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Regional neuromuscular regulation within human rectus femoris muscle during gait in young and elderly men

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ABSTRACT

Recently, we demonstrated region-specific electromyography (EMG) responses along the rectus femoris (RF) muscle during gait in healthy young men (Watanabe et al., 2014b). For the RF muscle, regional EMG response should be tested to characterize neuromuscular control and/or to assess its dysfunction and/or pathology during gait. We aimed to identify spatial distribution of EMG pattern within the RF muscle in elderly during gait. Seven young men (age: 20.4 ± 1.0 years) and 8 elderly men (age: 73.8 ± 5.9 years) walked on treadmill with three different speed: slow (preferred -1 km/h), preferred, and fast (preferred $+1$ km/h). The spatial distribution of surface EMG was tested by central locus activation (CLA), which is calculated from 18 surface electrodes along the longitudinal line of the muscle. CLA were not different between the groups for slow and preferred gait speed ($p > 0.05$) during a gait cycle. In fast gait speed, CLA at 80% of a gait cycle (swing phase) for the elderly were significantly located at more distal site than the young group ($p < 0.05$) (13.0 ± 2.1 cm and 10.2 ± 2.2 cm from most proximal electrodes for the elderly and young). This difference in CLA reflected a significantly lower EMG activity at the proximal regions in the elderly group ($p < 0.05$). These results suggest the elderly manifest characteristic regional EMG responses within the RF muscle for leg swing movement of fast speed gait.

1. Introduction

Age-associated changes in joint kinematics and/or kinetics during gait have been well investigated and biomechanical basis of them has been described (DeVita and Hortobagyi, 2000; Judge et al., 1996a, 1996b; Kerrigan et al., 1998; Riley et al., 2001; Winter et al., 1990). Although this is thought to be by neuromuscular adaptations with aging, very little work is currently available in the published literature on neuromuscular control in elderly during gait (Schmitz et al., 2009).

Surface electromyography (EMG) has been widely used for understanding neuromuscular control during gait (Winter and Yack, 1987). Comparison of EMG profiles to normal pattern is one of the standard methods for assessing dysfunction and/or pathology in gait for the elderly or patients (Annaswamy et al., 1999; Kerrigan et al., 1991; Reinbolt et al., 2008). Normal EMG patterns during gait for major lower extremity muscles were identified in previous studies (Barr et al., 2010; Byrne et al., 2005; Di Nardo and Fioretti, 2013; Nene et al., 2004). However, in rectus femoris (RF) muscle, large deviations were found among the previous studies. Two different EMG patterns, i.e., double bursts around swing-to-stance transition and stance-to-swing transition and single burst around stance-to-swing transition, were reported in the previous studies (Annaswamy et al., 1999; Winter and Yack, 1987; Yang and Winter, 1985). Since RF muscle is one of the key muscles in pathological gait patterns (Kerrigan et al., 1991; Knuppe et al., 2013; Reinbolt et al., 2008; Riley and Kerrigan, 1998; Sung and Bang, 2000), it is necessary to resolve this issue and to standardize normal EMG pattern for this muscle in the able-bodied adults, children, the elderly, and patients.

Recently, we demonstrated region-specific EMG responses within the RF muscle during gait by using multi-channel surface EMG technique (Watanabe et al., 2014b). In our study, proximal and distal regions of the RF muscle preferentially activate around stance-to-swing transition and swing-to-stance transition, respectively. Also, these two bursts were found at middle regions. Since two different patterns, which were reported in the previous studies (Annaswamy et al., 1999; Winter and Yack, 1987; Yang and Winter, 1985), were found within a muscle in our study, we suggested that variations in EMG patterns for the RF muscle among the studies is partly caused by region-specific neuromuscular regulation within a muscle. For the RF muscle, multi-channel

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surface EMG technique could strongly contribute to characterize neuromuscular control and to assess its dysfunction and/or pathology during gait. In the present study, we aimed to identify spatial distribution of EMG pattern within the RF muscle in elderly during gait. Since the effects of age upon joint kinetics/kinematics and/or neuromuscular activation were often found with an increase in gait speed (Judge et al., 1996a; Kerrigan et al., 1998; Riley et al., 2001), effect of gait speed on spatial distribution of surface EMG within the muscle was assessed in order to characterize age-related changes in neuromuscular responses in the present study.

2. Methods

2.1. Subjects

Seven young men (age: 20.4 ± 1.0 years, height: 169.5 ± 5.3 cm, body mass: 64.1 ± 6.9 kg) and 8 elderly men (age: 73.8 ± 5.9 years, height: 166.5 ± 6.6 cm, body mass: 63.1 ± 8.7 kg) volunteered for the present study. The subjects in both groups gave written informed consent for the study after receiving a detailed explanation of the purposes, potential benefits, and risks associated with participation. All subjects were healthy with no history of any musculoskeletal or neurological disorders. All study procedures were in accordance with the Declaration of Helsinki and research code of ethics of Chukyo University and were approved by the Committee for Human Experimentation of the Chukyo University and the Graduate School of Human and Environmental Studies, Kyoto University.

2.2. Experimental design

Walking on a treadmill (MEDTRACK ST65, Quinton Instrument Co., WA, USA) was performed in three different speeds. We applied (1) preferred speed (PS) for each subject, (2) slow speed, which is equivalent to 1 km/h slower than PS, and (3) fast speed, which is equivalent to 1 km/h faster than PS. PS was determined from average speed of normal gait for 10 m on floor. Gait speed of treadmill was gradually increased until reaching the specified conditions for 15–20 s and then maintained for 30–45 s. The subjects performed different gait speeds with ≥ 2 min rest between trials.

2.3. Multi-channel surface EMG recording

Multi-channel surface EMG signals were recorded from the RF muscle of the right thigh with 24 liner electrode array (1 × 5 mm, 10-mm inter-electrode distance) (ELSH004, OT Bioelectronica, Torino, Italy) (Fig. 1). Electrode array was arranged along longitudinal line of the muscle, since region-specific neuromuscular activation was mainly demonstrated along the RF muscle longitudinally in our previous studies (Watanabe et al., 2012, 2013, 2014a, in press). A conductive gel was applied within the cavities of the grid electrodes to ensure appropriate skin contact. Prior to attaching the electrode grid, the skin was shaved, abraded, and cleaned with alcohol. To determine electrode positions, the edge of the superficial region of the RF muscle was identified using ultrasonography (FAZONE CB, FUJI FILM, Tokyo, Japan). We identified the border between the RF muscle and neighboring muscles, i.e., vastus lateralis, vastus medialis, sartorius, and tensor fasciae latae muscles, was identified, and marks were applied to the skin above the border using a felt-tip pen under the guidance of the real-time axial ultrasonographic images. Electrodes were attached within superficial regions of the RF muscle were surrounded by the marks on the skin. Subjects' hip and knee joint angles were both 90° (180° is fully extended) during this procedure. The columns of electrodes were placed on the longitudinal axis of the RF muscle along a line between the anterior superior iliac spine and the superior edge of the patella. We defined the line between these two points as the longitudinal line of the RF muscle based on anatomical data on the human RF muscle (Sung et al., 2003; Yang and Morris, 1999). The center of the second electrodes from the proximal side placed in the proximal third along the longitudinal line of the RF muscle (Fig. 1). A reference electrode was placed at the iliac crest. These procedures for recording surface EMG from the RF muscle were used in our previous studies (Watanabe et al., 2012, 2013, 2014a, in press).

Detected EMG signals with monopolar recording were amplified by a factor of 1000, sampled at 2048 Hz with an 8th order Bessel band pass filter at 10–750 Hz (anti-aliasing filter), and converted to digital form by a 12-bit analog-to-digital converter (EMG-USB, OT Bioelectronica, Torino, Italy). In off-line analysis (MATLAB 7, MathWorks GK, Tokyo, Japan), the detected monopolar signals were filtered (band-pass filter, 20–400 Hz). In order to remove motion artifacts, the high pass frequency was set at 20 Hz (De Luca et al., 2010). Bipolar surface EMG signals were calculated from the electrode pairs between neighboring electrodes along the rows (Watanabe et al., 2013). Eighteen bipolar surface EMG signals were obtained (3 pairs × 6 array electrodes) from two lines of 6 × 4 array electrodes in this study (Fig. 1).

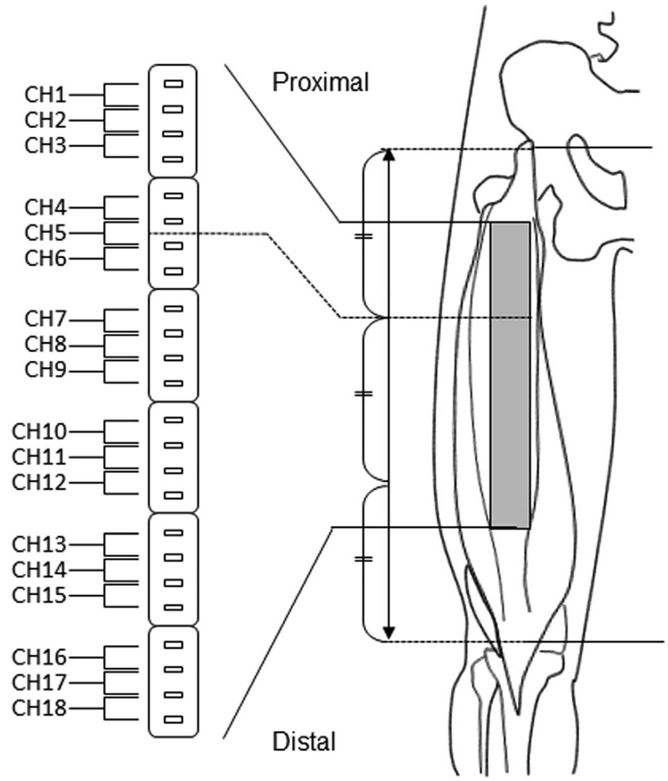

Fig. 1. Electrode positions and definitions of channel numbers for the rectus femoris muscle.

During trials, electrical signals from the footswitches which are taped to the heel and toe of the right foot were synchronized with surface EMG signals using an analog-to-digital converter. From these signals, timings of heel contact and toe-off were identified for analysis of surface EMG. Heel contact of right foot was defined as start and end of a gait cycle. Twenty consecutive strides were sampled for analysis during the constant phase of each trial. For each gait speed, each electrode pair and each subject, rectified EMG amplitude were averaged (averaged rectified value; ARV) every 2% of a stride across the 20 strides. The peak ARV among three different gait speeds was determined for each channel for each subject. ARVs for each 2% of a gait cycle were normalized by the peak ARV for each subject. We consequently calculated 18 normalized ARV and they were defined as CH1 to CH18 from the proximal side as shown in Fig. 1. Spatial distribution of ARV along the muscle was assessed and quantified from central locus activation (CLA). At every 2% of a gait cycle, CLA was calculated as the centroid of the normalized ARV along the longitudinal line of the muscle in inter-electrode distance units (Watanabe et al., 2012, 2014b, in press). Our electrode arrangement includes blanks between arrays. We thus replace ARVs at these blanks with the averaged values between CH3 and 4, CH and 7, …, CH15 and 16. The results are shown as the distance (cm) from the most proximal edge of electrodes.

Knee joint angle of right leg was also measured with an electro-goniometer (FA-DL-210, 4 assist, Inc. Tokyo, Japan). The knee joint angle was defined as the angle included between the two lines assumed as thigh and shank segments using anatomical markers, i.e., the greater trochanter, lateral femoral epicondyle, and lateral malleolus (Watanabe and Akima, 2011). This signal is synchronized with the EMG data.

2.4. Statistics

We used non-parametric tests since the sample size was not large (n=7 and 8) and data distribution was partly non-Gaussian. CLA values at every 2% of a gait cycle were equally divided into ten phases (Watanabe et al., 2014b). In each gait speed, CLA values at each phase were compared between the young and elderly groups using the Wilcoxon rank sum test. When there is a significant difference in CLA between the groups, normalized ARV for proximal, middle, and distal regions were calculated and compared between young and elderly groups using the Mann–Whitney test at the phase for detailed analysis of spatial distribution of EMG. ARV for proximal, middle, and distal regions are mean values of CH 1–6, CH 7–12, and CH 13–18, respectively. Gait speed, cadence, the timing of toe-off, and mean knee