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Effects of hip abductor muscle fatigue on gait control and hip position sense in healthy older adults

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ABSTRACT

We experimentally investigated whether unilateral hip abductor muscle fatigue affected gait control and hip position sense in older adults. Hip abductor muscles were fatigued unilaterally in side-lying position in 17 healthy older adults (mean age 73.2 SD 7.7 years). Hip joint position sense was assessed by an active–active repositioning test while standing and was expressed as absolute and relative errors. Participants walked on a treadmill at their preferred walking speed, while 3D linear accelerations were collected by an inertial sensor at the lower back. Gait parameters, including step and stride time, local divergence exponents and harmonic ratio were quantified. In fatigued gait, stride time variability and step-to-step asymmetry in the frontal plane were significantly increased. Also a significantly slower mediolateral trunk movement in fatigued leg late stance toward the non-fatigued leg was observed. Despite these temporal and symmetry changes, gait stability in terms of the local divergence exponents was not affected by fatigue. Hip position sense was also affected by fatigue, as indicated by an increased relative error of 0.7° (SD 0.08) toward abduction. In conclusion, negative effects of fatigue on gait variability, step-to-step symmetry, mediolateral trunk velocity control and hip position sense indicate the importance of hip abductor muscles for gait control.

1. Introduction

Balance control is essential for successful performance of daily activities and requires adequate sensory integration of visual, vestibular and somatosensory information [1]. Somatosensory information provides feedback about joint position and movement (kinesthesia). Muscle spindles impart the most important proprioceptive input [1]. Rat studies suggested that with aging, the sensitivity of muscle spindles in static and dynamic conditions, the total number of intrafusal muscle fibers and the number of nuclear chain fibers per spindle decrease [2]. Moreover, in older adults, proprioceptive acuity and muscular output may be reduced as a result of an increased fatigability because of loss of muscle strength and central activation failure [3]. Muscle fatigue is defined as an acute impairment in the ability to produce maximum force [4]. In general, muscle fatigue may arise during muscular contractions due to failure at one or more sites along the pathway of force production from the central nervous system to the contractile apparatus [5]. Muscle fatigue has also been associated with decreased proprioception [6,7].

During human gait, hip abductor muscles can stabilize the trunk over the stance leg in mediolateral (ML) direction [8]. In addition, biomechanical modeling showed that the hip abductor muscles apply a moment to accelerate the center of mass (COM) medially toward the swing leg in late stance [9]. Hence, hip abductor weakness might explain impaired gait control in ML direction. For example, following hip replacement surgery, asymmetrical loading has been shown between left and right legs, with a more variable gait pattern [10]. Additionally, the incidence of hip fractures due to sideway falls in older adults might be the consequence of hip abductor weakness and impaired ML balance control [11]. As age-related impairment in hip abduction/adduction torque-time capacity is evident [12], it is important to gain more knowledge on the consequences of hip abductor muscle

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impairment in ML gait control in older adults. One way to experimentally induce muscle impairment is by means of local muscular fatigue, in which the isolated role of hip abductor muscles could be investigated. For instance, hip abductor muscle fatigue is known to increase postural sway velocity in both the sagittal and frontal planes during unipedal stance in young adults, indicating a decline in postural control in static standing [13].

The aim of the present study was to experimentally explore whether unilateral hip abductor muscle fatigue affects gait control in older adults. Since decreased muscle force is the direct effect of fatigue [4], we hypothesized that unilateral hip abductor fatigue changes gait control mainly in ML direction by decreasing stability and symmetry as well as by increasing spatiotemporal variability, possibly because of changes in the trunk COM velocity. The second aim of the study was to investigate the effect of hip abductor fatigue on hip position sense. As muscular fatigue negatively affected proprioceptive acuity in the knee and ankle [7], we hypothesized that it might have the same effect in the hip, possibly contributing to impaired gait.

2. Methods

2.1. Participants

Seventeen healthy volunteers (twelve females; mean age 73.2 SD 7.7 years; height 166.1 SD 9.2 cm; weight 68.2 SD 8.8 kg) with no neurological or musculoskeletal impairments participated in this study. The local ethics committee approved the protocol and participants signed a consent form.

2.2. Assessment of joint position sense (JPS)

First we assessed JPS during hip abduction, for which participants were asked to stand on their unfatigued leg on a 10 cm high block, with their fatigued leg unsupported but aligned with the supported leg (starting position). They were allowed to touch a horizontal bar in front of them at hip height for support. The JPS for hip abduction was assessed unilaterally by an "active–active" and reliable repositioning test [14]. Four trials of hip abduction were performed with target angles randomly varying within a range of 10–40°. Kinematics of the lower limbs were recorded using 6 LED markers (Optotrak, Northern Digital Inc, Waterloo, Canada) at 100 Hz. The markers were attached bilaterally at the apex of the iliac crest, greater trochanter and lateral femur epicondyle to calculate the 3D angle between the vector from the greater trochanter to knee marker and the vector from the greater trochanter to the iliac crest marker. Hip JPS was quantified as the absolute angular error (AE) between target and reproduced angle and the relative angular error (RE) between the two angles with consideration of the direction of the error. A positive relative error indicates an overshoot toward abduction. Because data of the individual trials showed no correlation of AE and RE with the target angle, final values for both AE and RE were obtained by averaging the error over four trials at different target angles.

2.3. Gait and fatigue protocol

The participants walked on a treadmill at preferred walking speed while looking at a wall in front of them and without holding the handrails. A safety harness was used to support full body weight in case of an impending fall. Preferred walking speed was determined by gradually increasing the speed of the treadmill until the participants indicated that the walking velocity was experienced as comfortable. Then, the examiner further increased the speed by 0.5 km/h and checked whether it was experienced as

Fig. 1. Set-up to induce hip abductor muscle fatigue, by 30° abduction cycles at 20 repetitions per minute.

comfortable again. If so, this procedure was repeated until participants indicated that it was too fast. Then, the speed was gradually decreased to the point that participants were sure about their preferred walking speed.

After a period of 5 min familiarization and setting the preferred walking speed, they were instructed to walk for 5 min in order to collect 3D linear accelerations of the trunk during walking at a sampling rate of 100 Hz by an inertial sensor (Dynaport Hybrid, McRoberts, The Hague, The Netherlands). The sensor was attached with an elastic belt with Velcro fixation at the lower back at the level of L5.

Then, acute fatigue in the hip abductor muscle group was induced. Participants were placed side lying on a firm mattress with their body and leg straightened (Fig. 1). Randomly, half of the participants were fatigued on their right side and the other half on their left side. A plastic bar was positioned over the participant's foot and set to a height that corresponded to 30° of hip abduction with their knee fully extended. The bar gave the participant a fixed target for the amount of hip abduction required for every repetition and also offered tactile feedback. We instructed participants to raise their leg to touch the plastic bar on each repetition. A metronome was used to pace the lift-and-lower phases, yielding a rate of 20 lifts per minute for all participants. Weight resistance was used to promote the onset of fatigue and was equal to 20% of the participant's total leg moment as estimated from anthropometrical data. Fatigue was defined as the instant at which the subject failed either to reach the target range of motion or went out of sync with the pacing for 3 consecutive repetitions. Immediately after the fatigue protocol, we assessed the subjective exertion level through the Borg CR-10 scale. To ensure maximal effort from each subject, strong verbal encouragement was given throughout the protocol.

Immediately after the fatigue protocol, the JPS post-test was repeated. Since the effect of fatigue can quickly disappear, we repeated the fatigue protocol once more, followed by the post-fatigue walking test.

2.4. Analyses of gait parameters

First, to avoid inaccuracies due to misalignment of the sensor to the anatomical coordinate system, a correctional rotation was made [15]. Then, after low-pass filtering the forward acceleration signal, heel contact (HC) was determined based on the anteroposterior acceleration signal [16]. Mean stride time and variability (standard deviation) was calculated based on HC detection over the first 150 strides.

ML position data were calculated by a double integration of ML acceleration data to discriminate between fatigued and non-fatigued leg HCs, based on an analysis of ML movements of the lower trunk [16]. Fatigued leg stance time was then determined as the time from fatigued leg HC until the following non-fatigued leg HC, whereas non-fatigued leg stance time was determined from