BIOMECHANICAL ANALYSIS OF LOWER EXTREMITIES DURING CONVENTIONAL AND FUNCTIONAL ELECTRICAL STIMULATION (FES) ROWING IN NON-DISABLED INDIVIDUALS

Masaaki Takeshima1, Yoichi Shimada1, Toshiki Matsunaga2, Takehiro Iwami3 and Kazuhiko Hiramoto3

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1)Department of Orthopedic Surgery, Akita University Graduate School of Medicine, Akita 010-8543, Japan
2)Department of Rehabilitation Medicine, Akita University Hospital, Akita 010-8543, Japan
3)Department of Mechanical Engineering, Akita University Faculty of Engineering and Resource Science, Akita 010-8502, Japan

Abstract
Functional electrical stimulation (FES) rowing is a whole-body exercise in which the lower extremities are moved by electric stimulation and the upper extremities are moved voluntarily by paraplegics. The purpose of this study was to identify the kinematic factors of the lower extremities required to perform FES-rowing through the biomechanical analysis. Eighteen healthy adult men participated in this study. A mathematical model was developed to analyze the conventional rowing with or without handle and FES lower extremity extension exercise. During rowing with handle, maximum hip, knee extension and ankle plantar flexion moment was 23.5±4.5 Nm (Mean±SD), 2.5±3.1 Nm, and 11.9±2.5 Nm. During rowing without handle, maximum hip, knee extension and ankle plantar flexion moment was 8.0±3.9 Nm, 4.1±4.0 Nm, and 7.9±2.6 Nm. During FES lower-extremity extension exercise, maximum hip, knee extension and ankle plantar flexion moment was 4.7±2.7 Nm, 4.8±2.9 Nm, and 7.5±1.8 Nm. The calculated data has the potential to be applied in the low-load and safe FES rowing exercise for the paraplegics.

Key words: Functional electrical stimulation (FES), rowing, biomechanical analysis

Introduction
In patients with paraplegia caused by upper motor neuron damage, such as spinal cord injury or stroke, there is marked muscular atrophy as a result of severely compromised lower extremity function1. In the paraplegic patients, the onset of metabolic syndrome and increased cardiovascular diseases have been reported as the long-term health management problems caused by inadequate exercise due to a decreased amount of activity, and the importance of fitness exercise has been recognized2. Exercises that focus on the upper extremities, such as ergometer exercises3 and wheelchair sports4,5, and exercises that concentrate on the lower extremities, such as bicycle exercises6,7 using functional electrical stimulation (FES), are currently available to paraplegic patients.

FES rowing is a full-body exercise in which the lower extremities are moved by electric stimulation and the upper extremities are moved voluntarily by paraplegic patients. This exercise is recognized as a superior endurance training program for both the cardiovascular and musculoskeletal systems. Wheeler and colleagues re-
ported that FES rowing is safe in patients with spinal cord injury and is capable of reducing the risks associated with cardiovascular diseases\(^{10-15}\). The kinematics of FES rowing, however, has not been fully analyzed. Suzanne and colleagues instructed one paraplegic patient to perform FES rowing and five healthy patients to carry out conventional rowing and subsequently compared the changes in joint angles and indirect moments\(^{16}\). Another report about kinematics of FES rowing has not been reported in the literature to the best of our knowledge. There are, further, no basic data indicating the force required for paraplegic patients to perform FES rowing. For estimating safety and accelerating availability of FES rowing, it is necessary to kinematics of FES rowing.

The objective of the present study was to identify the joint moment and load of the lower extremities and the intensity of the stimulation required to perform FES-rowing for non-paraplegic volunteers through the kinematic study.

**Materials and Methods**

**Subjects**

Eighteen non-disabled adult males with a mean age of 24 years (range: 18-30 years), mean height of 171.9±4.5 cm (mean±SD) and mean weight of 67.7±8.4 kg were volunteered in this study. The subjects had no previous disease or injuries to their musculoskeletal systems. All subjects provided informed consent to participate in this study.

**Methods**

**The exercise device and Sitting conditions**

The Akita FES-rowing machine that we developed for the exercise was used in this study (Fig. 1)\(^{17,18}\). The machine measures 2,492×303 mm and weighed 80 kg. The machine was based on a rowing machine in which the air resistance on the fly-wheel fan. The subjects were instructed to sit in the seat and were positioned so that both feet were on the footrest and the trunk was belted to the backrest to prevent the subjects from leaning their body forward. The inclination angle of the seat rail of the rowing machine from the ground was set at 4°. All measurements were made under these sitting conditions.

**Conventional rowing exercise**

The subjects were instructed to rowing exercise without electrical stimulation of the lower extremities. Experimental conditions were as follows:

1. Conventional rowing with pulling handle
2. Conventional rowing without pulling handle, the subjects were instructed to cross their arms over their chests in exercises

A single rowing cycle consisted of knee flexion, extension and back to flexion, and each subject was instructed to three rowing trial at a pace of 23 cycles/min.

**FES lower extremity extension exercise**

The subjects were instructed to lower extremities exercise with electrical stimulation of the lower extremities. For electrical stimulation, subjects were instructed to fully relax their lower extremities. Surface electrodes (Sekisui Plastics Co. Ltd., Japan) were attached to the subjects’ bilateral quadriceps femoris. From the flexed knee position, electrical stimulation was used to extend the knee joints using an electrical stimulator (Dynamid, DM2500, MINATO MEDICAL SCIENCE Co. Ltd., Japan). The simulation was applied with a frequency of 40 Hz, power distribution of 4 s and rise time of 1.5 s. The intensity of stimulation was initially set at 20 mA and increased in 5-mA increments until the knees were completely extended. The intensity of the stimulation at maximum knee extension was subsequently recorded. The timing of the electrical stimulation was determined using the hand switch that each
subject was instructed to hold in their hands. In each subjects, three trials were collected.

**Motion analysis**

Using 6-axial tension transducer (IFS 105M50A220, Nitta Co. Ltd., Japan) attached to the footrest where both feet were fixed, measurements of reaction force were made at a sampling frequency of 60 Hz. Retroreflective markers were attached to the seventh cervical spinous process, the tenth thoracic spinous process, the sterno-clavicular joint space, the xiphoid process of the sternum, the greater trochanter, the posterior superior iliac spine, the lateral side of the thigh, the lateral side of the knee, the lateral side of the lower leg, the lateral malleolus of the ankle, the calcaneal region and the second metatarsal bone head. Kinematic data were collected with 7 cameras using the VICON 370 system (Vicon Motion Systems Ltd., Oxford, U.K.). From the experimental data, the moment of joints was calculated by a model calculation. During conventional rowing, electromyographic data were collected from 3 muscles (gastrocnemius, rectus femoris, and vastus medialis).

**Calculation of joint moment using a link model**

Modeling of the human body was performed and the joint moment was calculated using a link model. The dynamic model consisted of a two-dimensional 4-link model (trunk, thigh, shank and foot) (Fig. 2). For simplification, the following assumptions were made regarding the model:

1) Each link can be represented by a rigid-body link.
2) Each joint is a single-axis rotational joint.
3) The center of gravity of each link is positioned on the articulating axis.

Considering gravity and acceleration of each link, dynamic joint moment on the ankle (M_a), knee (M_k), and hip (M_h) were calculated from balancing horizontal and vertical forces and moments at each joint.

\[
\begin{align*}
\text{[Link : L1 (Foot)]} \\
\text{ }f_a &= -F + m_1 \vec{x}_1 \\
\text{ }n_a &= -N + m_1 (\vec{y}_1 + g) \\
\text{ }M_a &= -I_1 \dot{\theta}_1 + F y_1 - f_a (y_a - y_i) \\
& \quad + N (u - x_i) - n_a (x_i - x_a) \\
\text{[Link : L2 (shank)]} \\
\text{ }f_k &= -f_a + m_2 \vec{x}_2 \\
\text{ }n_k &= -n_a + m_2 (\vec{y}_2 + g) \\
\text{ }M_k &= -I_2 \dot{\theta}_2 + f_a (y_2 - y_a) - f_k (y_k - y_2) \\
& \quad - n_k (x_2 - x_d) + n_a (x_k - x_2) - M_k \\
\text{[Link : L3 (Thigh)]} \\
\text{ }f_h &= -f_k + m_3 \vec{x}_3 \\
\text{ }n_h &= -n_k + m_3 (\vec{y}_3 + g) \\
\text{ }M_h &= -I_3 \dot{\theta}_3 + f_k (y_3 - y_k) - f_h (y_h - y_3) \\
& \quad - n_h (x_3 - x_d) + n_k (x_h - x_3) - M_h
\end{align*}
\]

where \( F \) and \( N \) are horizontal and vertical components of floor reaction force, \( f \) is the joint force in horizontal plane, \( n \) is the joint force in vertical plane, \( u \) is the horizontal distance from the center of gravity of the link, \( y \) is the vertical distance from the center of gravity of the link, \( m \) is mass of the link, \( g \) is gravity, \( M \) is joint moment, \( I \) is inertia moment of the link around the center of gravity, \( n \) is horizontal displacement at the acting point of floor reaction force, \( \theta \) is angle of the link, 1, 2 and 3 indicate, respectively foot, shank, and thigh, and \( a, k \) and \( h \) indicate ankle joint, knee joint, and hip joint, respectively.

**Estimating leg muscle tension**

In the present study, a new method for estimating muscle tension is used that considers myoelectric data, muscle length, and muscle contraction velocity. The optimization problem for estimating the muscle tension can be formulated as a minimization problem of an evaluation function with constraints given by linear matrix inequalities (LMI). For the optimization problem the global optimal solution that minimizes the evaluation function can
be always obtained with the well-established convex optimization method. Muscle tension was obtained using the musculoskeletal model shown in Fig. 3. In creating this model, nine muscles that serve major roles in leg exercises were selected.

The muscle tension produced in each muscle and the maximum and minimum tension are defined as $F_i$, $F_{i}^{\text{max}}$, and $F_{i}^{\text{min}}$ ($i=1, \ldots, 9$). The length of the lever arm of the joint attached to each muscle is taken to be $L_{ij}^i$ ($i=1, \ldots, 9$, $j=a, k, h$). The lever arm length is the length from the center of rotation of the joint to the attachment of the muscle, obtained anatomically. The lever arm for joints to which a given muscle does not attach is taken to be 0.

Then the moment of each joint is given as the following:

$$M^j = \sum_{i=1}^{9} F_i s_{ij}^j L_{ij}^i, \quad j=a, k, h$$

(10)

where, $s_{ij}^j$ ($i=1, \ldots, 9$, $j=a, k, h$) is a constant that takes a value of either 1 or −1 depending on coincidence of the moment acting on the joint. The number of muscles for which muscle tension is estimated is nine, but the number of muscles that can be used in Eq. (10) alone is only three, and a greater number of unknowns remain. Left like this, muscle tension is a mathematically indeterminate problem that is underspecified.

Pedotti established the evaluation function $J$ as shown in Eq. (11), and showed that actual muscle tension can be obtained if this function is minimized. Most previous studies established an evaluation function to express efficient muscle activity during exercise in order to obtain combinations of individual muscle tension, which they did by optimizing the evaluation constant. In the present study, as well, we used the method of Beale, a nonlinear programming method shown in Eq. (11), as a means of optimizing the estimation of muscle tension using an evaluation function:

$$J := \sum_{i=1}^{9} \left( \frac{F_i}{F_{i}^{\text{max}}} \right)^2$$

(11)

The problem of estimating the muscle tension of the legs is formulated as follows.

**Muscle tension estimation problem**

For different time segments sampled between the commencement and completion of the rowing exercise, minimized muscle strain is obtained using Eq. (11) under equality constraint conditions for balancing the moment given in Eq. (11) and inequality constraint conditions for muscle tension.

The formulated estimation problem results in an optimization problem with inequality constraints given as linear matrix inequalities (LMIs). The detailed solution procedure is described as follows.

Expression (11) can be rewritten as the following:

$$J = G^T G$$

(12)

Where, $G$ is a 9-dimensional column vector defined as the following form:

$$G := \begin{bmatrix} F_1 & F_2 & \cdots & F_8 & F_9 \end{bmatrix}^T$$

(13)

Note that vector $G$ is the function on the muscle tension $F_i$ ($i=1, \ldots, 9$).

From Eq. (13) and Schur complement lemma, the condition $\mu - G^T G > 0$, that means $J$ is smaller than a positive number $\mu > 0$, becomes an LMI on $G$ ($F_i$, $i=1, \ldots, 9$) and $\mu > 0$ as the following:

$$\begin{bmatrix} \mu & G^T \\ G & I_9 \end{bmatrix} > 0$$

(14)

In addition, the inequality constraint condition for $F_i$ can be rewritten as the following linear inequality constraint condition for $G$:

$$G_{\text{min}} \leq G \leq G_{\text{max}}$$

(15)
Here, $G^{\min} = \left[ \begin{array}{c} F_1^{\min} \\ F_2^{\min} \\ \vdots \\ F_9^{\min} \end{array} \right]$, and 
$G^{\max} = [1 \ldots 1]^T$.

Moreover, the balance condition for joint moment (Eq. (10)), which is an equality constraint condition, can be approximated by the following linear inequality condition on $G$ with the use of very small positive numbers $0 < \varepsilon << 1$:

$$0 < \varepsilon \leq M^j - G^j \ H \ N^j \leq \varepsilon, \quad j = a, k, h$$

(16)

Where, $H := \text{diag}(F_1^{\max}, \ldots, F_9^{\max})$ and $N_j := [s_j L_i \ldots s_9 L_i]^T$ ($j = 1, \ldots, 9$). Equations (14), (15), and (16) are simultaneous LMI's for unknown parameter vector $G$, including unknown muscle tension $F_i$ ($i = 1, \ldots, 9$). Consequently, while fulfilling constraint condition equations (15) and (16), it is possible to obtain $G$ such that $\mu$ of equation (14) is globally minimized efficiently with a convex optimization program code that is commercially available. The unique combination of muscle tension $F_i$ ($i = 1, \ldots, 9$) that globally optimizes equation (11) can always be found.

**Myoelectric data**

In conventional muscle tension estimations, when evaluation indices such as equation (11) are optimized with the use of gradient methods or other techniques, the minimum muscle tension $F_i^{\min}$ produced in muscle $i$ is taken to be 0. The following equation is presumed to hold for muscle $i$ for which myoelectric data have been acquired.

$$q_i = F_i \ F_i^{\max} = \frac{EMG_i}{EMG_i^{\max}}$$

(17)

Where, $q_i$, $EMG_i$, and $EMG_i^{\max}$ represent the activity level ($0 \leq q_i \leq 1$), myoelectric data, and maximum myoelectricity, respectively, for muscle $i$. When evaluating muscle activity level by muscle tension and myoelectricity, equation (17) indicates that the two are equal. Equation (17) may be rearranged to get the following equation:

$$F_i = \left( \frac{EMG_i}{EMG_i^{\max}} \right) F_i^{\max}$$

(18)

$F_i$, of Eq. (18), obtained from myoelectric data and maximum muscle tension, is taken to be the minimum muscle tension of muscle $i$ during optimization. From this, the muscle tension estimated for muscle $i$ is at least $\left( \frac{EMG_i}{EMG_i^{\max}} \right) F_i^{\max}$ or greater, and by adding equation (17) to conventional muscle tension estimation problem, a method of estimating muscle tension that takes myoelectric data into consideration is possible.

**Muscle length and muscle contraction velocity**

Muscle tension changes due to the natural length of the muscle and its extension and contraction velocity.

The muscle tension $F_i$ of muscle $i$ is given as follows:

$$F_i = k(\xi_i) h(\eta_i, \xi_i) F_i^{\max} q_i$$

(19)

Here, $F_i^{\max}$ is the maximum muscle tension of a constant that does not depend on the strain or velocity of muscle $i$ determined from the anatomical viewpoint. $\xi_i$ and $\eta_i$ are the normalized length change and velocity of muscle $i$, respectively. Those normalized variables are defined as follows:

$$\xi_i = \frac{l_i - l_i^*}{l_i^*}, \quad \eta_i = \frac{\xi_i}{v_i^{\max}}, \quad v_i^{\max} = 3.0$$

(20)

In Eq. (20), $l_i$ and $l_i^*$ are the length and natural length of muscle $i$, respectively, and $v_i^{\max}$ is the maximum muscle extension and contraction velocity. Functions $k(\xi_i)$ and $h(\eta_i, \xi_i)$ are defined in the following equations (Figure 4-a, b):

$$k(\xi_i) = 0.32 + 0.71e^{(-1.112(\xi_i-1))} \times \text{sin}(3.722(\xi_i - 0.656))$$

(21)

$$h(\eta_i, \xi_i) = \frac{1 + \text{tanh}(a_1 \eta_i)}{a_2} - a_3 e^{(-2.6(\eta_i-1))}$$

(22)

$a_1$, $a_2$, $a_3$ : Constants determined by the type of muscle fiber

Each muscle $\xi_i$ is at its natural length in an upright position, and, by assuming that the muscle turns around a pulley as shown in Figure 3, each muscle $\xi_i$ can be sought from the angle of each joint obtained from exercise data. Considering equation (19), maximum muscle tension $F_i^{\max}(\xi_i, \eta_i)$ in a given posture and movement velocity $(\xi_i, \eta_i)$ is obtained from the following equation:

$$F_i^{\max}(\xi_i, \eta_i) = k(\xi_i) h(\eta_i, \xi_i) F_i^{\max}$$

(23)

By placing maximum muscle tension as in equation (22), it is possible to estimate muscle tension taking into consideration muscle length and muscle extension and contraction velocity.
Biomechanical analysis of conventional and FES rowing

Calculation of joint reaction force using a musculoskeletal model

Joint reaction force is the force on the contact surfaces of the joint; a force that breaks down the joint. The joint reaction force can be obtained as the vector resultant force of the intersegmental penetration force and muscle tension obtained above. The intersegmental penetration force is the force added to the body segment focused on from the adjacent body segment, but this force is an apparent force that appears in the process of calculating the joint moment. The joint reaction force is obtained by combining each intersegmental penetration force and muscle tension. The joint reaction force obtained from the intersegmental penetration force and muscle tension of the nine muscles is obtained from the following equations. In these calculations, it is assumed that the muscle tension is attached and acts parallel to the link:

\[ F_i = f_i + \sum F'_m \]  

\[ F_i(i = a, k, h) \] (a: ankle joint, k: knee joint, h: hip joint)

\[ f_i(i = a, k, h) \] (a: ankle joint, k: knee joint, h: hip joint)

\( F_i \) is the joint reaction force, \( f_i \) is the intersegmental penetration force of the joint, and \( F'_m \) is the muscle tension that straddles and is attached to joint \( i \).

Evaluations

In this study, changes in joint angle of hip, knee and ankle during rowing with or without handle and FES lower extremity extension exercise were measured at a rate of 60 Hz. The changes in joint moment of hip, knee, and ankle during rowing with or without handle and FES lower extremity extension exercise were calculated using a four link model. The changes in joint reaction force of hip, knee, and ankle during rowing with handle and without handle were calculated using a musculoskeletal model. The Stimulation voltages required to extend the lower extremities were also measured.

Results

Joint angle of lower extremity during rowing

During rowing with handle, joint maximum hip flexion and extension angles were 85.1±4.3° and -35.9±3.4° (Mean±SD), maximum knee flexion and extension angles were 98.0±4.2° and -4.0±3.9°, and maximum ankle plantar and dorsi flexion angles were 34.7±2.3° and 12.0±2.5°. During rowing without handle, joint angles, maximum hip flexion and extension angles were 78.6±2.3° and -36.4±3.1°, maximum knee flexion and extension angles were 97.2±3.8° and -7.9±3.7°, and maximum ankle plantar and dorsi flexion angles were 32.5±2.7° and 12.3±3.0° (Fig. 5). During FES lower extremity extension exercise, the lower extremity joint angles, maximum hip flexion and extension angles were 79.6±2.7° and -36.8±4.8°, maximum knee flexion and

![Graph](image-url)
extension angles were $93.8\pm3.9^\circ$ and $-2.7\pm3.0^\circ$, and maximum ankle plantar and dorsiflexion angle $37.4\pm2.2^\circ$ and $8.1\pm2.7^\circ$ (Fig. 6).

**Joint moment of the lower extremity during rowing**

During rowing with handle, maximum hip extension moment was $23.5\pm4.5$ Nm, maximum knee extension moment was $2.5\pm3.1$ Nm and maximum ankle plantar flexion moment was $11.9\pm2.5$ Nm. During rowing without handle, maximum hip extension moment was $8.0\pm3.9$ Nm, maximum knee extension moment was $4.1\pm4.0$ Nm and maximum ankle plantar flexion moment was $7.9\pm2.6$ Nm (Fig. 7). During FES lower extremity extension exercise, the lower extremity, maximum hip extension moment was $4.7\pm2.7$ Nm, maximum knee extension moment was $4.8\pm2.9$ Nm and maximum ankle plantar flexion moment was $7.5\pm1.8$ Nm (Fig. 8).

**Joint reaction force of the lower extremity during rowing**

During rowing with handle, maximum hip joint reaction force was $987.3\pm493.8$ N, maximum knee joint reaction force was $469.4\pm224.5$ N and maximum ankle joint reaction force was $423.0\pm164.4$ N. During rowing without handle, maximum hip joint reaction force was $651.1\pm172.7$ N, maximum knee joint reaction force was $293.7\pm106.4$ N and maximum ankle joint reaction force was $234.6\pm67.8$ N (Fig. 9).

**Stimulation intensity for FES rowing**

During FES lower extremity extension, the mean stimulation voltage of the electrical stimulator at which the lower extremities could be completely extended was $28.75\pm7.05$ mA.

**Discussion**

To develop effective and safe rehabilitation programs in
FES rowing, the biomechanical analysis of the lower extremity during rowing is very important. The change of joint angles of hip, knee and ankle joint during rowing was similar to the result of another report\(^{[16]}\), so that the rowing motion performed in this study was appropriate.

Derivation of joint moments and joint contact forces based on physical musculoskeletal models are the important biomechanical approaches to analyze the load on the body. There are few reports of biomechanical analysis of FES rowing\(^{[16,21]}\). In the past study, the optimization problem for estimating muscle tension for the calculation of joint moments and joint reaction forces was resolved by the conventional estimation methods with measured myoelectric data. In this study, the optimization problem was resolved by the linear matrix inequality (LMI) problem related to muscle tension. This is the first report to estimate joint moments and joint reaction forces during rowing exercise using the method of the LMI problem.

### Joint moments during rowing

In this study, differences in both displacement trends and values of joint moments were seen with and without the use of a handle. More specifically, moments occurred in the extension and plantarflexion directions in the hip and ankle joints in the early phase of leg extension both with and without the use of the handle. The values when the handle was used were about 20 Nm in the hip joint and about 10 Nm in the ankle joint, nearly twice as large as when the handle was not used. In the late phase of leg extension, when the handle was not used, joint moments acted on the extension and flexion directions in the hip and ankle joints, and extension direction in the knee joint. However, when using the handle, joint moments acted on the extension and plantarflexion directions in the hip and ankle joints and on the flexion direction in the knee joint during the period when the handle was being pulled; from the late phase of leg extension to the early phase of leg flexion. This is supposed to be because, in the hip and ankle joints, the leg is...
extended and plantar flexed and the knee joint is held firm in order to pull the handle. Afterward, in the late phase of leg flexion, joint moments showed similar displacement trends both with and without the use of the handle, and the value was about two times greater when the handle was used than when it was not used. In summary, the value of the leg joint moment showed a nearly 2-fold difference between using and not using a handle during the rowing exercise. Differences in the joint moment displacement trend when the handle was being pulled were thought to occur in order to pull the handle.

The displacement trends of the joint moments in the legs with and without the use of FES were nearly identical. More specifically, extension was mainly observed in the hip and ankle joints in the early phase of leg extension, while joint moment in the plantar flexion direction was seen in the knee joint in the late phase of leg extension. It was determined that moments occur in the flexion and dorsiflexion direction in the hip and ankle close to the early phase of leg flexion. The maximum joint moment generated in the flexion direction during flexion in the hip joint was about 30 Nm. It is thought that this is because a larger joint moment is needed here than in other joints in order to bend the trunk during leg flexion. The joint moments acting on the legs tended to be the same during leg extension regardless of whether or not FES was used.

**Joint reaction force during rowing**

The overall joint reaction force was larger when using the handle than when not using the handle. This is thought to be due to the influence of gripping the handle, and an increase in the joint reaction force was seen when the handle was pulled. Moreover, during this time, all of the leg joints exhibited the maximum joint reaction force. Differences in the joint reaction forces measured during normal rowing report maximum joint reaction forces in the hip, knee, and ankle joints of about 4,500, 4,000, and 2,000 N, respectively. At first glance, these forces may seem too large. However, the lever-arm, the distance to the bone attachment of the muscle exerting force to bend the joint, is only several centimeters. If the principle of the lever is considered, it is easy to comprehend, but since this distance is short the force that the muscle must generate theoretically increases, and the joint reaction force also increases. This result is from a 3-dimensional musculoskeletal model with 32 muscle attachments. Moreover, since leg extension and handle pulling are done almost simultaneously during rowing competitions, the reaction force of the foot is greater than when conducting these two motions separately. In the
present study, subjects were instructed to perform the motions separately, and the maximum reaction force measured from the foot was 50-100 N. In previous studies, in which rowing was performed with a competitive attitude, the maximum reaction force measured from the foot was 400-500 N. From this perspective, the value of the reaction force in the present study is about 1/4 to 1/5 of previous studies, and so a simple calculation of the joint reaction force is estimated to be about 1/4 to 1/5 that in previous studies. Through this calculation, the joint reaction forces are nearly equal, and so the results have validity.

Conclusions

Our results suggest that joint reaction force of lower extremity during rowing was lower than during walking and FES has the potential to be applied in the low-load exercise for the paraplegics. Further study is necessary to develop of FES rowing exercise.

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References


