Kinematics in Newly Walking Toddlers Does Not Depend Upon Postural Stability

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INTRODUCTION

In contrast with humans, many animal species can stand and walk within hours after birth. To deal with an inherently unstable upright posture, bipedal human locomotion involves a significant dependence on descending pathways (Capaday 2002; Ivanenko et al. 2000; Orlovsky et al. 1999) that are not mature in infants at the age of 1 yr (Paus et al. 1999). One oft-cited hypothesis is that “what allows the infant to begin to walk independently at the end of the first year is not necessarily maturation of the stepping pattern but instead maturation of the system that enables successful balance control” (Pearson and Gordon 2000). Indeed, when supported, infants can step long distances (forward, backward, sideways), and with appropriate coordinated behavior in response to external perturbations (Lam et al. 2003; Lamb and Yang 2000; Pang and Yang 2001). Nevertheless, the question remains as to what extent immaturity of gait kinematics in toddlers is due to postural instability as opposed to immaturity of the generating networks. Few previous studies have dealt with the effects of postural stability on the development of walking and these have reported mainly the degree of variability in joint rotations or various phase characteristics (Clark et al. 1988; Lasko-McCarthy et al. 1990). However, direct quantitative demonstration of the effect of posture-stabilizing maneuvers on the global aspects of the toddler’s gait, such as the inverted pendulum mechanism of walking (Cavagna et al. 1983), planar co-variance of the angular motion of the lower limb segments (Lacquaniti et al. 1999, 2002) or foot trajectory control (Ivanenko et al. 2002), is lacking.

When children start to walk without support, their bodies display considerable oscillations in different directions revealing postural instability (Assaiante et al. 1993; Bril and Brenière 1993; Yaguramaki and Kimura 2002). Postural instability in turn might affect the state of the control system, change reliance on vestibular (Fitzpatrick et al. 1994) or proprioceptive (Ivanenko et al. 1999) sensory cues, and strengthen participation of the cortical motor areas in equilibrium maintenance (Ouchi et al. 1999; Solorpova et al. 2003). Therefore equilibrium instabilities could reorganize a coordination pattern and augment kinematic variability in walking toddlers as occurs in adults under unstable walking conditions (Cham and Redfern 2002; Lejeune et al. 1998; Menz et al. 2003). Subjects with a high risk of falling typically exhibit reduced temporal-spatial gait parameters and increased step timing variability, the features typical for toddlers. Moreover, drastic changes in lower extremity behavior might occur even when there is a perceived potential risk of falls (Cham and Redfern 2002). Therefore one can expect that postural instability represents a perturbing factor that changes the state of the control system and prevents the expression of the mature coordination pattern in toddlers.

Here we tested this hypothesis by attempting to stabilize the child’s body and thus give the toddlers greater confidence in walking. In adults, hand contact with an external surface results in a significant decrease of trunk and limb segment oscillations when posture is unstable (Dickstein and Laufuer 2004; Ivanenko et al. 1999; Jeka and Lackner 1994). In toddlers, hand support represents a common strategy used by the parent to prevent the
child’s falls. We used a similar approach to increase postural stability in toddlers who were just beginning to walk independently. We also recorded walking at different speeds in adults (including stepping in place) and older children to highlight common kinematic principles that might be responsible for shaping the toddler’s gait. To compare kinematic patterns, we used the methods developed earlier for adult’s gait (Bianchi et al. 1998; Borghese et al. 1996; Ivanenko et al. 2002). The results show that at the moment of transition to independent walking, immaturity of inter-segmental coordination and high foot trajectory variability do not depend on postural imbalance. Instead, our data suggest that a few months of unsupported locomotion experience are apparently necessary for calibration of the innate stepping mechanics to the unconstrained walking conditions. We also argue that the precursor of the mature kinematic pattern consists of a locomotor strategy combining forward progression with elements of stepping in place.

METHODS

Subjects

We recorded surface locomotion in 7 toddlers (3 males, 4 females, 12–15 mo of age), 7 older children (2–7 yr old), and 10 healthy adults [5 females and 5 males, 28 ± 7 (SD) years old]. Informed consent was obtained from all the adults and from the parents of the children. The procedures were approved by the ethics committee of the Santa Lucia Institute and conformed with the Declaration of Helsinki. The laboratory setting and the experimental procedures were adapted to the children so as to result in absent or minimal risk, equal or lower to that of walking at home. Both a parent and an experimenter remained alongside the younger children to prevent them from hurting themselves during falls. For the toddlers, daily recording sessions were programmed around the parents’ expectation of the very first day of independent walking until unsupported locomotion was recorded.

Walking conditions

WALKING WITHOUT SUPPORT. For the recording of the very first steps, one parent initially held the toddler by hand. Then the parent started to move forward, leaving the toddler’s hand and encouraging her or him to walk unsupported on the floor. For each subject, ~10 trials were recorded under similar conditions. Short trials (~3 min, depending on endurance and tolerance) were recorded with rest breaks in between. Only sequences of steps executed naturally by the toddler (e.g., no stop between steps) and while looking forward, were considered to avoid initiation and braking phases and head movements due to looking in other directions. The mean walking speed in toddlers was 1.4 ± 0.7 km/h. Adult subjects were asked to walk at a natural, freely chosen speed (on average, 3.8 ± 0.4 km/h) and at lower speeds in additional trials and to step in place. Typically, we analyzed two to five consecutive step cycles in each trial for toddlers, older children, and adults.

WALKING WITH HAND CONTACT. One hand of the child was held in the hand of a parent, while the other parent (or an experimenter) encouraged the child to walk straight ahead. This condition was recorded in all seven toddlers and older children.

WALKING WITH TRUNK SUPPORT. As an alternative way to reduce the effects of postural instability on lower limb kinematics, in four toddlers, additional trials were recorded while an experimenter (or a parent) firmly held the trunk of the child with both hands and supplied only limited vertical force (typically <20–30% of the body weight, as estimated from the mean vertical force on the platform) during stepping attempts.

WALKING BEFORE THE ONSET OF UNSUPPORTED LOCOMOTION. Supported steps of four toddlers (2 males and 2 females) were also recorded between 1.5 and 4 mo before the onset of unsupported walking as well as after this event to follow the early gait maturation. The infants walked firmly supported by one or both hands of one of their parents.

Data recording

Bilateral kinematics of locomotion was recorded at a digitizing rate of 100 Hz by means of the VICON-612 motion analysis system (Oxford, UK). The positions of selected points on the body were recorded by attaching passive infrared reflective markers (diameter: 1.4 cm) to the skin overlaying the following bony landmarks on both sides of the body: gleno-humeral joint (GH), the tubercle of the anterosuperior iliac crest (IL), greater trochanter (GT), lateral femur epicondyle (LE), lateral malleolus (LM), and fifth metatarso-phalangeal joint (VM).

In children and in adults at low speeds, the ground reaction forces under both feet were recorded at 1,000 Hz by a force platform (0.9 × 0.6 m, Kistler 9287B, Zürich, Switzerland). Toddlers generally performed two to three steps on the force platform in each trial. At natural, higher speeds in adults, the ground reaction forces under each foot were recorded separately by means of two force platforms (0.6 × 0.4 m, Kistler 9281B), spaced by 0.2 m between each other in both the longitudinal and the lateral direction.

Electromyographic (EMG) activity was recorded by means of surface electrodes from the rectus femoris (RF), hamstring (HS), tibialis anterior (TA), and soleus-gastrocnemius (GC) muscles. EMG signals were preconditioned at the recording site (active electrodes from BTS, Milan, Italy or DelSys, Boston, MA), transmitted to the remote amplifier (bandwidth was 20–200 Hz), and sampled at 1,000 Hz. Some crosstalk from nearby muscles is inevitable in tiny limbs of young infants. Nevertheless, due to the low skin impedance and preconditioning at the recording site, no artifacts appeared due to movement of electrode cables and cross talk from the antagonistic muscles seems to be limited (Forssberg 1985; Okamoto et al. 2003; Sündemier et al. 2001). Sampling of kinematic, kinetic, and EMG data was synchronized.

Data analysis

We analyzed and separately presented the effect of support on postural stability and general gait characteristics (such as walking speed, percent of falls, step length and width, trunk oscillations) and on the kinematic patterns of walking (intersegmental coordination, EMG patterns, and foot trajectory control).

Deviations of gait trajectory relative to the x direction of the recording system were corrected by rotating the xz axes by the angle of drift computed between start and end of the trajectory. The body was modeled as an interconnected chain of rigid segments: GH-IL for the trunk, IL-GT for the pelvis, GT-LE for the thigh, LE-LM for the shank, and LM-VM for the foot. The main limb axis was defined as GT-LM. The elevation angle of each segment corresponds to the angle between the segment projected on the sagittal plane and the vertical (positive in the forward direction, i.e., when the distal marker falls anterior to the proximal one).

To evaluate trunk stability with respect to the vertical axis, we measured the peak-to-peak angular deviation of the long axis of the trunk in both sagittal and frontal planes. The long axis of the trunk was defined by connecting the midpoint of the two (left and right) IL markers with the midpoint of the two GH markers. Percent of falls was computed as the number of trials in which the toddlers fell divided by the total number of recording trials. Walking speed was measured by computing the mean velocity of the horizontal IL marker movement. The length of the lower limb (L) was measured as thigh (GT-LE) plus shank (LE-LM) length.
Gait cycle duration was defined as the time interval between two successive maxima of the elevation angle of the main limb axis of the same limb and stance phase as the time interval between the maximum and minimum values of the same angle (Borghese et al. 1996). Thus a gait cycle (stride) referred to a cyclic movement of one leg and equalled two steps. When subjects stepped on the force platforms, these kinematic criteria were verified by comparison with foot strike and lift-off measured from the changes of the vertical force around a fixed threshold. In general, the difference between the time events measured from kinematics and kinetics was <3%. However, the kinematic criterion sometimes produced a significant error in the identification of stance onset in toddlers if there was an unusual forward foot overshoot at the end of swing (cf. Forsberg 1985). In such cases, foot contact was determined using a relative amplitude criterion for the vertical displacement of the VM marker (when it was elevated to 7% of the limb length from the floor).

Raw EMG data were numerically high-pass filtered (cutoff: 30 Hz) to remove motion artifacts, rectified and then low-pass filtered with a zero-lag Butterworth filter (cutoff: 15 Hz). Data from several steps were ensemble-averaged after time-interpolation over individual gait cycles to a normalized 200-point time base.

The time-varying coordinates of the center of pressure (CoP) were derived from the force plate measurements. Step length (variation of CoP, between 2 foot contacts with the floor) and width (variation of CoP,) were calculated from the force plate data (Ledebt and Bril 2000) and normalized to the limb length. Duration of the single support phase (i.e., time of the step spent with only 1 foot on the ground) was normalized with respect to the duration of the step.

Ensemble averages with SD of kinematic variables were computed at each point of the normalized time base. The mean SD values were computed to characterize deviations of the individual step traces from the ensemble average.

Intersegmental coordination

The intersegmental coordination was evaluated in position space as previously described (Bianchi et al. 1998; Borghese et al. 1996). In adults, the temporal changes of the elevation angles at the thigh, shank, and foot co-vary during walking. When these angles are plotted one versus the others in a three-dimensional graph, they describe a path that can be fitted (in the least-square sense) by a plane over each gait cycle. Here, we studied the development of the gait loop and its associated plane in children. To this end, we computed the covariance matrix of the ensemble of time-varying elevation angles (after subtraction of their mean value) over each gait cycle. The three eigenvectors \( u_1-u_3 \), rank ordered on the basis of the corresponding eigenvalues, correspond to the orthogonal directions of maximum variance in the sample scatter. For each eigenvector, the parameters \( u_1, u_2, \) and \( u_3 \) correspond to the direction cosines with the positive semi-axis of the thigh, shank, and foot angular coordinates, respectively. The first two eigenvectors \( u_1-u_2 \) lie on the best-fitting plane of angular co-variation. The third eigenvector \( u_3 \) is the normal to the plane and defines the plane orientation.

Foot trajectory

The shape of the endpoint path was compared by computing the vertical excursion of the VM marker (during swing) and correlating it with the corresponding ensemble average in adults. VM trajectories were time-normalized over the swing phase duration.

Foot-trajectory variability

Foot-trajectory spatial variability in the sagittal plane was quantified in terms of spatial density and normalized tolerance area of VM, computed over the swing phase (Ivanenko et al. 2002). These indices describe the integrated variability of foot path, including variability in both the vertical and horizontal directions. To compare subjects of different heights, VM trajectories (relative to the instantaneous position of GT) were first scaled by the limb length (in proportion to the mean limb length of adults) and then re-sampled in the space domain by means of linear interpolation of the \( x,y \) time series (1.5-mm steps) over all gait cycles. All steps (typically 7–15) from different trials under the same walking conditions were pooled together for this analysis.

Spatial density was calculated as the number of points falling in \( 1 \times 1 \, \text{cm}^2 \) cells of the spatial grid divided by the number of step cycles. The density of each cell was depicted graphically by means of a color scale (empty cells were excluded): the lower the density (toward the blue in the color-cued scale), the greater the variability. Normalized tolerance area was derived as follows. The mean length of foot trajectory over the swing phase across all steps was calculated from the corresponding path integral. For every interval corresponding to 10% of the maximal horizontal excursion, we computed the two-dimensional 95%-tolerance ellipsis of the points within the interval. The typical number of points in each interval ranged from 500 to 1,200 (depending on the number of gait cycles). The areas of all tolerance ellipses were summed and normalized by the mean length of foot trajectory. This normalized area provides an estimate of the mean area covered by the points along 1 cm of path. A greater tolerance area indicates greater variability.

Statistics

Statistical analyses (Student’s \( t \)-test) were used when appropriate. Reported results are considered significant for \( p < 0.05 \). Statistics on correlation coefficients was performed on the normally distributed, Z-transformed values. Spherical statistics on directional data were used to characterize the mean orientation of the normal to the co-variation plane (see preceding text) and its variability across steps. To assess the variability, we calculated the angular SD (called spherical angular dispersion) of the normal to the plane.

RESULTS

Effect of hand contact on postural stability and general gait characteristics

When children start to walk without support, their bodies display considerable oscillations due to postural instability: peak-to-peak sway of the trunk was 14.1 ± 3.4° in the sagittal plane and 9.5 ± 1.9° in the frontal plane, both values being significantly \((P < 10^{-5})\) higher than in adults \((6.2 ± 0.8 \text{ and } 2.7 ± 0.6°, \text{respectively})\). As a rule, toddlers take relatively few steps and readily fall. Percent of falls in the children was relatively large \((37 ± 23\% \text{ in the recording trials})\).

At the onset of unaided walking, general gait parameters display high variability. There are considerable variable lateral trunk displacements (both GT and GH markers, Fig. 1B, left). To minimize disequilibrium, toddlers take short steps with a wide base of support and a prolonged time in double support (Fig. 2A) (see also Bril and Brenière 1993). The walking speed freely chosen by toddlers was considerably lower than in adults or in older children when expressed in dynamically equivalent terms—Froude number \( (Fr = V^2 g^{-0.5} L^{-1}) \), where \( V \) is the average speed of locomotion, \( g \) the acceleration of gravity, and \( L \) the limb length). The Froude number is a dimension-less parameter suitable for the comparison of locomotion in subjects of different size walking at different speeds, under the assumption of the gravity-related pendulum mechanism of movement (Cavagna et al. 1983). \( Fr = 0.04 ± 0.03 \text{ in} \)
toddlers versus 0.21 ± 0.08 in adults and 0.23 ± 0.03 in older children (Fig. 2, A and B).

To improve postural stability, in a series of trials the child walked with a hand lightly held by the hand of one parent. The procedure resulted in minimal additional contact forces but gave the toddlers greater confidence in walking than without hand contact (as shown by their increased willingness to walk in the laboratory). In contrast to older children (Fig. 2 B), the step length and the walking speed increased significantly and the step width decreased when walking with hand contact (Fig. 2A). The relative duration of the single support phase increased slightly but significantly in all toddlers. Trunk oscillations were significantly reduced: the reduction of sway in the sagittal plane amounted to 25% on average (Fig. 2A). Variability in the mediolateral GT oscillations, estimated as the mean SD from the ensemble averaged GTz waveform, was also reduced (being 0.041 ± 0.005 L with hand contact and 0.052 ± 0.009 L without, where L is the limb length; Fig. 2A, right). Finally, percent of falls in these supported trials decreased drastically (7 ± 6%, Fig. 2A, left).

In summary, hand contact had a strong impact on general characteristics of infant’s stepping (Fig. 2A). The same hand holding procedure in older children did not show any significant influence on these parameters (Fig. 2B).

Kinematic patterns of walking with and without hand contact

One question addressed in this study is whether postural instability represents a major factor inhibiting the expression of the mature stepping pattern in toddlers. To this end, we compared the kinematics of unaided unstable stepping with that of supported walking. Figure 3 shows a typical example of a stepping pattern in one adult and one toddler at the beginning of independent walking. Various general gait features are described in this figure.

First, in adults, the temporal changes of the elevation angles of lower limb segments co-vary along a plane, describing a characteristic loop over each stride (Fig. 3A, far right). In toddlers, the gait loop departed significantly from planarity and the mature pattern. Planarity was quantified by the percentage of variance accounted for by the third eigenvector (PV3) of the data covariance matrix: the closer PV3 is to 0, the smaller the deviation from planarity. PV3 was significantly higher in toddlers (3.1 ± 0.7%) than in adults (0.8 ± 0.3%, P < 0.001 Student’s unpaired t-test), in agreement with previous findings (Cheron et al. 2001b). Also because the amplitude of thigh movements was relatively higher with respect to that of shank and foot movements in toddlers, the gait loop was less elongated than in adults, as shown by the smaller contribution of the first eigenvector (PV1). Moreover, the step-by-step variability of plane orientation (estimated as the angular dispersion of the plane normal) was considerably higher in toddlers (16.1 ± 2.4°) than in adults (2.9 ± 1.0). In toddlers, neither the percent of variance PV1, PV2, and PV3, nor the orientation of the co-variation plane changed significantly across the two postural support conditions (Fig. 4A, right). Furthermore, the

FIG. 1. An example of unsupported and supported walking. A: unilateral sagittal stick diagrams when the toddler walked unsupported (left) and with the hand held in the hand of a parent (right). B: superimposed trajectories of the right gleno-humeral (GH) and greater trochanter (GT) markers across 10 steps (after subtracting the mean values) in the horizontal plane for unsupported (left) and supported walking (right).

FIG. 2. Effect of hand contact on global gait parameters. Left to right: percent of falls, normalized walking speed (Froude number = V^2g/L^1), step length, step width, relative single support phase duration, peak-to-peak pitch trunk oscillations, and variability in the lateral displacement of GT across 10 steps (estimated as the mean SD from the ensemble average). Step length, step width, and GT_lateral variability were normalized by the limb length to provide comparisons between children of different sizes. All parameters were significantly different in toddlers during walking with hand contact vs. unsupported walking. Older children (B) never fell during the test.
step-by-step variability of plane orientation remained unchanged (15.0 ± 2.8° with hand contact and 16.1 ± 2.4° without) reflecting a high degree of instability in the phase relationship between limb segments.

Second, in walking adults, the hip vaults over the stance leg as an inverted pendulum. As a result, we found two peaks in the temporal profile of vertical hip position (GT™ and IL™) over each gait cycle, in coincidence with midstance of the right and left leg, respectively (Fig. 3A, right). Fourier series expansion of GT™ revealed a clear dominance of the second harmonic: the percent of GT™ variance explained was 13 ± 7 and 80 ± 7% for the first and second harmonic, respectively. In toddlers, GT™ oscillations were variable from step to step. Their mean profile systematically differed from that of the adults. Thus independent of support conditions, the first peak in the adult GT™ (corresponding to the stance phase of the ipsilateral leg) was often absent in toddlers (Fig. 3A). Instead, toddler GT™ activity typically exhibited a peak corresponding to the second peak of the adult GT™ profile, and reflected a lift of the hip joint during swing relative to the contra-lateral hip joint of the load-bearing leg. This behavior was observed both during bilateral kinematic recordings and in the motion of the IL markers. Therefore we can exclude the possibility that the toddler GT™ peak originates from a misplacement of the GT marker relative to the center of joint rotation. In toddlers, the percent of GT™ variance explained by the first and second harmonic was 56 ± 9 and 23 ± 5%, respectively, indicating a dominance of the first harmonic (Fig. 4B). A lack of the pendulum behavior of the hip in toddlers was conserved across support conditions. Thus over the 0.07–0.20 range of Fr values, the percent of variance explained by the second harmonic of GT™ was 27 ± 7% with hand contact and 23 ± 5% without (P > 0.7 in all cases).
Third, foot trajectory characteristics differed systematically in toddlers as compared with those in adults. The dominant template of the foot motion is illustrated in the stick diagram of Fig. 3A (left). All toddlers moved the leg in such a way that the foot lift had only one maximum at midswing. Often, the toe reached its maximal height in front of the body at the end of swing. This behavior was entirely opposite to that observed in the typical adult gait, which was characterized by prominent foot lift in early swing, a minimum foot clearance during midswing, and another separate toe lift at the end of swing (Fig. 3A, right). As a result, the correlation coefficient between the time series of the VM_y during swing in toddlers and the corresponding ensemble average in adults was typically negative (−0.59 ± 0.22). The spatial variability of the endpoint (foot) path in the sagittal plane was considerably greater than in the adults (Fig. 3C): the normalized tolerance area (scaled to the mean limb length of adults) was 20.7 ± 3.1 cm²/cm in toddlers and 4.8 ± 1.9 cm²/cm in adults (P < 0.0001, Student’s unpaired t-test). Hand contact did not influence appreciably the shape (Fig. 3A) and spatial variability (Fig. 3C) of the endpoint path. The correlation coefficient between VM_y in toddlers and adults remained negative (−0.55 ± 0.20). The VM tolerance area remained high (17.9 ± 4.6 cm²/cm).

Finally, EMG activity in toddlers was variable across steps due to augmented step-by-step variability in the kinematics and in the speed of progression, though it comprised many features of adult gait. Nonplantigrade gait and foot placements were often accompanied by an atypical burst of activity in the gastrocnemius muscle at foot touchdown (Fig. 3B) (see also Forssberg et al. 1985; Okamoto et al. 2003), this burst was never observed in adults. The mean level of activation of leg muscles did not change significantly with hand contact. In the hand contact condition, the TA, GC, HS, and RF activity was 22 ± 10, 33 ± 11, 21 ± 10, and 18 ± 14 µV; in the unsupported condition, it was 20 ± 11, 27 ± 10, 20 ± 9, and 16 ± 15 µV, respectively. Characteristic EMG bursts at foot contact were also observed during supported stepping.

In summary, experimental interventions leading to increased postural stability and reduced trunk oscillations did not result in significant amelioration of the kinematic dis-coordination and EMG patterns in toddlers. Walking with hand contact also did not affect the kinematics of stepping movements in older children (Fig. 4B), which remained similar to that of adults across support conditions.

**Effect of trunk support**

In four toddlers, we also performed a more drastic maneuver to stabilize the body during walking. In some trials, an experimenter (or a parent) firmly held the trunk of the child with both hands while the child stepped. The mean walking speed (1.3 ± 0.6 km/h) was similar to that during unsupported stepping (1.4 ± 0.7 km/h). Supporters were instructed to avoid influencing the toddlers’ forward motion. However, we were unable to completely control for the possibility that some aspects of stepping were influenced by external forces generated by the supporter on the toddler’s forward motion. Thus the movement of the trunk (and in particular the vertical displacement of the hip joint) could not be analyzed reliably due to the presence of the external support. Nevertheless, leg movements and EMG activity were comparable to those during unsupported stepping (Fig. 3). The gait loop departed from planarity and from mature pattern in the two support conditions. The percent of variance accounted for by the third eigenvector (planarity, PV_3) was 2.5 ± 0.5% during walking with trunk support and 2.7 ± 0.3% without (cf. 0.8 ± 0.3% in adults). The step-by-step variability of plane orientation remained high (13.6 ± 3.6° with trunk support and 15.7 ± 3.2° without). The correlation coefficient between the time series of the VM_y during swing in toddlers and the corresponding ensemble average in adults remained negative (−0.51 ± 0.15) and the VM tolerance area remained high (19.2 ± 3.9 cm²/cm with trunk support and 17.0 ± 5.5 cm²/cm without). In summary, neither inter-step kinematic variability, nor the index of planarity (PV_3), nor the orientation of the plane of angular co-variation changed significantly (P > 0.05 in all cases).
Interestingly, the shape of the foot path, bilateral coordination of the two hip joints and EMG patterns in the toddlers were reminiscent of those observed for stepping in place in the adults (Fig. 5). The correlation coefficient between the time series of the vertical foot (VM) displacements during swing in the toddlers and the ensemble average in the adults for stepping in place was typically very high and positive (0.92 ± 0.09), while it was typically negative (−0.59 ± 0.22) when comparing with normal adult walking. As in the case of the toddler gait, GTv oscillations during stepping in place in adults displayed a prominent peak during swing and a tiny peak during stance (Fig. 5); as a consequence, the percent of GTv variance explained by the first and second harmonic was 52 ± 22 and 31 ± 13%, respectively.

In additional trials, adults were asked simultaneously to perform stepping-in-place-like vertical leg movements while moving slowly forward. The adults executed this task easily and the kinematics were very similar across adults. Again, in this task, the VMv and GTv behavior was similar to that of the toddlers (Fig. 5). Moreover, during stepping in place in adults, we detected bursts of EMG activity in the calf muscles at foot touchdown similar to those often observed in toddlers (Fig. 5), and the HS muscle regularly exhibited activity in the middle of swing phase. Furthermore, the gait loop during adult stepping in place dwindled to a line because all three segments moved in phase. However, when the adults were asked to simultaneously step in place and to move forward, this manipulation created phase shifts between segment rotations, and the shape of the gait loop became very similar to that of the toddlers (Fig. 5).

**Behavior before and after the onset of unsupported locomotion**

Four infants were tested repeatedly over a period between 4 mo before and 13 mo after the onset of independent walking (Fig. 6). During the experiments, the infants walked firmly supported by the hand of one of their parents before they could walk independently. In all recording sessions performed before the onset of unsupported locomotion, the pattern of inter-segmental coordination, the pendulum-related pattern of vertical hip displacement and foot trajectory characteristics all did not differ significantly from those recorded at the onset of unsupported locomotion. The percentage of variance accounted for by the second Fourier harmonic of the vertical GT displacement (denoting the double-peaked profile of pendular oscillations of COM) exhibited inter-step variability but did not change systematically as a function of age up to the time of onset of unsupported locomotion, when it started to increase rapidly over the first few months of independent walking experience (Fig. 6B). A similar trend was exhibited by the time course of the VMv (Fig. 6D), by the endpoint (VM) spatial variability (Fig. 6E), by the index of planarity of the gait loop (PV3, Fig. 6C) and by the step-by-step variability of plane orientation (angular dispersion of the plane normal, not shown).

**DISCUSSION**

There are three main findings in this study: immaturity of global gait parameters did not depend on postural stability, the toddler pattern shared fundamental features with adult stepping in place, and idiosyncratic gait parameters remained basically.
like or single-peak trajectories of the foot during swing. Trunk support also did not significantly improve gait kinematics (Fig. 3), although one can never be sure of having totally removed the problem of postural instability because translational and rotational trunk oscillations are inertially coupled with leg motion. However, if we take a realistic operational approach to the problem, we are confident that our results do reveal the absence of a strong relationship between postural instability and gait immaturity at the onset of independent walking.

Clearly, the task of maintaining stability when walking is considerably different from that of standing because the former necessitates an appropriate lower limb coordination pattern. Furthermore, the residual trunk oscillations during walking with hand support (Fig. 2, right) may likely be a result of the immature intersegmental coordination. The planar co-variance of the elevation angles of the lower limb segments is weak and variable at the time of the first unsupported steps (Cheron et al. 2001a,b). The maturation of the planar co-variance is functionally significant for the mechanics of walking (Bianchi et al. 1998) and is likely important for higher postural stability. The parallel development (similar time constants) of trunk stabilization, planar co-variance of the elevation angles (Cheron et al. 2001b), and the gravity-related pendulum mechanism of walking (Ivanenko et al. 2004) suggests that a dynamic integration of a gravity-centered reference emerges for equilibrium and forward propulsion.

An optimal cadence in walking roughly corresponds to the eigen (resonance) frequency of the swinging limbs coupled with inverted pendulum motion of the stance limb, as predicted by the pendulum mechanism of walking (Cavagna et al. 1983) and by the “ballistic walking” model proposed by Mochon and McMahon (1980), in which the swing limb behaves like a compound pendulum. Therefore matching of neural and mechanical oscillators might be essential both for minimization of energy consumption and for a higher dynamic stability because one of the benefits of pendular rhythmic movements is cycle-to-cycle stability and reproducibility (Goodman et al. 2000). In adults, the minimum variability of the foot trajectory and of the elevation angles occurs \(~\sim 3-4\) km/h and defines the optimal kinematic walking speed from the point of view of minimization of positional variance (Ivanenko et al. 2002) and \(~\sim 4.5\) km/h from the point of view of minimization of energy consumption (Cavagna et al. 1983). Optimal speed cannot be easily determined in newly walking toddlers as they usually walk over a limited range of speeds and steps. However, the kinematic pattern differed from the adult pendulum behavior and the kinematic variability was always considerably higher in the toddlers than in the adults, independent of speed or support conditions, suggesting that toddlers do not properly use the gravity-related properties of limb mechanics (transfer between potential and kinetic energies). In addition, the upper extremities in newly walking toddlers are held away from the body, whereas, as a child grows older, reciprocal arm swinging emerges (Sutherland et al. 1980).

Idiosyncratic kinematic features of a toddler’s gait

Prewalking children are typically able to stand up and maintain static equilibrium from \(~\sim 10\) mo of age (Zernicke et al. 1982). At the age of \(~\sim 1\) yr, maturation of central neuronal pathways arrives at a point whereby the necessary integrative capacity for balance and rhythmic leg activity allows unaided walking to take place even though the posture is unstable. Equilibrium instabilities, however, could reorganize a coordination pattern and augment kinematic variability in walking toddlers as occurs in adults under unstable walking conditions (Cham and Redfern 2002; Menz et al. 2003). However, our results show that even when supported through hand contact, which reduced balance difficulties and thereby increased confidence of stepping, toddlers still exhibited their idiosyncratic gait pattern, characterized by undeveloped phase coupling of limb segment motion, bilateral hip discoordination, lack of the pendulum behavior of the COM, characteristic EMG bursts at foot contact, high kinematic variability, and distinctive elliptic-
stepping, in particular, the vertical movement of the hip joint and of the foot, the shape of the gait loop, and the bursts of EMG activity on foot placements, all suggest that toddlers implement a mixed locomotor strategy, combining forward progression with elements of stepping in place. In fact, as noted in RESULTS, this toddler pattern is highly reminiscent of adult stepping in place accompanied by supplementary slow forward translation (see Fig. 5). It is also worth noting that stepping in place usually precedes independent walking because infants spend time stepping on the spot, when supported by an adult or while holding onto an object. Obviously, the linear component is fundamental for sustaining the locomotor pattern because unsupported toddlers do not typically step in place. Illustrations of the simple integration of stepping movements with forward translation can be found in the typical strategy of gait initiation in toddlers, whereby gait is initiated by letting the body fall forward, and in a form of early walking, whereby the toddler puts the swing foot forward to break the fall and bring the body back to the original stance foot (McCollum et al. 1995). Often, the “faller” cannot stop without something to bump into, such as a wall or friendly adult (McCollum et al. 1995). The linear component of foot motion in supported infants can also be evoked when stepping on a moving treadmill (Lamb and Yang 2000; Pang and Yang 2001; Yang et al. 1998). However, the behavior of the body as a compound inverted pendulum appears later with walking experience.

In infant stepping, several leg muscles emit short-latency EMG bursts when the foot contacts the ground (Fig. 3B). These bursts were attributed to hypersensitive stretch reflexes distributed to several muscles (Forssberg 1985, 1999; Okamoto et al. 2003). In contrast, our findings suggest that this characteristic EMG activity might be a result of nonplantigrade gait rather than hyperactivity of stretch reflexes because these bursts are always observed during adult stepping in place accompanied by supplementary linear translation (Fig. 5).

The occurrence of the prominent single-peak foot lift during swing corroborates Sherrington’s views on the involvement of the spinal flexion reflex in step generation (Sherrington 1910). A similar form of locomotor-like alternating kicking and stepping movements is present at birth and even during the prenatal period in humans (Forssberg 1985; Zelazo 1983). Thus one can recognize the above-mentioned features of toddler stepping in the stick diagrams, video recordings and EMG traces documented by others during supported neonatal stepping (≤4 wk after birth) and throughout the first year of life (1–12 mo after birth) (Forssberg 1985; Lamb and Yang 2000; Okamoto et al. 2003; Pang and Yang 2001; Yang et al. 1998), namely: high single-peak foot lift, short step length, disordered vertical hip displacements, and characteristic EMG bursts at foot contact.

Role of walking experience in gait maturation

Progressive changes of gait kinematics and kinetics depend on musculoskeletal growth (including foot shape modifications and ossification of the soft bones of the feet) (Bertsch et al. 2004), development of the vestibular system (Wiener-Vacher et al. 1996), central conduction delays (Eyre et al. 1991), and maturation of central neuronal pathways that are important for postural and locomotor control, the latter resulting in part from myelination of descending tracts (Paus et al. 1999). In addition to these factors, however, walking experience under unsupported conditions may act as a functional trigger of gait maturation because characteristic gait parameters were basically conserved until independent walking and then rapidly mature (Fig. 6). Thus the most dramatic phase of maturation takes place during the first months of independent walking (Sundermier et al. 2001), though anthropometrical changes and developmental tunings go on for many years. It is also worth noting that infants undergoing daily stepping exercise exhibit an earlier onset of independent walking than untrained infants (Zelazo et al. 1972). Consistent with learning of other motor skills, rapid maturation of the infant’s gait is accompanied by a similar rapid reduction in kinetic variability (e.g., Fig. 6E).

In a computational context, high variability may reflect the attempts of the CNS to explore a wide range of different kinematic solutions during development (Forssberg 1999; Konczak and Dichgans 1997; McCollum et al. 1995; Thelen and Smith 1994), and walking experience may act to accelerate the motor system’s ability to identify the optimal solution.

Underlying changes in information accessibility due to the enhanced freedom to explore the world beyond the territory at hand, coupled with improved cognitive capacity to generate complex associations (Butterworth 1998), may also be important for fully unaided walking to develop (Zelazo 1983). For instance, at the onset of independent walking, appropriate control of foot placement is greatly lacking: infants often neglect obstacles (e.g., toys located on the floor) when walking in a play area and a toddler’s ability to walk on uneven terrain or slopes is very limited (Adolph 1997). Although postural control constitutes a necessary ingredient of independent walking, our results clearly show that the onset of walking itself leads rapidly to stabilization of the locomotor pattern.

ACKNOWLEDGMENTS

We thank Dr. W. Miller for comments on the manuscript and V. Sabia for the toddler drawing (Fig. 3).

GRANTS

The financial support of Italian Health Ministry, Italian University Ministry (PRIN and FIRB projects), and Italian Space Agency is gratefully acknowledged.

REFERENCES


FIRST STEPS IN TODDLERS

763


