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Accuracy and concurrent validity of a sensor-based analysis of sit-to-stand movements in older adults

G. Ruben H. Regterschot [a,*], Wei Zhang [b], Heribert Baldus [b], Martin Stevens [c], Wiebren Zijlstra [a,d]

[a] University of Groningen, University Medical Center Groningen, Center for Human Movement Sciences, PO Box 196, 9700 AD Groningen, The Netherlands
[b] Philips Research Europe, Personal Health Department, High Tech Campus 34, 5656 AE Eindhoven, The Netherlands
[c] University of Groningen, University Medical Center Groningen, Department of Orthopaedics, PO Box 30.001, 9700 RB Groningen, The Netherlands
[d] Institute of Movement and Sport Gerontology, German Sport University Cologne, Am Sportpark Müngersdorf 6, 50933 Cologne, Germany

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ABSTRACT

Body-fixed motion sensors have been applied for the assessment of sit-to-stand (STS) performance. However, the accuracy and concurrent validity of sensor-based estimations of the body's center of mass (CoM) motion during STS are unclear. Therefore, this study investigated the accuracy and concurrent validity of sensor-based measures of CoM motion during STS in older adults. Accuracy and concurrent validity were investigated by comparing the sensor-based method to a force plate method. Twenty-seven older adults (20 females, 7 males; age: 72–94 years) performed five STS movements while data were collected with force plates and motion sensors on the hip and chest. Hip maximal acceleration provided an accurate estimation of the center of mass (CoM) maximal acceleration (limits of agreement (LOA) smaller than 5% of the CoM maximal acceleration; estimated and real CoM maximal acceleration did not differ ($p = 0.823$)). Other hip STS measures and the chest STS measures did not provide accurate estimations of CoM motion (LOA ranged from −155.6% to 333.3% of the CoM value; sensor-based measures overestimated CoM motion (range p: <0.001 to 0.01)). However, the hip sensor did not overestimate maximal jerk of the CoM ($p = 0.679$). Moderate to very strong associations were observed between sensor-based estimations and actual CoM motion (range $r = 0.64$–0.94, $p < 0.001$). Hence, sensor-based estimations of CoM motion during STS are possible, but accuracy is limited. The sensor-based method cannot replace laboratory methods for a mechanical analysis of CoM motion during STS but it may be a practical alternative for the clinical assessment of STS performance in older persons.

1. Introduction

Leg muscle power is a determinant of movement execution and an important parameter for measuring intervention effects in older adults [1–4]. However, available methods for the measurement of leg muscle power, such as force plates and isokinetic dynamometers, have practical disadvantages (e.g. costs, difficult to transport, uneasy to use) that limit the application of these methods in clinical settings. Therefore, it is important that practical methods are developed for the measurement of leg muscle power in older adults.

Zijlstra et al. developed an alternative method for the measurement of leg muscle power based on small body-fixed motion sensors [5]. Results indicated fair to excellent concurrent validity of sensor-based estimations of the body's center of mass (CoM) peak power during sit-to-stand (STS) based on a comparison with force plate measurements. In addition, results showed that a sensor on the hip provided more accurate estimations of vertical CoM acceleration during STS than a sensor on the chest, which overestimated CoM accelerations and peak powers [5].

However, Zijlstra et al. [5] had a limited number of older subjects and only evaluated the accuracy and concurrent validity of sensor-based estimation of CoM peak power during STS, not the accuracy and concurrent validity of other sensor-based measures of CoM motion during STS. Recent studies developed sensor-based measures of CoM motion during STS in addition to peak power [6,7]. However, the accuracy and concurrent validity of these additional sensor-based measures of CoM motion during

* Corresponding author. Tel.: +31 50 363 8846; fax: +31 50 363 3150.
E-mail addresses: rubenregterschot@hotmail.com (G. Ruben H. Regterschot), wei.zhang01@philips.com (W. Zhang), heribert.baldus@philips.com (H. Baldus), m.stevens@umcg.nl (M. Stevens), zijlstra@dshs-koeln.de (W. Zijlstra).

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STS (e.g. maximal vertical velocity, maximal vertical acceleration) are unclear. Accuracy and concurrent validity indicate respectively the closeness of agreement and the association of a measured value with an accepted reference value. When accuracy and concurrent validity are adequate, the sensor-based method can be used for a mechanical analysis of CoM motion during STS instead of laboratory-based stationary methods (e.g. force plates). Therefore, the aim of this study was to investigate the accuracy and concurrent validity of sensor-based measures of CoM motion during STS in older adults. For this purpose, we compared the sensor-based method to a standard laboratory method consisting of force plates under the chair and feet of the participants. Based on the findings of Zijlstra et al. [5], we hypothesized that hip STS measures have adequate concurrent validity and accuracy, and that chest STS measures have adequate concurrent validity but overestimate CoM kinematics resulting in inadequate accuracy.

2. Methods

2.1. Participants

Participants were recruited from a health care center, a residential care home and sheltered houses. Older adults could participate in this study when they were able to rise from a chair in one attempt without using the hands, walk at least 10 m (with or without a cane or wheeled walker), and when they were at least 70 years of age. Participants were excluded when they had any cognitive, neurological, cardiovascular or respiratory disorder, lower extremity orthopaedic surgery or a stroke within the six months before the study, severe comorbidity, significantly reduced vision.

In this study 27 older adults (20 females) participated on a voluntary basis. Age ranged from 72 to 94 years (81.7 $\pm$ 5.6 years), body mass ranged from 48.0 to 98.9 kg (75.7 $\pm$ 13.3 kg), and body height was between 1.46 and 1.84 m (1.63 $\pm$ 0.09 m).

The present study was approved by the Medical Ethical Committee of University Medical Center Groningen, the Netherlands (METc2011.054). The study protocol is in agreement with the Helsinki declaration. An informed consent was signed by all participants.

2.2. Procedures

Participants performed five chair rise movements at a normal speed from a standard chair (height: 0.47 m). Prior to standing up participants were leaning against the back of the chair. Participants stood up with their arms crossed in front of the chest. After rising from the chair participants stood still for 5 s before sitting down again. Between each stand-to-sit and sit-to-stand movement participants sat still on the chair for 10 s.

2.3. Data acquisition

2.3.1. Body-fixed motion sensors

During the sit-to-stand movements participants wore two body-fixed motion sensors (π-Node, Philips). Both sensors consisted of a 3D accelerometer ($\pm$2 g), a 3D gyroscope ($\pm$300°/s) and a 3D magnetometer ($\pm$2 G) [8]. One sensor was worn on the right side of the hip (just above the trochanter major femoris; see Fig. 2 in Regterschot et al. [6]) because a previous study demonstrated that a sensor at this location provides a more accurate estimation of the vertical CoM acceleration during STS than sensors at other locations [5]. The other sensor was worn on the chest (sternum) because this location seems preferable for activity monitoring [9]. Hereafter we refer to the sensors as hip sensor and chest sensor. Data were collected with 50 Hz sampling frequency and wirelessly transmitted to a PC for storage [8].

2.3.2. Force plates

Measurements were performed with two force plates (Bertec; each plate measured 0.60 m $\times$ 0.40 m). One force plate was positioned under both feet of a participant, the other force plate was located under the chair. Force plate data were sampled with 100 Hz frequency.

2.4. Data processing

Processing of the sensor data and the force plate data was performed using Matlab (The Mathworks, Inc.; version 7.12).

2.4.1. Body-fixed motion sensors

Quaternions were applied to estimate the accelerations of the hip sensor and the chest sensor in the global coordinate system using the accelerometer, gyroscope and magnetometer data in the sensor coordinate system [8]. Data were filtered with a low-pass Butterworth filter (cut-off frequency of 3 Hz [5]).

2.4.2. Force plates

The vertical data of both force plates were filtered with a low-pass Butterworth filter (cut-off frequency of 3 Hz [5]). Subsequently the vertical force data of both force plates were summed to calculate the vertical force of the body's CoM (F_{com}). Vertical acceleration of the body's CoM (a_{com}) was computed by applying the following formula: $a_{com} = F_{com}/m$. In this formula, m represents body mass.

2.5. Data analysis

Data analysis was performed using Matlab (The Mathworks, Inc.; version 7.12). The vertical acceleration data of the hip sensor in the global coordinate system (a_{hip}), the vertical acceleration data of the chest sensor in the global coordinate system (a_{chest}), and the vertical acceleration data of the body's center of mass as determined based on force plates (a_{com}) were separately used for the calculation of the following STS measures:

1. *STS duration (s):* Interval between the initiation of the forward trunk rotation prior to STS and the first intersection of the vertical acceleration data with the gravitational acceleration, after the deceleration phase (see Fig. 3 in Regterschot et al. [6]).
2. *Maximal acceleration (m/s^2):* Maximal vertical acceleration during STS.
3. *Maximal jerk (m/s^3)*: Maximal positive jerk during the acceleration phase of the STS movement. Jerk was calculated as $\text{jerk}_i = (a_{i+1} - a_i)/(1/f_s)$ with i indicating sample number, a vertical acceleration, and f_s sampling frequency.
4. *Maximal velocity (m/s):* Maximal vertical velocity during STS. Velocity was estimated by numerical integration of the vertical acceleration during STS. We assumed that vertical velocity was 0 m/s at the initiation of STS.
5. *Peak power (W):* Maximal vertical power generated during STS. Force (F) and velocity (v) were multiplied to estimate power: $P_i = F_i \cdot v_i$ [5]. Force (F) was computed using: $F_i = m \cdot a_i$. In this formula m represents body mass and i indicates sample number.
6. *Scaled peak power (dimensionless):* Peak power corrected for body mass (m), body height (l) and gravity (g): $P_{scaled} = P/(m \times g^{1.5} \times l^{0.5})$ [10].
7. *SD stabilization phase (m/s^2):* SD of the vertical acceleration data during the stabilization phase. Since in most older persons the duration of the stabilization phase is shorter or equal to 0.8 s