 2004). Indeed, Sadeghi et al. found that the knee extensors contribute to propulsion by controlling knee flexion in later stance into early swing (Sadeghi, Prince, Zabjek, Sadeghi, & Labelle, 2002). Thus, the effect of knee extension moment on propulsive force at terminal stance could be captured by TLA. For the same amount of force produced, a greater proportion of the force is directed horizontally to help drive the body center of mass forward when TLA is greater.

Our biomechanical-based model provides the ability to quantify the impact of each factor on propulsive force. In the present study, the propulsive force was described as a product of ankle moment, $\sin(TLA)$, and a constant C. Based on eq.(5), ankle moment contributes linearly to AGRF. For example, our model predicts a subject with 7° TLA and 80 N-m ankle moment would have a propulsive force of 68.38 N. For this subject, a 10% increase in ankle moment with no change in TLA would increase peak AGRF by 10% to become 75.21 N. Interestingly, a 10% increase in TLA with no change in ankle moment would also increase peak AGRF to 75.17 N, suggesting the same percentage of increase in ankle moment and TLA would result in similar peak AGRF regardless of the sine function on TLA. This is because the sine function operates nearly linearly within the normal TLA range (i.e. 5°–20°). Indeed, the percent change in TLA and $\sin(TLA)$ is very close (Table 2). Thus, ankle moment and TLA each contributes linearly to propulsive force, indicating that interventions targeting either ankle plantar flexor strength or trailing limb position can be effective rehabilitation strategies for improving propulsive force. It is important to note that based on this model, a target peak AGRF can be reached by unlimited combinations of TLA and ankle moment. For the previous example with a subject with 7° TLA and 80 N-m ankle moment, if the goal is to achieve a peak AGRF of 100 N, one strategy might be to increase ankle moment from 80 N-m to 116 N-m, which is an increase of 36 N-m or 45%. However, functionally, because the subject has a relatively low TLA (i.e., the average TLA of healthy subjects in the present study was 13°), it would probably be more practical for the subject to increase TLA from 7° to 10° and increase ankle moment by only 2 N-m to achieve the target peak AGRF. Thus, because both TLA and ankle moment have the same impact on propulsive force, a rehabilitation strategy can be designed to emphasize the factor that is more modifiable based on the subject's baseline abilities.

Interestingly, although ankle moment and TLA potentially have similar impacts on propulsive force, the experimental results showed that 65% of the change in propulsive force was due to the change in TLA while 33.7% of the change in propulsive force was due to the

change in ankle moment during healthy gait speed modulation. Previous studies of human locomotion have suggested that humans tend to adopt motor behavior that minimizes the metabolic cost (Bertram, 2005; Zarrugh, Todd, & Ralston, 1974). Thus, one of the reasons why our subjects produced greater changes in TLA may be because it is physiologically advantageous. However, we do not have evidence to support this statement in the present study. Thus, investigation is needed to determine whether modifying TLA would be more metabolically efficient strategy compared with increasing ankle moment.

Recently, there have been a number of studies that have suggested that increasing TLA is important for improving walking speed for individuals poststroke. Tyrell et al. suggested that TLA is biomechanically important for forward propulsion of the body and treadmill training at a speed faster than self-selected speed could produce beneficial changes in TLA for individuals poststroke (Tyrell et al., 2011). Mulroy et al. recommended emphasizing hip extension in the late stance during body weight support treadmill training and training at increased walking speeds to facilitate more rapid and appropriately phased muscle activation for individuals poststroke (Mulroy et al., 2010). Emphasizing paretic hip and knee extension for increasing paretic propulsion was also suggested by Peterson et al. (Peterson et al., 2010). Thus improving TLA by attempting to increase hip and knee extension are becoming important strategies in stroke rehabilitation.

There are limitations of this study that should be considered when interpreting the results. First, our results only explained the mechanism of increasing propulsive force within a limited speed range (SS to 120% SS). It is possible that the relative contribution of TLA versus ankle moment to the increase in propulsive force will change at a different speed range due to functional limitations (e.g. range of motion, ankle plantar flexor muscle strength). Second, although TLA and ankle moment are mathematically separable variables, they may not be physically independent. For example, changes in TLA resulting from changes in knee or ankle joint angles can alter ankle plantar flexors' muscle length and therefore change ankle moment without changing muscle excitation. Thus, TLA and ankle moment can potentially impact each other. However, we did not find a direct correlation between TLA and ankle moment at either walking speed or during speed modulation in the present study. Third, the root-mean-square error of 11% indicated that factors other than ankle moment and TLA do affect AGRF. One source of error in the present model could be due to the quasi-static approach, which neglected the effects of foot linear and angular accelerations and the weight of the foot segment. However, previous literature has shown that this error is negligible for estimating ankle moment (Wells, 1981). Another source of error in our model could come from the coefficient, C , which was associated with the perpendicular distance from the force vector to the ankle joint (d) and the offset in alignment between θ and TLA . The distance, d , was related to the position of the center of pressure (COP), which could vary among individuals. Thus, using a constant to account for this distance could induce error in the model. For able-bodied individuals, the COP is close to the second metatarsal head at terminal stance (Rodgers, 1988). In theory, if the offset in alignment between θ and TLA is small, C should be close to $1/d$. Indeed, the coefficient in our model was 7.013, suggesting a corresponding d value of 0.14m, which was within a reasonable range considering the distance between the second metatarsal head and the ankle

joint. The results from model validation suggested that this coefficient was generalizable and most of the variability in propulsive force can be accounted by ankle moment and TLA in able-bodied individuals. Future studies applying the present model to individuals poststroke will help determine whether other variables (i.e., COP position) should be considered for modeling pathological gait.

5. Conclusions

This is the first study that quantified the relative contribution of ankle moment and TLA to propulsive force. Applying a biomechanical-based model, the present results showed that while ankle moment and TLA both contribute linearly to AGRF, the increase in TLA contributes almost twice as much as the ankle moment to the increase in propulsive force during speed modulation for able-bodied individuals. Future work is needed to examine whether these metrics can be applied to poststroke individuals.

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Appendix

$$\begin{aligned}\Delta F_{r,a} &= F_{r,a2} - F_{r,a1} \\ &= 7.0133 M_{a2} \sin TLA_2 - 7.0133 M_{a1} \sin TLA_1\end{aligned}$$

Let $\sin TLA_2 = \sin TLA_1 + \Delta \sin TLA$ and $M_{a2} = M_{a1} + \Delta M_a$

$$\begin{aligned}\Delta F_{r,a} &= 7.0133 [(M_{a1} + \Delta M_a) (\sin TLA_1 + \Delta \sin TLA) - M_{a1} \sin TLA_1] \\ &= 7.0133 [M_{a1} \Delta \sin TLA + \Delta M_a \sin TLA_1 + \Delta M_a \Delta \sin TLA] \\ \frac{\Delta F_{r,a}}{F_{r,a1}} &= \frac{7.0133 [M_{a1} \Delta \sin TLA + \Delta M_a \sin TLA_1 + \Delta M_a \Delta \sin TLA]}{7.0133 M_{a1} \sin TLA_1} \\ \frac{\Delta F_{r,a}}{F_{r,a1}} &= \frac{\Delta \sin TLA}{\sin TLA_1} + \frac{\Delta M_a}{M_{a1}} + \frac{\Delta M_a \Delta \sin TLA}{M_{a1} \sin TLA_1}\end{aligned}$$

Research Highlight

1. We determine the contribution of ankle moment and trailing limb angle to propulsion.
2. Ankle moment and trailing limb angle each contributes linearly to propulsion.
3. Trailing limb angle is the major contributor to increase propulsion during gait.

Figure 1.

Illustration of variables. (A) Trailing limb angle was defined as the angle between the laboratory's vertical axis and a vector created between the greater trochanter and the fifth metatarsal head. (B) From top to bottom: anterior-posterior GRF, ankle moment, and TLA. HS and TO represent the events of heel strike and toe-off, respectively. The vertical line indicates the instant of peak AGRF, defined as the maximum AGRF during stance between the onset of the propulsion (anterior) phase of anterior-posterior GRF and toe-off.

Figure 2.

Free-body diagram of the rigid body describing the foot. M_a is the ankle joint moment about the medial-lateral axis of the ankle joint center, F is the ground reaction force, F_a is the anterior ground reaction force, θ is the angle between the F vector and the laboratory's vertical axis, and d is the perpendicular distance from F to the ankle joint center.

Figure 3.

The relationship between $M_a \sin(TLA)$ and F_a . The slope 7.013 was then used as C in our model.

Figure 4.

Relationships between the measured and predicted peak AGRF. (a) Group1, FS session. (b) Group2, SS session. (c) Group2, FS session.

Table 1

Subject information

	Age (yrs)	SS Speed (m/s)	FS Speed (m/s)	Height (m)	Weight (kg)
	Group 1				
Mean	28.7	1.08	1.3	1.65	62.68
SD	17.2	0.19	0.23	0.11	15.08
	Group 2				
Mean	38.2	1.08	1.3	1.68	68.11
SD	23.1	0.16	0.19	0.1	10.47

Table 2

Mean values of measurement variables for all subjects. *% change* was calculated as $\frac{(FS-SS)}{SS} \%$

	SS mean (SD)	FS mean (SD)	% change
<i>TLA</i> (°)	13.5 (2.69)	15.18 (2.58)	12.44
<i>sin(TLA)</i>	0.23 (0.05)	0.26 (0.04)	12.45
Ankle Moment (N-m)	75.93 (21.8)	80.8 (22.57)	6.37
AGRF (N)	125.23 (38.37)	149 (44.13)	18.9