Analysis of Rapid Stopping During Human Walking

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Hase, K. and R. B. Stein. Analysis of rapid stopping during human walking. J. Neurophysiol. 80: 255–261, 1998. The mechanisms involved in rapidly terminating human gait were studied. Subjects Indeed, termination of gait has been rarely studied (Frank et al. 1995; Jaeger and Vanitchatchavan 1992; Jian et al. 1993). Jian et al. (1993) studied a controlled stop in which the subject came to rest with the feet side by side on two force plates. However, in our studies we asked the subjects to stop quickly, and they almost always stopped with one foot in front of the other. Jaeger and Vanitchatchavan (1992) collected data in which subjects stopped either with the feet together or with one foot in front of the other. The latter strategy saved approximately half a second in the time to final foot placement and changed the point in the cycle at which the subject decided to take another step.

Termination of human gait requires a deceleration of the forward momentum of the body and assumption of a stable posture. Jian et al. (1993) considered that stopping biomechanically was merely a mirror image of initiating gait, but this cannot be true of rapid stopping, when the feet end up displaced from one another. Certainly, the muscles involved in the torque generation must be different. Jaeger and Vanitchatchavan (1992) felt that the two major mechanisms in stopping were a reduction in anterior force during the fast brake period and an increased braking force in the posterior direction. Frank et al. (1995) suggested that deceleration of gait during the fast brake period was due to activation of plantarflexor muscles.

None of these studies recorded electromyographic (EMG) activity so the neurophysiological mechanisms for stopping are really unknown. Because a stimulus can occur at any time in step cycle, different mechanisms may be used depending on the time of the stimulus. The factors involved in a decision to take another step or not remain obscure. In this study, we will provide the first EMG analysis of these issues in normal, adult human subjects.

METHODS

Subjects and general procedure

Eight normal subjects (4 male and 4 female) with ages ranging from 26 to 57 yr participated in this experiment with informed consent. The subjects were asked to walk up and down a room 8-m long with a comfortable gait speed (2.2–3.3 km/h) and to stop walking as soon as they got a cue. We used an electrical stimulation of the superficial peroneal nerve in the anterior surface of the right leg just near the crease of the ankle joint as the cue for the subjects to stop walking. Flexible, disposable, Electrotrace Ag/AgCl surface EMG electrodes (Jason, Huntington Beach, CA) for stimulation were placed over a location where subjects reported a strong radiating paresthesia in the dorsal surface of the right foot. This simulates an object hitting the top of the foot. The stimulation was adjusted to the strongest intensity that was described as nonnoxious by the subjects in order not to miss the cue during walking (~3
times perceptual threshold and twice radiating threshold). The stimulus was given irregularly zero to two times during each 8 m of gait by a manual switch. To get enough data to assess various phases of the step cycle, walking continued for 20 min.

**Data recording and analysis**

After cleansing of the skin with alcohol, disposable Ag/AgCl surface electrodes were placed on the right side over the tibialis anterior (TA), soleus (SOL), biceps femoris (BF), vastus lateralis (VL), glutaeus medius (GM), and erector spinae (ES) muscles at 30 mm lateral to L₄ to record the EMG activities. Ground electrodes were placed over electrically neutral portions such as the knee. The signals from each muscle were amplified, filtered (high-pass 30 Hz) and full-wave rectified. The rectified signals were then low-pass filtered at 30 Hz.

The pressure under both feet was measured during walking with force-sensitive resistors (FSRs) connected with a nylon self-fastening (Velcro) band were adjusted to lie under the heel and the ball of the foot near the medial and lateral metatarsal joints. They were used to establish step cycle parameters, to calculate the center of pressure (COP), and to estimate the total force under the foot. Because the three FSRs only measured the forces under parts of the foot (~4.9 cm² each), the total force estimated from the sum of the three sensors underestimated body weight substantially. Zehr et al. (1995) calibrated the results with FSRs against force plate records and good agreement was seen. However, to capture the total force under the foot, the sensors were sandwiched between larger metal plates so that the subject was in effect walking on three force plates. The added weight and discomfort was not suitable for the prolonged periods walking used here (>30 min of data collection and analysis per experiment). The values obtained here show the typical pattern of vertical force generation observed using force plates (Figs. 4 and 6, ⋯⋯, the total force for control steps), although reduced magnitude. An angular position of hip, knee, and ankle were recorded with flexible electrogoniometers placed over each joint (Penney and Giles, Gwent, UK). Angles were defined according to the conventions in Winter (1991) with flexion angles being positive. Signals were amplified and recorded, together with a stimulus trigger, using a program AXOTAPE (Axon Instruments) on a computer system.

A custom software program (SELPOS) was used to detect the step cycles in which the subjects were stimulated to stop. The unstimulated step before the stimulated one was identified as the control step. Custom software programs also were used to separate the step cycle into 16 parts, beginning with the right heel contact. We obtained data for 5–25 perturbed steps for each part and >10 control steps. To obtain smoother records, stimulus artifacts were removed and the EMG activities were filtered with a three-point digital moving average filter. Six subjects were videotaped during the task in the sagittal plane, with a red flash indicating the cue to stop walking and two subjects were studied similarly in the frontal plane.

**RESULTS**

**Termination style**

Figure 1 shows all the recordings in one subject from force sensors under the heel and ball of the right foot (near the first metatarsal joints) before and after the cue to stop walking was presented (time 0). The step cycles, which lasted 1,319 ms on average for this subject, were divided into 16 equal parts. In part 1, the heel made contact just before the stimulus. In each successive part the stimulus is shown later in the cycle until in part 16, the stimulus is displayed shortly before the next heel strike. The slight variations in timing that are seen arise from the fact that each part of the cycle had stimuli occurring during a period of 1,319/16 = 82 ms.

In parts 1–4, there was no weight bearing on the heel in termination (Fig. 1A) except for one response in part 2. However, force was maintained on the ball of the foot after the first peak (Fig. 1B). This means that the right leg was kept behind the body with weight on the toes and the left leg was in front. Parts 5 and 6 were a transition period in which the right leg either remained behind the body during termination or had an extra short step due to inertia. In part 5 of Fig. 1A, for example, heel contact is seen in 2 of 12 trials after a short step. In parts 7–12, the right leg stepped forward after the cue and the left leg was kept behind the body during termination (not shown). Similarly, the transition periods in which the terminal position of the left leg could be behind or in front of the body were parts 13 and 14 of the step cycle.

Thus the termination styles for rapid stopping can be divided into two common positions. In one, the subject stopped with the right leg forward, and in the other, the subject stopped with the left leg forward. These findings were verified by video in the sagittal plane. When the stimulus was applied in parts 7–12 (~35–70% of the gait cycle from right late stance to mid swing), the rapid stopping ended with the right leg forward (Fig. 2). In parts 15–4 (85–20% of the gait cycle from right late swing to early stance phase), the mirror image was seen with the left leg forward. Parts 5 and 6 (20–35% of the gait cycle) and parts 13 and 14 (70–85% of the gait cycle) were transitional in which a decision had to be made whether to take an extra short step. Very rarely did the subject stop with the feet together, except after having taken a particularly short step in the transition periods. Even during parts 15 and 16 and 7 and 8, the stride length in the final step was often shorter than in the control step. In the two subjects studied in the frontal plane, the foot was not placed laterally in the final stance. Rather, the foot tended to step toward the midline when the cue was applied during early stance phase. Because most of the weight was taken up by this leg after stopping, it provided a better balance point for the center of gravity.

Figure 3 shows the average responses for the trials in Fig. 1 (⋯⋯) and the averages for control steps (⋯⋯⋯) in which no cues were given. In most parts where heel contact is seen (parts 7–16), the force comes off the heel and onto the ball of the foot 100–200 ms sooner than in the absence of stimulation. Similarly, when an additional step is taken with the right leg (parts 7–12), heel contact occurs ~100 ms sooner than in control steps. Thus the step cycle is generally faster after the stimulus to stop is given until the body comes to rest. Similar results on termination style and timing of the final steps were seen in all subjects studied.

**Reduction in push-off power**

Because of the symmetry of the termination styles when either the left or right leg stops in front of the body, the decelerating system in rapid stopping can be revealed by analyzing parts 15–6 for ease of presentation. Each response
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FIG. 1. Forces under the heel (A) and ball (B) of the right foot before and after a cue for a representative subject. Top: responses in part 1 (shortly after heel strike). Bottom: responses in part 16 (shortly before the next heel strike). Numbers on the right give the number of responses recorded in each part of the step cycle. Schematic diagrams show the kinematics in parts 1 or 9.

in the right leg during parts 7–14 corresponds to one in the left leg during parts 15–6. Figure 4 shows data for one subject, giving the total force measured from the three sensors of the right leg, together with the corresponding joint movement and EMG of relevant muscles in part 4 of the step cycle (cue applied in right midstance phase of the step cycle).

There was a strong inhibition of SOL EMG activity for push-off in the perturbed steps (---) compared with the control steps (····). Also, TA was strongly activated starting 150–200 ms after the cue reaching levels more than four times those seen in control step cycles. Because the right leg is kept behind the body and the weight-bearing is on the ball of the foot, the TA activation and SOL inhibition reduce the forward moment. As a result the second peak of the total force, which represents push-off power, was decreased and plantar flexion of the ankle was slowed. All subjects showed this reciprocal activity in SOL and TA during parts 1–4 of the step cycle. The decelerating effect produced by the inhibition of the plantarflexor activity for push-off resulted in a reduction of the forces on the ball of the right foot (Fig. 5A). These forces in parts 1–4 were decreased significantly (according to t-test at the 1% level of confidence). The ones in parts 13–1 on the lateral surface of the foot were reduced.

FIG. 2. Schematic diagram of the termination style. ---, right leg. Termination style in parts 15–4 was identical to this except for the left leg.
but parts 2–4 were not (Fig. 5B). This may result from the strong TA activation (Fig. 4) because this muscle produces inversion as well as dorsiflexion.

Associated with the termination style in part 4, in which almost all weight-bearing was on the left leg, a large (almost 3 times that in control steps) long-lasting discharge was seen in BF as well as in GM to hold the right leg behind the body. These discharges were decreased in parts 15 and 16 and the transitions because the weight-bearing during termination was more evenly distributed over the two legs. Finally, to prevent the torso from falling forward due to the deceleration, ES was activated bilaterally at a latency of 200 ms. Thus although the latencies are short (150–200 ms), the stopping response involves a coordinated activity of many muscles of the leg and trunk.

**Braking force in the stepping leg**

Figure 6 shows data from part 12 for the same subject, corresponding to the activities of the left leg from part 4 of the step cycle. A complementary pattern is seen compared with Fig. 4 with a large burst of SOL EMG and little activity in TA. VL and ES also were activated just before the right heel contact to maintain knee extension and stabilize the trunk. All subjects had similar tonic activity of knee and ankle extensor muscles in the final stance during rapid stopping. Continuous activity in these muscles helps to stabilize the right leg after rapid weight bearing. Moreover, the long-lasting SOL activity in the forward leg provides backward momentum after heel contact, similar to TA action in the opposite leg as described previously.

Figure 7 shows the VL and SOL EMG in various parts of the step cycle in another subject. The times when the right foot contacts the ground are indicated by arrows. VL was typically activated 50–100 ms earlier than SOL, and both were activated (——) before heel contact as in a normal step. The amplitude of both EMGs also was increased substantially in parts 6–13, and the timing of the EMG was correlated not with the cue but with the time of foot contact. Note also that the peak in the EMG activity is shifted much earlier in the stance phase than in the control steps (⋯⋯). This will prevent the forward rotation of the body over the leg, once it strikes the ground, and hence provide additional braking force by maintaining the knee and ankle in an extended position. The early SOL activity in parts 14–2 also will produce the braking force.

In the transition (parts 13 and 14), the momentum may be too great to maintain the center of mass (COM) behind the forward leg. From the videos, we saw that the body rose
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Fig. 7) extends the ankle at the time of foot contact so that the foot will rapidly become flat on the ground. Once that occurs, the extensor torque generated by the SOL will oppose any further forward movement of the leg and maintain the knee extended. The tonic activity in the SOL as well as the VL helps to keep the body behind the forward leg. Other members of the triceps surae group are two-joint muscles and will generate some knee flexor torque before foot contact. However, the activation of the VL and other members of the quadriceps group also will provide knee extensor torque. The action of the GM at the hip and the ES will work to prevent hip flexion and forward trunk movement. These actions will tend to stabilize the COM behind the forward foot at a statically stable position if the forward momentum is not too great.

Reduced push-off power in the other leg

As well as braking mechanisms in the forward leg, the stance leg, which would normally propel the body up and forward, has a complementary pattern of actions (Fig. 8). The large burst in TA and the reduced activity in SOL (Fig. 4) will reduce the plantar flexion moment at the ankle. This provides the largest thrust for push-off (Winter 1991). The ankle collapses down by the strong activity in TA and the backward momentum will be produced. The action of BF and GM will tend to keep the hip extended so the leg will stay behind the body. Again, a statically stable position with

Discussion

Depending on the point in the step cycle at which the stimulus is given, the body generally comes to rest in one of two symmetric positions. In the one illustrated in Fig. 2, the right leg is in front of the left leg. Foot placement is the key to stop walking and we identified three mechanisms that appear to play a role in quickly reaching a stable position. They will be discussed in turn.

Braking mechanisms in the forward leg

Figure 8A schematically illustrates the braking mechanisms in the forward leg. The strong activation of the SOL (Fig. 7) extends the ankle at the time of foot contact so that the foot will rapidly become flat on the ground. Once that occurs, the extensor torque generated by the SOL will oppose any further forward movement of the leg and maintain the knee extended. The tonic activity in the SOL as well as the VL helps to keep the body behind the forward leg. Other members of the triceps surae group are two-joint muscles and will generate some knee flexor torque before foot contact. However, the activation of the VL and other members of the quadriceps group also will provide knee extensor torque. The action of the GM at the hip and the ES will work to prevent hip flexion and forward trunk movement. These actions will tend to stabilize the COM behind the forward foot at a statically stable position if the forward momentum is not too great.

Reduced push-off power in the other leg

As well as braking mechanisms in the forward leg, the stance leg, which would normally propel the body up and forward, has a complementary pattern of actions (Fig. 8B). The large burst in TA and the reduced activity in SOL (Fig. 4) will reduce the plantar flexion moment at the ankle. This provides the largest thrust for push-off (Winter 1991). The ankle collapses down by the strong activity in TA and the backward momentum will be produced. The action of BF and GM will tend to keep the hip extended so the leg will stay behind the body. Again, a statically stable position with
the COM between the two legs will result if the forward momentum is not too great.

**Conversion of kinetic to potential energy**

If the effects of the first two mechanisms (preceding paragraphs) are too weak or too late in the step cycle, the momentum of the COM will carry the body over the extended forward leg. As shown in Fig. 8C, the effect also will bring the body up onto the toes. In effect, some kinetic energy will be converted into potential energy as the COM rises. This is the transition region in which either the body is stopped before the COM goes in front of its COP on the supporting foot or the body will continue to move forward (dashed arrow in Fig. 8C) and another step will be needed. At the comfortable speed that the subjects chose to walk, reaching the decision point required about one-half of a complete step cycle. In other words, if the cue occurred before the COM passed the stance foot (midstance), the subject could stop by using the swing leg to act as a forward brake and reduce the push-off power enough to maintain the stance leg on the ground. If it occurred after midstance, another step was needed.

We did not ask our subjects to use faster or slower gait speeds, but the decision point would presumably come earlier in the step cycle with the increased momentum associated with higher speeds. Pai and Patton (1997) found that subjects can tolerate higher forward velocities, if the COM is more posterior, without causing a fall. A shortening of stride length has been reported for the elderly (Ferrandez et al. 1990; Guimaraes and Issaccs 1980; Winter et al. 1990) and is considered a reflection of reduced postural competence. It would be of interest to test whether older people also change the position of the decision point in the step cycle.

Our analysis was carried out mainly in the sagittal plane, although some video recordings were done in the frontal plane. Walking is a multidimensional activity with significant lateral movements. MacKinnon and Winter (1993) re-
ported that the medial acceleration of the COM during the single support phase of gait was primarily generated by a gravitational moment about the supporting foot. Its magnitude is decided by the medial distance between the COP and the COM under the support foot so the swing phase muscles assume the frontal balance control during the next stance period (Frank et al. 1995). Considering the complexity of the stopping strategies, the speed with which the set of EMG changes can be initiated is impressive. Because the cue could come at any time in the step cycle, choosing the appropriate strategy represents a complex reaction time task. Yet, the EMG changes were evident in 150–200 ms in most muscles of the leg; this is about the lower limit for simple reaction times.

There was also little variability in the time at which the strategies were initiated and in the coupling between muscles. This suggested that the basic strategies are stored as motor programs for rapid access. The decision on what strategy to choose will depend on information about subsequent postural changes that are appropriate to each part of the step cycle to reduce the disturbance of equilibrium (Duysens et al. 1992; Hirschberg and Forssberg 1991; Nashner and Forssberg 1986). Hirschberg and Forssberg (1991) felt that the phase-dependent postural modulations during human gait suggest a continuous updating of the internal representation of the body. This representation includes the position of the body segments, support surfaces or COM, the inertia of all segments, and the forces caused by the ongoing muscle activity for locomotion. A reduction in competence to integrate all this information or a reduced power output and hence safety margin may contribute to falls in the elderly and patients with problems of the CNS. How and where this postural information and the motor programs are stored and how they can be accessed to make rapid stopping movements remain fascinating topics for future study.

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