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Determination of ankle muscle power in normal gait using an EMG-to-force processing approach

R.A. Bogey [a,b,*], A.J. Gitter [c], L.A. Barnes [d]

[a] Rehabilitation Institute of Chicago, Chicago, IL 60611, United States
[b] Northwestern University Feinberg School of Medicine, Chicago, IL, United States
[c] Department of Rehabilitation Medicine, University of Texas at San Antonio, San Antonio, TX, United States
[d] B&L Engineering, Tustin, CA 92780, United States

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ABSTRACT

The purpose of this study was to determine the contribution of individual ankle muscles to the net ankle power and to examine each muscle's role in propulsion or support of the body during normal, self-selected-speed walking. An EMG-to-force processing (EFP) model was developed which scaled muscle-tendon unit force output to gait EMG, with that muscle's power output being the product of muscle force and contraction velocity. Net EFP power was determined by summing individual ankle muscle power. Net ankle power was also calculated for these subjects via inverse dynamics. Closeness of fit of the power curves of the two methods was used to validate the model. The curves were highly correlated ($r^2 = .91$), thus the model was deconstructed to analyze the power contribution and role of each ankle muscle during normal gait. Key findings were that the plantar flexors control tibial rotation in single support, and act to propel the entire limb into swing phase. The dorsiflexors provide positive power for swing phase foot clearance, negative power to control early stance phase foot placement, and a second positive power burst to actively advance the tibia in the transition from double to single support. Co-contraction of agonists and antagonists was limited to only a small percentage of the gait cycle.

1. Introduction

The role of individual ankle muscles during normal, self-selected speed gait is controversial (Perry, 1992; White and Winter, 1992). This lack of consensus is important, as rehabilitation protocols are directed towards recovery of near-normal motion when possible. Our understanding of the factors that contribute to normal locomotion would be advanced if the force and power, and thus their role could be determined *in vivo*.

Human muscle forces at the Achilles tendon have been recorded by placing force transducers directly on muscle-tendons (Komi et al., 1984, 1987). This type of assessment is neither easy nor practical for clinical use. The number of muscles that cross a joint are greater than the number of solution set equations (mathematically indeterminate). Since it is difficult to directly measure muscle forces, other approaches have been used. Muscle force estimates have been made via an electromyogram-driven model (Bogey et al., 2005; Heintz and Gutierrez-Farewik, 2007; Hof and van den Berg, 1978, 1981a,b,c,d) and via optimization theory (Chao and Rim, 1973; Davy and Audu, 1987; Hatze, 2000; Heintz and Gutierrez-Farewik, 2007; Patriarco et al., 1981). The first approach has the advantage of providing a unique solution set. Another technique is to examine synergistic muscle groups. Muscle force approximations are inferred from joint moments. The joint moment yields important information regarding the magnitude of muscle activity. With small changes in muscle moment arms – generally the case in human moment – the joint moment changes in parallel with muscle forces. Yet the type of muscle contraction – eccentric, concentric, isometric – cannot be determined from the joint moments.

Joint power may be used to establish the capability of muscle groups to generate or restrain movement (Sadeghi et al., 2002; Siegel et al., 2004). Joint power seems to be a valid indicator of a person's ability to control their limbs (Olney et al., 1994; Vardaxis et al., 1998). Joint power and its relationship to muscle work during activities such as normal adult gait (Norris et al., 2007; Sadeghi et al., 2001; Winter et al., 1977), amputee gait (Gitter et al., 1991; Powers et al., 1994) and after central nervous system disease (Olney et al., 1991) have been examined.

The power calculation approaches, described above, are based on inverse dynamics techniques. This method solves the actuator-redundancy problem. However, conventional power analysis cannot define the unique power contribution of a single muscle, except in those rare occasions where only a single muscle is responsible for the observed movement. The presence of co-contraction further confounds the analysis. Also, conventional power

* Corresponding author. Address: Northwestern University Feinberg School of Medicine, Chicago, IL 60611, United States. Tel.: +1 312 238 2879, fax: +1 312 238 4516.
E-mail address: rbogey@ric.org (R.A. Bogey).

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analysis methods assume that joint power accurately reflects the magnitude and type of muscle contraction during a purposeful movement. However, it is difficult to define with certainty the role, if any, of a muscle when there is no joint power. Non-zero power suggests the presence of eccentric (negative power) or concentric muscle contractions (positive power). Previous studies of amputee locomotion (Czerniecki et al., 1990) demonstrate non-zero power at the prosthetic ankle. These factors show that the muscle power estimates from inverse dynamics can include muscle force, joint angular velocities and other, undefined variables.

The purpose of this study was to determine the contribution to the ankle joint power generated solely by individual muscles, and by extension to give insight into the unique role of each muscle during normal, self-selected-speed walking.

2. Methods – overview

Ankle power values at self-selected speed walking were determined by (i) inverse dynamics and (ii) an EMG-to-force processing (EFP) model during the same gait trial for each study participant. Net power for each method was statistically compared at one percent gait cycle intervals. Closeness of fit between the EFP and inverse dynamics power curves was used to validate the EFP model. If the resulting power curves were sufficiently similar, the EFP power estimates were deconstructed to analyze the power contribution and role of individual ankle muscles in normal gait.

3. Subjects

Eighteen adult males with no history of neuromusculoskeletal disease were recruited to participate in the study. Two subjects did not participate in repeat gait trials, and their data were not further examined. The sixteen remaining study participants had a mean age of 27 ± 3.2 years (range 23–34 years), and a mean mass of 73.9 kg (±6.6 kg). Due to the comprehensive number of muscles examined each study participant performed two self-selected speed walking trials, with data selected from each trial. In normal adults the examined gait parameters are highly consistent across trials (Bogey et al., 2003), and the use of multiple sessions did not confound the results. The mean gait velocity for these subjects was 82.0 ± 3.6 m/min. Subjects consented to participate following explanation of the procedure and review of the informed consent, as approved by the Institute Review Board, and signed the Rights of Human Subjects form.

3.1. Kinematics and kinetics (KIN)

Gait data acquisition and processing steps used in this study have been described in greater detail elsewhere (Bogey et al., 2003). The primary focus of this analysis was the net power generated at the ankle through the entire gait cycle. Kinematic, kinetic, footswitch and EMG data were simultaneous acquired while study participants walked at their preferred, self-selected speed. Study participants began each trial standing at the end of the 12 m walkway. After a few strides they had reached their self-selected walking speed, which they maintained across the middle 6 m of the walkway. A few strides were needed to decelerate at the end of the walkway. As our goal was to examine "typical" (i.e. constant velocity) gait the data analysis was thus limited to the portion of the gait trial where study participants were within the middle 6 m of the walkway. Study participants wore appropriate clothing, consisting of a light shirt, shorts and flat-soled shoes. Contact-closing footswitches were placed in the participant's shoes during the trials to determine stance and swing times. Twenty round reflective markers were placed over anatomic landmarks to determine joint centers (Kadaba et al., 1989).

Study participants performed several preliminary trails to become acclimated to the test environment. As normal adult gait is essentially symmetric data analysis was restricted to the right leg, only. Data were averaged across five typical trials for each study participant.

Motion data were sampled at 100 Hz with an eight camera system (Motion Analysis Corporation Model Hawk). Marker coordinates were bi-directionally smoothed with a fourth-order Butterworth filter with an effective cutoff frequency of 6 Hz. Linear velocities and accelerations, and angular position, velocity, and acceleration of segments were determined as described elsewhere (Gitter et al., 1991). Ground reaction force (GRF) data was collected at 600 Hz with paired (AMTI type OR6-7) force platforms. GRF data was smoothed, the center of pressure location was determined (Gitter et al., 1991), and motion and ground reaction force data were temporally matched. Mass and center of mass locations were determined (Dempster, 1955). Three-dimensional joint moments and power were computed at the ankle (plus knee and hip) via Newtonian mechanics using in-house software.

Mechanical power was based on the kinetics and kinematics of the individual body segments, allowing no transfer of energy between segments. While mechanical power can include the transfer of energy between limb segments and the trunk (Unnithan et al., 1999), this constraint does not substantially affect ankle power and is consistent with the accepted convention. Ankle power (P^{KIN}) was defined as dot product of the ankle moment and angular velocity ($P^{KIN} = M^{KIN} \cdot \omega^{KIN}$). Joint position data were also used to determine muscle lengths via a force processor model (*below*).

4. EMG-to-force processing – force processor

The neuromusculoskeletal model included lower extremity skeletal structures plus twelve muscles crossing the ankle. Full details of the force processor model are presented elsewhere (Delp et al., 1990). The model consisted of three-dimensional representations of the bones and muscle–tendon paths, kinematic descriptions of the ankle and knee, and a nominal biomechanical model of each musculotendinous unit (MTU).

The isometric force generating properties of each of the twelve muscles were derived by scaling a Hill-based model (Hill, 1938). The scaled model was matched to each study participant's unique limb segment-mass and segment-length characteristics. The effect of muscle force–velocity relations, muscle contraction history and contraction type were considered (Bogey et al., 2005).

5. METHODS: EMG-to-force processing – dynamic electromyography (EMG)

Electromyographic (EMG) was obtained from 12 muscles crossing the ankle – soleus (SOL), medial and lateral gastrocnemius (MGAST, LGAST), posterior tibialis (POSTTIB), flexor digitorum longus (FDL), flexor hallucis longus (FHL), peroneus brevis (PB), peroneus longus (PL), peroneus tertius (PTERT), anterior tibialis (ANTTIB), extensor digitorum longus (EDL), and extensor hallucis longus (EHL). Electromyographic activity was recorded with bipolar 50 μm stainless steel wire electrodes insulated with polyimide except for 2 mm exposed tips. The electrodes were inserted into the tested muscles with a 20-gauge hypodermic needle using the techniques initially described by Basmajian and Stedio (1962). Insertion sites for each muscle were as described by Delagi et al. (1994). A single ground electrode was placed over a bony landmark on the lower extremity (i.e. greater trochanter). Electrode placement was confirmed by electrical stimulation of the muscle via the indwelling electrode, and by voluntary muscle contraction.