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Changing the demand on specific muscle groups affects the walk–run transition speed

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SUMMARY

It has been proposed that muscle-specific factors trigger the human walk–run transition. We investigated if changing the demand on trigger muscles alters the preferred walk–run transition speed. We hypothesized that (1) reducing the demand on trigger muscles would increase the transition speed and (2) increasing the demand on trigger muscles would decrease the transition speed. We first determined the normal preferred walk–run transition speed (PTS) using a step-wise protocol with a randomized speed order. We then determined PTS while subjects walked with external devices that decreased or increased the demand on specific muscle groups. We concurrently measured the electromyographic activity of five leg muscles (tibialis anterior, soleus, rectus femoris, medial and lateral gastrocnemius) at each speed and condition. For this study, we developed a dorsiflexor assist device that aids the dorsiflexor muscles. A leg swing assist device applied forward pulling forces at the feet thus aiding the hip flexors during swing. A third device applied a horizontal force near the center of mass, which impedes or aids forward progression thus overloading or unloading the plantarflexor muscles. We found that when demand was decreased in the muscles measured, the PTS significantly increased. Conversely, when muscle demand was increased in the plantar flexors, the PTS decreased. However, combining assistive devices did not produce an even faster PTS. We conclude that altering the demand on specific muscles can change the preferred walk–run transition speed. However, the lack of a summation effect with multiple external devices, suggests that another underlying factor ultimately determines the preferred walk–run transition speed.

Key words: gait transition, locomotion, EMG, electromyography.

INTRODUCTION

At speeds slower than ~2.0 m s^{-1} adult humans prefer to walk, and at faster speeds they prefer to run. The slowest speed at which people prefer to run is defined as the walk–run transition speed (Thorstensson and Roberthson, 1987). The trigger for the walk–run transition is controversial (Raynor et al., 2002). Some have proposed that underlying factors, such as metabolic cost (Mercier et al., 1994), perceived exertion (Hreljac et al., 2002; Noble et al., 1973), pendular mechanics (Kram et al., 1997) or body size (Hreljac, 1995b) determine the walk–run transition speed. Others have implicated local, muscle-specific factors such as overexertion/fatigue (Hreljac, 1993; Hreljac, 1995a; Hreljac et al., 2001; Prilutsky and Gregor, 2001) or muscle force-velocity-length relationships (Neptune and Sasaki, 2005) as triggers of the walk–run transition.

One of the local factors thought to trigger the walk–run transition speed is the greater ankle angular velocity during late swing and after heelstrike at faster walking speeds (Hreljac et al., 2001). The greater ankle angular velocity apparently causes the primary dorsiflexor muscle (tibialis anterior) to reach a critical level of muscle activity and subsequently a sense of overexertion (Hreljac, 1995a; Hreljac et al., 2001; Segers et al., 2007). By switching to running, the dorsiflexors can comfortably operate below their maximal capacity (Hreljac et al., 2001).

Building upon Hreljac's dorsiflexor argument, Prilutsky and Gregor (Prilutsky and Gregor, 2001) found a similar pattern of muscle activity in other swing-phase flexor muscles. The flexor muscles hypothesized by Prilutsky and Gregor to trigger the walk–run transition include the tibialis anterior, biceps femoris (knee flexor) and the rectus femoris (hip flexor) (Prilutsky and Gregor, 2001). At or above the preferred transition speed, they found that muscle activity in these flexor muscles was less in running than walking, which supported their hypothesis.

In contrast to the overexertion/fatigue hypothesis, Neptune and Sasaki (Neptune and Sasaki, 2005) suggested that the reduced force-generating ability of the plantar flexors during fast walking triggers the gait transition. Some or all of the plantar flexors contribute to body weight support and propulsion during gait (Gottschall and Kram, 2003; Neptune et al., 2004). Neptune and Sasaki's computer simulation based on empirical data demonstrated that force production in the soleus and medial gastrocnemius muscles is impaired at walking speeds faster than the preferred transition speed (Neptune and Sasaki, 2005). At the walk–run transition speed and faster, they found that running improves the contractile conditions of the plantar flexor muscles.

The purpose of this study was to determine if external devices that change the demand on specific trigger muscles would alter the preferred walk–run transition speed. We hypothesized that: (1) reducing the demand on trigger muscles would increase the transition speed and (2) increasing the demand on trigger muscles would slow the transition speed. For the current study, we developed a new device that reduces the demand on the dorsiflexor muscles during walking (Fig. 1). This dorsiflexor assist (DFA) device externally exerts a flexor torque at the ankle that reduces demand on the dorsiflexor muscles. Some recent studies from our laboratory have used other external devices to alter the muscle activity needed while locomoting on a treadmill. Modica and Kram (Modica and Kram,

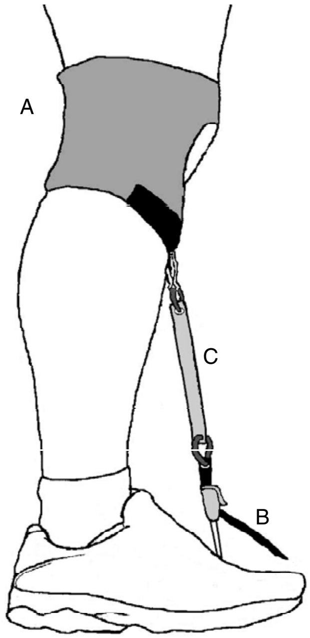

Fig. 1. Dorsiflexor assist (DFA) device used to reduce dorsiflexor muscle demand. The device secures around the subject's knee (A) with a Velcro-compatible surface. The subject dorsiflexed the foot allowing strap (B) to be pulled through the cam-style buckle. This allowed the rubber tubing (C) to be stretched applying a force and thus a torque around the ankle.

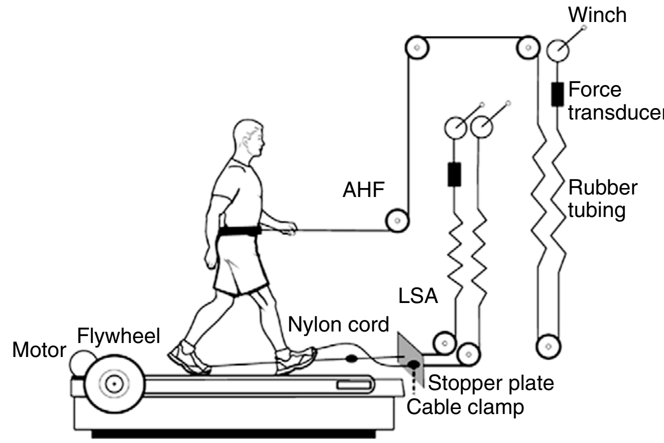

Fig. 2. Leg swing assist (LSA) device and aiding horizontal force (AHF) device. Schematic adapted from Gottschall and Kram (Gottschall and Kram, 2005). The system of rubber tubing and pulleys apply forces at either the feet or near the center of mass.

2005) utilized a leg swing assist (LSA) device to apply forward pulling forces at the feet, effectively aiding the hip flexor muscles during the swing phase (Fig. 2). Similarly, Gottschall and Kram (Gottschall and Kram, 2003) utilized a device to apply a horizontal force at the waist, near the subject's center of mass (Fig. 2). This device can pull the subjects forward providing an aiding horizontal force (AHF) or pull backwards providing an impeding horizontal force (IHF), thus decreasing or increasing the demand on the propulsive muscles (e.g. plantar flexors) during walking.

MATERIALS AND METHODS

Subjects

Ten men and ten women (age=28.3±5.9 years, height=1.73±0.10 m, mass=70.4±13.3 kg, leg length=0.87±0.19 m; mean ± s.d.) volunteered to participate in this study. The volunteers were recreationally active individuals and free of musculoskeletal injuries. Subjects gave written informed consent in accordance with the Human Research Committee at the University of Colorado.

Determination of preferred transition speed

The preferred transition speed (PTS) is defined experimentally as the slowest speed at which an individual prefers to run *versus* walk (Thorstensson and Roberthson, 1987). This transition can be reliably determined using ramp or step-wise increases of the treadmill speed (Hreljac et al., 2007). Our protocol allowed the subject to decide on a gait for each particular speed. We utilized a protocol that arranged the treadmill speeds in random order. The purpose of randomizing the treadmill speeds was to control for any influences of fatigue and to avoid subject perception of a monotonically increasing speed sequence.

Before testing, subjects became familiar with walking and running on the treadmill. Each subject walked for 5 min at 1.25 m s^{-1} and jogged for 5 min at 3.0 m s^{-1}. After the initial warm-up, a practice trial allowed subjects to become familiar with the testing protocol. With the treadmill stopped, subjects received instructions that they should walk or run at their preference. After 30 s of locomotion, the investigator asked the subject whether they preferred to walk or to run. Subjects took as much time as needed to decide on a gait, sometimes trying both gaits before reaching a decision. Once the subject verbally responded without doubt to the question, the investigator started collecting muscle activity data. Next, the investigator stopped the treadmill and set the motor to another speed. We repeated the same process with random speed settings of 0.1 m s^{-1} intervals ranging between 1.5 and 2.5 m s^{-1}.

Measurement of muscle activity

After the familiarization, we prepared each subject's right leg for placement of surface electromyographic (EMG) electrodes. We chose five muscles implicated in the determination of the walk-to-run transition speed: tibialis anterior (TA), soleus (SOL), medial gastrocnemius (MGAS), lateral gastrocnemius (LGAS), and rectus femoris (RF). Electrode placement location for each muscle followed the recommendations of previous researchers (Rainoldi et al., 2004). These electrodes remained in place for the entire experimental session. We prepared the skin at each muscle site by shaving and lightly abrading the skin with a fine grain sand paper. We cleaned the site with alcohol and after the surface dried, placed two bipolar, silver-silver chloride electrodes (1 cm diameter disks) 2 cm apart over the belly of each muscle. To optimize the EMG signal and minimize crosstalk, we instructed subjects to selectively activate each muscle, monitored the strength of the signal, and re-positioned electrodes if needed (Cram and Kasman, 1998). After verifying the EMG signal observations, we secured the electrodes and cables with tape and leg wraps.

During the experimental conditions, we sampled EMG activity at a rate of 1000 Hz for 10 s at each speed. Raw data were collected using a telemeteric amplifier system (Noraxon, Scottsdale, AZ, USA) with a gain of 1700. To remove movement artifact, we high-pass filtered using a fourth-order Butterworth filter with a cutoff of 7 Hz. A foot switch insole in the right shoe indicated heelstrikes and toe-offs. We analyzed and compared EMG activity for subjects while walking at their PTS during each condition. For the impeding horizontal force condition, all but three of the subjects transitioned at speeds slower than their normal PTS. Therefore, we compared the EMG activity while walking at the preferred transition speed for the impeding horizontal force condition to the EMG activity for normal walking at the same slower speed. We analyzed five steps that displayed normal EMG burst patterns for each muscle (i.e. no obvious gaps in data or unusual spikes). The percentage of gait cycle was calculated from the initial heelstrike of the right leg at 0% and the following heelstrike of the right leg at 100%. Burst onset and off times were determined by visual inspection of EMG signals *versus* time plots. For the TA muscle, this burst period occurred between 60–110% of the gait cycle. For the RF muscle, this burst period occurred between 50–85% of the gait cycle. For the plantar flexors, this burst period occurred between 0–60% of the gait cycle. We calculated burst durations for each muscle and condition. For each condition, we full-wave rectified and band pass filtered (16-499 Hz) the EMG signals using a software routine written specifically for this project (MatLab, Math Works, Natick, MA, USA). We calculated the integrated and mean EMG amplitude for each muscle and averaged five steps in each condition.

Experimental protocol

Subjects completed ten testing conditions. Each condition used the same randomized speed order used during the familiarization. The first condition determined normal PTS for locomotion on the level without any external devices (PTS1). Eight experimental conditions followed PTS1 in the following order, allowing for quick changes between each device: impeding horizontal force (IHF), aiding horizontal force (AHF), leg swing assist (LSA), aiding horizontal force combined with leg swing assist (LSA+AHF), dorsiflexor assist (DFA), dorsiflexor assist combined with aiding horizontal force (DFA+AHF), dorsiflexor assist combined with leg swing assist (DFA+LSA), and dorsiflexor assist combined with aiding horizontal forces and leg swing assist (COMBO). Last, we performed a re-test of the normal preferred transition speed without external devices (PTS2) to detect any possible effects of testing fatigue as well as establishing repeatability of the PTS measurement.

For each condition, subjects initially walked at $1.25\ \mathrm{m\ s^{-1}}$ to become comfortable with the configuration of external devices. Once comfortable (~5 min), we stopped the treadmill and began the randomized speed protocol. Rest periods of 2 min occurred between each condition as the external devices were attached and the proper amount of applied force established. At the end of each condition, subjects walked normally at a speed of $1.25\ \mathrm{m\ s^{-1}}$ for 2 min to 'wash out' any effects of the previous condition. This time period seemed reasonable and practical for the long experimental protocol.

External devices

Specifically for this study, we developed a dorsiflexor assist (DFA) device. This device was designed to reduce the need for muscle activity in the dorsiflexors (e.g. TA) during gait. Fig. 1 shows the attachment of elastic rubber tubing that assists dorsiflexion after toe-off and slows plantar flexion at heelstrike. The tubing was attached to the leg by means of a piece of specialized material (TheraTogs, Telluride, CO, USA) wrapped around the subject's knee and secured with Velcro straps (Fig. 1, point A). The underside of the TheraTog material has a non-skid surface for secure placement against the skin. The wrap has a Velcro compatible top surface. At the location of the tibial tuberosity, the elastic rubber tubing fastened to the knee wrap (Fig. 1, point C). The rubber tubing connected distally to a piece of nylon strap webbing that laced through a cam-style buckle (Fig. 1, point B). We secured the buckle to the mid-sagittal line of the subject's shoe, distally at the fifth metatarsal phalangel joint of the foot. To stretch the elastic portion of the DFA, the subject flexed their knees while standing with their feet flat. This allowed the webbing to be pulled through the buckle and secured in place. The subjects then extended their knees to a neutral standing position, stretching the elastic portion of the DFA.

We pilot tested the effectiveness of the DFA device in reducing tibialis anterior (TA) muscle activity. When subjects walked with the DFA, the elastic tubing stretched during late stance just prior to toe-off. When toe-off occurred, the stretched DFA dorsiflexed the ankle, which lifted the toes for ground clearance during the swing phase. During pilot testing of the DFA device, we found reduced TA activity during both toe-off and heelstrike. We also noted no obvious increases in plantar flexor EMG activity while using the DFA device (i.e. no co-contraction).

We quantified DFA force using the spring constant of the device. To do this, with the ankle joint at 90°, we measured the initial length of the elastic tubing prior to stretching. After stretch, we measured the tubing length. Having calibrated ahead of time, force could be calculated using the equation $F=k\Delta x$, where k is the spring constant of the elastic tubing and Δx is the tubing stretch distance. Leg and foot length affect the placement of the proximal and distal attachment points of the DFA. Thus, on a taller person, the DFA stretched further and consequently the DFA assisted ankle flexion with more force. A longer foot results in a longer moment arm for the DFA force and thus the torque applied by the DFA was proportionally greater for the taller subjects. At heel strike for an average 70 kg subject, the DFA applied a moment of ~0.52 Nm. At the end of stance when the ankle plantarflexes approximately 20° further, the DFA applied a moment of ~1.33 Nm. This moment is relatively small compared to the maximal muscle moment seen in normal walking (Winter, 1991).

We used the method previously described by Modica and Kram (Modica and Kram, 2005) to assist with leg swing. This device (the LSA: Fig. 2) applied a forward pulling force to each leg at the beginning of the swing phase, effectively reducing the hip flexor muscle activity (e.g. rectus femoris) needed to swing the leg forward after toe-off (Gottschall and Kram, 2005; Modica and Kram, 2005). These previous studies also reported that the LSA did not increase biceps femoris or vastis lateralis activity during late stance but did increase activity to decelerate the leg during late swing. A force of 3% of body weight was found to be optimal for reducing metabolic cost while walking at a preferred walking speed ($1.25\ \mathrm{m\ s^{-1}}$) (Gottschall and Kram, 2005). However, using the leg swing assist for walking at faster speeds can become awkward. We determined that applying 1.5% of body weight was more comfortable for subjects at fast walking speeds yet it still decreased the demand on hip flexor muscles (RF).

We used the method previously described (Gottschall and Kram, 2003) to apply aiding and impeding horizontal forces (AHF, IHF). In short, elastic bands applied a constant pulling force at approximately the center of mass (Fig. 2). A force of 10% of body weight was applied for both conditions. Gottschall and Kram demonstrated that plantar flexor muscle activity decreases with aiding horizontal forces and increases with impeding horizontal forces (Gottschall and Kram, 2003).

Statistical analysis

We performed a repeated-measures ANOVA with a Bonferroni adjustment using a computer-based statistical package (SPSS Inc, Chicago, IL, USA) to make pair-wise comparisons of the preferred transition speed and muscle activity for the experimental conditions.

Table 1. Mean preferred transition speed for all subjects

Condition	Preferred transition speed (m s^{-1})	Condition	Preferred transition speed (m s^{-1})
Normal	1.94±0.03		
DFA	2.02±0.04*	DFA+LSA	2.03±0.03*
LSA	2.01±0.03*	LSA+AHF	1.99±0.03*
AHF	2.06±0.04*	AHF+DFA	2.08±0.04*
		COMBO	2.05±0.03*
IHF	1.80±0.03*		

Values are means ± s.e.m. (*N*=20).

DFA, dorsiflexor assist; LSA, leg swing assist; AHF, aiding horizontal force; IHF, impeding horizontal force; COMBO, DFA+LSA+AHF.

*Significantly different compared with normal PTS ($P<0.05$). PTS for combined assists were not significantly different from single assistive device conditions ($P>0.277$).

$P<0.05$ was taken as statistically significant. All mean data are given followed by standard error of the mean.

RESULTS

There was no significant difference ($P=0.104$) between the preferred transition speed before the experimental conditions PTS1 (1.93±0.03 m s^{-1}) and the preferred transition speed after the experimental conditions PTS2 (1.95±0.03 m s^{-1}). Therefore, each subject's average preferred transition speed was used to compare the effects of the following experimental conditions. Table 1 presents the mean preferred transition speeds for all conditions.

Dorsiflexors

With the DFA device while walking at PTS, integrated and mean muscle activity at heelstrike significantly decreased in the TA (–33, –34% respectively; $P=0.004$, 0.003) and transition speed was significantly faster; 2.02±0.04 m s^{-1} (Fig. 3; $P<0.001$). Burst duration for the TA muscle was not significantly different between walking normally and walking with the DFA device ($P>0.50$). While running normally at PTS, both integrated and mean TA muscle activity were significantly less than for normal walking (–20, –28% respectively; $P=0.011$, 0.002). Burst duration was significantly less for normal walking compared to normal running at PTS ($P=0.021$).

Hip flexors

With the LSA device while walking at PTS, integrated and mean muscle activity during the swing phase decreased in the RF (–26%, –39%; $P=0.010$, 0.006) and transition speed was significantly faster, 2.01±0.03 m s^{-1} (Fig. 4; $P<0.001$). Burst duration for the RF muscle was not significantly different between walking normally and walking with the LSA device ($P>0.83$). While running normally at PTS, mean RF activity was significantly less than for normal walking (–26%; $P=0.005$).

Plantar flexors

With the AHF device while walking at PTS, the integrated and mean muscle activity during stance significantly decreased in the SOL (–24%, –29%; $P=0.003$, 0.001), MGAS (–29%, –33%; $P=0.002$, 0.001) and LGAS (–36%, –32%; $P<0.0001$), and transition speed was significantly faster at 2.06±0.04 m s^{-1} (Fig. 5; $P<0.0001$). Conversely, when using the IHF device, the integrated and mean muscle activity during stance significantly increased in the SOL (+42%, +30%; $P<0.012$), MGAS (+43%, +42%; $P<0.003$) and LGAS (+53%, +40%; $P<0.001$), and transition speed was significantly slower at 1.80 ±0.03 m s^{-1} ($P<0.0001$). For all the plantar flexor muscles, the EMG burst durations were not significantly different between walking normally and walking with the AHF device ($P>0.867$). While running normally at PTS, integrated muscle activity during stance was significantly greater than for normal walking for the SOL (+23%; $P=0.040$), MGAS (+38%; $P<0.001$) and LGAS (+35%, $P=0.004$). EMG burst duration was not significantly different for the SOL ($P=0.279$) MGAS ($P=0.124$) or LGAS ($P=0.280$) between normal walking and running. Plantar flexor activity did not significantly increase while using the DFA device ($P=0.199$).

Combined assists

Four of the experimental conditions consisted of combinations of external assist devices. With the two-device combinations (DFA+LSA, LSA+AHF, AHF+DFA), subjects did not have significantly faster preferred transition speeds (2.03±0.03 m s^{-1}, 1.99±0.03, 2.08±0.03 m s^{-1}) compared to walking with one assistive device ($P>0.277$). Preferred transition speed with all three assistive devices (COMBO, 2.05±0.03 m s^{-1}) was not significantly faster compared to the assistive devices individually or in pairs (all $P>0.366$).

DISCUSSION

Our goal was to determine if altering the mechanical demand on specific leg muscles would affect the walk–run transition speed. The results support the hypotheses that: (1) reducing the demand on proposed trigger muscles increases the transition speed and, (2) conversely, increasing the demand on trigger muscles decreases the transition speed. Furthermore, the combination of assistive devices revealed a limit on faster transition speeds.

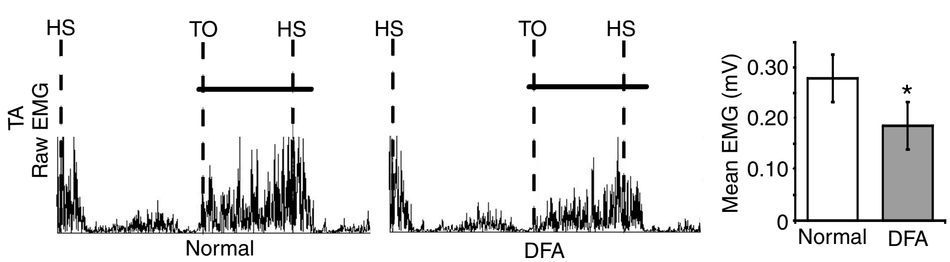

Fig. 3. Rectified electromyographic (EMG) signals (after 7 Hz high-pass filter) *versus* time [normal walking, walking with the dorsiflexor assist (DFA) device] for the TA muscle for a representative subject. Vertical broken lines indicate heel strikes and toe-off for the right leg. Black horizontal bars indicate the burst analyzed during swing and just post heel-strike (between 60–110%). The bar graph on the right depicts mean muscle activity (± s.e.m.) during the burst for all subjects while walking normally (white) and walking with the DFA device (grey). The asterisk denotes a significant difference compared with normal walking ($P=0.003$).

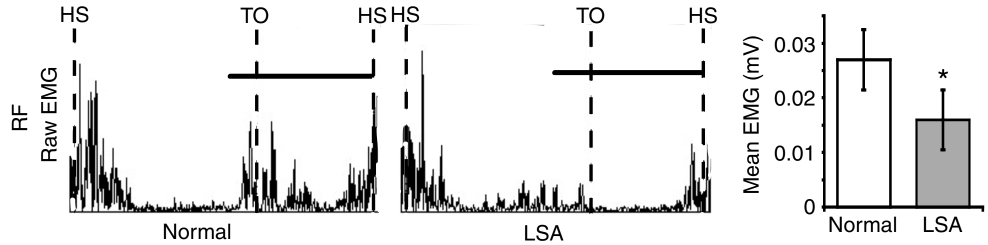

Fig. 4. Rectified electromyographic (EMG) signals (after 7 Hz high-pass filter) *versus* time [normal walking, walking with the leg swing assist (LSA) device] for the rectus femoris (RF) muscle for a representative subject. Vertical dashed lines indicate heel strikes and toe-off for the right leg. Black horizontal bars show the portion of the stride analyzed (between 50–85%). The bar graph on the right depicts mean muscle activity (± s.e.m.) for the burst during swing initiation for all subjects while walking normally (white) and walking with the LSA device (grey). The asterisk denotes a significant difference compared to normal walking (P=0.006).

Dorsiflexors

The dorsiflexor muscles activate during swing primarily for ground–toe clearance and at heelstrike to reduce foot slap. Hreljac (Hreljac et al., 2001) deduced that dorsiflexor overexertion is an important trigger of the gait transition based on increased peak muscle activity and ankle angular acceleration during fast walking. Typically, when subjects walk at their preferred transition speed or faster, they feel local discomfort or fatigue in their dorsiflexor muscles. In a recent study, it was shown that after pre-fatiguing the TA muscle, subjects preferred a slower transition speed (Segers et al., 2007). Using our DFA device, we decreased the demand of the dorsiflexors (measured in the TA muscle) and increased the preferred transition speed. In Fig. 6A, we show that mean EMG increases in the TA muscle with increasing walking speed and decreases when the person switches to a run. The DFA device clearly reduces TA EMG activity across walking speed, however, if one extrapolates the data for the EMG amplitude with DFA (open square symbols, Fig. 6A), that line would not intersect the 1.0 normalized EMG threshold until a much faster speed than the observed PTS with DFA. This suggests that indeed there is another factor that triggers the transition.

Hip flexors

Just before the beginning of the swing phase, the hip flexors activate to initiate the leg swing movement. We found that, at PTS, the hip flexors were more active when walking *versus* running. By decreasing the demand on the hip flexors with a leg swing assist device, we were able to increase the preferred transition speed. In

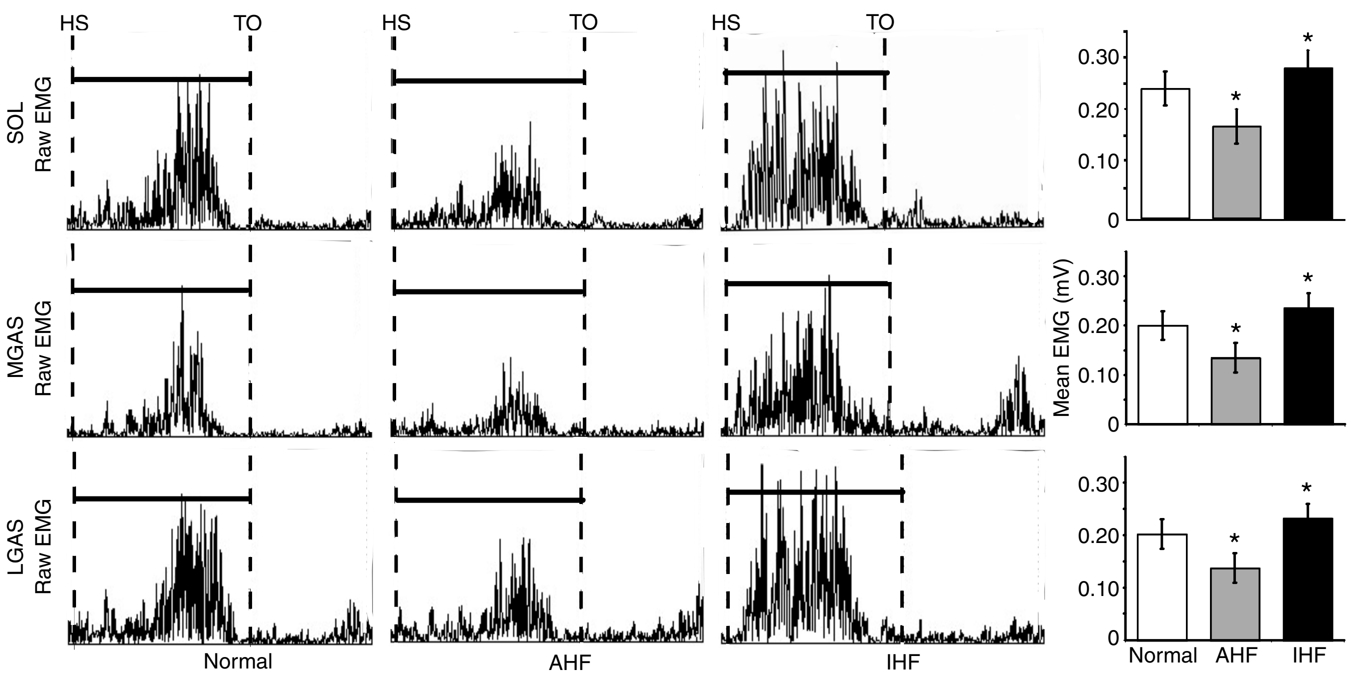

Fig. 5. Rectified electromyographic (EMG) signals (after 7 Hz high-pass filter) *versus* time [normal walking, walking with the aiding horizontal force (AHF) and impeding horizontal force (IHF) devices, respectively] for the soleus (SOL), medial gastrocnemius (MGAS) and lateral gastrocnemius (LGAS) muscles for a representative subject. Vertical lines indicate heel strikes and toe-off for the right leg. Black horizontal bars indicate the burst analyzed during stance (between 0–60%). The bar graphs on the right depict mean muscle activity (± s.e.m.) during stance for all subjects while walking normally (white), walking with the AHF (grey), and walking with the IHF (black). The asterisk denotes a significant difference in mean muscle activity compared to normal walking (all P values <0.012).

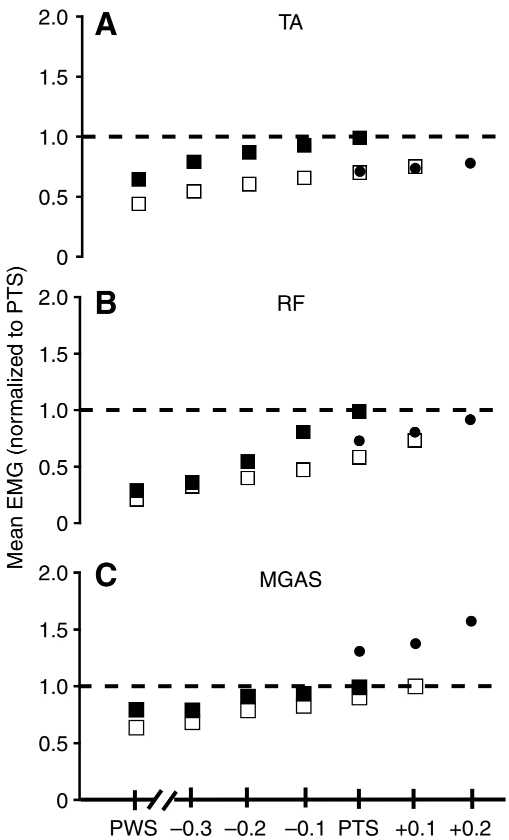

Fig. 6. Mean electromyographic (EMG) activity normalized to walking at PTS for three muscles: (A) tibialis anterior (TA), (B) rectus femoris (RF), (C) medial gastrocnemius (MGAS) during normal walking (filled squares), normal running (filled circles) and walking with an assistive device: (A) dorsiflexor assist (DFA; open squares)), (B) leg swing assist (LSA; open squares), (C) aiding horizontal force (AHF; open squares) at preferred walking speed (PWS; 1.25 m s^{-1}), preferred transition speed (PTS; 1.94±0.03 m s^{-1}), at speeds less than PTS (–0.3, –0.2, –0.1 m s^{-1}) and speeds greater than PTS (+0.1, +0.2 m s^{-1}).

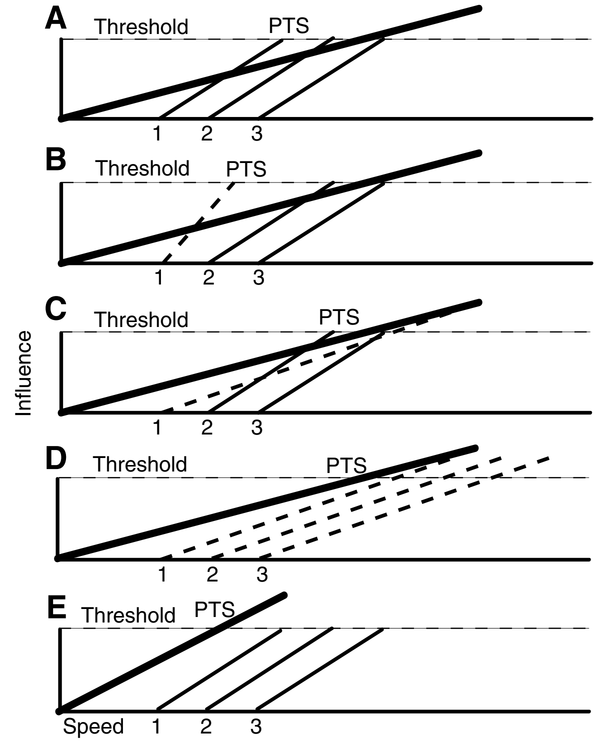

Fig. 7. Conceptual hypothetical framework illustrating the factors that influence the preferred walk–run transition speed (PTS). The horizontal axis represents increasing speed and the vertical axis represents the influence of each factor. A critical threshold for the PTS is represented by the broken horizontal lines. The thin lines 1,2,3 arbitrarily represent muscle groups. The thick line in each figure represents a proposed underlying factor that ultimately limits the walk–run transition speed. (A) PTS under normal conditions; (B) PTS at a slower speed as a result of increased demand in muscle group 1; (C) PTS at a faster speed as a result of decreased demand in muscle group 1; (D) PTS at the same speed as in C despite decreased muscle demand in all groups; (E) PTS at a slower speed as a result of another underlying factor that occurs during simulated reduced gravity.

Fig. 6B, we show that mean EMG activity increases in the RF muscle with increasing walking speed and decreases when the person switches to a run. While walking with the LSA device, RF EMG activity is less than in normal walking (Fig. 6B). Further, if one extrapolates the EMG amplitude data with the LSA (open square symbols) to faster speeds, the intersection of the RF EMG corresponds to the new PTS. This concurs with Prilutsky and Gregor (Prilutsky and Gregor, 2001), who concluded that swing phase-related muscles (notably the RF muscle) are an important determinant of the walk–run transition.

Plantar flexors

The plantar flexor muscles activate during stance for body weight support and forward propulsion. Unlike the EMG pattern of the dorsiflexors and hip flexors, the plantar flexor activity increases at faster walking speeds and continues to increase when gait is switched to a run at PTS (Fig. 6C). This pattern does not meet the criteria proposed by Hrlejac (Hrlejac, 1993) for a parameter to be considered a gait transition trigger. Those criteria are, that a trigger variable should increase with increasing walking speed and be decreased by switching to a run at PTS. However, Neptune and Sasaki (Neptune and Sasaki, 2005) still concluded that the plantar flexor muscles trigger the walk–run gait transition because of their reduced force-generating capacity during fast walking. By applying horizontal forces near the center of mass, we decreased (AHF) and increased (IHF) the force demanded from these muscles. This resulted in faster (AHF) and slower (IHF) preferred transition speeds. Applying horizontal forces near the center of mass causes similar responses in muscle activity as walking down (AHF) or up (IHF) an incline but without changing the vertical movements of the center of mass. Prior research has shown that walking down an incline (decreasing plantar flexor demand) increased the preferred transition speed (Minetti et al., 1994). Conversely, increasing plantar flexor demand by walking up an incline decreases the preferred transition speed (Diedrich and Warren Jr, 1998; Hreljac, 1995a; Hreljac et al., 2007; Minetti et al., 1994).

Local muscle triggers *versus* other factors

The present study supports the hypotheses that local sensory information in the muscles around the ankle and the hip are important proximate triggers of the walk–run transition. Each assistive device (DFA, LSA, AHF) alone significantly increased transition speed from 1.94 m s^{-1} to 2.02, 2.01, 2.06 m s^{-1}, respectively. When we combined the external assists in pairs, we expected an additive effect with the fastest preferred transition speed occurring with all three devices. Despite further decreases in muscle activity during the combined assists, we did not observe faster preferred transition speeds compared to each assist alone.

The lack of a summation effect in this study leads us to speculate about an underlying factor that ultimately determines the walk–run transition speed. In Fig. 7, we have illustrated a hypothetical relationship of local and underlying factors. The horizontal axis depicts an increase in speed, while the vertical axis depicts an increasing level of 'influence'. When a critical threshold of influence is reached, the preferred transition from a walking gait to a running gait occurs. With normal conditions (Fig. 7A), a speed is reached at which people generally prefer to transition from a walk to a run (denoted as PTS). When the demand is increased in a local muscle trigger (Fig. 7B) the preferred transition speed is slower than normal. When demand is reduced in one of these local triggers (Fig. 7C), transition speed is faster than normal and local factors such as those tested in this study, seem to adequately explain this gait transition. However, when demand is reduced in all of these local triggers (Fig. 7D), a different factor appears to triggers the gait transition. While walking in simulated reduced gravity, the dynamics of the inverted pendulum system (Kram et al., 1997) appear to trigger the walk–run transition before local triggers become a factor. In Fig. 7E, this relationship is depicted as the underlying factor triggering transition at a slower speed than normal.

It is unclear what underlying factor(s) ultimately determined preferred transition speed in the COMBO conditions. We were only able to increase transition speed up to about 2.2 m s^{-1} (an increase of 0.2 m s^{-1}). However, humans can walk up to 2.5 m s^{-1} without training (Bohannon, 1997). Humans normally choose to transition slower than the speed at which walking becomes metabolically more expensive than running (Minetti et al., 1994). Other non-muscular factors such as rates of visual flow (Mohler et al., 2007), training type (Beaupied et al., 2003), mental activity (Daniels and Newell, 2003) and perceived exertion (Noble et al., 1973) can also influence PTS. Both local factors (i.e. muscle activity, force–velocity relationships) and other underlying factors (i.e. perceived exertion, metabolic cost) have been shown to affect PTS. The subjective nature of choosing PTS might be the result of previous experience combined with input from each of these factors. Alternatively, there may be another local factor that triggers PTS at ~2.2 m s^{-1}.

Limitations

Our approach of using external assistive devices involves simplifying assumptions to identify the contributions of specific muscle groups during walking. Muscles perform multiple functions but we have categorized them as single functions (e.g. MGAS is a propulsive muscle but it is also involved in weight support and arresting leg swing). The DFA device developed for the present study effectively reduced muscle activity in the dorsiflexors. By stretching during late stance as the ankle extends, it is intuitive that plantar flexor muscle activity would increase to maintain walking speed. However, we could detect no significant increases in any plantar flexor muscle activity during DFA conditions. Similarly, the use of the LSA device may have affected other muscle groups not measured in this study. Our focus was on the muscle activity in the hip flexor muscle (rectus femoris) but recognize that the hamstrings may have been affected by the LSA device when decelerating the leg in swing. Prilutsky and Gregor determined that the hamstrings are relevant to triggering gait transition and in hindsight, we regret not measuring hamstring EMG activity. However, Gottschall and Kram (Gottschall and Kram, 2005) measured bicep femoris activity while using the LSA device and found that activity increased in late swing but not in late stance. Further, despite potentially increased bicep femoris activity while using the LSA device, we found faster gait transition speeds, suggesting a more dominant role of the hip flexors over the hamstrings.

Previously it has been shown that the use of the AHF and the LSA devices reduce metabolic cost (Gottschall and Kram, 2003; Gottschall and Kram, 2005). We hypothesized that reducing muscle demand while using these devices would indicate a trigger for the walk–run transition. It is certain that by using these devices, we reduced metabolic cost and that itself may have contributed to the faster preferred transition speeds. However, within the scope of this study, we can only speculate about the interaction of local muscle triggers and an underlying trigger such as metabolic cost.

Conclusions

We have shown that altering the demand on specific muscles can change the preferred walk–run transition speed. However, the small increases in transition speed observed and the lack of a summation effect with multiple external devices, suggests that a stronger, more underlying factor ultimately limits the preferred walk–run transition speed. Both the local and other underlying factors hypothesized to determine PTS seem to operate in a redundant system that controls gait preference.

LIST OF ABBREVIATIONS

AHF	aiding horizontal force
AHF+DFA	aiding horizontal force and dorsiflexor assist
COMBO	dorsiflexor assist, leg swing assist and aiding horizontal force
DFA	dorsiflexor assist
DFA+LSA	dorsiflexor assist and leg swing assist
EMG	electromyographic
IHF	impeding horizontal force
LGAS	lateral gastrocnemius
LSA	leg swing assist
LSA+AHF	leg swing assist and aiding horizontal force
MGAS	medial gastrocnemius
PTS	preferred transition speed
RF	rectus femoris
SOL	soleus
TA	tibialis anterior

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