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A novel approach to study human posture control: "Principal movements" obtained from a principal component analysis of kinematic marker data

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ABSTRACT

Human upright posture is maintained by postural movements, which can be quantified by "principal movements" (PMs) obtained through a principal component analysis (PCA) of kinematic marker data. The current study expands the concept of "principal movements" in analogy to Newton's mechanics by defining "principal position" (PP), "principal velocity" (PV), and "principal acceleration" (PA) and demonstrates that a linear combination of PPs and PAs determines the center of pressure (COP) variance in upright standing. Twenty-one subjects equipped with 27-markers distributed over all body segments stood on a force plate while their postural movements were recorded using a standard motion tracking system. A PCA calculated on normalized and weighted posture vectors yielded the PPs and their time derivatives, the PVs and PAs. COP variance explained by the PPs and PAs was obtained through a regression analysis. The first 15 PMs quantified 99.3% of the postural variance and explained 99.60% ± 0.22% (mean ± SD) of the anterior–posterior and 98.82 ± 0.74% of the lateral COP variance in the 21 subjects. Calculation of the PMs thus provides a data-driven definition of variables that simultaneously quantify the state of the postural system (PPs and PVs) and the activity of the neuro-muscular controller (PAs). Since the definition of PPs and PAs is consistent with Newton's mechanics, these variables facilitate studying how mechanical variables, such as the COP motion, are governed by the postural control system.

1. Introduction

The human body is a multi-segmental mechanical system whose inter-segment movements are generated and modified by actuators (muscles) controlled by a complex neuronal network. How this system achieves and maintains postural stability has been an important question in biomechanics and neuroscience over many decades.

The center of pressure (COP) excursion is a frequently used variable to assess balance and stability in humans. The COP offers a direct measure of mechanical stability in the sense that a COP position too close to the border of the base of support indicates an instability that must be corrected in order to prevent a fall. Furthermore, the characteristics of the COP motion provide information about the neuro-muscular control, particularly in cases of neuro-muscular deficits, for example, cerebral palsy (Donker et al., 2008; Rose et al., 2002), stroke (Corriveau et al., 2004; Roerdink et al., 2006), concussion (Cavanaugh et al., 2005, 2006; Rubin et al., 1995), or frailty (Lipsitz, 2002) and fall risk (Maki et al., 1994) in the elderly.

How postural movements govern the COP has been described for the inverted pendulum model (Winter et al., 1996, 1993). In this model the COP motion is determined by two aspects. First, the COP position depends on the position of the center of mass (CM) – if the body sways forward, then the COP will also move forward. Second, the COP depends on the acceleration of the body – when leaning forward, the neuro-muscular postural control system needs to produce a moment of force that pushes the body back into an upright position. This moment is created by muscle action moving the COP further forward. Hence, even in this simplified model a forward motion of the COP can be caused by either a forward sway or a backward acceleration of the body. In actual postural movements the COP motion is additionally influenced by other motion patterns such as hip-, knee-, or upper body strategies (Hsu et al., 2007; Pinter et al., 2008), physiologic movements such

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as breathing (Hodges et al., 2002), and movements triggered by cognitive processes such as arousal level (Maki and McLlroy, 1996) or emotional state (Hillman et al., 2004).

The neuro-muscular control of the COP motion has been analyzed by correlating magnitudes of muscle synergies [M-modes (Krishnamoorthy et al., 2003a)] with changes in COP position. Muscle synergies are calculated by performing a principal component analysis (PCA) on normalized electromyographic (EMG) data obtained from several muscles. For voluntary postural sway, M-modes explained 71% (Klous et al., 2011) and 88% (Krishnamoorthy et al., 2003b) of COP variance, however, explained variance dropped markedly when sway frequency was increased (Danna-dos-Santos et al., 2007).

Kinematic synergies obtained from performing a PCA on, for example, joint angles (Alexandrov et al., 1998; Freitas et al., 2006; Tricon et al., 2007; Vernazza et al., 1996) or marker coordinates (Federolf et al., 2013a, 2012b), were also used to study aspects of postural control. When applied to marker coordinates, the PCA transforms the complex, high-dimensional movements of all markers into a set of one-dimensional movement components. These PCA-generated movement components have been called "principal movements" (PMs) (Eskofier et al., 2013; Federolf et al., 2014, 2012b; Maurer et al., 2012). To date, kinematic synergies or PMs are usually considered as theoretical constructs that relate to, but that do not directly quantify the mechanics of the postural control system.

The purposes of the current paper are to define postural PMs consistent with Newton's mechanics; to validate that these PMs represent the mechanics of human postural motion by testing the hypothesis that a linear combination of PMs explain the COP variance; and to outline implications of this methodologic approach for postural control research.

2. Methods

2.1. Participants

Twenty-one volunteers (11 males, 10 females, age 26.4 ± 2.4, height 176 ± 8 cm, weight 71 ± 10 kg [mean ± standard deviation]) with good self-reported general health and no recent injury or other condition that that could affect balance were recruited. All subjects provided written informed consent prior to participating and the study protocol was approved by the Norwegian Regional Ethical Committee.

2.2. Measurement procedures

Measurements started with the volunteers standing in front of the force plate. The subjects were instructed to step onto the force plate into a comfortable, hip-wide, bipedal stance upon a signal from the experimenter. Then the subjects stood on the force plate with their hands on their hips until the experimenter signaled that the measurement was complete. For each subject, 1 trial of 2 min duration was collected. Subjects were not explicitly required to "stand as quiet as possible," however, they were asked to avoid any movements not required for postural control such as scratching or turning the head.

2.3. Instrumentation

The volunteers were equipped with 27 retro-reflective markers placed on the participant's head (3 markers on a custom-build adjustable helmet), C7, manubrium, and placed bilaterally on the acromion, lateral epicondyle, dorsal side of the wrist joint, crista iliaca, trochanter major, thigh, lateral femoral condyles, tibial shaft, lateral malleoli, posterior on the calcaneum, and on the 1st metatarsophalangeal joint. The positions of these markers were sampled at 300 Hz using a motion tracing system consisting of 10 Oqus 400 cameras (Qualisys, Gothenburg, Sweden). The ground reaction forces were recorded at 1500 Hz using an AMTI Optima force plate (AMTI, Watertown, MA, USA). The cameras and the force plate were controlled by a computer running the software Qualisys Track Manager (Qualisys, Gothenburg, Sweden), which synchronized the data acquisition devices and calculated the 3D positions of the markers and the COP position. All further data processing and analyses were conducted in Matlab (The MathWorks Inc., Natick, MA, USA). The data from one minute standing on the force plate, from 20 s to 80 s, was selected and the COP data was down-sampled to 300 Hz.

2.4. Normalization of the data

In analogy to previous studies (Daffertshofer et al., 2004; Federolf et al., 2012a; Troje, 2002; Verrel et al., 2009), the current study interpreted the 3D coordinates (x,y,z) of all markers at a given time t as a posture vector

$$p(t) = [x1(t), y1(t), z1(t), x2(t), \ldots, yj(t), zj(t)] \tag{1}$$

where j is the number of markers ($j=27$ in the current study). [Notation: bold printed, small-letter variables represent vectors; bold printed, capital-letter variables represent matrices; normally printed variables represent scalars; a bar over a variable indicates the mean over time.]

The normalization procedure applied to these posture vectors was designed to allow pooling the posture vectors of all subjects into one matrix $\boldsymbol{M}$ such that (i) every subject contributes an equal share to the variance in $\boldsymbol{M}$, (ii) the influence of anthropometric differences on the variance in $\boldsymbol{M}$ is minimized, (iii) the relative amplitude of the marker motion is preserved, (iv) the fraction of body weight that each marker represents is adequately represented. Pooling the data of all subjects into one matrix has the advantage that results can be directly compared between subjects. Thereto the following steps were conducted: (1) For each subject, subj, a mean posture vector $\overline{\boldsymbol{p}^{\boldsymbol{subj}}} = [\overline{x_1(t)}, \overline{y_1(t)}, \ldots, \overline{z_j(t)}]$ was subtracted from each posture vector:

$$\boldsymbol{p}'(t) = \boldsymbol{p}(t) - \overline{\boldsymbol{p}^{subj}} \tag{2}$$

Thus, the PCA was conducted on deviations from a subject's mean posture, i.e. on postural movements, not on the postures themselves. This procedure is a first step towards removing anthropometric differences.

(2) For each subject the postural movement vectors $\boldsymbol{p}'(t)$ were divided by their mean Euclidian norm $\overline{d^{subj}} = \overline{\|\boldsymbol{p}'(t)\|_2}$ (Federolf et al., 2013a):

$$\boldsymbol{p}''(t) = {}^{1}/\overline{d^{subj}}\ \boldsymbol{p}'(t) \tag{3}$$

This normalization step ensures that each subject contributes the same variance to the pooled matrix $\boldsymbol{M}$ and minimizes amplitude differences due to subjects' anthropometric differences.

(3) Finally, for each marker i a weight factor w_i was defined according to the relative body mass that this marker represented. Specifically, w_i was calculated by dividing the relative weight of the segment to which the marker was attached, m_s, by the number n_s of markers on this segment. For markers placed on joints, the masses of both segments were added. For example, w_i for the knee markers was calculated as $w_i = m_{thigh}/n_{thigh} + m_{shin}/n_{shin}$ with $n_{thigh} = n_{shin} = 3$, $m_{thigh} = 14.16\%$, and $m_{shin} = 4.33\%$ for men and $m_{thigh} = 14.78\%$ and $m_{shin} = 4.81\%$ (De Leva, 1996). Thus, the normalized postural movement vectors had the form

$$\boldsymbol{p}'''(t) = \frac{1}{\overline{d^{subj}}} \left[w_1 \left(x_1(t) - \overline{x_1{}^{subj}} \right), w_1 \left(y_1(t) - \overline{y_1{}^{subj}} \right), \ldots, w_j \left(z_j(t) - \overline{z_j{}^{subj}} \right) \right] \tag{4}$$

2.5. Principal component analysis and kinematics in posture space

The normalized $\boldsymbol{p}'''(t)$ of all participants were concatenated into a 378,000 × 81-matrix $\boldsymbol{M}$ (participants (21)*trial duration (1 min)*measurement frequency (300 Hz) × number of markers (27)*3D; i.e. observations × dimensions), which was then submitted to a PCA. The PCA has three types of results (Daffertshofer et al., 2004; Troje, 2002): a set of orthogonal eigenvectors $\boldsymbol{v_k}$, a set of associated eigenvalues ev_k, and, for each participant, a set of time series $\xi_k^{subj}(t)$ obtained by projecting the normalized postural movement vectors $\boldsymbol{p}'''$ onto the eigenvectors $\boldsymbol{v_k}$.

The whole set of eigenvectors $\{\boldsymbol{v_k}\}$ form an orthonormal basis in the vector space of postural movements. Each eigenvector $\boldsymbol{v_k}$ represents a specific postural movement pattern where the vector components in $\boldsymbol{v_k}$ describe how the movements of the individual markers are correlated with the movements of the other markers (Federolf, 2013c; Federolf et al., 2013b). The scores $\xi_k^{subj}(t)$ quantify the subject's postural movements according to the motion patterns defined by the associated $\boldsymbol{v_k}$ (Daffertshofer et al., 2004). The vectors $\boldsymbol{v_k}$ have been referred to as *principal movements* (PM) (Federolf et al., 2012b). However, to define the PMs consistent with Newton's mechanics, the following new variables are introduced: the amplitude of the PM$_k$ that a subject *subj* shows at time t is given by the scores $\xi_k^{subj}(t)$. In other words, the scores $\xi_k^{subj}(t)$ quantify a position in posture space (i.e. how much the posture at time t deviates from the mean posture in direction of $\boldsymbol{v_k}$). The $\xi_k^{subj}(t)$ could thus be referred to as *"principal position"* (PP$_k$). The rate at which a postural configuration changes can then be quantified by the *principal velocity* (PV$_k$), given as the first time derivative $\frac{d}{dt}\xi_k^{subj}(t)$ of PP$_k$. The acceleration of postural movements can be quantified by *principal accelerations* (PA$_k$), calculated as the second time derivative $\frac{d^2}{dt^2}\xi_k^{subj}(t)$ of PP$_k$. Since all $\boldsymbol{v_k}$ are linear combinations of the original marker coordinates, the definitions of the PP, PV and PAs is consistent with standard differentiation rules and the laws of Newton's mechanics. In the current study, an additional filtering of the PPs with a Butterworth filter (5th order, 2 Hz

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