Dynamic Elbow Flexion Force Estimation Through a Muscle Twitch Model and sEMG in a Fatigue Condition

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Abstract—We propose a joint force estimation method to compute elbow flexion force using surface electromyogram (sEMG) considering time-varying effects in a fatigue condition. Muscle fatigue is a major cause inducing sEMG changes with respect to time over long periods and repetitive contractions. The proposed method composed the muscle-twitch model representing the force generated by a single spike and the spikes extracted from sEMG. In this study, isometric contractions at six different joint angles (ten subjects) and dynamic contractions with constant velocity (six subjects) were performed under non-fatigue and fatigue conditions. Performance of the proposed method was evaluated and compared with that of previous methods using mean absolute value (MAV). The proposed method achieved average 6.7±2.8 %RMSE for isometric contraction and 15.6±24.7 %RMSE for isokinetic contraction under fatigue condition with more accurate results than the previous methods.

Index Terms—Muscle fatigue, force estimation, surface electromyography, elbow flexion, dynamic contraction

I. INTRODUCTION

In many applications, surface electromyogram (sEMG) is used to noninvasively measure muscle activation [1]. This noninvasive technique provides simple and reliable estimation of human body motion for healthy subjects as well as disabled subjects who can contract their muscles in prosthetics control, exoskeleton control, and biomechanical analysis [2]–[6]. In general, to estimate muscle force, surface electrodes are attached to the skin over the target muscles that are related to a target movement. An EMG-force relationship is used as a linear correlation between the recorded sEMG and the measured force in isometric condition [7]–[11]. In real-world applications, muscle fatigue, changes in joint angle, angular velocity, muscle moment arms, and the location of electrodes corresponding to a muscle should be considered to compensate for the dynamic effects (angle-varying or force-varying contractions) [12]–[15].

An important issue is the effect of muscle fatigue, which has not been significantly considered in previous force estimation studies. Muscle fatigue was defined as the point of failure to maintain a required or expected force [16]. The force generation mechanism changes when muscle fatigue develops during a fatiguing contraction [17]. Although myoelectric manifestations of fatigue have been used to track the progression of muscle fatigue during contraction, previous studies have not taken into account compensation to improve force estimation accuracy under fatigue condition. They have focused on developing a measure for detecting muscle fatigue in static and periodic contractions [18]–[25].

The magnitude of generated muscle force is determined by electrical behavior, mechanical muscle behavior, and muscle fatigue [26]. Electrical behavior refers to the properties of the firing rates of action potentials and the number of recruited motor units (MU), each of which is composed of a motor neuron and the muscle fibers it innervates [1]. Mechanical muscle behavior indicates the twitch force, which is a generated force in an MU by a single action potential [27]. Muscle fatigue influences electrical behavior and mechanical behavior. The magnitude of the twitch force is decreased even though the same action potential is transmitted to an MU as muscle fatigue progresses. In static force contraction, more MUs that are not used in non-fatigue condition are recruited, and this phenomenon is captured by changes in sEMG amplitude. Therefore, the effects of muscle fatigue should be considered for long-term contractions or strong contractions that cause muscle fatigue.

In this paper, we propose a force estimation method reflecting the force generation mechanism using the muscle-twitch model and the recorded sEMG in a fatiguing elbow contraction. In our previous work, we evaluated the proposed method capable of estimating isometric finger contraction force under non-fatigue condition [28] and using as an indicator for muscle fatigue [29]. Here, we extend our method for elbow flexion conditions; 1) multi-muscle-related contraction, 2) angle-varying contractions, and 3) fatiguing contraction. Changes in joint angle and joint velocity are compensated using the angle- and the velocity-based coefficients. The proposed method was evaluated for isometric...
contractions at six different joint angles and isokinetic contractions with two joint velocities under fatigue condition.

II. METHODS

A. Subjects

We recruited ten healthy volunteers (right-handed, 26.4±1.8 years, males) who had no previous experience with our experiments. All subjects participated in isometric contraction tests, and six subjects among the ten subjects participated in isokinetic contraction tests. Subjects in a relatively narrow age range were recruited to minimize the potential confounding effects of age. All participants had no history related to upper extremity or other musculoskeletal complaints and confirmed their informed written consent prior to participation. The protocol (KH2010-25) was approved by the Institutional Review Board of Korea Advanced Institute of Science and Technology (KAIST).

B. Data collection

The experimental setup is illustrated in Fig. 1. A dynamometer (Con-Trex MJ dynamometer, CMV AG, Switzerland), which provides analog signals for joint torque, angle, and direction, was used for isometric and dynamic elbow contractions. The participants lay on the dynamometer with their forearm secured to it using straps. The weight of the forearm was supported by the bracket for the forearm. The sEMG results for the biceps (BB) and brachioradialis (BR) for flexion at elbow joint were recorded using wireless EMG sensors (Trigno, Delsys, USA). The signal from the brachialis was not used in this study because the brachialis, which lies under the BB, cannot be measured using surface electrodes on the skin. The positions of the electrodes were chosen to be on the bellies of the BB and BR when the subjects performed isometric flexion of the elbow joint.

All signals in this study were measured through a real-time data acquisition board (Quarc Q8-USB, Quanser, Canada) at 1 kHz sampling rate. Analog data from the dynamometer was directly connected and data from the wireless EMG sensors was extracted from the EMG systems and then connected to the Q8-USB. Three signals from the dynamometer were low-pass filtered using a finite impulse response (FIR) filter with a corner frequency of 20 Hz. The signals from the EMG sensors were band-pass filtered using an FIR filter with a frequency range between 20 and 400 Hz.

C. Experimental procedure

The subjects were asked to lie comfortably on a testbed and to relax their upper limbs. The elbow joint was synchronized with the rotation joint of the dynamometer for elbow flexion from 20° to 95° as shown in Fig. 2. Two straps were fixed to the hand and forearm. The experimental protocol is illustrated in Fig. 3. There were three sessions: 1) a non-fatigue session, 2) a fatiguing contraction was performed until exhaustion. Third, the same contractions that were performed in the non-fatigue session were repeated in 3) the fatigue session.
fatigue-inducing session, and 3) a fatigue session. Visual feedback was provided to a subject for target force, measured force, and current joint angle using a monitor.

First, in the non-fatigue session, subjects were instructed to perform two times the maximum voluntary contraction (MVC), six times the isometric set, and ten times the isokinetic set. The isometric set was composed of five contractions for a half sine wave with 0.1 Hz and 20% MVC and was repeated at six joint angles as shown in Fig. 2. The isokinetic set was composed of five contractions at five predefined angles that were used in isometric contractions from 20° to 80° with a constant velocity (5 deg/s and 10 deg/s) and was repeated five times for two angular velocities, respectively. Isokinetic contraction indicates that the dynamometer was moved at a constant angular velocity without being affected by human movement. The subjects were asked to perform the same contraction level (20% MVC) that was used in the isometric set for as long as possible. After each set was finished in the non-fatigue session, three minutes’ rest time was provided to the subjects to avoid the muscle fatigue. In addition, 20% MVC was selected to reduce the development of muscle fatigue during repetitive contractions.

Second, in the fatigue-inducing session, subjects performed 80% MVC isometric contraction until exhaustion at the fixed joint (80°). When 80% MVC was not maintained over three seconds, the fatigue-inducing session was ended by an experiment instructor. Third, after the fatiguing session, subjects repeated the isometric and isokinetic contractions that were performed in the non-fatigue session. There was no rest time to prevent recovery from muscle fatigue. It was assumed that the same level of muscle fatigue was maintained in the fatigue session.

Figure 4 shows the datasets for the isometric contraction and isokinetic contractions. The sEMG signals recorded from the BB and BR are represented in first and second row plots. While the joint angle was maintained at 35°, a single contraction was performed at about 20% MVC. In isokinetic contractions, each subject performed five contractions at about 20% MVC, while the joint angle was changed with a constant angular velocity (5 deg/s). Figure 5 shows the data measured from the fatiguing session. The measured force by a subject was maintained for about 60 s with 80% MVC in the top plot. The manifestations of muscle fatigue were investigated based on sEMG feature changes during contraction (mean absolute value (MAV) and mean frequency (MNF)) and used to confirm that muscle fatigue had progressed. Increase in the MAV slope and decrease in the MNF slope observed during fatiguing, continuous, and isometric muscle actions due to muscle fatigue [20].

III. EMG-FORCE ESTIMATION

A. EMG-force mapping using a twitch model

The physiological muscle force generation model was reported by Contessa and de Luca [30] as shown Fig. 6(a). This mechanism for single muscle is described by
Physiological force generation

Spike trains in MUs | Twitch force | MU force
--- | --- | ---
MU1 | | |
MU2 | | |
MU3 | | |

(a)

Single joint with single muscle

Spike trains from sEMG | Twitch model | Muscle forces
--- | --- | ---
Total force | |

(b)

Elbow joint with multi-muscles

Spike trains from each muscle | Twitch model | Muscle forces
--- | --- | ---
BB | | |
BR | | |
Total force | |

(c)

Fig. 6. Force estimation method using spike trains extracted from sEMG and a twitch model during voluntary contraction. (a) Physiological muscle force generation model reported by Contessa and de Luca[30]. (b) Our previous method that modified from (a) to estimate muscle force using sEMG in a single joint with a single muscle. (c) The method proposed in this study for elbow joint with multiple muscles. The same mechanism that was used in (b) was applied for the BB and the BR.

\[
F_{tot}(t) = \sum_{i=1}^{N} F_i(t) \quad (1)
\]

and

\[
F_i(t) = \sum X_i(t) \cdot f_i(t), \quad (2)
\]

where \(F_{tot}(t), F_i, X_i, f_i, N\), and \(*\) indicate the total muscle force summing the generated forces by MUs, the force generated by the \(i\)th MU, the spikes of the \(i\)th MU, the twitch force of the \(i\)th MU, the number of MUs, and the convolution process, respectively.

In our previous work, we implemented an muscle force estimation method inspired by Fig. 6(a) using sEMG and a twitch model when index finger abduction was performed in isometric contraction [28]. Figure 6(b) represents our method for a single joint with a single muscle. To apply the force generation model to voluntary contractions, the spikes, \(X_i(t)\), transmitted to MUs were replaced with extracted spikes, \(x(t)\), from sEMG. The twitch force, \(f_i(t)\), was modified to the twitch model, \(T(t)\), which describes the properties of the twitch force. The proposed method is described by

\[
F_{tot}(t) = \sum x(t) \cdot T(t), \quad (3)
\]

where \(F_{tot}(t), x(t),\) and \(T(t)\) are the joint force, the spike trains extracted from sEMG, and the twitch model, respectively. Here, two important assumptions were used in our method. First, it was assumed that the magnitude and rates of spikes from sEMG could be used as indicators to estimate how many and how fast MUs are recruited because sEMG is sum of all motor neuron spikes represented by a series of motor unit action potentials (MUAPs). Second, the identical twitch model was applied corresponding to each spike.

Figure 6(c) shows the scalability of the proposed method for the elbow joint. Our method was expanded for the elbow joint with multiple muscles that have different distributions of MUs. The identical principle that was used in a joint with a single muscle was applied to each muscle with respect to the elbow joint. The twitch model was characterized for the BB and BR using the recorded sEMG signals. The number of twitch models was determined depending on the number of sEMG electrodes. For elbow joint force estimation in this study, (3) was expanded as follows:

\[
F_{tot}(t) = \sum x_{BB}(t) \cdot T_{BB}(t) + \sum x_{BR}(t) \cdot T_{BR}(t), \quad (4)
\]

where \(BB\) and \(BR\) indicate the biceps and the brachioradialis, respectively.

B. Spike extraction in sEMG

The greater the number of MUs recruited in generating the force, the larger the spike of the summed MUAPs appears. When two MUAPs occur within close proximity, they produce a signal with a greater spike; this spike resembles a larger MUAP. Accordingly, the number of recruited MUs was approximated as the sEMG magnitude at a spike of summed MUAPs. In addition, the moment at which a spike occurred was used to approximate the time of MUAP occurrence.

We defined a threshold value so that small peaks were not identified as input to avoid misidentification, because sEMG signals include small amplitudes under inactive conditions. The threshold was determined as three times the standard deviation of the recorded signals (noise) when the volitional force was zero and the muscle was inactive [31]. The spikes, \(x(t)\), in (3) were found using the following equation, which was used to extract the morphological features of the spikes:

\[
x_i = \begin{cases} 
  y_{i-1}, & \text{if } (y_i > a) \text{ and } (y_{i+1} - y_{i-2}) \cdot (y_i - y_{i-1}) < 0, \\
  0, & \text{otherwise}
\end{cases}
\]
where \( y_i, x_i, \alpha \) are the normalized sEMG signal at time \( i \), the extracted spikes, and the threshold value to avoid noise, respectively. The magnitude of the spikes reflects the percentages of available MUs that were recruited for force generation because the sEMG was normalized to the maximum value for each subject.

C. Twitch force model characterization

A typical characteristic of the twitch force was investigated using electrical stimulation of several muscles. Although force is a nonlinear response of the neuromuscular system, twitch force has been regarded as an impulse response of the linear system [32]–[35]. In (5), the twitch model, \( T(t) \), was calculated by

\[
T(t) = \frac{P \cdot t \cdot e^{(1/t_{CT})}}{t_{CT}},
\]

where parameters \( P, t_{CT} \), and \( t \) indicate the twitch peak, the contraction time to peak, and the time, respectively [34]. We calculated the twitch model using the relationship between the extracted peaks and the measured force [28]. The parameter changes in \( P \) and \( t_{CT} \) indicate how much muscle fatigue developed because the properties of the twitch model changed as a function of muscle fatigue [36].

D. Effects of joint angle and joint angular velocity

Muscle properties changed in relation to changes in joint angle because muscle length and muscle moment arm varied. As elbow joint angle decreases from fully extension angle, flexor torque increases with a maximum value near 90° and then decreases [37]. In addition, the location of the sEMG sensor was moved with respect to the underlying muscle compared with location under isometric condition.

To compensate the effect of angle-varying contraction on sEMG and muscle contraction properties, the angle-based coefficients for the BB and the BR were used for the non-fatigue session and the fatigue session, respectively. The coefficients were average values of the peak parameters of the twitch models at each angle. The calibrated coefficients for the peaks of the twitch models were used as follows:

\[
F_{tot}(t) = \sum_{i=1}^{6} x_{BB}(t) \cdot P_{BB}(\theta_i) \cdot t \cdot e^{(1/t_{BR,CT})} \cdot t_{BR,CT} + \sum_{i=1}^{6} x_{BR}(t) \cdot P_{BR}(\theta_i) \cdot t \cdot e^{(1/t_{BR,CT})} \cdot t_{BR,CT},
\]

where \( P_{BB}(\theta_i) \) and \( P_{BR}(\theta_i) \) are the angle-based coefficients for the BB and the BR at joint angle \( \theta_i, i = 1, ..., 6 \), respectively. The contraction time \( t_{CT} \) was neglected because the range of changes was narrower than that of the twitch peaks. The calibrated angle-based coefficients were used to estimate the force in isometric contractions. For isokinetic contractions, the slope of the angle-based coefficients was used because degree of angle was varied with constant velocity. All data from isometric contractions was included for calculating the slope. The average coefficients and the slope of the angle-based coefficients were represented in Fig. 7.

In addition, the angular velocity influences on muscle properties in relation to contraction types. As the contractile velocity increases, the magnitude of muscle force exponentially decreases [38]. On the other hand, the force increases as the negative velocity increases (lengthen direction). In this experiment, we simplified the contraction condition for concentric contractions in the entire experimental procedure. The slope of the angle-based coefficient can compensate the changes in angle. A constant velocity-based coefficient was added in (7) on the isokinetic contraction to compensate the effects of velocity on the isokinetic contraction. A linear regression was used to optimize the velocity-based coefficient using the estimated force data from (7) (without the velocity-based coefficient) and the measured force data.

E. Performance evaluation

Two parameters were calculated, namely, the coefficient of determination \( (R^2) \) [39] and the relative root mean-squared error \( (% RMSE) \) [12] as follows:

\[
R^2 = 1 - \frac{\sum_{i=1}^{N} (\hat{f}(i) - f(i))^2}{\sum_{i=1}^{N} (f(i) - \bar{f}(i))^2},
\]

and

\[
% RMSE = \frac{\sum_{i=1}^{N} (\hat{f}_i - f_i)^2}{\sum_{i=1}^{N} f_i^2} \times 100,
\]

where \( f_i, \hat{f}_i \), and \( N \) are the measured force, the estimated force, and the number of samples, respectively.
sEMG and force and accurate knowledge of the physiological human system. The regression method optimizes a quadratic equation and the ANN acts as a black box model to approximate complex nonlinear mappings directly from MAV to the finger force.

Configuration of training and testing for both methods was different compared with the proposed method to prevent poor performance on tasks outside those defined by the training sets. Four trial data was used to train a model and single trial data not included in training data was used to test the force estimation at each fixed angle in isometric contraction. In isokinetic contraction, a single set composed of five contractions was used for training data and four other sets were tested at each velocity condition. Same number of data was obtained for isometric contractions compared with the proposed method. In isokinetic contractions, other methods used greater number of data than the proposed method for each velocity condition.

The mathematical definition of MAV is

\[ y_{MAV,i} = \frac{1}{N} \sum_{k=1}^{N} |y_i(k)|, \]

where \( y_{MAV,i} \), \( y_i(k) \), \( i \), and \( N \) denote the MAV values of the sEMG signal, the \( k \)th signal sample, the \( i \)th segment, and the number of samples in a segment (window length), respectively [40]. We optimized the best window length (\( N \)) for each subject because there was a trade-off between the signal-to-noise ratio and responsiveness (rapid detection of onset or offset of muscle activation) [1]. When we used large time window for MAV, it reduced not only variability in sEMG but also rapid change, which could be intentional muscle activation. When we used a short window to reduce the delay effect, it increased the signal variability in sEMG, by contrast. The window size was varied from 150 ms to 250 ms with 1 ms increment and selected when it showed the best force estimation performance based on the \( R^2 \) value.

IV. RESULTS
A. Isometric contractions
The average \( R^2 \) and \( \%\text{RMSE} \) at the six joint angles under non-fatigue and fatigue conditions was shown in Fig. 8. For the proposed method, there were no statically significant differences at each angle between non-fatigue and fatigue (\( p > 0.05 \)). The data of one subject was an outlier data which affected the force estimation performance (large standard deviation) at 35°. The average \( R^2 \) achieved was over 0.8, and the average \( \%\text{RMSE} \) obtained was less than 15% for all joint angles. Statistical analysis on this study was carried out using the two-sample t-test (\( p < 0.05 \)).

Figure 9 shows the boxplots of the \( R^2 \) and \( \%\text{RMSE} \) for comparison of the proposed, the regression, and the ANN methods.
method in isometric contractions. The proposed method showed superior results to the regression method and comparable results to the ANN methods. When compared to the regression, the proposed method obtained significant improvement for all conditions. There were no significant differences between the proposed method and the ANN for all conditions except %RMSE in fatigue session (p < 0.05).

B. Isokinetic contractions

Figure 10 shows the boxplots of $R^2$ and %RMSE for the proposed method in isokinetic contractions. The variables $w1$, $w2$, NF, and F denote angular velocity 5 deg/s, 10 deg/s, non-fatigue, and fatigue condition, respectively. As velocity increased, there were no significant changes in $R^2$ and %RMSE ($p > 0.05$) regardless of muscle fatigue. However, there were significant decreases because of muscle fatigue development ($p < 0.05$) for all conditions except %RMSE for $w2$ ($p = 0.59$).

Figure 11 shows the boxplots of $R^2$ and %RMSE for the proposed, the regression, and the ANN method in isokinetic contractions with $w1$ velocity. The most noticeable change was the reduction in performance for all methods because of muscle fatigue. Variation of data increased in fatigue session based on the magnitude of upper quartile and lower quartile. Compared with other methods, there were significant differences between the proposed and the ANN method ($p < 0.01$) and no differences between the proposed and the regression method ($p > 0.05$) for all conditions. Unlike the proposed and the ANN, the regression showed outliers as shown '+' marker in Fig. 9.

V. DISCUSSION

This study evaluated the force estimation method for elbow flexion movement during isometric and dynamic contractions under non-fatigue and fatigue conditions. The muscle-twitch model was characterized at six fixed joint angles, and angle-based coefficients were used to compensate the effects of angle changes. The velocity-based coefficient determined by the values of angular velocity were imposed in dynamic contraction. Based on its force estimation performance during elbow flexion, the proposed method was demonstrated to be suitable for other joints with multiple muscles in isometric/isokinetic contractions and under muscle fatigue condition.

Most previous studies have estimated muscle force with the assumption that sEMG is used under stationary condition [7], [8], [10], [14], [6]. This assumption is only suitable for limited experimental conditions, such as static contraction (constant posture or cyclic-varying contraction) because the sEMG-force relationship is affected by changes in muscle fatigue, joint angle, and contractile velocity [12]. Although the manifestation of muscle fatigue has been reported using features of sEMG in many studies, few studies have been implemented to compensate the effects of muscle fatigue in force estimation [15], [41]. In this study, we used the muscle-twitch model, capable of indicating an index for muscle fatigue, for force estimation with spikes extracted from sEMG.

Non-fatigue condition and fatigue condition were considered in our experimental setup. The non-fatigue condition was maintained during the time interval from the start of the experiment to the time prior to the fatigue-inducing session. The fatigue condition was covered by the time interval from the time of exhaustion because of isometric 80% MVC contraction to the end of the experiment. We tried to quantitatively evaluate the proposed method under the stringent conditions. Therefore, only two sessions such as non-fatigue and fatigue (exhaustion) session were included in this study. The experimental protocol and upper-limb movements for reflecting real-world scenarios is remaining issues for future work.

A downward trend of MNF was used to verify whether muscle fatigue had developed because contraction time in a fatiguing contraction was determined by participant motivation. Changes in MNF have been considered as the gold standard for assessing muscle fatigue under static condition [42]. All subjects showed minus slopes of MNF during the fatigue-inducing session, and a sample data set is presented in Fig. 5. The signals from BB and BR showed the same trend of MNF.

Regarding isometric contraction, as shown in Fig. 8, the proposed method showed consistent results despite of the effects of joint angles and muscle fatigue, but the regression method decreased under fatigue condition compared with non-fatigue condition. The reason for this difference is the biomimetic structure of the proposed method, including
electrical behavior (spikes from sEMG) and the mechanical muscle behavior (twitch model) [26]. However, the MAV process including a rectification process altered the frequency component of sEMG without comprehensive physiological justification [43]. These results are supported by previous work which also concluded that the proposed method showed better performance than the MAV method in isometric contraction and under non-fatigue condition [28].

The angle-based calibration was used to compensate the angle-dependent factors on the twitch model. Because the properties of the twitch model were determined by spikes extracted from sEMG and the measured force, variations of the sEMG-force relationship were critical factors to produce changes in the twitch model. The effects of joint angle changes on the sEMG-force relationship have been reported; the MVC force and MVC EMG were changed, and the amount of influence differed according to the experiment setup [14], [44], [45]. The force-angle relationship is very specific to each muscle. In this study, changes in the twitch model were used to compensate the effects of joint angle instead of MVC measurement, because MVC is affected by uncontrolled factors, such as subject motivation, posture, and experimental setup [35].

For isokinetic contractions, the velocity-based coefficients and the slope of the angle-based coefficients were used to estimate joint force. Although these coefficients had different values across subjects, same trend of changes were obtained. For the angle-based coefficients, the absolute value of slope of the angle-based coefficients decreased in fatigue sessions rather than in non-fatigue sessions. Reduction of the absolute slope value across angle changes indicated the decrease of twitch force in fatigue condition. The velocity-based coefficient decreased as the magnitude of velocity increased regardless of muscle fatigue for all subjects. This trend indicated that the role of velocity-based coefficient was compensation of the overestimated force because the magnitude of muscle force physiologically decreases as the contractile velocity increases. The effects of changes of angle and velocity should be compensated in isokinetic contractions.

In force-velocity relationship of muscle, the concept of power, can be expressed as a product of force and velocity, is constant [37]. Force increases while velocity proportionally decreases, or vice versa. Although this effect was considered using the constant gain for each velocity in this study, the effects of velocity were not fully compensated. Regarding isokinetic contractions, as shown in Fig. 10, the performance decreased compared with that for isometric contractions. However, there were not significant differences because of velocity changes in isokinetic contractions.

In a related previous study [12], the experimental setup was slightly different, but force estimation performance was similar under non-fatigue condition. Hashemi et al. reduced the effects of angle changes using angle-based calibration, which was used in this study during angle-varying elbow contractions, and achieved minimum 33.3 %RMSE for fully dynamic contractions. However, muscle fatigue was neglected in their work. To the best of our knowledge, this study was the first attempt to estimate dynamic force using sEMG under fatigue condition. The performance of the proposed method was clearly superior to that of the MAV-based regression method.

Isokinetic contraction was performed using the dynamometer in this study [4]. The dynamometer was moved with a constant velocity in a continuous passive mode. Subjects generated the elbow flexion force starting at predefined angles that were used in isometric contractions, while their elbow was moved with the dynamometer at a constant velocity. Although limited elbow movements were conducted compared with general elbow movement in everyday life, the improvement of force estimation performance using the proposed method could be helpful for researchers who aim to develop a force estimation method that can be used in real life.

VI. CONCLUSION

In this study, force estimation in isometric and isokinetic elbow concentric contractions under fatigue condition using sEMG was evaluated with the muscle-twitch model and spikes extracted from sEMG signals. A major contribution of this work is that elbow flexion force estimation was performed with high accuracy for isometric and isokinetic contractions. The performance of the proposed method was compared with that of conventional methods (regression and ANN) under isometric static contractions at six joint angles and isokinetic contractions. The proposed method showed consistent results under fatigue condition and superior performance than the regression method and comparable results with the ANN, respectively. Our results will allow for comparison with future force estimation studies and help for sEMG-based applications: myoelectric prosthesis control, exoskeleton control, and rehabilitation robot control.

REFERENCES