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Muscle compensatory mechanisms during able-bodied toe walking

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Abstract

The purpose of this study was to use muscle-actuated forward dynamic simulations to quantify individual muscle contributions to body support (vertical ground reaction force) and propulsion (horizontal ground reaction force) and the mechanical energetics of the body segments during toe and heel-toe walking performed by able-bodied subjects to identify possible compensatory mechanisms necessary to toe walk. The simulations showed that an increased magnitude of plantar flexor power output in early stance, which was necessary to maintain the equinus posture during toe walking, contributed to body support and acted to brake (decelerate) the center-of-mass in the horizontal direction. This in turn required a reduction in the contributions to support from the vastii, gluteus maximus and biarticular hamstring muscles and decreased contributions to braking from the vastii and to a lesser extent the gluteus maximus. In late stance, the soleus contributed less to body support and forward propulsion during toe walking, which when combined with the increased braking by the plantar flexors in early stance, required a prolonged contribution to forward propulsion from the hamstrings from mid- to late stance. The multiple compensatory mechanisms necessary to toe walk have important implications for distinguishing between underlying pathology and necessary compensatory mechanisms, as well as for identifying the most appropriate treatment strategy for equinus gait.

Keywords: Equinus gait; Mechanical energetics; Modeling; Computer simulation

1. Introduction

Toe walking is a gait pattern often associated with several musculoskeletal and neurological disorders including cerebral palsy, muscular dystrophy, traumatic brain injury and stroke. In an effort to distinguish primary from secondary gait deviations, previous biomechanical studies have analyzed able-bodied subjects performing both toe and heel-toe walking and have noted distinct differences between the two gait modes (e.g., [1–5]). For example, Kerrigan et al. [4] found that the peak ankle torque and power in terminal stance and pre-swing were reduced in toe walking compared to heel-toe walking. Similarly, Hampton et al. [3] performed a quasi-static analysis of the foot and tibia and found that the increased equinus posture results in reduced plantar flexor force requirements. The reduced plantar flexor force (primarily from the gastrocnemius and soleus) was attributed to the closer proximity of the resultant ground reaction force (GRF) vector to the ankle joint center with greater angles of plantar flexion.

These results are significant since during the second half of stance the plantar flexors are the primary contributors to vertical and horizontal GRFs during heel-toe walking and provide body support and forward propulsion, respectively [6–8]. Thus, if plantar flexor force output is reduced during toe walking, this may require compensatory action from other muscle groups to provide the necessary body support and/or forward propulsion to walk at a given speed. In addition, the ability of a muscle to provide body support and forward propulsion (i.e., contribute to the GRFs) is dependent on the current state of the system (i.e., the body segment orientations; [9]). Thus, it is not clear if the equinus posture associated with toe walking alters the ability of the plantar flexors to contribute to these functional tasks.

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Another significant difference between toe and heel-toe walking is the greater muscle excitation and internal ankle moment required in early stance during toe walking in order to maintain the plantar flexed posture throughout stance (e.g., [1,4,5]), which would require greater plantar flexor force output. During heel-toe walking, the hip and knee extensors are the primary contributors to the vertical and horizontal GRFs in the first half of stance [7,8]. Thus, the increased plantar flexor force output in early stance during toe walking may require decreased contributions from the hip and knee extensors if the plantar flexors also contribute to the GRFs as they do in the second half of stance. This would be consistent with previous studies showing a decrease in the knee extensor moment in early stance during toe walking [1,4,5]. Identifying the compensatory mechanisms required for toe walking at the individual muscle level would help distinguish between underlying pathology and necessary compensations, which has important implications for identifying the most appropriate treatment strategy for equinus gait.

The purpose of this study was to use muscle-driven forward dynamic simulations of able-bodied subjects performing toe and heel-toe walking at the same speed to identify differences in individual muscle contributions to body support (vertical GRF) and propulsion (horizontal GRF) between the two gait modes. Specifically, we tested whether (1) the decreased plantar flexor force output in late stance during toe walking requires increased contributions from other muscle groups to provide the necessary body support and forward propulsion, and (2) the increased plantar flexor contributions to the ground reaction forces in early stance during toe walking requires decreased contributions from the hip and knee extensors to offset the increased plantar flexor contributions. To further understand the potential compensatory mechanisms, a segment power analysis [10] was used to identify differences between toe and heel-toe walking in the biomechanical mechanisms used by individual muscles to contribute to the GRFs.

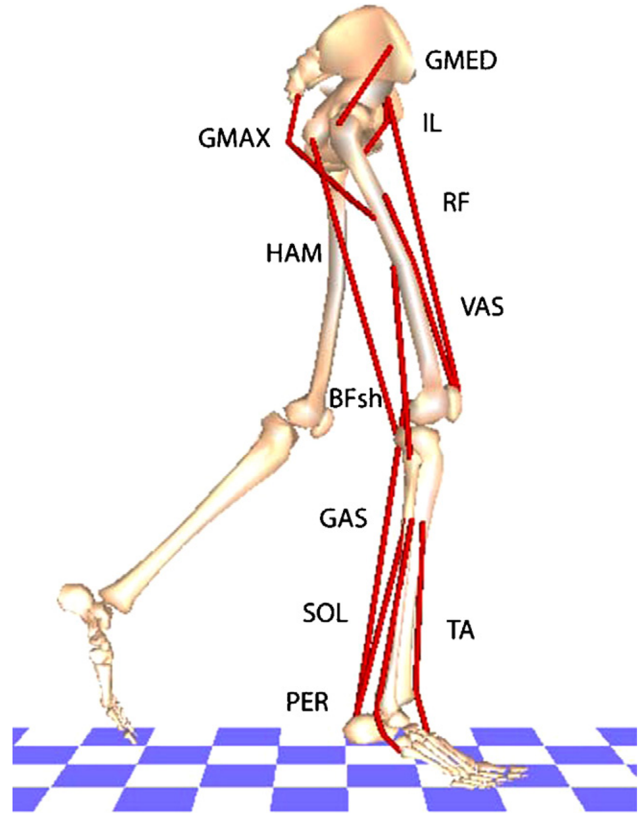

Fig. 1. The 2D musculoskeletal model consisted of the trunk (head, arms, torso and pelvis) and both legs (femur, tibia, patella, rear-foot, mid-foot and toes). Twenty-five Hill-type musculotendon actuators per leg were combined into the 11 muscle groups shown (only right leg muscle groups are shown for clarity). The muscle groups included GMED (anterior and posterior portion of gluteus medius), IL (iliacus, psoas), RF (rectus femoris), VAS (three components of the vastii), TA (tibialis anterior, peroneus tertius, extensor hallucis longus, extensor digitorum longus), PER (peroneus longus, peroneus brevis), SOL (soleus, tibialis posterior, flexor hallucis longus, flexor digitorum longus), GAS (medial and lateral gastrocnemius), BFsh (biceps femoris short head), HAM (medial hamstrings, biceps femoris long head) and GMAX (gluteus maximus, adductor magnus). The muscles within each muscle group received the same excitation pattern.

2. Methods

2.1. Musculoskeletal model

A 2D bipedal musculoskeletal model was generated using SIMM (MusculoGraphics, Inc., Santa Rosa, CA). The model included a trunk representing the head, torso and arms and two legs consisting of the thigh, shank, patella, rear foot, mid-foot and toes (Fig. 1). The model had 13 degrees of freedom including two translations and one rotation for the trunk, flexion–extension for hip, knee, ankle, rear-mid-foot and toe joints. The position and orientation of the patella were prescribed using a polynomial as a function of knee flexion [11]. Twenty-five Hill-type musculotendon actuators per leg were used to drive the model. The actuators were combined into 11 functional groups based on anatomical classification (Fig. 1), with muscles within each group receiving the same excitation pattern. Block excitation patterns specified by onset, duration and magnitude were used to excite the muscles, except for three muscle groups (SOL, GAS and VAS) for which EMG were available and used as the excitation inputs (see Section 2.4). The muscle activation dynamics were described with a first-order differential equation [12], with activation and deactivation time constants derived from Winters and Stark [13]. For those muscles that data were not explicitly available in Winters and Stark [13], nominal values of 12 and 48 ms, respectively, were used. The foot-ground contact was modeled using 38 visco-elastic elements per foot distributed among the foot segments [14].

2.2. Forward dynamic simulation and dynamic optimization

The model's equations-of-motion were generated using SD/FAST (PTC, Needham, MA) and the forward dynamic simulation code was generated using the Dynamics Pipeline (MusculoGraphics, Inc., Santa Rosa, CA). Dynamic optimization using a simulated annealing algorithm [15] was used to fine-tune the muscle excitation patterns to generate the simulations of toe and heel-toe walking. Constraints were placed on the excitation timing in the optimization to assure the timing was consistent with previous studies of toe and heel-toe walking [1,2,5,16]. The tracking variables included trunk rotation and translations, hip, knee and ankle joint angles, and the vertical and horizontal ground reaction forces during a full gait cycle from right foot-strike to the subsequent right foot-strike. The optimization process was continued until the tracking variables were within two standard deviations of the experimental average values (see Section 2.4).

2.3. Muscle contributions to body support and propulsion

Muscle contributions to body support and propulsion (or braking) were quantified by decomposing the vertical and horizontal GRFs, respectively, into individual muscle force contributions (e.g., [8,17]). To understand the biomechanical mechanisms contributing to body support and propulsion, a segment power analysis [10] was used to quantify how mechanical energy is generated, absorbed or transferred among the body segments (i.e., the trunk, and ipsilateral and contralateral legs) by each muscle. Positive power indicates that the muscle accelerated the segment in the direction of its motion, while negative power indicates the muscle decelerated the segment in the direction of its motion.

2.4. Experimental data

Previously collected experimental data [5] were used in the dynamic optimization to generate the simulations and will be briefly described here. Ten able-bodied subjects (eight males and two females: age 36.9 ± 11.2 years old, height 171.7 ± 8.3 cm, body mass 67.2 ± 12.4 kg) participated in the experiments after signing informed consent at the Pathokinesiology Laboratory at Rancho Los Amigos National Rehabilitation Center in Downey, CA, where all data were collected. The subjects performed toe walking at their freely selected speed on a straight walkway after becoming comfortable with toe walking. Three-dimensional GRFs (Kistler Instrument Corp., Amherst, NY) and kinematic motion data (Vicon Motion Systems, Oxford, UK) were collected at 2500 and 50 Hz, respectively. EMG from the soleus, medial gastrocnemius and vastus intermedius were collected using fine wire-electrodes. Following the toe-walking trials, the subjects performed heel-toe walking at ±5% of their toe walking speed, and the same data were collected. All data were normalized over the gait cycle, averaged across trials and then across all subjects to obtain group averaged data. Further details of experimental data collection and processing can be found in Perry et al. [5].

3. Results

The simulations of toe and heel-toe walking emulated well the group-averaged experimental kinematic and ground reaction force tracking variables within ±2 S.D. (Fig. 2). The excitation timing in the simulations also compared well with data in previous toe and heel-toe walking studies (see Fig. 4) [1,2,5,16].

3.1. Muscle contributions to body support and propulsion

In early stance during toe walking, SOL had the largest contribution to both the vertical and horizontal GRFs

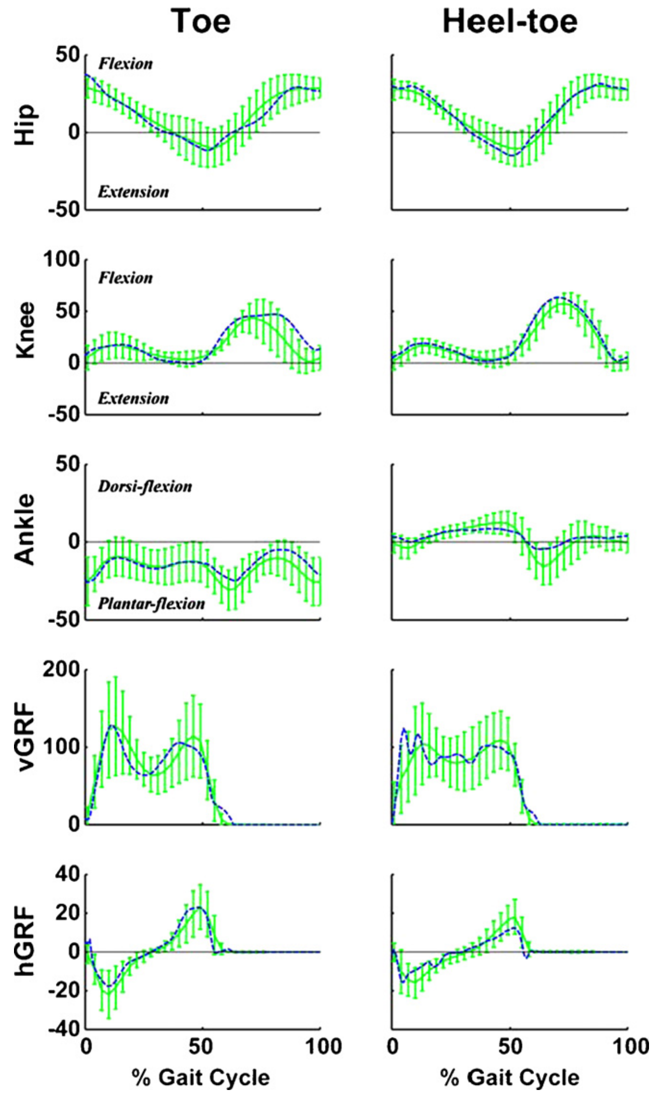

Fig. 2. Experimental joint angles (units: degrees), vertical and horizontal GRFs (vGRF and hGRF, respectively, units: % body weight) (solid line; average ± 2 S.D.) and corresponding simulation data from the toe and heel-toe walking simulations from right foot contact to right foot contact (dashed line).

(Fig. 3A and C: SOL, 0–30% stance). The positive vertical GRF provided body support (i.e., accelerated the body center-of-mass (COM) upward) while the negative horizontal GRF acted to decelerate or brake the COM in the anterior direction. Similarly, but to a lesser extent, GAS also acted to support and brake the COM in early stance during toe walking (Fig. 3A and C: GAS, 0–20% stance). In contrast, during heel-toe walking the contributions from SOL and GAS to the GRFs were minimal during early stance (Fig. 3B and D: SOL, GAS, 0–30% stance).

VAS, GMAX and HAM contributed little to body support during early stance in toe walking compared to heel-toe walking, where these muscles were the primary contributors to support (Fig. 3B: VAS, GMAX, HAM, 0–40% stance). VAS contribution to braking also decreased in toe walking during early stance (Fig. 3C and D: VAS, 0–25% stance).

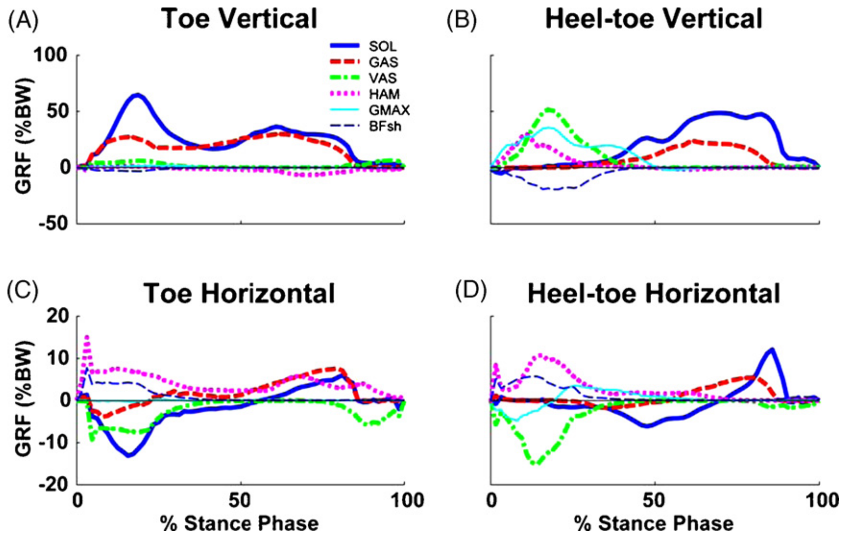

Fig. 3. Individual muscle contributions to the vertical and horizontal GRFs (units: % body weight) during toe and heel-toe walking. The contributions from RF, IL, GMED, TA and PER were minimal and not shown. Stance phase is defined from right foot contact to right toe-off.

GMAX contributed little to forward propulsion or braking after foot-strike in toe walking, in contrast to providing initial braking and then propulsion in heel-toe walking (Fig. 3D: GMAX, 0–50% stance).

HAM and BFsh provided propulsion in early stance in both toe and heel-toe walking, with the contribution from HAM extending into late stance in toe walking (Fig. 3C: HAM, 0–100% stance, BFsh, 0–25% stance; Fig. 3D: HAM, BFsh, 0–30% gait cycle). During late stance, SOL and GAS were the primary contributors to body support and propulsion in both toe and heel-toe walking (Fig. 3: SOL, GAS, ~60–90% stance). However, SOL contributions were smaller in toe walking than in heel-toe walking. VAS acted to brake the COM in late stance before toe-off in both walking conditions (Fig. 3C and D: VAS, ~80–100% stance).

3.2. Muscle contributions to body segment power

In general, the muscle contributions to the body segment mechanical power were similar between toe and heel-toe walking, with the exception of distinct differences in *early stance* (Fig. 4). In toe walking, eccentric SOL action absorbed power from the leg and delivered that energy to the trunk in early stance (Fig. 4A: solid and dash–dot lines <0, 0–20% gait cycle, dashed line >0, ~10–30% gait cycle). The leg power absorbed was primarily the result of SOL acting to decelerate the leg in the horizontal direction (Fig. 5A: SOL, dotted line <0, 0–50% gait cycle). In the horizontal direction, the amount of energy absorbed from the leg was greater than the amount of energy delivered to the trunk, which had the net effect of decelerating the COM (Fig. 3C: SOL, GRF <0, 0–30% stance). Simultaneously, SOL delivered positive vertical power to the trunk (Fig. 5A: SOL, solid line >0, 5–20% gait cycle) to provide body support (Fig. 3A: SOL, 0–30% stance). In contrast to SOL, GAS acted to brake the COM during early stance in toe walking by decelerating the trunk horizontally (Fig. 5A: GAS, dashed line <0, 0–25% gait cycle).

VAS and GMAX absorbed power from the leg and delivered that energy to the trunk in early stance in both toe and heel-toe walking (Fig. 4A and B: VAS, GMAX, dashed lines >0, dash–dot lines <0, ~0–25% gait cycle). The contribution from VAS and GMAX to trunk support was much less in toe walking compared to heel-toe walking (Fig. 5A and B: VAS, GMAX, solid lines >0, ~10–25% gait cycle), which was consistent with the decreased VAS and GMAX contributions to body support (Fig. 3A).

HAM and BFsh delivered power to the leg in early stance in both toe and heel-toe walking with the HAM contribution extending into early terminal stance in toe walking (Fig. 4A: HAM, dash–dot lines >0, 0–50% gait cycle, BFsh, 0–20% gait cycle; Fig. 4B: HAM, BFsh, dash–dot lines >0, 0–20% gait cycle). The positive leg power acted to accelerate the leg primarily in the horizontal direction (Fig. 5A: HAM, dotted line >0, 0–20% gait cycle, BFsh, 0–40% gait cycle; Fig. 5B: HAM, BFsh dotted lines >0, 0–20% gait cycle). HAM also acted to accelerate the trunk horizontally from early to late stance in toe walking (Fig. 5A: HAM, dashed line >0, 0–50% gait cycle). Thus, during toe walking HAM provided body propulsion nearly the entire stance phase (Fig. 3C) by accelerating both the leg and trunk in the horizontal direction from early to mid-stance, and then primarily by accelerating the trunk from mid- to late stance.

In *late stance*, the biomechanical mechanisms used by SOL and GAS to provide body support and propulsion were similar in both toe and heel-toe walking. SOL and GAS acted to decelerate the downward motion of the trunk (Fig. 5A and B: SOL, GAS, solid lines <0, ~25–50% gait cycle), which provided body support (Fig. 3A and B). SOL and GAS simultaneously accelerated the trunk in the horizontal direction from mid- to late stance, and then also

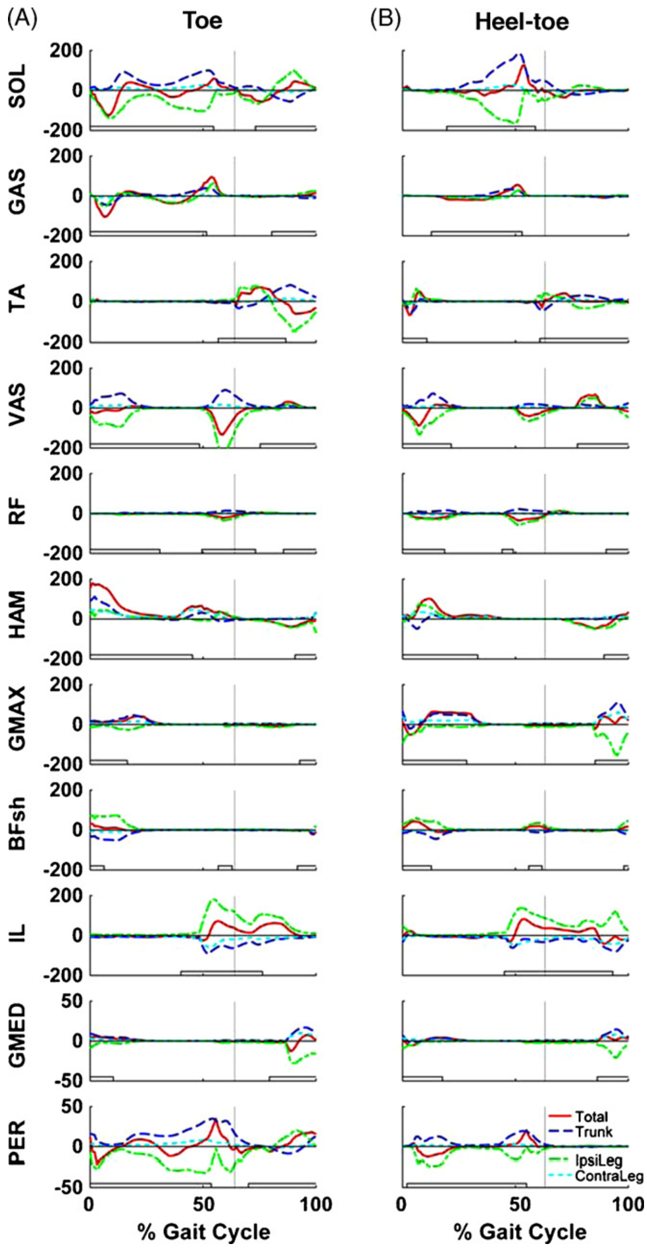

Fig. 4. Musculotendon mechanical power (total, solid line) and its distribution to the trunk (trunk, dashed-line), ipsilateral leg (ipsileg, dash–dot line) and contralateral leg (contraleg, dotted line) during (A) toe and (B) heel-toe walking over the gait cycle (from right foot contact to the subsequent right foot contact). The vertical lines indicate toe-off. The horizontal bars indicate muscle excitation timing. All units are in Watts.

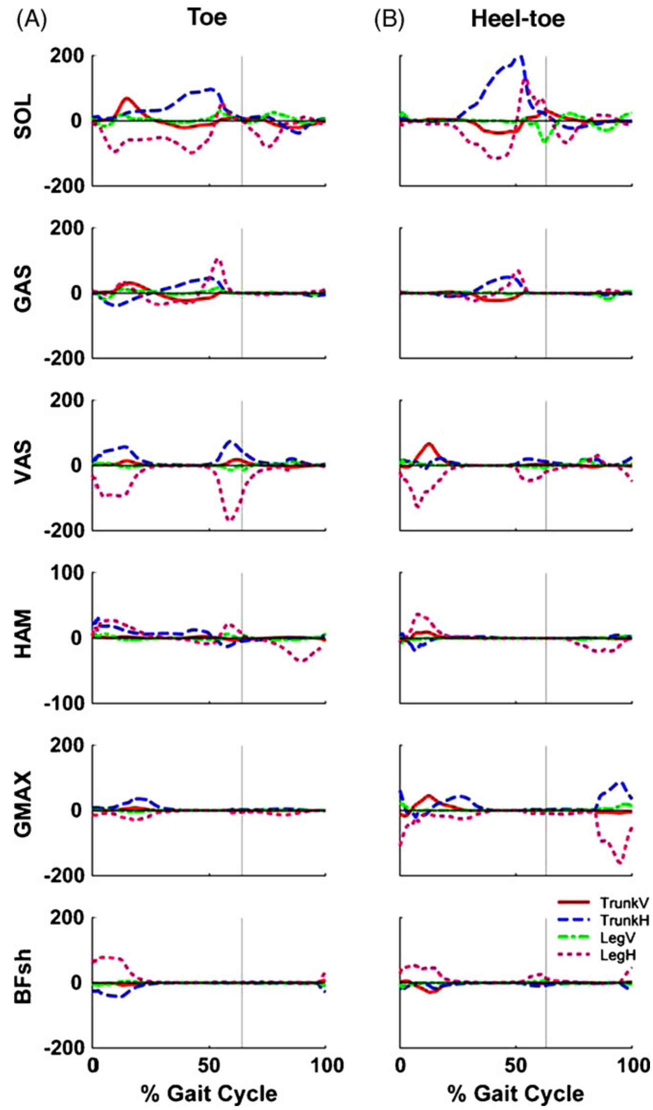

Fig. 5. Trunk and ipsilateral leg power in horizontal and vertical directions delivered by each muscle during toe and heel-toe walking over the gait cycle. TrunkV: trunk vertical power, TrunkH: trunk horizontal power, LegV: leg vertical power, LegH: leg horizontal power. The vertical lines indicate toe-off. All units are in Watts.

accelerated the leg at the end of stance (Fig. 5A and B: SOL, GAS, dashed lines >0, ~25–60% gait cycle, dotted lines >0, ~50–60% gait cycle).

4. Discussion

The objective of this study was to use muscle-driven forward dynamic simulations of toe and heel-toe walking to identify the differences in muscle contributions to the vertical (body support) and horizontal (braking/propulsion) ground reaction forces and identify potential compensatory mechanisms necessary for able-bodied subjects to toe walk. The results showed that muscles contribute to body support and forward propulsion in a similar manner between the two gait modes with a few notable exceptions (Fig. 3). During toe walking, the increased excitation and force output by SOL and GAS in early stance necessary to maintain the plantar flexed posture [18] resulted in greater contributions to body support (Fig. 3A) by acting to accelerate the trunk vertically to a greater extent than in heel-toe walking (Fig. 5A). However, the increased SOL activity also acted to decelerate the forward motion of the body to a greater extent (Fig. 3C), primarily by decelerating the leg in the horizontal direction (Fig. 5A: dotted line <0, 0–25% gait cycle).

Because of the increased SOL contributions to support and braking in toe walking during early stance, decreased contributions to support from VAS, GMAX and HAM, and braking from VAS were required (Fig. 3A: VAS, GMAX, 0–30% stance; Fig. 3C: VAS, 0–25% stance), which were due at least in part to a decrease in musculotendon force output (Fig. 6). Although HAM decreased its contribution to support, it still contributed to the horizontal ground reaction force to provide body forward propulsion in early stance, and to a greater extent from mid- to late stance (Fig. 3C). This additional propulsion was generated as HAM continued to deliver power to the trunk to accelerate it forward, (Fig. 5A: HAM, dashed line >0, 20–50% gait cycle). In contrast, during heel-toe walking the contribution by HAM to body propulsion ceased by mid-stance (Fig. 3D). The prolonged HAM contribution was consistent with previous EMG analyses of able-bodied subjects toe walking, which showed extended HAM excitation into late stance (e.g., [1]). Thus, the increased HAM contribution to propulsion from mid- to late stance appears to be an important compensatory mechanism for the reduced force output by SOL and GAS and greater braking produced by SOL in early stance to provide necessary forward propulsion in toe walking. It is critical to distinguish this compensatory HAM function from

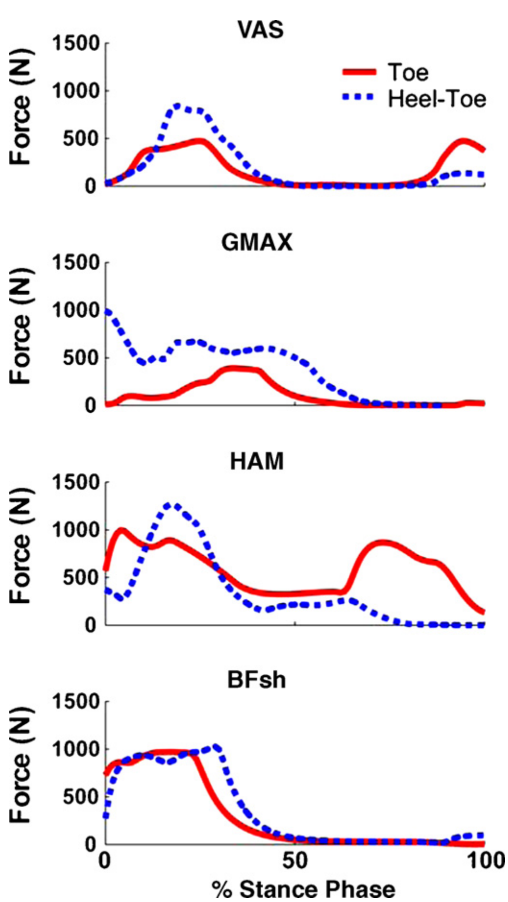

Fig. 6. Musculotendon force output during stance in toe and heel-toe walking from VAS, GMAX, HAM and BFsh.

primary obstructive activation that could be an indication for surgical intervention [19].

The decrease in VAS and HAM force output in early stance, while BFsh remained unchanged (Fig. 6), resulted in a lower net knee extensor moment during toe walking, which is in agreement with previous studies showing decreased knee extensor moment and power during early stance in able-bodied toe walking (e.g., [1,4,5]). The lower knee moment required in toe walking may provide a benefit for those individuals with weak knee extensors [20]. However, there is a cost associated with toe walking. SOL and GAS excitation levels were previously found to be 75% (SOL) and 83% (GAS) higher in toe walking compared to heel-toe walking [5] because of the reduced force generating capacity of the plantar flexors as a result of the shorter fiber length associated with the increased plantar flexion angles [18]. In the present simulations, despite a 55% increase in excitation, the SOL force decreased in late stance when it normally provides its greatest propulsion in heel-toe walking (Fig. 3D).

In addition to these differences between toe and heel-toe walking, previous EMG studies have observed prolonged VAS excitation (e.g., [1,5,16]) co-contracting with HAM during the stance phase in toe walking [1], which was also observed in our toe walking simulation. Davids et al. [1] suggested that the co-contraction is necessary for gait stability. However, we performed a post–hoc muscle-induced acceleration analysis [9] and found that both VAS and HAM act to accelerate the knee and hip into extension from mid- to late stance (VAS accelerations were small), which is a phenomenon also observed in heel-toe walking particularly during the first half of stance [21]. The prolonged VAS excitation during toe walking appears to assist the decreased RF function in late stance, by absorbing power from the leg and delivering that energy to the trunk (Fig. 4A: VAS, RF, dash–dot lines <0, dashed lines >0, 50–70% gait cycle).

An important note is that the simulations in the present study were generated using a generic musculoskeletal model with nominal musculotendon parameters, and therefore caution should be taken when interpreting the quantitative results. However, we constrained the muscle excitation timing based on our own EMG data (i.e., SOL, GAS and VAS) and previously published data [1,2,5,16] and the relative excitation magnitudes between simulations were consistent with our EMG data [5]. Thus, we have confidence in the qualitative differences in muscle function between toe and heel-toe walking. To further assess the sensitivity of our results, an extensive number of optimizations and analyses were performed and our results and conclusions remained unchanged. However, another important note is that previous EMG studies have shown large inter-subject variability in the excitation patterns in some muscles during toe walking by able-bodied subjects (e.g., the tibialis anterior [1,5,16,22]). Thus, it is likely that other muscle coordination strategies than those in the present simulations exist. In addition, the kinematic and kinetic patterns of the lower extremity during toe walking have been shown to differ depending on the relative contributions of SOL and GAS contractures to the equinus posture [23]. Therefore, it is possible that the balance of contributions of these two muscles to toe walking differs between able-bodied subjects and individuals with cerebral palsy. Further study is needed to examine the possibility of different muscle coordination strategies in both able-bodied and pathological populations.

In summary, we found that there were distinct differences in how muscles contribute to body support and forward propulsion between toe and heel-toe walking. The most notable differences were the increased SOL and GAS contributions to body support and braking in early stance during toe walking that required decreased contributions to body support from VAS, GMAX and HAM, and decreased contributions to braking from VAS. In addition, the decreased SOL contributions to body support and forward propulsion in late stance and the increased SOL braking in early stance during toe walking required increased HAM contributions to forward propulsion from mid- to late stance. Thus, considerable compensatory mechanisms were necessary in toe walking, which has important implications for distinguishing between underlying pathology and necessary compensatory mechanisms and identifying the most appropriate treatment strategy for equinus gait.

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Conflict of interest

There is no conflict of interest.

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