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A flexed posture in elderly patients is associated with impairments in postural control during walking

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ABSTRACT

A flexed posture (FP) is characterized by protrusion of the head and an increased thoracic kyphosis (TK), which may be caused by osteoporotic vertebral fractures (VFs). These impairments may affect motor function, and consequently increase the risk of falling and fractures. The aim of the current study was therefore to examine postural control during walking in elderly patients with FP, and to investigate the relationship with geriatric phenomena that may cause FP, such as increased TK, VFs, frailty, polypharmacy and cognitive impairments. Fifty-six elderly patients (aged 80 ± 5.2 years; 70% female) walked 160 m at self-selected speed while trunk accelerations were recorded. Walking speed, mean stride time and coefficient of variation (CV) of stride time were recorded. In addition, postural control during walking was quantified by time-dependent variability measures derived from the theory of stochastic dynamics, indicating smoothness, degree of predictability, and local stability of trunk acceleration patterns. Twenty-five patients (45%) had FP and demonstrated a more variable and less structured gait pattern, and a more irregular trunk acceleration pattern than patients with normal posture. FP was significantly associated with an increased TK, but not with other geriatric phenomena. An increased TK may bring the body's centre of mass forward, which requires correcting responses, and reduces the ability to respond on perturbation, which was reflected by higher variation in the gait pattern in FP-patients. Impairments in postural control during walking are a major risk factor for falling: the results indicate that patients with FP have impaired postural control during walking and might therefore be at increased risk of falling.

1. Introduction

A flexed posture (FP) is characterized by an increased thoracic kyphosis (TK), protrusion of the head and, in severe cases, knee flexion [1], which is a postural correction to the increased TK [2]. In the elderly, TK is likely to increase over time [3] because of intervertebral disc deformities and/or spinal extensor muscle weakness [4,5]. Vertebral fractures (VFs) are characteristic of osteoporosis: VFs in the thoracic vertebral column may also increase TK. These impairments may affect motor function, thus increasing the risk of falling [6] and fractures.

A recent review showed significant differences in postural control during standing and walking between patients with osteoporosis and healthy controls, particularly when variables indicating postural stability were calculated from objective measurements using instrumented devices like force plates and accelerometers [7]. In the majority of studies reviewed, however, the presence and severity of FP, TK and/or VFs in the osteoporotic group were not specified, and the relationship between these clinical entities is not yet clear.

Impaired postural control during walking is a major risk factor for falls and new fractures in the elderly [8]; therefore, early recognition and quantification of balance disorders is important in osteoporotic patients. Analyses of time-dependent variability, using measures derived from the theory of stochastic dynamics [9], enable differences in postural control during walking to be detected between young and old patients, fallers and non-fallers,

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and those with and without cognitive impairments [9–11]. Measures to quantify time-dependent variations of postural control during walking, such as the detrended fluctuations analyses [12], sample entropy [13] and maximal Lyapunov exponents [11], were used in the current study as well as more conventional gait parameters (e.g. average gait speed, and stride times). In conventional measures, each gait cycle is treated as an independent event unrelated to previous or subsequent strides, whereas the methods used in the current study assess fluctuations throughout the gait cycle, thereby providing greater insight into movement behaviour.

The primary aim of the present study was to examine postural control during walking in elderly patients with FP, and the secondary aim was to examine the relationship with TK, VFs and grip strength (as an indicator for overall limb strength) as possible causes of FP [1,4,5]. Comorbid diseases, frailty, polypharmacy and cognitive impairments are often present in the elderly, so the association of these geriatric phenomena with FP was also examined. Patients with FP are hypothesized to have increased variability of gait parameters compared with patients with normal posture. Also, the presence of increased TK, VFs, muscle weakness and other geriatric phenomena might further worsen FP, and consequently worsen postural control during walking.

2. Methods

2.1. Participants

Patients were recruited among the elderly who visited the Diagnostic Geriatric Day Clinic at the Slotervaart Hospital in Amsterdam. Patients aged at least 70 years who could walk safely for 3 minutes without assistance were included in the study. Patients who had any asymmetric mobility problems and/or did not understand the researcher's instructions were excluded from the study.

The study was approved by the Medical Ethical Committee of the Slotervaart Hospital. All included patients gave their informed consent.

2.2. Gait analysis

Patients walked about 160 m at a self-selected speed in a well-lit, 80-m-long hallway. Walking time was recorded to determine gait speed. Trunk accelerations were measured with a tri-axial accelerometer (DynaPort Minimod Hybrid, McRoberts BV, The Hague, the Netherlands; sample frequency 100 Hz) attached with a band at the level of the lumbar vertebral column.

2.2.1. Stride-related parameters

Medio-lateral (ML) and anterior-posterior (AP) trunk acceleration signals were analysed using custom-made software in MATLAB (The MathWorks, Inc., Natick, MA, USA). Signals were corrected for horizontal tilt, and high-pass filtered using a Butterworth filter (4th order; cut-off frequency 0.25 Hz). Foot contacts were determined from the peaks of the AP-acceleration time-series. A median filter was used to exclude outliers in the data due to turning points in the gait assessment. Foot contact data were used to calculate stride times, which were defined as the time interval between two ipsilateral foot contacts. Mean stride time, coefficient of variation (CV) of stride time, and stride frequency were calculated for each patient.

Further parameters were calculated to assess variations throughout the gait cycle. Temporal variability was quantified by the variance of the relative timing between sequential ipsilateral foot contacts using the point estimate of the relative phase: $\varphi_i = (\mathrm{FCR}_{t(i)} - \mathrm{FCL}_{t(i)})/(\mathrm{FCL}_{t(i+1)} - \mathrm{FCL}_{t(i)}) \times 360^\circ$ [14], where FCL and FCR are the left and right foot contacts at time $t(i)$, respectively. The relative phase is a circular measure; therefore, circular statistics were applied to calculate the mean and variance of the relative phase over strides [15]. A temporally symmetrical gait pattern is denoted by $\varphi_i = 180^\circ$; a higher variance indicates a more variable gait pattern.

In addition, long-range correlations in stride time intervals were quantified by calculating scaling exponent α using the detrended fluctuation analysis (DFA) [12]. When $0.5 \geq \alpha \geq 1$, this indicates the presence of long-term correlations in the signal, which means future fluctuations are better predicted by past fluctuations. Therefore, α values closer to 1 represent a more structured pattern.

2.2.2. Trunk movement patterns

The magnitude of ML and AP trunk acceleration patterns was quantified by calculating the root mean square (RMS). The harmonic ratio (Hratio), sample entropy (SEn) [13], and maximal Lyapunov exponent (λ_{max}) [16] were calculated using open source software (UPMOVE version 0.2a; http://www.upmove.org), indexing the smoothness, degree of predictability, and local stability of the trunk acceleration patterns, respectively.

The Hratio was calculated using spectral dynamics to quantify the smoothness of the ML and AP trunk movements, with a higher Hratio representing a smoother trunk acceleration pattern. A discrete Fourier transform was used to estimate the power spectral density of the fundamental oscillatory frequency and of the six consecutive harmonics. The Hratio was defined by dividing the power spectral density of the fundamental oscillatory frequency by that of the first seven harmonics (the first seven harmonics were chosen because no additional information was obtained from spectral analysis of higher frequencies after low-pass filtering the data at 10 Hz).

The degree of predictability in ML and AP acceleration time-series was assessed by calculating the SEn, which is defined as the negative natural logarithm of an estimate of the conditional probability of epochs of length m ($m = 3$ in this study) that match point-wise within a tolerance r and repeats itself for $m + 1$ points. An optimization approach [17] was used to determine the tolerance parameter r and m, since the choice of r for given m is decisive. Smaller SEn values indicate greater regularity; larger SEn values are associated with a small chance of similar data being repeated. The ML and AP acceleration data were normalized to unit variance, so the outcome was scale-independent.

Local stability of the ML and AP trunk acceleration patterns was expressed by the λ_{max}, which was calculated by applying the Wolf algorithm [16]: this algorithm is most appropriate to evaluate local dynamic stability from relatively small data sets. The time-series was first low-pass filtered using a least squares finite impulse response filter (6th order; cut-off frequency 10 Hz) [11]. All stride-cycles were then resampled to 100 samples to enable comparison of trials between patients with FP and those with normal posture on the same time scale. The estimated time interval was 10% of the stride cycle for all reconstructed state spaces. An embedded dimension of 5 was chosen, following previous studies [9]. Larger λ_{max} indicates greater sensitivity to local perturbations.

2.3. Additional measurements

Age, gender, body mass index (BMI) and number of prescriptions were recorded for each patient. FP was defined as an occiput-to-wall distance (OWD) of 5.0 cm or more [1]; OWD was measured while subjects stood with their head in a natural position with heels and back touching the wall and knees extended.