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Function dictates the phase dependence of vision during human locomotion

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¹Department of Kinesiology, University of Maryland, College Park, Maryland; ²Neuroscience and Cognitive Science Program, University of Maryland, College Park, Maryland; ³Biomedical Engineering Graduate Program, University of Maryland, College Park, Maryland; ⁴Laboratory of Neuromotor Physiology, Santa Lucia Foundation, Rome, Italy; ⁵Center of Space BioMedicine, University of Rome Tor Vergata, Rome, Italy; and ⁶Department of Systems Medicine, University of Rome Tor Vergata, Rome, Italy

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Logan D, Ivanenko YP, Kiemel T, Cappellini G, Sylos-Labini F, Lacquaniti F, Jeka JJ. Function dictates the phase dependence of vision during human locomotion. *J Neurophysiol* 112: 165–180, 2014. First published April 9, 2014; doi:10.1152/jn.01062.2012.—In human and animal locomotion, sensory input is thought to be processed in a phase-dependent manner. Here we use full-field transient visual scene motion toward or away from subjects walking on a treadmill. Perturbations were presented at three phases of walking to test 1) whether phase dependence is observed for visual input and 2) whether the nature of phase dependence differs across body segments. Results demonstrated that trunk responses to approaching perturbations were only weakly phase dependent and instead depended primarily on the delay from the perturbation. Recording of kinematic and muscle responses from both right and left lower limb allowed the analysis of six distinct phases of perturbation effects. In contrast to the trunk, leg responses were strongly phase dependent. Leg responses during the same gait cycle as the perturbation exhibited gating, occurring only when perturbations were applied in midstance. In contrast, during the postperturbation gait cycle, leg responses occurred at similar response phases of the gait cycle over a range of perturbation phases. These distinct responses reflect modulation of trunk orientation for upright equilibrium and modulation of leg segments for both hazard accommodation/avoidance and positional maintenance on the treadmill. Overall, these results support the idea that the phase dependence of responses to visual scene motion is determined by different functional tasks during walking.

human locomotion; phase dependence; vision

IT IS GENERALLY HELD that sensory input influences locomotion in a phase-dependent manner. Evidence suggests that sensory inputs can be gated, by either facilitation or suppression, at certain phases of the locomotive behavior (Duysens et al. 2000; Rossignol et al. 2006). For example, the vertebrate *Xenopus laevis* (tadpole) has been shown to inhibit sensory pathways from modulating motor neurons at functionally relevant phases of its swimming cycle (Sillar and Roberts 1988). Such phase dependence may provide insight into the respective roles of different sensory modalities during locomotion.

Human (e.g., Capaday and Stein 1986; Duysens et al. 1990) and animal (e.g., Forssberg et al. 1975; Forssberg 1979) studies have emphasized phase dependence of somatosensory input, illustrating how responses to both tactile and proprioceptive stimuli are modulated during the gait cycle. The focus on proprioceptive/tactile inputs arises from the suppression or facilitation of their reflex pathways at different phases of the

gait cycle to enable fast corrective responses to unexpected disturbances of balance and walking (Zehr and Stein 1999).

The use of vision has also been suggested to be dependent on the phase of the gait cycle. Patla and colleagues analyzed the use of visual cues at different phases of the gait cycle for subject-initiated modifications during the subsequent step (Patla 1991; Patla et al. 1991). Subjects were able to avoid small obstacles and alter step length or step width, but only when presented with visual cues up to the end of stance in the previous cycle. In contrast, change of direction must be cued prior to the end of midstance. Additionally, denial of vision during a “critical period” of late stance has been shown to prolong stance duration during a task requiring subjects to step on light-emitting “stepping stones” (Hollands and Marple-Horvat 1996). These studies use distinct tasks such as obstacle avoidance or alteration of foot placement to identify phases of the gait cycle in which vision is critical.

In addition to obstacle avoidance and foot placement, visual input is used for many other functions during walking. With the use of continuous optic flow stimuli, vision has been shown to be important for adjustments in speed (Konczak 1994), stride length (Prokop et al. 1997), and navigation (Warren et al. 2001). Immersive, oscillatory visual stimuli illustrate that visual inputs are critical for upright postural stability (Logan et al. 2010; Warren et al. 1996) during walking and that the nervous system makes greater use of visual information for the control of frontal-plane motion, which is thought to be more biomechanically unstable (O'Connor and Kuo 2009). The visual stimuli in the present study probed functions underlying the task of treadmill walking such as speed control for maintaining position within the boundaries of the treadmill as well as upright postural stability.

Here we further investigate the phase-dependent effect of vision on walking with discrete disturbances of the visual scene toward or away from a subject walking on a treadmill. Transient ramp and hold virtual perturbations systematically probed control of treadmill walking while the right leg was in three different phases of the gait cycle (loading, midstance, and terminal stance). We examined segment angles and muscular activity to investigate the phase dependence of responses to visual perturbations. For the midline trunk segment we considered the perturbations at three phases, and recording of kinematic and muscle responses from both right and left lower limb allowed the analysis of six distinct phases of perturbation effects.

In studies of human locomotion, phase dependence more often takes the form of a response with fixed time delay and an

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amplitude dependent on the gait cycle phase of the perturbation (perturbation phase). This is due to the “local” nature of the perturbations typically used to probe phase-dependent responses. For example, investigations using somatosensory perturbations typically probe different phases of the gait cycle and observe phase dependence as the altered amplitude of a stereotyped waveform such as the H-reflex (Capaday and Stein 1986) or short-latency stretch reflex (Mazzaro et al. 2005; Sinkjær et al. 1996; Yang et al. 1991), which are known to occur at a given response latency. Vision is clearly different. First, a “visual reflex” with fixed response latency during locomotion, if possible to elicit, has not been established. Second, visual pathways involve multiple neural structures that modify the response to a visual perturbation. Third, vision is a more “global” input, with the ability to affect all segments quasi-simultaneously.

With both the insight from studies of somatosensory input and the differences in vision noted above, we suggest a framework for understanding phase dependence of vision in human locomotion. Figure 1 presents three possible scenarios of phase dependence with these transient perturbations presented at three phases (Φ_1 , Φ_2 , Φ_3) of the gait cycle. Figure 1A illustrates a linear response that has no phase dependence, that is, the response depends only on the time delay between perturbation and response. Figure 1B illustrates gating, in which the specific phase of the gait cycle during which a perturbation is presented will dictate the gait modifications (Duysens et al. 2000). An alternative type of phase dependence, shown in Fig. 1C, is when a response occurs at a characteristic phase of the gait cycle, a pattern we will refer to as a response-phase pattern of phase dependence. A response-phase pattern results from the state (position, velocity, etc.) of the limbs dictating when a response can be generated. As

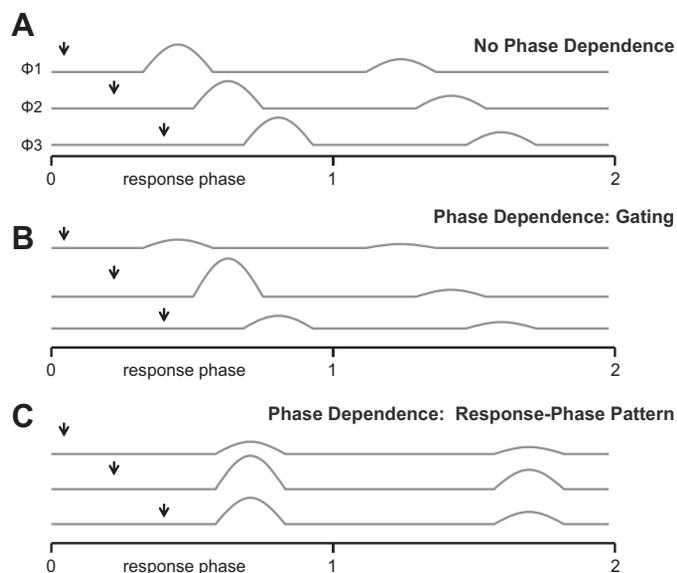


Fig. 1. Phase dependence schematic: a hypothetical data set observed across response phases of a full cycle of a continuous behavior such as walking with 3 perturbation phase presentations ($\phi_1 \dots \phi_3$) marked with arrows at perturbation onset. Responses can occur without phase dependence (A), while phase-dependent responses such as gating (B) or response-phase pattern (C) can occur. Response phase is indicated on x-axes with the perturbation cycle and a cycle following the perturbation cycle. In this hypothetical data set of potential responses the transient response diminishes as it continues into the cycle following the perturbation cycle.

response-phase pattern intrinsically dictates the latency of response from a perturbation, the resulting variable time delay and its effect on response amplitude will be observed in the response.

The use of transient scene motion in this study is crucial as it allows us to distinguish between possible phase dependence scenarios (e.g., gating vs. response-phase pattern). Transient scene motion presented at specific perturbation phases of the gait cycle allows a determination of the effective perturbation phase or phases that is not clear during continuous (throughout gait cycle) perturbations. Evidence from previous work (Logan et al. 2010) using continuous scene motion led us to hypothesize that trunk segment responses to transient visual scene changes in this study would be at most weakly phase dependent, whereas leg segment trajectories would be highly phase dependent. Here we provide support for these hypotheses with the additional finding that within-cycle modulation of the leg segments and associated muscle activity are observed only when changes in visual scene occur at midstance. This specific modulation due to visual input applied solely at midstance, in addition to a leg response observed when vision is applied at all phases, suggests that phase dependence is not strictly dependent on the specific segment but also depends on the function of that visual input.

METHODS

Ethical Approval

This study conformed to the Declaration of Helsinki and was approved by the Ethics Committee of the Santa Lucia Foundation. Informed consent was obtained from all participants according to the procedures of the Ethics Committee of the Santa Lucia Foundation.

Subjects

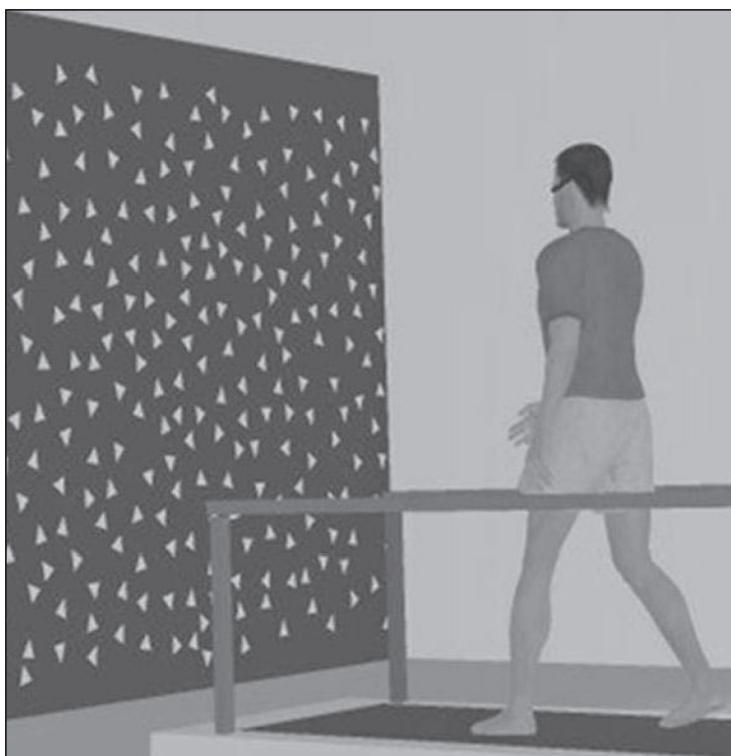
Eleven healthy subjects [6 men and 5 women, between 20 and 34 yr of age with age 23.1 ± 4.3 (mean \pm SD) yr, weight 64.1 ± 11.0 kg] received modest monetary compensation for participating in this study. All subjects were self-reported to have normal (or corrected to normal) vision and no history of neurological disorders or surgical procedures involving the feet, ankles, knees, hips, back, brain, spinal cord, or inner ear.

Apparatus

Virtual reality environment. Subjects walked at 3.6 km/h on a treadmill (EN-TRED 1475.911, Enraf-Nonius) 1 m in front of a translucent screen (4×3 m) with a rear-projected virtual display, as shown in Fig. 2A. The display consisted of 500 randomly distributed white triangles ($3.7 \times 3.7 \times 3.5$ cm) on a black background, updated at 60 Hz. The display was 3.7 m wide by 2.54 m high when static prior to trial initiation (*position 0*), and subjects wore goggles with occluded sides to prevent them from seeing the border of the visual display, allowing a 1.7-m-wide by 1.7-m-high field of view ($\sim 81^\circ$ of visual angle). The virtual display was created with CaveLib software (Mechdyne) with projection through a digital projector (MP3135, HP) synched to a desktop computer (Precision T5500, Dell). Visual signals were created off-line (MATLAB, MathWorks) and were generated via LabVIEW (National Instruments) on a desktop computer (Precision T5500, Dell).

Visual scene perturbations. During the experimental trials, the virtual scene translated either toward (negative/approaching) or away (positive/receding) from the subject in the anterior/posterior (A/P) direction. The perturbations were ramp and hold, reaching an ampli-

A



B

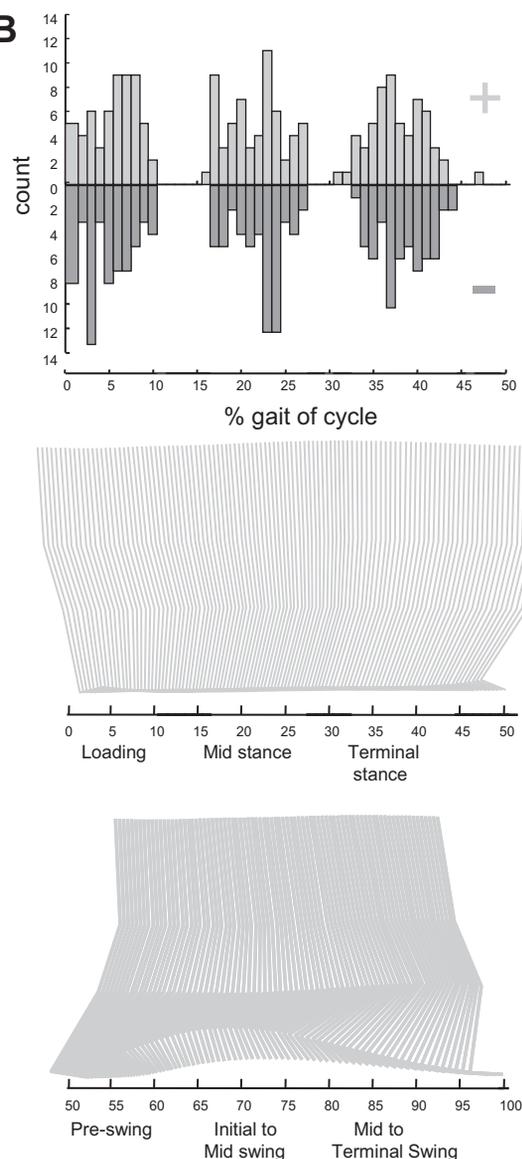


Fig. 2. Experimental setup. *A*: subjects walked on a treadmill in front of a virtual display that would translate unpredictably in the sagittal plane using ramp and hold perturbations timed to right heel strike (RHS). *B*: an exemplar histogram (*top*) of the number of positive and negative stimuli at each phase of the gait cycle that were presented to a subject and representative stick diagrams illustrating the position of the ipsilateral/right (*middle*) and contralateral/left (*bottom*) limbs at the times of perturbation. It is implicit in this exemplar stick diagram that the foot lands at zero, or the ground, in the vertical plane.

tude of 13.5 cm in 60 ms. Direction from *position 0* was chosen at random for all odd-numbered perturbations. A negative perturbation always followed a positive perturbation (and vice versa) in order to keep the range of scene motion between ± 13.5 cm. The virtual scene was constructed with a fixed perspective point at the subject's eye height, with the assumption that the subject was 1 m from the screen. The scene was created so that subjects would see a fixed visual scene with the entire scene occasionally moving coherently toward or away from them. Perception of scene motion was not quantified in an objective manner in this study. Informally, subjects typically reported that something strange occasionally happened to the visual scene without clear indication of the direction of scene motion. The scene was constructed in this way as a probe to understand how fast changes in visual scene motion are used [via kinematics, electromyogram (EMG)] at specific phases of the gait cycle. Visual display generation and data collection software were synchronized via an external trigger.

Kinematics. Body kinematics were measured with a nine-camera VICON-612 motion analysis system (VICON, Oxford, UK). Reflec-

tive markers (diameter 1.4 cm) were placed on the right and left sides of the body at external landmarks corresponding to base of the fifth metatarsal, posterior calcaneus (heel), lateral malleolus (ankle), lateral femoral condyle (knee), greater trochanter (hip), acromion process (shoulder), mastoid process (head), and frontal eminence (head). Additionally, markers were placed at the mediolateral center of the back of the head and the midline of the spine at the level of T₁, T₇, and L₁ vertebrae. All markers were attached to the skin of these bony prominences, except those placed on the shoes at the fifth metatarsal and heel. All kinematic data were collected at 100 Hz.

Our analysis focuses on the leg and trunk segments. Sagittal-plane foot, shank, and thigh segment angles relative to the vertical were computed from angles formed by the fifth metatarsal to ankle, ankle to knee, and knee to hip with the most inferior point as the origin. We use segment angles rather than anatomical joint angles because of our interest in how the body maintains its orientation relative to the vertical during walking. Moreover, because joint angles can be computed from segment angles with a linear transformation, qualitatively similar results would be obtained with joint angles. As these segment angles are relative to the

vertical, 0° in control waveform plots indicates that the superior marker on the segment is above and vertically aligned with the inferior marker on the segment. Positive values in these plots indicate that the superior marker on the segment is more forward of the inferior marker in the sagittal plane. Trunk orientation relative to the vertical in the sagittal plane was computed as the angle formed by the L₁ to T₁ markers. To characterize whole body motion on the treadmill, A/P displacements of all markers were analyzed.

EMG. Muscular activity from both legs was measured with surface EMG recordings (Zerowire, Aurion). Recordings of the following eight muscles were made from each side of the body: tibialis anterior (TA), gastrocnemius lateralis (LG), soleus (SOL), vastus lateralis (VL), rectus femoris (RF), biceps femoris (long head, BF), semitendinosus (ST), and erector spinae (ESL, recorded at L₁-L₂). Electrodes were positioned at the muscle belly with placement carefully chosen to minimize cross talk (Cappellini et al. 2006). Recording sites were shaved, lightly abraded, and cleaned with isopropyl alcohol prior to electrode application. The EMG processing consisted of high-pass filtering, rectification, and low-pass filtering with the same frequency cutoff values used in several previous studies of locomotion (Cappellini et al. 2006, 2010a, 2010b; Maclellan et al. 2012). These EMG signals were band-pass filtered (analog, built in to Zerowire EMG system) between 10 and 1,000 Hz and were sampled at 2,000 Hz. Observation of low-frequency noise (<20 Hz) in the recorded signal necessitated a high-pass digital filter prior to rectification. With MATLAB, these signals were high-pass filtered with a zero-lag forward-backward cascade of a 4th-order Butterworth filter with a 20-Hz cutoff frequency, full-wave rectified, and then low-pass filtered with a zero-lag forward-backward cascade of a 4th-order Butterworth filter with a 10-Hz cutoff frequency.

Procedures

Prior to experimental trials, subjects walked in front of a static visual display at the experimental locomotion speed in darkness with goggles on. This familiarization was followed by two trials 2 min long to determine mean gait period, which was used to create subject-specific timing of perturbations. Mean gait period was defined as the average time between each successive right heel strike (RHS) (see *Data Analysis*). In all trials, subjects were instructed to look straight ahead and were given ~30 s to reach steady state before recordings were made. For safety, an experimenter was behind the treadmill in close proximity with a push button to immediately halt the treadmill if needed (never used).

Using subject-specific mean gait period, perturbation signals were designed so that onset of the ramp perturbations was timed to occur at about 0%, 16.66%, or 33.33% of the gait cycle (see below). RHS was defined as 0% of the gait cycle. Subjects experienced 14 experimental trials lasting ~4 min each, with ~24 perturbations in each trial. These 24 perturbations were 4 repeats of the 6 possible ramp and hold perturbations (2 directions × 3 perturbation phases) and were randomized within each trial, subject to the direction constraints mentioned above (see *Visual scene perturbations*). These perturbations were initiated via a foot switch whose force sensor was placed 1.25 cm anterior to the heel on the sole of the right foot. The foot switch used was a pressure-sensitive resistor (Zerowire, Aurion) that would indicate RHS and was integrated into the visual display system. As the motion capture system was not integrated into the visual display system, the foot switch allowed initiation of the subject-specific perturbations designed to occur at specific phases of the gait cycle. Perturbations were applied pseudorandomly throughout the trial. Across subjects, the mean number of gait cycles between perturbations was 8 cycles and the mean minimum and maximum gait cycles between perturbations were 4.3 and 27 cycles. There were never <3.6 cycles between perturbations.

Data Analysis

Perturbations. Prior to data analysis, the phase of the gait cycle where perturbation onset occurred (initiation of visual scene motion) was identified for each ramp and hold perturbation. The phase of the cycle when the perturbation occurs is the perturbation phase, while the response phase is the phase at which a response occurs after perturbation onset. Gait cycle phase of perturbation onset was identified as the percentage of the mean control cycle (see *Statistics*) elapsed between RHS prior to perturbation and perturbation onset. Each heel strike was computed as the local minimum of the heel marker in the vertical plane (Borghese et al. 1996; Ivanenko et al. 2004) occurring after each cycle's maximum angle formed by the fifth metatarsal-hip axis in the sagittal plane with the hip as the angle's origin.

Because of variability of RHS predicted by the foot switch relative to that measured from kinematics, ranges of perturbation onset were used. Those perturbations that occurred at 0–10%, 16–26%, and 33–43% of the gait cycle were considered to occur at the three phases of perturbation onset. An exemplar histogram based on percentage of gait cycle from a single subject can be seen in Fig. 2B, and the mean (SD) percentage of the gait cycle across subjects in these three phases was 6.1% (0.6), 22.5% (0.6), and 39.2% (0.6) for the 0–10%, 16–26%, and 33–43% groupings, respectively. On average, 52, 49, and 52 perturbations occurred in these 0–10%, 16–26%, and 33–43% groupings for each perturbation direction for each subject.

Note that we applied perturbations in the first half of the gait cycle. To infer the effects of perturbations in the second half of the gait cycle, we assumed that walking has left-right spatio-temporal symmetry, namely, that reversing left and right sides of the body is equivalent to shifting time by half a gait cycle. For example, we assumed that 1) the right-side response to a perturbation at phases 50–60% with respect to RHS equals the left-side response to a perturbation at phases 50–60% with respect to left heel strike (LHS) and 2) phases 50–60% with respect to LHS equals phases 0–10% with respect to RHS, which corresponds to one of our experimental perturbations.

In this way, we were able to infer the responses of both sides of the body to perturbations at six phases of the gait cycle. In particular, left-side responses to perturbations at phases 0–10%, 16–26%, and 33–43% with respect to RHS were used to infer right-side responses to perturbations at phases 50–60%, 66–76%, and 83–93% with respect to RHS. Figure 2B, *middle*, shows an example control gait cycle trajectory of the right side of the body to illustrate the normative configuration of the ipsilateral side of the body during the perturbation phases. In Fig. 2B, *bottom*, the configuration of the contralateral (left) side of the body at the same time is provided to illustrate the position of the contralateral limbs when these perturbations were to occur.

For clarity in presenting our results, we will refer to perturbation phases with respect to heel strike as loading (0–10%), midstance (16–26%), terminal stance (33–43%), preswing (50–60%), initial to midswing (66–76%), and midswing to terminal swing (83–93%) (Perry 1992). Percentages refer to the range of perturbation onsets.

Statistics. As seen in Fig. 3A, large variability in a kinematic or EMG signal may mask the effect of the visual scene perturbation and how it depends on perturbation phase. Therefore, to quantify perturbation effects, we computed residual waveforms as follows. First, we defined a perturbation cycle as a gait cycle (heel strike to heel strike) during which a perturbation occurred and a control cycle as a gait cycle just prior to a perturbation cycle. We then used linear interpolation to compute response signals as a function of phase, where phase in increments of 0.005 ranged from –1 to 0 for control cycles, from 0 to 1 for perturbation cycles, from 1 to 2 for the first postperturbation cycle, etc. For each trial, we averaged over all control cycles to obtain an unperturbed mean control waveform. For displacements, we computed a linear trend based on the first and last value of the mean control waveform and subtracted this trend from the mean control waveform. EMG signals were normalized by the maximum value of

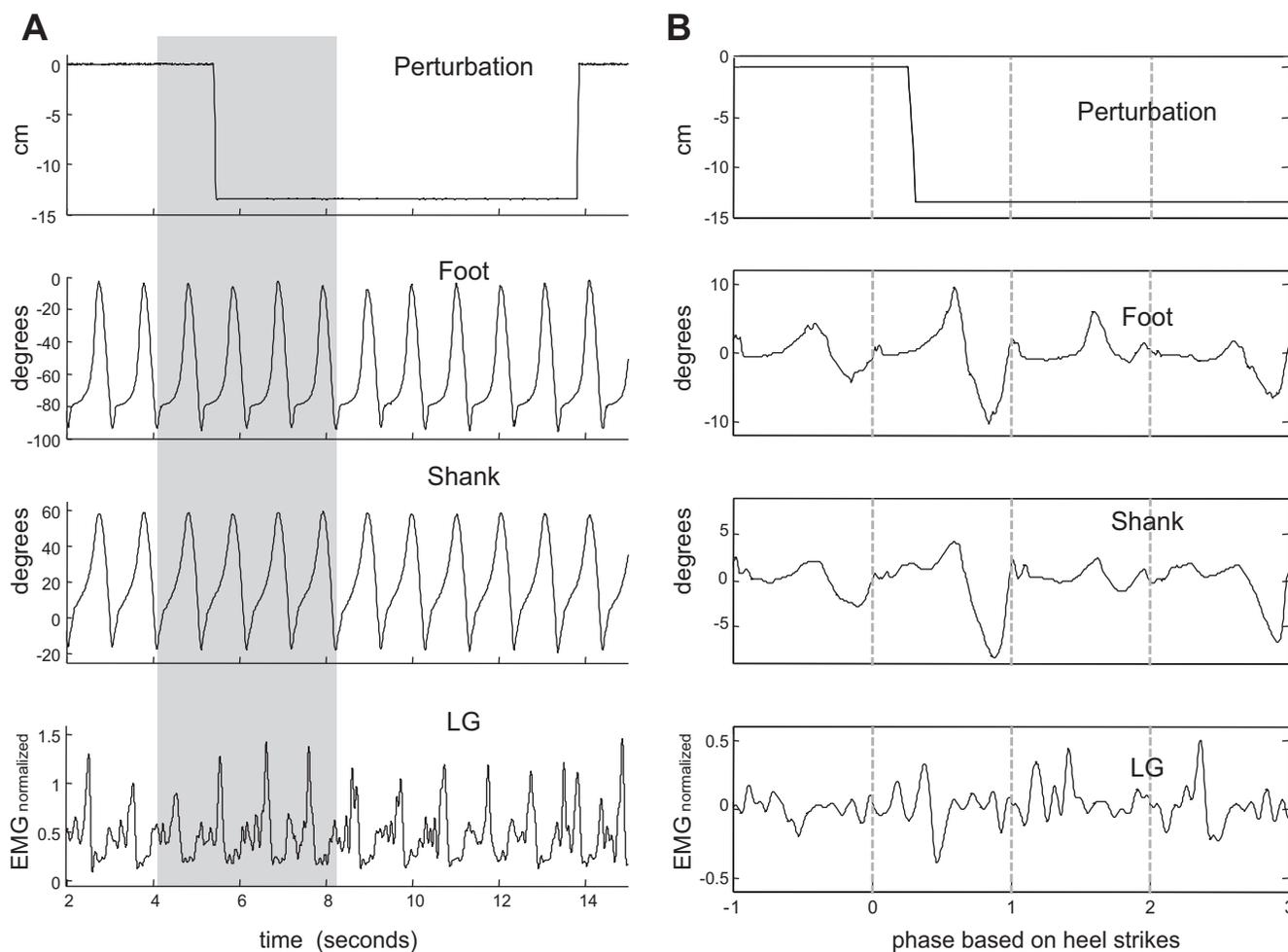


Fig. 3. Epoch extraction and normalization. *A*: example time series of perturbation, foot segment angle, shank segment angle, and rectified lateral gastrocnemius (LG) EMG recordings. *B*: the shaded region in *A* after normalization to the gait cycle and subtraction of mean control cycle waveforms (see METHODS). Vertical dashed lines in *B* denote heel strikes. EMG has been normalized to control cycle maximum (see METHODS). An *x*-axis value of 0 in *B* is the heel strike prior to visual scene motion. The perturbation isolated for observed responses in *B* occurs at midstance and is toward the subject in the sagittal plane (-).

the mean control waveform (Nieuwenhuijzen et al. 2000). For each gait cycle, the residual waveform as a function of phase was computed by subtracting the mean control waveform from the given signal (Fig. 3*B*). To correct for a slow drift in the subject's A/P position on the treadmill over the course of a trial, we computed the least-squares linear fit of the residual control cycle waveform of the L₁ marker, extrapolated this linear fit over the perturbation and postperturbation cycles, and then subtracted the linear fit from all A/P displacement signals. A residual waveform significantly different from 0 indicates that the visual perturbation had a significant transient effect (see EFFECT OF PERTURBATIONS). The effect of visual perturbations on cycle periods, which is related to phase resetting, was analyzed separately (see CHANGE IN GAIT MEASURES).

PHASE DEPENDENCE. To quantify phase dependence, we considered residual waveforms $r_k(\phi)$ as a function of delay ϕ from the perturbation in units of cycles, where the index $k = 1, \dots, 6$ indicates the perturbation phase. We computed the mean $\bar{r}(\phi)$ of $r_k(\phi)$ across the six perturbation phases and defined $d_k(\phi) = r_k(\phi) - \bar{r}(\phi)$. Nonzero $d_k(\phi)$ correspond to phase-dependent responses. We defined R^+ and R^- as the root mean square (RMS) of $r_k(\phi)$ across positive delays $\phi \in (0, 2]$ and negative delays $\phi \in (-2, 0]$, respectively, and across the six perturbation phases. Since true responses to a perturbation occur at positive delays, a value of $R = R^+ - R^-$ significantly greater than 0 indicates a significant response. Similarly, we defined D^+ and D^- as the RMS of $d_k(\phi)$ for positive and negative delays,

respectively. Then a value of $D = D^+ - D^-$ significantly greater than 0 indicates a significant phase-dependent response. To quantify the degree of phase dependence, we used the normalized measure $P = D/R$. $P = 0$ indicates no phase dependence, and P approaches 1 as the degree of phase dependence increases. Thus P evaluates phase dependence along a continuum. A low but significant P characterizes a response as “weakly phase dependent,” which is synonymous with being primarily dependent on a time delay from the perturbation. We computed 95% confidence intervals for values of P using the bootstrap percentile-*t* method with 4,000 bootstrap resamples and 400 nested bootstrap resamples for variance estimation (Hall 1988; Zoubir and Boashash 1998).

EFFECT OF PERTURBATIONS. Significant deviations of the normalized residuals from 0 were considered the effect of the perturbation. Characterizing a dependence on perturbation phase relies on observing the presence or absence of an effect for each perturbation phase. For each of the 12 perturbation types (2 directions \times 6 perturbation phases), epochs of one cycle prior to the perturbation cycle and two cycles after the perturbation cycle were extracted from the normalized residual waveforms. For displacements, three cycles after the perturbation cycle and one cycle prior were extracted from normalized residual waveforms. These epochs were averaged within condition for each subject and binned in 5% intervals for two cycles after the end of each perturbation phase (e.g., from 10% for perturbations during loading). Because muscular activity shows more transient, shorter-

lasting differences from 0, bin sizes of 1% for a single cycle after perturbation onset were used for EMG waveforms. Effects of the perturbation were considered significant if they were different from 0 by a *t*-test at each bin in each of the 12 perturbation conditions (2 directions \times 6 perturbation phases). To test for asymmetry between the effects caused by positive and negative perturbations, postperturbation data were summed across direction at each bin and then tested for significance from 0 with a *t*-test at each bin. To account for multiple tests within each perturbation condition, we controlled the false discovery rate (FDR) at a level of 0.05, using the method of Benjamini and Hochberg (1995) applied to the *P* values from the *t*-tests. This method is valid for independent *P* values or *P* values with positive dependency (Benjamini and Yekutieli 2001).

CHANGE IN GAIT MEASURES. Changes in the gait measures of stride length and gait period in each condition (2 directions \times 6 phases) from preperturbation control cycle values were computed. For changes in gait period, mean gait period (from heel strike to heel strike, in seconds) from preperturbation control cycles was subtracted from gait periods of the perturbation cycle (and successive cycles) for each subject prior to averaging across subjects. To observe changes in stride length over the same time period, deviations in A/P displacements of the heel marker from preperturbation control cycles were computed in the same manner as above for segment angles. To compute change from mean displacement of the same foot (change in stride length), the deviation of heel displacement at each heel strike of the previous cycle was subtracted from deviation of heel displacement at each heel strike. RHS timing and deviations of A/P displacement of the right heel marker were used for the perturbation phases of loading, midstance and terminal stance. Spatio-temporal symmetry was assumed, and these calculations for the perturbation phases of preswing, initial to midswing, and midswing to terminal swing were computed with LHS timing and deviations of the left heel marker. Changes in these gait measures were computed for the cycle in which the perturbation occurred and for two successive cycles. Correction for multiple tests within each condition was performed by controlling for FDR (as above).

RESULTS

Four main results emerged in response to visual scene perturbations. First, translations of the visual scene that were toward the subject (approaching/negative) led to more consistent and larger deviations from mean waveforms than the opposite translations (receding/positive) and are the focus of our results below. Second, significant deviations occurred in the trunk, but these deviations were not tied to a specific response phase or perturbation phase. Third, an additional, within-cycle gated response was observed in all leg segments when the perturbation was presented at midstance (perturbation phase) and was accompanied by significant deviations in distal leg muscles. Finally, responses in the leg segments were found to be highly phase dependent. More specifically, a response-phase pattern in the legs was observed as significant, stereotyped deviations of the foot and shank that consistently occurred prior to and after the stance to swing transition (response phase) of the gait cycle following the perturbation cycle.

Trunk Shows Little Phase Dependence, Legs Show Strong Phase Dependence

We quantified the phase dependence of kinematic responses to approaching (negative) visual perturbations on a scale from 0 to 1 (see METHODS). The trunk (Fig. 4) had only a low phase dependence of 0.24 with a 95% confidence interval of [0.10,

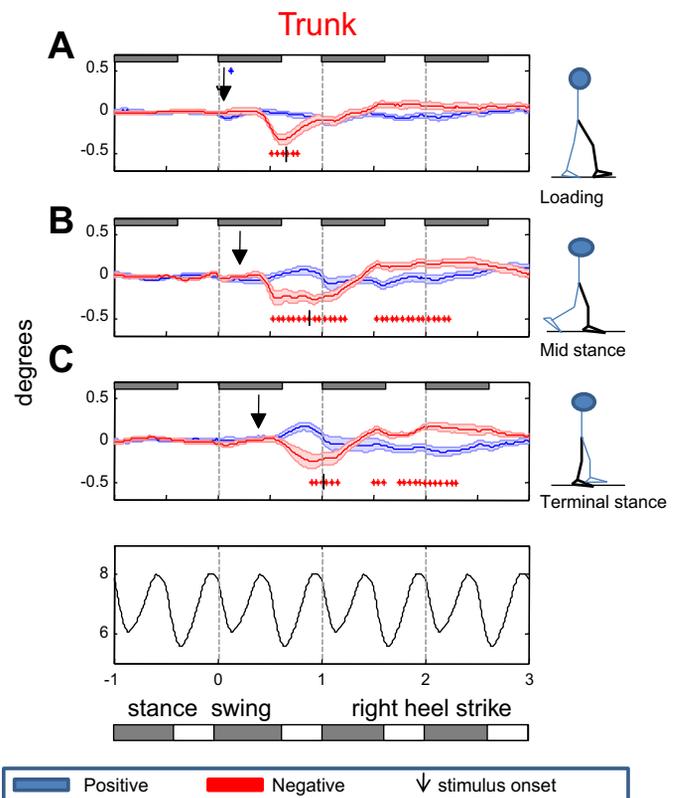


Fig. 4. Trunk orientation. Residual waveforms averaged across subjects are presented here for all directions and phases of visual scene motion. Arrows have been appropriately placed at the average stimulus phase for each condition, and stick figures represent position of the body during the 3 stimulus phases. The phase of perturbation is illustrated on right of each plot: loading (A), midstance (B), and terminal stance (C). Blue/red asterisks denote that the positive/negative condition was different from zero at that bin, and these asterisks have been placed at the midpoint of the bin, as these continuous traces were binned when performing statistics [$n = 11$, false discovery rate (FDR) < 0.05]. The vertical line within the asterisks for the negative perturbation denotes response latency for these trunk responses. An *x*-axis value of 0 is the heel strike prior to visual scene motion. Heel strike times of response phase are noted appropriately above or below each respective limb's response, and inset bars represent stance phase. Shaded error bars correspond to SE. Across-subject means of the control cycle waveforms are concatenated and presented below residual waveforms. A more positive segment angle indicates a more forward deviation from vertical in the sagittal plane that could result in a more flexed trunk.

0.31], indicating that perturbations at all phases of the gait cycle produced similar responses primarily as a function of the delay from the perturbation. Leg segment responses (Fig. 5), however, were highly phase dependent: 0.92 for the foot with a 95% confidence interval of [0.86, 0.96], 0.89 [0.84, 0.93] for the shank, and 0.91 [0.84, 0.97] for the thigh.

Approaching Perturbations Yield Three Distinct Responses

Segment angles. Figure 4, A–C, show responses of trunk orientation to perturbations applied at loading, midstance, and terminal stance, respectively. These trunk segment angles are aligned to the heel strike prior to the perturbation onset after removing mean waveforms from preperturbation gait cycles. These across-subject averages of the residual waveforms in Figs. 4–6 are the effect of the perturbations, and significant differences from zero are denoted with asterisks (FDR < 0.05 , see *Statistics*). To view normative segment angles at all response phases, these figures also contain concatenated control

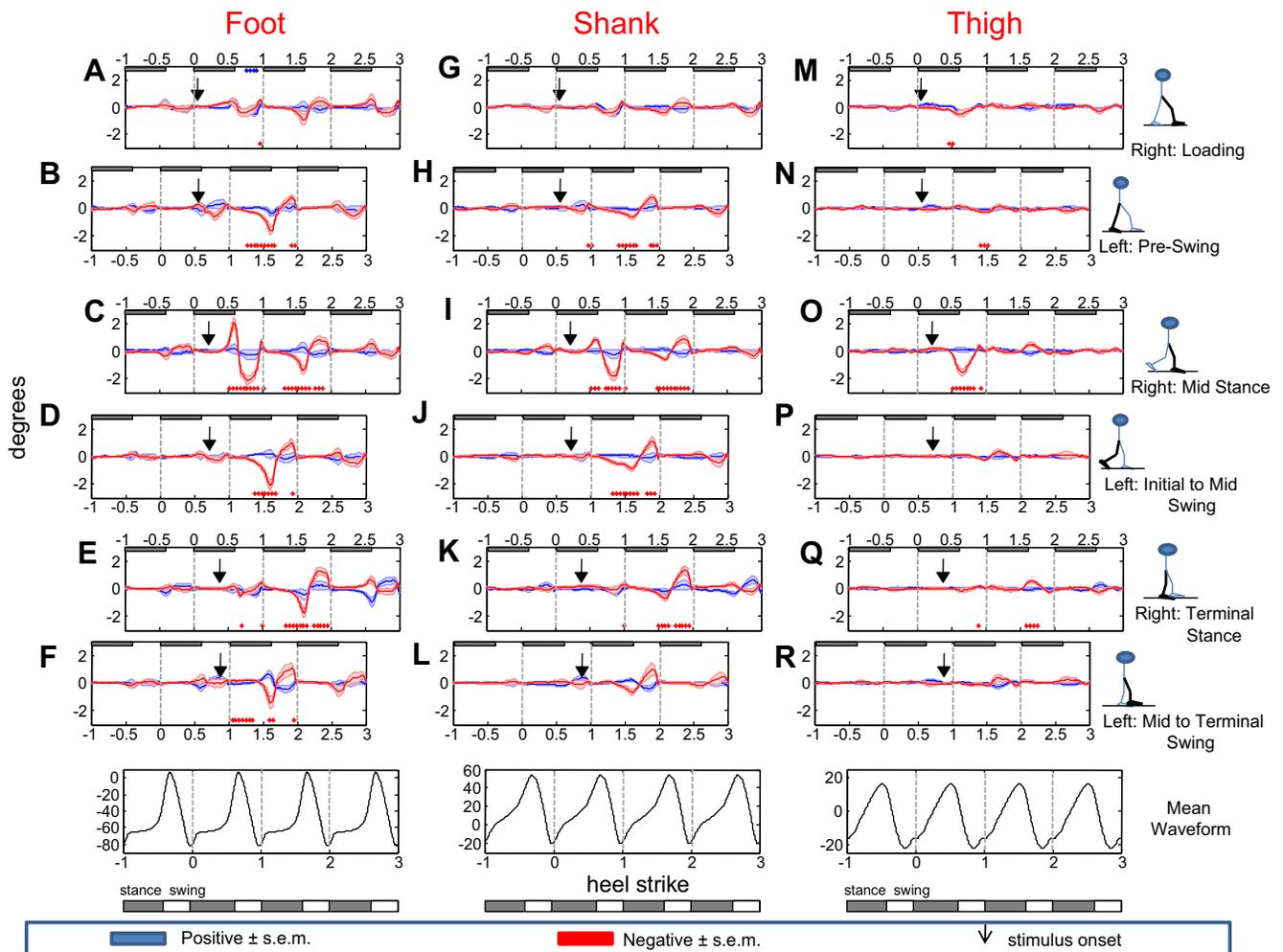


Fig. 5. Leg segment angles. Residual waveforms averaged across subjects are presented here for all directions and phases of visual scene motion. A–F: foot. G–L: shank. M–R: thigh. Arrows have been appropriately placed at the average stimulus phase for each condition, and stick figures represent position of the body during these 6 stimuli phases. Underneath each stick figure are the limb used for those data and the gait cycle phase it was in when the perturbation occurred. Blue/red asterisks denote that the positive/negative condition was different from zero at that bin, and these asterisks have been placed at the midpoint of the bin, as these continuous traces were binned when performing statistics ($n = 11$, $FDR < 0.05$). An x -axis value of 0 is the heel strike prior to visual scene motion. Heel strike times of response phase are noted appropriately above or below each respective limb’s response, and inset bars represent stance phase. Shaded error bars correspond to SE. Across-subject means of ipsilateral right control cycle waveforms are concatenated and presented below residual waveforms. A more positive segment angle indicates a more forward deviation from vertical in the sagittal plane that could result in a more extended thigh, flexed shank, or extended foot (plantar-flexion).

cycle waveforms (mean of preperturbation cycles) below the residual waveforms.

Negative perturbations applied at all perturbation phases led to a decrease (i.e., backward tilt) in trunk orientation, but at no particular response phase. Looking at the control waveform, these decreases in trunk orientation (backward tilt) occur when the trunk is in various states of the more positive flexion and less positive extension. Perturbations applied at midstance (Fig. 4B), for example, caused significant decrements in the response phase of preswing of the perturbation cycle until midstance of the postperturbation cycle and were largest at the stance to swing transition of the perturbation cycle. In comparison, decreases in trunk angle due to perturbations applied at terminal stance displayed in Fig. 4C began and reached their largest point in the response phase of terminal swing of the perturbation cycle and continued through midstance of the following cycle. To supplement our quantification of phase dependence reported above, the latency of the trunk response was

quantified as the midpoint of significant bins in the initial backward trunk tilt response. As the trunk response is not a stereotyped waveform but does share a significant, initial backward tilt of varying width across perturbation phases, the measure best details a single point when the response specific to each perturbation phase has occurred. The midpoints of the initial, significant backward tilt in trunk orientation occur at 0.55, 0.62, and 0.6 cycle lengths from perturbation onset in loading, midstance, and terminal stance, respectively, and can be observed in Fig. 4 among the significance asterisks. After correction for gait period in each condition, these cycle length latencies correspond to response latencies of 633 ms, 713 ms, and 693 ms for loading, midstance, and terminal stance, respectively. These are the values of response latencies we refer to in the remaining text. For reference, significant decrements first occurred in preswing ($\approx 53\%$, Fig. 4A, midpoint of bin), preswing ($\approx 54\%$, Fig. 4B), and terminal swing ($\approx 91\%$, Fig. 4C) for loading, midstance, and terminal stance perturbations,

respectively. Although we do not consider these values as response latencies, the first instances of significance corrected for gait period and stimulus onset correspond to 489 ms, 309 ms, and 543 ms from perturbation onset for loading, midstance, and terminal stance, respectively. Overall, the subplots in Fig. 4 are consistent with the phase dependence quantification presented above; for the trunk, phase does not dictate when a response happens and responses occur with a similar delay from stimulus onset.

Assuming left-right spatio-temporal symmetry, bilateral kinematic and EMG responses to perturbations at three phases of the gait cycle were used to infer responses at a total of six perturbation phases (see METHODS).

In contrast to responses observed in the trunk segment, responses in the leg segments occur at characteristic response phases and appear as two responses. First, significant within-cycle deviations of all leg segments were observed when negative perturbations were applied solely at midstance. At this perturbation phase specifically, the largest-magnitude deviations across all perturbation phases in the foot, shank, and thigh segment angles were observed in the gait cycle in which the perturbation occurred. As seen in the third row of plots in Fig. 5, significant increases in the foot and shank segment angle peaked at the stance to swing transition (≈ 0.6 , Fig. 5, *C* and *I*) while significant decreases followed and were at their greatest magnitude in the response phase of midswing for both the foot (≈ 0.79 , Fig. 5*C*) and the shank (≈ 0.84 , Fig. 5*I*). Looking at control waveforms of foot and shank, it is clear that the negative perturbation presented at midstance caused a heightened increase in the foot and shank angle until the stance to swing transition followed by larger decrease in these segment angles during midswing. These deviations result in a net increased plantar-flexion of the foot prior to toe off and an increased dorsi-flexion after toe off, while flexion of the shank increased prior to the stance to swing transition, followed by net decreased flexion in swing until heel strike. A similar negative deviation was observed in the thigh angle in Fig. 5*O* during the response phase of initial swing (≈ 0.69) corresponding to a larger decrease in the thigh angle at this phase (when compared with the control waveform), resulting in a more pronounced thigh flexion throughout the swing phase of the perturbation cycle.

Making up the second response of the leg segments, plots of the foot and shank angles illustrate that approaching (negative) visual perturbations across a range of perturbation phases caused deviations in the cycle following the deviation cycle. As seen in Fig. 5, *B–F*, in the gait cycle (heel strike to heel strike) following the perturbation cycle (postperturbation cycle, *heel strike 1–2*, Fig. 5) significant decreases and ensuing increments were observed in the foot angle when perturbations were applied at midstance, terminal stance, preswing, initial to midswing, and midswing to terminal swing. These decrements in the postperturbation cycle are largest in magnitude at the response phase of the stance to swing transition (≈ 1.6 on the *x*-axis for all perturbation phases in Fig. 5, *B–F*). The significant increments that follow are all largest in magnitude in the response phase of midswing to terminal swing (range: 1.83–1.92 in Fig. 5, *B–F*). Although not significant, peak decrements occurred at the same response phase when perturbations were applied at loading (Fig. 5*A*). Comparing the residual waveforms in Fig. 5, *B–F*, to the control waveform of foot angle

(postperturbation cycle, *heel strike 1–2*, Fig. 5), the effects of the perturbation correspond to a decrease in the increasing foot angle prior to and including the stance to swing transition and an increase in the decreasing foot angle during swing. These deviations translate to a net decreased plantar-flexion of the foot prior to toe off and a net decreased dorsi-flexion after toe off.

In the shank, significant decrements were observed when negative perturbations were applied at midstance, terminal stance, preswing, and initial to midswing. Shown in Figs. 5, *H–K*, these decrements occurred in the postperturbation cycle (*heel strike 1–2*, Fig. 5, *H–K*) and were largest in magnitude at the stance to swing transition (≈ 1.6 in Fig. 5, *H–K*). These decrements were followed by a more pronounced increment whose peak occurred during the response phase of terminal swing (≈ 1.9 in Fig. 5, *H–K*, respectively) when perturbations were applied at midstance, terminal stance, preswing, and initial to midswing. Comparing these deviations to the control waveform of shank angle, the effects of the perturbation correspond to a decrease in the increasing shank angle prior to and including the stance to swing transition and an increase in the decreasing shank angle during swing. These deviations result in a net decreased flexion of the shank at the knee prior to the stance-swing transition followed by an increased flexion at the knee as the knee is extending in midswing. In sum, negative perturbations, when applied across different phases of the gait cycle, characteristically alter the trajectories of the lower leg segments of foot and shank at specific response phases of the postperturbation cycle (*heel strike 1–2*, Fig. 5, *H–K*).

Muscle activation. As observed in Fig. 6, the largest deviations in EMG waveforms were observed after negative, midstance perturbations. Nonsignificant increases in plantar-flexor muscles precede foot plantar-flexion, while significant decrements precede foot dorsi-flexion, illustrating a coordination between kinematic and EMG responses to visual scene motion. As decrements in LG and SOL were significant, we focus on their functional role in causing a dorsi-flexion from mean waveform during midstance perturbations. Significant decreases in both LG and SOL can be observed in Fig. 6, *C* and *I*. In both cases, these decreases were largest in magnitude during the response phase of terminal stance of the perturbation cycle, with LG and SOL reaching sharp declines of -0.094 (fraction of maximum control activity) and -0.074 at 48% and 49% of the gait cycle, respectively. As seen in the control waveforms, these decrements in LG and SOL occur as the activity of these muscles is decreasing in amplitude from peak activity, suggesting an increased decline in activation of these muscles prior to push-off. Of these two muscles, SOL reliably decreased first, with the midpoint of the significant decrements occurring at 47% of the perturbation cycle (Fig. 6*I*). When corrected for gait cycle timing and perturbation onset, this corresponds to a response latency of 300 ms to the midpoint of significant bins of this decrement. For comparison with a previous study (Marigold et al. 2007), the response latency of SOL decrements to the negative, midstance perturbations of this study was 338 ms if each latency was computed on a single-trial basis and deviations had to be greater than two standard deviations for at least 30 ms (method of Marigold et al. 2007). The negative perturbations applied at midstance also elicited a significant increase in TA activity at the stance to

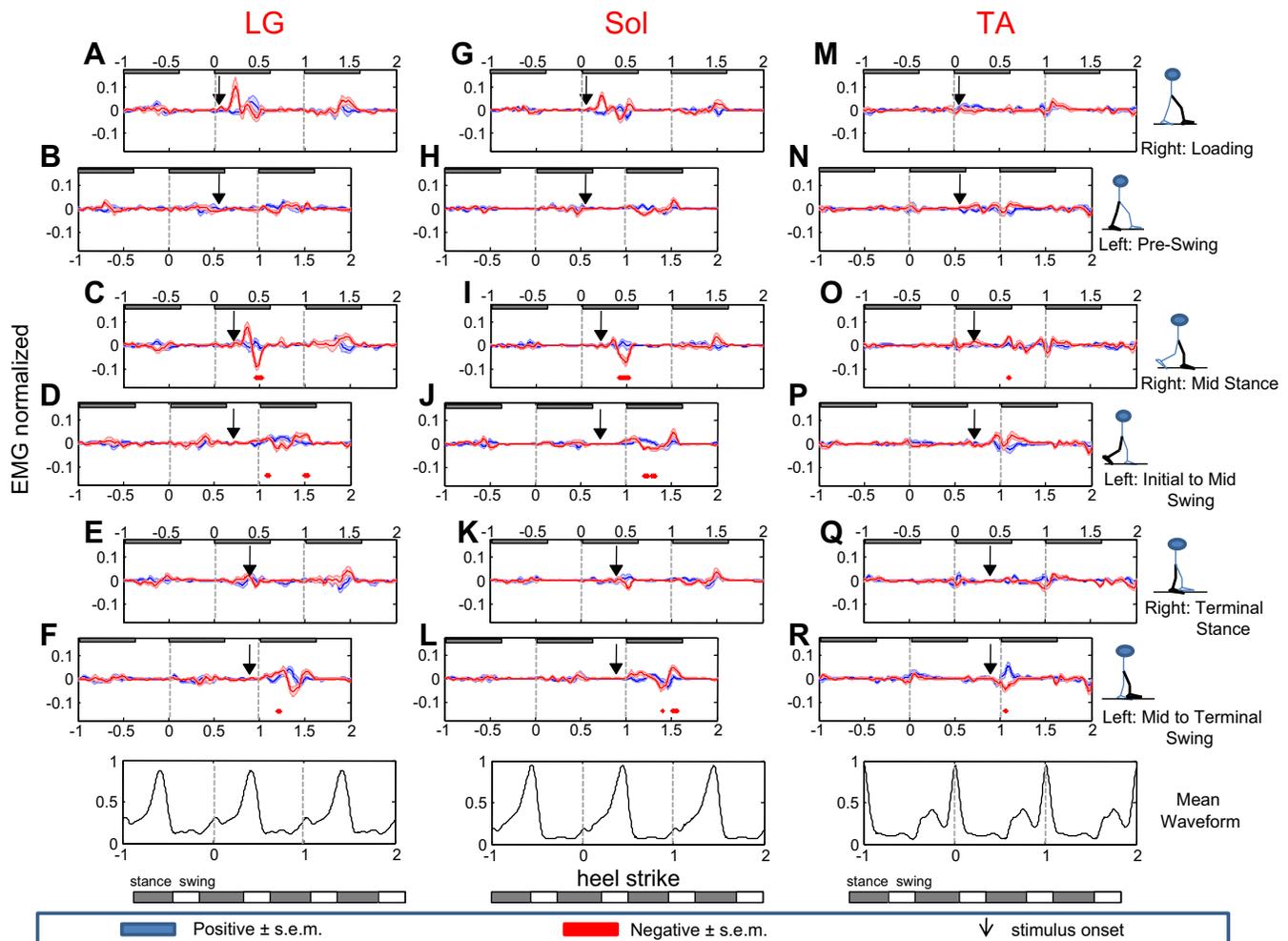


Fig. 6. EMG waveforms. Residual waveforms averaged across subjects are presented here for all directions and phases of visual scene motion. A–F: lateral gastrocnemius (LG). G–L: soleus (Sol). M–R: tibialis anterior (TA). Arrows have been appropriately placed at the average stimulus phase for that condition, and stick figures represent position of the body during these 6 stimuli phases. Underneath each stick figure are the limb used for those data and the gait cycle phase it was in when the perturbation occurred. Blue/red asterisks denote that the positive/negative condition was different from zero at that bin, and these asterisks have been placed at the midpoint of the bin, as these continuous traces were binned when performing statistics ($n = 11$, FDR < 0.05). An x -axis value of 0 is the heel strike prior to visual scene motion. Heel strike times of response phase are noted appropriately above or below each respective limb's response, and inset bars represent stance phase. Shaded error bars correspond to SE. EMG values are in normalized units (to control cycle maximum for each trial). Across-subject means of the ipsilateral right control cycle waveforms are concatenated and presented below residual waveforms.

swing transition of the perturbation cycle as shown in Fig. 6O. In sum, perturbations applied only at midstance modulated the amplitude of all of these distal leg muscles during late stance response phases of the perturbation cycle.

In addition to those observed in Fig. 6, there were sporadic, significant deviations observed in ESL, BF, and ST due to these negative perturbations. In ESL, a decrease (-0.078 at its lowest) was observed in preswing (54–58%) of the perturbation cycle when negative perturbations were applied at midstance. There were three instances where significant deviations were observed in BF: an increase (0.036 at its highest) was observed in midswing (84–87%) of the perturbation cycle when perturbations were applied at preswing; an increase (0.027 at its highest) was observed in loading (2–6%) of the postperturbation cycle when negative perturbations were applied midswing to terminal swing; and a decrease (-0.041 at its lowest) was observed in terminal swing (87–91%) of the postperturbation cycle when these negative perturbations were applied at midswing to terminal swing. In ST, a decrement

(-0.036 at its lowest) was observed in early stance (9–14%) of the postperturbation cycle when negative perturbations were applied at initial to midswing. Although significant, these changes were generally much smaller than those observed in the distal leg muscles when perturbations were applied at midstance, diminishing their functional significance.

Receding Perturbations Had Little Effect on Segment Angles and EMG

As shown by Figs. 4–6, receding (positive) perturbations generally yielded smaller and inconsistent changes in segment angle trajectories and EMG waveforms compared with approaching (negative) perturbations. Figure 5A shows small, but significant, decrements in the foot angle observed during the swing phase of the perturbation cycle when positive perturbations were presented at loading (FDR < 0.05). In the trunk (Fig. 4A), however, a single significant decrement was observed when positive perturbations were applied at loading. In all muscles, only a single instance of a significant deviation

occurred during positive perturbations. A small but significant increase (0.042 at largest) in ST activity was observed in midswing (85–87%) in the perturbation cycle, followed by a significant decrease (−0.038 at lowest) in late swing (91–94%) when positive perturbations were applied at loading.

When the effects due to the two directions were directly tested for asymmetry, significant differences were typically observed when negative perturbations caused significant deviations (FDR < 0.05). Overall, 88% (129/147) of those significant asymmetries in the leg and trunk segment angles were associated with significant responses to negative perturbations. Similarly, 71% (10/14) of asymmetries observed in the muscles were associated with significant responses to negative perturbations.

Changes in Stride Length and Gait Period

Deviations from mean stride length and gait period are presented in Fig. 7, illustrating that positive (receding) perturbations had little effect on the gait of subjects on the treadmill.

Negative perturbations in the majority of perturbation phases caused a decreased stride length in both the cycle in which the perturbation occurred (perturbation cycle) and the cycle afterward (+1 cycle/postperturbation cycle). As seen in Fig. 7A, a decreased stride length was observed in the perturbation cycle in the first four perturbation phases, with increases observed when perturbations were applied at initial to midswing and midswing to terminal swing. In the postperturbation cycle, however, solely decreases in stride length were observed and four of these decreases were statistically significant. Figure 7B shows that the cycle after the perturbation (+1 cycle) was lengthened in time for the majority of phases in which a negative perturbation was used. Interestingly, both increases in gait period and decreases in stride length were observed in the +1 cycle during the three perturbation phases of midstance, terminal stance, and preswing. This combination of changes in stride length and gait period corresponds to a stride shorter in distance and longer in time, which effectively slows the subject on the treadmill in the postperturbation cycle.

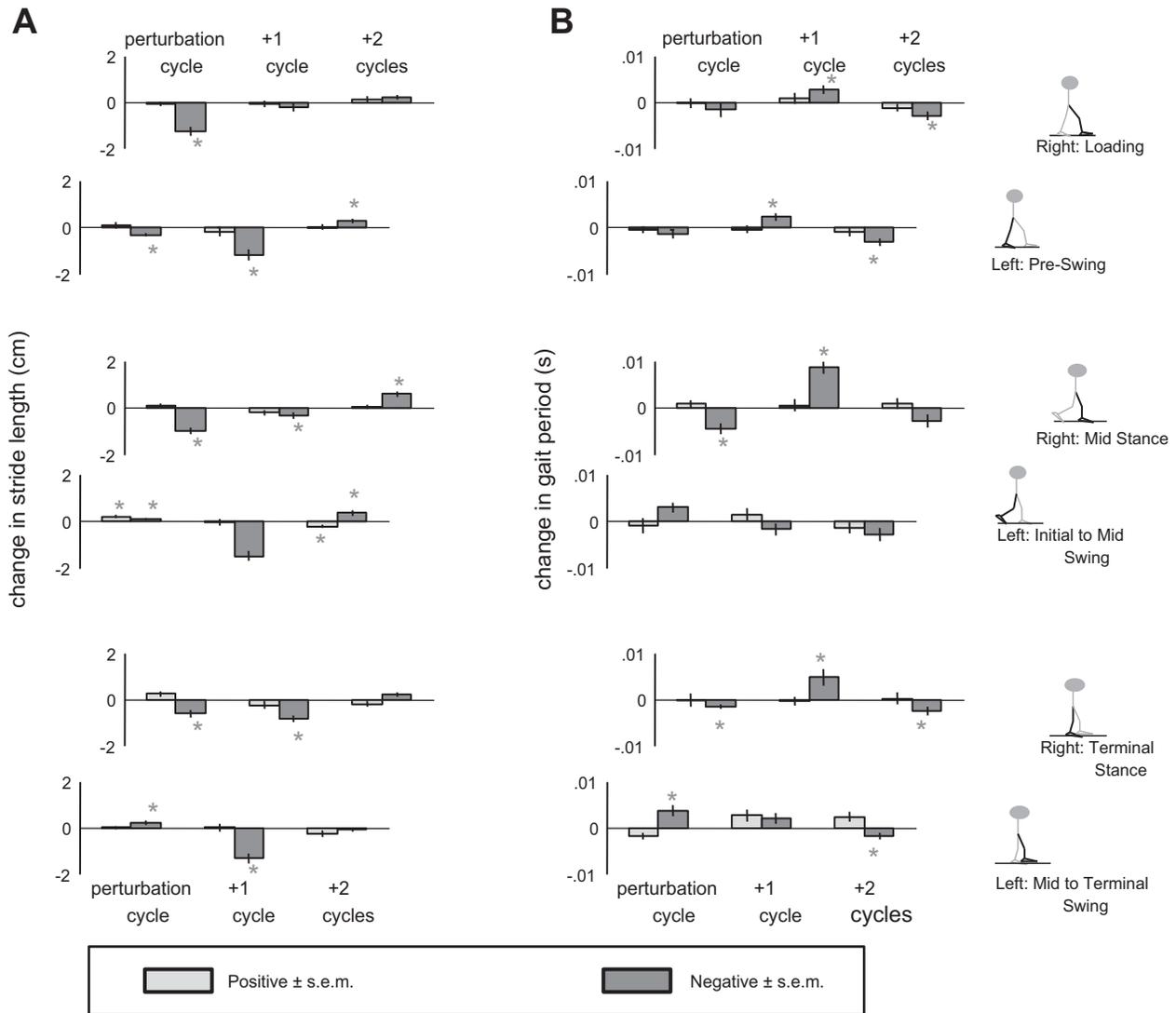


Fig. 7. Changes in gait measures. Changes in stride length (A) and gait period (B) from mean pre-perturbation cycle are presented here for all phases of perturbation and direction. Underneath each stick figure are the limb used for those data and the gait cycle phase it was in when the perturbation occurred. Perturbation cycle corresponds to the gait/stride cycle in which the perturbation occurred, and +1 cycle corresponds to the following cycle. Asterisks denote a significant change from mean pre-perturbation cycle ($n = 11$, FDR < 0.05). Error bars correspond to SE.

Displacement of the Body on the Treadmill

By assuming that responses in each leg would be the same if perturbations were presented at an identical phase of the gait cycle (spatio-temporal symmetry), it has been shown above that the negative visual perturbations applied at many phases of the gait cycle result in slowing on the treadmill. To investigate this slowing further, displacements of the body with both legs and trunk on the same normalized timescale are plotted in Fig. 8.

Figure 8 shows mean residual waveforms of displacements of both legs and trunk (T1). They were displaced backward significantly by the negative perturbation at all three phases of perturbation. Similar to the weak phase dependence previously observed in the trunk orientation (Fig. 3), displacement of T1 backward begins later as perturbations occur later. Significant backward deviations begin at $\approx 48\%$ (midpoint of bin), $\approx 64\%$, and $\approx 86\%$ of the perturbation cycle with perturbations at

loading (Fig. 8A), midstance (Fig. 8B), and terminal stance (Fig. 8C) of the right leg, respectively.

In the legs, backward displacements began in the left leg prior to the right leg when perturbations occurred at loading of the right leg (Fig. 8A) while backward displacements occurred first in the right leg when perturbations occurred during both midstance (Fig. 8B) and terminal stance (Fig. 8C) of the right leg. Interestingly, the largest backward deviations of all markers (ankle, toe) of both feet occurred when each leg was in its swing phase of the cycle following the perturbation cycle (postperturbation cycle). Consistent with responses of segment angles, deviations observed in the right leg when the perturbation occurred during midstance of the right leg were unique relative to the other phases of perturbation. As seen in Fig. 8B, significant forward displacements in all markers of the right leg occurred at the stance to swing transition and, for most markers of the leg, continued throughout the entirety of the swing phase. In sum, all segments were eventually displaced back-

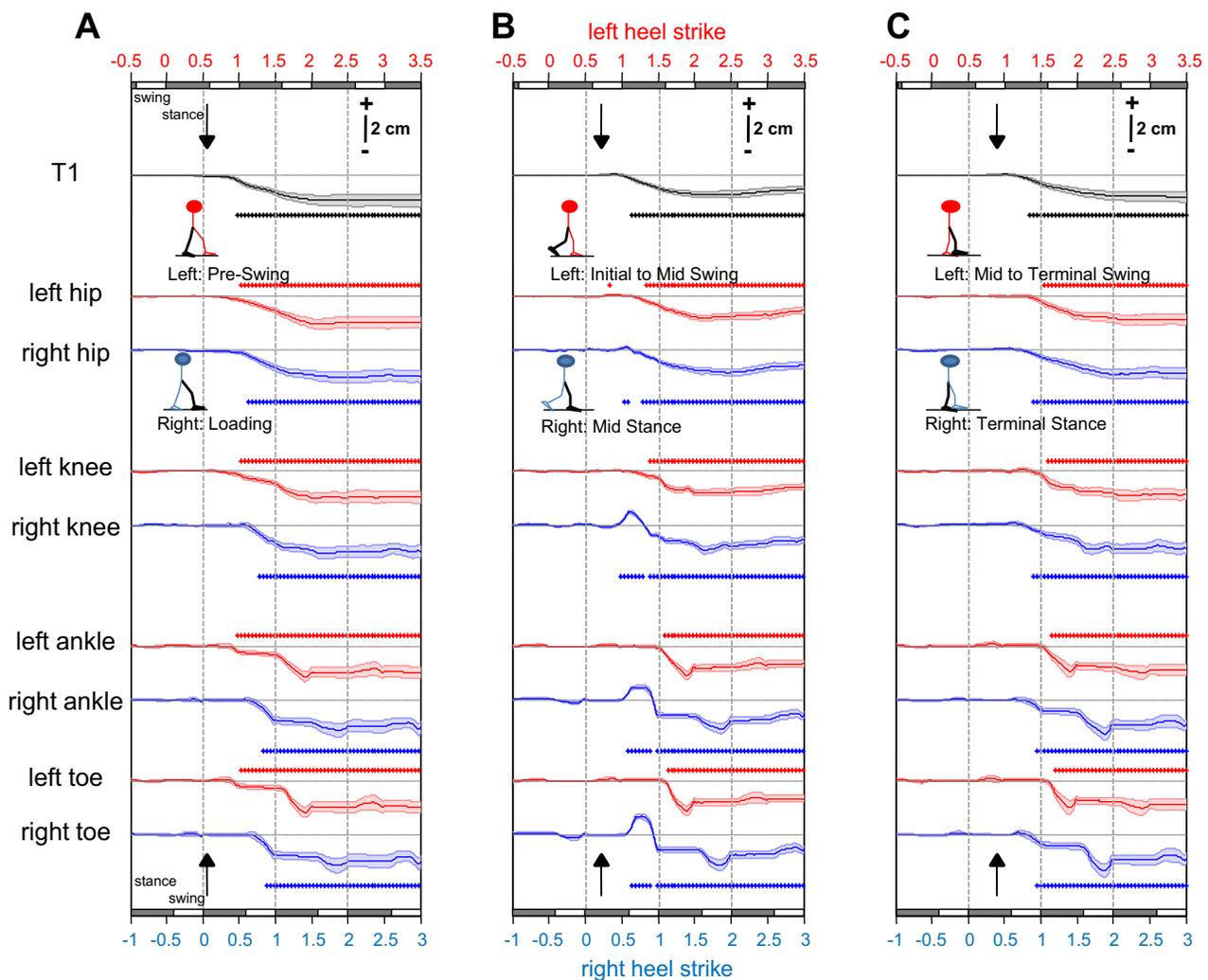


Fig. 8. Whole body displacements during negative perturbations. Residual anterior/posterior (A/P) displacements averaged across subjects are presented here for the 3 phases of negative (backward) visual scene motion. Displacements of right/left side of the body are in blue/red and midline displacement (T1) in black. Dashed vertical lines denote 0 difference from mean displacement, while the inset scale bar notes $-/+$ indicating a backward/forward displacement of the body on the treadmill. Arrows have been placed at the average perturbation phase, as the right leg is in the gait cycle phase of loading (A), midstance (B), and terminal stance (C) while the left leg is in preswing (A), initial to midswing (B), and midswing to terminal swing (C). Blue/red asterisks denote that the right/left side displacement was different from 0 at that bin, and black asterisks denote that the midline marker on T1 was different from 0 at that bin. These asterisks have been placed at the midpoint of the bin, as these continuous traces were binned when performing statistics ($n = 11$, FDR < 0.05). An x -axis value of 0 is the heel strike prior to visual scene motion. Shaded error bars correspond to SE. Heel strike timing for right/left legs is on the *bottom/top* x -axis of each subplot.

ward by the negative visual scene motion at all phases and the timing profile of displacement responses is consistent with responses observed in the segment angles.

DISCUSSION

Transient visual scene motion was used here to investigate phase-dependent responses to visual input during human locomotion. Trunk responses to approaching (negative) perturbations were only weakly phase dependent and instead primarily depended on the delay from the perturbation. In contrast, leg responses were strongly phase dependent. Leg responses during the same gait cycle as the perturbation exhibited gating, occurring when the perturbation was applied at midstance. Leg responses during the postperturbation gait cycle, however, exhibited a response-phase pattern over a range of perturbation phases. These two types of leg responses likely serve separate functions during the locomotion task used in this study. Overall, these results support the notion that the phase dependence of responses to visual input is determined by the functions, or subtasks, associated with vision during walking.

Directional Asymmetry

Across segments, responses to perturbations when the virtual scene was moving toward the subject (negative) were greater than responses, if they occurred at all, to visual perturbations moving away (positive) from the subject. These results are consistent with previous studies using continuous optic flow stimuli, which show that visual perturbations approaching a subject on a treadmill have a larger effect on step velocity (Konczak 1994), stride length (Prokop et al. 1997), and variability of ankle-knee relative phase and variability of center of mass displacement (Guerin and Bardy 2008) compared with perturbations moving away from the subject.

Although we have used transient perturbations in our study, the nervous system places an increased emphasis on approaching changes in visual scene motion, suggesting a functional aspect regardless of being a continuous flow or a transient change from fixed scene motion. Both approaching and receding visual scene motion are everyday occurrences during walking. It is normal for both directions of motion to occur, yet an approaching visual scene is more likely to require a functional change in the locomotive behavior. Transient changes in scene motion toward the subject in this study may be interpreted as an impending obstacle to avoid (due to large peripheral component), a wall that the approaching body will collide with, or a change in position on the treadmill, necessitating a response. In the alternative case of receding perturbations, a response is functionally less critical.

Strong Phase Dependence of Leg Responses but Not Trunk Responses

We quantified phase dependence along a continuum in this study, and trunk responses were found to be weakly phase dependent. To supplement this finding, latency to the midpoint of an initial, significant backward tilt was similar at all perturbation phases to approaching perturbations. Portions of stereotyped response waveforms have also been used to investigate phase dependence during walking in other sensory systems. Modulation of waveforms such as the “medium latency response” (e.g., Blouin et al. 2011; Iles et al. 2007) for galvanic

vestibular perturbations or the H-reflex (e.g., Capaday and Stein 1986; Dietz et al. 1990) for perturbations to proprioceptive afferents support the use of a feature of responses to determine dependence on phase of gait cycle. The midpoint of a stereotyped backward trunk tilt was used here to supplement the quantification of phase dependence, and it shifted later as perturbations were applied later. In sum, the trunk’s response is primarily dictated by the time delay from a change in scene motion and is most similar to the “no phase dependence” scenario in Fig. 1A rather than the gating (Fig. 1B) or response-phase pattern (Fig. 1C) scenarios.

The weak phase dependence of trunk responses in the present study is consistent with the responses of the trunk to continuous visual perturbations found in a previous study, which could be approximated with linear, time-invariant frequency response functions (Logan et al. 2010). In the task of treadmill walking and most forms of walking, there is a critical underlying subtask of maintaining postural control upright that the nervous system must continually perform to keep the oscillating trunk from toppling over the legs as they propel the body from place to place. Previous studies have shown that oscillatory motions of the trunk couple with oscillatory visual scene motion to stabilize the motion of the trunk at low frequencies (Logan et al. 2010; Warren et al. 1996). The long-lasting and weakly phase-dependent responses observed to these transient perturbations support the notion that trunk responses in this experiment also reflect a continuous, time-invariant maintenance of postural equilibrium during bipedal locomotion.

Responses observed in the leg, on the other hand, were strongly phase dependent, in two distinct ways. First, a phase-dependent response was observed in all leg segments within the same cycle as the visual perturbation. As seen in the third row of Fig. 5, within-stride deviations of the foot, shank, and thigh angles are observed when perturbations are presented solely at midstance, indicating that this response is gated (as in Fig. 1B). In the foot, the visual perturbation yielded a net increased plantar-flexion prior to toe off and an increased dorsi-flexion after toe off. At the same time, deviations in the shank render an increased flexion of the shank prior to the stance to swing transition followed by decreased flexion at the knee as the shank is extending in swing until heel strike. The thigh also displays more pronounced flexion throughout the swing phase of the perturbation cycle. Such flexion of the thigh and foot coupled with extension of the shank reveals a combination of segment angles that result in an overall higher position of the leg segments during swing. Significant decreases in SOL and LG activation observed in the response phase of preswing (third row of Fig. 6) likely account for the net decreased angle (dorsi-flexion) of the foot in ensuing swing when perturbations occur at midstance. Additionally, the significant increases in TA observed in early swing (Fig. 6O) also likely play a role in this dorsi-flexion observed in the foot during swing. In sum, the combined action of a decrease in lower limb plantar-flexor muscle activity and increased dorsi-flexor activity causes the eventual foot dorsi-flexion that occurs in midswing when perturbations occur during midstance perturbations.

This early phase-dependent response observed in the leg segments serves a within-cycle function in response to midstance perturbations. Midstance may mark a visually sensitive period in which a fast pathway “hazard detector” (Marigold

2008) is activated to avoid or accommodate hazards such as impending collisions with a wall or obstacles in the ensuing swing phase. Although there has been work in animal models (Graziano and Cooke 2006; Sherk and Fowler 2001; Simmons et al. 2010) regarding defensive postures prior to collisions, there is not, to our knowledge, a definitive account of kinematic or muscular strategies in response to an incoming whole body collision during human locomotion. It is premature to assume that fast pathways that may initiate obstacle avoidance and whole body collision avoidance are separable. However, common responses observed in this study and the studies reported below suggest that fast, approaching changes in visual scene motion trigger a fast, generalized hazard accommodation response regardless of the features (wall, obstacle, etc.) of the visual scene motion.

As responses in the leg segments to these types of perturbations have not been reported elsewhere, studies of obstacle avoidance support the idea that the kinematic features and timing of this response are a response to an approaching hazard. Both the foot dorsi-flexion and thigh flexion observed in the swing response phase in this investigation (in response to midstance perturbations) are main components of the “elevating strategy” for avoiding small obstacles occurring within the same gait cycle (Eng et al. 1994; Patla 1991). Although our attention to gait cycle phase dictates normalizing by gait cycle, mean response latencies calculated for muscular activity in real time reveal latencies with timescales comparable to those found in a previous study whose focus was avoidance of actual obstacles (Marigold et al. 2007). Despite the absence of “real” obstacles in this study, our approach is similar to that of Marigold and colleagues, as both studies introduce a dramatic change in the peripheral visual field with responses observed in the same cycle. They found that the BF is a reliable first muscle to be activated, with a mean response latency of 134 ms when obstacles are released in late stance. In our study, the SOL was the most reliable first muscle to be activated, with a mean response latency of 338 ms (method of Marigold et al. 2007) when negative visual scene perturbations were presented at midstance. In the study of Marigold et al., response latencies grew larger as obstacles were presented earlier in the gait cycle. With this caveat in mind, the 134-ms BF response latency to a late stance obstacle could be extrapolated to a 314-ms latency for a midstance obstacle in our study (obstacle presentation 15% earlier in the cycle), suggesting muscle response latencies on similar timescales. Thus the kinematic features and muscular timing of the responses to the approaching, midstance perturbations display a response that is functionally distinct from other trunk and leg responses we observed, indicating an alteration of leg segment trajectory within cycle in a manner resembling a hazard accommodation/avoidance response.

An alternative explanation to this proposed hazard accommodation/avoidance response would be that this response is caused purely by a visual startle during walking. This distinct, within-cycle response observed in the leg segments to a transient perturbation is reminiscent of auditory startle and its interaction with obstacle avoidance in human locomotion (Nieuwenhuijzen et al. 2000; Queralt et al. 2008). An auditory startle paired with an obstacle has been shown to improve the obstacle avoidance response and associated modulation of EMG amplitude (Queralt et al. 2008), indicating that startle to

sensory perturbations can have functional implications and may play a role in the response observed in this study when midstance perturbations are presented. Auditory startle responses in the musculature have been shown to occur at all perturbations, however, with amplitude modulated by perturbation phase without changes to the fast (<150 ms) response latencies (Nieuwenhuijzen et al. 2000). Here we observe little consistency across perturbation phases in measuring the existence of significant EMG responses. The additional response in leg segments that does not rely on time from stimulus, but instead shows a response-phase pattern, indicates that we have not measured a modulation of fixed-latency startle response for the whole body. Interestingly, auditory startle has been shown to shorten the gait period of both the perturbation and postperturbation cycles while decreasing both the maximal peak flexion and extension of both the ankle and knee (Nieuwenhuijzen et al. 2000). The increases in both the maximal extension and flexion of foot and shank angles observed here (Fig. 5, C and I) are actually the opposite of that found in auditory startle. Moreover, the gait period of the postperturbation cycle (+1 cycle in third row of Fig. 7B) is increased. These discrepancies suggest that the response to midstance perturbation is not a visual startle response.

For this specific response in the legs, the time-domain measures we used allowed us to observe an EMG response to visual scene motion that precedes a kinematic response during midstance perturbations. However, we did not observe clear EMG responses during the additional phase-dependent response of the legs or the response in the trunk. There could be different reasons for this discrepancy. It is well known that muscle activation patterns for unperturbed walking show large intersubject variability in addition to variability stride to stride, while kinematics remain relatively invariant (Winter and Yack 1987). Indeed, the results reported here also show more consistent, interpretable results in responses of kinematics compared with EMG during walking with perturbed visual scene motion. The lack of a systematic relationship for all responses could be due to missing action of muscles that were not recorded, not performing enough perturbations, and the potential of synergistic coactivation of muscles for function.

The second phase-dependent response occurred when perturbations presented at all phases elicited consistent responses of the leg segments in the gait cycle subsequent to the cycle in which a perturbation occurred (postperturbation cycle, *heel strike 1–2* in Fig. 5). As seen in the postperturbation cycle in Fig. 5, B–F, deviations in the foot segment angle translate to a net decreased plantar-flexion of the foot prior to toe off and a net decreased dorsi-flexion after toe off. Meanwhile, shank angle deviations translate to a net decreased flexion of the shank at the knee prior to the stance-swing transition followed by an increased flexion at the knee as the knee is extending in midswing. We consider this to be indicative of a response-phase pattern and in line with Fig. 1C, as it consistently occurred at the same response phase with characteristic waveforms in the foot and shank. Together, these deviations illustrate decreased range of motion of the lower leg on both sides of the stance to swing transition in the cycle following the perturbation. In other words, the legs adjust to slow the subject so that a central position can be maintained after a forward translation is determined from approaching scene motion. Such changes occur in the postperturbation gait cycle, which is

consistent with previous studies that show subjects use vision for changes in planned stepping adjustments approximately two steps, or a full gait cycle, in advance of a change in ground terrain (Marigold and Patla 2007; Patla and Vickers 2003).

This interpretation is also supported by the changes in stride length and gait period presented in Fig. 7. Various combinations of stride length and gait period allow maintenance of the same speed during treadmill walking (Dingwell et al. 2010). Increasing stride length could be counteracted with increasing gait period (as well as both decreasing) to not change the subject's speed or position on the treadmill. In the postperturbation cycle (+1 cycle in Fig. 7), the two gait measures never changed in the same direction, indicating that the approaching (negative) visual scene motion caused a change in position on the treadmill. Five of the six perturbation phases yielded either a decreased stride length or an increased gait period to slow subjects in the postperturbation cycle (+1 cycle in Fig. 7). Furthermore, there were three perturbation phases (midstance, terminal stance, preswing) with both decreases in stride length and increases in gait period. This combination of changes in stride length and gait period strongly supports slowing on the treadmill. These results coupled with the backward displacements of leg segments in the postperturbation cycle (*heel strike 1–2*) observed in Fig. 8 further support the notion that the major adjustments in speed occur primarily in the postperturbation cycle regardless of perturbation phase.

Smaller stride lengths coupled with counteracting shorter gait periods (or increases in both), however, would maintain similar speed on the treadmill. Indeed, significant changes of both stride length and gait period in the same direction are observed in the perturbation cycle (perturbation cycle in Fig. 7) in three perturbation phases (midstance, terminal stance, midswing to terminal swing), while the other three perturbation phases are accompanied by nonsignificant trends of gait period in the same direction that aid in counteracting significant changes in stride length. These findings are generally consistent with the lack of positional maintenance changes in segment angles observed in Fig. 5 in the perturbation cycle (0–1 in Fig. 5) and consistent with little change in displacements in the perturbation cycle (0–1 in Fig. 8). Although changes in segment angles occur in the perturbation cycle when perturbations occur at midstance (Fig. 5, *C*, *I*, and *O*), the deviations in leg displacements occurring in the perturbation cycle are corrected within cycle (Fig. 8*B*, blue lines), further supporting the notion that the within-cycle responses to midstance perturbations do not support the function of changes in speed over a stride. Additionally, changes in gait measures and displacements inform as to why deviations in the postperturbation cycle are not significant in the legs during perturbations presented at loading (Fig. 5, *A*, *G*, and *M*). Decreases in stride length in the perturbation cycle in addition to increases in gait period in the postperturbation cycle (first row of Fig. 7, *A* and *B*) coupled with the backward displacements in the perturbation cycle (Fig. 8*A*, 0–1, blue lines) in addition to the postperturbation cycle suggest that adjustment for maintaining position is spread across both the perturbation cycle and postperturbation cycle when perturbations were presented at loading. We speculate that this is the reason for a lack of significant, stereotyped kinematic deviations in the postperturbation cycle (1–2 in Fig. 5, *A*, *G*, and *M*) when perturbations were applied at loading. Further experimentation will be needed to test this idea that

adjustments in speed are enacted across both the perturbation and postperturbation cycles when visual scene motion is presented at the onset of the gait cycle.

Phase Dependence of Vision Dependent on Function

Visual perturbations delivered at different phases of the gait cycle revealed several responses in the trunk and legs that reflect distinct functions during locomotion. Changes in visual scene motion were shown to 1) modulate trunk orientation for upright equilibrium, 2) alter leg motion for hazard accommodation/avoidance, and 3) alter leg motion to control position on the treadmill. Such results are consistent with perturbations using other modalities during locomotion. For example, Bent et al. (2004) showed distinct leg and trunk responses due to galvanic vestibular stimulation; foot placement responses were dependent on phase of perturbation, while trunk responses showed no such dependence. Bent and colleagues suggest that these responses are indicative of independent upper and lower body control during walking (Bent et al. 2004, 2005). As distinct responses were observed both within the legs and between the trunk and leg segments in this study, the nature of phase dependence cannot simply be ascribed to the level of segments but relies on the functional subtask for which vision was used.

Subtasks, or necessary elements underlying the task of safe walking, typically consist of support maintenance by the stance limb, control of posture upright in sagittal and frontal planes, and proper foot trajectory control (Winter 1989). In more dynamic environments, however, additional subtasks such as hazard accommodation/avoidance or positional maintenance on a treadmill must be successfully performed. Recently, there has been renewed support for the idea that subtasks of human locomotion are modular in nature and are enacted by distinct combinations of muscle activations (McGowan et al. 2010; Neptune et al. 2009). Such studies echo previous work in human locomotion suggesting that the muscular activations performing the “subtasks” of locomotion are a few, underlying temporal components (Ivanenko et al. 2004).

Such subtasks often occur within the same gait cycle of the overarching task of bipedal walking, and visual input at a phase of the gait cycle critical for one subtask is not necessarily relevant for another subtask. Subtask-dependent timing is likely mediated by separate neural control pathways. For example, the within-cycle subtask and the subtask occurring in the postperturbation cycle of the leg observed in this study are accomplished through parallel online feedback and slower, feedforward pathways (Marigold 2008), respectively. Candidate pathways for parallel activations are cortical area MST and associated areas VIP and CSv, which have been found to have varying degrees of egocentric and allocentric tuning to visual scene motion (Wall and Smith 2008). Such pathways may underlie the distinction between changes in self-motion and motion of hazards in the external world. As these neural control pathways are further explored in animal models and patient populations, it is critical to take gait cycle phase into account. The nervous system's modulation of responses to sensory perturbations throughout the gait cycle is particularly important when using sensory inputs such as vision to aid in the restoration of specific motor functioning (subtasks) underlying locomotion.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: D.L., Y.P.I., T.K., F.L., and J.J.J. conception and design of research; D.L., G.C., and F.S.-L. performed experiments; D.L. and T.K. analyzed data; D.L., Y.P.I., T.K., F.L., and J.J.J. interpreted results of experiments; D.L. prepared figures; D.L. and J.J.J. drafted manuscript; D.L., Y.P.I., T.K., F.L., and J.J.J. edited and revised manuscript; D.L., Y.P.I., T.K., F.L., and J.J.J. approved final version of manuscript.

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