Variability of ground reaction forces during treadmill walking

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Masani, Kei, Motoki Kouzaki, and Tetsuo Fukunaga. Variability of ground reaction forces during treadmill walking. J Appl Physiol 92: 1885–1890, 2002. First published January 11, 2002; 10.1152/japplphysiol.00969.2000.—The purpose of this study was to investigate whether or not the neuromuscular locomotor system is optimized at a unique speed by examining the variability of the ground reaction force (GRF) pattern during walking in relation to different constant speeds. Ten healthy male subjects were required to walk on a treadmill at 3.0, 4.0, 5.0, 6.0, 7.0, and 8.0 km/h. Three components (vertical (F_v), anteroposterior (F_a), and mediolateral (F_m) force) of the GRF were independently measured for ~35 steps consecutively for each leg. To quantify the GRF pattern, five indexes (first and second peaks of F_v, first and second peaks of F_a, and F_v peak) were defined. Coefficients of variation were calculated for these five indexes to evaluate the GRF variability for each walking speed. It became clear for first and second peaks of F_v and F_a peak that index variabilities increased in relation to increments in walking speed, whereas there was a speed (5.5–5.8 km/h) at which variability was minimum for first and second peaks of F_a, which were related to forward propulsion of the body. These results suggest that there is “an optimum speed” for the neuromuscular locomotor system but only for the propulsion control mechanism.

LOCOMOTION IS AN ACTION THAT is executed by repeating the same movement cyclically. It was suggested that a specific neural system, a pattern generator, located in the spine, contributes to the generation of the cyclical motor command automatically (9, 10, 13, 17). However, this pattern generator is not the only determinant of walking movements. Walking movements emerge as a consequence of the interaction, or self-organizing process, of neural and mechanical dynamic systems, including musculoskeletal dynamics, the pattern generator, modulation from the supraspinal neural system, and afferent modulation (1, 25). These multiple modulations in the neuromuscular locomotor (NML) system may induce variability in walking movements.

However, excess variability could be one factor involved in falling and could at least prevent smooth walking, because, according to system theory, excess variability implies system instability in general. Therefore, gait variability should be as small as possible. Actually, a healthy NML system suppresses gait variability well, whereas gait variability is high when parts of the NML system lose function, as in gait-disabled patients (5, 11, 14, 21). In addition, high gait variability is related to fall risk in older adults (15, 20). Thus, with regard to the gait system, low gait variability suggests that the NML system is well stabilized, and, inversely, high gait variability suggests that the NML system is poorly stabilized. Therefore, it is possible to evaluate the stability of the NML system by measuring gait variability (12, 15, 20, 21, 24).

Numerous studies have suggested that the metabolic cost per unit distance walked is minimized at usual walking speeds (6, 16, 26, 31, 32) and that mechanical efficiency is maximized at usual walking speeds (2, 7). These phenomena are generally known as gait optimization. According to this concept, it is also hypothesized that the NML system is best stabilized at the usual walking speed, that is to say, gait variability is also minimized at the usual walking speed. There have been a few studies that have addressed the minimization of gait variability. Yamasaki et al. (30) reported that the variability of step length was minimum at the usual walking speed during a treadmill walking task; the same result was confirmed by Sekiya et al. (24), who studied normal ground walking.

However, no study to date has analyzed the variability of the ground reaction force (GRF) to evaluate gait performance. The GRF is regarded as a representative measurement of gait, because it is the external force involved in walking and it affects the acceleration of the body’s center of mass (29). The goal of locomotion is to drive the center of mass stably in the desired direction. It is at the kinetic level that we can see the cause of movement rather than at the kinematic level, and, therefore, GRF may be a more appropriate global parameter to characterize gait than kinematics such as step length or step width. However, it is difficult to obtain the GRF for a number of steps at constant speed by using a standard ground-mounted measurement

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system. In recent years, several types of force platforms mounted on treadmills have been developed (3, 4, 8, 18, 19). We are now able to obtain three-dimensional GRF data easily for an adequate number of steps by using the most recent model (3, 4).

Therefore, the purpose of this study was to investigate whether or not the NML system is optimized at a unique speed with respect to the variability of the GRF during treadmill walking.

**METHODS**

**Subjects.** Ten healthy male subjects who were well experienced in treadmill walking (age 28.8 ± 5.2 yr, height 170.2 ± 6.6 cm, weight 65.2 ± 7.9 kg; mean ± SD) volunteered for the experiment. Subjects were asked to be free from pathologies likely to affect gait. They gave informed consent before taking part in the experiment, and this study was approved by the ethics committee of this university.

**Materials.** GRF was measured with a treadmill ergometer (ADAL, Tecmachine) specially equipped with four three-dimensional piezoelectric sensors (3, 4). The apparatus had two walking belts that were virtually two independent treadmills placed side by side and separated by 4 mm. The treadmills were mechanically independent to allow for separate measurements of the GRF induced by each lower limb during the support phase. For that purpose, each treadmill was built on a metallic frame and driven by its own motor, which was also fixed on the same support. All of the treadmill components, including the motors, were tightly fixed to the ground through four crystal transducers (type KI 9067, Kistler). The natural frequency of this measurement system was >300 Hz, and the linealities were ensured by the manufacturer from 0 to 3,000 N for the vertical components and 500 N for the horizontal components. In addition, Belli et al. (3) recently examined the same kind of treadmill in detail. The maximum nonlinearity was 0.2–0.7%, except for a 1.45% nonlinearity of the anterioposterior force ($F_y$). We also tested nonlinearity, and it was sufficiently <1%, similar to the level given by the manufacturer.

**Procedure and data analysis.** Subjects were required to walk on the treadmill at 3.0, 4.0, 5.0, 6.0, 7.0, and 8.0 km/h after adequate practice (1–2 min) walking at each speed. Each trial executed successively after practice was 60 s, and, therefore, we could obtain the GRF for >40 steps for each leg per each trial. Three orthogonal GRF components were recorded: vertical force ($F_z$), horizontal- mediolateral force ($F_x$), and horizontal-anteroposterior force ($F_y$). Signals from crystal force transducers were sampled at 100 Hz and stored on a personal computer via a 12-bit analog-to-digital converter. All data were low-pass filtered (cutoff frequency = 8 Hz, according to Winter (27)) with a fourth-order Butterworth filter by using a zero-phase lag (29).

Figure 1 shows a typical example of the GRF during treadmill walking at 4.0 km/h for one subject. $F_z$ typically showed two peaks with a trough at midsupport. We defined the first peak as $F_zp1$ and the second peak as $F_zp2$, $F_x$ showed a brief, laterally directed peak at foot contact, followed by a mainly medially directed reaction force during the main part of the support phase, which was defined as $F_xp$. $F_y$ had a small, initial force peak in the anterior direction, followed by a posterior-directed braking force, defined as $F_yp1$, and a propulsive horizontal force before takeoff, defined as $F_yp2$. The amplitudes were normalized to multiples of body weight. These indexes were determined automatically with the computer algorithm that we created for this study. At first, the stance phase was detected by a threshold for $F_y$, which was set at 10% of the maximum value of $F_y$ during the first 5.0 s of each trial. Then midstance was determined by the zero crossing of $F_y$. Next, peaks for $F_x$ and $F_y$ during the pre-midstance phase were measured as $F_xp$, $F_yp$, respectively, and peaks for $F_z$ and $F_y$ during the post-midstance phase were measured as $F_zp$, $F_yp$, respectively. One peak for $F_z$ was also measured during the pre-midstance phase as $F_zp$. An experimenter visually checked whether appropriate points were detected or not. Incorrectly measured force curves, due to both legs being on one force plate simultaneously, were removed by visual checks. Therefore, data from 35 steps for each leg for each trial were collected for all subjects and were subjected to statistical analysis. Steps for each leg were dealt with separately, because there was a significant bilateral asymmetry, i.e., there were significant differences for 15% of all indexes (54 of 360) between variances of right leg indexes and left leg indexes ($F$ test, $P < 0.05$), and there were significant differences for 78% of all indexes (281 of 360) between the mean values of the right leg indexes and left leg indexes ($t$-test with Welch’s correction, $P < 0.05$).

Conventional statistical methods were used to calculate means and SDs of GRF indexes, and then the variability of...
these GRF indexes were evaluated with the coefficient of variation (CV).

The effects of different speeds on CVs were tested by a one-way analysis of variance with repeated measures followed by a Tukey post hoc test (SAS, version 6.12) for each leg. The level of significance was set at $P < 0.05$.

**RESULTS**

Figure 2 shows examples of GRF variability over 20 steps at three different speeds in one subject. We can see that the variability of $F_z$ and $F_x$ increased with increments in walking speed. However, with respect to $F_y$, variability at the middle speed (5.0 km/h) was smallest compared with those at both the low (3.0 km/h) and high (8.0 km/h) speeds.

Figure 3 shows CVs of GRF indexes in relation to walking speed for all subjects and for both legs. The effects of speeds on CV were significant for all GRF indexes, except for the left $F_y$ index. In addition, Tukey's test indicated that there were increment trends of CV with speed for $F_z$ and $F_x$, whereas, for $F_y$, CVs at middle speeds (6.0 and 7.0 km/h for right $F_y$ index, and 5.0 and 6.0 km/h for both $F_y$ indexes) were significantly lower than those at both the slowest and fastest speeds. These statistical results indicated that kinetic variability increased with speeds for $F_z$ and $F_x$, whereas there was a speed at which variability was minimum for $F_y$.

To obtain the minimum variability speeds for the $F_y$ indexes, we applied quadratic regression analyses by the least squares method to each CV of the $F_y$ index. The results of these regression analyses were as follows: right $F_y$ index, $y = -37.2 - 10.7x + 0.912x^2$, $R^2 = 0.864$; left $F_y$ index, $y = 24.8 - 6.18x + 0.528x^2$, $R^2 = 0.948$; right $F_y$ index, $y = 23.8 - 6.99x + 0.623x^2$, $R^2 = 0.900$; left $F_y$ index, $y = 29.1 - 8.94x + 0.810x^2$, $R^2 = 0.991$, where $x$ is walking speed and $y$ is CV. According to these equations, the minimum variability speeds were 5.8, 5.6, 5.8, and 5.5 km/h for right $F_y$, left $F_y$, right $F_y$, and left $F_y$, respectively.

**DISCUSSION**

Before we turn to our main result, a few remarks should be made concerning measurement errors. Belli et al. (3) reported a measurement error due to running belts above the 17-Hz frequency region. We removed this noise with a low-pass filter (see METHODS). There was also measurement error $<1\%$ due to nonlinearity of this force plate, as mentioned in METHODS. This level

![Fig. 2. Qualitative visualization of GRF variability. Twenty GRF curves from 1 subject for right leg at 3 different speeds are superimposed on each graph. Nos. near each peak indicate the coefficient of variation (CV) for each index of these 20 steps. Superimposed vertical scale bar, 0.1 BW; horizontal scale bar, 0.1 s.](image-url)
of error was low compared with the range of CVs in this paper (from 1.76% for left $F_z$ at 3 km/h to 20.8% for right $F_z$ at 8 km/h). We also emphasize here that this small error did not critically affect the speed dependency of CV, because the error mixed in indexes equally for all speeds. Therefore, we consider our CV data to be reliable for the speed dependency of GRF variability during treadmill walking.

The main finding of this study was that there was a speed dependency for GRF variability. For $F_z$ and $F_y$, there was an increasing trend in variability with speed from 3 to 8 km/h. However, there was a speed at which the variability of GRF was minimum for $F_y$. The minimum variability speeds of 5.5–5.8 km/h were within the limits of usual walking speeds. This result indicates that the NML system was most stabilized at usual walking speed, that is to say, there was an optimum speed for the NML system. This result is similar to those of previous studies, which revealed that there was an optimum speed for walking in terms of energetics (2, 6, 7, 16, 26, 31, 32) and that these optimum speeds were also within the limits of usual walking speeds. These optimization phenomena suggest that we usually choose the most energetically efficient speed when we walk, and our result also suggests that, at this usual speed, the gait control system is most stable.

It should be noted that an optimum speed was found only for $F_y$ and that the variability of $F_x$ and $F_z$ increased with speed. Whereas $F_y$ affects the propulsion of the body, $F_x$ affects lateral sway, and $F_z$ affects the vertical sway of the body. Therefore, the variability of $F_x$ and $F_z$ can be regarded as representing the stability of the balance control mechanism. Thus our results suggest that optimization for the NML system is only observed in the case of the propulsion control mechanism, whereas the instability in the balance control mechanism increases with speed.

Fig. 3. Relationship between the CVs of GRF indexes and speed of walking. A: $F_z$; B: $F_y$; C: $F_x$; D: $F_y$; E: $F_y$. Each bar indicates group average ± SD. *Significantly different from CV at 3.0 km/h for corresponding leg; †significantly different from CV at 8.0 km/h for corresponding leg: $P < 0.05$. 

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It was already reported that variability in step length was minimum at around the usual speed (90–100 m/min) during treadmill walking (30) and that variability in step length was also minimum at a preferred speed during overground walking (24). The step length and $F_y$ values may interactively relate to each other, because, as the push-off force, $F_yp2$ is estimated to be the main force moving the leg forward, and $F_yp1$, as a braking force, may be strongly affected by the magnitude of step length. Actually, both step length and $F_y$ increased linearly with increments in walking speed (22, 23), suggesting that there is a linear relationship between the two. In addition, Sekiya et al. (24) also reported that variability of step width increased linearly with speed. $F_x$ and step width may also be related to each other, because kinetics would be the source of kinematics, as mentioned above. Therefore, it is possible that our major results and the results of the kinematic studies were attributable to the same mechanism.

However, the previously reported kinematic variability was comparatively smaller than kinetic variability observed in this study. From Fig. 5 in the previous study by Yamasaki et al. (30), the minimum CV of step length was $\sim 2.5\%$ for male subjects at 100 m/min, and CVs for the slowest speed and the fastest speed were $\sim 4.9\%$ at 60 m/min and $\sim 3.8\%$ at 130 m/min, respectively. These values were considerably smaller than our CVs for $F_y$, i.e., the minimum value was 4.0% for right $F_yp2$ at 5 km/h, and CVs for the slowest speed and the fastest speed were 12.6% for right $F_yp1$ at 3 km/h and 11.3% for right $F_yp1$ at 8 km/h, respectively (Fig. 3). Winter (28) investigated kinematic and kinetic variabilities simultaneously, although he measured only one subject and nine steps, and concluded that kinetic variability was larger than kinematic variability. Therefore, our speculation that our results and the results of kinematics were attributable to the same mechanism is not rejected. However, further work with simultaneous recording of kinematics and kinetics for populations is needed to test this matter.

In conclusion, we quantified the variability of the GRF during treadmill walking at different walking speeds. We discovered that the variability of the $F_y$ was minimized at the usual walking speed, whereas those of the other two components increased with increments in walking speed. This finding suggests that there is an optimum speed for the NML control system but only for the propulsion control mechanism. In general, a system that is designed for a specific condition does not necessarily work well for other conditions. Our results suggest that the gait system is adapted to execute walking at the usual speed. However, it remains unclear how this optimization occurs or why the optimization occurs only for the propulsion control mechanism. Further investigation is needed to clarify these points.

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