Interlimb coordination evoked by unilateral mechanical perturbation during body-weight supported gait

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Abstract— **During locomotion, motor strategies can rapidly compensate for any obstruction or perturbation that could interfere with forward progression. Here we studied the contribution of interlimb pathways for evoking muscle activation patterns in the case where body weight is externally supported and vestibular feedback is limited. The experiments were conducted using a novel device intended for gait therapy: the MIT-Skywalker. The subject's body weight was supported by an underneath saddle-like seat, and a chest harness was used to provide stabilization of the torso. Eight neurologically healthy individuals were asked to walk on the MIT-Skywalker, while one side of its split belt treadmill was unexpectedly dropped either before heel-strike or during mid-stance. Leg kinematics will be reported. We found that unilateral perturbations evoked responses at the contralateral limb, which were observed in both kinematic and neuromuscular level. The latency of most responses exceeded 100 msec, which suggests a supraspinal (i.e. not local) pathway.**

Keywords-component; gait perturbation, interlimb coordination

I. Introduction

Gait in humans have been investigated at least as early as in ancient Greece by Aristotle (384-322 BC). Interlimb coordination, particularly the maintenance of stability under various environmental perturbations, is a problem that has been intriguing researchers for more than one century [1]. Coordination of limbs during locomotion is closely linked to rhythmic activity of circuits that control different muscles [2]-[4]. On the other hand, it has been proven that sensory feedback not only assists the transition between the gait phases, but it also affects corrective responses to external perturbations [5]. Indeed, interaction of sensory inputs with those circuits' activity can determine the coordinated pattern of agonist and antagonist muscles [6]. This sensory activity contributes to motor control in two ways. It may carry "error signals" following sudden external perturbations, and it may contribute to the pre-programmed motoneuronal activity such as the Hoffman reflex (H reflex) [7] and cutaneous and stretch reflex responses [8], [9].

Various platforms and experimental protocols have been used to investigate reflex mechanisms during different phases of the gait with the majority of the experimental protocols focusing on overground walking and dropping the supportive surfaces at distinct gait phases [9]-[12].

Although most of the aforementioned studies focused on the effect of unilateral perturbations at the ipsilateral leg muscles, the bilateral response has also been studied [9], [10], [13]. During posture maintentance, experiments with powerful unilateral displacement of one leg produced bilateral responses both in adults [14], [15], and in healthy human infants [16].

Although there is enough evidence to justify the influence of unilateral perturbation to the contralateral limb, little is known whether this influence is exclusively based on the mechanisms for body stabilization and balance maintenance, or if it is also brought about from interlimb connections from gait pattern generators. Here we studied the mechanisms of interlimb coordination by applying unilateral mechanical perturbations when the body weight was supported, the position of the body center of mass was fixed, and excitation to the vestibular system was limited. A novel device designed for robot-assisted gait therapy, the MIT-Skywalker, was used.

II. Materials and Methods

A. Subjects

Eight neurologically intact subjects [6 men, 2 women, age 27 (3 SD) years] who were naive to the experimental goals were enrolled into this study. Data was collected from both the dominant and non-dominant lower limb of each participant. The dominant lower limb was determined for each subject by asking which lower limb he/she would use to kick a ball. The experimental protocol was approved by the MIT Committee on the Use of the Humans as Experimental Subjects (COUHES).

B. Apparatus

The MIT-Skywalker is a unique device intended for providing robot-assisted gait therapy [17]. It consists of a split treadmill and a body weight support, which supports subject's weight from underneath in a saddle-like manner. This system can provide support ranging from zero to 100% of the patient's weight and keep the subject safe from falls, yet not interfere with the required ranges of leg motion. The body weight support system includes a chest harness providing stabilization for the torso. Each side of the split treadmill can also be vertically actuated through a cam system. In other words, each side of the treadmill can be lowered from the ground level and raised back in a controlled fashion (see Fig. 1).

To measure the hip flexion-extension and knee flexion-extension of both legs in real time, we used a custom-made, camera-based motion tracking system. Two low-cost cameras (Logitech Quickcam Pro 9000, Logitech Inc.) were mounted on the sides of the platform at an appropriate distance to cover the whole range of leg movement (about 70 cm from each side). Two battery-powered systems equipped with two infrared LEDs were placed on each of the subject's limbs, one on the shank and the other on the thigh. Fig. 2 depicts the configuration of the sensors. The cameras were modified in order to be able to see the infrared light filtering out the visible spectrum[1]. Standard image processing techniques were used to monitor the position of the LEDs on the sagittal plane, which was aligned with the camera view plane.

C. Experimental Protocol

Subjects were fully supported with their heel barely touching the treadmill at heel-strike, i.e. fully extended knee and hip flexed at about 25 degrees with respect to the vertical axis [18], and ankle foot orthoses were attached to both ankles to minimize ankle motion during the experiment.

Four battery operated LED modules were placed on the subjects' leg using elastic Velcro straps. Electromyographic (EMG) signals were acquired from muscles of both legs using single differential bipolar surface EMG electrodes (DE 2.1, Delsys Inc.) connected to an EMG signal conditioning device (Bagnoli 16, Delsys Inc.) but these will not be reported here. The vertical motion of the treadmill was controlled by the rotation of the cam and lowered by 3.5 inches (8.9 cm) from the ground level height [19].

The subjects were first asked to familiarize themselves with the device and walk on the Skywalker at preferred speed (approx 1.7 MPH for all subjects). Experimental procedure consisted of 4 blocks lasting 5 minutes each, at speeds of 1.6, 1.7, 1.8, 1.9 MPH selected in random fashion. Gait perturbations were introduced in a random manner with the interval between two successive perturbations randomly varying between 4 and 10 steps. The perturbations were randomly presented in two different instances of the gait cycle, immediately before heel-strike and at the mid-stance of the perturbed side.

D. Data analysis

All data was analyzed offline, except that derived through the computation of the leg kinematics which was deduced from the images captured by the camera system and which was done in real-time.

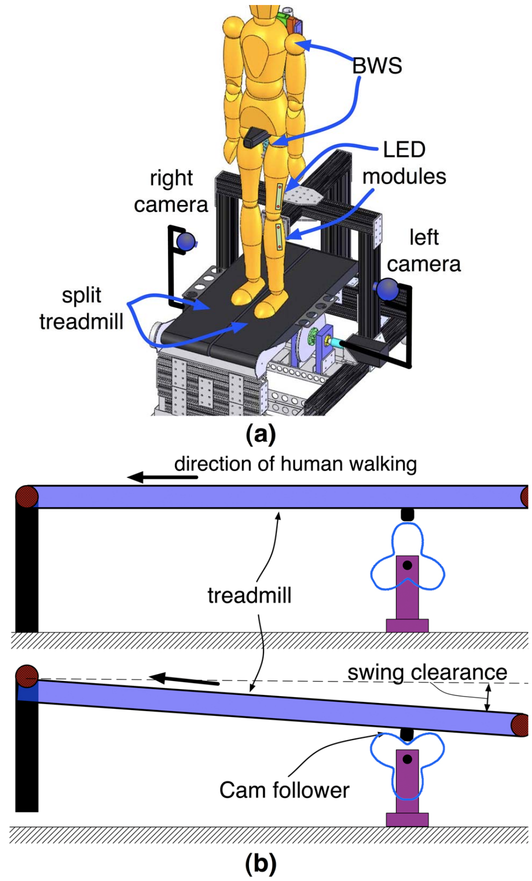


Figure 1. (a). The MIT-Skywalker platform used for the perturbation experiments. The subject's weight is supported by the body-weight support (BWS) including a saddle-like seat and a chest harness. Two cameras and four LED modules were used for leg motion tracking. (b). Cam-based mechanism of the vertical actuation of the treadmill. As the cam rotates, the follower comes into the low dwell of the cam and a swing clearance of 8.9 cm is created.

1) Leg kinematics: Only saggital plane motion was considered. Assuming that the pelvis of the subject was constrained by the body weight support of the device, the hip flexion-extension (q_1) and knee flexion-extension (q_2) angles were computed by

$$\begin{aligned} q_1 &= a\tan 2(x_2 - x_1, y_2 - y_1) \\ q_2 &= a\tan 2(x_4 - x_3, y_4 - y_3) - q_1 \end{aligned} \tag{1}$$

where x_i, y_i, $i = 1,2,3,4$ are the coordinates of the LEDs' position at the image plane (see Fig. 2). Similar equations were used for both limbs. The x-coordinate of the heel was computed from

$$x_h = L_1 \sin(q_1) + L_2 \sin(q_1 + q_2) \tag{2}$$

[1] We removed the infrared filter from the camera lens, while a photographic black thin film was placed on top of the lens to block human-visible light.

where L_1, L_2 were the lengths of the thigh and shank respectively measured before the experiment. The x-coordinate of the heel was used for determining the gait phase in real-time in order to introduce the perturbation accordingly.

III. Results

A typical set of ensemble-averaged data from a single subject is shown in Fig. 3. The results shown are from a typical subject walking at the preferred speed (1.7 MPH). The knee and hip joint angles from both the perturbed (right) and the unperturbed (left) legs are shown, along with the timing of perturbation. Traces for averaged trials for unperturbed (U), perturbed before heel-strike (PH), and perturbed at mid-stance (PM) of the right leg are shown. All data are depicted with respect to time, averaged for each of the three cases (U, PH, and PM). The gait phases for both legs are also shown, while the vertical displacement of the support surface is shown for both kinds of perturbations.

In the case of the perturbation before heel-strike (PH), the perturbed hip kinematics were significantly affected, compared to the knee kinematics. More specifically, the perturbed hip is extended at the same time the leg looses contact with the floor, but it converges to the normal pattern as soon as the floor comes up again. The perturbed knee is close to full extension as soon as the PH happens, and it continues to follow closely the normal trajectory after the perturbation. In the case of the perturbation at mid-stance (PM), the perturbed hip is not significantly affected, while the dropping of the support surface causes extension on the knee joint, which soon converges to the normal trajectory though. We must note that the differences in leg kinematics presented here, with respect to normal kinematics found in the literature [18], are caused by the fact that the motion of the pelvis is constrained by the device used.

Unilateral perturbations affected the contralateral (unperturbed) leg too. The effects on the contralateral leg can be best observed examining the joint angular velocities. Fig. 4a depicts a typical set of ensemble-averaged data of angular velocities for the contralateral hip and knee in the three cases (U, PH, PM). Moreover, the timing of the perturbations is also noted. We observed a consistent change of the angular velocities in both hip and knee joints after both perturbations. Fig. 4b shows the unperturbed hip and knee angular velocities just after the unilateral perturbations. Initial values (i.e. at the time of perturbation) have been subtracted from the profiles to allow for better comparison. In the case of PM, the left leg is at mid-swing. Therefore, when the right leg looses the walking surface, the left leg was observed to try to land earlier on the walking surface. This was realized by reducing the knee extension and hip flexion, phenomena that would both lead the left foot to touch the walking surface earlier than in the unperturbed case. In the case of PH, the contralateral leg is at mid-stance. Being in contact with the ground and essentially driven by the treadmill when the perturbation occurred, the contralateral leg did not significantly alter its pattern right after the perturbation. However, the toe-off phase was accelerated and the leg left the ground earlier. This is based on the early hip flexion and knee extension shown in Fig. 4b. Fig. 4b also shows the timing of the perturbation and the response of the

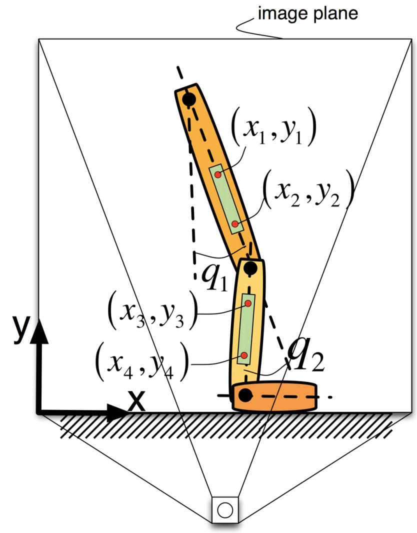


Figure 2. The computation of hip flexion-extension (q1) and knee flexion-extension (q2) is done using the coordinates of each LED at the image plane reference system, included in the LED modules. Offset values of joint angles are discarded through an initial calibration procedure.

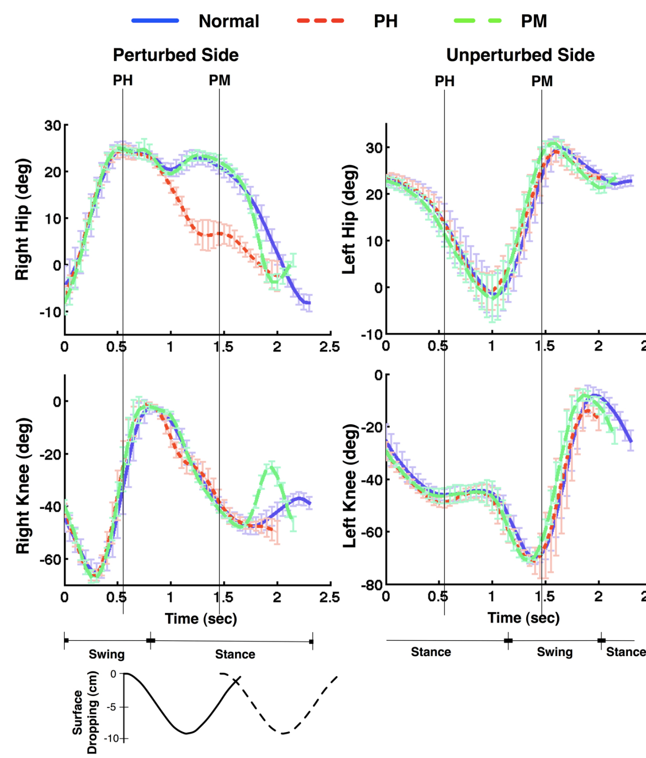


Figure 3. A typical set of ensemble-average data from a single subject at the preferred speed (1.7 MPH). The knee and hip joint angles from both the perturbed (right) and the unperturbed (left) legs are shown, along with the timing of perturbation (thin vertical lines). Traces for averaged trials for normal-unperturbed, perturbed before heel-strike (PH), and perturbed at mid-stance (PM) of the right leg are shown. Mean and standard deviation values are plotted. All data are depicted with respect to time, averaged for each of the three cases (normal, PH, and PM). The gait phases for both legs are also shown (bottom), while the vertical displacement of the support surface is shown for both kinds of perturbations (bottom left).

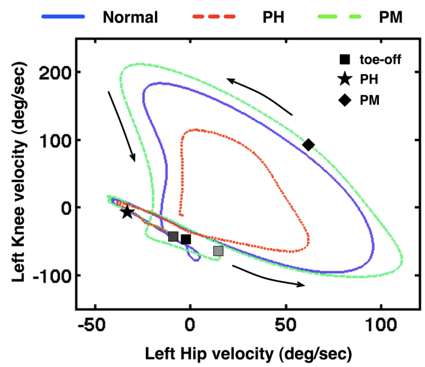
contralateral hip and knee in the angular velocity space, compared with the angular velocities in normal-unperturbed gait. The timing of response was computed from the time instance of perturbation to the point when a significant (more than 15%) difference of the angular velocity was observed. The values for the response time for the unperturbed knee and hip joints in both perturbations are listed in Table I. The response time for the unperturbed leg exceeded 100 msec, even in the PM case where the unperturbed leg was mid-swing, i.e. no external constrains were present.

IV. Conclusion

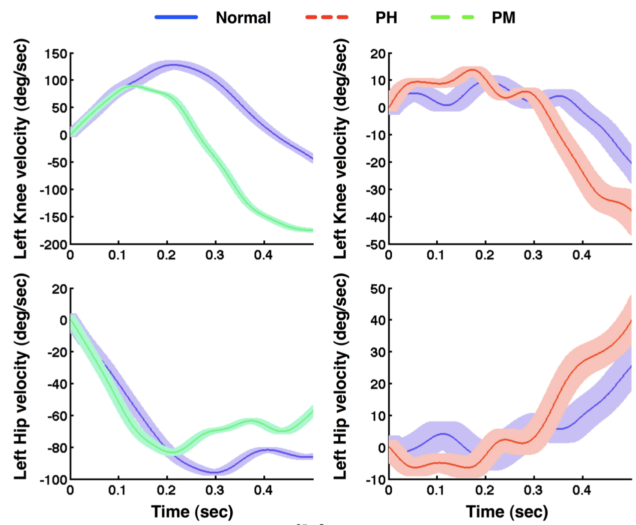
The aim of this study was to investigate the interlimb coordination by applying unilateral perturbations and analyzing the response of the contralateral limb. Unexpected lowering of the walking surface by 10 cm was induced at one leg in two different gait phases of that leg, i.e. at terminal swing before heel-strike and at mid-stance. Although the kind of the induced perturbation is similar to ones used in previous studies [9]-[12], [20], our experimental paradigm included full body weight support and torso stabilization, in order to limit as much as possible the vestibular feedback, and loading of the legs. This can reveal interesting aspects of inter-limb coordination that are not affected by mechanisms for body stabilization. In particular, it reveals that the latency of the effect of the perturbations was larger in our experiments than in any previous work. Contrary to others, our results lead to the adoption of a supraspinal, i.e. not locally-mediated pathway regulating inter-limb coordination at least for the case of body-weight-supported gait. In other words, the experimental paradigm used in this study revealed an inter-limb coordination mechanism that controlled the bilateral occurrence of walking phases and events, based mainly on proprioceptive responses, and this mechanism is more likely to be in a supraspinal level.

TABLE I. Response time for the unperturbed knee and hip joints in both perturbations.

	PM	PH
Contralateral knee response (msec)	120 (SD 22)	312 (SD 42)
Contralateral hip response (msec)	205 (SD 32)	325 (SD 54)



(a)



(b)

Figure 4. (a). A typical set of ensemble-averaged data of angular velocities for the contralateral hip and knee, in the three cases (normal, PH, PH). The starting points of the gait cycle (toe-off) measured at the ipsilateral leg for the three cases are denoted by squares, the start of the PH perturbation by a star, and the start of the PM perturbation by a diamond shape on the figure. Arrows show the direction from the start (toe-off) of the ipsilateral leg gait cycle to the end (next toe-off). Positive (negative) hip velocity denotes hip flexion (extension), while positive (negative) knee velocity denotes knee extension (flexion). (b) Ensemble-averaged data for contralateral hip and knee angular velocities after the unilateral perturbations. Initial values (i.e. at the time of perturbation) have been subtracted from the profiles to allow for better comparison. Time equal to zero denotes the start of the perturbation.

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