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Modeling margin of stability with feet in place following a postural perturbation: Effect of altered anthropometric models for estimated extrapolated centre of mass

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ABSTRACT

Background: Maintaining the centre of mass (CoM) of the body within the base of support is a critical component of upright balance; the ability to accurately quantify balance recovery mechanisms is critical for many research teams.

Research question: The purpose of this study was to investigate how exclusion of specific body segments in an anthropometric CoM model influenced a dynamic measure of postural stability, the margin of stability (MoS), following a support-surface perturbation.

Methods: Healthy young adults (n = 10) were instrumented with kinematic markers and a safety harness. Sixteen support-surface translations, scaled to ensure responses did not involve a change in base of support, were then issued (backwards, forwards, left, or right). Whole-body CoM was estimated using four variations of a 13-segment anthropometric model: i) the full-model (WFM), and three simplified models, ii) excluding upper limbs (NAr); iii) excluding upper and lower limbs (HTP); iv) pelvis CoM (CoMp). The CoM calculated for each variant was then used to estimate extrapolated CoM (xCoM) position and the resulting MoS within the plane of postural disturbance.

Results: Comparisons of simplified models to the full model revealed significant differences ($p < 0.05$) in MoS for all models in each perturbation condition; however, the largest differences were following sagittal plane based perturbations. Poor estimates of WFM MoS were most evident for HTP and CoMp models; these were associated with the greatest values of RMS/maximum error, poorest correlations, etc. The simplified models provided low-error approximates for frontal plane perturbations.

Significance: Findings suggest that simplified calculations of CoM can be used by researchers without reducing MoS measurement accuracy; however, the degree of simplification should be context-dependent. For example, CoMp models may be appropriate for questions pertaining to frontal plane MoS; sagittal plane MoS necessitates inclusion of lower limb and HTP segments to prevent underestimation of postural stability.

1. Introduction

Maintaining the centre of mass (CoM) of the body within the base of support (BoS) boundaries is a critical component of upright balance [1]; however, quantifying the ability to balance is a challenging task. Hof et al. [2] proposed a dynamic measure of postural stability, the dynamic margin of stability (MoS), which accounts for both position and velocity of the CoM (i.e. extrapolated CoM; xCoM). While their model is not the first to consider CoM position and its time derivative [3,4], it provides a single measure of dynamic stability that is relatively simple to implement and has been frequently used to quantify stability for a variety of tasks (e.g. obstacle avoidance [5], perturbed balance [6]) and clinical populations [7,8].

A primary factor associated with calculations of MoS is the ability of researchers to estimate whole-body CoM position. One commonly used approach combines kinematic analyses with anthropometric models to estimate segmental CoM; these are weighted and summed to provide an estimate of whole-body CoM [1]. Often researchers will simplify anthropometric models to include a subset of body-segments (e.g. head, trunk and pelvis) for ease of use (i.e. reduced number of markers, decreased setup time) in addition to fewer steps in data processing [5,9–15]. One well-known and commonly used anthropometric model

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within many gait and posture-based research laboratories [5,13,16,17] is the Winter et al. [1] model, which considers the "whole-body" to be fourteen rigid segments, each defined by anatomical landmarks and a proportion of total body mass.

The ability to accurately quantify balance recovery mechanisms is critical for many research teams. Previous work has explored the effectiveness of simplified marker setups in reproducing "whole-body" CoM/xCoM kinematics derived from a full anthropometric model during volitional activities [9–13,15]; however, there remains a limited understanding of how they impact the study of reactionary responses [9,10]. Reducing the number of segments used to examine whole-body stability (via kinematic analyses) may be necessary when equipment limitations (e.g. camera angles), time constraints, or setbacks within collected data sets (e.g. marker occlusion) do not permit the use of a detailed model. As suggested by Jamrakang et al. [11], simplifying a model may also permit a detailed analysis of single segment kinematics (e.g. trunk) while retaining similar "whole-body" estimates.

Therefore, the purpose of the current study was to explore the impact of simplifying a single anthropometric model [1] used to estimate "whole-body" CoM on calculations of MoS. The fidelity of these simplified CoM estimates was challenged further as these calculations were applied to data acquired following a support-surface perturbation which evoked rapid fixed-support postural strategies. Given the results of Yang and Pai [9] and Tisserand [10], we hypothesized that increasingly simplified estimates of "whole-body" CoM would decrease accuracy of the estimates of full anthropometric model MoS during the postural task. As our focus was on the resulting measures of stability, our analyses were conducted within, rather than between the different perturbation conditions present.

2. Methods

2.1. Participants

Ten healthy young adults (5 males; mean ± SD, age: 22.5 ± 1.78 years; height: 1.71 ± 0.09 m; weight: 72.4 ± 12.0 kg) participated in the current study. Individuals were free from self-reported musculoskeletal or neurological conditions that could affect their ability to maintain balance. They did not report taking any medications that could impact motor control and had normal or corrected to normal vision. All participants gave written consent to participate; the study was approved by the institutional research ethics board.

2.2. Experimental protocol

Data collected and analyzed were a subset of a larger experimental protocol that included 104 trials. Participants were fitted with a suspended safety harness and asked to stand barefoot on a robotic motion platform (5 × 2 m; Shelley Automation, Cambridge, ON, Canada). Foot position was traced and kept approximately hip-width apart throughout. Sixteen trials were analyzed for the current study; all included a support-surface perturbation that evoked a fixed-BoS postural response (displacement, peak velocity, and acceleration adapted from Maki et al. [18] within the sagittal [forward/backward] or frontal plane [left/right]; Table 1). Participants were instructed not to take a step and were free to use their arms as necessary (excluding grasping the harness). Randomization of experimental trials into four blocks mitigated the effects of anticipation on postural responses; conditions were evenly distributed amongst blocks and demonstrations were given prior to data collection.

Kinematic data were collected using a 12-camera Optitrack system (100 Hz; NaturalPoint, Corvallis, OR, USA). Rigid bodies of four reflective markers were fastened to body segments (e.g. trunk, thigh, pelvis, etc.) and single markers covered in retro-reflective tape were then digitized relative to these fixed rigid bodies. Markers were placed on anatomical landmarks following the criterion outlined in Winter et al. [1]; six additional markers were used to digitize BoS boundaries (e.g. toe, head of fifth metatarsal, heel).

Table 1
Each direction of perturbation (and the respective magnitude: displacement, Δx; peak velocity, $\dot{x}$; acceleration, $\ddot{x}$) utilized in the experimental protocol (see Maki et al. [17]).

Perturbation Direction	Perturbation Magnitude		
	Δx (cm)	$\dot{x}$ Peak (cm/s)	$\ddot{x}$ (cm/s2)
Forward	7	22	73
Backwards	9	30	100
Left	9	29	96
Right	9	29	96

2.3. Data processing and centre of mass estimation

Data analyses were completed with Visual3D software (Version 6, C-Motion Inc., Germantown, MD, USA). Kinematic data was first interpolated then low-pass filtered using a 4th order dual-pass Butterworth filter (cut-off: 6 Hz). Within each trial, four models consisting of N body segments (i.e. variations of the 14-segment Winter et al. [1] model) were used to estimate position x of the "whole-body" (net) CoM using the following equation:

$$\mathrm{CoM}(x) = \sum_{i}^{N} (x_i * k_i) \tag{1}$$

where x_i (segment CoM position) and $\mathrm{CoM}(x) \in \mathbb{R}^3$ (anteroposterior, AP; mediolateral, ML; vertical) and k_i is the mass-proportion constant of the i^{th} segment (Table 2). The first variation was the Winter full model (WFM) that used 13 segments in place of the original 14; the second and third trunk segments were combined due in part to the safety harness

Table 2
Whole-body center of mass (CoM) used for the Margin of Stability calculations were derived from the following segments/mass proportions (k, adapted from Winter et al. [1]) for each model variant (full model, WFM; full model excluding arm segments, NAr; head, trunk and pelvis, HTP; pelvis, CoMp). "Trunk #" indicates one of four trunk segment(s), as defined by Winter et al. [1], that were utilized for estimation of total CoM. The resulting CoM positions during quiet standing are also presented (calculated during ten seconds of quiet standing with respect to the xiphoid process). Negative values in the sagittal plane indicate the posterior direction; in the frontal plane, they indicate the left direction.

	Full Model	Simplified Models		
	WFM	NAr	HTP	CoMp
Segment Mass Proportions (k)				
Head	0.081	0.090	0.140	–
Trunk 4	0.136	0.151	0.234	–
Trunk 2 + 3	0.143	0.159	0.247	–
Trunk 1	0.078	0.087	0.134	–
Pelvis	0.142	0.158	0.245	1.000
Upper Arm (*2)	0.028	–	–	–
Forearm (*2)	0.022	–	–	–
Thigh (*2)	0.100	0.110	–	–
Shank (*2)	0.060	0.067	–	–
TOTAL	1.000	1.000	1.000	1.000
CoM Directions (cm)				
Anteroposterior (AP)	−12.19 (0.36)	−10.95 (0.34)	−8.89 (0.30)	−5.36 (0.49)
Vertical	−25.19 (1.06)	26.09 (1.04)	−4.19 (0.77)	−24.64 (1.29)
Mediolateral (ML)	0.07 (0.27)	1.01 (0.26)	0.05 (0.23)	0.24 (0.41)