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Quantitative and Qualitative Evaluation of Knee Electromyograms by a Bluetooth-communication Gait Analyzer: Integration and Power Spectral Analysis of Surface Electromyograms

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Abstract. [Purpose] We observed the action of muscles during gait using a Bluetooth Gait Analyzer (sampling frequency 1 kHz), by integrating electromyograms and performing power spectral analysis. Analyzer was described in the quantitative and qualitative evaluation of these muscles. [Subjects] The subjects were 15 healthy adult males. [Methods] A Bluetooth electromyography gait analyzer was used to observe the action of the vastus medialis muscle (Quad muscle) and the long head of the biceps femoris muscle (Hamstrings) during slow, medium, and fast gait. Electromyograms were subjected to integration and power spectral analysis evaluation. [Results] The integrated electromyograms of the Quad muscles and the Hamstrings increased significantly with increases in gait speed. Power spectral analysis showed the mean power frequency of the Quad muscles tended to decrease, and the high frequency component (81–250 Hz) was significantly reduced. The mean power frequency of the Hamstrings increased with increases in gait speed, but the high frequency component was significantly increased. [Conclusion] The observed tendencies might have resulted from the induction of positive action in type I muscle fibers in the Quad muscles with increased gait speed and of the muscle action of type II fibers in the Hamstrings. We suggest that, as the gait speed increases, the Quad muscles are forced to play the role of a braking muscle.
Key words: Bluetooth Communication, Surface Electromyography, Quantitative/Qualitative Gait Evaluation

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INTRODUCTION

Our joint Japan-US research project is investigating a comprehensive anthropometric evaluation system. Our aim is to prepare a cost-effective system that is highly reliable, reproducible, and valid in clinical practice. This evaluation tool includes an electromyograph, an accelerometer[1–6)], a goniometer, a dynamometer[7,8)] a thermometer, a muscle hardness meter[9–11)], and a pain meter[12,13)]. We are also aiming to develop a system in which these sensors are connected wirelessly to reduce the restrictions imposed by the use of connection cords. In addition, the system will be simple to wear, and capable of employing various sensors in various combinations.

Currently, a variety of evaluation sensors are wired together for use in many of the evaluation tools employed for clinical evaluation and investigation, and many of the evaluation tools used in clinical practice or in research employ large numbers of sensor wires. This restrictive wiring can limit subjects' dynamic movements, for example, when walking. On the other hand, in many wireless communication tools, an individual sensor is used. Various sensors including accelerometers and electromyographs are connected for simultaneous analysis at a sampling frequency of 1 kHz[14]. The frequency characteristics of these sensors are such that a sampling frequency as high as 1 kHz is not required, even when many sensors are used in combination. However, the Nyquist theorem states that, for analogue-to-digital conversion of a surface electromyogram, the highest frequency component of the signal to be sampled should be less than half the sampling frequency. Therefore, the sampling frequency of electromyogram signals needs to be 1 kHz or higher. This is especially important in frequency analysis.

We used a Bluetooth device for gait analysis. Bluetooth is a near field communication technology, which can communicate at a rate of 1 Mbps via a 2.45 GHz band radio wave. The 0.5-square inch compact transceiver used in this device reduces power consumption and production costs, and provides higher speed communications than other communication devices. We previously reported the reliability and application of a gait analyzer with a Bluetooth-communication 3-axis accelerometer and EMG[15]. In that previous study, time factors of the gait cycle were calculated according to the acceleration waveforms obtained at the third lumbar vertebra and knee joint by the system. We demonstrated the reliability, reproducibility, and validity of these factors and discussed future clinical applications.

In this study, we observed the action of the knee joint muscles during gait using the gait analyzer, at a sampling frequency of 1 kHz, by integrating the electromyogram and performing power spectral analysis. Use of the analyzer is described in the following quantitative/qualitative evaluation of these muscles.

SUBJECTS AND METHODS

The subjects were 15 healthy adult male subjects who had no history of trauma that might affect the lower extremities in gait or cerebral functions (mean age 21.8 years, ranging from 19 to 27, mean body weight 63.1 ± 7.9 kg, mean height 174.3 ± 5.8 cm). The subjects all provided their informed written consent.

The subjects were dressed in shorts so that their knees were exposed and had bare feet. Our analyzer was capable of performing 8 channel measurements from various sensors including an acceleration sensor and an electromyograph. In this study, measurements were obtained through 4 channels: of (i) the right quadriceps femoris muscle (vastus medialis muscle) surface electromyogram, (ii) the right hamstrings (long head of the biceps femoris muscle) surface electromyogram, (iii) a right heel pressure sensor (A201-25, Nitta), and (iv) the examiner's push switch (custom-made, Holonic).

The activity of the quadriceps femoris muscle and hamstrings were measured using a balanced input with an input resistance of 100 MΩ or more, frequency characteristics of DC to 500 Hz, a 1000-fold amplification factor (fixed), a CMMR of –95 db or more, and an anti-polarization potential of ±500 mV AMP (custom-made, Holonic). The electrodes were pre-treated to a skin impedance of 5 kΩ or less and placed on the right vastus medialis muscle [(i) the middle of the lower 1/3 between the knee-joint fissure and the greater trochanter, (ii) the mid point between the femoral midline and the most medial edge, and the intersection point between (i) and (ii)] and the long head of the right biceps femoris muscle [(i) the bottom of the mid 1/3 between the knee-joint fissure and the greater trochanter, (ii) rather lateral to the femoral midline, and the intersection point between (i) and (ii)] with a distance of 3 cm between electrodes. The pressure sensor of the heel contact timing switch was placed on the right heel, and the examiner used a hand switch to identify acceleration and deceleration during the early and end phases of gait, respectively. These 4 sensors were connected to a Bluetooth transmitter (103 × 59 × 27 mm; 126 g; A/D converting part with a resolution of 12 bits, MSP430FG439 Texas Instruments, USA, sampling frequency 1 kHz; transmitter/receiver part with Bluetooth class 2 transmission, transmit/receive frequency of 2.4 GHz, MES-01, Holonic) and then

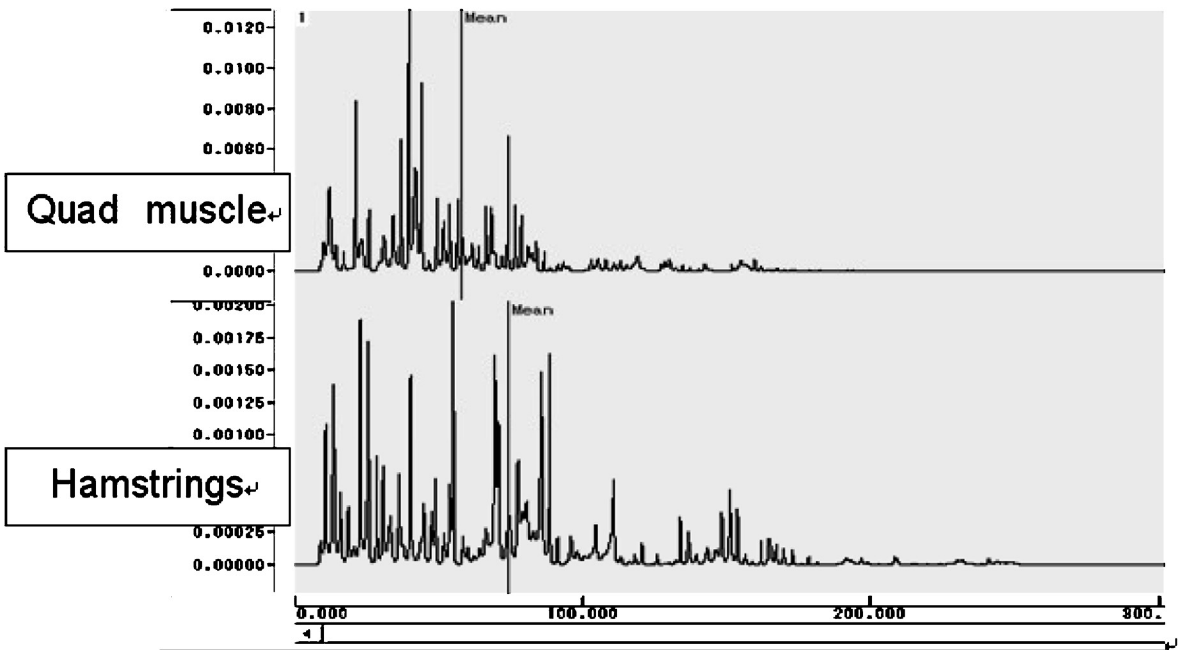

Fig. 1. Maximum Entropy Method.
The upper graph shows a raw data sample for the power spectral analysis of the Quad muscle (vastus medialis muscle), while the lower graph shows one for Hamstrings (long head of the biceps femoris muscle).

fixed to the subjects' anterior lumbar region with a belt. After A/D conversion with a quantized 12-bit converter, the data were sent to a laptop PC equipped with a receiver (Hagiwara HNT-UB03, Hagiwara Sys-Com) via a Bluetooth transmitter.

Subjects walked along a 16-m course (a 3-m acceleration stage followed by a 10-m measurement stage and a 3-m deceleration stage) a total of 9 times: three periods at a fast gait (FG), three at a medium pace gait (MD), and three at a slow gait (SL). In accordance with previous studies[1–6], the subjects were instructed as follows: (i) FG: walk as fast as possible without running, (ii) MD: adopt your usual walking speed, and (iii) SL: walk slowly as if, for example, waiting at a bus stop or window-shopping. Then, training was provided to enable the subjects to learn these gait velocities. During the gait analysis, when the anterior surface of a subject's trunk passed a line at the end of the acceleration stage and at the beginning of the deceleration stage, this was identified with a signal from a wired push button operated by an accompanying second examiner. Although an infrared gate sensor is normally used as a switch to remove the acceleration phase occurring early in the gait cycle and the deceleration phase occurring toward the end of the cycle, here we used a manual switch because of difficulty in synchronizing the sensor with the Bluetooth device.

Signals sent to the laptop PC equipped with a Bluetooth receiver were identified as heel contact and were captured with communication information collection software (HolonicBio HOL-01, Holonic) and a waveform viewer program (VitalTracer D20071120, KISSEI COMTEC) using a multi-purpose bio-information analysis system (BIMUTAS II H0504011, KISSEI COMTEC) at a sampling frequency of 1 kHz. According to the myoelectric data collected at each gait speed, an upstroke at the pressure sensor on the heel employed for heel contact timing was used to sample 10 consecutive steps, which constituted a gait cycle, followed by band-pass (10–250 Hz) and band-stop (49.5–50.5 Hz) processing.

The integrated EMG (I-EMG) of a complete gait cycle was performed after full wave rectification to average the integral value of 10 steps between one heel contact and the next.

Power spectral analysis of each muscle was undertaken using the maximum entropy method (MEM) for the data of 10 gait cycles (Fig. 1). Then, the mean power frequencies (MePF) were calculated at the respective speeds and the frequency content for the respective muscles were calculated and categorized as low (10–40 Hz), medium (41–80 Hz), or high frequency (81–250 Hz) according to Nagata[16].

SPSS (SPSS16.0 for Windows) was used for the statistical analysis. The Friedman test was used to test the validity of the difference in the amounts of muscle action for the different velocities and the change in MePF and the content of the frequency components in electromyograms obtained for each muscle at different walking speeds. This was followed by a multiple comparison using Wilcoxon's signed rank test (significance level of < 5%).

Subjects were informed about the objectives and details of this study, the benefits and risks, the protection of their personal information, and their rights to refuse or withdraw from participation. They then provided their signed written consent. This study was submitted for review and approved by the Institutional Review Board (IRB) of Hachiohji Health Cooperative Shiroyama Hospital (Registered No. A000610231).

RESULTS

The I-EMG of the vastus medialis muscle and the long head of the biceps femoris muscle (Table 1) increased with gait speed, and all the increases were revealed to be significant by Friedman's test and Wilcoxon's test (p<0.01).

The mean power frequency of the vastus medialis muscle showed a tendency to decrease with increases in gait speed (Friedman test, p<0.07). Wilcoxon's test revealed significant differences between slow and medium gait and between slow and fast gait (p<0.05). The mean power frequency of the long head of the biceps femoris muscle increased significantly with an increase in gait speed (Friedman's test, p<0.05). Wilcoxon's test revealed significant differences between slow and fast gait (p<0.05) and between medium and fast gait (p<0.01) (Table 2).

Table 3 shows the frequency content of the vastus medialis muscle EMGs. The mean frequency content tended to increase in the low frequency component (10–40 Hz) and medium frequency component (41–80 Hz) with increases in gait speed, but Friedman's test revealed no significant difference (p<0.6 to p<0.8). The high frequency component (81–250 Hz) was significantly reduced with an increase in gait speed (p<0.05). Wilcoxon's test revealed significant differences between the high frequency components for slow and medium gait and for slow and fast gait (p<0.05).

The frequency content of the low frequency component of the long head of the biceps femoris muscle EMGs decreased with increases in gait

Table 1. Integrated Electromyography

	Slow	Medium	Fast
Quad (mV·msec)	134.0 ± 79.6	181.8 ± 81.6	426.6 ± 158.8
Ham (mV·msec)	105.0 ± 61.1	119.3 ± 71.4	283.4 ± 101.0

The I-EMGs of the Quad muscle (vastus medialis muscle) and Hamstrings (long head of the biceps femoris muscle) increased significantly with increases in gait speed (Friedman's test, Wilcoxon's test; p < 0.01).

Table 2. Mean power frequency (MePF)

	Slow	Medium	Fast
Quad (Hz)	65.7 ± 4.9	61.3 ± 6.1	60.5 ± 7.0
Ham (Hz)	73.7 ± 9.9	72.8 ± 10.6	79.4 ± 11.1

The mean power frequency of the Quad muscle (vastus medialis muscle) tended to decrease with increases in gait speed (Friedman's test; p<0.07). Wilcoxon's test revealed a significant difference between slow and medium gait and between slow and fast gait (p<0.05). The mean power frequency of Hamstrings (long head of the biceps femoris muscle) increased significantly with increases in gait speed (Friedman's test; p<0.05). Wilcoxon's test revealed a significant difference between slow and fast gait (p<0.05) and between medium and fast gait (p<0.01).

Table 3. Frequency content

	Slow	Medium	Fast
LF-quad (%)	39.2 ± 6.4	42.7 ± 8.7	42.8 ± 9.3
LF-ham (%)	38.6 ± 9.8	39.0 ± 10.2	32.6 ± 9.8
MF-quad (%)	33.7 ± 6.6	33.9 ± 6.7	35.0 ± 6.1
MF-ham (%)	28.1 ± 4.3	28.3 ± 4.9	29.5 ± 4.5
HF-quad (%)	27.0 ± 4.9	23.3 ± 6.1	22.3 ± 6.6
HF-ham (%)	33.3 ± 8.3	32.7 ± 8.0	37.9 ± 8.2

The content of low (10–40 Hz) and medium (41–80 Hz) frequency components increased with increases in gait speed, while that of the high frequency component (81–250 Hz) was significantly reduced in the Quad muscle (vastus medialis muscle, p<0.05). In Hamstrings (long head of the biceps femoris muscle), the content of the low frequency component was significantly reduced (p<0.01), while that of the high frequency component significantly increased with increases in gait speed (p<0.01).

speed ($p<0.01$), while that of the high frequency component increased significantly ($p<0.01$). Wilcoxon's test results showed significant differences between the low frequency component content for slow and fast gait ($p<0.05$) and that for medium and fast gait ($p<0.01$), and a significant difference was also observed in the content of the high frequency components for medium and fast gait ($p<0.01$).

DISCUSSION

In this study, we achieved a sampling frequency of 1 kHz, which is rare for conventional electromyogram measurements via wireless communication, using Bluetooth electromyography for gait analysis. The I-EMGs obtained during gait were measured using this apparatus and the power spectral analysis results were valid. The Bluetooth-communication electromyography that we used for gait analysis achieved a sampling frequency of 1 kHz, which is rare for conventional wireless measurement of myopotential. We discuss the integral value of the electromyogram obtained by using this system during gait and the results of the frequency analysis.

Using a multi-purpose bio-information analysis program, we randomly sampled data of 10 gait cycles, which we defined as one heel contact to the next heel contact, during the 10-m gait stage at various gait speeds. The I-EMG of the vastus medialis muscle and the long head of the biceps femoris muscle (Table 1) increased significantly as the gait speed changed from slow to fast ($p<0.01$). The muscle strength is considered to depend on the motor unit recruitment and motor neuron firing frequency. The I-EMG increases linearly with an increase in muscle contraction[17)], while the muscle strength increases somewhat compared with the mean electromyogram at around 80% of the maximal voluntary contraction (MVC)[18–20)], suggesting that the change in muscle contraction can be determined quantitatively. Our results may accurately show the quantitative change in muscle activity with change in gait velocity. However, we did not measure the maximum muscle strength at 100% MVC, the muscle strength at 50% MVC, or the % integrated electromyogram (%I-EMG) at these contractions. Moreover, we did not compare it with the I-EMG obtained when the subject was walking. We should investigate their relationships in future studies.

The mean power frequency of the vastus medialis muscle showed a tendency to decrease with increasing gait speed (Table 2). The content of the low (0–40 Hz) and medium (41–80 Hz) frequency components increased, while that of the high frequency component (81–250 Hz) decreased significantly ($p<0.05$) (Table 3). Meanwhile, the mean power frequency of the long head of the biceps femoris muscle increased significantly with increases in gait speed (Table 3). The content of the low frequency component decreased significantly with increasing gait speed ($p<0.01$), while that of the high frequency component increased significantly ($p<0.01$) (Table 3). In previous studies in which the signal waveform of an electromyogram was resolved into its frequency components [16,21,22)], it was suggested that the slow (type I fiber) and fast (type IIa and IIb fibers) muscle fibers were responsible for the low and high frequency band components, respectively, enabling quantitative evaluation. Therefore, our results suggest that a positive action of type I muscle fiber might be induced in the vastus medialis muscle by increases in gait speed, while the action of type II muscle fiber might be induced in the biceps femoris muscle.

As described above, muscle strength may increase as a result of adjustment of fiber types the total number of the motor units recruited and an increase in the motor neuron discharge frequency (rate coding). Milner-Brown et al.[23)] noted that motor unit recruitment is dominant during weak muscle contraction, while an increased discharge frequency is increasingly apparent during stronger muscle contraction. Also, Henneman et al.[24)] proposed the size principle that type I fibers with a smaller diameter are initially recruited at the beginning of muscle contraction followed by the recruitment of type II fibers with a larger diameter with increased contraction strength. In this study, we obtained a result in which there was increased gait velocity despite increased amounts of muscle action in the vastus medialis muscle, which is inconsistent with these earlier reports[23,24)]. This suggests that an increase in gait speed forced the vastus medialis muscle to act as a braking muscle, while the biceps femoris muscle acted as an accelerating muscle. The vastus medialis muscle exhibits an eccentric contraction upon heel contact in the gait cycle to absorb impact with the floor. The mode of muscle contraction at this moment[25)] and exercise may be influential factors. Yao et al.[26] noted that the low frequency component became dominant with increased synchronization of the motor units. De Luca[27] reported that different muscles showed different motor unit recruitment and different proportions of discharge frequency. Our results may have arisen from muscle contraction (concentric or eccentric)[25], motor unit recruitment, and proportions of discharge frequency[27], in contrast to the report by Yao et al.[26].

Hof et al.[28] and Otter et al.[29] conducted a detailed analysis of the relationship between gait speed profile and muscle action in each phase of the gait cycle. However, our subjects did not wear a goniometer, so we could not identify the phases of the gait cycle. Analysis of each phase of the gait cycle is also required for qualitative examination through frequency analysis of the electromyogram in each phase.

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