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Influence of normative data's walking speed on the computation of conventional gait indices

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ABSTRACT

The pathology's impact on gait pattern may be overestimated by conventional gait indices (Gillette Gait Index – GGI, Gait Deviation Index – GDI, Gait Profile Score – GPS), since impairments' consequences on kinematics may be amplified by a change in walking speed. The objectives of this study were to evaluate the influence of walking speed on the computation of gait indices and to propose a corrective method to cancel the effects of walking speed. Spatiotemporal parameters and kinematics of fifty-four asymptomatic participants (30 M/24 W, 37.9 ± 13.7 years, 72.8 ± 13.3 kg, 1.74 ± 0.10 m) were collected at four speed conditions (C_1:[0,0.4] m s^{-1}, C_2:[0.4,0.8] m s^{-1}, C_3:[0.8,1.2] m s^{-1}, C_4:spontaneous). Four values of each index were computed for each trial using successively the four conditions as normative data repository. Mean values over all participants were statistically compared (paired *t*-tests, 95% confidence level). Indices values computed with normative at equivalent walking speed were not statistically different from reference values. Meanwhile, deviations appeared when the walking speed discrepancy between conditions and normative increased. These drifts related to walking speed mismatch have been quantified and fitting functions proposed. A correction was applied to indices. GGI was efficiently adjusted while GDI and GPS remain different from their reference values for C_1 and C_2. Gait indices must be interpreted cautiously in function of the normative data repository's walking speed used for computation. Furthermore, a coupled use of conventional and corrected gait indices could lead to a better comprehension of the contribution of impairments and walking speed on gait deviations and overall gait quality.

1. Introduction

Nowadays, instrumented gait analysis is widely used to quantify movement patterns of individuals during walking, especially to understand the deficits related to a pathology with intricate gait deviations. This analysis provides joint kinematics, kinetics and ground reaction forces in three dimensions, and electrical muscular activity. However, the interpretation of this complex set of data is not trivial, and several gait indices have been introduced in the literature to summarise them and to assess treatment outcomes. The most common are the Gillette Gait Index (GGI) or Normalcy Index (Schutte et al., 2000), the Gait Deviation Index (GDI) (Schwartz and Rozumalski, 2008), and the Gait Profile Score (GPS) (Baker et al., 2009). The GGI, based on a principle components analysis, defines a distance between 16 discrete gait parameters (*i.e.* temporal, spatial and kinematic parameters) and averaged normative data (Schutte et al., 2000). Instead of using discrete variables, the GDI and GPS take into account 15 kinematic time-series along the whole gait cycle and tend to give a more overall measure of gait deviations (Baker et al., 2009; Schwartz and Rozumalski, 2008). In all cases, the computation of these indices is based on a comparison between gait characteristics of a participant and those of a normative data repository established on an asymptomatic population. These three indices have been validated and used in children with cerebral palsy (Baker et al., 2009; Massaad et al., 2014; Rasmussen et al., 2015), and to a lesser extent in children with various pathologies (McMulkin and MacWilliams, 2015; Romei et al., 2004). In adults, the GGI has been validated for an asymptomatic population (Cretual et al., 2010), and each of these indices has been used for various pathologies, such as spinal cord and brain injuries (GGI: (Cretual et al., 2010)), Parkinson's disease (GDI and GPS: (Speciali et al., 2014)), spastic cerebral palsy

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(GDI: (Maanum et al., 2012)), and multiple sclerosis (GPS: (Pau et al., 2014)).

Whatever the investigated population, the spontaneous walking speed of participants (e.g. ranged between 0.18 and 1.03 m s^{-1} for stroke (Olney et al., 1994)) is often slower than for asymptomatic participants (ranged between 1.04 and 1.60 m s^{-1} (Salbach et al., 2015)). However, a modification of walking speed influence gait parameters, and several studies have described its effect on spatiotemporal parameters, kinematics, kinetics and muscle activity in asymptomatic children (Schwartz et al., 2008; van der Linden et al., 2002) and adults (Hanlon and Anderson, 2006; Kirtley et al., 1985; Kwon et al., 2015; Lelas et al., 2003; Murray et al., 1984). In particular, it has been shown that a decrease in walking speed implies a decrease in cadence and swing phase relative duration (*i.e.* expressed as percentage of gait cycle) (Kirtley et al., 1985; Murray et al., 1984; Schwartz et al., 2008), in the range of hip flexion–extension (Murray et al., 1984; Schwartz et al., 2008; van der Linden et al., 2002), in the maximum knee flexion during early stance and swing phase (Hanlon and Anderson, 2006; Kwon et al., 2015; Lelas et al., 2003; Schwartz et al., 2008; van der Linden et al., 2002), as well as in the maximum plantarflexion (Kwon et al., 2015; Schwartz et al., 2008; van der Linden et al., 2002). Gait indices, based on gait parameters or kinematic curves and being usually computed using normative data repositories established on asymptomatic participants walking at spontaneous walking speed (Baker et al., 2009; Romei et al., 2004), may thus also be influenced by walking speed. To our knowledge, only one study has been reported in a conference abstract to highlight the impact of the normative data's walking speed on the computation of GDI in a child population (Rozumalski and Schwartz, 2012).

While an impairment highlighted during clinical examination may have a direct impact on the related joint kinematics, it can also affect walking speed and thus indirectly the kinematics of other joints. For example, in case of spasticity of the triceps surae, walking speed may be decreased willingly to avoid muscle spasms, and hip joint kinematics may thus be altered indirectly because of this reduced walking speed. Hence, since the impact of impairments on spatiotemporal parameters and kinematics may be amplified by the consequences of a change in walking speed, the impact of a pathology on gait may be overestimated by conventional gait indices. Not adjusting the walking speed of the two compared populations (*i.e.* pathological vs. asymptomatic) may thus influence the analysis of gait deviations, and lead to misinterpretations regarding the impact of the pathology on gait. Unfortunately, this levelling if often impossible in a clinical setting, since established normative data repositories has been obtained at a spontaneous walking speed (Pinzone et al., 2014), and the development of a new normative data repository requires a lot of resources and time. To overcome this issue, several authors have thus proposed some methods aiming to adapt existing normative data repositories for slow-walking participants. Lelas et al. proposed linear and quadratic regression equations between walking speed and 27 kinematic and kinetic parameters (Lelas et al., 2003). Hanlon and Anderson considered kinematic time-series along the whole gait cycle and confirmed correlations and linear regressions between walking speed and lower limb kinematics (Hanlon and Anderson, 2006). However, these correction methods have not been applied yet to the gait indices computation.

Based on a recently established data repository of asymptomatic participants walking at different speeds (Schreiber et al., 2016a, 2016b), the objectives of the present study were (1) to evaluate the impact of the normative data's walking speed on the computation of GGI, GDI and GPS in asymptomatic adults and (2) to propose non dependent-velocity indices based on corrective methods of conventional indices to limit the effect of walking speed discrepancy.

2. Methods

This study uses the data of a previous protocol aiming to establish a data repository of asymptomatic participants walking at different speeds (Schreiber et al., 2016a, 2016b). Details about participants, protocol, data acquisition and treatment are given below.

2.1. Participants

Fifty-four adults (24 women and 30 men, 37.9 (SD 13.7) years, 1.74 (SD 0.10) m, 72.8 (SD 13.3) kg) with no neuro-orthopaedic trouble were included in this study. They all gave informed written consent prior to their inclusion. Data were collected and anonymised before analysis during an ongoing internal measurement campaign in the Centre National de Rééducation Fonctionnelle et de Réadaptation – Rehazenter of Luxembourg, aiming to provide a normative data repository for our clinical practice. The protocol was approved by the Institutional Review Board.

2.2. Protocol

The participants were asked to walk on a 10-m straight level walkway and four conditions of walking speed were recorded. During conditions C1, C2 and C3, the participants were asked to adapt their walking speed respectively between 0 and 0.4 m s^{-1}, 0.4 and 0.8 m s^{-1}, and 0.8 and 1.2 m s^{-1}. These ranges correspond to the three walking speed groups described by Perry et al. (1995) (*i.e.* household ambulators, limited community ambulators and community ambulators, respectively). It was assumed that this rhythmic auditory stimulation does not significantly impact spatiotemporal parameters and kinematics as previously demonstrated by Schreiber et al. (2016a). Condition C_4 corresponded to the participants' spontaneous walking speed. A minimum of 5 gait trials were recorded on each participant and for each condition. A full description of the protocol has been previously provided by Schreiber et al. (2016a).

2.3. Data acquisition and processing

Spatiotemporal parameters and kinematics were recorded simultaneously, using a 10-camera optoelectronic system (OQUS-4, Qualisys AB, Göteborg, Sweden) sampled at 100 Hz. The marker set was based on the Leardini's protocol (Leardini et al., 2007) and composed of 26 reflective cutaneous markers placed on anatomical landmarks. The markers were not removed between trials and conditions to avoid any difference in their placement.

Data were imported and processed under Matlab (Matlab R2011b, The MathWorks, Inc., Natick, Massachusetts, United States) using the Biomechanics ToolKit (BTK) (Barre and Armand, 2014). Marker trajectories were interpolated when the gap size did not exceed 10 frames (*i.e.* 50 ms) (cubic spline) and smoothed (4th-order low pass Butterworth filter, cut-off frequency of 6 Hz). They were finally normalised to 100% gait cycle. In order to take into account leg length differences between participants, walking speed was reported non-dimensionalised by dividing the raw walking speed by the Froude velocity, as proposed by Hof (1996). This factor corresponds to the product of the leg length and the gravitational constant. The computation of the three gait indices (*i.e.* GGI, GDI and GPS) followed their original procedure (Baker et al., 2009; Schwartz and Rozumalski, 2008) except for the GGI that was computed over 15 parameters instead of 16 (*i.e.* excluding the time of peak flexion) following the recommendations of Cretual et al. for adults (Cretual et al., 2010). Under the assumption that

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