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Multiple sclerosis influences the precision of the ankle plantarflexon muscular force production

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ABSTRACT

Objective: To quantify the precision of the steady-state isometric control of the ankle plantarflexors musculature of individuals with multiple sclerosis (MS), and to evaluate if the precision is related to the mobility impairments.
Methods: Individuals with MS and healthy adults performed a submaximal steady-state isometric contraction with the ankle plantarflexors. The coefficient of variation was used to assess the amount of variability or error in the precision of the torques generated by the ankle plantarflexor musculature. The participants also walked across a digital mat at their preferred and fast-as-possible walking speeds, which recorded their spatiotemporal gait kinematics.
Results: The individuals with MS: (1) had reduced maximal voluntary torques at the ankle, (2) a greater amount of variability in the precision of the isometric ankle torques, (3) altered and more variable spatiotemporal gait kinematics, and (4) a greater amount of variability in the isometric ankle torques were related to a slower walking speed and cadence, shorter step length and a greater amount of gait variability.
Conclusions: These results further fuels the impression that a reduction in control of the ankle joint musculature may be a key factor in the mobility and balance impairments seen in individuals with MS.

1. Introduction

Multiple sclerosis (MS) is a demyelinating disease that occurs in young adults and often affects the control of the leg musculature. Numerous individuals with MS experience mobility and balance impairments that limit their activities of daily living [1]. Historically, the clinical impression was that these impairments were due to weaker muscles that fatigue at a faster rate [2–7]. Although this is likely a factor, there has been limited attention to how MS impacts the precision of the ankle musculature control. Precise control of the ankle joint is important for correcting the postural sway, clearing the foot during the swing phase of gait and push-off at terminal stance [8,9]. It has been shown that individuals with MS with higher Kurtzke Expanded Disability Status Scores (EDSS) tend to generate less power by the ankle joint during the stance phase of gait [10]. Additionally, spasticity in the gastrocnemius and soleus muscles has been shown to impact the gait and balance in individuals with MS [11]. Taken together, these results suggest that a reduction in control of the ankle joint musculature may be a primary factor that leads to the mobility and balance impairments seen in individuals with MS.

Variability or error is present in all voluntary contractions and impacts the precision and control of the motor performance [12–14]. Several investigations have shown that aging results in greater variability in the steady-state isometric performance of the ankle joint, and that these variations may be a result of the inability to properly activate the motor unit pool that innervates the ankle musculature [11,14]. Despite this insight, limited efforts have been made to determine if MS further amplifies the amount of variability that occurs while attempting to control the precision of the ankle musculature. A previous study has shown that individuals with MS may have an increased amount of variability in the motor unit firing rate [15]. Given that the variability of the motor unit discharge rate is known to be associated with increased force variability during isometric force tasks [16], it is possible that individuals with MS may display an increased variability while trying to control the precision of the muscular force. Potentially, a greater amplification of the variability at the ankle joint may be a key factor for the mobility impairments often reported in individuals with MS.

The primary purpose of this study was to quantify the amount of variability or error in the precision of the steady-state ankle

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plantarflexion isometric muscular forces generated by individuals with MS. We hypothesized that (1) compared with controls, individuals with MS will have an amplified amount of variability when they attempt to precisely match a low level isometric target with their ankle plantarflexors. Secondarily, we hypothesized that (2) individuals with MS will have weaker isometric ankle plantarflexion muscular strength, (3) the spatiotemporal gait kinematics will be altered, and (4) the spatiotemporal gait kinematics will be related to the amount of variability seen in the precision of the ankle plantarflexor target matching task.

2. Experimental procedures

Twenty-two adults (Age: 49.3 ± 8 years; Female = 14) with relapsing-remitting or secondary progressive MS participated in the study. The subjects had an average EDSS of 5.3 ± 1 (median = 5.75), which indicates that on average each subject could walk independently for at least 100 m with an assistive device (e.g., cane). Twenty normal, healthy adults served as a control group (Age: 45.1 ± 14 years; Female = 16). All testing was done at the University of Nebraska Medical Center. This study was approved by the Institutional Review Board and was performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki. Additionally, all participants provided informed consent prior to participation in the study.

The subjects performed the isometric ankle plantarflexion contractions seated in an isokinetic dynamometer (Biodex Inc., Shirley, NY). The chair of the isokinetic dynamometer had the backrest set at an angle of 90°, and the participant had their knee fully extended with their ankle in a neutral position. A foot strap was used to secure their foot to the metal footplate. The largest torque generated from two maximum isometric contractions was used to establish the participant's maximum voluntary torque (MVT) and was normalized by body weight (kg) prior to comparison. For the experiment, the participant performed two steady-state isometric contractions at 20% of their MVT. The target and the torque exerted by the participant was displayed as a bar graph on a large monitor that was positioned ~1 m away from the subject at eye level. The participant was instructed to produce and hold a plantarflexion force that matched the 20% MVT target. The participant was given ample time to practice achieving the target torque before the two actual trials were recorded. These two trials were then averaged together for all data measures. The voltage output from the torque motor was read by custom LabVIEW (National Instrument Inc., USA) software and sampled at 1 kHz by a 14-bit National Instruments analogue-to-digital converter. The voltage output from the Biodex dynamometer was converted to Nm and displayed in real-time to the participant. The maximum on the vertical scale of the bar graph was twice the target value [13]. Each steady-state contraction was performed for 30 s. The coefficient of variation (CV = [Standard Deviation of Torque/Mean Torque] × 100) was used to assess the amount of variability present in the middle 15 s of the steady-state torque. A greater CV value was an indication of a larger amount of error in the joint steady-state torque control [17].

Prior to the completion of the ankle plantarflexion task described above, the participants walked across a digital mat (GaitRITE, Sparta, NJ) at their preferred and fast-as-possible walking speeds. The mat quantified the participant's spatiotemporal kinematics and was used to calculate the walking velocity, step width, step length, and cadence. In addition, the standard deviation of the step length, and step width were used to quantify the gait variability. Each participant completed two walking trials at the respective speeds and the data from these two trials was averaged together.

Independent *t*-tests were used to examine the differences between the MS and control groups for the maximum torque, CV and the spatiotemporal kinematics. Spearman rho correlations were used to evaluate the relationship between the CV of the steady-state torque and the spatiotemporal kinematics, as well as MVT and the spatiotemporal kinematics. All statistical analyses were conducted using SPSS version 22 (IBM, Armonk, NY), with an alpha level of 0.05.

3. Results

A representative time series for an individual with MS and a control performing the ankle plantarflexion motor task is shown in Fig. 1. Qualitatively it is apparent that the individual with MS had greater variability when trying to control the precision of the ankle joint plantarflexor musculature. This observation was confirmed by the CV for the steady-state torques, where the CV was greater for the individuals with MS compared to the controls ($p = 0.03$; Fig. 2A). Hence, indicating that the participants with MS generated more errors when attempting to control the precision of their ankle plantarflexor muscular force production. The maximum torque generated by the ankle plantarflexors was also significantly lower for the individuals with MS compared with the controls ($p = 0.03$; Fig. 2B). This indicated that the individuals with MS also had weaker isometric ankle plantarflexor strength compared to the controls.

The spatiotemporal gait kinematics were notably different between the two groups for all variables. At preferred walking speeds, the individuals with MS had a slower walking velocity (MS = 0.68 ± 0.22 m/s, controls = 1.28 ± 0.14 m/s; $p < 0.01$), wider step width (MS = 0.16 ± 0.04 m, controls = 0.11 ± 0.02 m; $p < 0.01$), shorter step length (MS = 0.44 ± 0.08 m, controls = 0.67 ± 0.07 m; $p < 0.01$), and slower cadence (MS = 92.2 ± 21.4 steps/min, controls = 114.5 ± 9.4 steps/min; $p < 0.01$). In addition, the step lengths (MS = 3.39 ± 1.81 cm, controls = 1.91 ± 0.86 cm; $p < 0.01$), and step widths (MS = 2.81 ± 1.39 cm, controls = 1.74 ± 0.92 cm; $p = 0.02$) were more variable in the MS group.

The same was true at fast-as-possible walking speeds, with individuals with MS having a slower velocity (MS = 0.93 ± 0.36 m, controls = 1.98 ± 0.27 m; $p < 0.01$), wider step width (MS = 0.14 ± 0.04 m, controls = 0.10 ± 0.03 m; $p < 0.01$), shorter step length (MS = 0.51 ± 0.12 m, controls = 0.80 ± 0.09 m, $p < 0.01$), slower cadence

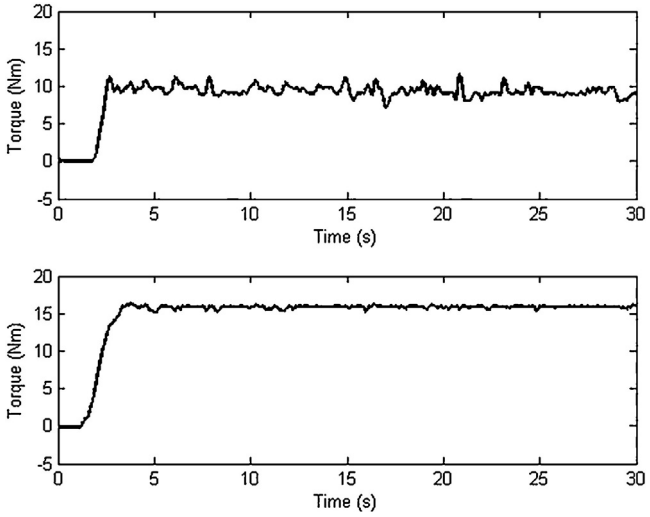

Fig. 1. Exemplary ankle joint torques for a subject with MS (top panel) and control subject (bottom panel).