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Displacement of the pelvis during human walking: experimental data and model predictions

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Abstract

Displacements of the pelvis during treadmill walking were studied in dependence of walking speed, stride frequency and stride length. Displacement curves per stride cycle were described by means of harmonic analysis. Simple mechanical, or geometrical models of the body's center of mass (COM) trajectory during walking were used to predict amplitude and timing of pelvic displacements. As predicted by inverted pendulum models, the amplitude of pelvic displacement in vertical direction depended on stride length. Anterior–posterior displacements were predicted by assuming equal maxima of potential and forward kinetic energy of the body. Predictions for left–right displacements were based upon a model that assumed a constant stride width and a sinusoidal movement pattern. In agreement with this model, the amplitude of pelvic displacement in left–right direction depended on stride frequency. The presented models give insight into mechanical mechanisms which determine the pelvic trajectory during walking. The presented data may be useful in the clinical evaluation of gait disturbances. © 1997 Elsevier Science B.V.

Keywords: Locomotion; Motor control; Modeling

1. Introduction

The general goal of any form of animal locomotion can be defined as the displacement of the body's center of mass (COM). In terrestrial locomotion this is achieved by limb movements generating propulsive forces which result in a displacement of the trunk [1]. In human bipedal locomotion any trunk displacement is achieved by means of lower limb movements which primarily result in a forward displacement of the pelvis which is the supportive structure for the head, arm, and trunk (HAT) segments. During walking, the pelvis moves forward and downward with each step, and it subsequently moves upward when body weight is transferred from one leg to the other. Thus, with each step, the pelvis moves up and down and, relative to the mean forward velocity, it displays speed fluctuations in the direction of progression [2]. The weight transfer from one supporting leg to the other also coincides with cyclic left–right movements of the pelvis. Due to an active stabilization of the HAT [3,4], the movement pattern of the HAT largely follows that of the pelvis [5,6]. The basic pattern of displacements of both pelvis and HAT, thus is a movement pattern which, during one stride, displays two oscillations along the vertical axis, two oscillations in anterior–posterior direction, and one oscillation in left–right direction.

It is well known that shape and amplitude of the pelvic and trunk displacements vary with speed of locomotion [5–7], but as the above discussion already indicates, these variations may ultimately be determined by step length and stride frequency. When the walking pattern of an individual is studied, spontaneously chosen combinations of stride length and frequency are observed which usually have a high degree of re-

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peatability [2,8]. However, humans are capable of voluntary adaptation of stepping movements (e.g. [8,9]), and in case of pathology, deviations from the normally observed combinations of stride length and frequency in healthy subjects may be seen as a result of the specific pathologic condition. It is very likely that such deviations will result in changes of the normal displacement pattern of the pelvis and HAT. Thus, although displacements of the pelvis, trunk and head have been studied with varying speeds of locomotion, it may be more appropriate to study them in their dependency of both stride length and stride frequency.

In order to get more insight into the mechanisms which determine the trajectory of pelvis and HAT during walking, the present study was directed at an analysis of pelvic displacement during walking. As the COM lies within the pelvis (in anatomical position the COM is close to a position in front of the second sacral vertebra), models of the COM trajectory during walking can possibly be used to predict patterns of pelvic displacement. It will therefore be attempted to make predictions of the amplitudes and timing of the oscillations of the pelvis by means of simple models of the COM trajectory. To this purpose, it will be assumed that the displacement patterns of COM, and pelvis, can be approximated by sinusoidal functions. This has as an advantage that their trajectories can be modeled more easily. A harmonic analysis, as already applied by Capozzo [5,10], will be used for describing patterns of pelvic displacement, and for determining the validity of the assumption that patterns of pelvic displacement to a large part correspond to sinusoidal displacement patterns. This method uses the stereotypical movement pattern of the pelvis and trunk to define an intrinsic and extrinsic pattern of upper body movement. Ideally, the pelvis and HAT movements should be characterized by a perfect symmetry of displacement with respect to the anatomical planes, and be cyclo-stationary with a periodicity of one stride in left–right direction and half a stride (one step) in vertical and anterior–posterior directions. A frequency decomposition of the displacement functions of pelvis and HAT in vertical and anterior–posterior directions should, in this ideal case, consist of even harmonics only, and similarly, the movements in left–right direction should consist of odd harmonics only. These are the intrinsic harmonics of locomotion. The first intrinsic harmonic of the displacement function for a particular movement direction indicates the basic pattern of displacement during walking, it is therefore called the fundamental intrinsic harmonic. All harmonics other than intrinsic harmonics are called extrinsic harmonics, they may be regarded as indicative for an asymmetry in the act of locomotion.

The objective of the present study is to evaluate the extent to which patterns of pelvic displacement of healthy subjects can be predicted by simple models of the COM trajectory. To this purpose we present a new model for left–right movements of the COM during walking, and we use variants of existing models that have not yet been systematically applied for the prediction of vertical and anterior–posterior displacement patterns of the pelvis during walking. A second purpose of our study is to present normative data for the amplitudes and timing of displacements of the pelvis at a broad range of speeds, and stride lengths/frequencies.

2. Methods

2.1. General procedures and experimental conditions

Ten healthy male subjects volunteered to participate in the experiments. Their ages ranged from 21 to 33 years (mean age 25.4 years). Leg length was measured as the distance from the top of trochanter major femoris to the floor, it ranged from 0.92 to 1.02 m (mean 0.98 m). The subjects' heights ranged from 1.72 to 1.89 m, mean height was 1.82 m. Body weight ranged from 57 to 92.5 kg (mean weight 72.7 kg).

Subjects were asked to walk on a treadmill at six different speeds in an increasing order (0.5, 0.75, 1.0, 1.25, 1.5 and 1.75 m/s). At each treadmill belt speed data were collected in three conditions: first, during normal walking at the particular speed; then with the subject making larger steps than his usual step length; and lastly, with the subject making smaller steps than his usual step length. Subsequently, the next speed condition was offered. All subjects were given the opportunity to practice treadmill walking (and modulating their step length) before the experiments started. Subjects were instructed to look straight ahead and to avoid changing position in left–right or anterior–posterior direction on the treadmill during data recording.

2.2. Data recording

In each condition data were collected during 25 s of walking. Data acquisition involved measuring the signals of eight force transducers placed under the surface of the treadmill and a kinematic acquisition, both by means of an Elite Movement Analysis System sampling at 50 Hz. The force transducers were placed under a left and a right supporting surface which were physically separated by a distance of 0.01 m. The treadmill belt moved over these supporting surfaces. Kinematic data were obtained by attaching small light reflecting markers on different anatomical locations and recording the marker positions by means of two Elite cameras. For the present study, only use was made of two markers placed on the left and right Spina Iliaca Posterior Superior (SIPS). The two cameras were installed behind the walking subjects. The orientation of the reference frame was conform the I.S.B. recommendations for reporting of kinematic data [11]. Consequently, increasing x values correspond to a forward movement, increasing y values to an upward movement, and an increase in z values corresponds to a displacement to the right, relative to the subjects' line of progression.

2.3. Data analysis

The force signals obtained from the transducers under each walking surface were summed to obtain an approximation of the vertical components of the left and right ground reaction forces. These force traces were used to analyze the temporal characteristics of the stride cycle. The onset of an increase in the force signal of the right leg was used as a trigger for determining the beginning and end of a stride cycle. Thus, it was possible to determine the duration (T_c) of each recorded stride cycle. Durations of support phases were measured as the period during which a leg exerted force on the treadmill. Swing duration was measured as the period in which no force was exerted. Durations of both double support phases were measured as the durations of simultaneous support of the left and right leg. Stride length (S_c), and the displacement during single (S_{sup}) and double (S_{bip}) support, were estimated by multiplying the duration of the stride cycle (T_c), single support, or double support, by treadmill belt speed, respectively.

From the recorded kinematic signals, the 3D co-ordinates of the left and right SIPS were used to calculate the 3D trajectory of a point on the pelvis located halfway the line which connects the left and right SIPS. For the purposes of the present study, these kinematic data were left unfiltered before further processing. The recorded force signals were used for defining the beginning and end of the kinematic signal within a stride cycle. The next step in analyzing our data was a harmonic analysis of the kinematic signals within each recorded stride cycle. After detrending the kinematic signal over a stride cycle, the parameters of the first ten harmonics were estimated as a Fourier-series. For displacements in anterior–posterior direction the following Fourier-series was calculated:

$$x(t) = a_{x0} + \sum_{k=1}^{10} a_{xk} \cos(2\pi k f_c t + \phi_{xk}) \tag{1}$$

a_{x0} represents the mean position over the stride cycle; a_{xk} and ϕ_{xk} are the amplitude and phase of the k-th harmonic, respectively; and f_c is stride frequency. Stride frequency was determined as the inverse of the duration of the particular stride cycle under consideration. Thus, for each analyzed stride cycle its stride duration was taken as the fundamental period of the harmonic analysis. Fluctuations in stride frequency during the recording period could, therefore, not influence the harmonic spectrum. The subscript x in function (1) can be replaced by y or z to obtain the functions for movements in vertical or left–right direction. To conform to a usual mode of presentation in studies of locomotion, the values for phase will be presented as percentages of the stride cycle instead of degrees or radians.

Apart from the harmonic analysis of the displacement curves per stride cycle, amplitudes of displacement were also determined by calculating the difference between minimum and maximum displacement for each stride cycle. In all conditions, at least ten stride cycles were analyzed and used for the calculation of individual mean values and standard deviations of spatio-temporal parameters, and the parameters of the first ten harmonics of each displacement function. To obtain measures of the intra-individual variability of the walking pattern in a particular condition, the standard deviations of spatio-temporal parameters and harmonic amplitudes were expressed as a percentage of their mean values in that particular condition. These coefficients of variation, the individual mean spatio-temporal parameters, and the individual means of the parameters of each displacement function were all used for calculating group mean values. The subsequent data analysis was directed at establishing the degree of correspondence of individual and group mean data to the models which will be described in the next sections. Linear regression functions, between the amplitudes of displacement as predicted by a model, and as described by the fundamental intrinsic harmonic of the displacement function, were calculated to obtain an indication of the validity of the models.

2.4. Predictions of the displacements of the pelvis

2.4.1. Displacement in vertical direction

The vertical movement of the COM during walking can be modeled as resulting from a compass gait type during single support and an approximately horizontal trajectory during the periods of double support [12]. When this inverted pendulum model is taken as a starting point for estimating the vertical movement of the pelvis (Δy), a prediction can be made based upon the geometric relations shown in Fig. 1 (upper part):

$$\Delta y_1 = l(1 - \cos\theta) \tag{2}$$

Δy_1 represents the vertical displacement of the pelvis during single support, l is leg length, and θ is the angle between the vertical axis and the leg.

As the assumption of a constant height of the pelvis during the double support phases did not seem to be valid for most of our subjects, an underestimation of the real vertical displacement of the pelvis during the stride cycle will be introduced by this calculation. To correct for this, we made an additional assumption about the vertical displacement during the period of double support. When it is assumed that the pelvic displacement during double support can also be approximated as an (inverted) circular path, and that no discontinuities exist in the trajectory during the transitions from single to double support (and vice versa), the total vertical displacement of the pelvis can be estimated as the sum of Δy_1 and Δy_2 (see Fig. 1, lower part). Δy_1 can be calculated according to Eq. (2), and Δy_2 being the vertical displacement during double support, becomes:

$$\Delta y_2 = l'(1 - \cos\theta) \tag{3}$$

Thus, the amplitude of the fundamental intrinsic harmonic for vertical displacement (a_{y2}) can be predicted according to two models:

$$a_{y2} = 0.5\Delta y_1 \quad \text{(model I) or,} \tag{4}$$

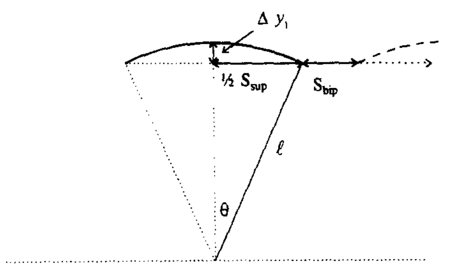
$$a_{y2} = 0.5(\Delta y_1 + \Delta y_2) \quad \text{(model II).} \tag{5}$$

In both models θ can be calculated from:

$$\sin\theta = \frac{S_{\mathrm{sup}}}{2l} \tag{6}$$

From Fig. 1 (lower part) it can be seen that:

$$\frac{l'}{l} = \frac{S_{\mathrm{bip}}}{S_{\mathrm{sup}}} \tag{7}$$



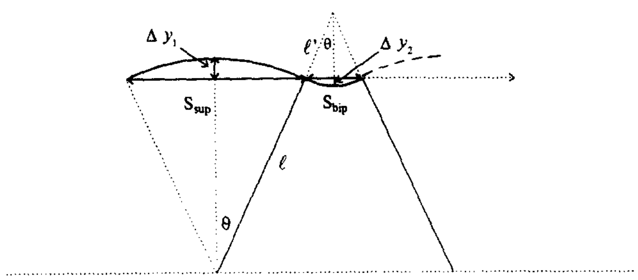


Fig. 1. Models of the vertical displacement of the pelvis during walking. S_{sup} and S_{bip} represent the forward displacement of the pelvis during single and double support, respectively. The upper figure represents model I, the lower model II. The trajectory of vertical displacement of the pelvis is, in both figures, indicated by a bold line. The direction of progression is to the right. In both models, Δy_1 represents the vertical displacement of the pelvis during single support, l is leg length, and θ is the angle between the vertical axis and the leg. In model II, Δy_2 represents the vertical displacement during double support.

Thus, the only variables determining the predicted amplitude of vertical displacement are S_{sup} in model I, and S_{sup} and S_{bip} in model II. As both S_{sup} and S_{bip} are determined by stride length, it is clear that in both models the predicted vertical displacements depend on stride length only. A prediction for the phase of the fundamental intrinsic harmonic can be obtained based upon the timing of the maximal, or minimal, vertical position of the pelvis. According to both models, a maximal vertical position should be reached during the middle of single support, and a minimum should be reached during the (middle of the) double support phase.

2.4.2. Anterior–posterior displacement

During walking the potential and kinetic energy of the trunk vary due to up and down movements and speed fluctuations in forward direction, respectively. In energetically optimal walking, a 100% recovery of potential and kinetic energy would lead to a constant sum of potential and kinetic energy levels throughout the stride cycle [12]. This would require that the changes in potential and kinetic energy levels during the stride cycle are exactly out of phase and equal in amplitude and shape. Although this is not entirely true during normal walking, it can be assumed, in agreement with experimental results [12,13], that the maxima of potential and kinetic energy of the trunk are approximately equal. Thus, a prediction of the fluctuations in relative anterior–posterior position of the pelvis during walking can be made as follows:

$$\Delta(m\,g\,y) = \Delta(\tfrac{1}{2}mv_x^2) \tag{8}$$

and

$$m\,g\,\Delta y \approx m\bar{v}_x\,\Delta v_x \tag{9}$$

Δy is the vertical displacement calculated according to one of both models from the previous section, $\bar{v}_x$ is treadmill belt speed, and Δv_x indicates the fluctuations in forward speed of the COM. m and g, respectively, represent the subjects' mass, and the acceleration due to gravity (9.81 m s^{-2}). As we use the amplitude of the fundamental intrinsic harmonic (a_{x2}) to describe the relative anterior–posterior displacement of the pelvis, the displacement (x) and velocity ($\dot{x}$) functions can be deduced from Eq. (1):

$$4\pi f_c a_{x2} = \tfrac{1}{2}\,\Delta\,v_x \tag{10}$$

Combining this with Eq. (9), and substituting the product of stride frequency and stride length for mean walking speed, results in:

$$a_{x2} = \frac{g\,\Delta y}{8\pi f_c^2 S_c} \tag{11}$$

In which Δy has been calculated according to Eq. (4) or Eq. (5).

From Eq. (11), it is clear that the amplitude of anterior–posterior displacement depends on both stride frequency and stride length. Assuming a 100% recovery of energy, the timing of the maximal vertical position of the pelvis should coincide with a minimal forward velocity, and the timing of the minimal vertical position should coincide with a maximal forward velocity. Consequently, the timing of the maximal forward position of the pelvis should be reached halfway the middle of single and the middle of double support.

2.4.3. Displacement in left–right direction

During locomotion the COM trajectory has to be controlled within narrow limits to maintain dynamic equilibrium. It has been shown that during walking the COM projection on the horizontal plane shows an approximately sinusoidal trajectory between the medial borders of the supporting feet [3,4,14]. The COM trajectory ultimately is determined by the ground reaction forces at the feet and the changes in position of body segments relative to each other. Foot placement, therefore, plays a decisive role in the control of the COM trajectory during walking [15–18]. As the calculated trajectory of the point intermediate to both SIPS approximately will correspond to that of the body's center of mass, the pelvic trajectory in the frontal plane can be estimated based upon a simple model: assuming that the sinusoidal left–right trajectory of the COM solely is determined by foot placement and the ground reaction forces working on the foot, the relevant parameters for predicting the trajectory in left–right direction of the COM (and also the pelvis) can be modeled as in Fig. 2. The model makes use of two basic assumptions: the center of pressure (COP) always lies directly under the foot and stride width is approximately constant. Furthermore, it can be assumed that to ensure lateral stability the force vector $\boldsymbol{F}_{yz}$ should be directed from the COP (at the foot) towards the COM. It can then be deduced from the model in Fig. 2 that:

$$\frac{a_z}{g} = \frac{z_0 - z}{h} \tag{12}$$

in which, a_z represents the acceleration in left–right direction, g is the acceleration due to gravity (9.81 m s^{-2}), h is the approximate height of the COM, z_0 is half the stride width, and z is the left–right displacement of the COM. As we expect the amplitude of the first harmonic (a_{z1}) to describe the left–right displacement of the pelvis, it can be deduced from Eq. (1) and Eq. (12) that:

$$\frac{4\pi^2 f_c^2 a_{z1}}{g} = \frac{z_0 - a_{z1}}{h} \tag{13}$$

Therefore a_{z1} can be estimated as follows:

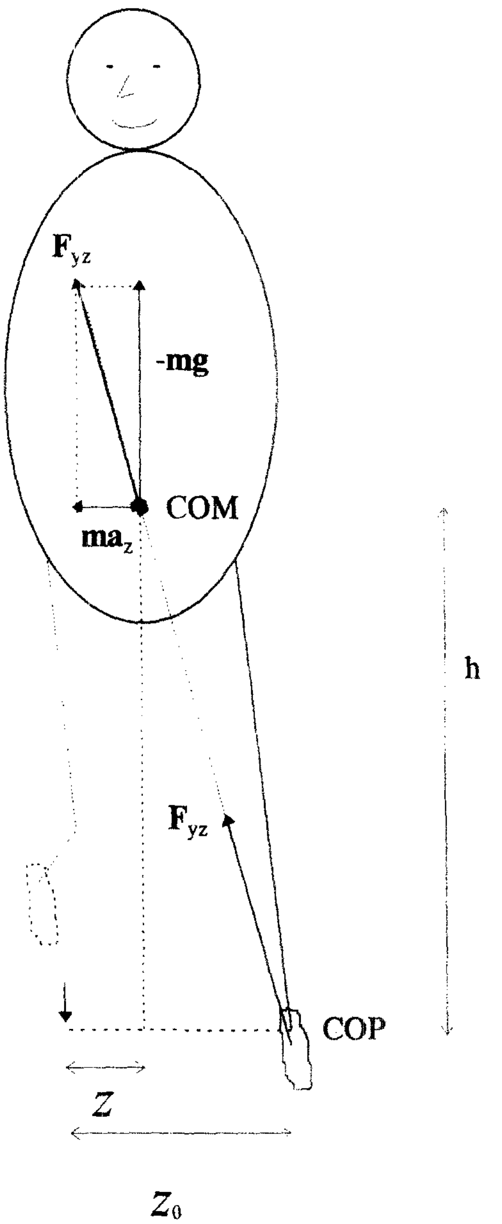


Fig. 2. Model of the control of left-right movements of the COM during walking. The ground reaction forces acting on the foot are indicated by the force vector F_{yz} Its direction is from the COP at the foot towards the COM. a_z represents the acceleration in left-right direction, g is the acceleration due to gravity (9.81 m s^{-2}), m is the subject's mass, h is the height of the COM, z_0 is half the stridewidth, and z is the left–right displacement of the COM. The figure depicts the lateral movement relative to the midline of locomotion (indicated by an arrow), in a subsequent step (indicated by the footprint) a similar lateral movement will be made in contra-lateral direction.

$$a_{z1} = \frac{z_0}{1 + 4\pi^2 f_c^2 \dfrac{h}{g}} \tag{14}$$

Eq. (14), thus, predicts that the amplitude of the sinusoidal left–right movements of the pelvis depends on stride frequency, and stride width only. The equation is very similar to the equations used by Kodde et al. [19,20], for predicting the lateral movements of the COM during quiet stance on a force platform. Eq. (14)

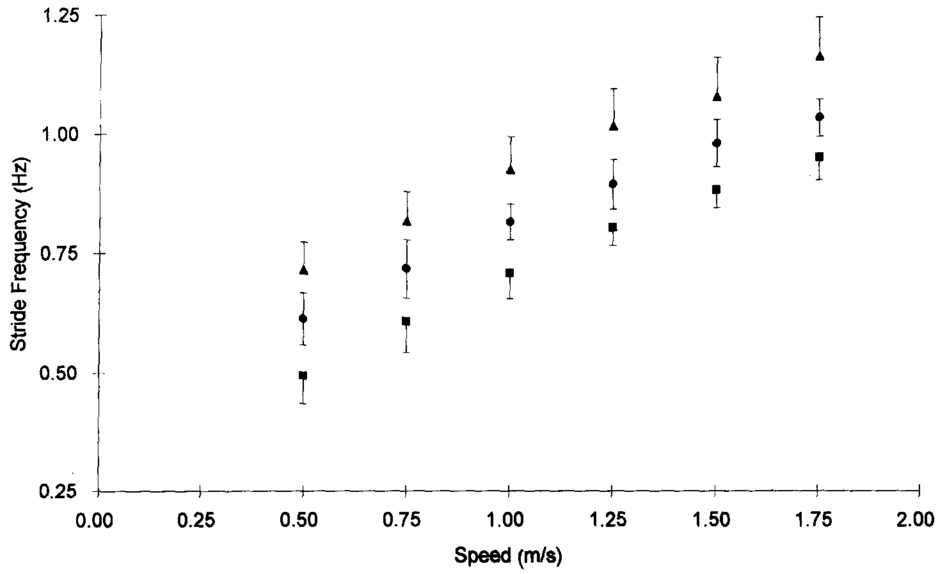


Fig. 3. Mean stride frequencies during normal treadmill walking and in conditions where step length had to be modulated. The different symbols indicate group means ($n = 10$) during normal walking (●), walking with large steps (■) and walking with small steps (▲). S.D.s are indicated by a bar.

has one unknown parameter; as we did not measure stride width, z_0 has to be fitted to experimental data in order to obtain optimal predictions for the amplitude of left–right movement. A prediction for the phase of the fundamental intrinsic harmonic can be obtained based upon the expected timing of the maximal left or right position of the pelvis. These maxima will be reached at approximately the middle of single support of the left and right leg, respectively.

3. Results

3.1. Adaptation of spatio-temporal parameters of the stride cycle

Although the amount of adaptation of stride length and frequency clearly differed between individuals, none of our subjects experienced any difficulties in performing this task. As illustrated by Fig. 3, the mean voluntary increase or decrease in stride frequency (and vice versa stride length) at a particular speed was similar over all treadmill belt speeds. The adaptation of the walking pattern was largest when step length had to be decreased. The mean adaptation of stride frequency was $+0.12$ Hz in the conditions were the subjects had to make small steps and -0.10 Hz in the large steps conditions. Only marginal differences in the amount of adaptation were observed between conditions with different speeds.

With changes in speed, stride frequency, and stride length, variations in the durations of support, swing and double support phases could be observed. The latter changes resulted in changes in the relative timing of the stride cycle which depended on stride length. In all subjects and regardless of the condition, increases in stride length resulted in a decreasing relative duration of the support phase, and an increase in the relative duration of the swing phase. For the group mean values, the relationship between stride length (S_c) and the relative duration of the support phase ($T_{sup\%}$) could be described by a linear regression function: $T_{sup\%} = -6.61 * S_c + 76.6$ ($R^2 = 0.90$, $P < 0.001$).

3.2. The pattern of displacement of the pelvis during walking

A typical example of the individual data obtained during treadmill walking is presented in Fig. 4. The dashed vertical lines in Fig. 4 indicate the start and end of a stride cycle. The patterns of displacement between these lines represent examples of the displacement curves which were described by means of the harmonic analyses. The figure clearly indicates the oscillatory nature of pelvic displacement during walking, but also slight deviations from a perfect periodicity and symmetry of pelvic displacement can be seen. Small fluctuations in the pattern of displacement from stride cycle to stride cycle will be reflected in the calculated coefficients of variation for the harmonics. Any asymmetry of pelvic displacement will manifest itself in a contribution of extrinsic harmonics in the description of the displacement curve. The degree of correspondence of displacement curves to the hypothetical perfect periodicity and symmetry of pelvic movements during walking will be elaborated in the next sections.

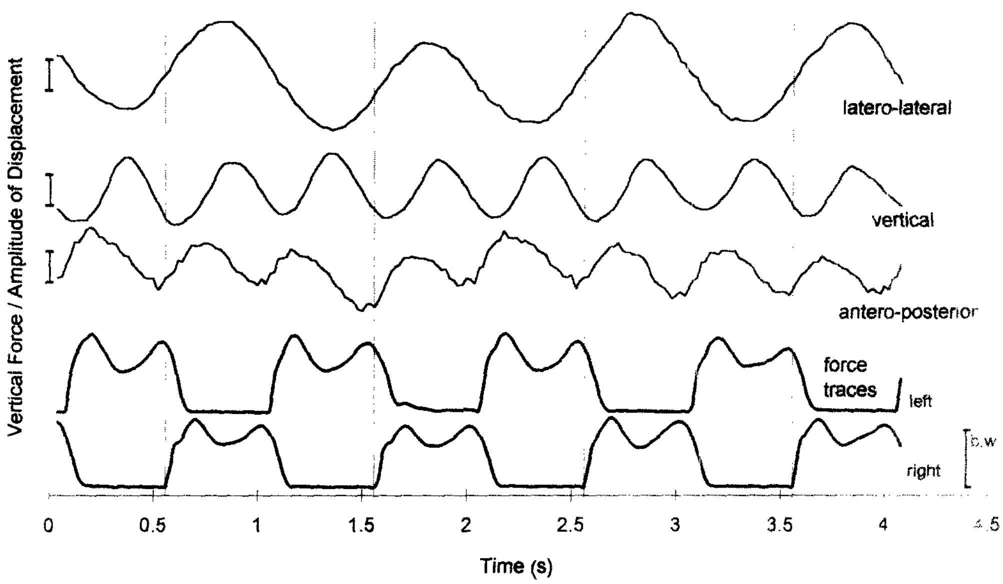


Fig. 4. Representative example of data obtained during treadmill walking. The subject walked with a spontaneously chosen combination of stride length and frequency at a treadmill belt speed of 1.5 ms^{-1}. The upper three traces indicate displacements of the pelvis, the lower two traces indicate left and right vertical force components. The dashed vertical lines in the figure indicate the start and end of a stride cycle. In the upper left part of the figure, the three bars each indicate 20 mm. In the lower right part of the figure, a bar indicates body weight (b.w.).

3.3. The harmonic analyses: contributions of intrinsic and extrinsic harmonics

Before giving a full presentation of the variations in amplitudes of intrinsic and extrinsic harmonics in the experimental conditions, we will present a general report of the results of the harmonic analyses in this section. The calculated x, y and z periodic displacement functions yielded good descriptions of the actual displacement curves with only a limited amount of harmonics. We generally found that the first five harmonics (or less) sufficed for describing at least 99% of the variance of the original displacement curve. The fundamental intrinsic harmonics, i.e. the first harmonic for left–right displacement and the second harmonics for vertical and anterior–posterior displacements, always were largest, and always showed the least cycle-to-cycle variability. As will become clear in the subsequent sections, the contributions of higher intrinsic harmonics mostly served to modify the shape of the calculated displacement curves, they did not substantially contribute to the amplitude of displacement. Their variability generally increased with increasing harmonic number. Extrinsic harmonics (of lower order) appeared to play a role only in the vertical and anterior–posterior displacements. When the amplitudes of the fundamental intrinsic harmonics were compared to the displacement amplitudes as calculated directly from the displacement curves, all trends were similar, and no differences larger than 5 mm were observed.

Fig. 5 is a summary of the results of the harmonic analyses of the displacement curves, it presents the mean amplitudes and the timing of the first maximum of the fundamental intrinsic harmonics ($x2$, $y2$, and $z1$) in all conditions. Systematic influences of conditions are obvious in all fundamental harmonics and in both amplitude and timing. The influence of deviations of the preferred walking pattern and treadmill belt speed upon the amplitude of fundamental harmonics can be deduced from Fig. 5, but will be discussed further in the next three sections. The timing of maxima of the fundamental harmonics shows fairly consistent values for all three movement directions. As indicated by Figs. 4 and 5, maximal left–right displacement and the highest vertical position are reached during single support, in both cases approximately at the middle of the single support phase. Maximal forward displacement is reached at the end of double support or during the transition from double to single support (indicated by (–) in Fig. 5). This is a consistent departure from the model predictions indicated by (×). Although small effects of adaptation of stride frequency (and stride length) upon the phase of the fundamental harmonics can be seen in all movement directions, these effects can largely be attributed to the changes in relative timing of the stride cycle (as indicated in Fig. 5). Only for the displacement in left–right direction are there some small deviations from the timing as predicted in the figure.

3.4. Displacement in vertical direction

The fundamental intrinsic harmonics of the displacement functions for vertical directions clearly accounted for the largest part of the variance in the actual displacement curves. Regardless of the frequency/amplitude combination, the fundamental harmonic on average accounted for 70–75% of the variance in vertical position at the lowest speed (0.5 m s^{-1}). This percentage increased to 86–89% at a treadmill belt speed of 0.75 m s^{-1}, and further increased with increasing speed to approximately 95–98% at the four highest treadmill belt speeds. The lower percentages at the two lowest speeds were largely due to a relatively high contribution of the first harmonic at these speeds. The contribution of this extrinsic harmonic reflects an asymmetry of the vertical pelvic trajectory which was present in all subjects. However, in eight out of ten subjects the first harmonic only made a minor contribution to the displacement function at speeds higher than 1 m/s. Adding the contributions of the first, the second and the fourth harmonics yielded descriptions which always accounted for 98% or more of the variance. The third and fifth harmonics, thus, only marginally contributed to the description of the displacement curves.

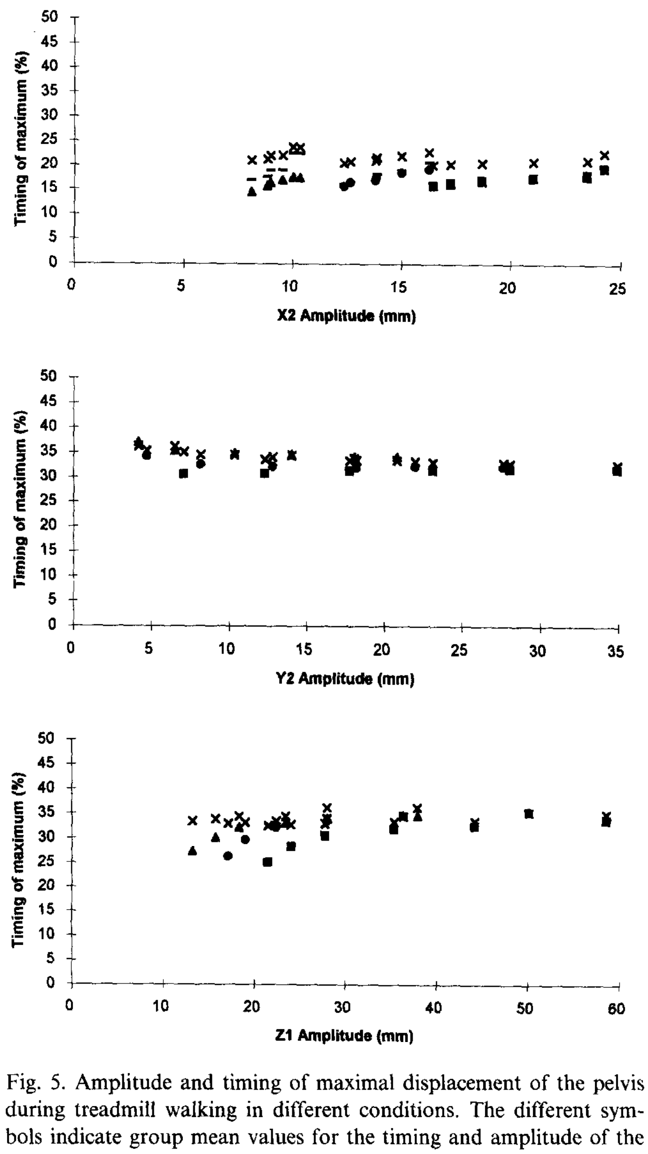


Fig. 5. Amplitude and timing of maximal displacement of the pelvis during treadmill walking in different conditions. The different symbols indicate group mean values for the timing and amplitude of the fundamental intrinsic harmonics ($x2$, $y2$, and $z1$) during normal walking (●), walking with large steps (■) and walking with small steps (▲). (×) indicates the timing of the mid of single support in the two lower figures, and an instant halfway the middle of single and the middle of double support in the upper figure. The upper figure also indicates the transition from double to single support (–). The vertical axis indicates the timing of the stride cycle starting with a right support phase (0%) and ending with the start of the left support phase (50%). The upper figure represents the timing and amplitude of the maximal forward position of the pelvis (as indicated by the fundamental harmonic), the middle figure indicates the highest vertical position, and the lowest figure indicates the maximal displacement to the right.

Fig. 6 shows mean amplitudes of the fundamental intrinsic harmonic in the different conditions and, as two black lines, the predicted amplitudes of the vertical displacement according to the two different models discussed in a previous methods section. To obtain smooth lines indicating the predicted values, a regression equation, linking the group mean values for S_{sup} and S_{bip} to stride length, was used rather than the actual measured values for these variables. As predicted, a stride length dependency of the amplitude of vertical movement of the pelvis is seen. As already foreseen, the first model clearly underestimates the actual values. However, the second model (which additionally takes into account a downward displacement of the pelvis during the double support phases) tends to overestimate the actual values. In fact, it was seen in the individual data that three subjects behaved more according to model I, while four other subjects showed better correspondence to the values predicted by model II. The remaining three subjects showed an intermediate type of behavior. That the mean values for all subjects lie between the predictions of both models is therefore not surprising. Again, the predictive value of both models was tested by means of linear regression equations of the observed mean amplitudes upon the predicted values. The individual predictions generated R^2 values ranging from 0.94 to 0.99 (for the subjects which behaved according to model I) and from 0.97 to 0.99 (for the subjects which behaved according to model II). The remaining subjects showed R^2 values which ranged from 0.79 to 0.99.

Only a small contribution of the fourth harmonic was seen; its amplitude varied between 0.6 and 2.2 mm. The amplitudes of the first harmonic were low also, but they increased with stride length from 1.7 to 3.8 mm. The third harmonic also increased with stride length, but the amplitudes of this harmonic did not exceed 1 mm. Amplitudes of the fifth harmonic were always lower than 0.5 mm. The (relatively) large amplitudes of the first (and third) harmonic at low speeds, together with the strong increase of the second harmonic (Fig. 6), and marginal increases in the first (and third) harmonic amplitudes with stride length, explain the in-

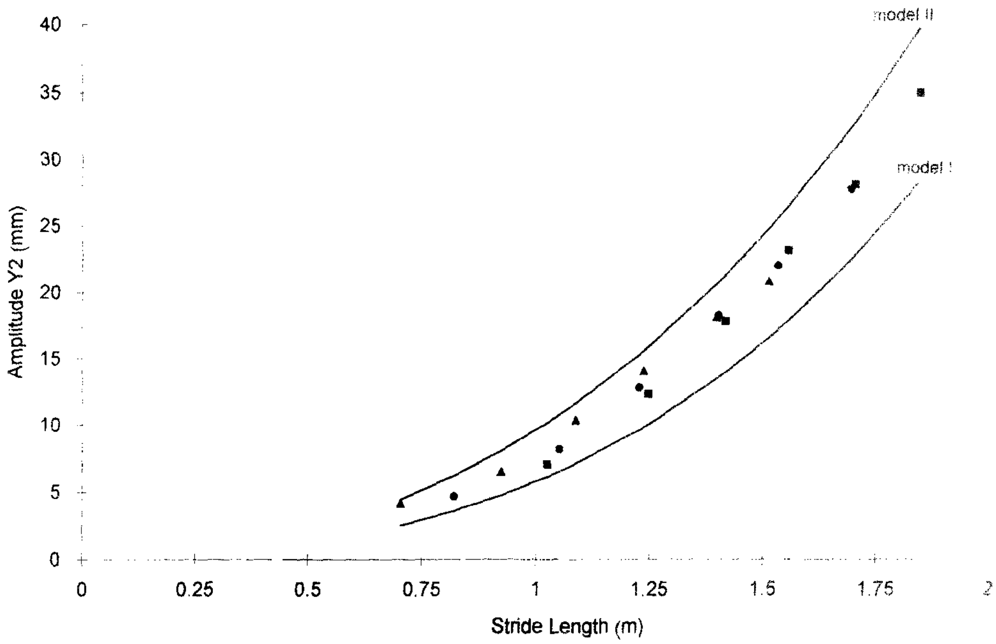


Fig. 6. Mean amplitudes of the fundamental intrinsic harmonic for vertical displacement ($y2$) in the different conditions. The amplitudes predicted according to two different models of vertical displacement during walking are represented as two black lines. Symbols indicate group means ($n = 10$) during normal walking (●), walking with large steps (■) and walking with small steps (▲).

creasing percentage of variance accounted for by the fundamental intrinsic harmonic.

The cycle-to-cycle variability in the amplitudes of the first five harmonics was analyzed by means of the calculated coefficients of variation. These coefficients were only moderately low for the fundamental intrinsic harmonic: for $y2$ the coefficients of variation were highest at the lowest speed (between 30–33% at 0.5 m s^{-1}), and remained approximately constant within a range of 10–20% at all higher speeds. At these speeds, the high frequency (small steps) conditions consistently had a higher variability. Within the range already mentioned, the coefficients of variation in these conditions had values which were at least 5% higher than the values for normal walking and large steps conditions. The latter two conditions always had very similar values. The coefficients of variation for the fourth harmonic exceeded 50% in all conditions, values for the extrinsic harmonics ($y1$, $y3$, $y5$) were excessively large; they always were larger than 100%.

3.5. Displacement in anterior–posterior direction

The mean percentage of variance in anterior–posterior position accounted for by the fundamental intrinsic harmonic ($x2$) fluctuated between 86–94% and did not show an influence of treadmill belt speed. However, within this range, the displacement curves in conditions where the subjects had to make small steps were less well (approximately 5% less) accounted for by the fundamental intrinsic harmonic. Many subjects showed a contribution of the first (extrinsic) harmonic over the whole range of speeds, the mean values of the contribution of this harmonic ranged between 4–12% over all conditions. In all conditions a small contribution of the fourth harmonic was seen. The magnitude of its contribution varied between 1–4%, and tended to increase with speed. As the third and fifth harmonics hardly contributed to the descriptions of the displacement curves, an accurate description was obtained by the sums of the first, the second and the fourth harmonics. The mean percentage of variance explained by these harmonics was minimally 99%.

Fig. 7 shows the mean amplitudes of the fundamental intrinsic harmonic ($x2$) as a function of treadmill belt speed. Different conditions are indicated by the different symbols. Closed symbols represent empirical data, open symbols represent the predicted amplitudes. For the model predictions shown in this figure the predicted vertical displacements according to model II, which included vertical displacements during both single and double support, were used. Although the model predictions tend to overestimate the actual values in the conditions where subjects had to make large steps, the observed trends in empirical data are well predicted by our model. Determination of the predictive value of the model this time yielded an R^2 value of 0.97 for the group means. For the individual predictions the R^2 values varied between 0.88 and 0.97.

Particularly, when subjects made large steps at lower speeds, the first harmonic significantly contributed to the displacement functions. At the two lowest speeds the amplitude of this harmonic varied between 2.9 and 8.1 mm. At higher speeds, values never exceeded 4.2

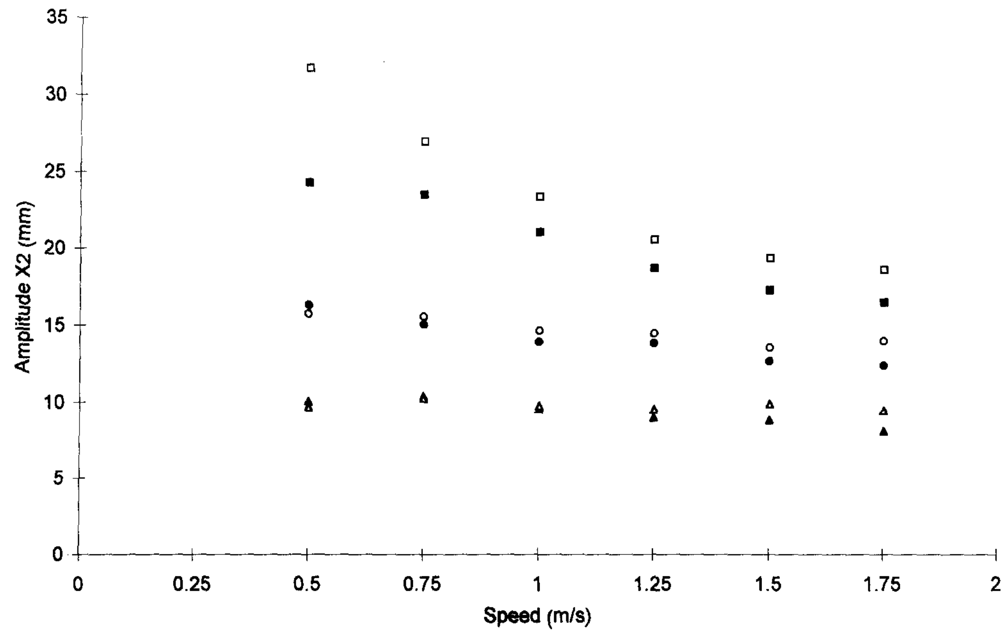


Fig. 7. Mean amplitudes of the fundamental intrinsic harmonic for anterior-posterior displacement ($x2$) as a function of treadmill belt speed. Different conditions are indicated by the different symbols. Symbols indicate group means ($n = 10$) during normal walking (●), walking with large steps (■) and walking with small steps (▲). Closed symbols represent empirical data, open symbols represent the predicted amplitudes.

mm, and minimal values were obtained in conditions where subjects walked normally or made small steps (for the latter conditions the minimal value was 2.2 mm). The fourth harmonic slightly increased with speed in all conditions. Overall, its amplitude varied between 1.1 and 3.2 mm. Within this range, consistent differences were seen between conditions: the amplitudes of the fourth harmonic were lowest in the small steps conditions, highest in large steps conditions and always intermediate for normal conditions. Contributions of the third and fifth harmonics can be neglected, their amplitudes never exceeded 0.75 mm.

Coefficients of variation of the fundamental intrinsic harmonic ($x2$) were always lowest during normal walking, and during the conditions where subjects made large steps. At treadmill belt speeds above 0.5 m s^{-1} the values for these conditions remained constant within a range of 12–14.5%. Within the same speed range the high frequency (small steps) conditions had a variability between 17.5 and 22.5%. At a treadmill belt speed of 0.5 m s^{-1} the variability coefficients were higher for all conditions (16.9, 24.3 and 25.2% for normal walking, large and small step conditions respectively). Coefficients of variation for the fourth harmonic were larger than 50%, and the values for the extrinsic harmonics were again excessively large (more than 150%).

3.6. Displacement in left–right direction

On average, extrinsic harmonics did not play any role of importance in the displacement functions for left–right movements. In all conditions, the displacement curves in left–right direction were almost totally accounted for by the fundamental intrinsic harmonic. At the four lowest treadmill belt speeds, this harmonic, on average, accounted for at least 97% of the variance of the displacement curves. At the two highest speeds this percentage decreased somewhat, but it remained higher than 90%. When the third harmonic was taken in addition to the fundamental intrinsic harmonic, the percentages accounted variance increased to a minimum level of 98%. In the few conditions where the first and third harmonics explained 'only' 98%, the fifth harmonic often added the last percents.

Fig. 8 shows mean amplitudes of the fundamental intrinsic harmonic ($z1$) in the different conditions and, as a black line, the predicted amplitudes of left–right displacement as predicted for the group mean stride frequencies in different conditions. The predictions were made assuming a constant stride width of 0.22 m and a mean pelvic height of 1.10 m. Directly apparent is the confirmation of the predicted dependency of left–right pelvic movements of stride frequency. The validity of the model predictions made for the individual and group mean values was tested by means of a linear regression of the observed mean amplitudes upon the predicted values. For the group means this yielded an R^2 value of 0.98, for the individual predictions the R^2 values ranged between 0.88 and 0.99. Consistent between subject differences were observed in stride width values which yielded the best predictions of left–right movement in an individual. The stride widths used for the individual predictions varied between 0.14 and 0.28 m.

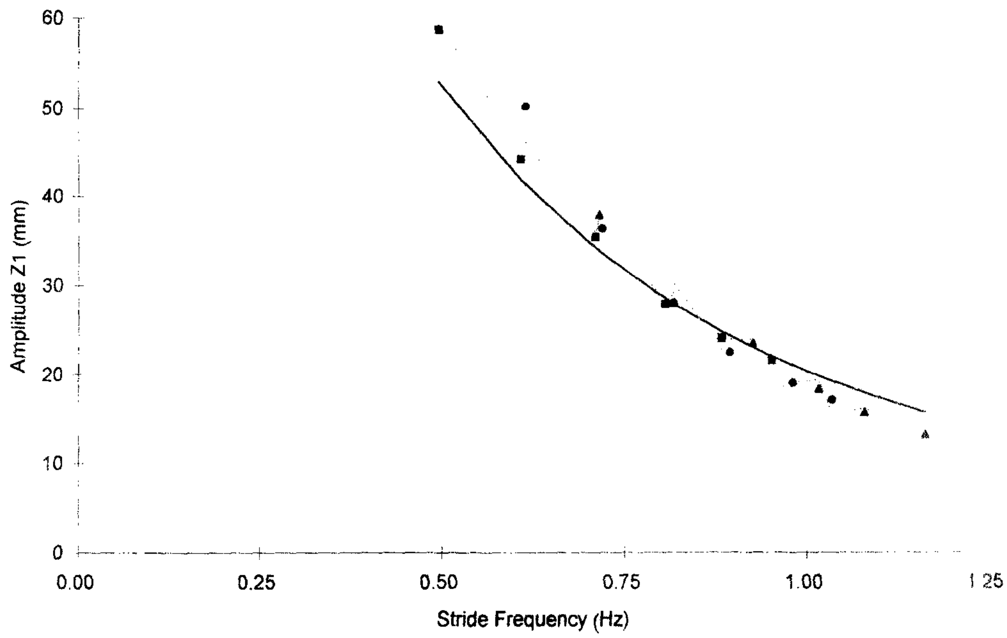


Fig. 8. Mean amplitudes of the fundamental intrinsic harmonic for left-right displacement ($z1$) in the different conditions. The predicted amplitudes of left-right displacement are shown as a black line. Symbols indicate group means ($n = 10$) during normal walking (●), walking with large steps (■) and walking with small steps (▲). The dashed line indicates an improved model prediction to which is referred to in the discussion.

The amplitudes of harmonics other than the fundamental harmonic were always small. Amplitudes of the third and fifth harmonic varied between 1.1–3.6 and 0.4–1.7 mm, respectively. Within these ranges, the amplitudes of both third and fifth harmonic tended to increase with stride frequency. This explains the earlier mentioned decrease in the percentage of variance accounted for by the fundamental harmonic. Amplitudes of the extrinsic harmonics ($z2$, $z4$) were very small: the second harmonic varied between 0.6 and 1.6 mm, and the fourth harmonic always was lower than 0.5 mm.

Only for the fundamental intrinsic harmonic were the coefficients of variation moderately low: for $z1$ the coefficients of variation increased with speed from approximately 10–15% at 0.5 m/s to 20–30% at 1.75 m/s. The high frequency (small steps) conditions most often were characterized by a somewhat higher variability. The coefficients of variation for $z3$ and $z5$ generally exceeded 50%, and they were excessively large for $z2$ and $z4$ (always larger than 100%).

4. Discussion

Our analysis of the 3D displacements of the pelvis clearly shows that the displacements of the pelvis can, to a large extent, be described by the fundamental intrinsic harmonics. Whether subjects walked with spontaneously chosen or voluntarily adapted combinations of stride length and frequency, the fundamental intrinsic harmonics always accounted for most of the actual displacement of the pelvis. This was true regardless of the direction of pelvic displacement. Considering this, and considering the absence of large differences with actual displacements, the fundamental intrinsic harmonic seems to be an adequate description of at least the amplitude of pelvic displacements. Nevertheless, in all movement directions a contribution of higher harmonics was seen when it was attempted to adequately describe the shape of the oscillatory movements. However, an adequate description did not require more than the first five harmonics. A relevant contribution of extrinsic harmonics to the description of pelvic displacement was only observed in the vertical and anterior–posterior displacement of the pelvis, particularly at the lower speeds. The magnitude of this contribution not only varied strongly between individuals, but, as indicated by the coefficients of variation, also from stride-cycle to stride-cycle. The contribution of extrinsic harmonics to the description of anterior–posterior and vertical displacements is in agreement with the notion that extrinsic harmonics reflect asymmetries due to differences in left and right step duration or length, as was also suggested by Capozzo [5].

For the comparison of model predictions with actual displacement of the pelvis we used the amplitude of the fundamental intrinsic harmonic as a measure of the actual displacement. The results of our harmonic analysis, and, particularly, the comparison of the amplitude of the fundamental intrinsic harmonic with the calculated displacement amplitudes, already underline the validity of this approach, but further support for the validity of taking the amplitude of the fundamental intrinsic harmonic as a measure of the actual amplitude of pelvic oscillations is seen when the fundamental intrinsic harmonic amplitudes presented in our results are compared with the amplitudes of left–right, vertical, and anterior–posterior displacement of the pelvis [7], or displacement data of the lower spine (L3) [6] during treadmill walking. Our results on normal treadmill walking and the results of these studies generally show good agreement. There is, however, one major exception; increases in the amplitude of left–right movements were reported [6,7] during walking at higher treadmill belt speeds than those in our study. The latter increase in the amplitude of left–right movements of the pelvis with treadmill belt speed can not be predicted from the empirical and model data presented here. The increase in left–right movements at treadmill belt speeds beyond the speed where usually the transition from walking to running is observed, suggests that our model predictions may only be valid within a limited range of walking speeds, and that additional mechanisms play a role at higher speeds. For normal adult men, a transition from treadmill walking to running usually takes place at a speed of about 1.88 m s^{-1} ([21]), the speed range in our study (0.50–1.75 m s^{-1}) thus at least includes the full range of normal walking speeds.

A good correspondence is seen between our data on normal treadmill walking and those of Capozzo [5] who studied displacements of the head and trunk during free walking at different speeds by means of the same methodology as we used in our study. When Capozzo's and our results are compared within a similar speed range, the changes in amplitudes of the fundamental harmonic with walking (or treadmill belt) speed are very alike for all directions of displacement. Only for the vertical displacement some differences are seen, the fundamental intrinsic harmonic amplitudes as reported by Capozzo are consistently higher than ours. A likely explanation for this difference is the somewhat higher step length which has been reported for free walking in comparison to treadmill walking at similar speeds [22], and possibly also the fact that Capozzo reports individual data on five subjects (and few stride cycles) only. Thus, despite this difference with Capozzo's data on free walking, it seems reasonable to assume that the presented kinematics of treadmill walking do not differ from those who would be obtained during free walking with similar stride lengths and frequencies. Although the biomechanical similarity between treadmill and overground walking is also emphasized by other authors [23], small differences between kinematic data on treadmill walking and overground walking are possible as a result of speed fluctuations of the treadmill [23,24].

Whereas earlier studies only described the displacements of the pelvis (or HAT) as a function of walking speed, our results unequivocally demonstrate that the pattern of displacements of the pelvis is predominantly determined by either stride frequency (left–right movements) or stride length (movements along the vertical axis). The models used for predicting left–right and vertical movements of the pelvis generally give good predictions of the amplitudes of the oscillatory pelvic movements. Displacements in anterior–posterior direction were also predicted reasonably well by a model. They are dependent on both stride frequency and stride length.

Despite the generally good validity of the presented models, a number of shortcomings are apparent. First of all, we are aware that, although COM and pelvic displacement may approximately correspond to sinusoidal movement patterns, the underlying pattern of linear acceleration will certainly not be perfectly described by a sinusoidal function ([4]). However, in experiments were we used accelerometers on the pelvis, we were able to calculate similar displacement patterns as described in the present study, despite the fact that the frequency content of the measured acceleration signals was much higher than that of the displacement signal.

Secondly, in the model for left–right displacement, the simplifying assumption of a constant vertical force vector equal, but opposite in direction, to the acceleration due to gravity also needs discussion. An analysis of the ground reaction forces acting on the feet shows that the real vertical force component changes during the stride cycle, and that the interaction between horizontal and vertical forces changes with speed [25,26]. We therefore tried to improve the model predictions by making the additional assumption that changes in the vertical force vector with speed can be estimated by subtracting the centripetal forces from the gravitational forces working on the body. The centripetal acceleration is due to the circular trajectory of the body over the supporting leg, and it is equal to v^2/l. Thus when g in Eq. (14) is exchanged for $(g - \bar{v}_x^2/l)$, new predictions can be made as indicated by the dashed line in Fig. 8. It can be seen that this model correction indeed leads to a greater accuracy in predicting the amplitude of the fundamental harmonic for left–right displacement, albeit at the cost of some simplicity.

Thirdly, by means of the models, we have made predictions for the mean values of displacement during a number of stride cycles. For the predictions of left–right displacement, we had to assume a constant stride width whereas from stride cycle to stride cycle and, perhaps even more so, from condition to condition stride width may have varied. It is likely that a large part of the variance in left–right movements, as expressed by the coefficients of variation of the fundamental harmonic, reflects variations due to a cycle-to-cycle control of the COM trajectory by means of small variations in foot placement. In our analysis of pelvic displacements during walking we did not specifically pay attention to non-periodic changes in the movement pattern which evolved over a number of stride cycles. This is partly due to the fact that we did not measure certain aspects of the movement pattern (like, for instance, foot placement), but this must also be attributed to the chosen methodology; by means of our harmonic analyses we focused on the periodicity of the kinematic patterns.

In our prediction of the timing of anterior–posterior displacements we made the simplifying assumption of a constancy in the mechanical energy of the COM. As the possibility of speed fluctuations of the treadmill can not be ruled out [26], the kinetic energy of the COM may have shown small deviations from the kinetic energy which can be calculated when a constant treadmill belt speed is assumed. Not only this factor would introduce a small error, during walking the COM does show slight variations in mechanical energy [12,13]. It has been shown that the peak forward velocity is attained at, approximately, heel contact [13]. This means that the phase of the changes in velocity is slightly advanced compared to what is predicted when a 100% recovery of mechanical energy is assumed. The maximal forward position would therefore also be reached earlier than predicted. This probably explains the small error in the predicted timing of maxima in Fig. 5. Experimentally we observed that the transition from double to single support was a better predictor for the timing of maximal forward displacement (Fig. 5).

5. Conclusion

In this study the normal pattern of pelvic displacements has been described for a large range of walking speeds and a large number of different combinations of stride length and stride frequency. The results of the harmonic analysis of the displacement curves per stride cycle not only confirm the sinusoidal character of the pelvic trajectory during walking, they also extend existing data of pelvic kinematics to locomotion modes which deviate from the normal preferred walking pattern. The presented data define the normal movement behavior of the pelvis during walking, and thus may be useful as normative data for situations where normal and pathological locomotion are to be compared. The presented models give insight into the underlying mechanical mechanisms which determine the pelvic trajectory. As the head, arms, and trunk all are supported by the pelvis, the pelvic trajectory determines to a large extent the effect any movement of head, trunk or arm segments may have on the COM trajectory. Knowledge of the pelvic kinematics during walking thus is essential in understanding the regular locomotor pattern and the mechanisms which play a role in the equilibrium control during walking.

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