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The 3D trajectory of the body centre of mass during adult human walking: Evidence for a speed–curvature power law

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ABSTRACT

During straight walking, the body centre of mass (CM) follows a 3D figure-of-eight ("bow-tie") trajectory about 0.2 m long and with sizes around 0.05 m on each orthogonal axis. This was shown in 18 healthy adults walking at 0.3 to 1.4 m s^{-1} on a force-treadmill (Tesio and Rota, 2008). Double integration of force signals can provide both the changes of mechanical energy of the CM and its 3D displacements (Tesio et al., 2010). In the same subjects, the relationship between the tangential speed of the CM, Vt, the curvature, C, and its inverse—the radius of curvature, r_c, were analyzed. A "power law" (PL) model was applied, i.e. log Vt was regressed over log r_c. A PL is known to apply to the most various goal-directed planar movements (e.g. drawing), where the coefficient of log r_c, β, usually takes values around $\frac{1}{3}$. When the PL was fitted to the whole dataset, β was 0.346 and variance explanation, R^2, was 59.8%. However, when the data were split into low- and high-curvature subsets (LC, HC, arbitrary cut-off of $C=0.05$ mm^{-1}, $r_c=20$ mm), β was 0.185 in the LC (R^2 0.214) and 0.486 in the HC (R^2 0.536) tracts. R^2 on the whole dataset increased to 0.763 if the LC–HC classification of the forward speed and their interaction entered the model. The β coefficient, the curvature C, and the pendulum-like recovery of mechanical energy were lower during the double foot-ground contact phase, compared to the single contact. Along the CM trajectory, curvature and muscle power output peaked together around the inversions of lateral direction. Non-zero torsion values were randomly distributed along 60% of the trajectory, suggesting that this is not segmented into piecewise planar tracts. It is proposed that the trajectory can be segmented into one tract that is more actively controlled (tie) where a PL fits poorly and another tract which is more ballistic (bow) where a PL fits well. Results need confirmation through more appropriate 3D PL modelling.

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1. Introduction

In a recent paper (Tesio et al., 2010), the 3D path of the body centre of mass (CM) during walking on a force-treadmill (Tesio and Rota, 2008) was described in healthy adults. The cyclic path of the CM was a figure-of-eight (described as a "bow-tie" trajectory), which changed as a function of the average forward speed. In the present study, the speed/curvature relationship of the 3D trajectory of the CM was investigated. For planar movements this kind of relationship is known to be invariant for a wide variety of human actions, according to a "power law" (PL) (Lacquaniti et al., 1983).

Let ω be the angular or curvilinear speed, r_c the radius of curvature, $C=1/r_c$ the curvature, and k a constant. In its simplest form the law predicts that

$$\omega = kC^{2/3} \tag{1}$$

where k is a constant. Hence the popular term "$\frac{2}{3}$ power law".

An equivalent form relates to the tangential velocity, $Vt=\omega r_c$

$$Vt=\omega r_c = k(1/r_c)^{2/3} \quad r_c = kr_c^{1/3} \tag{2}$$

In log–log form, the relationship becomes linear with slope of $\frac{1}{3}$

$$\log Vt = k' + \tfrac{1}{3}\log r_c \tag{3}$$

where $k'=\log k$ and it is still a constant which has been dubbed the velocity gain factor (Viviani and Cenzato, 1985). This is also known as "the $\frac{1}{3}$ power law". Since being found to subtend drawing movements, the PL has been confirmed in a wide variety of human planar movements such as arm-reaching and slow eye pursuit (de'Sperati and Viviani, 1997; Harris and Wolpert, 1998). The fore-aft trajectory of the foot during straight walking (Ivanenko et al., 2002) and the horizontal path of the head during curved walking have been shown to comply with the PL (Vieilledent et al., 2001; Hicheur et al., 2005). These results cannot be safely extrapolated to the motion of the CM during straight walking. In fact: (a) the CM is not coincident with any anatomical landmark and (b) the trajectory of the CM is deployed within a 3D space (see Fig. 2). This study tested the 2D PL compliance of the motion of the CM within the step, as a premise to further 3D modelling, so far applied to upper

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limb motions (Bennequin et al., 2009; Maoz et al., 2009; Pollick et al., 2009; Schaal and Sternad, 2001).

2. Subjects and methods

The instrumental setting has been described in a dedicated paper (Tesio and Rota, 2008). The subjects and the methods of data recording and analysis were as reported in another companion paper, where the geometry of the 3D path of the CM, only, was dealt with (Tesio et al., 2010) . What follows is a shortened description of the experimental protocol.

2.1. Subjects

Nine men and 9 women were enrolled, 35.9 $\pm$ 10 years, 70.7 $\pm$ 14.2 kg weight, and 1.69 $\pm$ 0.09 m height. Right leg length (anterior–superior iliac spine to ground while standing) was 0.89 $\pm$ 0.06 m, with a difference between the two lower limbs $\leq |0.01|$ m.

2.2. Instrumentation

Ground reaction forces were recorded while subjects walked on a force-mounted split-belt treadmill (model ADAL 3D COP Mz, Medical Development, France) made up of two longitudinal components, 1.26 m $\times$ 0.3 m each. Force signals were sampled at 250 Hz and analyzed off line.

2.3. Walking trials

The subjects were requested to walk at a constant average speed for at least 10 strides. Increasing speeds were tested, from 0.3 up to 1.4 m s^{-1} in 0.1 m s^{-1} increments. Higher speeds, implying longer strides, were deemed unsafe for several of the subjects owing to the risk of falling off the treadmill.

2.4. Ethics

Subjects provided informed consent. The study was approved by the ethical committee of the Istituto Auxologico Italiano-IRCCS, Milan, Italy (statement CE 30-5-2006)

2.5. Computations

i. *Normalizing speeds.* Absolute walking speed, V, was normalized to the a-dimensional Froude numbers (Fr) which take into account the length of the lower limb (l) and the gravity acceleration (g) according to the formula

$$Fr = V^2/gl \tag{4}$$

ii. *Step period definition.* The beginning of each step was taken as the moment the forward speed of the CM reached a maximum (Cavagna et al., 1976; Cavagna and Kaneko, 1977; Tesio et al., 1985).

iii. *Foot-ground contacts.* The phases of single and double stance were recognized by vertical forces exceeding 10 N on either or both half-treadmills, respectively (Tesio and Rota, 2008).

iv. *Geometry of the CM path.* The double integration of the ground reaction forces makes it possible to compute the velocity changes and the displacements of the CM within a 3D frame as per the ''Newtonian'' method (Cavagna, 1975). Fig.1 gives representative original tracings of the fore-aft forces along two strides from one subject (top row), and the corresponding changes of speed (middle row) and position (bottom row) of the CM, coming from successive integrations of the force signals.

v. *Energy transfer and power in the CM motion.* In previous studies, the force-plate method was adopted to measure the pendulum-like changes of kinetic and gravitational potential energy, hence, of the sum of the total energy E_{tot} of the CM during the step (Cavagna et al., 1976, 1983; Tesio et al., 1998a, 1998b, 2010). Changes of E_{tot} imply muscular work (W) and power ($\dot{W}$) output, either positive (contraction while shortening) or negative (contraction while resisting elongation). The kinetic-potential energy transfer abates the changes of E_{tot}, hence saving muscle work. The efficiency of the transfer can be summarized by a ''recovery'' index (Cavagna, 1975) ranging from 0 (no saving) to 1 (fully ballistic motion). In this study, the instantaneous version of the index, r_t, and its average during the step, r_{int}, were computed (Cavagna et al., 2002).

vi. *3D shape of the average trajectory.* For graphic representation, the spatial coordinates of the CM were averaged in a series of 6 strides and the periods were normalized to 100. Fig. 2 gives the average CM trajectory during six strides (panels A) at a walking speed of about 1 m s^{-1} in two representative subjects (upper and lower row, respectively, see legend). The thickened segments mark the phases of double foot-ground contact. For graphic clarity, the figure is rescaled and modified in panels B to D in order to show other features of the path (see the Results section).

vii. *Power law within the CM trajectory: statistical modelling.* The CM undergoes a curvilinear motion in a 3D space. On the matrix of temporally ordered coordinates, Vt, r_c and torsion, τ, were computed throughout the complete trajectories according to Frenet–Serret formulae, with reparameterization from time to arc-length, as implemented in MATLAB software. The compliance of the 3D trajectory of the CM with a power law in the form of Eq. (3) above was tested by linear regression of Vt over r_c, in log–log form. Ninety-five percent confidence limits (95% c.l.) were computed for both slopes and intercepts. Significance was set at $p<0.05$.

viii. *Planarity.* A topic under debate is whether voluntary movements are executed in a piecewise planar way, rather than along continuously changing planes (Morasso, 1983; Soechting and Terzuolo, 1987; Pollick et al., 2009). In analogy with the approach followed by Pollick et al. (2009), torsion data from 2 representative speeds, 0.5 and 1.0 m s^{-1}, were analyzed across the 18 subjects. Each series of 6 strides was segmented into 100 point (0.4 s) bins. The Wilcoxon sign-rank statistics were applied to each bin. According to the piecewise planarity hypothesis, in the great majority of the bins the median torsion should not differ statistically from 0, while significant non-zero values should only come from a minority of bins in inter-planar transition regions.

ix. *Estimation of noise in curvature and torsion measures*. The computation of curvature and, in particular, of torsion of the CM trajectory is highly sensitive to noise. Noise from force signals is attenuated by the double integration procedure (see Fig. 1). Nevertheless, noise on torsion was estimated through an experimental simulation, and it was found to have little impact on the experimental tracings (Fig. 3).

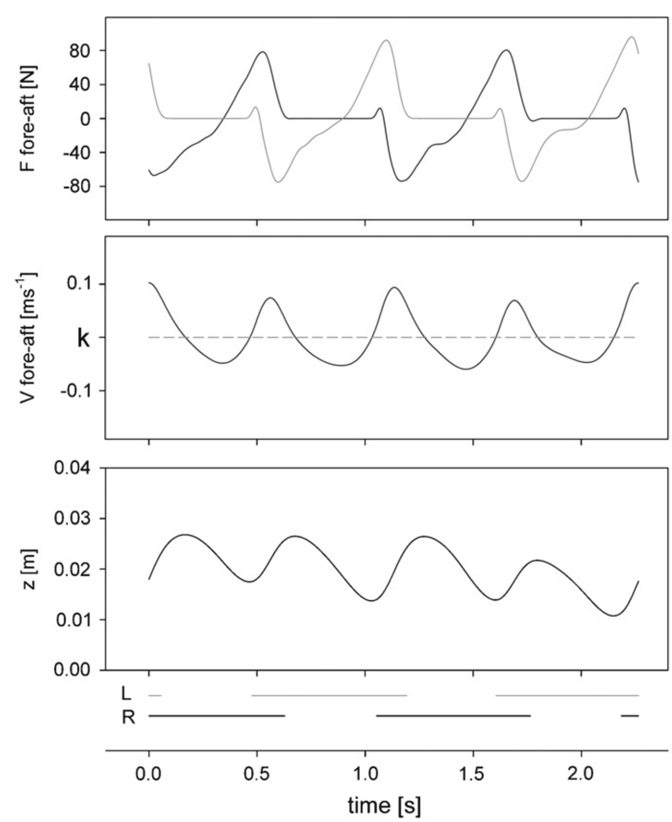

Fig. 1. The ''double-integration'' method for computing the displacements of the body centre of mass (CM) during walking on the force-treadmill. Subject BG was a man, 30 years, 1.73 m, 66 kg, and leg length (superior iliac spine to ground) 0.885 m. He was performing entire successive strides on a force-treadmill. The whole device rests on 3D force transducers placed on firm ground. The upper panel refers to the fore-aft force signals, usually the noisiest ones, recorded during two strides at a sampling rate of 250 Hz, with no filtering, as a function of time given in the abscissa. Integrating these force signals gives, in the middle panel, the fore-aft speed changes over four subsequent strides, with respect to a constant of integration (k; this is equal to the average walking speed, here 1.0 m s^{-1}). Further integration of the speed changes (bottom panel) gives the sagittal oscillations of the CM, z, with respect to the (unknown) initial position. The same algorithms make it possible to compute the lateral and the vertical speed changes (constant of integration assumed to be 0 in both cases) and displacements, hence the 3D coordinates of the CM (see Tesio and Rota, 2008, for technical details).

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