Journal of Electromyography and Kinesiology 25 (2015) 339–346

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Decrease in ankle–foot dorsiflexion range of motion is related to increased knee flexion during gait in children with spastic cerebral palsy

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ARTICLE INFO

Article history:
Received 17 January 2014
Received in revised form 16 October 2014
Accepted 26 October 2014

Keywords:
Cerebral palsy
Gait
Ankle
Knee
Equinus deformity
Range of motion

ABSTRACT

Purpose: To determine the effects of decreased ankle–foot dorsiflexion (A-Fdf) range of motion (ROM) on gait kinematics in children with spastic cerebral palsy (SCP). *Methods:* All participants were children with spastic cerebral palsy ($n = 10$) who walked with knee flexion in midstance. Data were collected over 2–5 sessions, at 3-monthly intervals. A-Fdf ROM was quantified using a custom-designed hand-held ankle dynamometer that exerted 4 Nm at the ankle. Ankle–foot and knee angles during gait were quantified on sagittal video recordings. Linear regression (cross-sectional analysis) and General Estimation Equation analysis (longitudinal analysis) were performed to assess relationships between (change in) A-Fdf ROM and (change in) ankle–foot and knee angle during gait. *Results:* Cross-sectional analysis showed a positive relationship between A-Fdf ROM and both ankle–foot angle in midstance and terminal swing. Longitudinal analysis showed a positive relationship between individual decreases in A-Fdf ROM and increases of knee flexion during gait (lowest knee angle in terminal stance and angle in terminal swing). *Conclusion:* For this subgroup of SCP children, our results indicate that while changes in ankle angles during gait are unrelated to changes in A-Fdf ROM, changes in knee angles are related to changes in A-Fdf ROM.

1. Introduction

Limitations in ankle–foot dorsiflexion (A-Fdf) range of motion (ROM) are common in children with spastic cerebral palsy (SCP) (Benard et al., 2010; Nordmark et al., 2009; Wichers et al., 2009) and frequently result in altered gait patterns such as equinus gait, characterized by increased ankle plantar flexion at the ankle–foot (Becher, 2002; Gage, 2004; Rodda et al., 2004). As walking with equinus gait results in higher than normal levels of energy expenditure (van den Hecke et al., 2007), several treatment strategies have been developed to reduce equinus at the ankle during gait. These strategies focus on lengthening the calf muscles to increase A-Fdf ROM, as physicians assume that increasing A-Fdf ROM leads to an increase in ankle–foot dorsiflexion during gait (Morris and Condy, 2009). However, this assumption has received relatively little scientific attention and the few cross-sectional studies available have reported rather low correlation coefficients between these variables (Abel et al., 2003; Desloovere et al., 2006). Currently, a possible relationship between changes in A-Fdf ROM and ankle angle in gait is unknown.

As the gastrocnemius muscle exerts forces and moments at both ankle and knee joints, it is likely that its length and stiffness probably affect ROM of both A-Fdf and knee extension. Lengthening and/or reducing stiffness of gastrocnemius muscle is therefore expected to affect both joint angles. Surprisingly, earlier studies failed to show any relationship between A-Fdf ROM and knee angles during gait (Abel et al., 2003; Desloovere et al., 2006; McMulkin et al., 2000).

At least three factors could potentially explain a failure to show a relationship between A-Fdf ROM and gait knee angles. One factor may be the specific gait patterns seen in children with equinus gait. These children show distinguishable gait patterns. For instance, in some children the gait pattern is characterized by walking with knee hyperextension in midstance (Becher, 2002), which is

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http://dx.doi.org/10.1016/j.jelekin.2014.10.015
1050-6411/

presumably related to a prematurely active, short or stiff soleus muscle. Additional shortness, stiffness or hyperactivity of the gastrocnemius muscle may yield excessive plantar flexion in the ankle–foot (Becher, 2002; Matjacic et al., 2006). Another pattern is characterized by walking with knee flexion in midstance (Becher, 2002). Changes in A-Fdf ROM in this pattern may not only affect the ankle–foot angle during gait, but may also change knee flexion (Becher, 2002; Matjacic et al., 2006). Therefore, effects of altered A-Fdf ROM on ankle–foot and knee angles during gait may vary for both patterns. It is important to clearly distinguish these gait patterns, because gait type differences may be confounding the relationship between A-Fdf ROM and the ankle–foot and knee angle during gait research.

A second important factor is the sizable interindividual variation in factors, besides gait type and A-Fdf ROM, that affects knee angle during gait. Confounding factors such as hip flexion ROM and the length and stiffness of the hamstring muscles may affect the relationship between A-Fdf ROM and knee angle during gait (Desloovere et al., 2006; Gage, 2004). Longitudinal analysis of this relationship will help reduce confounding.

A third important factor is the very large variation due to measurement errors. Further improvements in measurement methods for A-Fdf ROM are therefore essential: (a) Rather than using simple goniometry for joint angle measurement (for limitations, see studies Kim et al. (2011) and Rome and Cowieson (1996)), joint angular measurement should be combined with techniques that standardize externally applied moments (Barber et al., 2011, 2012). (b) Particular attention should be paid to stabilization of the foot joints, as the measured A-Fdf ROM is a combination of talocrural and other foot bone joint movements rather than talocrural joint movement alone (Huijing et al., 2013; Iwanuma et al., 2011). For example, in the case of flexible valgus deformations, talonavicular movement should be prevented by supinating and adducting the forefoot. Applying these methods yields higher levels of precision (of up to 3° (Benard et al., 2010)) compared to use of simple goniometry to determine A-Fdf ROM.

The present study was designed to investigate relationships between longitudinal changes in A-Fdf ROM and concomitant changes in ankle–foot and knee angles during gait in children walking with equinus and flexed knees. These longitudinal results will be compared to results from a cross-sectional study for the same subjects and variables. It is hypothesised that: (1) A-Fdf ROM changes over time and is related to changes in A-Fdf and knee flexion angles during gait, and (2) children with smaller A-Fdf ROM walk with less A-Fdf and enhanced knee flexion.

2. Materials and methods

This study, being part of a larger study (Maas et al., 2012), was approved by the Medical Ethics Committee of VU University Medical Center. The parents of all participants provided written informed consent.

2.1. Participants and ethics

Ten children with SCP (9.2 ± 1.8 years old, range: 5.9–11.2) took part in a randomized controlled trial assessing the effects of orthotic treatment at rest [Maas et al., 2012]. Participant characteristics included: (a) a clinical diagnosis of SCP, (b) aged between 4 and 12 years old and (c) a GMFCS class of I–III. In addition, (d) a non-instrumented clinical assessment of at least 0° A-Fdf was attainable, also described as 90° between lower leg and foot, with extended knee (note that the result with instrumented measurement, used in the analysis below, may differ from this clinical assessment), (e) the knee joint could be fully extended on clinical examination, and (f) participants walked with knee flexion in midstance. Furthermore, participants (g) had been previously treated for reduced A-Fdf ROM, (h) participated in at least two consecutive sessions, and (i) were able to relax during measurements (meaning that EMG levels of mm. gastrocnemius lateralis and tibialis anterior had to be lower than 10% of maximum voluntary contraction values).

2.2. Experimental procedures

The participants were tested 2–5 times, at intervals of 3 months (see Table 1). Testing was performed at the VU University Medical Center, Department of Rehabilitation Medicine, Amsterdam and at the Medical Rehabilitation Center Groot Klimmendaal, Arnhem, the Netherlands.

2.3. Experimental protocols

2.3.1. Dynamometry

A hand-held ankle dynamometer (Fig. 1, for details see Benard et al. (2010)) was used to quantify A-Fdf ROM at each testing session. This hand-held dynamometer enables correction for mobile valgus or varus deformity of the foot (for details see Huijing et al. (2013)). During quantification of A-Fdf ROM, participants were prone on a bench with extended knees, both feet overhanging the edge. The lower leg was positioned horizontally by alignment of fibula head and lateral malleolus. Using the dynamometer handle, the foot was dorsiflexed slowly until a 4 Nm external dorsiflexion moment was exerted. This was maintained for 5 s ('holding' phase). The measurement was acceptable only if participants did not move during holding. The procedure was repeated 5 times, with 5 s intervals between trials.

The angle between footplate and tibia at the end of holding was used to quantify A-Fdf ROM ($\varphi_{\text{foot-4Nm}}$). Means of 5 trials were used as individual data points. Standard error of the mean (SEM) is 1.36° (Benard et al., 2010). Longitudinal changes of A-Fdf ROM ($\Delta\varphi_{\text{foot-4Nm}}$) were quantified by calculating the differences in $\varphi_{\text{foot-4Nm}}$ between consecutive sessions.

To quantify muscle activation during dynamometry measurements (i.e. determination of A-Fdf ROM), surface EMG levels of mm. gastrocnemius lateralis (GL) and tibialis anterior (TA) were recorded using a Porti 5 system (TMS-International, The Netherlands). Only EMG data from the last 1.5 s of the holding phase were analysed. EMG signals were recorded at a sampling rate of 1000 Hz using Portilab (TMS International, The Netherlands). According to Seniam Guidelines (Hermens et al., 1999; De Luca et al., 2010), raw EMG signals were filtered at 20 Hz using a 5th order high pass Butterworth filter. In addition, a 50 Hz 5th order band stop Butterworth filter was used to eliminate hum (from 50 Hz power line interference) and signals were full wave rectified for further processing. Subsequently, for visual inspection, the signals were filtered with a 5 Hz 5th order low pass Butterworth filter. To quantify possible events during holding phase, the moving average of a 200 ms timeframe (Merletti, 1996) around the maximum of the rectified signal during the last 1.5 s of the holding phase was calculated. Processing was performed using a custom software package programmed in Matlab (version 7.1, The MathWorks Inc., Natick, MA, USA). To express the EMG activity of dynamometry measurements as a percentage of maximal voluntary contraction (MVC), MVC of plantar- and dorsiflexion at the ankle was measured (with the ankle slightly plantar flexed and the knee almost extended) and recorded for each participant while the assessor provided resistance. This was performed twice. After processing MVC EMG signals of mm. gastrocnemius and tibialis anterior (similar to EMG processing during dynamometry measurements), EMG signals recorded during dynamometry