Wearable Data Collection System for Online Gait Stability Analysis

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■ **ABSTRACT**

We had shown in our previous research that the stability assessment and control are essential for generation of faster and more energy efficient functional electrical stimulation (FES) and/or crutch-assisted gait. The objective of our recent research work has been to design a wearable and portable system for gait stability analysis with online capabilities that is also applicable to crutch-assisted gait modes. The developed wearable stability assessment system for as yet only biped gait consists of foot switches and goniometers attached to the leg joints. The instantaneous static and dynamic stability is, within the wearable system, assessed from the trajectory of the estimated body center of gravity (COG_{HAT}) and the supporting area shape/size as derived from step length and foot-floor contact state. We used motion analysis system data as reference for testing the wearable system accuracy. The wearable system was tested on five healthy subjects and one above-knee amputee. It proved to be reasonably accurate if compared to the classical, motion analysis system based method. However, additional work is required to port the system to the FES assisted and/or crutch assisted gait. ■

KEY WORDS: biped gait, dynamic stability, online assessment, static stability.

INTRODUCTION

Voluntary controlled functional electrical stimulation (FES) and crutch-assisted gait of spinal cord injured (SCI) subjects is inferior to a normal free gait of healthy subjects (1). As it is unlikely for the upright body balance problem to be solved in the near future (2), we are therefore dealing with the problem of how to improve the existing four-point FES gait. The present FES gait of complete SCI subjects is a creeping gait pattern known as crawl, which exhibits superior static/kinematic stability properties; it is statically stable throughout the gait cycle (3). By definition, a statically stable gait consists of only statically stable states where each gait phase can last for an arbitrary amount of time. Contrary to a healthy person's gait, which is significantly faster than the FES-assisted one, body stability and posture is maintained by a mechanism of dynamic stability; it is

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the role of the gait to maintain the stability too. In a dynamically stable gait, it is the system dynamics that ensures the system stability, and thus the gait phases cannot last for an arbitrary amount of time (4). The stability, as described here, can also be understood as a resistance to tipping over.

From a mechanical point of view, the subject is in a statically/kinematically stable state if his center of gravity is inside the supporting area. The subject is dynamically stable if he is able to recover a statically/kinematically stable state with zero final center of gravity velocity without any action of the supporting legs. The two criteria are independent and thus both must be calculated to obtain full information about the instantaneous stability of the subject.

The stability is thus obviously the key element for faster gait synthesis as the gait control strategy heavily depends on stability (3), that is, a statically stable gait requires a different control strategy than a dynamically stable gait. The most important difference is that the former case requires a kinematics-only based controller while the latter one requires a full real-time dynamic model based control.

Up to now, a motion analysis system was required to assess the gait stability (3,5). Even the best such systems currently cannot provide reliable real-time gait data, thus we could calculate the gait stability only off-line and after the experiment. A real-time system for gait analysis is necessary to implement stability based gait control. Healthy persons, however, have no problem maintaining their balance in real-time while they use only "built-in" sensors. Therefore, one should also be able to obtain the necessary data for stability control by using artificial body-attached sensors.

The objective of our recent research work has been to design a wearable and portable system for gait stability analysis with online capabilities. In the first step, we considered only biped gait modes where no walking aids such as crutches, walkers, etc., were utilized. Typical cases include healthy person gait or gait of an amputee with a prosthetic limb. The proposed approach also can be, with certain enhancements, extended to crutch-assisted gait modes, such as the FES-assisted one. Analysis was performed for a gait on a flat, hard, and level surface. The following sections discuss in detail the design, application, and results of the wearable system for biped gait stability analysis as tested on several healthy subjects and one amputee.

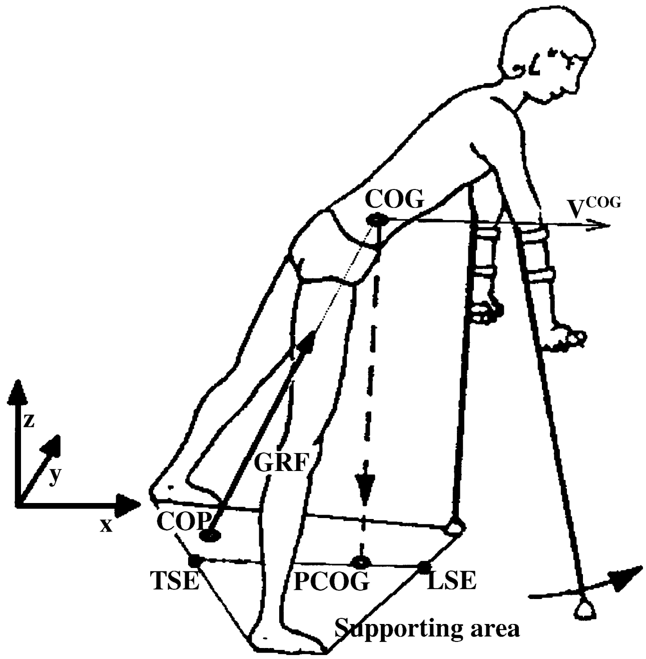

Figure 1. Biomechanical parameters used in assessment of gait stability. Abbreviations: LSE, leading stability edge; TSE, trailing stability edge; COG, center of gravity; PCOG, vertically projected center of gravity; COP, center of pressure; GRF, ground reaction force. Adapted from Ref. 4.

MATERIALS AND METHODS

Gait Stability

So far, we have mainly used two stability indices for gait stability assessment: $RKSI_1$ (relative kinematic/static stability index) and AVI (absolute velocity index) for dynamic stability (4). It is important to emphasize that both indices describe instantaneous gait characteristics.

The $RKSI_1$ is calculated as:

$$RKSI_1 = \frac{d(PCOG, CS)}{\frac{|d(TSE, LSE)|}{2}} \quad (1)$$

Parameters are explained in details in Figure 1 and Ref. 4. Metric d denotes the distance between the two points. The distance is positive if the vertical projection of the center of gravity on the ground plane (PCOG) is behind the center of the supporting area (CS) and vice versa. Thus the $RKSI_1$ is positive if PCOG is behind CS. The system is statically/kinematically stable if $|RKSI_1| \leq 1$ and PCOG is within the supporting area. The quantities used in Equation 1 change if the direction of the center of gravity (COG) instantaneous velocity changes what also affects the stability index. In the case of a wearable system we assume that the velocity direction is constant and is limited to A-P direction.

Dynamic stability index AVI is defined as:

$$AVI = \sqrt{\frac{g}{z^{COG}(t)}}\, d(PCOG, LSE) - v_x^{COG}(t) \quad (2)$$

Height of the COG above the ground level is indicated by $z^{COG}(t)$ and moves forward in the x direction. The first term is also known as the critical velocity. The AVI units are *m/s*. The system is dynamically stable if and only if AVI is positive. AVI indicates how much faster the subject could move and still remain in a dynamically stable state.

Experiments

The key element of a classical experimental setup is a motion analysis system (OPTOTRAK, VICON) providing full 3D stability assessment as described in detail in Ref. 4. Conversely, the proposed system for real-time stability analysis utilizes an approach based on movement assessment in the sagittal plane only. This sagittal plane approach and model suffice for stability assessment assuming no movement occurs in the M-L plane. The proposed system is suitable only for biped gaits, as it provides no information on position of the crutches; for crutch-assisted gait additional sensors are required.

In addition to anthropometric tables, where parameters such as segment lengths, weights, and inertial characteristics are found, the wearable system for stability analysis relies on two sets of sensors: 1) goniometers placed bilaterally on main leg joints to measure the joint angles, and 2) footswitches under both feet to determine the type of foot-floor contact,

Figure 2 schematically shows the experimental setup of the wearable sensory system. In our experiments we used Penny & Giles goniometers (Biometrics, Ltd., Gwent, UK) attached bilaterally to main leg joints. Two Tekscan pressure insoles (Tekscan Inc., South Boston, MA) provided data on foot-floor contact to model.

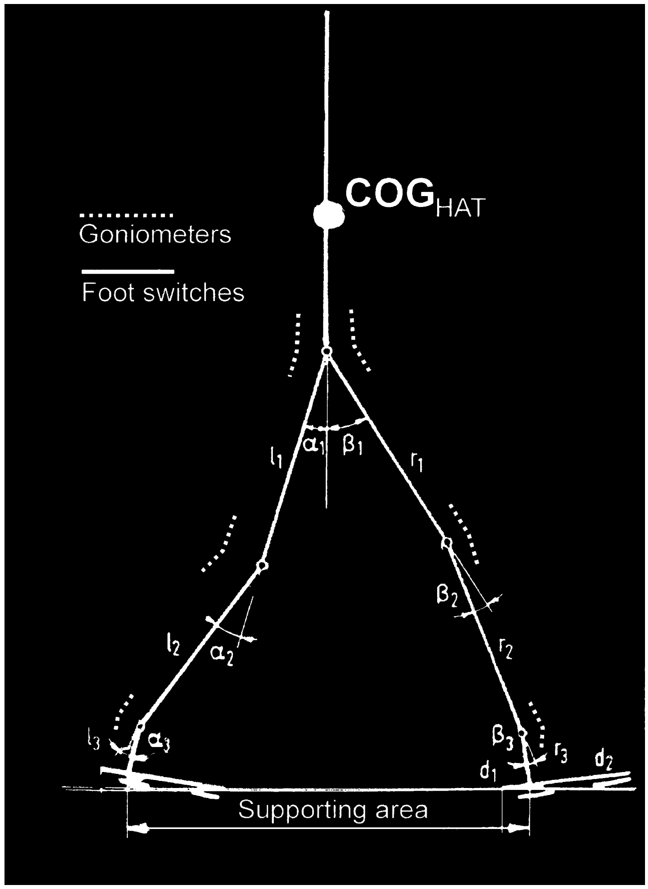

Figure 2. Wearable system setup: goniometers at main leg joints and foot switches under both feet.

Basically, two quantities are required to calculate stability indices: 1) supporting area size/length in the direction of walking (x coord.), and 2) COG trajectory (position, velocity) relative to the supporting area.

The step length, which together with foot length in ground contact corresponds to the stability area length, is easily determined from the goniometer and segment length data (6). The foot-floor contact is modeled in three ways: heel contact only, foot flat, and toe contact only. Combining these data yield supporting area length.

The COG trajectory is calculated in a similar way. Obviously, it is not possible to directly assess the COG trajectory as required by Equation 2. The COG changes with body posture and can only be determined by a full 3D motion analysis. Therefore, we have substituted the COG by a fixed point COG_{HAT}. It is defined as a COG in quiet standing, typically between 18 and 25 cm above the hip joint. Additionally, a single, rigid HAT segment, as indicated in Figure 2, substitutes the head, arms, and trunk. Movement in the M-L direction is thus neglected. Based on the model from Figure 2 we calculate the position of the COG_{HAT}, given that all leg joint angles, segment lengths and foot-floor contact types are known (6). The COG_{HAT} velocity is obtained by deriving and filtering the position data. The COG_{HAT} position relative to the supporting area is thus always known, as there is always at least one foot in ground contact during the entire gait cycle. With all that information available both stability indices can be calculated.

The data from the wearable system was processed offline, but in a way that could be used also for online stability assessment without any modification. The VICON motion analysis system was used in parallel as a reference to validate the results of the wearable system. The VICON data were processed in a classical way not suitable for real-time analysis. Prior to the gait experiments, the goniometers were calibrated, again with VICON used as a reference.

Initial testing of the wearable system was performed on five able-bodied individuals. Afterwards we proceeded with experiments on an above knee amputee, who was otherwise a well-trained sportsman. Amputee gait is asymmetrical and thus offers better verification of the proposed system as it includes irregularities while still being a biped gait.

RESULTS

Figures 3 and 4 present the results as obtained by wearable system (dashed line) vs. classical motion analysis (solid line) based methods of a gait of the above knee amputee. The subject was asked to walk as fast as possible (1.71 m/s) in order to stress the wearable system to a maximum extent. The results from the same experiment are presented in both figures. Approximately two gait cycles are shown. The basogram is shown in the bottom of each figure; black stripes indicate a

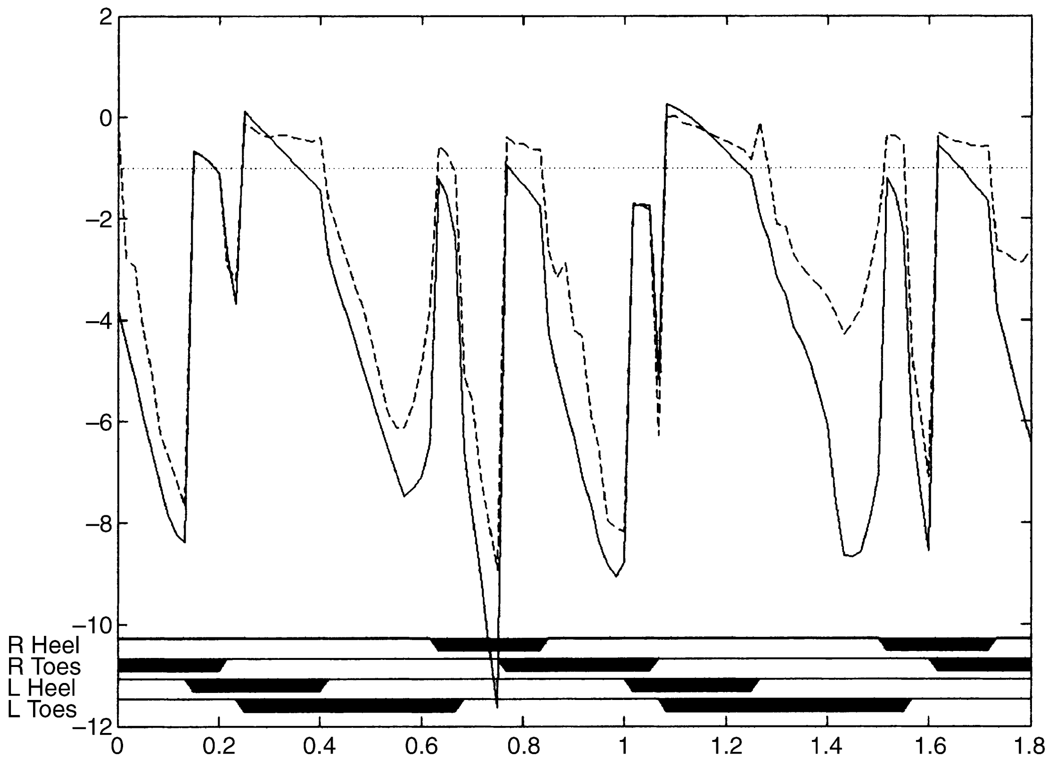

Figure 3. Static/kinematic gait stability in an above knee amputee as assessed by the wearable system (dashed line) and offline by VICON motion analysis system (solid line). Basogram in the bottom depicts gait phases, the dotted line divides stable and unstable region.

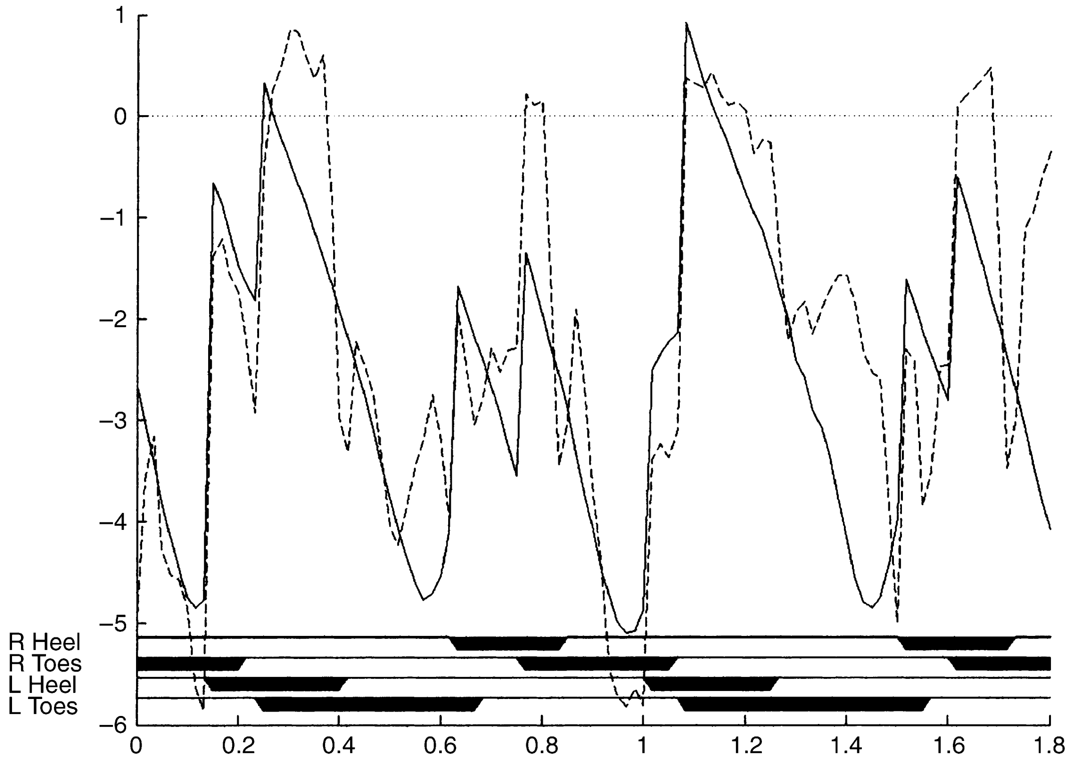

Figure 4. Dynamic gait stability in an above knee amputee as assessed by wearable system (dashed line) and offline by VICON motion analysis system (solid line). Basogram in the bottom depicts gait phases, the dotted line divides stable and unstable region.

ground contact. The dotted line divides stable and unstable regions.

Figure 3 presents the static/kinematic stability index $RKSI_1$ as measured in a classical way and as obtained by the wearable system. The $RKSI_1$ is negative most of the time. This indicates that the COG is ahead of the center of the supporting area (CS). This is typically observed in a fast gait. The peak negative values are mainly due to the low value of the denominator in Equation 1, resulting in high absolute values of index $RKSI_1$. This typically occurs during "heel only" or "toe only" foot contact during a single support phase. On the other hand, the figure clearly demonstrates pseudo-periodic characteristics in either result. The subject was statically/kinematically stable for a short period of time during double support phase and foot-flat ground contact.

Figure 4 presents dynamic stability index AVI as measured by both methods. The index is negative most of the time and the subject was in a dynamically unstable state. In the weight acceptance phase of the healthy (left) leg support phase, there are periodic, but short time intervals, when dynamically stable states occur. These roughly coincide with statically/kinematically stable states. There is also a clearly expressed asymmetry in the index value when comparing left and right leg. As expected, even high quality prostheses as used by the tested subject cannot compare to the normal intact leg.

The difference between the motion analysis and wearable system based stability assessment can be clearly seen from Figures 3 and 4. Generally the results from the wearable system resemble the reference ones based on motion analysis system data. This is particularly true for the static/kinematic index as shown in Figure 3. Typically the match between the two curves is worse during the "toe only" foot-floor contact in single support phase. For the dynamic stability index shown in Figure 4, an additional period of misalignment between the two results occurs during the initial "foot-flat" contact in single support phase. Important source of differences between the results generated by both methods originate from methods for foot-floor

Table 1. Average Correlation Coefficients *r* Between Stability Indices as Measured by 3D Motion Analysis Based Approach and the Wearable System[a]

	$RKSI_1$			AVI		
	Gait velocity			Gait velocity		
Subjects	Slow	Normal	Fast	Slow	Normal	Fast
H1	0.952	0.948	0.941	0.908	0.804	0.825
H2	0.947	0.949	0.936	0.851	0.876	0.795
H3	0.955	0.946	0.943	0.850	0.819	0.843
H4	0.948	0.947	0.936	0.880	0.795	0.789
H5	0.951	0.946	0.936	0.902	0.887	0.840
AMPUT	0.937	0.930	0.928	0.809	0.783	0.773

[a]Five healthy subjects and one amputee were tested several times under the same conditions so the numbers in the table are the average of correlation indices obtained under the same conditions.

contact detection. In the wearable system we use foot-switches for that purpose, while the classical method relies on motion analysis data. When events such as heel-strike or toe are not detected at the same time, a vastly different supporting area shape/size assessment results. That results in significant differences in stability indices between the two methods.

The results from the wearable system include high frequency components that are more pronounced. That can be attributed to the fact that the movement of the COG is the average movement of the entire body, while in the wearable system a fixed point COG_{HAT} on the HAT segment is used instead. Thus the difference between the two curves is more pronounced when the movement of the COG and COG_{HAT} differs most.

The differences between the two methods are best summarized in Table 1. The numbers represent the average correlation coefficients *r* between stability indices as measured by the classical approach and the wearable system in five healthy subjects and one amputee. Each subject was measured several times under the same conditions so the numbers in the table are the average of correlation indices obtained under the same conditions. Clearly, the correlation coefficients are higher for static/kinematic index $RKSI_1$ than for dynamic stability index AVI. Thus, the results of the wearable system are better for static/kinematic stability than for a dynamic stability assessment; the reason being that AVI calculation requires more complex data. Also as expected, the asymmetry in the amputee gait decreases the values of correlation coefficients.

DISCUSSION

The accuracy of the wearable system depends heavily on exact calibration of the goniometers used to measure the leg joint angles. Additionally the leg kinematics during the push-off phase is quite complex (6) when a limited degree of uncertainty is introduced in the calculation of the COG position. In that phase, the ankle goniometer does not provide sufficient information on the foot kinematics: the position of ankle depends on how much the heel has been lifted. There is no possible way to measure the angle between the foot and the ground with the given setup. So we have to use the simplified formula from Ref. 6 which results in less accurate results. This is also the reason why the inaccuracy in both $RKSI_1$ and AVI are so high during the push-off phase.

The inaccuracy in AVI during the foot-flat has similar cause. During the single support phase the human body forms an open kinematic chain. The COG_{HAT} trajectory is therefore calculated from the position of one leg only and the accuracy is worse if compared to the double support gait phase. Small inaccuracies and fluctuations occur in trajectory of the COG_{HAT} when calculated in that way; these in are significantly amplified by derivation required to calculate the COG_{HAT} velocity. The inaccuracy of the COG_{HAT} velocity as well as other parameters that enter Equation 2 are the main reason for inaccurate assessment of AVI.

However an important advantage of the wearable system is that it uses foot switches for detecting various gait phases. Foot switches are more accurate than motion analysis system derived data on foot-floor contact. So in terms of gait event timing the wearable system is more accurate than the classical system based on motion analysis system.

The most important question for us is whether the wearable system is accurate enough to be used in bio-feedback applications (7). It turns out that accurate measurement of the foot-ankle kinematics is of crucial importance, because even a small error results in a significant error in COG_{HAT} position; the length of the entire leg times the ankle angular error makes the difference. Subject-specific tuning through more sophisticated data processing does provide better results than those shown here, though at the expense of a more generalized approach. Specifically, the timing errors between the reference and the wearable system are still small enough even for the demonstrated case of a very fast gait where the deficiencies of the wearable system are most clearly pronounced. In slower gait modes the accuracy is of course better.

Regarding biofeedback applications, we think that the accuracy is sufficient for the example demonstrated in this paper. However, biofeedback is relevant only for more pathologic gait modes such as crutch-assisted gait in SCI subjects (7); this is also the way we foresee the usage of the developed system. The described wearable system is not directly applicable to such gait modes, as it does not provide any method for the assessment of crutch position. We believe that overall accuracy heavily depends on the successful estimation of crutch position relative to the feet, because that is the basis for calculation of the supporting area size.

Similar problems can be expected also in the COG_{HAT} vs. the COG estimation. Namely the posture in crutch-assisted gait is different from the posture in a healthy gait. Additionally there is an extensive movement of the pelvis in the A-P direction in SCI gait that heavily affects the COG_{HAT} but not the COG to the same degree. However, in FES-assisted SCI gait the trajectories of foot-ankle system are less complex as there is only a very limited plantar flexion at the push-off phase what is now an important error source. Currently, it is not possible to conclude anything reliable yet about the applicability of an extended system to the crutch and FES-assisted gait of SCI subjects.

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