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Stability and Energy Criteria in Healthy and Paraplegic Subject Gait

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Abstract: The functional electrical stimulation (FES) assisted gait of paraplegic patients is inferior to that of healthy subjects. The difference can be observed in terms of speed, upright balance, biomechanical energy consumption, and generation of propulsion forces in the direction of walking. The biomechanical structure of paraplegic subjects is the same as that of normal ones; however, the mode of walking differs significantly because of the reduced number of activated muscles and primitive control. The healthy subject is utilizing a 2-point dynamically stable gait. The paraplegic patient is using 4-channel FES and utilizing a 4-point statically stable gait. We believe that the FES gait can be improved if converted into a semidynamically or dynamically stable gait. The gait is considered statically stable if the center of gravity (COG) projection on the ground (PCOG) is inside the supporting area. For a quadruped, this is only possible if it is utilizing a creeping crawl gait. In this paper, the relationship between PCOG and the supporting area are discussed as a criterion for dynamic stability assessment. Results are shown for 3 different modes of 2-point and 4-point gaits. **Key Words:** Assisted gait—Restoration of gait.

In selected spinal cord injured patients the restoration of biped gait can be realized by means of functional electrical stimulation (FES) (1). The patients are using 4-channel surface FES. The swing phase is realized through an afferent FES provoked flexion reflex, resulting in the simultaneous flexion of hip and knee and the ankle dorsiflexion, providing clearance of the foot from the ground. The stance phase is achieved by stimulating the knee extensor bilaterally, providing sufficient support to the patient utilizing crutches for balance and partial support. This FES assisted gait of paraplegic patients is significantly inferior to a healthy subject's gait in terms of biomechanics. The following are the main drawbacks. The velocity is considerably slower. The average speed achieved by a paraplegic is about 0.15 m/s while a healthy person walks at about 1.5 m/s (1). Energy inefficiency results in high energy consumption, about 14 J/kg/m for a paraplegic walking at 0.15 m/s. A healthy subject's gait at 1.5 m/s requires only 3.3 J/kg/m (1). An insufficient horizontal propulsion impulse in the direction of walking is clearly demonstrated from the amplitude ratios of propulsion forces in paraplegic (40 N) and healthy subjects (200 N) (2). The upright balance is adjusted mainly by forces provided by the hands. In addition, there are other factors, and all of the factors are interrelated in a complex manner.

The gait cannot be improved in a subjective way or simply by copying a normal biped gait. Both normal and FES assisted gaits have the same biomechanical structures consisting of segments and rotational joints, but FES results in a different gait mode. The FES gait is a quadrupedal gait because the subject is utilizing crutches for balance, partial propulsion, and support. As it is unlikely that the balance problem will be solved in the near future, the FES enabled gait will remain quadrupedal; therefore, we are dealing with the problem of how to improve the existing 4-point FES gait (3).

The basic theoretical approach of multilegged gait analysis is derived from robotics, and several walking and running machines have already been constructed. The whole analysis is performed for walking on a flat, hard, level surface. In this study, several new parameters are introduced into the analysis.

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MATERIALS AND METHODS

The supporting area in a certain phase of gait is defined as the minimum convex point set in the ground plane such that all the leg contact points are contained. All quantities use a fixed global coordinate system as shown in Fig. 1. The location of the center of gravity (COG) is defined as the first normalized moment along the given axis with Eq. 1 presenting the exact formula for COG x component x_{COG}:

$$x_{\mathrm{COG}} = \frac{\sum_{M} x_i m_i}{\sum_{M} m_i} \tag{1}$$

where M is the total body mass, m_i is mass of segment i, and x_i is the x component of the segment i. In human walking the COG is, because of its difficult assessment, often replaced with the center of body (COB), which is defined as a fixed anatomical quantity. Although such replacement could introduce significant errors, mainly in nongait activities, it is often used because of its simple application. Another important parameter is the vertical projection of COG on the ground level (PCOG). The vertical projection of COB on the ground plane (PCOB) is similar. The parameters are explained graphically in Fig. 1. The sum of the ground reaction forces (GRF) is calculated as a vector sum of forces under each supporting leg/crutch. GRF is a vector quantity and has 3 orthogonal components as described in Eq. 2:

$$\overrightarrow{\mathrm{GRF}} = \vec{F}_x + \vec{F}_y + \vec{F}_z \tag{2}$$

GRF is always pointing from the point on the ground level called the center of pressure (COP) to the COG. Therefore, COP can be defined as the point projected on the ground along the direction of the resultant reaction force acting to the COG. GRF are related by Newton's second law to the acceleration of COG as shown in Eqs. 3, 4, and 5:

$$F_x = M\ddot{x}_{\mathrm{COG}} \tag{3}$$

$$F_y = M\ddot{y}_{\mathrm{COG}} \tag{4}$$

$$F_z = M\ddot{z}_{\mathrm{COG}} + F_g \tag{5}$$

where M is the total body mass; $\ddot{x}$, $\ddot{y}$, and $\ddot{z}$ are accelerations of COG in the respected directions, F_g is gravitational force, and F_x, F_y, and F_z are forces acting on COG. An important conclusion is that the COP is always inside the supporting area while this is not true for PCOG.

A statically stable gait is a gait pattern (4) for which the PCOG is inside the supporting area during the whole gait cycle. There are 5,040 nonsingular gait patterns out of which only 3 can be statically stable. Out of these 3 creeping patterns, the crawl gait offers superior static stability properties. This type of gait is adopted by 4-legged animals for slow walking and intuitively by FES assisted paraplegic subjects. The main characteristic of statically stable gait is that a system, in our case the paraplegic subject, can maintain its posture in any gait phase for an arbitrary amount of time. Such a gait must and can be as slow as desired but is consequently unfeasible at higher velocities.

Quite opposite is a healthy person utilizing a semidynamically stable gait. This is a gait mode with both statically stable and statically unstable phases. The statically unstable states in the gait cycle occur when PCOG is outside the supporting area. When the gait cycle consists of only statically unstable states, the gait is truly dynamically stable, e.g., running. The movement assures the stability of the system in statically unstable phases. If the system stops, it falls. In statically unstable gait phases, the dynamics of the mechanism, together with the propulsion forces, determine the gait velocity. As the statically unstable phases are mandatorily gravity- and inertia-driven, the gait cannot be arbitrarily slow. Thus, developing an FES assisted semidynamically stable gait would lead to a faster and more efficient gait.

So far, no quantity or index has been proposed to enable a quantitative analysis or description to be

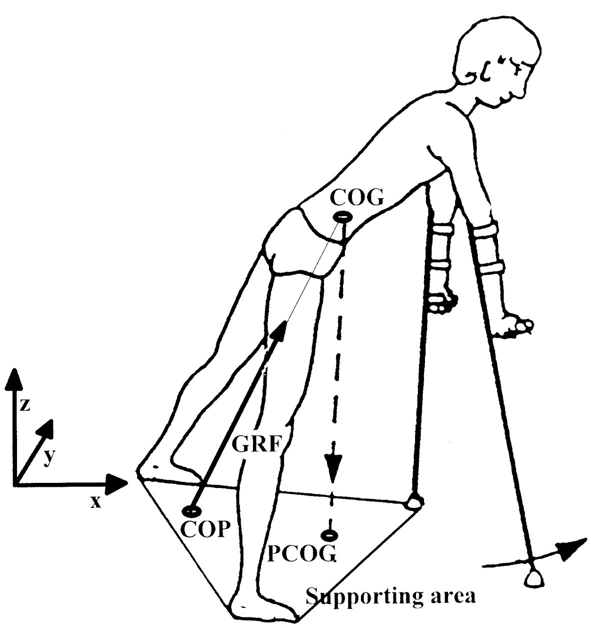

FIG. 1. The locations and relationships of the biomechanical parameters are shown.

universally valid for both static and dynamic gait modes. Therefore, numerous approximations have been introduced. When measuring system instability, the first approximation is to use the simple relationship between PCOG and the supporting area as a stability measure. This approach ignores all inertial and propulsive forces. Therefore, it is suitable only for a very slow statically stable gait. It is a reliable indicator whether the gait is statically unstable or not. In statically unstable phases of the gait, a system is definitely gravity/inertia driven and that reduces the energy consumption.

For assessment of system instability in faster but still statically stable gait, the comparison of the COP to the supporting area can be used instead of the comparison of the PCOG to the supporting area (5). COP position depends not only on system posture, but also on system inertial or ground reaction forces. In faster statically stable gait, the static stability margin is smaller than that of a slower gait. This means that the control system needs to react faster, especially when the static stability margin is small. The COP can come close to the supporting area edge, not only because the mechanism is actually close to the static stability edge or in transition to an unstable phase, but also because of the high propulsion/breaking forces. The COP-to-supporting-area relationship cannot be used in semidynamically stable gait because the COP is always within the supporting area and provides no information as to whether a system is statically stable or not.

For semidynamically stable gait, the only useful parameter is PCOG compared to the supporting area because the COP is always inside the supporting area and is therefore useless. PCOG determines whether the mechanism is, in a certain moment, statically stable or not.

To understand these basic stability criteria, we have measured the static stability index in a healthy subject biped gait, a healthy subject 4-point gait, and in a below-knee amputee gait. The dimensionless relative static stability index (SSI) is defined in Eq. 6:

$$\mathrm{SSI} = 1 - \frac{distance\,(PCOG,CS)}{\dfrac{distance\,(LSE,TSE)}{2}} \qquad (6)$$

The leading stability edge point (LSE) is the intersection point of the supporting area leading edge and the line from PCOG in the direction of COG velocity. The trailing stability edge point (TSE) is the equivalent point in the trailing supporting area edge. The center of stability area (CS) is the midpoint between LSE and TSE. Figure 2 explains the parameters used in the definition of SSI graphically.

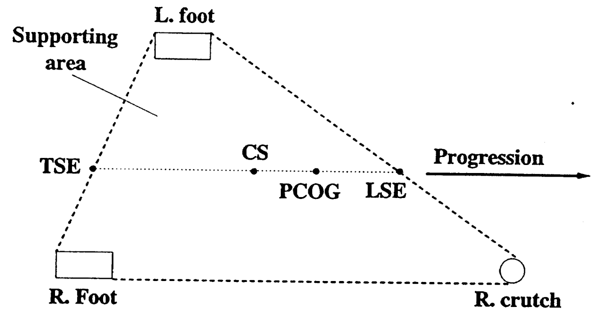

FIG. 2. Shown are the parameters for the definition of the static stability index (SSI).

When SSI is positive, a mechanism is statically stable. As SSI becomes increasingly negative, the static instability of the mechanism increases. Theoretically, there is no lower limit for SSI as the denominator in Eq. 6 can be arbitrary small. To assess the SSI, only the knowledge of the COG position and motion and the supporting area geometry is required.

RESULTS

For measuring purposes, we used the OPTOTRAK contactless motion analysis system. For easy assessment, we measured the motion of COB instead of COG. The foot-floor contact model has been characterized by 3 different types: heel contact only, toe contact only, and foot flat. The first 2 contacts were modeled as irregular triangles while the third included a rectangular area between the triangles. The actual dimensions were assessed individually. The crutch–floor contact was modeled as point contact. MATLAB was used for data processing and visualization.

Figure 3 shows the SSI for a healthy subject biped free gait. As expected, there are 2 statically unstable gait phases: the initial and final parts of the single support phase. The variance of SSI in the final part of the single support phase before the heel strike is due to the small foot–floor contact area. So even a

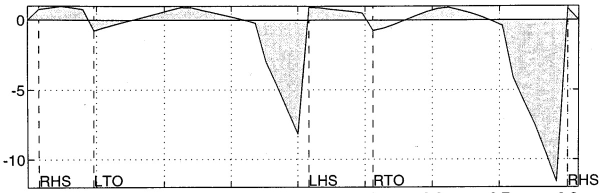

FIG. 3. The graph represents the SSI assessed in a healthy person biped gait. The events in the gait are the RHS: right heel strike; LHS: left heel strike; LTO: left toe off; and RTO: right toe off.

small variation from step to step causes significant changes in *SSI*. The distance between PCOG and CS rarely exceeds 10 cm, which is only about 12% of the vertical distance from COG to the floor. So even a small swing of COG around its vertical position enables semidynamically stable gait.

Figure 4 shows the *SSI* for a healthy subject 4-point gait in slow pace. As expected, the minimum value of *SSI* is greater than that in biped gait. However, the gait is still semidynamically stable. The variance of *SSI* is higher.

Figure 5 shows the *SSI* for a right leg below-knee amputee 4-point gait in normal pace. As expected, the negative peak of *SSI* occurs due to the rigid prosthetic foot. During this interval, the *SSI* is less than −50; the graph in Fig. 5 is clipped at −10 for better visibility. The gait is still semidynamically stable.

DISCUSSION

We hypothesize that the introduction of the unstable states into the paraplegic gait can improve its efficiency and increase its average speed. Because it is impossible for an FES walker to utilize a normal gait, we are forced to copy other 4-point gaits, which already incorporate unstable states. Above- and below-knee amputees walking with crutches are good 4-point walkers as shown in our results. They serve as a model for paraplegic FES gait. Still, an important difference exists. Amputees heavily utilize hip flexors and extensors, which are difficult to stimulate by surface electrodes and are not used in this study's described FES gait. In our future work we will try to find out if semidynamically or dynamically stable FES gait is possible at all, and if it is possible, we will try to synthesize it.

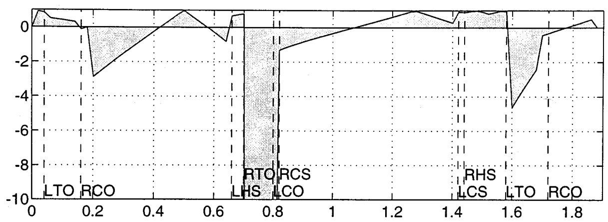

FIG. 5. The graph represents the SSI assessed in a below-knee amputee crutch-supported gait. The events in the gait are the RHS: right heel strike; LHS: left heel strike; LTO: left toe off; RTO: right toe off; LCO: left crutch off; RCO: right crutch off; LCS: left crutch strike; and RCS: right crutch strike.

To quantitatively assess the dynamic stability, a new criterion needs to be established that will include the mechanism's kinetic energy as well. *SSI* does not suffice for assessing the dynamic stability. Even if the mechanism is statically unstable, e.g., a biped walking at the beginning of the single-support phase, it can recover to a statically stable posture, e.g., the mid single-support phase, without any propulsion forces. The reverse is also true: a mechanism can slip out of the statically stable state just because of its kinetic energy, e.g., transfer from the mid to end of the single-support phase. These kinds of phenomena occur in faster multilegged gaits as well. Therefore, the understanding of dynamic stability control is essential particularly for the synthesis of semidynamically stable FES and crutch assisted gait.

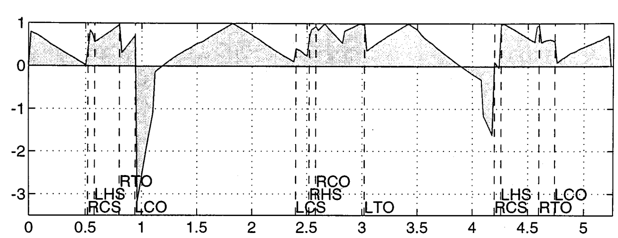

FIG. 4. The graph represents the SSI assessed in a healthy person forced quadrupedal gait. The events in the gait are the RHS: right heel strike; LHS: left heel strike; LTO: left toe off; RTO: right toe off; LCO: left crutch off; RCO: right crutch off; LCS: left crutch strike; and RCS: right crutch strike.

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