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Medio-lateral stability of sit-to-walk performance in older individuals with and without fear of falling

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ABSTRACT

Most falls in older people are due to loss of balance during everyday locomotion, e.g., when initiating walking from sitting; sit-to-walk (STW). It has been considered that the broader stride width in walking that is seen in many people with fear of falling (FoF) does not increase stability, but could be predictive of future falls because of increased medio-lateral (ML) velocity of the body centre of mass (CoM). This study was aimed to examine step-, velocity- and stability-related parameters, focusing on ML stability, in STW performance of people with and without FoF. Ten subjects with FoF and 10 matched controls, aged $\geq$70 years, were included. Kinematic and kinetic data were collected in a laboratory. Stability parameters were calculated from a formula implying that the vertical projection of the CoM extrapolated by adding its velocity times a factor $\sqrt{l/g}$ (height of inverted pendulum divided by gravity) should fall within the base of support (BoS). A related spatial margin of stability (SMoS), defined as the minimum distance from the extrapolated CoM (XCoM) to the boundaries of the BoS, was also calculated. In the phase 'seat-off–second-toe-off', the FoF group had significantly ($p < 0.05$) shorter and broader steps, lower forward but similar ML CoM velocity, and broader CoM and XCoM widths. The FoF group therefore exhibited a disproportionately large sideways velocity compared to the controls. This indicates that STW may be a hazardous transfer for older people with FoF, which should be relevant in assessment and training aimed at preventing falls.

1. Introduction

Most falls in older people occur during everyday locomotion [1,2], such as standing up from sitting (i.e. sit-to-stand, STS) and walking [3,4]. Considering how, in activities of daily living (ADL), STS is rarely aimed at standing alone, these two tasks are commonly merged into the sit-to-walk (STW) transfer [5]. Compared to STS, the STW performance is more heavily dependent on balance control, since it requires a greater propulsive impulse generating a higher horizontal momentum [6,7]. The STW is also characterised by an ongoing forward body transfer while initiating walking before reaching the fully upright position [8,9]. The phase of STW, from 'seat-off' to establishment of walking, is particularly critical, since the centre of mass (CoM) is then transferred forward/upward simultaneously as the base of support (BoS) is narrowed [8,10]. People with conditions such as Parkinson's disease [11] or stroke [9,12] may have difficulties in merging the two tasks of STS and walking with a fluid motor strategy. Healthy older adults also tend to use less fluid STW strategies, since these persons generate minor horizontal CoM momentum prior to rising up, and delay walking until standing more upright [13,14]. Reasons for these phenomena may be limitations of strength or ranges of motion or unwillingness to generate the required momentum, because of prioritisation of stability before mobility [13,15–17]. Additionally, it has been suggested that older people who show hesitancy and therefore reduced fluidity during the STW movement are also at increased risk of falling [16,18].

Older people who experience fear of falling (FoF) constitute a large proportion of the elderly population exhibiting an increased risk of falling during ADL [19]. The prevalence of FoF among older people with frail health is reported to be at least 50% [20,21]. Moreover, research aimed at determining the temporal relationship between falls and FoF indicates that individuals who develop one of these outcomes are at risk of developing the other, with a resultant spiralling of FoF, risk of falls, and functional decline [22]. People who express FoF, compared to other people in the same age group, have more functional limitations, a decreased quality of life [20,23], and have poorer balance-test performance [24].

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Some gait adaptations made secondary to FoF, i.e. a decreased step length and speed and prolonged double-support, are interpreted as attempts at stabilisation to reduce the risk of falling. Conversely, a broader step width does not necessarily increase stability but can predict future falls [25]. On the other hand, recent research [26] has indicated that side-fallers have narrower step widths than fallers in other directions, and that they may not adapt their gait to compensate for medio-lateral (ML) instability. Accordingly, it is considered that a wide step width cannot alone differentiate between people with and without balance impairments [27]. A more lateral foot placement may have the effect of exacerbating instability by increasing ML acceleration of the total body CoM [25,28]. The increased ML head motion, associated with a broad step, compromises balance by disturbing the stabilisation of the visual field [25]. Even so, the mechanisms underlying the relations between different step parameters and dynamic ML stability are poorly understood.

To address the questions as to what conditions should be fulfilled for balance to be maintained, and how good is balance in a certain situation, a new method that takes into account the dynamics of the inverted pendulum model has been suggested [29]. This extension of the standard inverted pendulum model implies that the vertical projection of the CoM extrapolated by adding its velocity times a factor $\sqrt{l/g}$ should be within the BoS. Here l corresponds to the height of the inverted pendulum and g to the acceleration of gravity [29]. This spatial variable is termed 'extrapolated centre of mass' (XCoM), as the CoM trajectory is extrapolated in the direction of its velocity. From this, a related spatial margin of stability (SMoS) is defined as the minimum distance from XCoM to the boundaries of the BoS. This model provides an opportunity to investigate how step width and the velocity and direction of CoM influence the margin of stability in mobility tasks related to ADL.

In this study we tested the hypothesis that performance of STW jeopardises ML stability in older people with FoF, and that this is detectable through application of XCoM and related stability parameters. For this purpose, we examined STW-related step, velocity and stability parameters, focusing on ML stability, among subjects with and without FoF.

2. Methods

2.1. Participants

Ten participants with FoF and 10 controls were recruited by co-operation with senior citizens' organisations and staff at municipal elderly care centres in a city in central Sweden. Inclusion criteria were: age ≥70 years, independence in personal everyday activity (PADL), and ability to walk ≥10 m without assistance. Fear of falling was identified as expressed experience of ADL-related FoF and function-related insecurity in at least two variables assessed by the insecurity subscale of the General Motor Function assessment scale (GMF) [30]. The GMF/insecurity scale allows a performance-based assessment of 11 mobility functions (see Appendix A) that are considered to influence the ability for ADL. The subjects carried out these functions one by one, while they were asked a "yes-or-no" question about insecurity caused by the execution of the task in question. The scoring of the scale is constructed as sum-scores, i.e. 11 implies that the subject experiences insecurity when executing all mobility functions (see Appendix A). More detailed descriptions of the GMF are provided elsewhere [31,32]. Controls, matched for age, gender, weight, and height, should not display any FoF or any function-related insecurity, according to the GMF.

Exclusion criteria for all participants were as follows: use of neuroleptics, obvious cognitive impairment, neurological diagnoses, and function-related pain (>2 points according to the pain subscale of the mobility variables of the GMF).

2.2. Data collection and processing

A test protocol was used to register the participants' medication, history of falls and illness, weight, height, and other anthropometric measures. Additionally, clinical assessments were used, i.e. function-related help dependence assessed by the GMF [30], timed up-and-go (TUG) [33], standing balance [34], 10 m timed walk [35], and the Swedish version of the Falls-Efficacy Scale (FES) [36].

The laboratory set-up and the basic data processing used correspond to a recent study by our research group [10]. Synchronised kinematic and kinetic were collected with a sample frequency of 100 Hz. For the kinematics, multiple body segment motions were recorded with an optoelectronic eight-camera motion capture system (ELITE 2000, BTS, Milan, Italy). Thirty-eight markers (Fig. 1) were used, to allow construction of a biomechanical model consisting of 13 segments: one head/neck; two upper trunk; three pelvis; 4–5 each upper arm; 6–7 each forearm; 8–9 each thigh; 10–11 each shank; and 12–13 each foot. Six extra markers, positioned medially at the elbow, knee and ankle joints, were used during reference measurements. To ensure good reliability [37], one person attached all the markers in all of the experiments. Surface forces were measured and registered by four AMTI force plates (Model MC818-6-1), and were used for identification of the seat-off and first- and second-toe-off events.

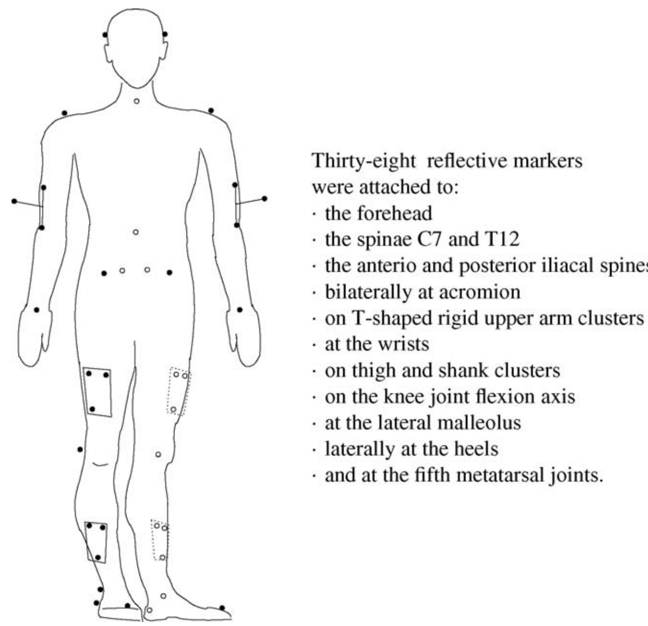

Fig. 1. Overview of the marker placement.

The participants wore tight sport shorts, a bra (in women), and ordinary walking shoes. Initially, they were sitting on a platform with a height of 0.45 m, with one force plate beneath each buttock and one beneath each foot. To avoid marker occlusion at the start, the arms were held out from the body and the feet were positioned slightly forward. An audio (telephone) signal was given and the subject was instructed to rise and walk 3 m straight ahead in the direction of the telephone. They were free to reposition their feet and use their arms if necessary. All participants conducted 10 STW trials at a self-selected speed, of which 5–6 (with the best data quality) per person were used for further analysis.

Markers were identified with the Tracklab, Biomech program. Marker trajectories were interpolated, when necessary, in a separate program (Qualisys Track Manager, QTM), and subsequently, data were exported to a biomechanics analysis programme (Visual 3D). A reference file, required for further data processing in Visual 3D, was created from a reference measure in a neutral standing position. Anatomical frames for each of the 13 segments were defined from the position of the markers in the reference standing trial and from anthropometrical measures. This model was used for data analysis of the kinematics and for calculating of the total CoM, which was obtained as a weighted sum of the CoM of each of the segments (calculated on the basis of Dempster's table for the relative mass and location of CoM in a body segment). Calculations of CoM from kinematics, compared to double-integrated horizontal ground reaction forces, show intra-class correlation factors between 0.92 and 0.99, indicating high validity of this method [38]. The CoM position and velocity were computed on the basis of filtered marker trajectories, for which a fourth order, zero phase-shift, Butterworth filter, with a cut off frequency of 6 Hz, was used.

Some anthropometric measurements were used to facilitate definition of the sizes of the body segments represented by the biomechanical model. These were (i) proximal circumference of the thigh, (ii) length of upper arm, (iii) distance between the left and right anterior superior iliac spine, and (iv) shank length.

Virtual markers, representing the lateral boundary of each foot, were defined from the positions of the heel and toe markers at the fifth metatarsal joint. The purpose was to correct for the diameter of the real markers, which places them at a distance slightly greater than their radius lateral to the boundary of the foot.

The following events were determined from the force platforms' vertical components: (1) Seat-off by the first time frame with no force (and no recurring force, i.e. retry) beneath the buttocks. (2) First-toe-off by the first time frame after seat-off with no force beneath either the left or the right foot. (3) Second-toe-off by