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RESEARCH ARTICLE

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Control of frontal plane body motion in human stepping

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Abstract During a step the body's centre of mass (CoM) typically remains medial to the supporting foot and therefore the body is unstable and falling (sideways) under gravity. This may make it difficult to adjust the frontal-plane body motion appreciably once the step is under way. We have therefore investigated whether this motion could be controlled largely in a ballistic manner, that is by setting the initial (toe-off) position and velocity of the CoM such that the fall develops as required for the particular step without the need for appreciable mid-step adjustment. Subjects stepped in different directions and from different postures, and the resulting motion of their CoM in the frontal plane was compared with that of a single-segment mathematical model of the body which falls freely under the influence of gravity. The lateral position and velocity of subjects' CoM at toe-off varied across the different step types in a manner consistent with a ballistic mode of control. Furthermore the model, given these positions and velocities as initial conditions, closely predicted the subsequent CoM motion. The results suggest that subjects may produce the different body trajectories required for different types of step largely in a ballistic manner. This would imply that the central nervous system must judge in advance the size and direction of the initial "throw" given to the body-mass.

Key words Balance · Walking · Movement

Introduction

Most studies of stepping or the initiation of gait which have considered the motion of the body-mass have focused on the antero-posterior component of this motion (e.g. Carlsöö 1966; Cook and Cozzens 1976; Brenière et al. 1987; Crenna and Frigo 1991; Burleigh and Horak 1996). However, in considering the control of balance during stepping, the medio-lateral component of body motion is particularly relevant since the change in the support conditions which occurs when a foot is lifted off the ground is predominantly in the frontal plane. This study begins an investigation into the control of balance in stepping by analysing the frontal plane motion of subjects' centre of mass (CoM) as they take a variety of steps.

In normal stance the CoM is midway between and some distance above the two feet, that is above the middle of the body's base of support. If one foot is lifted off the floor, the support conditions change. The body is now supported by only one limb and the base of support is greatly reduced in size, its area being that of the supporting foot's contact with the ground. It follows that lifting one foot off the floor (without any preliminary shift of the body-mass) leads to the body's CoM no longer being over its base of support. In these circumstances the body becomes unstable and begins to topple, pivoting about the ankle and falling downwards and sideways away from the supporting foot. In order to move from normal bipedal to stable unipedal stance this fall must either be avoided or be arrested once it has begun. Subjects appear to do the former. They accelerate their body-mass towards the forthcoming support side *before* lifting the foot (Rogers and Pai 1990; Mouchnino et al. 1992). The instability is avoided altogether and the CoM is brought to a position directly above the single supporting foot.

A similar preparatory lateral acceleration of the CoM to that seen in moving from bipedal to unipedal stance is observed when quietly standing subjects prepare to take a step forwards in order to initiate gait (Brenière et al. 1987; Nissan and Whittle 1990; Jian et al. 1993). It is often assumed that, in a manner analogous to moving into unipedal stance, this serves to bring the body into a position of stability over the single limb which will support it during the step. However Jian et al. (1993) report that during gait initiation, although the CoM of the body

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moves towards the support side, it does not do so sufficiently to bring it directly over the supporting foot. If correct, this has important implications for the way in which balance during a step is controlled. It implies that the body is being allowed to topple or fall over during the step and therefore that the extent to which the body motion can be altered once the step has begun is limited. Thus it seems unlikely that the wide range of body trajectories required for different types (e.g. direction, speed, length) of step could be produced wholly by adjusting the motion once the step has begun. The alternative is that broadly different trajectories are produced by changing the state (position and velocity) of the body at the start of the step. There is some indirect experimental evidence to support this idea. Patla et al. (1991) found that if walking subjects are instructed to change direction just before the beginning of a step, they are not able to do so during this step. They are, however, able to change direction if instructed to do so one step in advance. The authors concluded that a change of direction while walking needs to be planned in the previous step. Hollands et al. (1995) have shown that subjects, stepping over a series of irregularly placed "stepping stones", generally fixate the next stone to be acquired before the stepping foot has been lifted. Again this implies that the forthcoming step is being planned in advance.

This study looks for the possible existence of a "ballistic" strategy of control of body-motion in the frontal plane during a step. In particular it looks to see whether the lateral position and velocity to which subjects bring their body-mass at the start of the step is such that a more-or-less unconstrained fall during the step will take the body in the required direction. Such a strategy would keep to a minimum the need for potentially difficult mid-step adjustments. We have analysed the motion of subjects' CoM as they stepped in different directions and from different initial postures, and have used a mathematical model of the body falling freely under gravity about the ankle joint to help interpret the data. Part of these data have previously been published in brief form (Lyon and Day 1995).

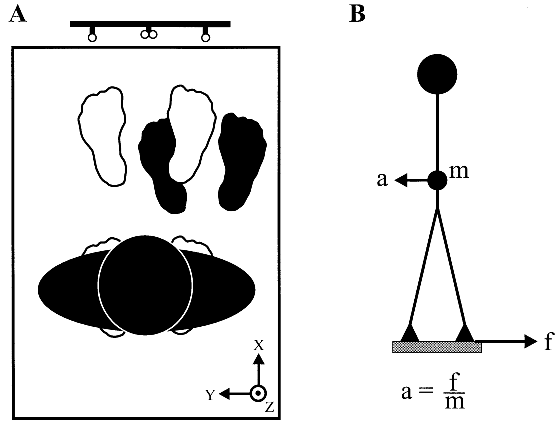

Fig. 1 A A view from above of the experimental set-up. Subjects stood on a force platform facing a row of four lights, illumination of which instructed them with which foot and in which direction to step. Approximate final positions of the feet for a movement forwards (*open feet*) and a movement diagonally beginning with the right foot (*shaded feet*) are shown. **B** For each dimension, the acceleration (*a*) of a subject's centre of mass (CoM) is calculated from the subject's mass (*m*) and the force (*f*) exerted on the platform

Materials and methods

Six normal subjects (four female and two male) with ages ranging from 23 to 36 (mean 28.5) years gave their informed consent to participate in the study, which was approved by the local ethics committee. Subjects stood barefoot on a large force platform (Kistler 9287) adopting one of two stance widths – narrow (intermalleolar distance 10 cm) or wide (20 cm) – and on an auditory cue stepped to a new position. Data collection for each trial began with the auditory cue and the illumination of one of four lights arranged in a row in front of the subject. These lights instructed the subject with which foot and in which direction to step. Illumination of one of the centre lights was an instruction to step forwards either with the right or left foot (right centre and left centre light respectively). Illumination of one of the outer lights was an instruction to step diagonally (forwards and out to the side) with the foot nearest to the light. These four conditions were presented to subjects in random order and therefore data collection began before subjects knew with which foot to start moving. This was to ensure that subjects did not stand with their weight more on one side than the other in preparation for a step with a "known" foot, so that all the preparatory movement normally necessary could be recorded. Data collection continued for 4 s after the signal to move. Subjects stepped in their own time and at their own speed. Each trial consisted of the initial step as instructed by the lights, followed by a step with the other foot to bring it alongside the first, such that subjects started and ended each trial in normal, quiet stance. Only the initial step (as instructed by the lights) is analysed in this study and all results and discussion refer to this. Typical final positions of the feet for movements forwards and diagonally beginning with the right foot are shown in Fig. 1A.

Subjects were asked to adopt the most comfortable starting positions of the feet (at the required intermalleolar distances) and these were marked by drawing around the feet with chalk. This enabled subjects to start all trials of a given stance width from the same position. Each subject's starting foot positions were captured onto computer by tracing around the chalk outlines with an infrared LED and following the path described with a Selspot system. Trials were grouped into four blocks of 16 and the initial stance width was alternated between blocks so that each subject performed two blocks of trials from each width, and a total of 64 trials. In order to record the precise time at which subjects' feet cleared and struck the surface of the force platform (at the start and end of steps respectively), the surface of the plate was divided into a number of isolated conductive areas. A small potential difference was applied between the subject and these areas which caused a current to flow only when a foot was in contact with the surface of the platform. The data from the force platform and the conductive areas on the platform were collected with a sampling frequency of 1000 Hz. Signal noise was reduced by averaging every five consecutive data points, which lowered the effective sampling frequency to 200 Hz.

Force platform calculations

We have adapted a previously described technique (Shimba 1984) in which data from a force platform are used to calculate the motion of the CoM of a body moving on the platform. The platform registers the force exerted on it independently in three orthogonal directions. Thus, using Newton's Second Law of Motion, the three-dimensional acceleration of the CoM of a body of known

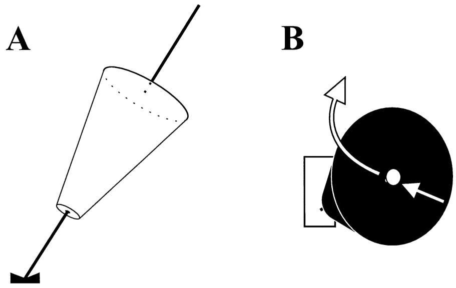

Fig. 2 A A model of the body supported on one leg during a step. It consists of a truncated cone which pivots on a base about fixed point. This point corresponds approximately to the subject's ankle (see Discussion). The starting position and velocity of the cone are set to the position and velocity of the subject's CoM at toe-off. Thereafter the cone falls freely under the influence of gravity. **B** A view from above of the motion of the cone (*curved arrow*) when given an initial velocity forwards and towards the support (*straight arrow*). This is representative of the direction of the initial velocity and type of motion observed in the present study. The *rectangle* represents the support foot

mass moving on the platform may be calculated (Fig. 1B). The acceleration records thus obtained may then be integrated numerically to obtain the velocity of the CoM, and then integrated again to obtain the displacement. This method of following whole-body CoM motion has some advantages over techniques which are based on a whole-body kinematic analysis. It does not rely on estimates of the mass and position of the COM of each of the major body segments, made using standard anthropometric data, which introduce errors that are difficult to quantify (Plagenhoef et al. 1983). Furthermore it does not require the time-varying positions of all the major body segments to be followed, which is technically difficult and prone to distortions. However, it does suffer from one drawback which is that very small errors in the force records are "amplified" by the double integration, giving rise to so-called integration drift. This can produce unacceptably large errors (Eng and Winter 1993). Our solution to this problem has been to design the experiment in such a way as to enable us to quantify and thus greatly reduce the error. The procedure is fully described in the Appendix.

Model

We have represented the body supported on one leg during a step with a truncated cone which pivots on a base about a fixed point (Fig. 2A). The use of a single-segment model such as this is suggested by the finding that during a step the body moves largely as a single unit (MacKinnon and Winter 1993; Jian et al. 1993). The cone falls freely about the fixed pivot under the influence of gravity except that it is constrained to have zero angular velocity about its long axis. For a structure such as this which is symmetrical about its principal long axis this means that the angular velocity and angular momentum vectors are always parallel. As a result the equation of motion simplifies to

$$\dot{\mathbf{J}} = I_1 \dot{\omega} = \mathbf{R} \times m\mathbf{g}$$

where **J** is the angular momentum, I_1 is the moment of inertia (short axes) about the pivot, ω is the angular velocity, **R** is the position of the CoM, m is the mass, and **g** is the gravitational acceleration. The model was given a subject's mass and moment of inertia about the ankle. This latter value was estimated using standard anthropometric data (Plagenhoef et al. 1983). Then, for each trial, the model was given as initial conditions the position and velocity of the subject's CoM at the start (toe-off) of each step (as calculated from the force platform data). The subsequent motion of the model's CoM as it fell about the pivot under the influence of gravity was then predicted by solving numerically the above equation of motion using a second-order Runge-Kutta algorithm. This motion was compared with that of the subject's CoM for each trial. This process was repeated for all subjects. Figure 2B shows diagrammatically a view from above of the sort of motion which resulted from a typical starting position and velocity.

The model was also used to investigate the importance of the lateral velocity (or momentum) of subjects' body-mass at toe-off to the subsequent motion of the body during the step. The process described above was repeated but this time the model's initial lateral velocity was set to zero. Thus it predicted how the subject's body would move if all conditions at the start of the step were as before except that the body now had zero lateral momentum.

Results

In all measurements made, no significant difference was found between right and left foot steps and so the data have been pooled across the levels of this factor. For ease of measurement and presentation left foot steps have been reflected so that all steps appear and may be analysed as right foot steps. Four experimental conditions remain made up of two initial stance widths (narrow and wide) and two step directions (forwards and diagonal). There was no significant interaction between these two factors. Although data collection and modelling were in three dimensions, the present analysis focuses almost exclusively on the motion in the frontal plane.

Figure 3 is a plan view of the paths described by the CoM and centre of pressure (CoP) with respect to the initial position of the feet. One trial of each of the four conditions is shown, all taken from a single subject. As discussed in the Appendix, the CoM was assumed to start at the same position in the horizontal plane as the CoP (forward of the ankles and midway between the feet). The CoP (dotted line) moves first towards the forthcoming stepping (right) foot and a little backwards. It then reverses direction and moves across under the support (left) foot. During the step it moves forwards under the support foot. The path of the CoP is not shown beyond the end (heel-strike) of the right foot step. The CoM (continuous line) initially moves towards the support foot and forwards. The thickened part of the line indicates when the right foot was not in contact with the

Table 1 Percentage of the total number of trials in each condition in which the centre of mass was on or over a line defining the medial border of the support foot (1) at toe-off, and (2) at any point during the step

	Narrow		Wide	
	Forwards (%)	Diagonally (%)	Forwards (%)	Diagonally (%)
At toe-off	36	3.3	1.2	0
At any point during step	52	11.5	5.2	2.1

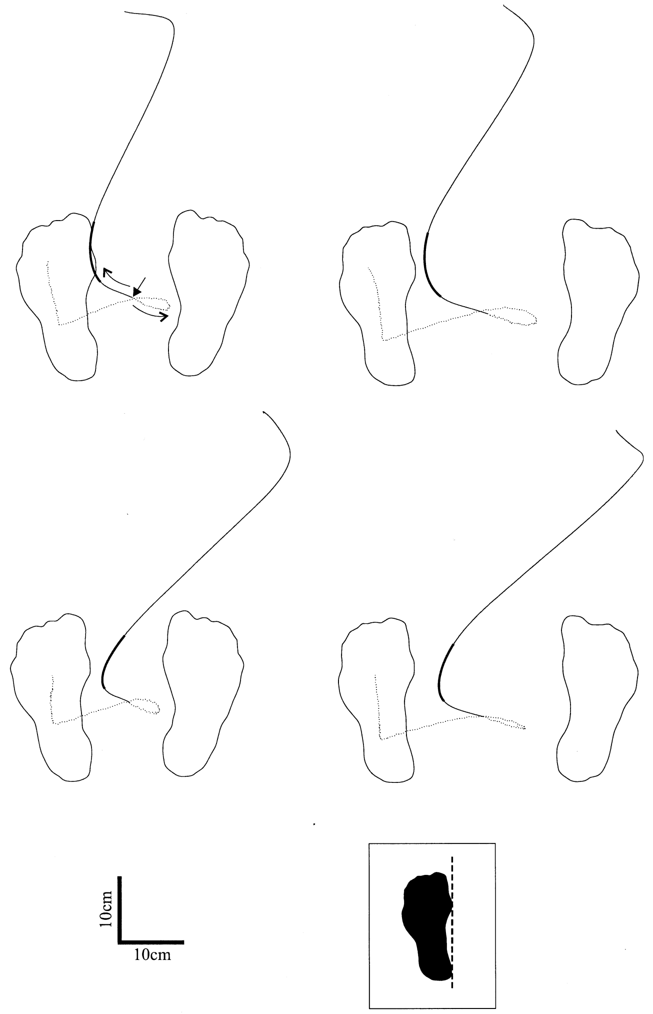

Fig. 3 A plan view of the paths described by the CoM (*continuous line*) and centre of pressure (CoP; *broken line*) with respect to the initial position of the feet for four trials. The *upper row* shows movements forwards, the *lower row* movements diagonally (forwards and to the right). The *left-hand column* shows movements from the narrow stance width, the *right-hand column* shows movements from the wide stance width. The CoM and CoP both start in the same position, forwards of the ankles and midway between the feet. This position is indicated by the *straight arrow* in the top left diagram; the *curved arrows* show the initial directions of CoM and CoP movement. The *thickened part* of the CoM line shows when the right foot was not in contact with the platform. The path of the CoP is shown only up to the right foot heel-strike. The *inset* shows the medial border of the support surface defined as a line joining the most medial points of the support foot

ground. At the moment the right foot clears the ground (toe-off), which is indicated by the transition from the thin to the thickened portion of the line, the CoM is not directly above the support foot in any of these trials. During the step the CoM moves forwards and initially continues to move towards the support side before then starting to move away. Only in the case of the step forwards from the narrow stance width (top left) does this extra displacement towards the support side bring the CoM over the support foot at any point during the step. We defined a medial border of the support surface (inset Fig. 3) and counted the number of trials in which the CoM was on or over this border (1) at the start of and (2) at any time during the step. The results are shown in Table 1. Only in steps forwards from a narrow stance width did either of these events occur in a high proportion of trials. In the vast majority of trials the CoM was not over the supporting surface at any time during the step.

Figure 4 depicts the same four trials but now showing the lateral component of motion of the CoM plotted against time. The traces are aligned to the initial (right foot) toe-off, shown by the vertical lines. Initially the CoM is accelerated towards the support (left) side. As a result, by toe-off it is displaced and has velocity towards this side. This lateral displacement and velocity at toe-off is greater when stepping forwards (continuous lines) than when stepping diagonally (broken lines). It is also greater when stepping from a wide (right-hand column) than from a narrow (left-hand column) stance width. We measured the lateral displacement and velocity of all subjects' CoM at toe-off in all trials and found these same trends to be present in general. The results are shown in Fig. 5. ANOVA with repeated measures showed both effects to be highly significant for both velocity and displacement. For velocity, $F(5,1) = 262.61$, $P < 0.001$ (step direction) and $F(5,1) = 187.32$, $P < 0.001$ (stance width), and for displacement, $F(5,1) = 152.64$, $P < 0.001$ (step direction) and $F(5,1) = 145.07$, $P < 0.001$ (stance width).

Model

In general the model was able to predict the motion of the CoM during the step with reasonable accuracy. Figure 6A shows the lateral motion of the CoM and the prediction of the model for a single trial from a typical subject. The model is set going at toe-off (first vertical line) and falls freely under gravity until heel-strike (second vertical line). When given the position and velocity of the subject's CoM at toe-off as initial conditions (upper dotted line), it predicts a trajectory of the CoM close to that observed in the subject. Also shown is the model's prediction when given the position and forwards velocity of the subject's CoM at toe-off as initial conditions but with the initial lateral velocity set to zero (lower dotted line). In this case it predicts that the mass falls away from the support side much more rapidly. We suggest that the difference between these two predictions gives an indication of the contribution of the velocity (or momentum) of the subject's CoM at toe-off to its subsequent motion during this step.

The performance of the model across all trials from each subject was assessed by measuring the displacement of the subject's CoM during the step and the displacement predicted by the model (Fig. 6B). Then, for each trial, the actual displacement was plotted against the model's prediction. Figure 6C (upper graph) shows the results from a typical subject. The filled symbols

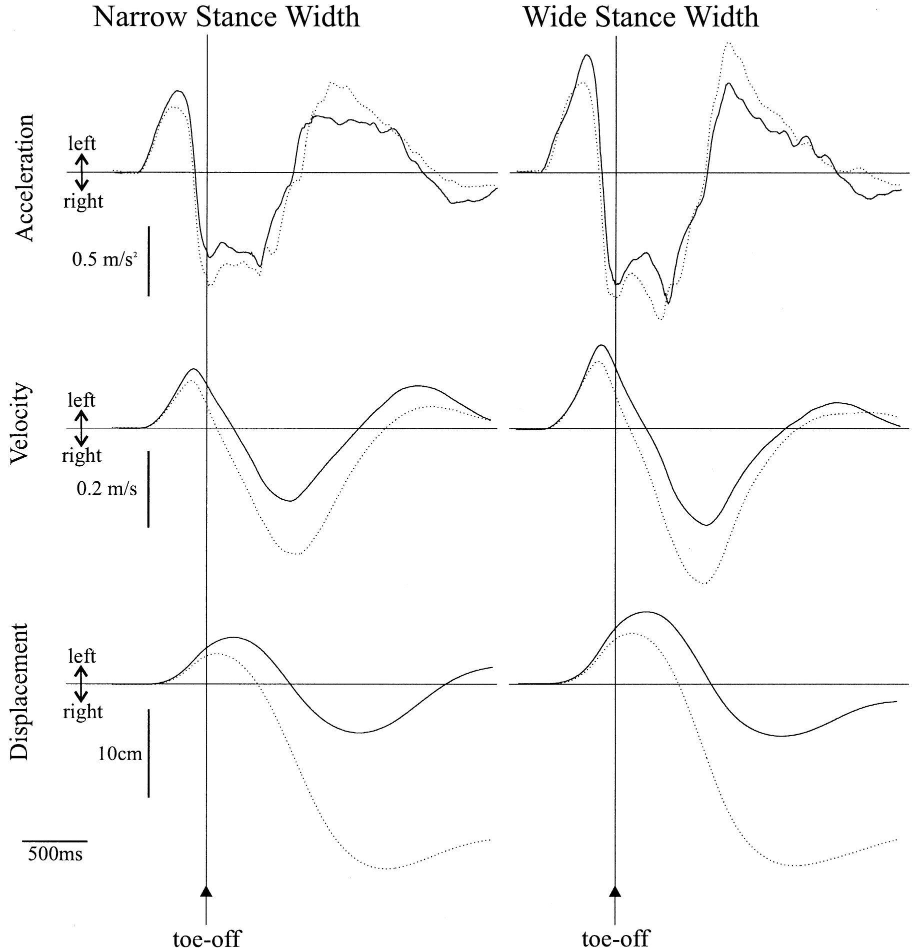

Fig. 4 Details of the lateral component of motion of the CoM for the same four trials depicted in Fig. 3. Approximately the first 3 s of each trial are shown. The traces are aligned to the initial toe-off shown by the *vertical lines*. Heel-strike occurred some 400–500 ms later. The *continuous traces* are forwards movements, the *broken traces* are diagonal. The *left-hand column* shows movements from the narrow stance width, the *right-hand column* shows movements from the wide stance width

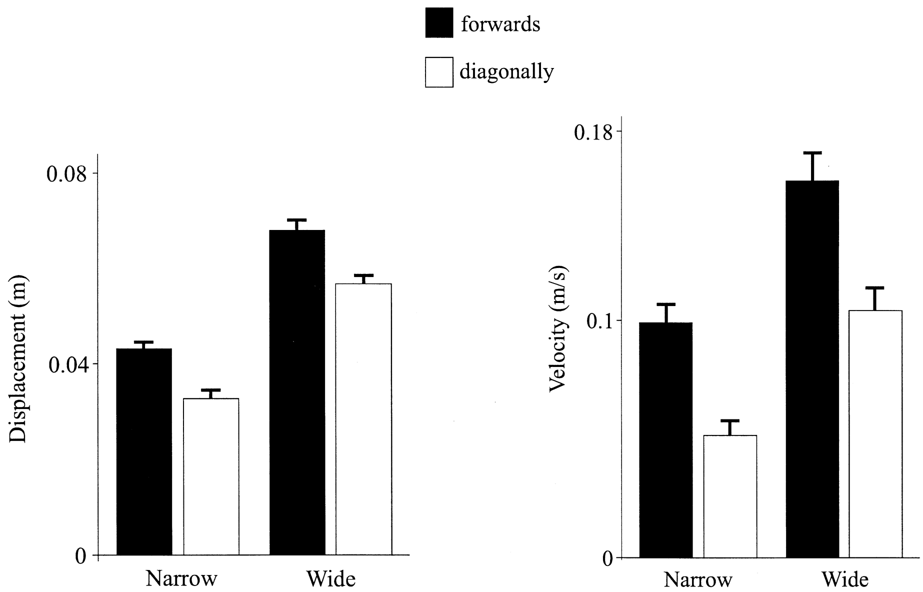

Fig. 5 Group data for the displacement and velocity of the CoM at toe-off (mean + SEM). The *filled bars* show forwards movements, the *open bars* show diagonal. The effects of step direction and stance width are significant ($P < 0.001$) for both displacement and velocity (ANOVA with repeated measures)

show the result obtained when the model was given all initial (toe-off) conditions. The regression line has a slope close to unity (0.94), indicating the existence of a relationship between predicted and actual displacement. Further, the points appear to be clustered about the regression line, and this is reflected in the high adjusted r^2 value of 0.93. This shows that most of the variation in the displacement of the subject's CoM during the step was accounted for by the model. This same pattern was found in all subjects. Because variation in the model's

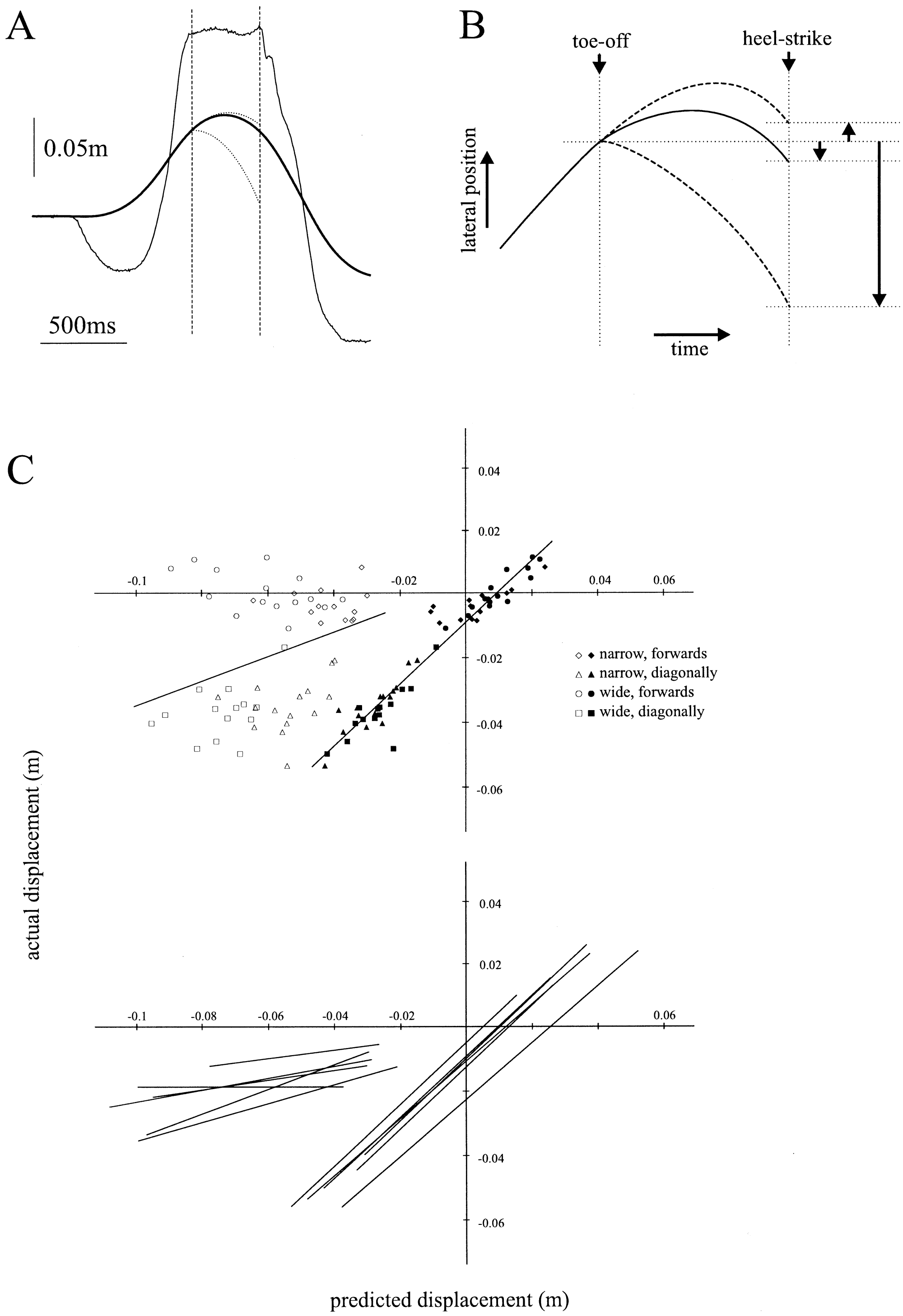

Fig. 6 A The lateral displacement of the CoM (*bold continuous line*) and the CoP (*thin continuous line*) during the first 2 s of a typical single trial. The *vertical dashed lines* mark the start (toe-off) and end (heel-strike) of the step. Also shown are two predictions of the CoM displacement during the step (*dotted lines*). The *upper* of these two lines is the prediction made when the model started with the same position and velocity as the subject, the *lower* when its initial lateral velocity was set to zero. **B** The method of measurement of the displacement of the CoM (actual and predicted) during the step. The sign convention is that towards the support side (upwards) is positive. **C** The *upper figure* shows the actual plotted against the predicted displacement for all trials from a typical subject. The various types of step (different stance widths and directions) are shown by the various *symbols*, explained by the key. *Filled symbols* show the result when the model started with the same position and velocity as the subject at toe-off, *open symbols* when its initial lateral velocity was set to zero. The *lower figure* shows the regression lines for all subjects

Table 2 Parameters of the regression of actual displacement against predicted displacement when the model was given (1) all initial conditions, and (2) all initial conditions except that its initial medio-lateral (*M–L*) velocity was set to zero. Because of non-normal distribution of data, inferential statistics are not given (Hays 1988)

	All initial conditions			Zero initial M-L velocity		
Subjects	Gradient	Intercept (m)	Adj. r^2	Gradient	Intercept (m)	Adj. r^2
1	0.95	–0.0051	0.94	0.29	–0.0062	0.05
2	0.88	–0.0108	0.94	0.01	–0.0185	–0.02
3	0.94	–0.0093	0.93	0.38	0.0035	0.09
4	0.96	–0.0100	0.96	0.13	–0.0020	–0.01
5	0.95	–0.0126	0.88	0.15	–0.0076	0.00
6	0.88	–0.0226	0.91	0.18	–0.0049	0.01

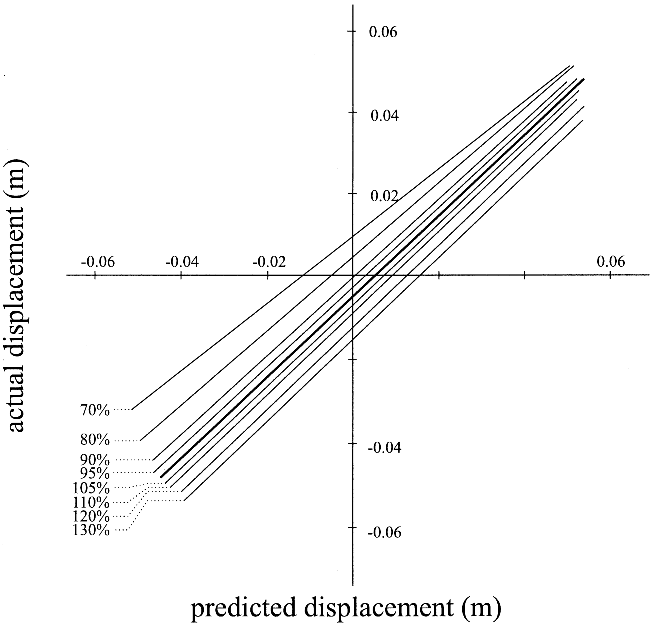

Fig. 7 Regression lines of actual against predicted displacement for one subject (BF). The *bold line* shows the result obtained with the original estimate of the subject's moment of inertia. The *other lines* show the results obtained with greater and smaller estimates, shown as percentages of the original estimate

behaviour is due entirely to changes in the initial (toe-off) conditions, these findings suggest that the displacement of a subject's CoM during a step is largely predicted by the position and velocity of the CoM at the start of it. The open symbols show the result obtained when the model was not given the lateral velocity of the subject's CoM at toe-off, but instead had its initial lateral velocity set to zero. The regression line has a slope of 0.38 and the points appear to be scattered widely about the line, reflected in the very low adjusted r^2 value of 0.092. Thus here, in sharp contrast to the previous result, the regression does not seem to describe the data in any meaningful way. The model seems now to account for virtually none of the variation in the displacement of the subject's CoM. This same pattern was found in all subjects. Regression lines for all subjects are shown in the lower graph of Fig. 6C. The group of lines to the left show the results when the model was given zero initial lateral velocity, those to the right when it was given all of the subjects' initial (toe-off) conditions. The regression parameters for all subjects are given in Table 2.

Table 3 Regression parameters (actual against predicted displacement) for various values of moment of inertia (MoI) expressed as percentages of original value. Data are for subject 1

Value of MoI (% of original value)	Gradient	Intercept (m)	Adj. r^2
70	0.79	0.0092	0.93
80	0.87	0.0042	0.96
90	0.92	–0.0006	0.96
95	0.94	–0.0029	0.95
100	**0.95**	**–0.0051**	**0.94**
105	0.96	–0.0071	0.93
110	0.96	–0.0090	0.92
120	0.96	–0.0124	0.89
130	0.96	–0.0152	0.86

In a plot such as that in Fig. 6C of actual against predicted behaviour, all the points lying on a regression line having unity gradient and zero intercept would indicate that the model perfectly predicted the behaviour of the real physical system. As shown in Table 2 the gradients found were all slightly less than but generally very close to unity, and the intercepts were all negative and on average approximately 1 cm. Thus across all subjects and trials the model tended to predict a slightly more positive displacement (<1 cm) during the step than that actually observed (for example as in Fig. 6A). The source of this error is not known. One possibility is that it is some form of measurement error, for example in determining the position of the CoP (Bobbert and Schamhardt 1990). Another possibility is that subjects were actively assisting the fall either by producing torques at the ankle or/and by accelerating one body segment on another. A further possible source of the error is an incorrect estimation of the subjects' inertial properties. These are estimated using subjects' height and weight and standard anthropometric data (Plagenhoef et al. 1983). Using such data introduces error due, for example, to the fact that no subject has the same body build and shape as those from whom the data were obtained. Plagenhoef et al. estimate that an error of between 10% and 15% is to be expected. To investigate the effect of error of this sort on the results obtained we ran the model on one subject's data using a range of values for the moment of inertia. The results are shown in Fig. 7 and Table 3. The model's behaviour is found to be sensitive to small variations in the moment of inertia. Reducing the value to 90% of the original estimate reduces the intercept almost to zero while leaving the gradient virtually unchanged. It also produces a small increase in the r^2 value. This indicates that inaccuracies (here an over-estimation) in the moment of inertia estimate could produce the modelling errors observed.

Discussion

This study begins an investigation into the control of whole-body motion during a single step by looking for the existence of a "ballistic" strategy of control in the frontal plane. In such a strategy the sideways fall which occurs during the step would be allowed to develop freely and the motion of the body-mass during the step would be controlled by setting the position and velocity of the CoM at the start of the step.

Position of the COM with respect to the base of support

As outlined in the Introduction, it has previously been noted that, during the first step of gait initiation and the single support phases of steady-state gait, the CoM of the body is not directly over the base of support. We have found that this is also generally the case when subjects step to a new position (and then come to a halt), either directly or diagonally in front (Fig. 3, Table 1). Only when stepping forwards from a narrow stance width did subjects' CoM cross the medial border of the support surface in a significant proportion (52%) of trials. When this did occur, the CoM typically just "clipped" the medial border of the foot (Fig. 3, top left), and we suggest that it is unlikely that a position in which the body was stable over the single supporting limb was attained. Thus, in keeping with earlier studies (Shimba 1984; Mackinnon and Winter 1993; Jian et al. 1993) we find that in general the body does not achieve a position of stability over the single supporting limb during a step.

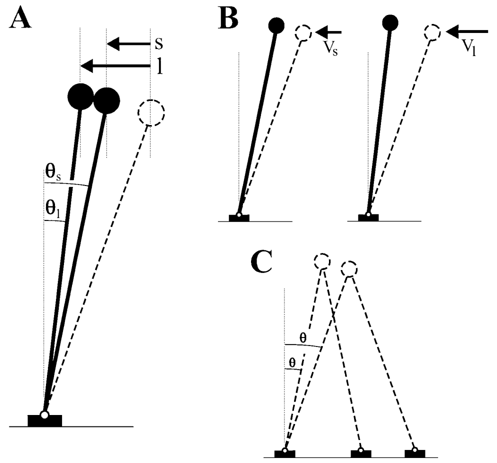

Fig. 8 **A** A frontal plane representation of the body supported on one limb at toe-off (starting position shown by *broken line figure*). A smaller preparatory lateral displacement (*s*) leads to a larger angle with the vertical at toe-off (θ_s) than a larger preparatory lateral displacement (l and θ_l). **B** A smaller lateral velocity at toe-off (V_s) leads to a smaller maximum displacement towards the support than a larger lateral velocity (V_l). **C** Assuming no preparatory lateral motion, increasing the stance width increases the angle of the body with the vertical at toe-off

Displacement and velocity at toe-off

Again in keeping with earlier studies (Brenière et al. 1987; Nissan and Whittle 1990; Jian et al. 1993) we have found that subjects accelerated themselves towards the support side before lifting the stepping foot. In all trials this resulted in the CoM being displaced and having velocity in this direction at toe-off. Both lateral displacement and velocity were greater in forwards than in diagonal steps, and when stepping from the wide compared with the narrow stance width (Fig. 5).

These findings are consistent with the notion that during the step the body falls freely and that therefore its trajectory is determined at the start of the step by setting appropriate initial conditions. For example, a diagonal step requires that the body ends up displaced laterally, whereas a forwards step does not. This means that the sideways fall during the step must develop more in a diagonal than in a forwards step. This could be achieved by reducing the preparatory lateral displacement which would increase the angle of the body with the vertical at toe-off (Fig. 8A). The medial acceleration of the CoM due to gravity also would be increased as it is proportional to the cosine multiplied by the sine of this angle and the angles involved are small. Thus the rate of the sideways fall is increased, causing it to develop more in a given time. Alternatively the fall during the step could be allowed to develop more by reducing the lateral velocity of the CoM at toe-off, which would amount to "throwing" the body mass less forcefully towards the support side (Fig. 8B). This would reduce the maximum displacement towards this side (attained during the step) and increase the overall rate of fall away from it. Both these differences were observed when comparing diagonal with forwards steps. The stance width effect may be considered in the same way. Assuming for a moment no preparatory motion, increasing the stance width increases the angle with the vertical of the body at toe-off (Fig. 8C). This therefore increases the medial acceleration of the CoM during the step. Thus for a given step direction (and therefore CoM trajectory) either an increase in the preparatory displacement or/and velocity is predicted when comparing steps from the wide with those from the narrow stance width. Again, both of these differences were observed.

Model

We sought to investigate to what extent the body might be falling freely during a step by comparing the behaviour of subjects with that of a mathematical model of a freely-falling structure. The model is three-dimensional but only the medio-lateral component of its motion is relevant and considered here. The different step directions and starting postures used in this study provide the model with a wide range of initial positions and velocities and, therefore, taken together, constitute a more rigorous test than would any one of the types of step individually. The model is found to account for most of the variation (across all steps and step-types) in displacement of subjects' CoM during a step. This lends preliminary support to the idea that subjects bring their body-mass to a position and velocity at the start of the step such that a more-or-less unconstrained fall during the step produces the particular body-motion required. There is some error in the model's prediction. In general it appears slightly to underestimate the rate of the sideways fall occurring during the step. This may be due to measurement/estimation errors (see Results) and/or to some violation of one or more of the assumptions of the model. For example the fall during the step may be actively assisted/retarded, the body may not be adequately described as a single rigid unit, or the pivot (ankle) may move significantly. Further insight into the source and nature of this error must await more accurate methods of determining subjects' inertial properties as well as more complex modelling.

When the model's initial lateral velocity was set to zero, rather than to that of the subject's CoM at toe-off, a strongly contrasting picture was produced (Fig. 6, Table 2) in which the model predicted that the CoM fell much more rapidly than it actually did, and in which very little of the variation (across all trials) in the displacement during the step was accounted for. The first of these findings suggests that, without the initial momentum towards the support side, the body would keel over sideways very rapidly as soon as the stepping foot was lifted. This fall would have to be arrested by quickly placing the stepping foot back down on the floor more laterally. In effect a quick and uncontrolled sideways step would have been taken. Thus, for this purely mechanical reason, we suggest that the initial sideways momentum is crucial to the execution of a controlled forwards step. In view of the apparent importance of this sideways momentum at toe-off, we suggest that the preparatory lateral movement is well described as a "throw" of the body-mass towards the support side. These findings also emphasise just how critical is the instability of the body at the start of a step. Although the horizontal position of the CoM may not be far outside the base of support, the body falls sideways rapidly unless it already has substantial momentum towards the support.

Previous ballistic models of gait and gait initiation

The idea that the motion of the body during a step results from a fall, and that this fall is modified by the body's state at the start of the step is not new. As Roberts (1978) puts it, "in locomotion . . . the mass of the body is alternately thrown and caught again at each step; thrown upward and forward at take off, and caught again on landing" (p. 145). Mochon and McMahon (1980) developed the simple inverted pendulum ballistic model of motion during the swing phase by adding a double pendulum to represent the swing limb. Their model was able to predict with reasonable accuracy a number of published experimental observations of normal gait. Another group has emphasised how motion at the end of the first step of gait initiation depends on that at the start (Brenière et al. 1987; Brenière and Do 1991). However, all these studies have restricted their analysis to the sagittal plane, presumably because they have been concerned mostly with the means and control of forward progression during locomotion. Here we are concerned with the control of balance during a step and have focused on motion in the frontal plane because the change in the conditions of support that occurs when one foot is lifted off the floor is predominantly in a medio-lateral direction.

Implications of the present findings for the control of frontal plane body motion in steady-state gait

As pointed out above the modelling presented here emphasises the importance of the lateral velocity at toe-off (which is always directed towards the support side) to the subsequent motion of the body during the step. The lateral velocity of the CoM at toe-off may be similarly important in steady-state gait. MacKinnon and Winter (1993) report a figure of approximately 0.1 m/s directed towards the support side. This is very close to our figure for steps forwards from the narrow stance width (Fig. 5), which, of the types of step we have studied, are the most similar to those taken during normal gait. If the lateral velocity at toe-off is crucial to the maintenance of balance during steady-state gait, it becomes important to know how it is produced and controlled. In taking an initial step from quiet stance (as studied here), the body-mass appears to be accelerated up to its toe-off lateral velocity by the action of the forthcoming stepping limb, and by a slight flexing (and thus withdrawal of the support) of the forthcoming stance limb (Nissan and Whittle 1990; Brunt et al. 1991). Control of the lateral velocity of the CoM at toe-off in steady-state gait may be achieved in an analogous way. During double support, a thrust delivered by the rear leg before it begins its swing accelerates the body forwards and upwards into the next swing phase (Elftman 1939; Bernstein 1967). This thrust will also have a medio-lateral component the size of which could be varied by adjusting the exact direction and size of the thrust by precise control of the torques about the various joints of the leg (van Ingen Schenau et al. 1992). Furthermore it is well established that during double support the knee of the leg in front (forthcoming support limb) flexes by between 10° and 15° (Rose and Gamble 1994).

However, Townsend (1985) has suggested an alternative means by which frontal plane motion of the body during steady-state gait could be controlled, which is by the setting of medio-lateral foot placement. The more laterally the foot of the swinging leg is placed down in front, the greater the angular acceleration of the body during the next step (in which this foot acts as the support). This "foot placement" strategy is clearly not available to subjects taking an initial single step, as in this study. However, in steady-state gait it is a possibility and represents another means by which lateral velocity at toe-off may be controlled.

Further work is needed to elucidate which of these strategies predominates in the control of frontal plane whole-body motion during gait. One situation which may produce problems for the "foot placement" hypothesis is that where there is little or no choice about foot position, as when walking on stepping stones. Interestingly, a recent study which looked at subjects' eye movements as they walked over a series of irregularly placed stepping stones found that the next target in the series was generally fixated just before the foot which was to be placed on it had been lifted (Hollands et al. 1995).

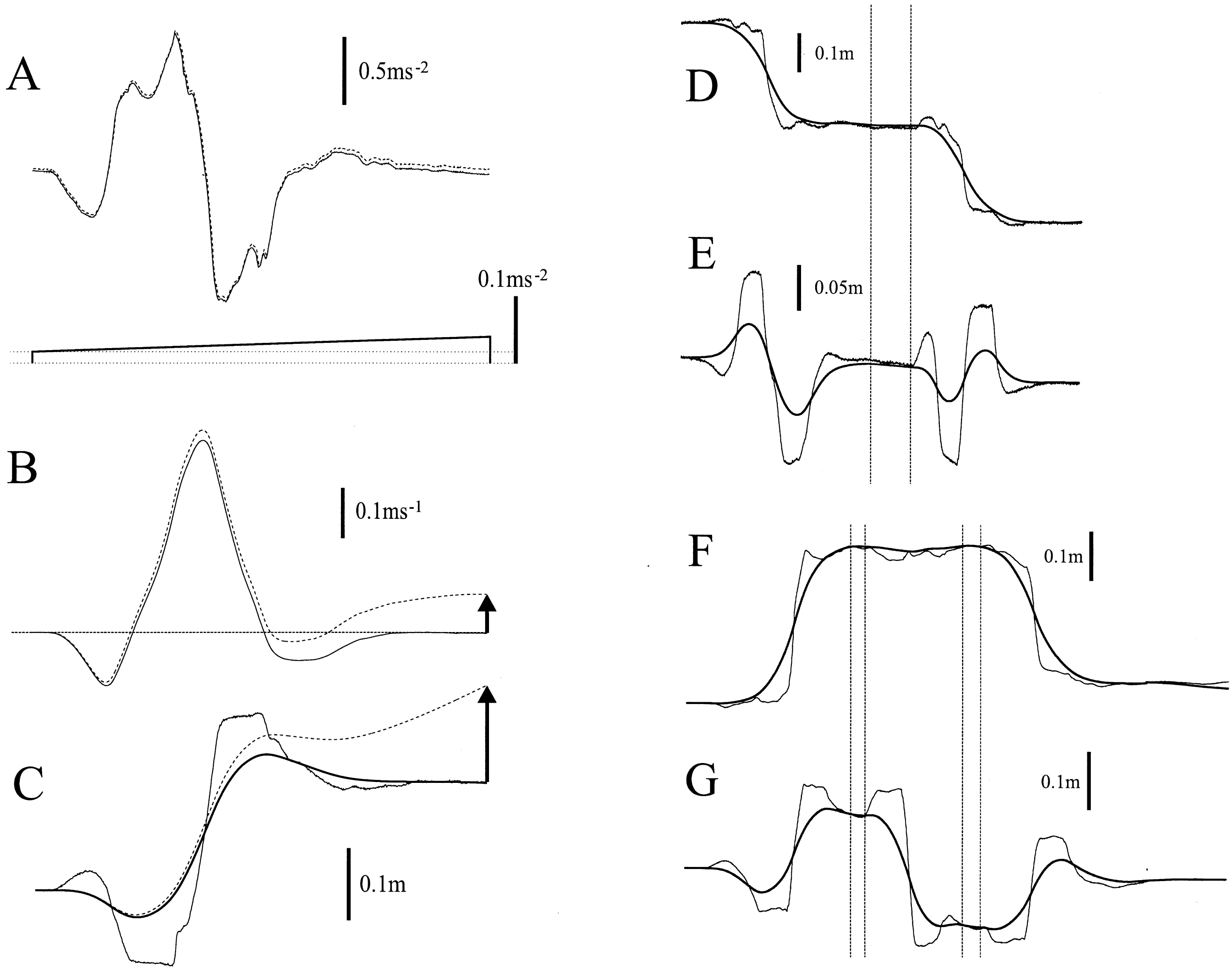

Fig. 9 **A** (*upper trace*) The *dotted line* shows a typical record of the medio-lateral acceleration of a subject's CoM. **B**, **C** *Dotted lines* show the medio-lateral velocity and displacement obtained from the acceleration record by numerical integration. The *light continuous line* in **C** shows the medio-lateral position of the CoP. The *vertical arrows* show respectively the velocity and displacement errors. These are used to calculate the offset and slope of a correction ramp (shown in large scale in the *lower trace* of **A**) which is subtracted from the acceleration record. The *continuous lines* in **A** and **B** and the *bold line* in **C** show the acceleration, velocity and displacement following correction of the acceleration record and subsequent integration. **A**–**C** all show a time period of 4 s. **D**, **E** The antero-posterior and medio-lateral position of the CoM (*bold line*) and CoP (*light line*), respectively, plotted against time for a single trial (8 s duration) in which a subject stepped forwards, came to a standstill, and then stepped forwards again. The position of the CoM is as calculated following correction of the acceleration traces (not shown). The intermediate period of stillness deduced from the force traces (not shown) is indicated by the area between the *vertical lines*. During this period, the positions in the horizontal plane of the CoM and CoP are very similar. **F**, **G** (subjects AP and ML respectively) show another similar trial, this time 10 s in duration. Here the subject steps three times, moving in a triangular path and ending up at approximately the starting position. The two periods of stillness are marked by the *vertical lines*

This finding is in keeping with the idea that the body motion during the step is being prepared and determined just before the swing foot is lifted, as predicted by the "thrust" hypothesis.

In conclusion we suggest that our hypothesis, that frontal plane body motion during stepping is controlled ballistically, accounts well for the experimental findings, although it is acknowledged that more experiments are required to test the hypothesis further. The use of a ballistic strategy would imply that the central nervous system is able to judge in advance the size and direction of the initial throw given to the body-mass. Although subjects appear not to adjust their body motion appreciably during a step, it remains to be determined whether this is because they cannot, or simply because (normally) they need not. Finally, we suggest that this work may have some clinical relevance, in that difficulty with a predictive mode of control may underlie some of the problems with walking experienced by neurological patients.

Appendix

Subjects started and ended each trial in normal, quiet stance. They were asked to try to stand as still as possible during these periods and examination of the force records showed that they were able to comply well with this instruction. This enabled us to make the approximation that the CoM started and ended each trial with zero velocity. Thus the initial velocity was set to zero and the change in velocity during the trial calculated by integrating the acceleration trace. This procedure inevitably results in a non-zero final velocity due to error in the acceleration trace (Fig. 9B). This error is assumed to consist of an offset and a superimposed drift. Thus we can write

$$\int_{t=0}^{t=T} \left[a(t) - (mt + c) \right] \mathrm{d}t = 0 \tag{1}$$

where $a(t)$ is the acceleration with time, T is the duration of the trial and m and c are the slope and offset of a correction ramp which is subtracted from the acceleration trace.

When a body is static (and no horizontal forces are exerted on it), its CoM is vertically above the point of application of the resultant of the ground reaction (CoP). Thus we were also able to make the approximation that the position of subjects' CoM in the horizontal plane was given by the position of the CoP during the initial and final still periods. In practice, because the CoP of a subject standing still oscillates around a mean position and only momentarily comes completely to rest (Murray et al. 1975), we used the mean position of the CoP over the initial and final 100 ms. Thus the initial position of the CoM was set to that of the CoP and the change in the CoM's position during the trial calculated by integrating the velocity trace. This results in a final position for the CoM different from that of the CoP due to error in the acceleration trace (Fig. 9C). Thus we can write

$$\int \int_{t=0}^{t=T} \left[a(t) - (mt + c) \right] \mathrm{d}t = \Delta\mathrm{CoP} \tag{2}$$

where ΔCoP is the change of position of the CoP during the trial. Re-arranging (1) and (2) and integrating we obtain

$$\int_{t=0}^{t=T} a(t)\,\mathrm{d}t = \frac{1}{2} mT^2 + cT \tag{3}$$

$$\int \int_{t=0}^{t=T} a(t)\,\mathrm{d}t = \frac{1}{6} mT^3 + \frac{1}{2} cT^2 + \Delta\mathrm{CoP} \tag{4}$$

The parameters of the correction ramp are obtained by solving (3) and (4) for m and c. This process is then repeated for the other (horizontal plane) dimension.

Correcting the acceleration traces in this manner assumes that the error consists of an offset and a superimposed linear drift. This assumption is based on the two known sources of error. The first of these is due to crosstalk in the force transducers. The presence of crosstalk means that if a vertical force is exerted on the platform, in addition to registering this, it also indicates that a small horizontal force is present. The vertical forces in this experiment are much larger than the horizontal and are dominated by the subject's weight, which is constant. Thus an approximately constant but fictitious horizontal force is indicated throughout each trial giving rise to an offset in the horizontal acceleration traces. The second source of error arises from the tendency of the piezo transducers and/or the charge amplifiers (used to convert the output of the transducers into a time-varying voltage) to drift. This gives rise to drift superimposed onto the acceleration records.

However, other sources of error showing more complex characteristics could presumably also be present and could contribute to the "integration drift" observed. We have therefore conducted a number of tests of this method of correcting the acceleration traces and these have shown that it can produce very credible records of CoM displacement in a variety of situations. For example Fig. 9D and E show a test in which a subject on the force platform stepped forwards and came briefly to a standstill before stepping forwards once more. The displacement of the CoM was calculated from acceleration traces corrected as described using the initial and final periods of stillness. It can be seen that during the intermediate period of stillness the positions of the CoM and CoP approximately coincide. This suggests that the correction has led to a reasonably accurate record of CoM position not only at the start and end of the trial, but throughout. This can be seen also to be the case in Fig. 9F and G which show CoM and CoP records from a trial in which a subject stepped along a triangular path, stopping twice and ending up at approximately the starting position. Both these trials were of considerably longer duration than the experimental trials in the main body of this study (8 and 10 s as opposed to 4 s). Because of the double integration procedure, the final error in the position trace before correction can become very large over such a long period of time. In these cases the correction required is large and generally not perfect (as judged by the correspondence between CoM and CoP positions during periods of stillness). However, in this study the trials were short and the latest event at which a measurement was made was at the heel strike of the foot which made the initial step. This event typically occurred at less than 1.5 s into the trial. As can be seen from Fig. 9C, the size of the correction made at this time (about one third of the way along the trace) is small, and this was the case in all trials. We suggest, therefore, that any error in the correction would have a negligible effect on the present results.

In addition, we have also tried fitting higher-order functions (quadratic and cubic) to the acceleration error and tested the results in the same manner as described above. However, we have found that on the whole this does not produce a better error correction than the linear function. We suggest that, together with the good performance of the linear function, this implies that most of the error in the acceleration records is simple in form.

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