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A Study on Human Musculoskeletal Model for Cycle Fitting: Comparison with EMG

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***Abstract*—**It is difficult to study the effect of various variables on cycle fitting through actual experiment. To overcome such difficulty, the forward dynamics of a musculoskeletal model was applied to cycle fitting in this study. The measured EMG data weres compared with the muscle activities of the musculoskeletal model through forward dynamics. EMG data were measured from five cyclists who do not have musculoskeletal diseases during three minutes pedaling with a constant load (150 W) and cadence (90 RPM). The muscles used for the analysis were the Vastus Lateralis (VL), Tibialis Anterior (TA), Bicep Femoris (BF), and Gastrocnemius Medial (GM). Person's correlation coefficients of the muscle activity patterns, the peak timing of the maximum muscle activities, and the total muscle activities were calculated and compared. BIKE3D model of AnyBody (Anybodytech, Denmark) was used for the musculoskeletal model simulation. The comparisons of the actual experiments with the simulation results showed significant correlations in the muscle activity patterns (VL: 0.789, TA: 0.503, BF: 0.468, GM: 0.670). The peak timings of the maximum muscle activities were distributed at particular phases. The total muscle activities were compared with the normalized muscle activities, and the comparison showed about 10% difference in the VL (+10%), TA (+9.7%), and BF (+10%), excluding the GM (+29.4%). Thus, it can be concluded that muscle activities of model & experiment showed similar results. The results of this study indicated that it was possible to apply the simulation of further improved musculoskeletal model to cycle fitting.

***Keywords*—**Cycle fitting, EMG, Musculoskeletal modeling, Simulation.

I. Introduction

THE cycling population is explosively increasing, keeping up with health and environmental issues. Cycling is a sport recommended for the aged, osteoporotic patients, and overweighted people. It helps to strengthen the muscular power of the lower body and the cardiopulmonary function, and maintain and improve physical strength [1]. However, if improper pedaling postures and pedaling loads are used, the possibility of injury increases.

Accordingly, many studies have been conducted on the cycle fitting to determine the proper pedaling postures. For example, Bae [2] developed a riding machine that can automatically control the frame size for proper pedaling posture. Umberger [3] compared the pedaling forces at four different seat tube angles (69, 76, 83, and 90°) and showed that the pedaling force increases as the angle decreases. Bini [4] proved that knee injury risk in pedaling can be decreased by adjusting the saddle height to achieve the knee angle of 25-30°. Choi [5] showed that an increased saddle height directly affects the range of motion of the joint and the muscle length. Macdermid & Edward [6] reported that the 170mm pedal arm produces higher peak power and reaches the peak power faster than 172.5 and 175mm pedal arms. According to such studies, the variables that can cause injuries and exercise effects during pedaling are diverse such as the seat tube angle, saddle height, and pedal arm length. To investigate the influence of all such variables, repeated experiments that control each variable are needed, and it is also difficult to obtain consistent test results.

The variables to be controlled by musculoskeletal model can be done more easily than by actual experiment. Musculoskeletal model has several advantages. For example, though the numbers of electromyography (EMG) channels are limited, a musculoskeletal model can identify all muscle activities. Due to such merit, studies have been conducted on finger muscles and tendons [7] and the muscle force after lower limb treatments [8]. There are mainly two kinds of musculoskeletal models. One is the forward dynamics model, in which the subject's physical segment length and experimental condition are directly used. The other is the inverse dynamics model, which uses the subject's physical segment length, motion, and pedal forces obtained through experiment. The inverse dynamics calculates several results by using the forces applied to the pedal, so more precise and diverse results can be estimated, but it is disadvantageous since it requires actual experiments. Therefore, in prior cycle fitting estimations, the forward dynamics is considered more effective than the inverse dynamics since diverse variables can be controlled and applied through simple anthropometric measurement without actual pedaling experiments.

The forward dynamics of a musculoskeletal model is based on its physical size and the kinematical condition. It analyzes muscle powers and joint moments numerically due to various movements. Furthermore, if it is applied to cycling, simulations are possible by controlling the body height, weight, and length of each segment of the subject, as well as the cycling variables to be fit. That is, if an optimal model applicable to cycle fitting is established, the actual experiments can be minimized.

There have been many studies on such musculoskeletal models applied to cycling. Jeffery [9] investigated the crank power and the timing of muscle activities depending on the

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chain type. Rasmussen [10] used the forward dynamics to study the changes in the tendon energy of the lower limb muscles and the foot angles according to the changes in the tendon elasticity values during pedaling. Liu [11] applied the inverse dynamics to study the increase in the iliac muscle power and a decrease in the soleus muscle power depending on the trunk bending. Raasch [12] showed that the rectus femoris is activated when the maximum pedaling changes from upward to downward, and the hamstring is activated when the maximum pedaling is changed from downward to upward. However, studies on musculoskeletal models have mainly focused on the changes in the pedaling power and muscle activity. There are insufficient studies on cycle fitting using a musculoskeletal model based on the changes in the physical sizes and the cycle fitting variables. Therefore, in this study, the measured muscle activities while pedaling are compared with those of a musculoskeletal model by the forward dynamics simulation.

II. Method

A. Musculoskeletal Model Simulation

The BIKE3D model of AnyBody (Anybodytech, Denmark) software was used in this experiment. The model consisted of 25 bones and 464-hill-type muscles. The physical information of the subjects (their height, weight, and the length of the trunk, thigh, tibia, upper arm and forearm, hand & foot sizes, pelvic width, and positions of saddle and handle, pedal arm length and width) were collected and applied to the model. The measured values obtained by controlling the cadence and load while pedaling were applied to the model to calculate the muscle activities under the same condition (Fig. 1).

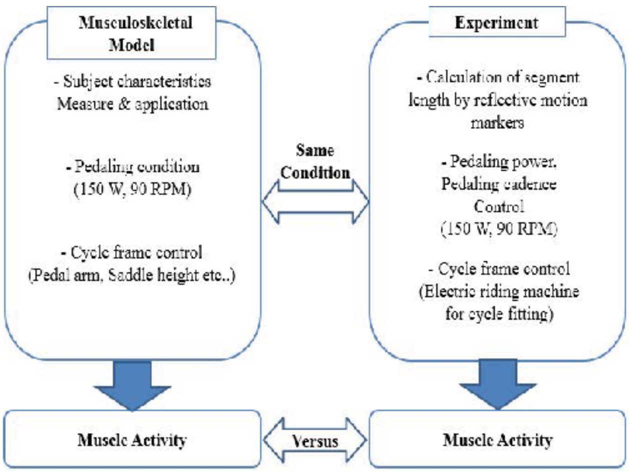


Fig. 1 Simulation and experimentation

TABLE I
PHYSICAL CHARACTERISTICS OF THE SUBJECTS

	Height (cm)	Weight (kg)	Thigh (cm)	Trunk (cm)	Tibia (cm)	Foot (cm)
Mean	174.8	82.2	43.0	88.0	42.8	26.6
SD	±4.7	±5.1	±1.1	±3.9	±1.1	±0.7

B. Subjects

In this study, the experiment was performed with five elite cyclists who do not have injuries for a recent year. Their physical characteristics are shown in Table I. Before the experiment, experimental procedures were explained to all subjects, and written consents were received.

C. Experiments

The subjects performed stretching and warm-up exercises before the experiment, and the experiment was conducted after their heart rates were stabilized. In this experiment, the subjects pedaled for three minutes at their own saddle heights with a constant cadence of 90 RPM and power of 150 W. The pedaling experiment was conducted by a riding machine [2] with the same saddle heights of the cycle that the subjects commonly use. For a constant cadence and power in pedaling, the simulation programs I Magic Trainer and SRM Powermeter (Schoberer Rad Messtechnik, Germany) were used.

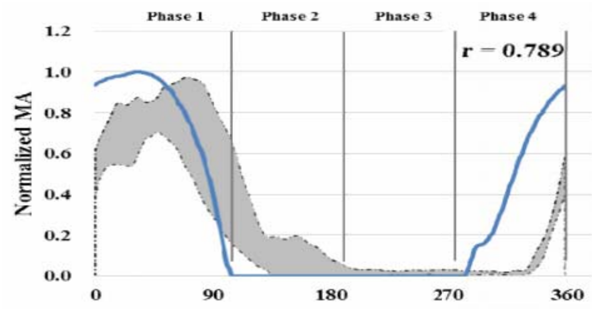
For the determination of the pedaling phase, a three-dimensional motion analysis system (Motion Analysis System Corp., USA) with six infra-red cameras was used. For crank rotation angles, the markers were attached to the crank axes and pedals, and the motion data were obtained with the sampling frequency of 120 Hz.

For the measurement of muscle activities, a wireless EMG measurement system [Trigno wireless EMG system with an operation range of 40 m, a transmission frequency of 2.4 GHz, and a CMRR (Common Mode Rejection Ratio) > 80 dB; Delsys Inc., USA] was used with the sampling frequency of 1,200 Hz. The muscle activities were measured at the mono-articular muscles [Vastus Lateralis (VL) and Tibialis Anterior (TA)] and the multi-articular muscles [Biceps Femoris (BF) and Gastrocnemius Medial (GM)], which are mostly used in pedaling (Fig. 2).

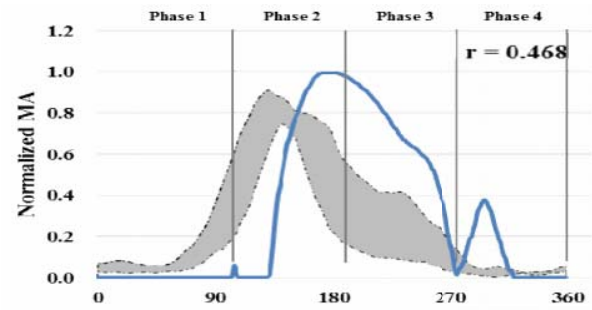
Fig. 2 Used muscles for this study

A. Data Analysis

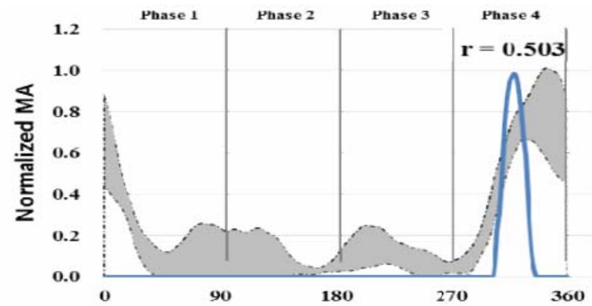
Total of two minute's data by excluding initial and final 30 seconds data were used for the analysis. A second-order zero-delay Butterworth filter with 10Hz cutoff frequency was used for low-pass filtering for noise removal of motion data [13]. The pedaling phases were divided into 4 phases such as Phase 1 (0-90°), Phase 2 (90-180°), Phase 3 (180-270°), and Phase 4 (270-360°) (Figs. 3 and 4).



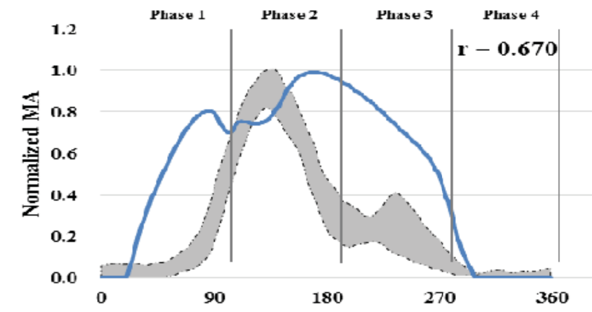
(a) Vastus Lateralis



(b) Tibialis Anterior



(c) Biceps Femoris



(d) Gastrocnemius Medial

Fig. 3 Muscle activities during a pedaling cycle (One cycle started from 0° to 360°)

$$\text{Normalized } MA = MA / \max MA \quad (1)$$

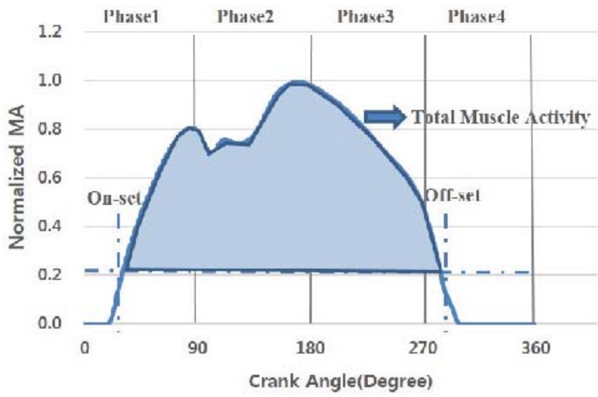


Fig. 4 Definition of total muscle activity

The EMG data were obtained through band-pass filtering with 15-500 Hz by using a fourth-order zero-delay Butterworth filter [13]. The muscle activity (MA) patterns were investigated through the correlation between measured EMG data and musculoskeletal model results. The peak timing for the maximum muscle activities was selected when the maximum muscle activity value appeared. For comparison between musculoskeletal model results and measured EMG data, both data were expressed was the normalized muscle activities (Normalized MA in (1)) [14]. Total muscle activity was defined as the summation of all muscle activities by excluding below 20% muscle activity values as shown in Fig. 4. All calculation was carried out with Matlab R2013a (Mathworks Inc., USA) software. Pearson's correlation coefficient (α = 0.05) was calculated by SPSS 19.0 (SPSS Inc., USA).

III. Results

A. Correlation

The muscle activity patterns that appeared in measured EMG and calculated musculoskeletal model result showed significant correlations (VL: 0.789, TA: 0.503, BF: 0.468, GM: 0.670) in all four muscles (Fig. 3 and Table II).

B. Peak Timing

The peak timing of the maximum muscle activities in measured EMG and calculated musculoskeletal model result showed differences of 20.1° and 26.3° in the VL and TA; 21.9° in the BF; and of 29.4° in the GM. Thus, the peak timing of all four muscles measured in this experiment showed about 20° differences. However, for phases, the peak timing of the maximum muscle activities of each muscle was distributed at the same phase (VL: Phase 1; TA: Phase 4; BF: Phase 2; and GM: Phase 2) (Table II and Fig. 5).

C. Total Muscle Activity

The total muscle activities during a pedaling cycle were VL (EMG: 22.4%, MODEL: 32.4%), TA (EMG: 14.7%, MODEL: 5.0%), BF (EMG: 20.6%, MODEL: 30.6%), and GM (EMG: 22.4%, MODEL: 51.8%) as shown in Table II. Among four muscles, VL, TA, and BF showed within 10% differences of the total muscle activities of measured EMG (10.0%, 9,7%, and 10.0%, respectively) excluding GM with a 29.4% difference.

TABLE II
CORRELATION COEFFICIENT, PEAK TIMING, AND TOTAL MUSCLE ACTIVITIES OF EACH MUSCLE (*P < 0.05) BETWEEN EMG AND THE MUSCULOSKELETAL MODEL RESULT

	VL		TA		BF		GM	
	EMG	Model	EMG	Model	EMG	Model	EMG	Model
Correlation Coefficient	0.789*		0.503*		0.468*		0.670*	
Peak Timing (°)	54° (±23.7)	34° (±1.7)	343° (±14.1)	317° (±0.8)	149° (±21.3)	171° (±6.1)	135° (±32.3)	164° (±2.6)
Phase	1	1	4	4	2	2	2	2
Total Muscle Activity	22.4% (±6.1)	32.4% (±0.3)	14.7% (±3.4)	5.0% (±0.2)	20.6% (±12.3)	30.6% (±.1)	22.4% (±3.5)	51.8% (±1.1)

Total muscle activity: Normalized MA for one cycle (mean±SD)

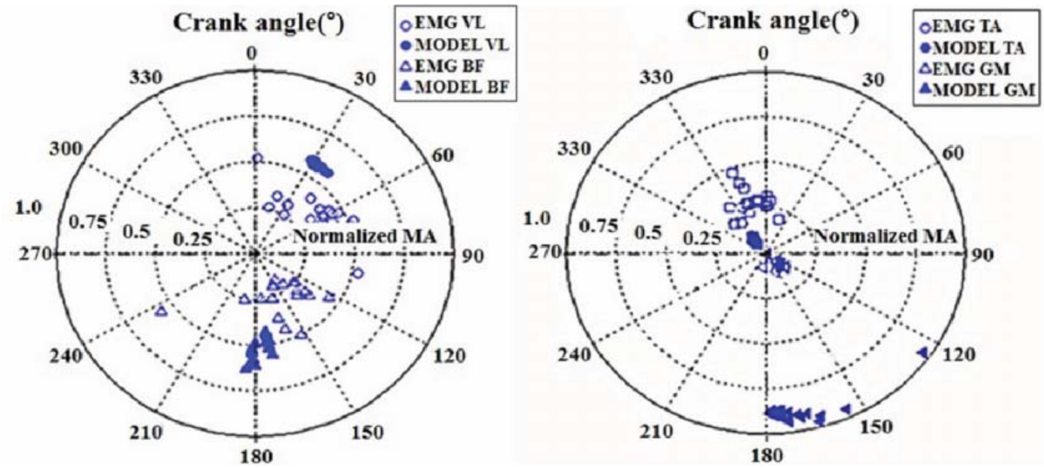


Fig. 5 Distributed peak timing and normalized MA of each muscle

IV. DISCUSSION

The forward dynamics of a musculoskeletal model was applied to this preliminary cycle fitting study. The measured EMG data were compared with the muscle activities of the musculoskeletal model through forward dynamics. The correlation coefficients of the muscle activity patterns, the peak timing of the maximum muscle activities, and the total muscle activities were calculated and compared.

There were some studies on musculoskeletal models of cycling to which the forward dynamics was applied. Neptune [15] examined the difference between the lower limb muscle activities and the joint powers that occurred due to the pedaling direction changes. Jeffery [9] studied the crank power and the muscle activities when the crank changed to the oval type. These studies investigated the results by the differences in the shapes of graph rather than by statistical and numerical comparisons between measured results with calculated musculoskeletal model results [10]. However, this study showed a statistical correlation between measured EMG and calculated musculoskeletal model results, and also showed the difference between measured EMG and calculated musculoskeletal model results with numerical values.

The correlation coefficients of the muscle activities showed positive correlations (VL: 0.789, TA: 0.503, BF: 0.468, GM: 0.670) in all four muscles. Therefore, it is considered that musculoskeletal models can be used to estimate muscle activity patterns in actual cycle pedaling. The peak timing of the maximum muscle activities in both measured EMG and musculoskeletal model result were distributed across particular phase for all muscles (VL: Phase 1, TA: Phase 4, BF: Phase 2, GM: Phase 2). The results indicate that the peak timing of the maximum muscle activities due to the posture changes can be evaluated using the forward dynamics of musculoskeletal models. In the meantime, the total muscle activities between measured EMG and musculoskeletal model result presented less than 10% differences in three muscles, VL, TA, and BF, excluding GM (29.4%). The results of musculoskeletal models by forward dynamics were statistically similar to the measured muscle activities. In addition, similarities were observed in the phases for the maximum muscle activities and the relative ratios of the total muscle activities. The merit of the forward dynamics is that it can minimize the number of the actual experiments and thus, save time and costs. Therefore, the forward dynamics is considered an effective method of cycle fitting that particularly requires the adjustment of diverse variables. However, the difference in the GM muscle activities shall be further considered. For the usage of musculoskeletal models with forward dynamics in cycle fitting, more precise human body must be considered. In the cases of multi-articular muscles such as BF and GM, these are involved in the precise control of the movement more than a joint. However, as the musculoskeletal models consist of straight-lined muscles, the muscle bending must be considered in the joint movements [16]. Such factors are considered to have comprehensively acted and caused the differences in the peak timing of the maximum muscle activities and the integrated muscle activities [17], [18].

If the forward dynamics of musculoskeletal models is used, the information on the EMG changes depending on the cycle fitting can be more simply collected than the actual measurement. Rankin [19] used the forward dynamics to show insignificant effects of the seat tube angle on the crank power, and Neptune [15] used the forward dynamics to explain the difference between the lower limb muscle activities and the joint powers generated depending on the pedaling directions. However, more studies have to be conducted that use musculoskeletal models for cycle fitting.

Therefore, studies that used the forward dynamics of musculoskeletal models are likely to be useful in evaluating the correlation of muscular patterns, the peak timing of the maximum muscle activities, and the total muscle activities. In addition, it is considered that more studies have to be conducted on the additional variables involved in cycle fitting, the pedal power directly related to pedaling, and the effect of cadence change on muscle activities.

V. Conclusion

In this study, the measured EMG data were compared with the calculated muscle activities of the musculoskeletal model through forward dynamics. The correlation coefficients of the muscle activity patterns, the peak timing of the maximum muscle activities, and the total muscle activities were calculated and compared. The muscle activity patterns showed significant correlation for all the data shown; the peak timing of the maximum muscle activities showed that each muscle changed in certain phases, and the total muscle activities showed similar values in the VL, TA, and BF except the GM. Thus, it can be concluded that muscle activities of model & experiment showed similar results. The results of this study indicated that it was possible to apply the simulation of further improved musculoskeletal model to cycle fitting.

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