Effect of Swing Limb Heel-Strike Accuracy on Force Modulation and EMG While Stepping over an Obstacle versus Initiating Gait from a Position of Quiet Stance

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국문요약

보통 보행과 장애물 보행 시작시 에서 발꿈치 닿기 (Heel-Strike)의 정확도가 힘 조절과 EMG 에 미치는 영향

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본 연구의 목적은 보통 보행과 장애물 보행 시작시에 accuracy constraints, 즉 발꿈치 닿기(swing limb heel-strike)의 정확도가 힘판(forceplate) 상에서 힘의 조절 (force modulation)과 EMG 에 어떠한 영향을 미치는지를 분석하는 것이다. 본 실험의 대상자는 힘판(forceplate)위에서 보통 보행과 장애물 보행을 하되, 대상자 앞에 놓인 표적(target)에 정확히 발꿈치 닫기 (heel-strike)를 하도록 유도되었다. 이 때 힘판 자료와 전경골근(tibialis anterior)및 가자미근 (soleus)의 근전도 (EMG)의 활동을 양쪽 다리에서 측정하였다. 대상자 앞에 놓인 표적 (target)에 정확한 발꿈치 닫기(heel-strike)가 요구되었을 때에는 발끝밀기(swing toe-off) 시간이 증가되었으며 힘판 (forceplate)상에서의 peak force 와 slope to peak force 가 감소되는 것으로 나타났다. 전경골근 (tibialis anterior)의 활동역시 큰 차이로 감소하는 것으로 나타났다. 하지만 보통 보행과 장애물 보행시의 근전도 혹은 힘판상의 자료에는 큰 차이점이 없는것으로 나타났다. 이러한 결과는 기존의 상지 (upper extremity)에서 보여준 운동 제어 (motor control)의 이론들이 하지(lower extremity)에서도 동일하게 적용될 수 있음을 보여주는 것이다.

Introduction

The investigation of programming of volitional movement has typically focused on muscle activation patterns around a single, upper extremity joint and the resultant forces or trajectories. How movement is programmed to reach a target or obtain a given force level may be simplified in terms of those variables that are controlled and those that remain invariant (Corcos et al. 1989; Corcos et al. 1990; Ghez and Gordon 1987; Gordon and Ghez 1987a; Gottlieb et al. 1989a; 1989b; Gotlieb et al. 1990). The one kind of movement is referred to as "speed-insensitive (SI) strategy", in which the rate of rise of torque profiles remains invariant and diverge only as a function of duration of movement (Gottlieb et al. 1990g). In contrast, in "speed-sensitive (SS) strategy" the rate of rise of torque profiles varies with the same speed of movement, but the duration of movements remains invariant (Cocos et al. 1989a).

The experiments for the single joint control of upper extremity (Corcos et al.1990a; Corcos et al.1990b; Ghez and Gordon 1987b; Gordon and Ghez 1987c; Gordon and Ghez 1987d; Gottlieb et al. 1989e; Gottlieb et al. 1989f; Gottlieb et al. 1990g) and non-weight bearing lower extremity movements (Monohar et al. 1998; Robichaud et al. 2000) have shown that a subject's response may be determined by variables that constrain the velocity of movement. That is, variables such as accuracy or an explicit instruction of speed will affect the kinetics of the movement. This is not the case with other variables such as distance. It appears to us that the forces associated with stepping should be analogous to those of moving different distances in the upper extremity experiments. With stepping, ground clearance is greater and swing time longer. That being the case then the invariance of the slope to the first peak Fx is in accordance with the upper extremity model.

Two previous studies (Brunt et al. 1999; Brunt et al. 2000) have provided preliminary data to suggest that principles of motor control that have been applied to upper motor extremity movements could be generalized to the lower extremity during GI.

However, there are two results of the above studies that are not consistent with the upper extremity model. First, a change in the slope to the second peak Fx of the stance limb with stepping was not predicted. Second, it has been clearly shown that the forces prior to swing toe-off determine the final velocity of GI (Breniere and Do 1986; 1991; Breniere et al. 1987; Cook and Cozzens 1976). Therefore, the slope to the first Fx peak should have changed with the accuracy constraint. There are two probable explanations for these data. The second peak force during single limb stance may be a consequence of the initial velocity and is modulated to control swing heel-strike and provide propulsive forces for the stance limb. Second, the target used for swing heel strike may have been too large to dictate the velocity of movement. This has been shown with an isometric plantar flexion task where the slope to peak force remained unchanged as the target size increased (Monohar et al. 1998).

There are two reasons that study of gait initiation may be a functionally appropriate task with which to make this comparison. First, gait initiation is motion about a single joint axis (at the ankle) where moments of force are generated that accelerate the center of mass, a movement resembling Breniere's model of an inverted pendulum (Breniere et al. 1987; Breniere and Do 1986; 1991). Second, the center of pressure in the sagittal plane, or the location from where the ground reaction force vector originates, is controlled by the interaction of the antagonist muscles at the ankle (Breniere and Do 1991; Brunt et al. 1991; Rogers and Pai 1990).

In gait initiation muscles of the lower extremities are activated stereotypically and create moments of force about the ankles and hip that rotate the body (Breniere and Do 1986; 1991; Elble et al. 1994; Lepers and Breniere 1995; Rogers and Pai 1990). Initially, there is an inhibition of tonic soleus, which is active during quiet stance followed by the onset of tibialis anterior of both the swing and stance limbs (see Figure 1). This combination is responsible for the backwards movement of the center of pressure (Breniere and Do 1986; Brunt et al. 1991; Crenna and Frigo 1991). Swing limb hip abductors also create movement of the center of pressure towards the swing limb (Rogers and Pai 1990). Thus, muscle activity at the ankle and hip tends to propel the center of mass forwards and towards the intended stance limb. Decoupling of the center of mass and center of pressure completes the first phase of gait initiation (Jian et al. 1993; Brunt et al. 1999; 2000). The second phase of gait initiation is a stepping motion where we note a fairly rapid increase in the forward velocity of the center of mass, which, in part, is controlled by the stance limb tibialis anterior followed by a burst of the stance soleus activity (see Figure 1).

In gait initiation two distinct peaks of acceleration forces are generated by the stance limb (Brunt et al. 1999; Brunt et al. 2000) (see Figure 1). The first peak (see line c in Figure 1) coincides with stance limb loading and swing toe-off while the second peak (see line e in Figure 1) approximately coincides with swing heel-strike and precedes stance toe-off (see line f in Figure 1).

<u>Insert figure 1 about here</u>

It has been shown that the slopes to these peaks can be modulated differently. For example, when stepping over an obstacle compared to GI the slope to the first peak remained invariant but the slope to the second peak increased for stepping (Brunt et al. 1999). It was thought that these results were not unlike upper extremity experiments (Corcos et al. 1989; Corcos et al. 1990; Fitts 1954; Fitts and Peterson 1964; Freund and

Budingen 1978; Ghez and Gordon 1987; Gordon and Ghez 1987a; 1987b) and non-weight bearing lower extremity movements (Gottlieb et al. 1989a; 1989b) where the modulation of the slope of force has explained how individuals may program a given movement.

Based on previous studies of GI and stepping (Brunt et al. 1999; Brunt et al. 2000) I undertook the present study to find how individuals modulate forces and EMG with manipulating limb trajectory, for example stepping over an obstacle as opposed to GI. By investigating should provide improved data that describes how we modulate forces during voluntary movement from quiet stance.

Methods

Subjects

This study sample consisted of 9 healthy subjects with no known neurological or orthopedic deficits. All participants signed an informed written consent form.

Equipment

Surface electrodes were applied to the center of the muscle bellies of the tibialis anterior (TA) and soleus (S) of the stance and swing limb. Each recording electrode consisted of two sliver-silver chloride 1-cm diameter electrodes embedded in an epoxy mounted preamplifier system (x35) whose centers are spaced 2 cm apart. A reference electrode was attached to the medial aspect of the tibia. The EMG signals were high-passed filtered (20Hz to 4KHz) (Therapeutics Unlimited, Iowa City, Iowa) and full wave rectified on-line. Final amplification was 10 K. Two force platforms, embedded in a level walkway (5m in length and 1.22m in width), measured ground reaction forces of the stance and swing limb. Foot switches (B & L Engineering, Los Angeles) were placed in the shoes to measure heel-strike of the swing limb and heel-off of the stance limb. EMG and force platform signals were sampled on-line at a rate of 1,000 Hz for 2 seconds (BIOPAC Systems, Goleta, CA).

Procedures

Experimental setup is shown in Figure 2. For each trial subjects stood in a predetermined position with each foot on a force platform. Subjects were asked to begin walking or stepping over a 10 cm high obstacle at a fast speed. Prior to experimental trials the average position of the swing heel-strike was determined for each subject through video analysis. For half of the experimental trials a 3 cm in diameter target was placed on the ground to dictate the position and accuracy of swing limb heel-strike. Subjects completed 12 trials in each of the following conditions: GI and GI to the target and stepping over an obstacle and stepping over an obstacle to the target. The few trials in which the subject missed the target were repeated.

Insert figure 2 about here

Data analysis

Analysis of variance techniques were used to determine main and interaction effects. Single degree of freedom mean contrasts ($p < 0.05$) were used to determine the source of any significant interaction effects (Keppel 1973). The independent variables were initiation condition (GI and stepping) and accuracy (target, no target). The dependent measures included EMG amplitude, slopes and peak of the acceleration ground reaction force, and timing events of GI. The amplitude of stance and swing limb TA was determined between the onset of movement and time to swing peak Fx (TAEMG1) and between time to swing peak Fx and swing toe-off (TAEMG2). The amplitude and duration of swing S activity were also calculated. The onset and offset of S activity was visually determined with the use of an interactive cursor of 1 ms resolution Time to swing limb toe-off, swing heel-strike, stance heel-off and stance toe-off were also determined. Timing data was referenced from the onset of movement, which was defined as the first detectable onset of force platform activity. The dependent measures except for swing heel strike and stance toe-off (see Figure 1) are shown in Figure 3.

Insert figure 3 about here

Results

Tibialis anterior and soleus EMG

The amplitude of the TA EMG was determined from both the onset of force platform activity to peak swing Fx and from peak swing Fx to toe off of the swing limb. These data are shown in Table 1. There was a significant main effect for target for swing TAEMG1 ($F\ (1, 8) = 7.75$, $P < 0.05$) with decreased amplitude for the target conditions. There was no significant effect for pre-swing TAEMG2. For stance TAEMGI there was a significant interaction ($F\ (1,8) = 12.12$, $P < 0,05$) (see Figure 4).

Insert figure 4 about here

GI stance TAEMG1 was greater than stepping with no target but there was no difference for the target conditions. Stance TAEMG1 was significantly greater for the no target conditions. For stance TAEMG2 the no target conditions were significantly greater than the target conditions ($F\ (1,8) = 13.31$, $P < 0.01$). The mean data demonstrates that, except for pre-swing EMG, the TA amplitude is clearly greater for the no target conditions. S EMG amplitude was similar for all conditions although its duration was 24 ms greater for the target condition ($F\ (1, 8) = 8.05$, $P < 0.05$).

Ground reaction forces

Mean data for the ground reaction forces are shown in Table 2. There was a significant main effect for target for both peak swing Fx ($F(1, 8) = 30.88$, $P < 0.001$) and the slope to peak swing Fx ($F(1, 8) = 18.2$, $P < 0.01$). Values were greater for the no target conditions. There was no significant difference between GI and stepping for either dependent variable. For stance peak Fx there was a significant interaction ($F(1, 8) = 13.64$, $P < 0.01$). As can be seen from Figure 5 the interaction was due to a larger effect between GI and stepping for the no target condition versus the target condition. However, both these comparisons were significant ($P < 0001$ for no target and $p < .05$ for the target condition). The results for stance peak Fx are therefore reported as main effects. These main effects indicate that peak Fx GI was greater than stepping ($F(1, 8) = 30.07$, $P < 0.01$) while the peak was also greater for the no target conditions ($F(1, 8) = 24.80$, $P < 0.01$) (Figure 5). For the slope to peak Fx there was a main effect for target only where the slope was greater for the no target condition ($F(1, 8) = 22.27$, $P < 0.01$).

<u>Insert figure 5 about here</u>

Temporal events

Mean data for the temporal events are shown in Table 3. Although there was an interaction ($F(1, 8) = 6.47$, $P < 0.05$) for time to peak swing Fx (the target condition for stepping was greater than that for GI, see Figure 6), the mean time to peak Fx for the target conditions were significantly greater than for the no target conditions. For the time to swing toe-off there were significant main effects for both initiation and target condition ($F(1, 8) = 10.87$, $P < 0.05$ and $F(1, 8) = 19.29$, $P < 0.01$ respectively). Time to swing toe-off was longer for stepping to the target but shorter for the obstacle condition. For time to swing heel strike and stance toe-off there were main effects for initiation and target condition. ($F(1,7) = 7.83$ to 18.24, $p < 0.01$ to < 0.05). Times increased with both the target and stepping.

Insert figure 6 about here

Discussion

In the present study the accuracy constraint clearly decreased the speed of gait initiation. Swing toe-off appears to be the first significant event that for the no target condition occurred approximately 90 ms earlier than the target conditions. This decrease in time was, in part due to a 15 % (61 ms) decrease in time to peak swing Fx for the no target conditions. This decrease in time to peak Fx was due to a 78% increase in slope to peak Fx that resulted in a 55% increase in peak Fx of the swing limb. It appears that subjects modulated the rate of rise of force and kept time to peak force relatively constant to achieve the increase in the peak acceleration force of the swing limb. As the TA controls the backwards movement of the center of pressure (Breniere and Do 1986; 1991; Breniere et al. 1987; Brunt et al. 1991) there was therefore a significant increase in swing TAEMG1 for the no target conditions. The relationship between the backwards movement of the center of pressure (Breniere et al. 1987; Lepers and Breniere 1995) or TAEMG (Cook and Cozzens 1976; Lepers and Breniere 1995) and gait velocity has previously been recognized. Of interest, is that the time from swing toe-off to swing heel-strike remained invariant for both GI and stepping, a finding supported by the earlier data of Breniere et al. (Breniere and Do 1986; 1991; Breniere et al. 1987). In the present study, the mean difference between the accuracy conditions was only 23 ms. In addition, there was no main effect for the amplitude of swing TAEMG2 or soleus. That the soleus creates heel rise in preparation for the first step but does not contribute towards gait velocity has been recognized previously (Lepers and Breniere 1995). The contribution of the swing limb to the velocity of initiation (GI or stepping) appears, therefore, to be determined by TAEMG1 and the slope to swing peak Fx (Lepers and Breniere 1995).

For the stance limb there was a significant target effect for both TAEMG1 and TAEMG2. However, for both stepping and GI the amplitude of TAEMGI was far greater than TAEMG2. The overall mean of TAEMGl was .024 mv.s compared to .0095 mv.s for TAEMG2. However, minimal acceleration force was generated until the slope to stance Fx (see Figure 3). The onset of this slope coincided with peak Fx of the swing limb, our selected time division between TAEMG1 and TAEMG2. During this phase of GI and stepping the swing limb is loaded and the stance limb unloaded. This transition from unloading to loading occurs slightly before peak Fx of the swing limb. Given the greater amplitude for TAEMG1 of the stance limb but less peak force then it is our belief that the smaller Fx is probably related to the unloading of the stance limb and slope to stance peak Fx to the loading of that limb prior to swing toe-off. The slope to stance peak Fx was the same for GI and obstacle but peak Fx was less for obstacle. Peak Fx coincides with swing limb toe-off and the earlier toe-off for stepping (and therefore a smaller peak Fx) is thought to be related to trajectory of the swing limb (Brunt et al. 1999). Because of the greater trajectory then the time to swing heel-strike from toe-off was longer for the stepping condition. These data support our analogy of swing limb trajectory to distance moved in the upper extremity experiments.

The slope to stance peak Fx increased with the no target condition. There was an increase of approximately 50% for the no target condition with only a 14% difference in time to peak Fx. The increase for the slope to peak Fx of the swing limb was 78%. For the stance limb these data differ from the previous paper (Brunt et al. 2000) which used both a large and a small target. However, the target size used in the present study was 50% smaller than the small target used in the previous study (Brunt et al. 2000). In an isometric plantar flexion task the slopes to peak forces were also found to be the same once the target size had reached a critical level. That is, the targets were too large to influence movement velocity (Monohar et al. 1998). The fact that an explicit instruction to increase the velocity of GI (as opposed to accuracy constraints) resulted in an increase in the slope to stance peak Fx supports this interpretation (Brunt et al. 1999).

The velocity of GI and stepping increased when no accuracy constraints were placed on swing limb heel-strike. This increase was a result of the modulation of both the stance and swing limb ground reaction forces prior to swing toe-off. The time to these peak forces remained relatively constant (a decrease of 100 ms for the no target conditions). That is, the slope to the swing and stance limb peak Fx was modulated while the time to peak Fx remained relatively constant. This notion is supported by the strong relationship between the slope to swing limb Fx and time to swing heel-strike ($r = 0.84$) and a more modest relationship ($r = 0.64$) for the slope to stance peak Fx. In addition, there were no differences in the slopes to either swing or stance peak Fx between GI and stepping. This concurs with the upper extremity literature where distance moved did not affect the rate of rise of force. Based on these data it does appear that GI and/or stepping may prove to be tasks by which to measure motor control. This would perhaps be preferable to upper extremity experiments where the subject is sitting and the extremity is stabilized. Voluntary movement from upright stance may be a useful task to assess changes in performance following rehabilitation.

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