Biomechanics of Rowing*
(I. A Model Analysis of Musculo-Skeletal Loads in Rowing for Fitness)

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Compared with the other exercise, such as walking and cycling, rowing was expected to have some fitness advantage, while there were some misgivings about the risk of injury. The objectives of this study were to quantify biomechanical characteristics of rowing for fitness and rehabilitation and to offer normative data for the prevention of injury and for determining effective exercise. An experiment was performed to collect the kinematic and kinetic data during rowing by experienced and non-experienced subjects. A three-dimensional whole-body musculo-skeletal model was used to calculate the biomechanical loads, such as the joint moments, the muscular tensions, the joint contact forces and the energy consumption. The results of this study indicate that rowing is an effective exercise for rehabilitation and fitness. However, the non-experienced rower should acquire considerable skill to obtain sufficient exercise. The rowing cadence should be decided according to the purpose of the exercise.

Key Words: Biomechanics, Human Engineering, Motion Control, Muscle and Skeleton, Ergometer Rowing, Inverse Dynamics

1. Introduction

Rowing a boat is whole-body exercise in which the movements of the upper and lower extremities cooperate together. Rowing is not only for competitive sport but also widely adopted as an exercise training method for general fitness using an indoor rowing ergometer. There are also some attempts to have a spinal cord injured patient row by using functional electrical stimulation in order to prevent cardiovascular disease(1,2), and rowing should be noticed

* Received 31st May, 2000 (No. 02-4091)
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difficulty is that a simple biomechanical model, such as a two-dimensional lower-extremity model, is insufficient and that a whole-body model is required to analyze the cooperative motion of rowing in the whole body.

The purposes of this study are twofold. One is to obtain biomechanical data in rowing, such as the joint moments and muscular tensions, from the measured motion data by using a three-dimensional whole-body musculo-skeletal model in order to quantitatively evaluate biomechanical characteristics of rowing for fitness and rehabilitation and to offer normative data for the prevention of injury and for determining effective exercise. The rowing data obtained are compared with the other exercise, such as cycling, to clarify the characteristics. Another is to examine the safety and effectiveness in taking up rowing as a fitness exercise based on the biomechanical data. The answers for the following questions are found: i) Are the energy consumption and joint loads for non-experienced rowers larger than those of experienced rowers? In other words, is there any question of safety and exercise effectiveness when non-experienced people row? ii) What can be expected when changing the rowing cadence? In this study, the problems are discussed based on biomechanical quantities, such as the muscular energy consumption and joint contact forces, calculated by the model analysis technique.

2. Method

Ten people were recruited as subjects. The experienced subjects were five male athletes (height = 1.76 ± 0.03 m; mass = 70.6 ± 3.1 kg; age = 20.4 ± 0.5 yr) who were members of the varsity crew in Keio University, Japan. The non-experienced subjects were five healthy male students (height = 1.75 ± 0.04 m; mass = 70.8 ± 2.3 kg; age = 22.6 ± 0.9 yr) who were active in other sports but did not row regularly with comparable physical fitness. Subjects with similar standing and height and body weight were selected to remove physical differences to elucidate the difference between the skilled and non-skilled motion.

The rowing ergometer (Model B, Concept 2, USA) for indoor training was used in the experiment. The ergometer consisted of the handle corresponding to an oar of a boat, the fan providing the mechanical loads by air damping resistance, the chain connecting the handle to the fan, the foot cradle fixing the feet, the seat that could move horizontally and the rail guiding the seat. The PSD (position sensitive detector) camera system (PSS-C5370, Hamamatsu Photonics, Hamamatsu, Japan) was utilized in the measurement of kinematic data. The LEDs (infrared emitting diodes) were mounted on the following anatomical points: the ankle, knee, hip, lumbar (top of iliac crest), neck, wrist, elbow, shoulder joints, the toe and the top of the head in the right side of the body in the sum total of 10 markers. Four markers were also fixed at the center of the fan of the ergometer, the upper and lower ends of the foot cradle, and the edge of the handle, respectively. Two PSD cameras were positioned unilaterally 5 m from the ergometer. They detected the movement of these LEDs, and three-dimensional positions were calculated by the DLT (direct linear transformation) method. The tension in the handle and the reaction forces in the foot cradle were measured as external forces acting on the body. The handle tension was measured by the tension transducer (T.K.K. 1269a, Takei Scientific Instruments, Japan) installed between the handle and the chain. The foot cradle was replaced with a custom-made force transducer to measure the foot reaction force. This force transducer was composed of 16 strain gauges, and could collect two components of force in the normal and anterior–posterior shear directions and the point of application in the anterior–posterior direction for the plate of the foot cradle. The measurement errors were 3 N in both forces and 3 mm in the point of application. In addition, in order to evaluate the muscular tensions calculated by the model analysis, electromyographic (EMG) data were collected using surface electrodes and the telemeter system (WEB-5000, Nihon Kohden, Japan) from seven muscles, i.e., the rectus abdominis, erector spinae, vastus, biceps femoris long head, latissimus dorsi, biceps brachii, and triceps brachii. The EMG signals were recorded on the digital audio tape recorder (PC208, Sony, Japan) at 1 000 Hz.

In the experiment, each subject rowed with his maximum effort at three different rowing cadences of 34, 26, 20 spm (stroke per minute). The experiment order was set randomly, and the enough time was taken between trials to collect data in the absence of fatigue. After the cadence reached steady state, kinematic and kinetic data were collected at 100 Hz for 7 sec to capture at least two successive strokes for each subject. In order to estimate the body segment parameters, the standing height, body weight, and segment lengths of each subject were also measured.

Since kinematic data of the right side only was measured, the motion in the left side was synthesized assuming motion symmetry. The kinematic and kinetic data measured were filtered with a Butterworth low-pass filter at 5 Hz to remove noise. The point in time in which the anteroposterior velocity of the handle changed from forward to backward was defined as a start point of the rowing cycle. These
points were used to define the rowing cycle, drive and recovery. The position of the hand and the direction of the chain of the ergometer were calculated from the positions of the markers on the edge of the handle and on the center of the fan, transforming the handle tension into the force vectors affecting each hand. By assuming that the mediolateral component of force could be disregarded, the foot reaction forces measured in two-dimension were transformed into three-dimensional force vectors with the mediolateral component zero. The foot reaction forces described in the local coordinate system on the foot cradle were then transformed into the global coordinate system by using the positions of the two markers fixed on the ends of the foot cradle. The EMG data recorded at 1000 Hz were full-wave rectified, smoothed with a Bartlett Window function and resampled at 100 Hz.

Various in vivo loads, such as joint moments and muscular tensions, were calculated from the measured kinematic and kinetic data by using a musculo-skeletal model (Fig. 1). Modifying our previous model\textsuperscript{10,11}, a three-dimensional whole-body musculo-skeletal model was used in the present study. Inertial properties and joint constraints of the body were represented as the three-dimensional 13-rigid-links system consisting of the feet, calves, thighs, pelvis, thorax, head, upper arms and forearms segments. The body segment parameters were estimated by using the regression formulae from the measured body weight and segment lengths\textsuperscript{13}. The joint constraints of the ankle, knee, lumbar, neck and elbow were assumed to have one degree of freedom, flexion-extension. The hip and shoulder joints were assumed to have three degrees of freedom, flexion-extension, abduction-adduction and internal-external rotation. In addition, the imaginary clavicle segments were introduced in the shoulder parts, connecting the upper arms to the thorax. The joints between the clavicle segments and the thorax were assumed to have two degrees of freedom. Therefore, the shoulders were modeled as the compound joints having five degrees of freedom.

The kinematic data were calculated from the positions of the markers by using the global optimization method described by Lu and O'Connor\textsuperscript{12}. In this method, the imaginary markers were initially fixed on the positions of the model dealing with the measured markers, and the joint angles and translations were then calculated using an optimization to minimize the total error in three-dimension between the measured markers and those of the imaginary markers. Segment velocity and acceleration were obtained by numerically differentiating the displacement. The joint moments were calculated using three-dimen-sional inverse dynamics based on the Newton-Euler method. In this method, the body segments were divided into the five groups of the straight-chain structure: the lower and upper extremities and the trunk. The equations of motion were solved recurrently from the terminal segment in each segment group, i.e., the foot, calf, thigh, forearm, upper arm, head, thorax and pelvis in that order. The moment and resultant reaction force at the lumbar joint were calculated from the equations of motion of the thorax segment. Therefore, it was theoretically unnecessary to measure the reaction force from the seat and to calculate the equation of motion of the pelvic segment.

The musculo-skeletal system was represented by 32 muscle models in the right lower and upper extremities and the trunk, while those in the left extremities were disregarded. The geometric configuration of the muscle was modeled as a series of line segments through the origin, insertion and via points. When it
Fig. 2 External forces in the experienced at 26 spm. The bold lines indicate the averages of the 10 trials (five subjects with two trials) and the thin lines indicate the one standard deviation of the data. The horizontal axis of each graph was time normalized to 0-100%, with drive starting at 0% and recovery finishing at 100%. The upper right graph demonstrates a relationship between the horizontal axis of each graph and rowing phases. The graph format is similar in the following Figs. 3-6. The handle tension simply means the tension force collected from the tension transducer. That is, it is equivalent to the resultant force acting on the both hands. The data for the foot reaction force represent the force acting on one leg for the global coordinate system. In this paper, the mechanical data were not basically normalized by the body weight, since the body weight was comparatively equivalent in all the subjects.

Fig. 3 Joint moments in the experienced at 26 spm. The ‘(F)’ and ‘(E)’ signs in the figure denote the directions of the flexion and extension moments, respectively. The stick diagram in the bottom figure indicates the direction of flexion (F) and extension (E) of each joint. The positive values of the joint moment show clockwise movements of the distal segments. Although the hip and shoulder joints have three degrees of freedom, the graphs illustrate only the flexion-extension moments. The directions of the flexion-extension axes in the hip and shoulder joints were defined as being parallel to the flexion-extension axes of the knee and elbow joints, respectively.

3. Results

Various biomechanical data measured and calculated in rowing by the experienced at 26 spm are shown in Fig. 2 to 6. Figure 2 denotes the measured handle tension and foot reaction force. The handle maximum tension reached approximately 850 N, and the peaks of the foot reaction force of the right leg were 450 N anteroposteriorly and 370 N vertically. Although the foot reaction forces peaked in the drive phase of rowing, they began to increase significantly in the latter half of the recovery phase. The joint
moments are shown in Fig. 3. The extensor moments of the ankle, knee, hip and lumbar joints all peaked at the beginning of the drive phase. The peak of the knee extensor moment was approximately 40 Nm. It was not so large; rather, the flexor moment appeared in the latter half of the drive phase. The elbow and shoulder flexor moment peaks occurred just after the extensor moment peaks of the lower extremity.

Figure 4 indicates the muscular tensions of 15 muscles of the 32 muscles modeled. In the figure, the EMG patterns collected from the seven muscles are superimposed on the graphs of the muscular tensions dealing with the muscles. The calculated muscular tension patterns roughly agreed with the EMG patterns. One exception was the biceps brachii, which did not agree with the EMG pattern. Considerable co-contraction was found in the antagonist muscles in the lower extremity.

Figure 5 shows the changes in the amplitudes of the joint contact forces. The peaks reached approximately 4,900, 4,400 and 4,100 N in the lumbar, hip and knee joints, respectively, during the drive phase. As for the ankle joint, its peak did not appear in the drive phase, but reached 2,800 N in the latter half in the recovery phase. Although these figures show only the amplitudes of the joint contact forces and not the directions, the forces always acted in the direction of compression. The handle tension pulled the joints in...
Fig. 5  The amplitudes of the joint contact forces in the experienced at 26 spm

Fig. 6  Muscular metabolic power in the experienced at 26 spm. The bold and thin solid lines indicate the average and the standard deviation of the muscular metabolic power summed in the whole-body. The line with circles shows the muscular power in the lower body. The line with asterisks shows the muscular power in the upper body.

the upper extremity, while the joint contact forces were compressive if they were computed based on the muscular excitation.

Figure 6 indicates the muscular metabolic power summed in the whole-body, upper-body and lower-body groups. In general, large muscular metabolic power was demonstrated for the drive phase. Dividing the body muscles into the lower-body and upper-body groups, the muscular metabolic power in the lower-body group peaked at the end of the recovery phase, however began to increase in the middle of the drive phase, and the peak power of the upper-body group appeared in the latter half of the drive phase.

The ratio of the total muscular metabolic energy between the lower and upper body groups was approximately 6:4.

The rowing patterns in the experienced and non-experienced were compared using the following evaluative indices (Fig. 7). i) The handle tension (HT): Time average of the handle tension. ii) The muscular metabolic energy (MME): Time average of the sum of the metabolic energy consumed in the muscles for the whole-body. iii) The joint contact force (JCF): Time average of the sum of the joint contact forces for the whole-body. iv) Ratio of MME to JCF (MME/JCF). Compared with these indices, the HT of the non-experienced was less than that of the experienced by 19%. The MME and JCF of the non-experienced were less than the experienced by 32% and 27%, respectively. The non-experienced was also less than the experienced in the ratios of the indices, MME/JCF. All the differences were significant (p<0.05).

Compared with the indices mentioned above, the effect of the change in the rowing cadence was examined for the metabolic energy consumption and joint contact forces (Fig. 8). As a result, all the HT, MME and JCF increased as the rowing cadence increased. As the rowing cadence increased, the MME/JCF also increased. All the variance was significant (p<0.05).

4. Discussions

The accuracy and validity of these results is dependent on the musculo-skeletal model used. The muscles around the shoulder joint have large attachment areas to the bones, i.e.: trapezius and deltoid.
Fig. 8 Comparison with the rowing cadences. The data for each rowing cadence included both the experienced and non-experienced. That is, the each graph denotes the average of the 20 data (ten subjects with one type of the rowing cadences with two trials) and the standard deviation. Each graph was normalized by the data of 26 spm.

muscles, and it is difficult to model them as simple line segments. The present model does not accurately express the geometric configurations of these muscles. The inaccuracy of the musculo-skeletal model might be a reason why the muscular tension pattern in the biceps brachii did not closely agree with the measured EMG pattern. We must realize that the model did not accurately calculate the mechanical load affecting the shoulder joint. Furthermore, the lumbar model was composed only of one pin joint, and did not include any spinal curvature or intra-abdominal pressure. It might, therefore, be difficult for the present model to analyze the mechanism of a backache, which is one of the most common injuries in rowing. The structures of the other joints were also simplified as pin or ball joints, making it impossible to evaluate the tangential and normal forces. Therefore, the results of the present study presented amplitudes of joint contact forces. Calculation of joint moments using an inverse dynamics model has been established by many researches\(^{16}\), and has widely been accepted in the biomechanical field. On the other hand, the calculation method of muscular tensions using a static optimization has not been established, and there are many opinions on determining the objective function and the calculation technique itself\(^{16,17}\). Although we recognize these limitations, we believe that one advantage of the present study is the comprehensive analysis of rowing. That is, the present study analyses the complex dynamic motion of rowing including the external forces using the complex model representing the three-dimensional musculo-skeletal structure of the whole-body. Since the whole-body musculo-skeletal structure was considered in the model, the model could evaluate both aspects of comprehensive human movement from a macroscopic point of view, and the precise muscular activity of the individual muscle from a microscopic point of view. There are few studies similar to the present one analyzing practical human movement using a complicated model\(^{16}\). We believe that the present study offers novel results that are difficult for the other biomechanical model and analytical method.

The mechanical loads on the musculo-skeletal system during rowing must be compared to the other exercises, such as walking and cycling. An impact external force, such as ground reaction force during running, did not appeared in rowing, though the peaks of the handle tension was larger than the subject's body weight. The foot reaction force in rowing was less than the subject's body weight or that in walking. As for the joint moments, although the large extension moments in the hip and lumbar joints are understandable, the extension moment at the knee joint was smaller than expected. The peak extension moment in the knee joint during walking has been measured as 40 Nm\(^{16}\). The peak extension moment during rowing was almost equivalent to walking. However, the knee joint contact force increased in relation to the muscular tensions, reaching 4.100 N (i.e., six times the body weight) during rowing. The peak contact force in the knee joint has been estimated at three times bodyweight (BW) during walking\(^{20,21}\), two times BW during cycling\(^{23}\) and seven times BW when rising from a chair\(^{20}\). The subjects rowed with their maximum effort in the experiment of this study, and we should consider the difference of the effort levels when comparing to the other exercises. However, the joint contact forces in rowing were expected to be small or similar to that in walking, as there is no need to support body weight on the lower extremities and the foot reaction force in rowing was less than walking. Actually, the results calculated in this study show the joint contact force amplitudes to be quite large. The reason might be caused by the powerful co-contraction of the antagonist muscles, such as the vastus and hamstrings, as shown in Fig. 4. Due to the co-contraction, the joint contact force was large in spite of the small foot reaction force and net joint moment similar to walking. From a rehabilitation point of view, rowing has lower impact force and therefore lower risk of injury, compared to running. However, appropriate exercise during rowing occurs, that allows muscle endurance, muscle strength and bony density to increase. It is not evident why the antagonistic muscles co-contract during rowing, and a detailed elucidation of the mechanism is one for future examination. The problem of co-contraction is important not only for basic biomechanics research, but also for
the clinical application of rowing exercise. It is necessary to understand the muscular activity patterns in rowing, in order to establish a rowing system for exercise therapy of spinal cord injured patients using functional electrical stimulation, which is one of the final applications of our research.

The muscular metabolic power patterns shown in Fig. 6 indicate that rowing is an ideal whole-body exercise in which the metabolic energy consumption of the upper and lower bodies was almost equivalent. There is no research to compare the muscular activity in the upper body with that of the lower body during walking and cycling to the best of our knowledge. We believe that walking and cycling will provide sufficient exercise for the upper body. Although a cycle ergometer for spinal cord injured patients have been proposed \(^{24}\), the machines are complicated and expensive, and many muscles are not always involved in the exercise. On the other hand, rowing enables many muscles to be adequately stimulated by a single motion with a comparatively simple machine. An interesting feature of rowing is that the energy consumption rate changed significantly between the drive and recovery phases. In the drive phase, a large power was generated over a short period, and in the recovery phase, the power was minimal over a long period. Seventy-five percent of the energy in one rowing cycle was consumed in the drive phase, which occupied 37% in the time of one rowing cycle. Such polarization of the energy consumption became more remarkable, as the rowing cadence became slower. Although the fitness effect of the polarization of the energy consumption in rowing is not always obvious, we suppose that rowing might provide a fitness effect similar to the interval training. In addition, note that the power and energy described above do not correspond to the energy consumption measured by the expiration gas analyzer, because they were consumed by the muscles only, and did not include the energy consumed in non-muscular activity, such as the viscera.

In the comparison between the experienced and the non-experienced rowers shown in Fig. 7, the larger HT found in the experienced rowers indicates that they generated a larger driving force for a boat than the non-experienced. The larger MME and JCF in the experienced rowers indicate that they are subjected to larger loads, such as the energy consumption and the joint contact force. The comparison between the experienced and the non-experienced was motivated by the anxiety that skill-less rowing resulted in excessive loads on the musculo-skeleton, and posed a danger for injury. However, the results indicated that the problem of the non-experienced rowers was not excessive loads on the musculo-skeleton but insufficient exercise to improve fitness. This agreed with the subjective evaluation by the non-experienced who reported no fatigue unexpectedly because they could not row well. It is concluded that non-experienced rowers should acquire considerable rowing skill even if the purpose is just fitness. Furthermore, the larger MME/JCF in the experienced rowers indicate that they consume larger metabolic energy in relative comparison to the joint contact force. Therefore, skilful rowing might be better in order to provide cardiovascular effect.

The most notable result in the graphs of Fig. 8 is the MME/JCF. The increasing MME/JCF with the rowing cadence means that the energy consumption increased with the joint contact forces, as rowing cadence increased. From a clinical point of view, the result suggests that the rowing cadence should be set higher if the purpose of the exercise is to lose weight or to improve cardiovascular endurance by consuming metabolic energy. On the other hand, the rowing cadence should be set lower if the purpose of the exercise is to improve bone density or muscular strength by increasing mechanical load on the body. The optimal cycling cadence has been estimated in the biomechanical studies of cycling \(^{23,26}\). Such an optimal cadence is also expected to exist for rowing, and it is important especially for competitive rowing to investigate it. It is also necessary for rowing for fitness to clarify an optimal cadence. However, the optimum might be dependent on the purpose of the exercise as mentioned above.

5. Conclusions

In conclusion, the present study has calculated the biomechanical loads, such as joint moments, muscular tensions and muscular energy consumption, in ergometer rowing for fitness by using a musculo-skeletal model. The muscular activity in rowing was most notable for co-contraction of antagonist muscles in the lower extremity. The muscular tensions and the joint contact forces were considerable large, such as 4100 N on the knee joint, because of the co-contraction. The energy consumption was almost equivalent in the upper and lower bodies, and its rate changed significantly between the drive and recovery phases. The non-experienced rower was lower than the experienced in energy consumption; therefore, it is necessary for the non-experienced to improve rowing skill before they can improve fitness. Rowing with a higher cadence consumed more energy for endurance training, and rowing with a lower cadence resulted in larger joint contact forces potentially for improving bone density.
References


