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Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds

Samuel R. Hamner [a], Scott L. Delp [a,b,*]

[a] Department of Mechanical Engineering, Stanford University, USA
[b] Department of Bioengineering, Stanford University, USA

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ABSTRACT

Running is a bouncing gait in which the body mass center slows and lowers during the first half of the stance phase; the mass center is then accelerated forward and upward into flight during the second half of the stance phase. Muscle-driven simulations can be analyzed to determine how muscle forces accelerate the body mass center. However, muscle-driven simulations of running at different speeds have not been previously developed, and it remains unclear how muscle forces modulate mass center accelerations at different running speeds. Thus, to examine how muscles generate accelerations of the body mass center, we created three-dimensional muscle-driven simulations of ten subjects running at 2.0, 3.0, 4.0, and 5.0 m/s. An induced acceleration analysis determined the contribution of each muscle to mass center accelerations. Our simulations included arms, allowing us to investigate the contributions of arm motion to running dynamics. Analysis of the simulations revealed that soleus provides the greatest upward mass center acceleration at all running speeds; soleus generates a peak upward acceleration of 19.8 m/s^2 (i.e., the equivalent of approximately 2.0 bodyweights of ground reaction force) at 5.0 m/s. Soleus also provided the greatest contribution to forward mass center acceleration, which increased from 2.5 m/s^2 at 2.0 m/s to 4.0 m/s^2 at 5.0 m/s. At faster running speeds, greater velocity of the legs produced larger angular momentum about the vertical axis passing through the body mass center; angular momentum about this vertical axis from arm swing simultaneously increased to counterbalance the legs. We provide open-access to data and simulations from this study for further analysis in OpenSim at simtk.org/home/nmbl_running, enabling muscle actions during running to be studied in unprecedented detail.

1. Introduction

As runners increase their speed, the magnitude of forces acting on their bodies increases. Researchers have observed changes in ground reaction forces, joint moments, muscle activities, leg stiffness, and body segment motions at different running speeds (e.g., Cappellini et al., 2006; Cavagna et al., 1976; McClay et al., 1990; McMahon and Cheng, 1990; Novacheck, 1998; Schache et al., 2011; Winter, 1983). Analysis of body segment motions and ground reaction forces during running has revealed that runners increase their forward speed by increasing their stride length and stride frequency (Cavagna et al., 1988; Hildebrand, 1960). At running speeds between 2 and 7 m/s, runners increase their stride length by generating larger ground reaction forces (Derrick et al., 1998; Mercer et al., 2005; Weyand et al., 2000). These experimental studies have characterized the larger ground reaction forces runners produce as they run faster, yet it remains unclear which muscles contribute to the production of larger ground reaction forces as running speed increases.

Musculoskeletal simulations enable examination of how muscles produce ground reaction forces. Muscles generate forces that are transmitted by bones and connective tissue to other body segments, causing the foot to apply a force to the ground. The ground applies an equal and opposite reaction force to each foot, which accelerates the mass center forward (i.e., propulsion), backward (i.e., braking), and upward (i.e., support). The mass center acceleration is equal to the ground reaction force divided by the subject's total body mass (Winter, 1990). Sasaki and Neptune (2006) used two-dimensional simulations to highlight a change in soleus function at the walk-run transition speed. Besier et al. (2009) estimated muscle forces during running with an electromyography-driven musculoskeletal model to characterize quadriceps forces in subjects with patellafemoral pain. Dorn et al. (2012) estimated muscle forces during running and sprinting using static optimization, and calculated muscle contributions

* Correspondence to: Clark Center, Room S-321, Stanford University, Mail Code 5450, 318 Campus Drive, Stanford, CA 94305-5450, USA. Tel.: +1 650 723 1230; fax: +1 650 723 8544.
E-mail address: delp@stanford.edu (S.L. Delp).

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to vertical mass center and hip accelerations. However, static optimization excludes effects of activation dynamics and tendon compliance on muscle force production. Achilles tendon compliance decreases metabolic cost during running (Alexander and Bennet-Clark, 1977) and affects muscle fiber lengths, fiber velocities, and force generation during running (Biewener and Roberts, 2000; Farris and Sawicki, 2012).

We previously developed a three-dimensional muscle-driven simulation of a single subject running at approximately 4 m/s that included activation dynamics and tendon compliance (Hamner et al., 2010). Analysis of the simulation revealed that quadriceps and plantarflexors are major contributors to mass center acceleration at this running speed. We observed that arm motion effectively counterbalanced angular momentum about the vertical axis passing through the body mass center from leg swing, but had little effect on mass center accelerations. In this study, we extend upon our previous work by developing and analyzing muscle-driven simulations of multiple subjects running over a range of speeds.

Our goal was to examine how muscle forces and arm swing affect dynamics of the body at different running speeds. Specifically, we sought to determine how muscle forces contribute to mass center accelerations during the stance phase of running, and how the arms act to counterbalance motion of the legs at different running speeds. We achieved this goal by creating and analyzing muscle-driven dynamic simulations of ten subjects running at different speeds. As the simulations are based on experimental data, we also report measured joint angles, joint moments, and ground reaction forces that occurred during running at different speeds.

2. Methods

We measured motions, forces, and electromyography (EMG) patterns of ten subjects running on a treadmill at four speeds: 2.0, 3.0, 4.0, and 5.0 m/s and used these data to create muscle-driven simulations of each subject at each speed. The simulations were analyzed to determine muscle contributions to vertical, backward, and forward mass center accelerations during the stance phase. Subjects were all male with an average age, height, and mass of 29 ± 5 years, 1.77 ± 0.04 m, and 70.9 ± 7.0 kg, respectively. Each subject was an experienced long distance runner who reported running at least 50 km/week. Seven subjects were consistent mid-to-rearfoot strikers and three subjects were consistent forefoot strikers at all running speeds examined in this study, except one forefoot striking subject who landed on his rearfoot while running at 2 m/s. The Stanford University Institutional Review Board approved the experimental protocol and subjects provided informed consent to participate.

2.1. Experimental data

Marker trajectories and ground reaction forces and moments were collected as each subject ran on a treadmill at different speeds. We placed 54 reflective markers on each subject and collected a static calibration trial. Functional joint movements were measured to calculate hip joint centers (Gamage and Lasenby, 2002). Marker positions were measured at 100 Hz using eight Vicon MX40+ cameras. Ground reaction forces and moments were measured at 1000 Hz using a Bertec Corporation instrumented treadmill. Marker positions and ground reaction forces were low pass filtered at 15 Hz with a zero-phase 4th order Butterworth filter and critically damped filter (Robertson and Dowling, 2003), respectively.

EMG signals were recorded using a Delsys Bangoli System with surface electrodes placed on 10 muscles: soleus, gastrocnemius lateralis, gastrocnemius medialis, tibialis anterior, biceps femoris long head, vastus medialis, vastus lateralis, rectus femoris, gluteus maximus, and gluteus medius. The raw EMG signal from each muscle was corrected for offset, rectified, and low-pass filtered at 10 Hz with a zero-phase 2nd order Butterworth filter (Buchanan et al., 2005). We then normalized the processed EMG signal from each muscle by the maximum voltage recorded across all trials for each subject.

2.2. Musculoskeletal simulations

Musculoskeletal simulations were generated using OpenSim (Delp et al., 2007). A 12 segment, 29 degree-of-freedom generic musculoskeletal model (Hamner et al., 2010) was used to create the simulations (Fig. 1; Supplemental Movie 1). Lower extremity and back joints were driven by 92 Hill-type musculotendon actuators (Anderson and Pandy, 1999; Delp et al., 1990) and arms were driven by torque actuators. We scaled the generic model to match each subject's anthropometry based on experimentally measured markers placed on anatomical landmarks and calculated hip joint centers. A virtual marker set was placed on the model based on these anatomical landmarks. Joint angles were calculated using an inverse kinematics algorithm that minimized the difference between experimentally measured marker positions and corresponding virtual markers on the model at each time frame. Joint moments were calculated using the residual reduction algorithm (RRA) (Delp et al., 2007). RRA allows for small changes in joint angles (RMS change $< 1.5^\circ$) and torso mass center location (RMS change < 5 cm) to minimize residual forces and moments applied to the pelvis (Kuo, 1998). Muscle excitations, activations, and forces needed to generate those moments and track the measured motion were estimated with the computed muscle control (CMC) algorithm (Thelen and Anderson, 2006; Thelen et al., 2003). CMC estimates muscle forces by minimizing the sum of the square of muscle excitations while accounting for muscle activation and contraction dynamics (Zajac, 1989). Constraints were applied to the muscle excitations for gluteus medius, semimembranosus, biceps femoris long head, vastus lateralis, gastrocnemius medialis, gastrocnemius lateralis, soleus, and tibialis anterior so that they better matched EMG recordings.

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To test the accuracy of the simulations, we compared simulated quantities to experimental data. The simulations tracked measured kinematics with a maximum RMS deviation of 2.5° for each joint angle over a gait cycle. Simulated muscle activations and experimental EMG data showed similar features (Fig. 2), including strong activation during the stance phase of soleus, gastrocnemius medialis, gastrocnemius lateralis, vasti lateralis, vasti medialis, biceps femoris long head, gluteus maximus, and gluteus medius. There was a delay of approximately

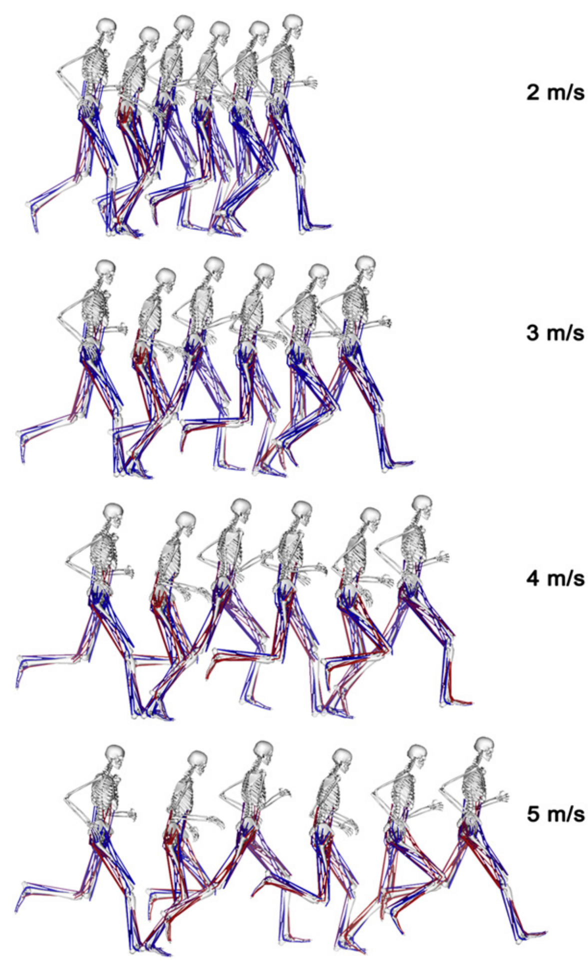

Fig. 1. Musculoskeletal model used to generate simulations of the running gait cycle for ten subjects at four running speeds: 2.0, 3.0, 4.0, and 5.0 m/s. Snapshots from the simulations of a representative subject illustrate a complete gait cycle at each speed. The gait cycle starts at right foot strike and ends at the subsequent right foot strike. Muscle color indicates simulated activation level from no activation (dark blue) to full activation (bright red).