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Maintaining gait balance after perturbations to the leg: kinematic and electromyographic patterns

Eleonora Croci, Roger Gassert, and Camila Shirota

***Abstract*—Maintaining balance following gait perturbations is difficult and still not well addressed in gait assistive devices. A challenge is in identifying perturbations, and whether and which responses are required to reestablish balance and walking. Here, we investigate the timing of changes in the kinematic and muscle activation patterns of unimpaired subjects to external perturbations. We used the ETH Knee Perturbator to lock the knee at different points of swing phase, and identified changes in the gait pattern with Statistical Parametric Mapping, adjusted for data containing perturbations. We show that kinematic patterns differ within approximately 100 ms of the perturbation, and that muscle activity changes later, much closer to foot-strike. Our results suggest that mechanical (joint angles and velocities) sensors are best suited to identify external perturbations, devices should change their behavior in response to such perturbations, and responses may not need to be initiated immediately following the perturbation.**

I. Introduction

GAIT assistive devices have greatly advanced in the past decades, enabling an ever-increasing range of activities to their users. However, most devices still lack the ability to respond to balance-disturbing perturbations, especially during gait. For example, the most advanced solutions available for above-knee amputees stiffen or lock the knee to prevent joint buckling [1]. However, such solutions are very limited compared to the abilities of unimpaired human knee joints to generate responses, and are further insufficient to address balance-disrupting disturbances, as indicated by the high incidence of falls in populations that use such devices [2].

We first need to better understand how unimpaired humans respond to different types of balance disturbances during gait. Although many studies in gait perturbations have been done [3, 4], they were limited by methods that can only analyze single points in the gait cycle, or single variables, as opposed to the multi-variate time-series that characterize gait. Here, we investigate the response of unimpaired human subjects to disturbances to the knee joint during different points in swing phase. We compare the muscle activity and kinematic responses to undisturbed walking using a multivariate time-series technique to identify when patterns change. These changes indicate the responses used to maintain balance, and could be used as target patterns for gait assistive devices in response to perturbations.

II. Methods

A. Experimental setup

Two subjects (30 and 25 years old, BMI of 22.1 and 22.5 kg/m^2, respectively) participated in this experiment. Subjects gave informed consent prior to their participation (swissethics, PB_2017-00160 / KEK-ZH: 2014_0049).

The ETH Knee Perturbator [5] was used to lock the right knee for 200 ms at 6 points in swing phase (10%, 22% 34%, 46%, 58% and 70%). Perturbations were repeated 15 times each, applied in random order, and with at least 30 s between perturbations to avoid anticipation. Data collection occurred in blocks of 20 min, with mandatory pauses between blocks to avoid confounding effects, e.g., due to fatigue.

EMG (Noraxon TeleMyo DTS, Noraxon, Scottsdale AZ USA) from 7 leg muscles (right and left *rectus femoris* (RF) and *tibialis anterior* (TA), and right *gluteus medius* (GMed), *semitendinosus* (ST) and *gastrocnemius* (GM)) were measured at 1500 Hz. Motion data of the pelvis and lower limbs were collected using reflective markers (Optitrack, NaturalPoint, Covallis OR USA) and exported to Visual3D (C-Motion, Germantown MD USA), where gait events (foot-strike and toe-off) and the angle and velocity of the hip and knee joints were calculated. All data were exported to and analyzed in MATLAB (The Mathworks, Natick MA USA).

EMG signals were high-passed at 50 Hz, rectified, then low-passed at 25 Hz, using 2nd-order Butterworth filters applied to have zero delay. Strides were defined between right foot-strikes. The perturbed strides and the strides immediately before each perturbed stride were extracted for analysis. Data from each stride were resampled to 101 points in time.

B. Data analysis

A modified version of statistical parametric mapping (SPM) was used to compare the perturbed and baseline (stride immediately before the perturbation) strides. SPM [6] is the multivariate time-series equivalent to a t-test, taking into account the dependencies between time points and multiple variables used to describe a system. Paired Hotelling's T^2 test ($\alpha = 0.05$) were used to compare data from each of the perturbation times to their corresponding baseline strides.

SPM outputs a test statistic as a function of time, and a threshold over which statistical significance is achieved. We adjusted the SPM threshold to the maximum value of the test statistic before the perturbation, to avoid the identification of differences that are statistically significant, but not relevant to the problem of identifying changes post-perturbation.

For each subject and each perturbation time, we compared perturbed strides to baseline strides based on EMG data only or kinematic data only. We measured the time between the perturbation (as indicated by the closing of the ETH Knee Perturbator clutch) and the first time point where the adjusted SPM indicated differences in the signal.

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III. Results and Discussion

Muscle activation (Fig. 1) and kinematic patterns differed between baseline and perturbed strides for both subjects for almost all perturbation times (Fig. 2).

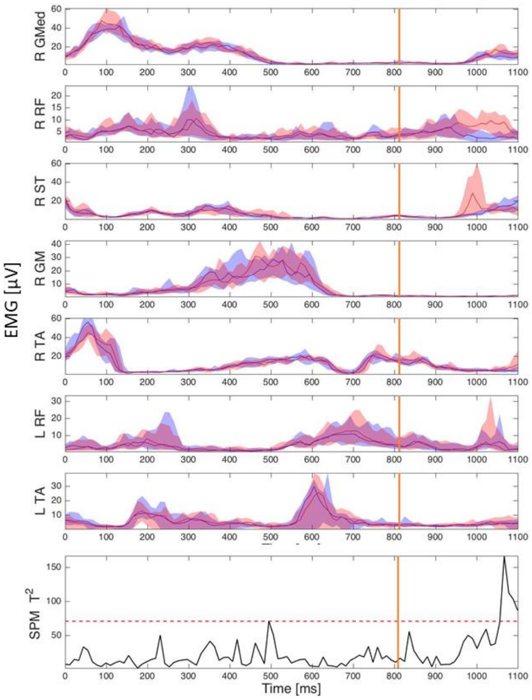

Fig. 1. Sample EMG data and SPM output for subject 1. Baseline (blue) and perturbed (red; 34% swing phase) muscle activation patterns, and the output of the adjusted SPM (bottom). Perturbation time is the vertical orange bar, and adjusted SPM threshold is the red dotted line.

Kinematic patterns differed earlier than EMG patterns in all cases. This is likely due to the nature of the perturbations, which should cause changes to the kinematics before there are changes to muscle activation patterns, as is generally expected to occur when there are external disturbances to gait. Differences in movement patterns were not immediately evident, generally occurring around 100 ms post-perturbation regardless of perturbation timing in swing phase.

In contrast, EMG patterns were much slower to change, but differences still occurred before the end of swing phase with a couple of exceptions (22% and 70% for subject 2). This indicates that active responses were used to counteract the perturbations, suggesting the need to react to external perturbations as opposed to the lack of responses in most gait assistive devices. More detailed comparisons between the kinematic and EMG responses across our perturbations would indicate whether multiple reaction patterns would be necessary, and what each would consist of.

Further, EMG responses tended to occur quicker with perturbations later in swing phase. This could be related to the amount of time available between the perturbation and the end of swing phase: the shorter the time available, the earlier reactions need to be to adequately restore balance and the baseline walking pattern. For perturbations in early swing, this delayed response could be further suggesting that responses do not need to occur immediately after the perturbation, but rather that some time can be 'spared' and dedicated to gather more information before selecting an adequate response pattern. However, data from more subjects and different types of perturbations would be needed to better understand these issues.

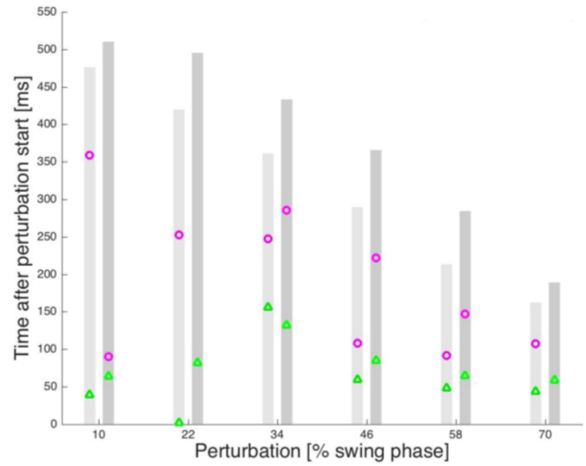

Fig. 2. Time after perturbation when perturbed strides are statistically different from baseline strides for EMG (circles) and kinematic (triangles) data, for subjects 1 (left) and 2 (right). Shaded areas represent the average time between perturbation start and end of swing phase.

IV. Conclusion

Maintaining balance after perturbations during gait requires adequate responses that are still not addressed by current gait assistive devices. In this paper, we showed that mechanical (joint angles and velocities) sensors indicate the occurrence of a perturbation earlier than muscle activity, regardless of perturbation timing. Our results suggest that mechanical sensors would be best suited to quickly indicate gait disturbances, and that assistive devices should change their behavior to help reestablish balance and the undisturbed walking pattern, based on the perturbations timing.

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