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Short communication

Local stability and kinematic variability in walking and pole walking at different speeds

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ABSTRACT

This study investigated the kinematic variability and the local stability of walking and pole walking using two tri-axial accelerometers placed on the seventh cervical (C7) and the second sacral (S2) vertebrae of twenty-one adults. Each participant performed three 1-min trials of walking and pole walking on a motorized treadmill (60, 80, 100% of the preferred walk-to-run transition speed). Forty strides per trial were used to calculate, in all directions of C7 and S2, the median of the stride-to-stride median absolute deviation (medMAD) and the local divergence exponent (λ). Generalised estimating equations and pairwise contrasts revealed, during pole walking, a higher medMAD (all directions, most speeds, C7 level only), and a lower λ (all directions, all speeds, both C7 and S2 level). As speed increased, so did medMAD (all directions, both walking with or without poles), with higher values at C7 compared to S2 level. A similar effect was observed for λ in the vertical direction (walking and pole walking), and in the anterior-posterior direction (only pole walking). An increase in speed brought about a λ reduction in the medial-lateral direction (C7 level only), especially during walking. Finally, both medMAD and λ were higher at C7 than S2 level (all directions, both walking and pole walking) except for λ in the anterior-posterior direction, which resulted higher in walking (C7 level only).

In conclusion, despite a higher kinematic variability, pole walking appears to be more locally stable than walking at any speed, especially at C7 level.

1. Introduction

During steady-state walking, even small movement perturbations can lead to falls if they are not attenuated by the control system [1]. Such perturbations are usually quantified by calculating the variability of kinematic parameters, with high values being related to an elevated risk of falls [2]. However, kinematic variability does not directly represent local stability, i.e. the ability to handle the movement perturbations usually occurring during walking [1]. This can be quantified by the local divergence exponent (λ) of kinematic parameters [3]. Nevertheless, kinematic variability and local stability reflect different aspects of walking stability and both concur in predicting the risk of falls [2].

Pole walking (PW), i.e. walking using of a pair of handheld poles, effectively increases body stability among hikers [4], thus possibly reducing falls, i.e. the major cause of muscle-skeletal injuries in this population [5]. However, the effect of using poles on walking stability has been investigated only during quiet standing [4], which is unlikely to be able to capture the dynamic nature of PW.

Accordingly, the present study aimed to measure the kinematic variability and local stability associated to both walking and PW performed at different speeds. This will contribute to determine whether the use of poles affects walking stability.

2. Methods

2.1. Participants

Twenty-one healthy adults (10 males and 11 females; age 30.9 ± 8.2 years; BMI 24.3 ± 3.2 kg m^{-2}) were enrolled in this study, which was approved by the Ethical Research Committee of the Sports, Health and Exercise Science Department of the University

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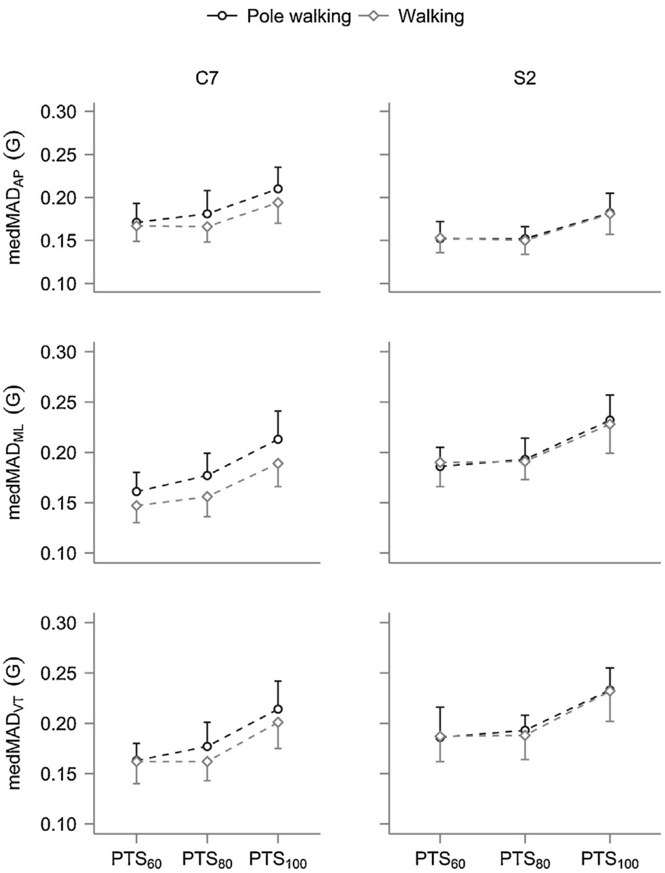

Fig. 1. Kinematic variability (medMAD) calculated in the anterior-posterior ($_{AP}$), medial-lateral ($_{ML}$) and vertical ($_{VT}$) directions. The accelerometers were placed at the level of the seventh cervical vertebrae (C7) and of the second sacral vertebrae (S2). Three different speeds are presented: 60% (PTS$_{60}$), 80% (PTS$_{80}$) and 100% (PTS$_{100}$) of the preferred transition speed during walking and pole walking.

of Hull (UK). The participants signed a written informed consent form before enrolment.

2.2. Experimental procedure and equipment setup

The participants visited the lab at least twice, 24 h apart. In the first visit, the preferred transition speed (PTS), i.e. the highest speed at which walking is preferred to running, was determined (mean ± standard deviation = 1.97 ± 0.17 m s^{-1}) on a motorized treadmill (Pulsar–h/p/cosmos Sports & Medical, Nussdorf-Traunstein, Germany) as described elsewhere [6]. The participants were then familiarised to PW and additional familiarization sessions were scheduled if needed to ensure comparable PW technique among them (see [6] for details). The length of the poles (Forclaz 500–Quechua, Passy, France) was adjusted to each participant's body size [7]. In the last visit, 2 tri-axial accelerometers (±6 G range; DTS–Noraxon USA Inc., Scottsdale, Arizona, USA) were secured over the seventh cervical (C7) and second sacrum (S2) spinous processes [8] and a heart rate monitor (RS800CX–Polar Electro Oy, Kempele, Finland) was worn by the participants. Participants sat for 5 min and the mean heart rate over the last minute was assumed to be the resting heart rate. Baseline accelerations were then collected while standing still for 30 s, prior to a 5-min warm-up period (PW at 60% of the PTS). Afterwards, six 1-min bouts of walking (with and without poles in random order) at 60, 80 and 100% of the PTS (PTS$_{60}$, PTS$_{80}$ and PTS$_{100}$, respectively) were performed with 1 min of rest in between. After the PTS$_{100}$ trials, the participants sat until the heart rate was steadily within the resting value ±10 bpm for 1 min.

2.3. Data collection, processing and parameters calculation

Acceleration data was collected synchronously at 1500 Hz and stored on a computer using a 16 bit resolution wireless system (Desktop DTS–Noraxon USA Inc., Scottsdale, Arizona, USA).

The baseline average roll and pitch angles of the acceleration data were calculated and removed [9]. Afterwards, the 40 central strides for each trial were detected [10].

Similar to Dingwell et al. [3], kinematic variability (medMAD) was calculated by: (a) cubic-spline interpolating the data of each stride to 101 points; (b) calculating the median absolute deviation between the strides for each of the 101 points; (c) computing the median value of the median absolute deviations. According to Bruijn et al. [11], λ was calculated by: (a) cubic-spline interpolating each signal to 4040 points (i.e. approximately 101 points per stride); (b) reconstructing a state space from each spatial direction using 5 embedding dimensions and 10 points of delay; (c) calculating the slope of the regression line fitting the stride time normalized curve of the mean logarithmic rate of divergence in neighbouring trajectories [12] over the 0–0.5 stride period.

Both medMAD and λ were calculated, in all directions, at both C7 and S2 levels, and for each participant and trial, using a custom-

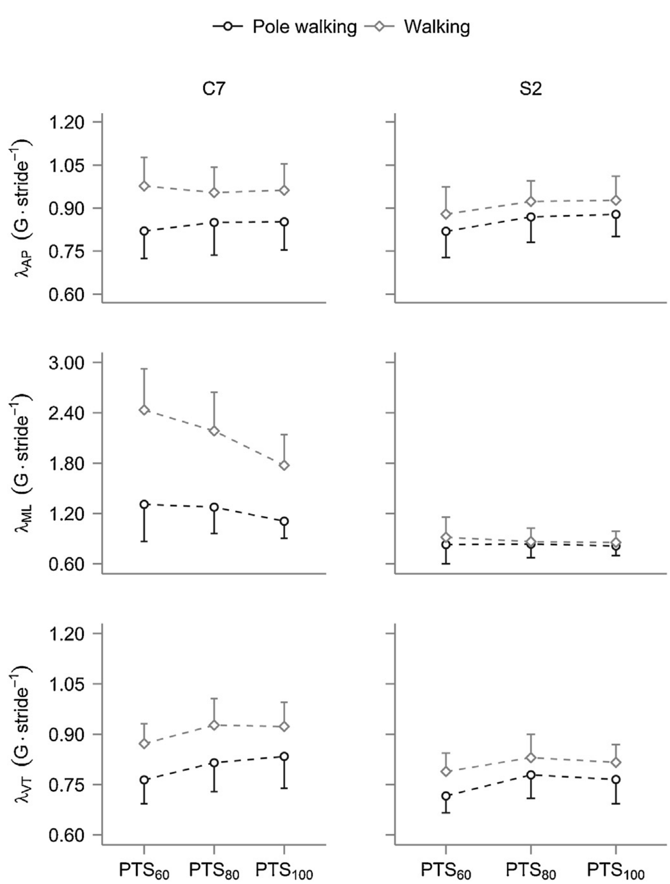

Fig. 2. Local divergence exponent (λ) calculated over the 0-0.5 strides interval in the anterior-posterior ($_{AP}$), medial-lateral ($_{ML}$) and vertical ($_{VT}$) directions. The accelerometers were placed at the level of the seventh cervical vertebrae (C7) and of the second sacral vertebrae (S2). Three different speeds are presented: 60% (PTS$_{60}$), 80% (PTS$_{80}$) and 100% (PTS$_{100}$) of the preferred transition speed during walking and pole walking.