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A body-fixed-sensor-based analysis of power during sit-to-stand movements

Wiebren Zijlstra [a,*], Robertus Wilhelmus Bisseling [a], Stephan Schlumbohm [b], Heribert Baldus [b]

[a] Center for Human Movement Sciences, University Medical Center Groningen, Groningen, The Netherlands
[b] Philips Research, Eindhoven, The Netherlands

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ABSTRACT

This study presents an analysis of power exertion for lifting the body's centre of mass (CoM) during rising from a chair. Five healthy young (21–44 years) and 12 healthy older (70–79 years) subjects performed sit-to-stand (STS) movements while data were measured with force-plates underneath chair and feet and motion sensors attached to different locations on the upper and lower trunk. Force-plate-data were used to determine the timing of STS movements and the vertical power for lifting the CoM from a sitting to a standing position. Data of three-dimensional hybrid motion sensors, consisting of accelerometers, gyroscopes and earth-magnetic-field sensors, were used to determine vertical accelerations and power. The comparison of sensor-based estimations of peak power with peak power calculated from force-plate-data demonstrated fair to excellent linear relationships for all sensor locations on the trunk. The best approximation of peak power was obtained by a weighted combination of data measured at different trunk locations. Results of the older subjects were consistent with those of the young subjects performing slow, normal and fast STS movements. The presented approach is relevant for monitoring fall risk and assessment of mobility in older people. Similar approaches for assessing power may be developed for other mobility related activities, such as stair walking, or sports related activities such as jumping.

1. Introduction

Muscle power, i.e. the speed with which muscular forces produce movements of body segments, is a determinant of the successful performance of sportive activities such as high jump, javelin throw, or weight lifting. However, also the performance of activities in daily life requires that enough muscle power is generated. The safe performance of mobility related activities, such as rising from a chair, walking and stair climbing, requires a control of position and orientation of body segments whilst the body's centre of mass (CoM) is moved from one position to the other. The ability to quickly produce sufficient muscle force is of paramount importance for controlling CoM movements during mobility related activities, and studies demonstrate that decreased muscle strength and/or power is associated with functional limitations [1,2].

In older people, loss of muscle strength is a strong predictor of falls. Falls are one of the largest health risk factors in older people and they can have serious consequences regarding both physical functioning (e.g. fractures) and psycho-social functioning (e.g. fear of falls leading to activity restriction and social isolation). Fall prevention requires an early identification of fall risk and the use of effective and targeted fall prevention strategies. Measures of muscle strength and balance performance are early indicators of fall risk [3], and studies of exercise-based interventions have shown that combinations of strength and balance training can reduce the risk of falling in older people [4]. Available field tests for assessing balance and mobility usually yield very basic parameters such as movement duration and speed, therefore objective methods that assess additional relevant aspects of movement performance may contribute to fall prediction and/or outcome assessment in older people.

A recent development is the use of body-fixed motion sensors for assessing motor functioning. Though suitable methods for assessing mobility related activities based on the use of motion sensors are available [5], these methods have not yet been applied to their full potential in older people [6]. Moreover, feasible methods to assess muscle strength or power during daily activities are missing. Currently, such measures can only be obtained by using a motion analysis system and/or force-plate in a laboratory environment, or by using devices which usually have been developed for assessing sports related performance in younger subjects or athletes (e.g. cycling or rowing ergometers). Thus, it would be an enormous advance if, in addition to other aspects, power can be assessed during mobility related activities. Particularly, the performance of the sit-to-stand (STS) transfer is

* Corresponding author at: Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, P.O. Box 196, 9700 AD Groningen, The Netherlands. Tel.: +31 050 363 7868; fax: +31 050 363 3150.
E-mail address: w.zijlstra@med.umcg.nl (W. Zijlstra).

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very relevant in this respect. The STS transfer is a regular mobility related activity in daily life, it is performed multiple times a day, and studies have demonstrated that measures of STS performance are important indicators of overall functioning and balance performance in older persons [7,8].

Assuming that trunk kinematics can be used to approximate CoM movement, this paper analyses whether the power to lift the body's CoM during the STS movement can be determined based on motion sensors on the trunk. The most critical aspect in this approach is an accurate estimation of vertical CoM accelerations. Therefore, hybrid sensors, consisting of accelerometers, gyroscopes, and magnetometers, were used to calculate vertical accelerations. Subsequently, power was calculated from vertical accelerations at each trunk location and from vertical accelerations of a virtual CoM position which was estimated by a weighted average of acceleration data. To evaluate the sensor-based approaches, a comparison was made to data obtained by a conventional laboratory approach; i.e. a camera-based motion analysis system in combination with two force-plates under chair and feet.

2. Methods

Seventeen healthy subjects participated in this study. Before participating, subjects received information about the nature of the study. All participants signed an informed consent. The Medical Ethical Committee of the University Medical Center Groningen approved of the study.

2.1. Subjects

Five healthy young subjects (3 males, 2 females) and 12 healthy older subjects (4 males, 8 females) volunteered to participate in this study. In the young subjects, age ranged from 21 to 44 years (mean 29.2 years), length ranged from 1.64 to 1.93 m (mean 1.79 m), and body mass ranged from 58.5 to 89 kg (mean 76.2 kg). All young subjects indicated to be free of any physical complaints that might interfere with the activities in this study. In the older subjects, age ranged from 70 to 79 years (mean 73.8 years), length ranged from 1.58 to 1.93 m (mean 1.705 m), and body mass ranged from 54 to 102.5 kg (mean 74.8 kg). All older subjects were able to stand up from a chair without help, and none of them had severe cardiovascular disease, orthopaedic problems, or a cognitive disorder that might interfere with comprehension or execution of the tasks in this study. None of the subjects had a fall history, a recent (<1 year) surgery or an unstable medical condition.

2.2. Experimental procedures

All subjects were asked to stand up from a chair of standard height (0.46 m). In the 5 young subjects an extensive measurement protocol was used. First, subjects were instructed to stand up at a self-chosen "natural" speed. Then subjects were instructed to perform STS transfers as slow as possible with the arms crossed in front of the abdomen. Thus, it was impossible to use arm rests while standing up. Subsequently, subjects were instructed to perform STS at a natural speed again. Thereafter, they were instructed to perform STS transfers as fast as possible, and finally subjects were asked to stand up from the chair at a self-chosen natural speed while using the left and right arm rests. For each set of instructions 5 trials were measured. The older subjects only performed STS transfers at a self-chosen natural speed. They were free to use arm rests. Three trials were measured in each older subject.

2.3. Data acquisition

Data acquisition included the measurement of analogue data of two Bertec force-plates (each 0.40 m × 0.60 m) by an Optotrak Data Acquisition Unit, 3D marker positions by an Optotrak motion analysis system (Northern Digital Inc., Canada), and 3D motion data by wireless inertial hybrid sensor nodes (Philips, Aquisgrain 2.0 [9]). One force-plate was beneath the chair and both feet were resting on the second force-plate. After amplifying, analogue force-plate signals were converted to digital by the 16-bit A/D converter of the Optotrak system. All force and position data were measured at a sampling frequency of 100 Hz.

Three matchbox-sized sensor nodes were attached to different trunk locations. Two sensor nodes were attached to a neoprene bandage worn around the hips just below the waist; one was attached to the right side of the pelvis, above trochanter major femoris, the other was attached at the dorsal side of the pelvis with the centre of the node located between both spina iliaca posterior superior (SIPS). The third sensor node was attached to the sternum. A fourth sensor node was placed on a nearby table and used for synchronization of sensor-data to force and position data. Each sensor node measured 3D-accelerations (±2 g), 3D-angular velocities (±300 deg/s), and 3D-orientation in the earth-magnetic-field (±2 Gauss). Data of each sensor node was transmitted to a nearby PC, equipped with a wireless receiver. Raw sensor-data was transmitted from the nodes to the host PC via a proprietary multi-point packetized radio protocol [10]. For node synchronization the flood time synchronization protocol was used: one node in the network is the time master and is updating periodically the time for every other node (cf. [11]). Synchronous data of all sensor nodes were obtained at a sampling rate of 50 Hz.

3D positions of all sensor nodes and of anatomical landmarks on the right side of the body were measured by the Optotrak motion analysis system. Synchronization of force and position data with the wireless sensor-data was realized by using the fourth sensor node with a mounted Optotrak marker. Before every single trial, the unit was quickly moved upwards. During data analysis, peak vertical accelerations determined from position data and from the fourth sensor node were synchronized.

2.4. Data analysis

A quaternion-based solution was used to calculate sensor orientations and accelerations in a global reference frame [9]. Thereafter, Matlab (The Mathworks, Inc.; version 7.1) was used for all further data processing. After synchronizing the force and sensor-data, force-plate-data were sub-sampled to a sample-rate of 50 Hz. Subsequently, both data sets were low-pass filtered with a cut-off frequency of 3 Hz. In addition to the three sensor locations on the trunk, CoM accelerations were approximated by calculating a weighted average based on segment weights [12] of accelerations ($\mathbf{a}$) measured at the right side of the pelvis and at the sternum:

$$\mathbf{a}_{\text{com}} = 0.603 \times \mathbf{a}_{\text{pelvis}} + 0.397 \times \mathbf{a}_{\text{sternum}} \tag{1}$$

Vertical power for lifting the body's CoM during STS was calculated from force-plate-data and based on the vertical accelerations at the different trunk locations. Subsequent data analyses focussed on the comparison of data at each trunk location to vertical accelerations and power as obtained from the force-plate-data. Measured position data were only used to verify trunk movements and sensor signals. These data are not reported.

2.4.1. Power calculations

For every STS trial, vertical power (P_y) was calculated by multiplying the vertical force acting on the body's CoM ($F_{\text{y-com}}$) with vertical velocity of the CoM ($\mathbf{v}_{\text{y-com}}$):

$$P_y = F_{\text{y-com}} \times v_{\text{y-com}} \tag{2}$$

2.4.1.1. Power calculated from force-plate-data. In the force-based method, vertical force (F_y) was calculated as the summed vertical force components of the ground reaction forces measured by the two force-plates. Vertical acceleration of the body's CoM ($\mathbf{a}_{\text{y-com}}$) was calculated by dividing F_y by body mass (m) and subtracting gravity:

$$\mathbf{a}_{\text{y-com}} = \frac{F_y}{m} - 9.81 \tag{3}$$

Vertical velocity of the CoM ($\mathbf{v}_{\text{y-com}}$) was calculated by a numerical integration of $\mathbf{a}_{\text{y-com}}$. An algorithm was used to determine the start and end of the numerical integration as indicated in Fig. 1(left). This start and end definition was also used for determining STS movement duration (cf. [13]).

2.4.1.2. Power calculated from sensor-data. In the sensor-based method, vertical force was determined by multiplying the vertical acceleration obtained from the sensors by body mass. A numerical integration of vertical acceleration produced vertical velocity. Similar to the analyses of force-plate-data, an algorithm was used to determine start and end of the numerical integration from vertical accelerations at the different trunk locations and the estimated CoM (see Fig. 1(right)).

2.4.2. Comparison of data based on force-plates and body-fixed-sensors

Since vertical accelerations are the sole inputs into the calculations of both the sensor-based and the force-based power calculations, vertical accelerations as determined from force-plate and sensor-data were compared by calculating root-mean-square (RMS) values of differences and Pearson's correlation coefficients between the vertical acceleration determined from force-plates and the vertical accelerations from every sensor location. RMS values were normalized by dividing the RMS value for a given trial by the range of force-plate-based vertical accelerations during that trial. Peak power was determined for each available trial and Pearson's correlation coefficients were calculated for peak power values calculated from the force-based method and sensor-based methods for all STS conditions.

3. Results

To avoid that loss of data packets affected data analysis, trials were excluded for further analysis if data samples were missing.

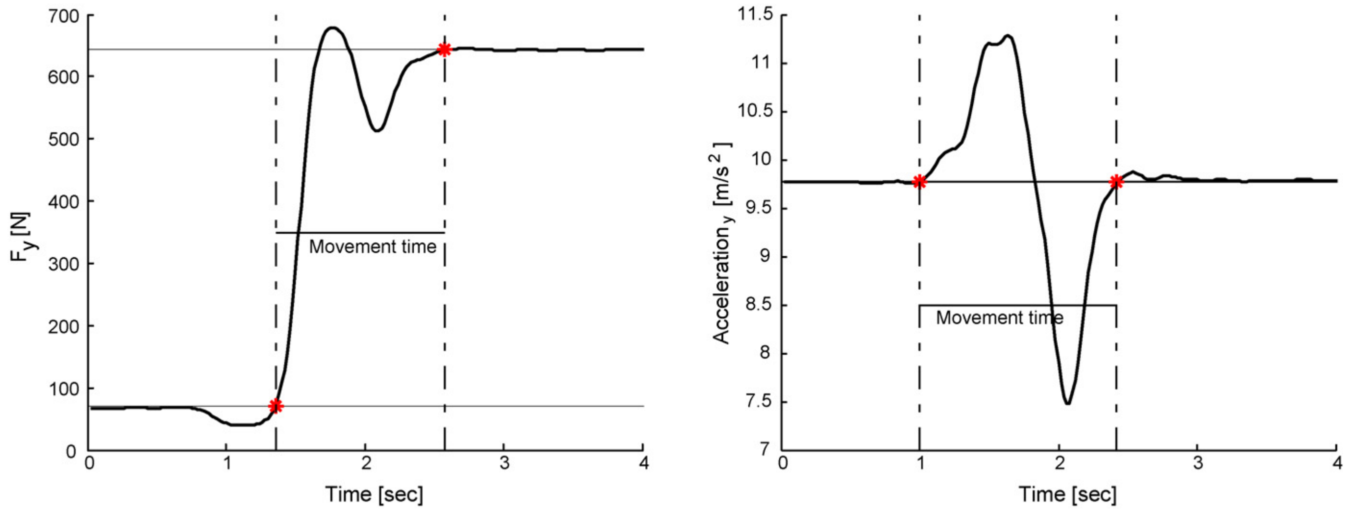

Fig. 1. Changes in the vertical ground reaction force measured by the force-plate under both feet (left) and vertical accelerations (right) determined from a wireless sensor node on the right side of the pelvis. The asterisk symbols indicate start and end definitions of the numerical integration for calculating power from force-plate-data (left), and from sensor-data (right). The durations between start and end indicate how STS movement duration can be determined from either force-plate or sensor-data.

The remaining trials comprised 16 slow, 20 natural, 22 fast and 20 STS movements with use of arm rest in the young subjects, and 32 STS movements at a natural speed in the older subjects. Without explicit instructions the young subjects never used an arm rest. In the older subjects, only one subject used arm rests in 2 out of 3 STS measurements.

3.1. STS performance

Table 1 presents STS durations and peak power. In the young subjects, duration and power vary with the instruction to perform the STS transfer at a slow, natural or fast speed. The use of arm rests while standing up resulted in a mean duration and power which are intermediate to STS performance at slow and natural speeds. STS durations in the older subjects are similar to those of the young subjects at a natural speed, peak powers are somewhat lower.

3.2. Comparison of vertical accelerations

Fig. 2 shows a representative example of vertical accelerations during STS. Typically, the closest correspondence was found between the force-based vertical acceleration and accelerations at the pelvis and the estimated CoM. Data measured at the sternum usually showed a first downward acceleration followed by a systematic overshoot in peak acceleration values, while the vertical accelerations of the SIPS usually started earlier than at the other locations.

Table 1 presents RMS values of differences in vertical accelerations for all available STS trials and correlation coefficients.

Table 1
Sit-to-stand (STS) durations and peak powers as determined from force-plate-data. Numbers indicate mean, standard deviation and range in young and older subjects. Data comparison per sensor location contrasts the results of the force-based method and sensor-based data at different trunk locations. For each STS condition, normalized root-mean-square values of differences in vertical accelerations (RMS-ACC), Pearson's correlation coefficients between vertical accelerations (R-ACC), and Pearson's correlation coefficients between vertical peak power as calculated by the force- and sensor-based methods are presented (R-peakpower).

	STS variables		Data comparison per sensor location				
	STS duration (s)	Peak power (W)	Sternum	Pelvis	SIPS	CoM	
STS in young subjects (n = 5)							
Natural	1.25 ± 0.25	504.6 ± 85.8	0.297 ± 0.039	0.106 ± 0.038	0.231 ± 0.065	0.115 ± 0.030	RMS-ACC
	(0.84–1.68)	(352.5–679.9)	0.93 ± 0.054	0.93 ± 0.045	0.66 ± 0.165	0.98 ± 0.016	R-ACC
			0.85	0.82	0.65	0.94	R-peakpower
Slow	2.52 ± 0.73	247.6 ± 61.3	0.356 ± 0.056	0.168 ± 0.034	0.328 ± 0.143	0.128 ± 0.023	RMS-ACC
	(1.4–3.66)	(185.3–409.5)	0.86 ± 0.114	0.84 ± 0.081	0.30 ± 0.180	0.97 ± 0.021	R-ACC
			0.90	0.56	0.67	0.87	R-peakpower
Fast	0.88 ± 0.11	910.7 ± 254.5	0.246 ± 0.047	0.090 ± 0.033	0.168 ± 0.059	0.116 ± 0.026	RMS-ACC
	(0.74–1.12)	(559.0–1363.5)	0.96 ± 0.053	0.96 ± 0.033	0.79 ± 0.125	0.98 ± 0.023	R-ACC
			0.91	0.95	0.96	0.99	R-peakpower
Use of armrests	1.47 ± 0.40	441.6 ± 113.4	0.265 ± 0.056	0.134 ± 0.042	0.209 ± 0.064	0.130 ± 0.025	RMS-ACC
	(0.88–2.14)	(312.3–711.4)	0.92 ± 0.064	0.94 ± 0.033	0.79 ± 0.082	0.98 ± 0.016	R-ACC
			0.90	0.94	0.92	0.99	R-peakpower
Over all conditions			0.286 ± 0.063	0.122 ± 0.047	0.227 ± 0.101	0.122 ± 0.027	RMS-ACC
			0.92 ± 0.078	0.92 ± 0.067	0.66 ± 0.236	0.98 ± 0.020	R-ACC
			0.97	0.96	0.92	0.99	R-peakpower
STS in older subjects (n = 12)							
Natural	1.28 ± 0.21	439.5 ± 158.8	0.281 ± 0.054	0.095 ± 0.023	0.229 ± 0.063	0.122 ± 0.034	RMS-ACC
	(0.86–1.84)	(209.0–872.6)	0.95 ± 0.038	0.96 ± 0.023	0.65 ± 0.164	0.99 ± 0.010	R-ACC
			0.93	0.94	0.96	0.94	R-peakpower

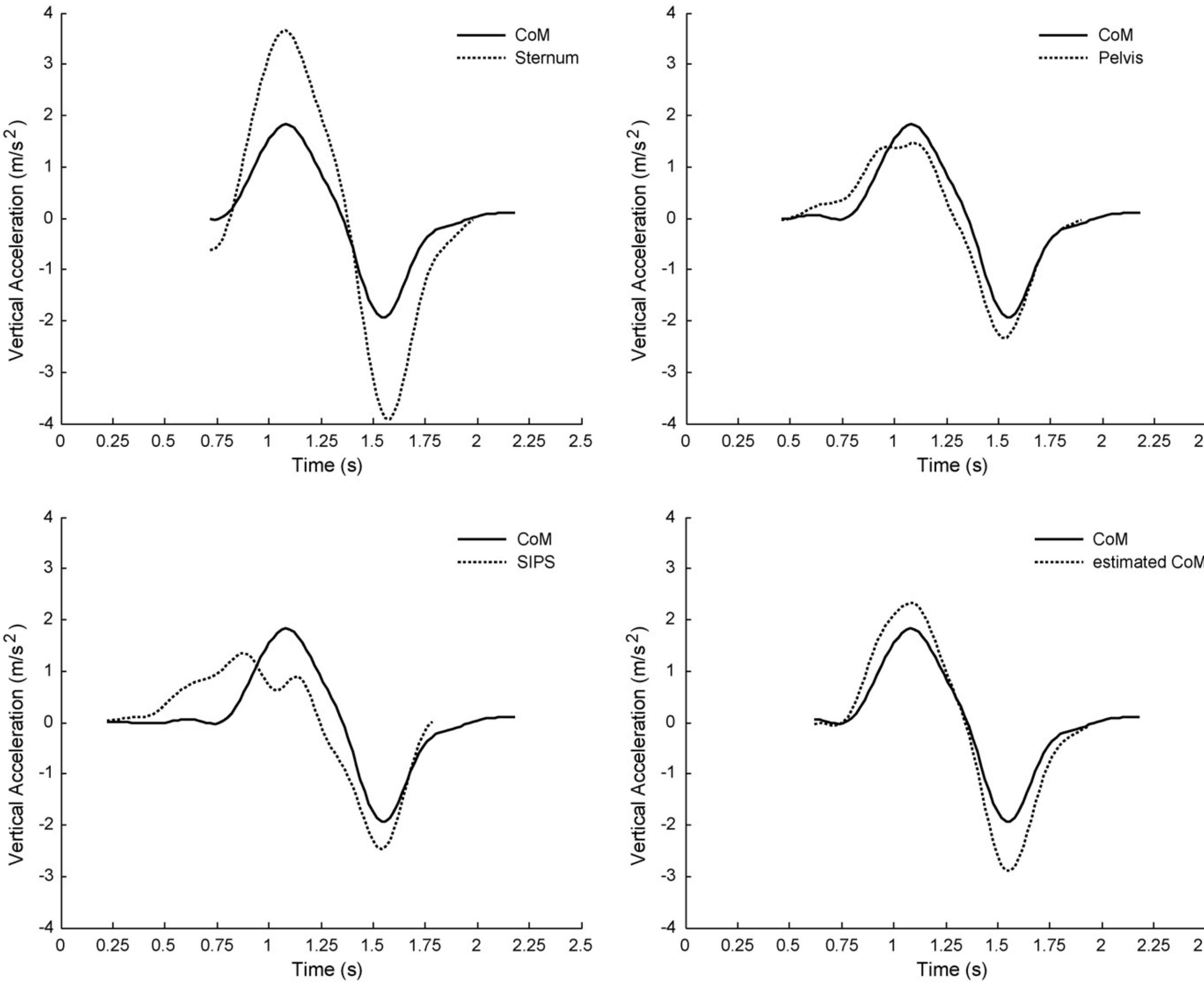

Fig. 2. Representative data traces of vertical accelerations of the centre of mass (CoM), as determined by the force-based method, three sensor locations at the trunk (i.e. sternum, pelvis, and SIPS), and the estimated vertical acceleration of the CoM. Data were collected from a young female subject while rising from the chair at a self-chosen natural speed.

The data reveal that vertical accelerations from the sensor node attached at the right side of the pelvis and the estimated CoM accelerations have much lower RMS values than acceleration data at the SIPS and Sternum. In all conditions, estimated CoM accelerations had the highest correlation to vertical accelerations from the force-based method.

3.3. Comparison of vertical power

Fig. 3 shows data of the same subject and STS instruction as in Fig. 2. The power curves show a pattern which corresponds to the observations in the vertical acceleration data. The sternum location overestimates vertical power, whereas pelvis and estimated CoM generally show good correspondence. For each STS performance by young and older subjects, Table 1 presents Pearson's correlation coefficients between peak power values as determined by the different methods. Fig. 4 presents a graphical representation of all individual peak power values. For the data of young subjects performing slow, normal or fast STS movements, the coefficients of determination (R^2) of linear fits between estimated peak power and peak power from force-plates ranged between 0.842 and 0.945 for the different sensor locations. The best approximation of peak power (R^2 = 0.984) was obtained at the estimated CoM.

4. Discussion

This study analyzed power exertion during STS based on motion sensors. To this purpose, young subjects were asked to perform different STS movements, and healthy older (70+) subjects were asked to stand up from a chair at a preferred speed. The instructions to the young subjects lead to a marked range of STS performances; mean movement durations varied between 0.88 and 2.52 s. Mean STS duration of the older subjects was close to mean STS duration of young subjects at their preferred speed, i.e. 1.28 s versus 1.25 s. The STS data demonstrate a wide range of CoM accelerations and peak power values; in the younger subjects peak powers ranged between 185 and 1364 W, in the older subjects this range was 209–873 W. The latter range includes the mean power values that have been reported for healthy older people [13]. All data of older subjects fitted very well with those of young subjects performing different STS movements.

4.1. Correspondence between sensor-based and force-based STS data

The results demonstrate fair to excellent linear relationships between sensor-based peak power and force-based peak power. Although the data do demonstrate that the best results were

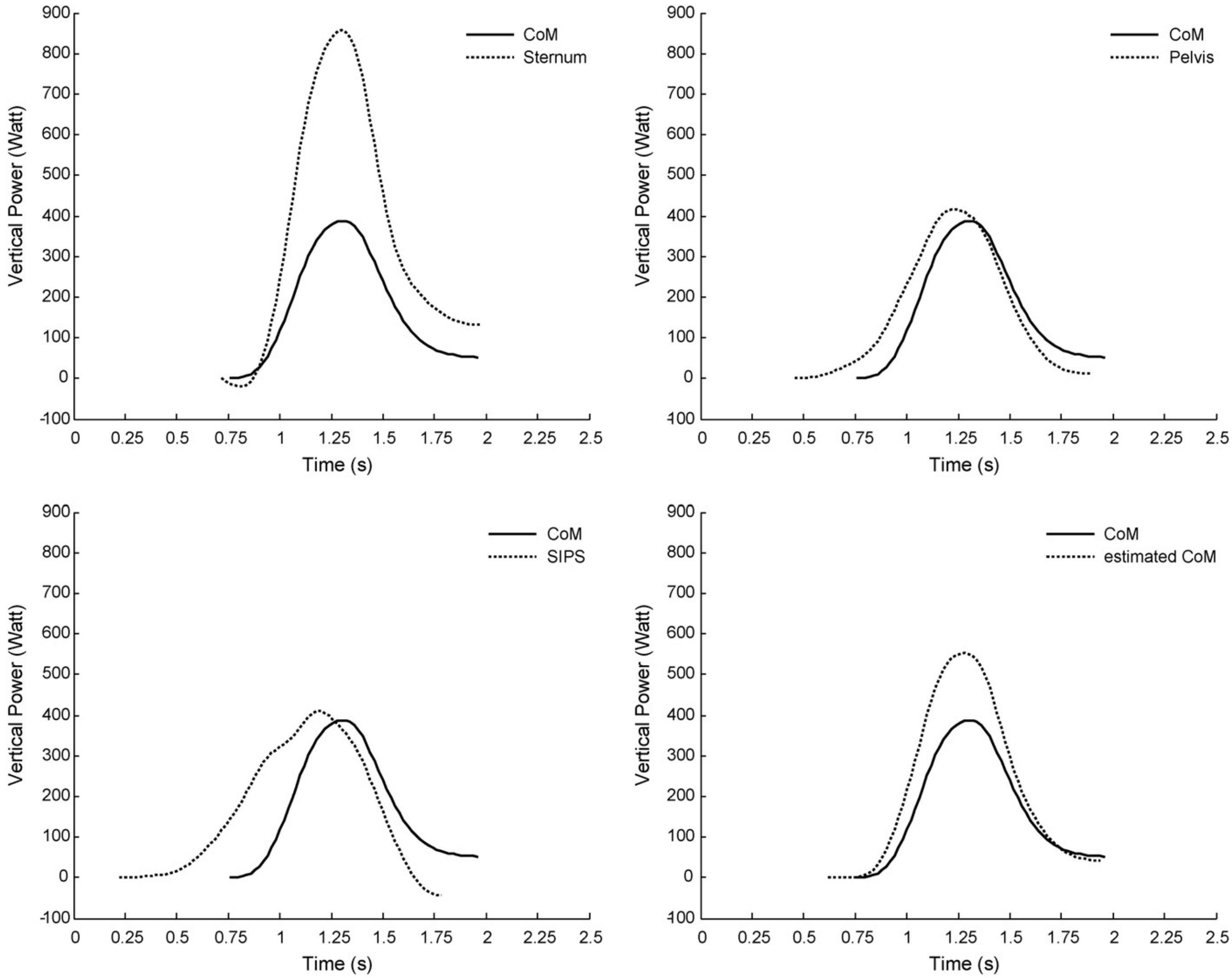

Fig. 3. Representative data traces of vertical power during rising from a chair. The presented power traces were based on the vertical accelerations determined by the force-based method (CoM), the vertical accelerations at a sensor location at the trunk (i.e. sternum, pelvis, or SIPS), or the estimated vertical acceleration of the CoM. Data are from the same subject as in Fig. 2.

obtained by using a weighted combination of acceleration data, peak power data obtained from each of the three sensor locations showed a good linear fit to peak power values as calculated based on a reference method using force-plates. The fact that different sensor locations produced different results can be understood based on the trunk kinematics during the STS movement.

Before initiating an upward movement of the CoM during STS, the trunk is moved forward by a flexion in the hips. Thus, the sternum typically moves forward and downward before an overall upward movement is initiated. In effect this leads to an early downward movement of the sternum which precedes a rising phase in which the upper trunk shows an upward movement which corresponds both to the upward CoM movement and to the hip extension movement which is needed to reach an upright end position of the trunk. Obviously, this typical movement pattern increases the vertical accelerations at the upper trunk, and therefore accelerations at the sternum are substantially larger than at the CoM. As a consequence, vertical CoM accelerations and peak power are consistently overestimated by a motion sensor at the sternum.

Data of motion sensors at pelvis and SIPS did not substantially overestimate the vertical CoM accelerations. However, since the initial upward accelerations at the SIPS usually preceded those of the CoM, differences between SIPS and CoM acceleration patterns were evident. Thus, also the estimated peak powers showed somewhat less correspondence to the reference data than the other trunk locations. The early upward accelerations measured at the SIPS may also be caused by the initial forward rotation of trunk; unlike the upper trunk, the lower trunk often starts moving from a backward lean, and thus produces a movement from lumbar kyphosis to lordosis. The latter leads to an immediate upward movement of the lower trunk during the forward movement of the trunk.

Accelerations and peak power as determined from the sensor at the right side of the pelvis showed the best correspondence with CoM accelerations and force-based peak powers. Since the pelvis sensor was placed close to the trochanter major femoris, data measured at this location suffered less from the initial forward movement of upper and lower trunk segments. Therefore, the upward accelerations measured at the pelvis showed a good correspondence to vertical CoM accelerations. The latter correspondence resulted in the consistently lowest RMS values of the three trunk locations and an excellent relation between estimated and real peak powers.

Although more fine-tuning would have been possible, the rough approximation of accelerations at a virtual CoM position by a

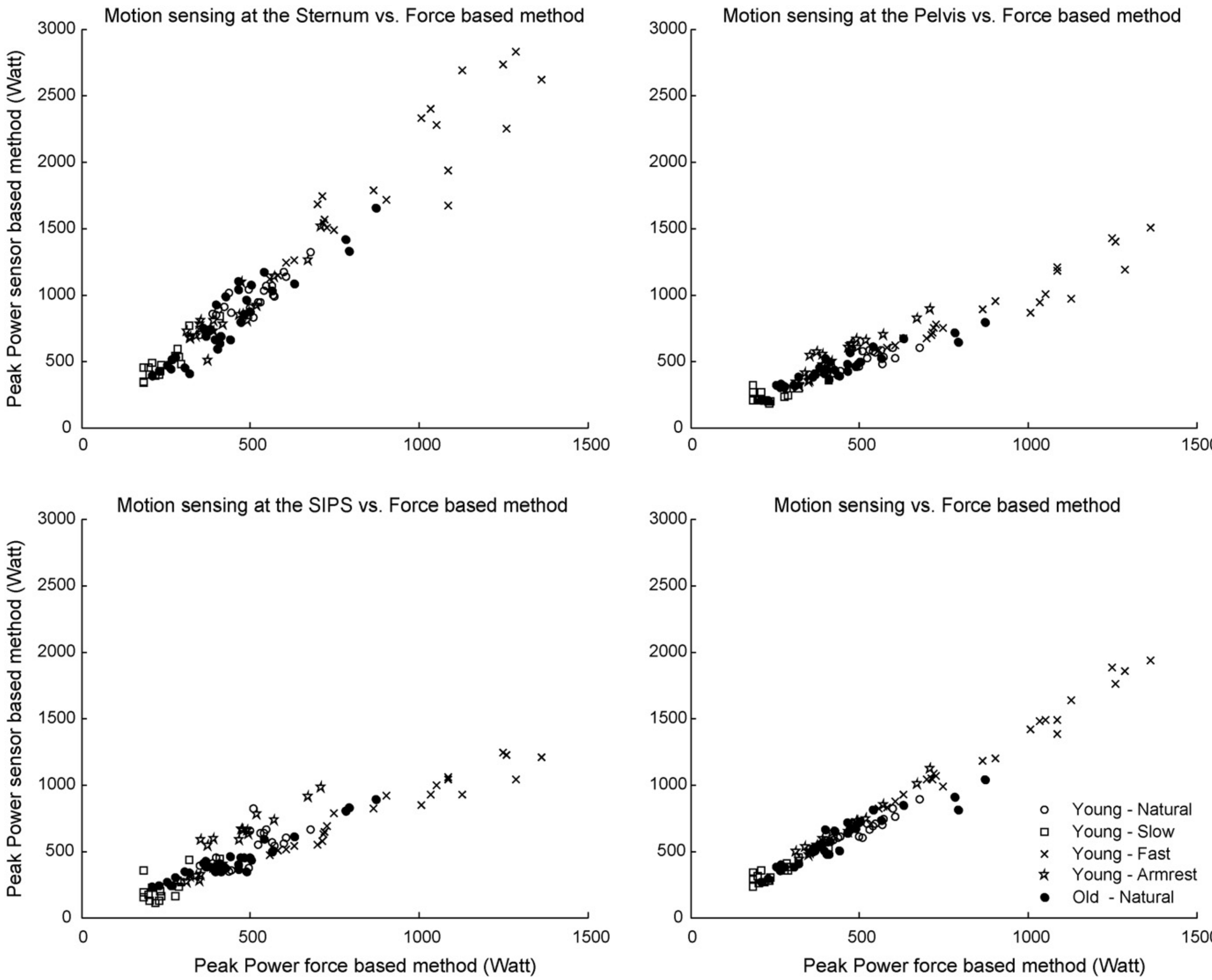

Fig. 4. Peak power values as calculated from force-plate-data and sensor-data. Each symbol represents the peak power for one specific trial. Young adults ($n = 5$) were given different instructions for performing the STS movement (see inserted legends and see text). Data of older adults ($n = 12$) represents STS performance at a preferred speed. The four plots indicate the data determined from the three sensor locations, and the estimated vertical accelerations of the CoM (lower right figure).

weighting of acceleration data of sternum and pelvis showed to produce remarkably consistent results in all STS conditions. In fact, despite the inclusion of accelerations at the sternum, the RMS values remained very close to those for the accelerations at the pelvis.

4.2. Implications for practical use of sensor-based estimations of power

Although better approximations of power curves, and possibly also peak power, would have been obtained if the force-based starts and ends of STS had been used for the numerical integration of the sensor-based accelerations, we deliberately did not use information obtained from force or position data in our sensor-based approaches. Thus, the presented power results are exactly as one would have obtained when the methods had been applied outside a laboratory.

The results demonstrate that all sensor locations can be used for estimating peak power during STS. Despite differences in goodness-of-fit, data from all three locations show strong linear relationships with peak power as determined from force-plate-data. Thus, predictions of real peak power are possible. For a single-sensor approach, a sensor location near the trochanter major femoris seems optimal for estimating vertical power. However, for other purposes, e.g. activity monitoring or gait analyses, this sensor location may not be optimal. If a multi-sensor approach is feasible, the combination of a sternum sensor and a sensor near the trochanter, or on the SIPS, can be used to estimate power. The former location can be the basis for activity monitoring [14] and the latter location allows for basic balance and gait assessments (e.g. [15,16]).

For interpreting the power estimates, it is important to realise that the sensor-based approaches do not give information about how the power to lift the CoM during an STS movement was produced. Whereas under controlled conditions, the power estimates can be used as a means to quantify the contribution of leg muscle activity to STS, our results from the condition where subjects used both armrests shows that similar STS movement durations and peak powers are obtained. This fact may present some limitation in applying the method and interpreting the data. However, under controlled conditions, e.g. STS assessment in the presence of an instructor, this limitation does not play a role.

Our results demonstrate that the vertical power for lifting the body's CoM from a sitting to a standing position can be estimated based on motion sensors on the trunk. The presented approach for assessing power is of high relevance for functional assessment of mobility in older people. Earlier studies have demonstrated that with ageing muscle strength and power decline [1,17]. It seems that after a certain threshold, the loss of muscle function leads to functional limitations and is highly predictive of falls [3]. Thus, the presented approach is relevant for monitoring fall risk and/or evaluation of the effects of interventions aiming to improve muscle function, balance and mobility. Since the approach to estimate power does not depend on one sensor location, a number of existing methods [5] for mobility assessment and/or activity monitoring may be extended to also include analysis of power exertion during movements in daily life. Similar approaches may be developed for other mobility related activities (such as stair walking) or sports related activities such as jumping. Thus, the presented approach may be an important contribution to monitoring the effects of exercise-based interventions on functioning in older people as well as in sportsmen.

Conflict of interest statement

The authors declare that they have no conflicts of interest.

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