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Regional neuromuscular regulation within human rectus femoris muscle during gait

Kohei Watanabe [a,*], Motoki Kouzaki [b], Toshio Moritani [c]

[a] School of International Liberal Studies, Chukyo University, Nagoya, Japan
[b] Laboratory of Neurophysiology, Graduate School of Human and Environmental Studies, Kyoto University, Kyoto, Japan
[c] Laboratory of Applied Physiology, Graduate School of Human and Environmental Studies, Kyoto University, Kyoto, Japan

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ABSTRACT

The spatial distribution pattern of neuromuscular activation within the human rectus femoris (RF) muscle was investigated during gait by multi-channel surface electromyography (surface EMG). Eleven healthy men walked on a treadmill with three gait speeds (4, 5, and 6 km/h) and gradients (0°, 12.5°, and 25°). The spatial distribution of surface EMG was tested by central locus activation (CLA), which is calculated from 2-D multi-channel surface EMG with 46 surface electrodes. For all conditions, CLA was around the middle regions during the swing-to-stance transition and moved in a proximal direction during the stance phase and stance-to-swing transition ($p < 0.05$). CLA during the stance-to-swing transition and early swing phase significantly moved to proximal site with increasing gait speed ($p < 0.05$). During the early stance and swing phases, with increasing grade, CLA significantly moved distally ($p < 0.05$). These results suggest that the RF muscle is regionally activated during a gait cycle and is non-uniformly regulated longitudinally.

1. Introduction

The rectus femoris (RF) muscle is one of the four components of quadriceps femoris muscle group and the only bi-articular muscle of the group, crossing the hip and knee joints. In gait analyses, the RF muscle has often been the subject of intense investigation, since it has been reported that an abnormality in regulation of neuromuscular activation of the RF muscle causes pathological gait in patients with upper motor neuron injuries such as stiff-legged (or -knee) gait (Chapman et al., 2008; Kerrigan et al., 1991; Reinbolt et al., 2008; Riley and Kerrigan, 1998; Sung and Bang, 2000). While numerous studies have investigated the neuromuscular activation pattern of the RF muscle during gait using electromyography (EMG) in able-bodied adults, children, the elderly, and patients (Annaswamy et al., 1999; Barr et al., 2010; Byrne et al., 2005; Chantraine et al., 2005; Di Nardo and Fioretti, 2013; Nene et al., 2004; Sung and Bang, 2000), little attention has been paid to the complicated anatomical characteristics of this muscle. It has been suggested that the RF muscle is comprised of two different muscle-tendon units or neuromuscular compartments, and this anatomical property is referred to as "muscle-within-muscle" (Balius et al., 2009; Gyftopoulos et al., 2008; Hasselman et al., 1995). Muscle fibers of the proximal one third and remainder of the RF muscle arise from two different proximal tendons which attach at the anterior inferior iliac spine and superior acetabular ridge, respectively (Balius et al., 2009; Gyftopoulos et al., 2008; Hasselman et al., 1995). Also, the proximal region and remainder of the human RF muscle are separately innervated by different motor nerve branches (Sung et al., 2003; Yang and Morris, 1999). Based on these anatomical characteristics, it can be assumed that the two muscle-tendon units within the RF muscle are controlled via different strategies by the central nervous system and play different functional roles. A previous study also suggested that the two muscle-tendon units are independently and/or regionally activated during human movements (Hasselman et al., 1995). Currently, no solid experimental evidence exists regarding the quantitative as well as qualitative differences in neuromuscular activation among regions within the RF muscle during gait based on the specific anatomical characteristics.

Recent studies employed multi-channel surface EMG to record and assess the neuromuscular activation of large areas of the muscle (Holtermann et al., 2008; Staudenmann et al., 2009; Vieira et al., 2010; Watanabe et al., 2012). This technique would provide important information to understand the neural control of bi-articular muscles, the normal EMG pattern of the RF muscle during gait, and mechanisms of gait disorders in the elderly and/or disabled persons.

We investigated the neuromuscular activation of the whole RF muscle during gait using multi-channel surface EMG. We hypothesized that the RF muscle is regionally activated and non-uniformly

* correspondence to: School of International Liberal Studies, Chukyo University, Yagotohonmachi, Showa-ku, Nagoya 466-8666, Japan.

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regulated along its longitudinal line during gait because of the potential region-specific activation based on the anatomical properties (Balius et al., 2009; Gyftopoulos et al., 2008; Hasselman et al., 1995).

2. Methods

2.1. Subjects

Subjects comprised 11 healthy men (age: 22.6 ± 2.5 years, height: 170.9 ± 6.6 cm, body mass: 66.7 ± 9.1 kg). We selected only male subjects because of thinner subcutaneous fat tissue than female subjects. They gave 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 Graduate School of Human and Environmental Studies, Kyoto University.

2.2. Experimental design

Subjects walked on a treadmill (MEDTRACK ST65, Quinton Instrument Co., WA, USA) set with five different combinations of various speeds and grades: (1) 4 km/h at a grade of 0°, (2) 5 km/h at a grade of 0°, (3) 6 km/h at a grade of 0°, (4) 5 km/h at a grade of 12.5°, and (5) 5 km/h at a grade of 25°. To investigate possible region-specific neuromuscular responses during gait, various speeds and grades were given in the present study. Each task involved a phase with a gradual increase in speed and/or grade for 15–20 s until reaching the specified conditions, which were then maintained for 30–45 s. Rest periods among the trials under the different conditions were ≥ 2 min.

2.3. Multi-channel surface EMG recording

Multi-channel surface EMG signals were recorded from the RF muscle of the left thigh with a matrix of 48 surface electrodes comparing 2 columns and 24 rows (1 × 5 mm, 10-mm inter-electrode distance) (ELSH004, OT Bioelectronica, Torino, Italy) (Fig. 1). This electrode arrangement is similar to that in our previous study (Watanabe et al., 2013). Since region-specific neuromuscular activation was mainly demonstrated along the RF muscle longitudinally in our previous studies (Watanabe et al., 2012–2014), electrodes were also arranged longitudinally.

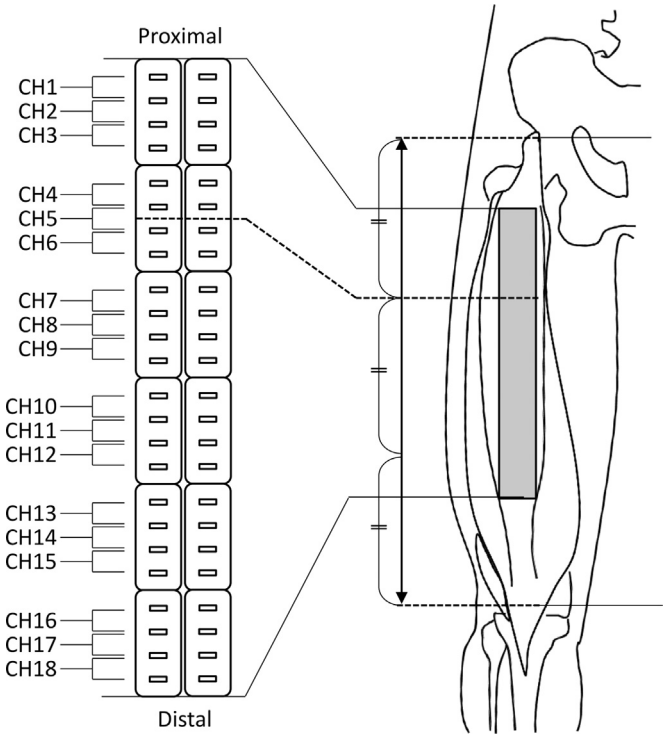

Fig. 1. Electrode positions and definitions of channel numbers for the rectus femoris muscle.

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). The ultrasonographic procedure to determine electrode positions was similar to that employed in our previous studies (Watanabe et al., 2012–2014). On real-time axial ultrasonographic images, 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. Consequently, superficial regions of the RF muscle were surrounded by the marks on the skin, and electrodes were attached within them. During this procedure, subjects' hip and knee joint angles were both 90° (180° is fully extended). 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. The line between these two points was defined 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. In our previous study, we confirmed that multi-channel surface EMG of the RF muscle, which was recorded with this procedure, demonstrated a region-specific activation pattern similar to intramuscular EMG pattern from multiple regions of the muscle (Watanabe et al., 2012).

Monopolar EMG signals 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). Monopolar surface EMG signals were off-line band-pass filtered (20–400 Hz) and transferred to analysis software (MATLAB 7, MathWorks GK, Tokyo, Japan). The high pass frequency was set at 20 Hz in order to remove motion artifacts (De Luca et al., 2010). From the electrode pairs between neighboring electrodes along the rows, bipolar surface EMG signals were calculated (Watanabe et al., 2013). Since we used two lines of 6 × 4 array electrodes in this study, 36 bipolar surface EMG signals were obtained (3 pairs × 6 array electrodes × 2 lines) (Fig. 1).

For analysis of surface EMG, timings of heel contact and toe-off were identified by signals from footswitches taped to the heel and toe of the left foot. Electrical signals from the footswitches were synchronized with surface EMG signals using an analog-to-digital converter. One heel contact to the next heel contact was determined as one step and considered as 100% of a gait cycle for each stride. During the constant phase of each trial, 20 consecutive strides were sampled for analysis. The root mean square (RMS) of surface EMG was calculated from the sampled surface EMG signal. To average and normalize RMS values, the following procedure was performed: (1) for each gait condition, each electrode pair and each subject, RMS values were averaged every 2% of a stride across the 20 strides; (2) RMS values of the same rows were averaged for each subject and gait condition; (3) for each subject and gait condition, the peak RMS value was determined from 50 averaged RMS values of each 2% of a gait cycle; (4) RMS values for each 2% of a gait cycle were normalized by the peak value for each subject and gait condition. Consequently, 18 normalized RMS values were obtained at every 2% of a gait cycle, defined as CH1 to CH18 from the proximal side (Fig. 1). To quantify the spatial distribution of RMS within the muscle, central locus activation (CLA) was also calculated at every 2% of a gait cycle. CLA was calculated as the centroid of the normalized RMS along the longitudinal line of the muscle in inter-electrode distance units. Since we used an array electrode with four electrodes, there were blanks between arrays where we were unable to detect a surface EMG signal. For CLA calculation, these blanks were considered and the results are shown as the distance (cm) from the most proximal edge of electrodes. Therefore, for calculating CLA we added the averaged values between CH3 and 4, CH and 7, …, CH15 and 16 as the estimated RMS values between arrays.

2.4. Statistics

Non-parametric tests were employed since the sample size was not large ($n=11$) and data distribution was partly non-Gaussian. CLA values at every 2% of a gait cycle were equally divided into ten phases. In each gait phase, CLA values were compared among the different speeds and grades using the Wilcoxon rank sum test. For each condition, CLA values in individual gait phases were compared with the CLA value in gait phase 1 using the Wilcoxon rank sum test. Cadence and the timing of toe-off were compared among conditions using Wilcoxon rank sum test. The level of significance was set at 0.05 and modified by Bonferroni correction, i.e., $\alpha=0.05$/number of pairs. Statistical analyses were performed using SPSS software (version 15.0; SPSS, Tokyo, Japan).

3. Results

The mean cadence and timing of toe-off for each condition are shown in Table 1. There were significant differences in cadence