Journal of Biomechanics 43 (2010) 938–944

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

The 3D path of body centre of mass during adult human walking on force treadmill

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ARTICLE INFO

Article history:
Accepted 31 October 2009

Keywords:
Centre of mass
Walking
Gait
3D-trajectory
Instrumented treadmill

ABSTRACT

Three-dimensional (3D) path of the body centre of mass (CM) over an entire stride was computed from ground reaction forces during walking at constant average speed on a treadmill mounted on 3D force sensors. Data were obtained from 18 healthy adults at speeds ranging from 0.30 to 1.40 m s^{-1}, in 0.1 m s^{-1} increments. Six subsequent strides were analyzed for each subject and speed (total strides = 1296). The test session lasted about 30 min (10 min for walking). The CM path had an upward concave figure-of-eight shape that was highly consistent within and across subjects. Vertical displacement of the CM increased monotonically as a function of walking speed. The forward and particularly lateral displacements of the CM showed a U-shaped relationship to speed. The same held for the total 3D displacement (25.6–16.0 cm, depending on the speed). The results provide normative benchmarks and suggest hypotheses for further physiologic and clinical research. The familiar inverted pendulum model might be expanded to gyroscopic, "spin-and-turn" models. Abnormalities of the 3D path might flag motor impairments and recovery.

1. Introduction

During walking the body system as a whole can be represented, from a mechanical standpoint, by its centre of mass (CM). This is a virtual point that is not coincident with any anatomical landmark. Indirect measurement of the CM motion can be obtained from anthropometric modeling and kinematic analysis of body segments (Cappozzo et al., 2005). This approach implies several assumptions (e.g., that the CM is stationary within the body) and approximations (e.g., a close matching between actual and modeled moments of inertia of body segments). Direct measurement of the CM motion can be done by recording the ground reaction forces (Cavagna, 1975, see also Methods). The indirect method may yield results close to those provided by the direct method (Gutierrez-Farewik et al., 2006); the latter, however, remains the gold standard in the framework of Newtonian mechanics. Direct measurements are usually performed while the subject walks on long platforms embedded in the floor because the whole body system has to load onto the sensitive surface (Tesio et al., 1998a). In the last decades the force-plate method has been used in research mostly oriented towards elucidation of the pendulum-like mechanics of walking in human adults (Cavagna et al., 1976), children (Cavagna et al., 1983a), athletes (Cavagna and Franzetti, 1981) and in legged animals (Cavagna et al., 1977). Most investigations focused on the sagittal plane. The method has been applied to various gait impairments in adults (Cavagna et al., 1983b; Tesio et al., 1985, 1998b) and children (Massaad et al., 2004). Clinical applications have been scarce. It may be that the CM appears too abstract from a clinical standpoint, a virtual point that is not anchored to any anatomical body landmark, even though it is a key target of neural control of movement (Massion, 1992). Another reason may be that the method necessitates expensive equipment and skilled personnel. Also, test sessions are lengthy and uncomfortable for most impaired subjects. In fact, because a pre-set speed and constant velocity are required in successive walks along the floor, several entire walks have to be discarded. The duration of the test can be dramatically shortened if walking occurs on a treadmill mounted on force sensors (so-called Gait Analysis on Force Treadmill—GAFT; Tesio and Rota, 2008). On this basis, the present study was devised, making use of the advantages conferred by the GAFT method and aiming at (a) providing normative data on the three-dimensional (3D) path of CM during walking at low and intermediate speeds and (b) exploring the potential usefulness of the path in physiologic and clinical research. In a few seconds several strides can be recorded, at constant and replicable speed. Substantial equivalence between the kinematics of ground and treadmill walking has been already reported in the literature

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doi:10.1016/j.jbiomech.2009.10.049

(Riley et al., 2007). Yet, to the authors' knowledge, the GAFT approach has been reported in just one clinical article (Mahaudens et al., 2009). At least two explanations may be proposed. First, concerns may arise with respect to the quality of force recordings from a long and vibrating treadmill. Second, there is still some debate on the mechanical and neurologic equivalence (both in terms of muscle sequencing and subjective perception) between treadmill and ground walking (for a review on this topic see Zanetti and Schieppati, 2007). The pros and cons of GAFT were discussed in two recent independent studies. In healthy adults, both spatio-temporal, kinematic and kinetic variables, as well as lower limb EMG patterns, were compared between walking on ground and on a treadmill (Tesio and Rota, 2008; Lee and Hidler, 2008). Both studies evidenced only minimal mechanic and neurologic differences, which were deemed of little relevance in view of the many advantages for clinical applications.

2. Subjects and methods

2.1. Subjects

Eighteen healthy adults were enrolled. They were staff members or graduate students from within a Hospital Unit of Rehabilitation Medicine. Subjects had to be free from neurologic and/or orthopedic impairments and from a history of major neurologic or orthopedic impairments. All participants provided informed consent.

2.2. Ethics

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

2.3. Instrumentation

Ground reaction forces were recorded while subjects walked on a force-mounted treadmill (model ADAL 3D COP Mz, Medical Development, Tecmachine Hef, Andrézieux Bouthéon, France). The treadmill consists of two parallel, independent half-treadmills, 1.26×0.3 m each (split-belt treadmill). Each of the two devices can run in opposite directions and has ground contact only through 3D piezoelectric force sensors (type KI 9048B, Kistler Co., Winterthur, Switzerland) placed under four corners of the steel frame. Force signals were sampled at 250 Hz, displayed on a PC screen and stored for offline analysis (see Tesio and Rota, 2008 for details). In this study, force signals from the two independent half-treadmills were summed vectorially, thus reproducing the signals generated from a single treadmill.

2.4. Experimental protocol

i. *Anthropometrics.* Subjects had to wear a t-shirt, shorts and light gym shoes. Then, height and weight were measured (height measure and weighing scales accuracy 2 mm and 50 g, respectively). Leg length was also measured bilaterally while standing and taken as the distance from the anterior superior iliac spine to the foot plant distal to the lateral malleolus.
ii. *Subject's mass calibration*. The force sensors' offset was zeroed. Then the subject was requested to stand quietly on the treadmill for 5 s. The average vertical force signal provided the calibration for the subject's mass.
iii. *Walking test.* The subject was allowed to get used to walking in the middle of the treadmill for about 2 min (Verkerke et al., 2003) at $0.3\ \mathrm{m\,s^{-1}}$. Visual anchoring was provided by a contrasted visual target on the wall in front of the subject, 3 m apart. Thereafter, force recording began. The goal was to analyze 6 subsequent strides with no visible changes in frequency or any sign of imbalance. Then, increasing average speeds were tested, in $0.1\ \mathrm{m\,s^{-1}}$ increments, up to $1.4\ \mathrm{m\,s^{-1}}$. The subject was warned before each speed change, which took 5 s in a ramp-like fashion. Adaptation required 4–6 steps (Choi and Bastian, 2007). This experimental protocol takes about 30 min, of which 10 min is required for the walking performance.

2.5. Computations

i. *Basic algorithms.* The length of the 3D path of the CM was computed on the basis of the algorithms proposed by Cavagna (1975) and Cavagna et al. (1983a, 1983b). In short, force signals provide accelerations of the body CM in the anterior–posterior (AP), the vertical (V) and the lateral (L) directions. From the V forces, the offset provided by the subject's weight is subtracted. Double integration of accelerations provides the CM displacements in a 3D space, representing changes with respect to an unknown initial state. The method implies some assumptions. Air and within-body frictions are neglected. For the analysis of steady-state gait, the average speed over the selected series of strides must be nil and constant (for A–P speed, net of the average treadmill speed). Consistencies of the observed data with the speed assumptions are tested on the tracings of velocity changes, where no drift should be present (see Tesio and Rota, 2008 for details). The CM path during the stride assumes a figure-of-eight shape (see Fig. 1). This figure shows spatial but not temporal information. It is fairly reproducible across subsequent strides (see below the section on Results) and is referred to, heretofore, as CM "cyclogram" (a term first coined by Wong et al., 2004).
ii. *Definition of step and stride period.* A–P speed changes of the CM provided the criterion for defining the step period (Cavagna et al., 1983a, 1983b). This was taken as the time interval between two successive maxima of forward speed. This criterion prevented the error variance coming from the variable timing between heel strike and CM events. The right step within the stride was defined as the cycle of mechanical events occurring between the two subsequent maxima of the A–P speed of the CM following the right heel strike. This was easily detected from V force tracings. Mirror reasoning applied to the left step.
iii. *Measuring the CM path length.* The path length of the CM along the stride and the right and left steps were computed as a function of gait speed. The left/right step ratio was given in log form (Cavagna et al., 1983b). The maximal amplitudes of A–P, V and L displacements of the CM along the stride cyclograms were computed as a function of speed.
iv. *Normalizations.* The walking speed was normalized through the non-dimensional Froude number:

$$\mathrm{Fr} = V^2/gl \tag{1}$$

where V is the average forward walking speed, g the gravity acceleration and l the lower limb length. The Froude number is commonly adopted when the mechanics of walking has to be compared across individuals with similar geometric shape but different overall size (Mc Neill Alexander and Jayes, 1983; Cavagna et al., 1983b). In this study, Fr normalization assigned individual, size-related speeds to each subject, although the same absolute speeds were imposed to all subjects. Fr