Foot dissipation during ankle push-off: human walking insights from a multiarticular EMG-driven musculoskeletal model

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1 Introduction

Humans perform a burst of push-off work with their trailing limb during the step-to-step transition in walking. This push-off helps redirect the body's center-of-mass, reducing collisional energy losses after heelstrike and enabling economical gait [1]. The majority of this positive push-off work is performed by muscles and tendons that cross the ankle joint [2]. However, the foot performs negative work during this phase of gait [3], which counteracts a fraction of the positive push-off work of the ankle; thus reducing the work done on the body's center-of-mass and potentially degrading gait economy. These opposing ankle and foot behaviors are difficult to reconcile with previous findings on energy-saving mechanisms used during gait, and dynamic walking principles emphasizing the importance of center-of-mass push-off [1, 2].

This study sought to investigate one plausible explanation for the enigmatic ankle and foot behaviors during push-off. We hypothesized that our current understanding of ankle and foot function is obscured by methodological limitations in commonly-used biomechanical estimates. A key assumption in inverse dynamics and other segment-based kinetics estimates [3] is that joint moments and powers originate from monoarticular sources (e.g., muscle-tendon units). These biomechanical estimates do not account for multiarticular muscles, such as the flexor digitorum and hallucis longus muscles, which articulate across the ankle and toe (metatarsophalangeal) joints. We predict that a failure to account for multiarticular muscles may skew current estimates of biomechanical work and power at individual body joints (e.g., ankle) and segments (e.g., foot), which could potentially alter our interpretation of the interplay between the ankle and foot during push-off.

The purpose of this study was two-fold. First, we sought to develop a data-driven musculoskeletal model of the ankle and foot that allowed us to approximate muscle-specific contributions to gait. Second, we aimed to estimate the magnitude of work and power errors that may be inherent in conventional biomechanical estimates which neglect multiarticular muscles, and determine if these errors might impact our current interpretation of ankle and foot function during push-off in walking.

2 Methods

Data collection and analysis

Three healthy male subjects (24±5 years, 88±14 kg, height: 1.8±0.1 m) participated in this gait analysis study. All subjects gave informed consent to the protocol, as reviewed by the Institutional Review Board at Vanderbilt University. Subjects walked at four speeds: 0.75, 1.00, 1.25 and 1.50 m/s on a split-belt instrumented treadmill (Bertec). Six degree-of-freedom kinematics and kinetics of the shank and foot were derived from a ten camera motion tracking system (Vicon) in conjunction with post-processing via C-Motion Visual3D. Prior to walking, four surface EMG sensors (Delsys) were placed on muscles that contribute to ankle plantarflexion: soleus, medial gastrocnemius, lateral gastrocnemius, and flexor digitorum and hallucis longus (FDHL, measured together due to limitations in surface EMG). Marker and force data was filtered at 10 Hz and 25 Hz, respectively. EMG data was demeaned, high-pass filtered at 150 Hz [4], rectified, and low-pass filtered at 10 Hz. All filters used were 3rd order, zero-lag Butterworth filters.

Musculoskeletal model

We developed a sagittal plane musculoskeletal model in order to estimate the individual ankle and foot muscle contributions to push-off during walking. To do so, we used a simple EMG to force mapping algorithm and integrated it with conventional gait analysis measurements. Below we briefly summarize our model.

We estimated muscle force, F_m, (eqn. 1) for each muscle (m). Each EMG profile was normalized to its peak activation magnitude during maximal contraction to obtain EMG_m. This profile was scaled by the muscle physiological cross sectional area, $PCSA_m$, to account for size differences between muscles. The pennation angle, θ_m, was incorporated into the model to account for muscle fiber direction, assuming a constant θ_m for each muscle.

$$F_m \propto PCSA_m * \cos(\theta_m) * EMG_m \qquad \text{[Eqn. 1]}$$

We then estimated the moment, M_m, (eqn. 2) that each muscle force created about the ankle using plantarflexion moment arms, r_m, from literature.

$$M_m = r_m \times F_m \qquad \text{[Eqn. 2]}$$

We next calculated the power from each muscle, by multiplying muscle moment with the angular velocity of the ankle joint, ω_{ank}. We summed the individual muscle powers to obtain the total power about the ankle, P_{ank}, (eqn. 3).

$$P_{ank} = \sum M_m * \omega_{ank} \qquad \text{[Eqn. 3]}$$

To ensure that P_{ank} computed here was directly comparable to inverse dynamics based power estimates we incorporated a time lag and an additional scaling factor.

The time lag accounted for the delay between muscle activity and mechanical force production; the scaling factor facilitated the mapping of EMG (μV) to force (N). Using this model we could then parse P_{ank} into contributions from each individual muscle-tendon unit.

To estimate the maximum potential errors in conventional ankle and foot power calculations, we assumed the multiarticular FDHL muscle-tendon units performed no mechanical work. In other words, we assumed the multiarticular FDHL acted isometrically during push-off, effectively like a cable across the ankle and toe joints. Using our model we then computed how much the FDHL contributed to the conventional ankle and foot power calculations (i.e., the inaccuracy error inherent in these measures if the multiarticular muscle were acting isometrically). Subtracting these multiarticular power contributions from the conventional ankle and foot estimates yielded updated estimates for ankle and foot power, which may better reflect the underlying physiology.

3 Results

We found that our EMG-driven musculoskeletal model was able to reproduce inverse dynamics based sagittal plane ankle power with high fidelity ($R^2 = 0.98 \pm 0.01$).

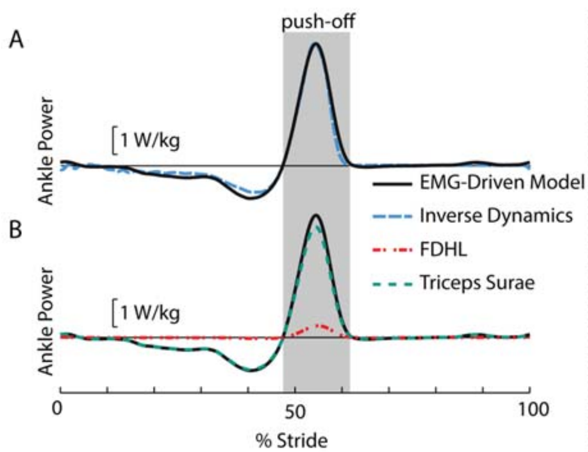

Figure 1: Ankle power at 1.25 m/s for one subject. (A) Musculoskeletal model (EMG-Driven Model) power reproduced the sagittal plane inverse dynamics estimate. (B) Muscle group contributions to plantarflexion power. The FDHL is the multiarticular muscle tendon unit acting about both the ankle and toe joints.

The model then enabled us to estimate muscle-specific contributions (Fig. 1), including contributions from multiarticular muscles. These muscle-specific estimates were also in good agreement with the power profiles reported in prior literature [5], which indicate the soleus (one of the three muscles in the triceps surae) as the primary contributor.

Next, we estimated the errors that might results from neglecting the multiarticular ankle-toe (FDHL) muscles in inverse dynamics. We found that inverse dynamics may over-estimate sagittal ankle work by about 1.9 ± 0.9 J (Fig. 1B: area under FDHL power curve), or about 6% of ankle push-off work, at 1.25 m/s.

We also discovered that the foot may not dissipate as much energy as previously estimated (Fig. 2). On average, we found -9.3 vs. -10.5 J (updated vs. conventional estimate) of negative foot work during push-off, and 2.2 vs. 1.5 J of positive foot work during push-off at 1.25 m/s.

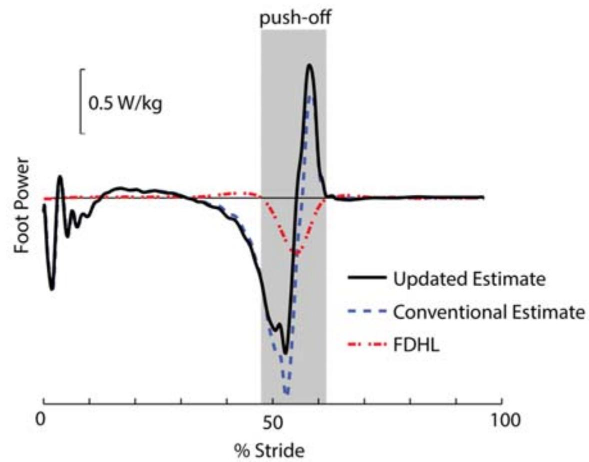

Figure 2: Updated estimate of foot power: the multiarticular muscle tendon unit power (FDHL assuming isometric contractions) subtracted from the Conventional power estimate based on a deformable foot model [3].

4 Discussion

The musculoskeletal model presented provides an estimate of muscle-specific power contributions to ankle plantarflexion push-off during walking, which can then be used to account for multiarticular muscle function. Preliminary findings suggest that positive ankle work may be over-estimated, and positive foot work under-estimated, by about 1.9 J. Rather than purely dissipating energy during push-off, these findings suggest that the foot may also undergo a cycle of viscoelastic energy storage followed by energy return, which may have implications for prosthetic foot design. However, the foot still appears to absorb more energy (~9 J) during push-off than it returns as positive work (~2 J) in terminal stance. This simple musculoskeletal modeling approach provides an updated estimate of ankle foot kinetics during gait, and could also be applied to other joints to improve our fundamental understanding of biological movement.

References

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