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Kinematic patterns during walking in children: Application of principal component analysis

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ABSTRACT

The relative displacements of body segments during walking can be reduced to a small number of multi-joint kinematic patterns, pm_k, through Principal Component Analysis (PCA). These patterns were extracted from two groups of children ($n = 8$, aged 6–9 years, 4 males, and n = 8, aged 10–13 years, 4 males) and 7 adults (21–29 years, 1 male), walking on a treadmill at various velocities, normalized to body stature (adimensional Froude number, *Fr*). The three-dimensional coordinates of body markers were captured by an optoelectronic system.

Five components (pm_1 to pm_5) explained 99.1% of the original dataset variance. The relationship between the variance explained ("size") of each pm_k and the *Fr* velocity varied across movement components and age groups. Only pm_1 and pm_2, which described kinematic patterns in the sagittal plane, showed significant differences (at $p < 0.05$) across pairs of age groups. The time course of the size of all the five components matched various mechanical events of the step cycle at the level of both body system and lower limb joints. Such movement components appeared clinically interpretable and lend themselves as potential markers of neural development of walking.

1. Introduction

1.1. Maturation of walking: disentangling age from body size and walking velocity

Walking appears to be a nearly effortless activity for humans. Nevertheless, it is a very complex task as it requires the integrated action of the neurologic, musculoskeletal, cardiovascular, and respiratory systems. To achieve adult-like walking, not only body growth but also central nervous system maturation is required. Body size and spontaneous walking velocity increase in parallel with age. The effect of these two variables and their interaction should be adjusted for, to appraise the role of age itself (i.e., of the neuromotor evolution) in the "maturation" of walking. Curiously enough, this was done for the pendulum-like motion of the body Center of Mass (CoM) (Malloggi et al., 2019) but not in studies on walking kinematics.

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In some studies focusing on CoM, the average forward velocity was normalized to body size through the adimensional Froude number (more on this later). The normalization allowed discovering that the pendulum-like energy-saving mechanism of walking is implemented from the age of five (Cavagna, Franzetti, & Fuchimoto, 1983; Dierick, Lefebvre, Van Den Hecke, & Detrembleur, 2004). Recently, changes in the CoM trajectory have been described in children from 5 to 13 years of age: in particular, an age-related decrease in the lateral oscillation of the CoM has been reported, suggesting a less effective and more prudent balance control in younger children (Malloggi et al., 2019).

On the kinematic side, many authors have studied age-related changes in lower limb motion in healthy children. Findings on spatiotemporal parameters (Hillman, Stansfield, Richardson, & Robb, 2009), joint kinematics (Cığalı, Uluçam, & Bozer, 2011; Smith, Louw, & Brink, 2016; Sutherland, Olshen, Cooper, & Woo, 1980), muscle activity (Agostini et al., 2010), and joint dynamics (Chester, Tingley, & Biden, 2006; Chester & Wrigley, 2008; Cupp, Oeffinger, Tylkowski, & Gage, 1999; Ganley & Powers, 2005) suggested that adult-like motion patterns are gradually attained between 4 and 13 years of age. In all these studies, however, the confounding influence of the walking velocity was overlooked. In contrast, it is well documented that walking velocity has a strong association with many gait features, including kinematics (Frigo & Tesio, 1986; Murray, Mollinger, Gardner, & Sepic, 1984).

The effect of velocity was specifically tested in other research on children and boys aged 4–17 years. Spatiotemporal parameters, electromyographic patterns, joint rotation angles, moments, work and power were found, as expected, to be velocity-dependent (Lythgo, Wilson, & Galea, 2011; Schwartz, Rozumalski, & Trost, 2008; Stansfield et al., 2001; Van Der Linden, Kerr, Hazlewood, Hillman, & Robb, 2002; Van Hamme et al., 2015). However, the confounding effect of age was not considered.

In the present study, synergistic kinematic patterns of children aged 6–13 years were analyzed as a function of age and the adimensional Froude velocity. This design helped to clarify the relationship between these patterns and age.

1.2. Summarising kinematic variables and making age-related changes interpretable

Nowadays, instrumented gait analysis produces, rather easily, a large amount of kinematic data to describe human movement. Principal Component Analysis (PCA) is a multivariate statistical technique that can be used to identify motion patterns by decomposing the kinematics of coordinated actions, like walking, into a few fundamental combinations of joint rotations (motor synergisms or structures, termed Principal Movements, pm_k) by which the motor system organizes an action (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Federolf, 2015; Troje, 2002). A relevant advantage of this approach is that it concentrates the overwhelming amount of information obtained from the motion of body segments into smaller variables. This approach helps to detect the functional structure of multi-plane kinematics by isolating its linear components. Recently, PCA has been applied to characterize and investigate pathological multi-joint walking and balance-postural patterns in children with Cerebral Palsy and Down Syndrome (Rethwilm, Böhm, Dussa, & Federolf, 2019; Zago et al., 2017; Zago et al., 2021; Zago, Federolf, Levy, Condoluci, & Galli, 2019).

In the present study, PCA was used to investigate kinematic walking patterns in a cohort of healthy children aged 6–13 years old, walking on a treadmill at progressively increasing speed. The study aimed to describe the development of the movement components underlying walking in children of various ages and adults. The goal was disentangling the role of age from that of velocity.

2. Methods

2.1. Participants

The current study exploits a dataset gathered in previously published studies from the same group (Tesio et al., 2017; Tesio, Rota, Malloggi, Brugliera, & Catino, 2017). Two men, only, and ten women between 21 and 33 years of age could be screened as adult controls. In order to consistently apply PCA (Daffertshofer et al., 2004), five consecutive walking cycles with the full marker set had to be available at each time frame for each participant and for each velocity. This constraint further unbalanced the women/men ratio of the adult controls: one man and six women could be compared with the children sample (see Discussion on this point). Data were analyzed from healthy children aged 6–9 years (Age1 group, $n = 8$, 4 males), and 10–13 years (Age2 group, n = 8, 4 males), and from healthy adults (Age3 group, 21–29 years, $n = 7$, 1 male).

All participants were volunteers; children's parents and adult participants returned a written informed consent. Children themselves, after proper explanation and parent's consent, gave their oral consent to participation. The ethical approval for this study was obtained from the Institutional Ethics Committee of the first Author. The study was conducted in accordance with the Declaration of Helsinki (Declaration of Helsinki, British Medical Journal, 1964).

2.2. Data collection

Participants were involved in walking trials conducted on a force treadmill. This approach previously returned results comparable with those obtained during overground walking in both adults (Tesio & Rota, 2008) and children (Tesio, Malloggi, et al., 2017). The imposed and steady treadmill velocity allowed for highly reproducible results. Briefly, data were collected at a sampling frequency of 250 Hz using an optoelectronic motion capture system. This consisted of 10 near-infrared stroboscopic cameras (Smart-D, BTS Bioengineering SpA, Milan, Italy) while participants walked on a split-belt force treadmill (1.26 m long; model ADAL 3D; Medical-Development, Andrézieux-Bouthéon, France). Each half-treadmill was mounted on four 3D piezoelectric force sensors (KI 9048B; Kistler, Winterthur, Switzerland). Twenty-two reflective skin markers (diameter: 15 mm) were positioned in correspondence to anatomical landmarks according to the Davis' protocol (Davis, Õunpuu, Tyburski, & Gage, 1991).

Participants wore a t-shirt, short pants, and light gym shoes to minimize the occlusion of optic markers and eventual artifacts during data collection. Participants' height and body mass were recorded. Then, each participant could adapt to treadmill walking for about 30 s. The experimental protocol consisted of walking on the treadmill for 30 to 40 s for each velocity level. Starting velocity was 0.3 m/s, incremented by steps of 0.1 m/s, up to each participant's maximum sustainable velocity. Velocity changes took 5 s in a ramp-like fashion. As a cautionary measure, children were allowed to hold the operator's hand during the velocity transients.

2.3. Data analysis

Publicly available Matlab™ (v. R2018b, The Mathworks Inc., Natwick, USA) routines were used for data processing (Haid, Zago, Promsri, Doix, & Federolf, 2019). Data processing involved three steps: pre-processing (postures registration), principal movements extraction, and post-processing (Federolf, 2015; Zago et al., 2019). In the following, vectors will be marked in bold and matrices in capital bold. For the subsequent analyses, the trajectories of sixteen markers were considered: seventh cervical vertebra, sacrum; right and left acromion, right and left anterior-superior iliac spines, right and left greater trochanters, right and left lateral femoral epicondyles, right and left fibular head, right and left lateral malleolus, right and left fifth metatarsal head.

2.3.1. Pre-processing and postures registration

Five walking cycles were selected for each participant and each walking velocity (strides conventionally started with a right heel-strike). Heel-strike events were identified by automatically detecting the onset of vertical ground reaction force (vertical force exceeding a 30-N threshold) (Tesio & Rota, 2008). Then, raw coordinates were interpolated (cubic splines), filtered (15 Hz low-pass, zero-lag 2nd order Butterworth filter), and each walking cycle was resampled over 100 time points. For each walking velocity, the trajectories of the corresponding markers were encapsulated in 500 × 48 posture matrices containing 500 posture vectors $p_i(t)$ (100 per walking cycle); such vectors described for the i-th participant at each time point t the position of the 16 (×3 dimensions) markers. Coordinates were resolved in a new right-hand orthogonal local reference system originating at the midpoint of the pelvic markers (sacrum and anterior-superior iliac spines), as in (Zago et al., 2017; Zago et al., 2019): the x-axis was anteroposterior and pointed forward; the y-axis was caudo-cranial and pointed upwards; the z-axis was oriented as the vector from the left to the right anterior superior iliac spine and pointed to the body's right side. The time average of each coordinate (1 × 48 vector) was subtracted from every posture vector.

A 1 × 48 weighting vector w_i was created based on the relative mass contribution of each marker j. Weights w_j were obtained by dividing the relative body mass of the corresponding segment by the number of markers on the same segment (Gløersen, Myklebust, Hallén, & Federolf, 2018; Zago et al., 2019). Segments' inertial parameters were adjusted to age, using specific tables (Muri, Winter, & Challis, 2008).

To complete postures registration, they were normalized by dividing each posture vector p_i by the mean Euclidean norm of all the n participants' posture vectors (where n was 500 time points multiplied by the number of velocity increments successfully sustained): $\overline{d_i} = \frac{1}{n}\sum_{t=1}^{n}\|\boldsymbol{p}_i(t)\|_2$. In sum, posture vectors contained the scaled deviations from a participant's mean posture during walking cycles. Postures' scaling and registration are meant to minimize the influence of size mismatches between participants (Federolf, 2015). This enabled the combination of different participants' posture matrices (n × 48) in a global matrix P, obtained by pooling individuals' posture matrices P_i at every available walking velocity.

2.3.2. Principal movements extraction

A Principal Component Analysis was performed on P; PCA returned three items: (1) a set of orthogonal eigenvectors (or principal components, pc_k, 1 × 48), (2) the corresponding eigenvalues ev_k, accounting for the variance explained by each movement component, and (3) a set of time series (scores) obtained by projecting the registered posture matrix on the principal components. Scores are referred to as *principal positions*, $\boldsymbol{pp}_k = \boldsymbol{P} \times \boldsymbol{pc}_k$ (where k is the order of the component), as they describe the temporal course of each movement component.

We retained and analyzed the first K components explaining together more than 99% of the original variance. The projection of principal components on the principal positions allowed to reconstruct and animate a movement component, which takes the name of principal movement (pm_k):

$$\boldsymbol{pm}_k(t) = \overline{\boldsymbol{p}}_i + \boldsymbol{S}_i\left[pp_k(t)\bullet\boldsymbol{pc}_k^{'}\right] \quad (1)$$

where S_i is a 48 × 48 matrix reversing the pre-PCA scaling, defined as $\boldsymbol{S}_i = diag\left(\overline{d_i}/\boldsymbol{w}_j\right)$.

2.3.3. Walking velocity normalization

A unique dependent variable was computed from $pp_k(t)$ curves, namely the Relative Amplitude (RA). RA ranges from 0 to 1, and it is the ratio between the area subtended by the principal position of each participant (weighted by the corresponding ev_k) at each different velocity and the weighted sum of the areas subtended by all components (Zago et al., 2019; Zago, Pacifici, et al., 2017):

$$RA_{i,k} = \frac{ev_k \sum_{t=1}^{n} |pp_k(t)|}{\sum_{k=1}^{K} ev_k \sum_{t=1}^{n} |pp_k(t)|} \quad (2)$$

where K is the number of retained components. RA was used in previous investigations to measure the impact of each pm_k on the overall movement and can be thought of as the relative "size" of each movement component (Zago et al., 2019; Zago, Pacifici, et al., 2017).

The Froude number (*Fr*, see also the Introduction) was used to express the dependent variable with respect to a dynamically normalized walking velocity. The *Fr* number is a dimensionless parameter given by

$$Fr = \frac{V^2}{g \bullet h} \quad (3)$$

where V is the treadmill velocity, i.e., the average walking velocity of the participants, g is the gravity acceleration, and h the participant's height (Malloggi et al., 2019). *Fr*, therefore, represents the ratio between inertial and gravitational forces of participants who are moving in environments mainly characterized by inertia and gravity. The *Fr* number allows for comparisons across individuals of different heights but similar geometry and kinematics (Alexander & Jayes, 1983; Cavagna et al., 1983).

2.4. Statistical analysis

Continuous variables were expressed as means and standard deviations (SDs), while categorical data were expressed as frequencies. The effect of age group and of *Fr* velocity on the RA computed per each principal movement and velocity level was evaluated with linear regression models with repeated measures. Some participants could not reach all the requested walking velocities. Nevertheless, all the participants entered the analysis and contributed to the outcome, each of them only for their available walking velocities. In these models, the within-subjects effect was the 'walking velocity' (Moser, 2004), and the fixed effects were 'age groups' (Age1 = 6–9 years, Age2 = 10–13 years, Age3 = 21–29 years) and '*Fr*'. Moreover, a flexible regression modeling approach based on first-order and second-order fractional polynomials was adopted to embrace a potential nonlinear relationship between *Fr* and the RA of each principal component (Royston, Ambler, & Sauerbrei, 1999). In brief, for a regression model involving a single continuous covariate x, first- and second-order fractional polynomial models can be written as follows:

$$\text{First-order}: \beta_0 + \beta_1 \times X^p \quad (4)$$

$$\text{Second-order}: \beta_0 + \beta_1 \times X^p + \beta_2 \times X^q \quad (5)$$

Values of p and q are typically restricted to the set $S \in \{-2, -1, -0.5, 0, 0.5, 1, 2, 3\}$, which provides practical flexibility. By convention, $X^0 = \log(X)$ and when $p = q$, then X^q is set to $X^p \times \log(X)$. The best functional form was selected based on Akaike's Information Criteria (AIC), with preference given to the model with the lowest AIC (Burnham & Anderson, 2004). This approach was repeated on 100 bootstrap samples, and the most frequently selected functional form was chosen. Finally, pairwise comparisons across age groups were conducted using Tukey's post-hoc tests to take type I error into account (alpha). The statistical significance of two-sided hypothesis tests was set at 0.05.

2.5. Software

Computations, statistics, and creation of plots were conducted with Matlab, SAS version 9.4 (SAS Institute, Cary, NC, USA), and SigmaPlot™ (Systat software Inc., version 14.0, San Jose, CA; USA).

3. Results

The tested walking velocities ranged from 0.3 m/s to 1.3 m/s. Only 7 out of 16 children and 6 out of 7 adults could afford the whole velocity range. We collected a total of 199 sets of five walking cycles each (on average, 12.4 walking velocities per participant), resulting in a global postures matrix P sized 99,500 × 60.

Table 1
Participants' age and anthropometrics, presented as mean (SD), and stratified by group.

	Age1 (6–9 years old)	Age2 (10–13 years old)	Age3 (21–29 years old)
No. of participants	8	8	7
Gender, male/female	4/4	4/4	1/6
Age, years	7.4 (1.1)	11.4 (1.2)	25.6 (2.7)
Weight, kg	29.2 (9.1)	42.2 (5.4)	60.1 (10.0)
Height, cm	129.6 (12.1)	148.0 (6.3)	166.6 (7.8)
BMI, kg•m^{-2}	16.9 (2.5)	19.3 (2.1)	21.5 (2.0)

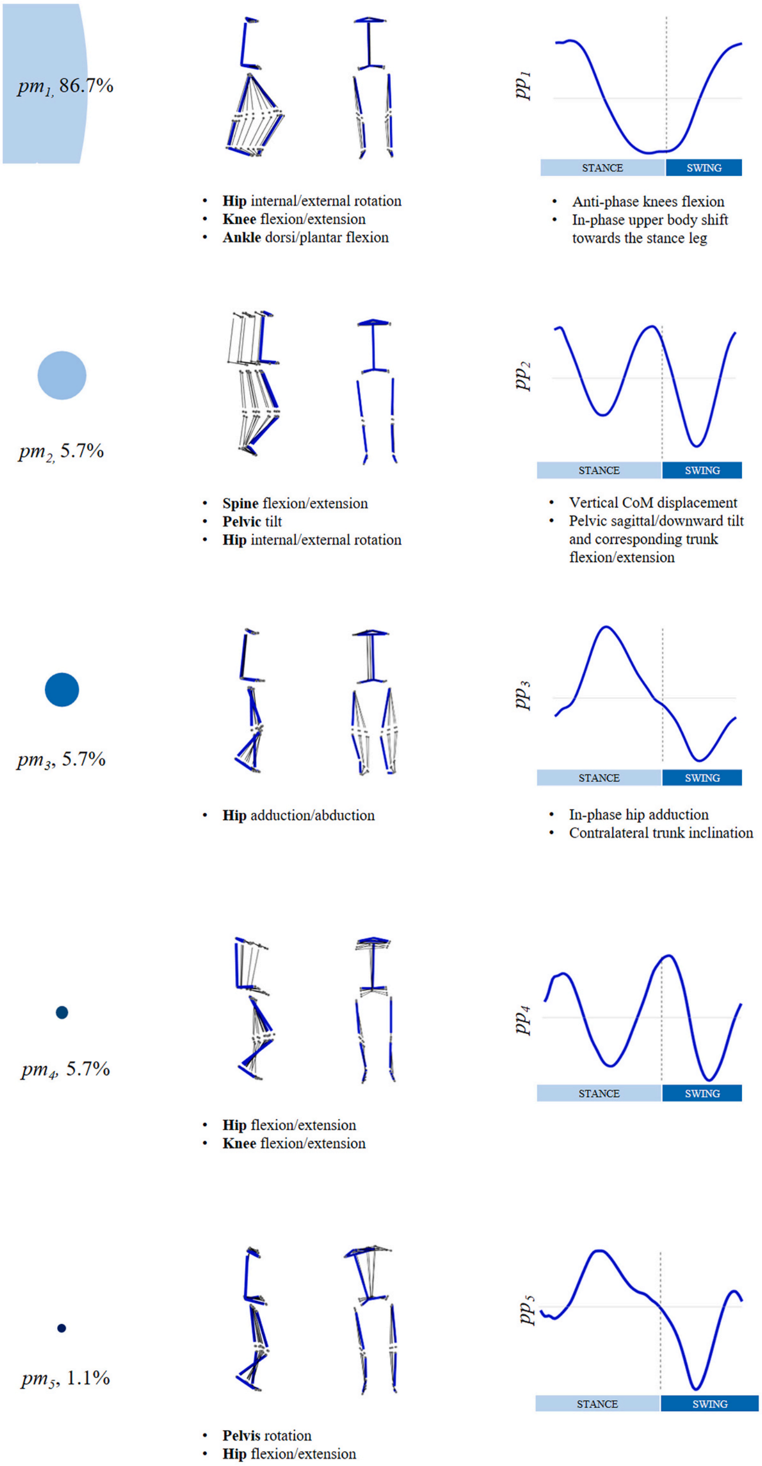

Fig. 1. Principal Movements (pm_k) characterization. Left panel: explained variance (in percentage and visually represented by circles diameter, shown only partially for pm_1); middle panel: stick diagrams (in grey, directions of movement in each posture's space, in blue the posture corresponding to maximum score) and involved joints; right panel: principal position (plots on the right), accompanied by a concise description of the related multi-joint movement pattern. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 1 presents demographic and anthropometric information for each group.

The first five movement components explained 99.1% of the original variance of the dataset. In particular, pm_1 described a sagittal-plane locomotor pattern with a periodicity equal to 1/gait cycle (cycle = 1 stride); pm_2 and pm_4 also occurred on the sagittal plane with a period of 2/gait cycle. pm_2 described vertical whole-body oscillations and represented external hip rotation and knee flexion during the load absorption phase, while pm_4 represented the bilateral knee flexion throughout the swing phase.

The third and fifth principal movements featured multiple-plane actions. In particular, pm_3 involved pelvis inclination, hip adduction, and knee extension during the single support phase, while pm_5 combined pelvis rotation and hip extension at push-off.

Fig. 1 visually represents the Principal Movements' directions and their correspondent interpretation (this will be expanded upon in the Discussion).

Fig. 2 depicts the relationship between the RA of each pm_k and the Fr velocity for each age group. Such relationship varied across movement components: the function was monotonically increasing for pm_1, monotonically decreasing for pm_3, and consisted in a cubic

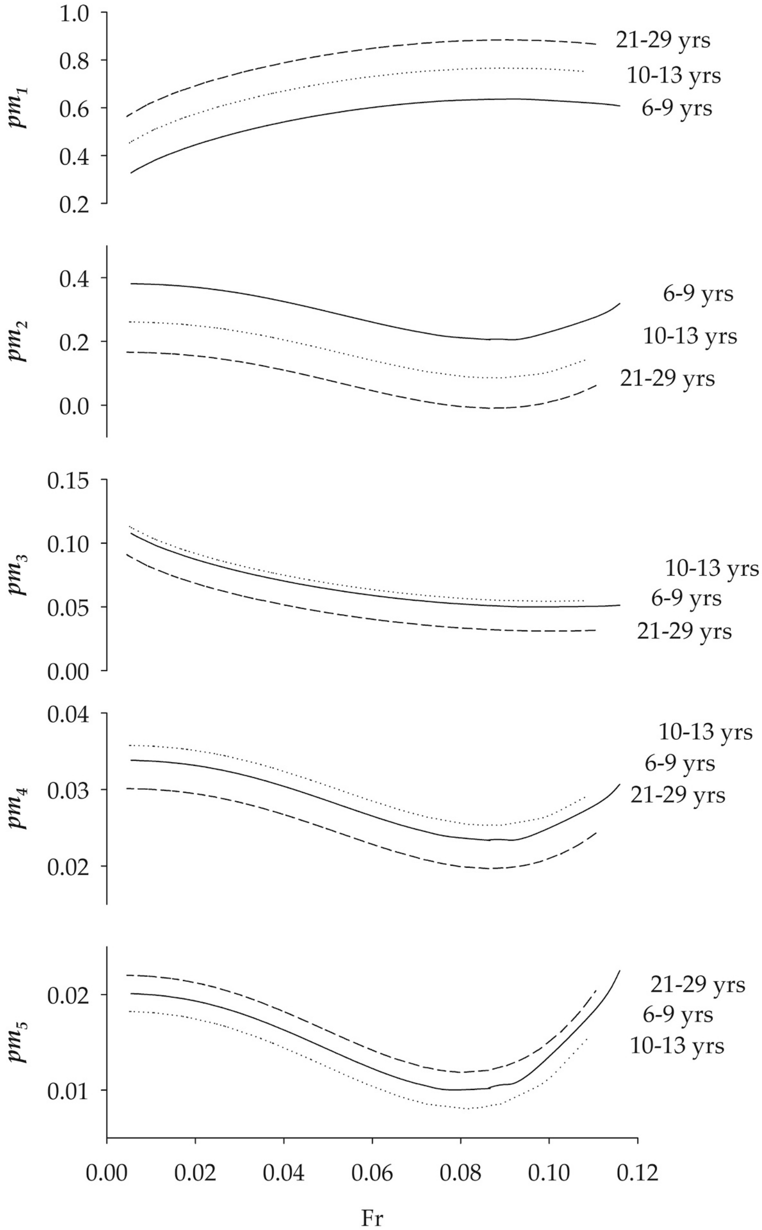

Fig. 2. Relative Amplitude (adimensional) of Principal Movements (pm_k) 1 to 5 as a function of walking velocity (in Froude units, Fr). Each parameter was best-fitted to a first- or second-order fractional polynomial function (see Methods). Curves from different age groups are superimposed and represented with distinct tracts. Note the different scales across the panels. Data were obtained for absolute walking velocities ranging from 0.3 to 1.3 m/s.

relationship for pm_2, pm_4, and pm_5. Values of pm_k depended on age-group. Specifically, pm_1, and pm_5 were higher in adults, while pm_2, pm_3, and pm_4 were higher in the younger group. The relationship between the pm_k and the *Fr* velocity displayed similar shapes for all age groups.

Table 2 shows the *p*-values of the post-hoc contrasts between pairs of age groups. Significant contrasts at $p < 0.05$ were only found between pm_1 and pm_2. Graphic representation of the overlap of the functions relating pm_k to *Fr* velocity across age groups is provided in Supplementary Fig. 1.

4. Discussion

In the present study, the kinematic walking patterns of two groups of healthy children (aged 6–9 and 10–13 years) and a group of healthy adults were compared in principal movement components. A range of increasing walking velocities (normalized to body size) was considered.

The approach embraced in this study has some advantages. First, it did not require any a-priori anthropometric modeling, which is implicit when the kinematic and kinetic variables such as joint angles are determined through additional post-processing steps involving a-priori biomechanical models (Eskofier, Federolf, Kugler, & Nigg, 2013). Also, the present study considered a wide range of sustainable walking velocities. This overcomes a limitation in most of the existing studies where only the spontaneous walking velocity was evaluated. Of note, in a previous paper (Eskofier et al., 2013), a classification method was obtained that correctly differentiated young and elderly participants based on 3D marker motion following a PCA. The participants chose their preferred walking velocity so that the classification might reflect the different velocities.

An interpretation must be attempted here for a) the different amount of variance explained by the movement components (pm_k); b) the relationship of the five components with other mechanical events of walking; c) the relationship of components with respect to walking velocity and, finally and most important, d) the relationship of such movement components with age.

a) As shown in Fig. 1, pm_1 and pm_3 explained together over 90% of the original variance. This fits with the large sagittal rotations of the hips and knees, bilaterally: in fact, this periodicity encompasses the whole stride. On the frontal plane, there is a whole-body translation from left to right and vice versa. Overall, despite a slightly different marker set adopted, the gross behavior of analyzed principal movements matched the kinematic patterns observed in previous papers focusing on healthy and pathological gait, especially concerning the first and second *pm* (Zago et al., 2019; Zago, Sforza, et al., 2017).
b) According to the studies where pm_k values were first used to quantitatively describe the patterns of human movement (Daffertshofer et al., 2004; Troje, 2002; Zago et al., 2019; Zago, Sforza, et al., 2017), the current analysis did not identify classical time-related events during the gait cycle. To widen the possible interpretation of our results, one can hypothesize systematic correlations between the time course of the components evidenced here (principal positions) and the dynamic events of the step cycle (Lacquaniti, Ivanenko, & Zago, 2012). In particular, it may be of interest to split each step into phases characterized by the power required to move the whole body system, represented by its CoM, like those corresponding to the four regions noted by Neptune et al. (Neptune, Zajac, & Kautz, 2004). In Fig. 1, pm_k are labelled 1 to 5 in descending order of variance explanation. Still, they may be made more interpretable if related to the temporal sequence of these phases, e.g., the well-known concurrent joint rotations during overground (Lencioni, Carpinella, Rabuffetti, Marzegan, & Ferrarin, 2019) and (with minor differences) treadmill walking (Fukuchi, Fukuchi, & Duarte, 2018).

Table 2
Results from the nonlinear regression model for the Relative Amplitude (RA) of each Principal Movement (pm_k) over Froude velocity and age group. The third column from left presents the RA_k estimates (means and Standard errors, SE). The fourth column shows 95% Confidence Intervals of RA_k estimates. The rightmost column presents the p-values on contrasts between pairs of age groups. Data were obtained for absolute walking velocities ranging from 0.3 to 1.3 m/s. *: $p < 0.05$.

Outcome	age groups	Difference of RA_k between age groups *mean (SE)*	95% C.I.	p-values (Tukey post hoc)
pm_1	age2 vs age1	0.130 (0.039)	(0.054 ÷ 0.206)	0.01*
	age3 vs age1	0.248 (0.040)	(0.170 ÷ 0.326)	<0.01*
	age2 vs age3	−0.118 (0.040)	(−0.196 ÷ −0.040)	0.02*
pm_2	age2 vs age1	−0.120 (0.043)	(−0.204 ÷ −0.036)	0.03*
	age3 vs age1	−0.215 (0.044)	(−0.301 ÷ −0.129)	0.00*
	age2 vs age3	0.095 (0.044)	(0.009 ÷ 0.181)	0.11
pm_3	age2 vs age1	0.005 (0.011)	(−0.017 ÷ 0.027)	0.90
	age3 vs age1	−0.019 (0.011)	(−0.041 ÷ 0.003)	0.21
	age2 vs age3	0.023 (0.011)	(0.001 ÷ 0.045)	0.10
pm_4	age2 vs age1	0.002 (0.003)	(−0.004 ÷ 0.008)	0.73
	age3 vs age1	−0.004 (0.003)	(−0.010 ÷ 0.002)	0.35
	age2 vs age3	0.006 (0.003)	(0.000 ÷ 0.012)	0.10
pm_5	age2 vs age1	−0.002 (0.004)	(−0.010 ÷ 0.006)	0.87
	age3 vs age1	0.002 (0.004)	(−0.006 ÷ 0.010)	0.87
	age2 vs age3	−0.004 (0.004)	(−0.012 ÷ 0.004)	0.59

Age1 = 6–9 years, Age2 = 10–13 years, Age3 = 21–29 years. RA_k = Relative Amplitude of each principal movement. SE = Standard error. 95% C.I. = 95% Confidence Intervals.

It can be seen that pm_1 and pm_3 were characterized by a larger size following foot-ground contact. This event implies a brisk deceleration of the CoM both downward and forward (as summarized by pm_1), and laterally towards the new supporting side (pm_3). The trunk tilts in the anteroposterior direction, and some "yielding/rebounding" occurs through the whole supporting lower limb. Different individuals may adopt different patterns of hip and knee rotations (Frigo & Tesio, 1986). It is of note that in the lateral direction, at foot impact, the new supporting hip is externally rotated so that the whole trunk is also forced to pivot on the new "pole" on a nearly horizontal plane. This mechanism may explain the hip rotation described by pm_1. The rear limb starts flexing at both hip and knee.

The pm_2 mostly reflects the sagittal kinematic pattern between foot strike and early single stance (roughly, Neptune et al.'s region 2). It is known that here the CoM rises more than it is allowed by its simultaneous deceleration so that a rigid inverted-pendulum model does not hold. The CoM motion reflects the joint movements, i.e., trunk extension, while the pelvis completes its rotation around the supporting hip.

The pm_4 mostly reflects the kinematics at late stance (Neptune et al.'s late region 2 and 3) of the front lower limb (coincident with the initial swing of the rear lower limb). The trunk is now erect, but further lifting is obtained by completing the extension of the supporting knee.

The pm_5 gives back the overall kinematic pattern seen at push-off (including the late stance and the initial double stance phases (Neptune et al.'s region 1 and (contralateral) region 4). At push-off, the rear lower limb extends, and the front hip rotates internally. The lower limb muscles provide most of the power needed to keep the CoM in motion (Tesio et al., 2018). Modest joint rotations occur (see the minimal variance explanation of this component). This consideration reminds us that the "size" of the kinematic phases (Relative Amplitude) may not be synchronous with the "dynamic" phases of the CoM motion. For instance, wide knee flexion occurs during the limb swing (see pm_1), when the body oscillation is nearly passive. Also, the present analysis captured no ankle motion, although the plantar flexors are the main drivers of the CoM translation (Tesio et al., 2018; Zelik & Adamczyk, 2016). Inertial and passive gravitational forces contribute to joint kinematics, not less than muscle forces. In the same vein, the components given in Fig. 1 do not correspond to the five basic muscle activation patterns (as revealed by surface electromyography) demonstrated for walking (Ivanenko, Poppele, & Lacquaniti, 2004).

c) How to explain the effect of velocity?

As shown in Fig. 2, pm_1 and pm_3 reached a plateau above ***Fr*** velocity 0.10. By contrast, pm_2, pm_4 and pm_5 showed a minimum around ***Fr*** velocity 0.08–0.10. Each component's "size" (RA and principal position) is proportional to the total variance explained. Therefore, at low velocities, the reduced "size" of pm_1 (summarising the body oscillations which are widest on the sagittal plane) is complemented by a maximum "size" of pm_2 and pm_4, mostly reflecting the vertical oscillations of the body (both its CoM and segments) and a maximum "size" of pm_5, reflecting hips' flexion-extension at push-off. It is well known that with increasing walking velocity, hence step frequency, the vertical lifts progressively decrease, while step length increases (Cavagna, Thys, & Zamboni, 1976; Cotes & Meade, 1960). Why did the latter rise again between ***Fr*** 0.08 and 0.12? It can be recalled that around ***Fr*** 0.08, the step length starts to saturate primarily because of the saturation of ankle rotation (see the plateau of pm_5). At this point, the rotation of the pelvis increases, thus partially elongating the step (rise of pm_1). Further compensation is provided by the cadence increasing more than step length. This process might explain why the relative "size" of pm_2 and pm_4, reflecting vertical lifts, also increased.

d) How to explain age-related differences?

At dynamically equivalent ***Fr*** velocities, components mainly related to sagittal movements (here, pm_1 and pm_5 were consistently larger in adults than in children (see Fig. 2). This result agrees with the known finding that, after normalization of velocity by stature, children have a lower length/velocity ratio (Alexander, 1984) or, stated otherwise, a lower length/cadence ratio (Hillman et al., 2009; Rota, Perucca, Simone, & Tesio, 2011) when compared to adults. By contrast, pm_2, pm_3, and pm_4 were consistently higher in children than in adults. These components, mainly reflecting vertical and lateral motions, are much less affected by the step length, agreeing with the above considerations.

4.1. Limitations

This study has relevant limits. The sample size was relatively small, thus constraining the generalizability of the results. The power issue cannot be underestimated. Statistical significance was only reached for some of the several contrasts across age groups. On the other hand, the consistency of differences across the various ***Fr*** velocities (Fig. 2) and with respect to biomechanical considerations suggests that the absence of significance in the other contrasts may just reflect a low statistical power.

Moreover, children's compliance and physical skills did not always allow them to walk at the highest requested velocities.

In addition, gender could not be matched for the adult group. The selection bias (6 women, 1 man) was unintentional, given that further constraints were applied retrospectively to a sample from a previous study. In any case, results on adults might be less generalizable to men compared to women. Evidence from the literature shows opposing results on the influence of gender on PCA-driven locomotion pattern analysis during walking (Kobayashi, Hobara, Heldoorn, Kouchi, & Mochimaru, 2016; Rowe, Beauchamp, & Astephen Wilson, 2021; Saito & Kobayashi, 2021; Troje, 2002) and running (Maurer, Federolf, von Tscharner, Stirling, & Nigg, 2012; Nigg, Baltich, Maurer, & Federolf, 2012; Phinyomark, Hettinga, Osis, & Ferber, 2014). Kobayashi et al. demonstrated a significant age-gender interaction in pelvic and hip joint motions in healthy adults aged 20 to 75 (Kobayashi et al., 2016). On the contrary, Rowe et al. identified gender-specific pattern differences independent of age in a sample of asymptomatic women and men between the ages of 20–75 years (Rowe et al., 2021). In addition, minor differences exist in gait kinematics between women and men; in particular, for any given speed, women adopt a lower step length (hence, a higher cadence) (Frimenko, Goodyear, & Bruening, 2015). Minor kinematic differences have also been repeatedly demonstrated, following the seminal work by Murray in 1970 (Bruening, Frimenko, Goodyear, Bowden, & Fullenkamp, 2015; Kerrigan, Todd, & Della Croce, 1998; Murray, Kory, & Sepic, 1970). On the other hand, the critical point of the study is the comparison between age groups in children. These were balanced by gender. In addition, it must be considered that gait maturation is gradually reached from 7 to 13 years, depending on the parameters considered (Agostini et al., 2010; Cavagna et al., 1983; Malloggi et al., 2019; Meyns et al., 2020). Gender differentiation is part of this maturation: therefore, gender related differences are probably of little relevance in the majority of the children participating in this study. For instance, some gender differentiation in EMG patterns only initiated around 9 years of age onwards (Di Nardo et al., 2017). It must be acknowledged, in any case, that the present study presumably did not have the statistical power to detect gender differences in the subsample of participants above the age of 9 years.

Doubts could be cast upon the comparability of treadmill walking with overground walking. However, kinematic, dynamic, and metabolic differences with respect to treadmill walking are minimal both in adults (Song & Hidler, 2008; Tesio & Rota, 2008; Willems & Gosseye, 2013) and children (Tesio, Malloggi, et al., 2017) if the same velocities are compared within groups.

The analyzed walking velocities did not exceed *Fr* 0.12. This is slightly below the expected "optimal" velocity, at which the energetic cost of walking is minimal. Above 5 years, this optimum is estimated at 0.25 if the *Fr* velocity considers lower limb length (Leurs et al., 2011). More precisely, in the inverted pendulum model, the CoM "leg" should be considered. The CoM lies at 57% of the ground-head vertex distance (Swearingen & Young, 1965). "Optimal" velocity can thus be estimated at 0.14 if the stature is considered. In short, care must be taken in extrapolating the present results to higher velocities, given that the kinematic relationships across body segments are velocity-dependent.

It should also be remembered that principal movements are linear decompositions of kinematic trajectories. In principle, the qualitative interpretation of Fig. 1 could be biased by wider movements, as smaller marker displacements encapsulate less variance and, therefore, are less easy to identify in visual representations.

Finally, the cross-sectional nature of the current report does not allow to conclude on specific individual's development, but only to argue the underlying phenomena on the behavior of age groups.

4.2. Methodological considerations

Some epistemological warnings on the extraction of movement components should be stated. First, any "factor" just stems from covariation across its variables: but this is far from demonstrating its causal role. Is the factor artificially created more than discovered? Any causal inference must be strengthened by providing why the covariation should reflect an underlying "cause", rather than merely describing (i.e., numerically summarising) the many properties of the sample at hand. Second, sensible and stable factors can be found more likely if the search is a hypothesis- (rather than data-) driven one, and the "factor" interpreted in the light of established knowledge. This prudent attitude may help to avoid the epistemic fallacy of "nominal realism", i.e., mistaking the name for the real object (McGrane & Maul, 2020).

These requirements seemed satisfied by the fruitful research on "patterns" (or synergies) underlying the muscle activations during gait in most vertebrates (Grillner & El Manira, 2020; Lacquaniti et al., 2012). The same approach seems rewarding also for the kinematic "components" evidenced so far in adults and, here, in children.

The extraction of "components" from walking kinematics may look cryptic to most clinicians. However, this "extraction" is what clinicians implicitly do when visually assessing the overall "normality" of walking or when they strive to focus on critical focal impairments merged into the general motion. However, generally accepted solutions for a formal and quantitative assessment of "dis-mobility" (Cummings, Studenski, & Ferrucci, 2014) are still lacking. This problem may explain the unsuccessful application of refined instrumental gait analysis techniques, providing a hefty amount of information which remains difficult to summarize to make clinical decisions (Benedetti et al., 2017).

4.3. Conclusions

This paper observed that specific multi-segmental gait patterns, which are velocity-dependent, changed from childhood to adult walking. Only kinematic patterns occurring on the sagittal plane showed significant differences as a function of age. These results confirm PCA as a tool for information on the coordination patterns characterizing locomotion across different growth phases. Extracting patterns might represent a useful intermediate step between CoM analyses (Malloggi et al., 2019) and traditional segmental analyses (Tesio & Rota, 2019), as long as they formalize the visual diagnostics performed by clinicians. In particular, the "components" may provide a summary index of the maturation of gait by allowing the assessment of the fit between age and kinematic locomotor patterns, complementing the analysis of the development of the CoM patterns (Malloggi et al., 2019).

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Declaration of Competing Interest

There are no conflicts of interest.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.humov.2021.102892.

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