INTRODUCTION

Rowing is a whole-body exercise that requires co-ordination of the motions of the upper and lower extremities and trunk. Indoor ergometer rowing has long been used by competitive rowers as a training method, but it has also been widely adopted by non-rowers as an exercise for general fitness. A knowledge of the loads in the muscles and joints of the body during ergometer rowing is important not only for studies of sports performance, but also for identification of the tissues potentially at risk of injury, particular in non-rowers who may have poor rowing technique. The purpose of this study is to propose three types of the musculoskeletal model for a comprehensive biomechanical study of ergometer rowing: a three-dimensional whole-body model based on inverse dynamics (Model 1), a quasi-three-dimensional whole-body model based on forward dynamics (Model 2), and a two-dimensional lower extremity model with anatomical joint models based on inverse dynamics (Model 3).

METHODS

Experimental Methods.

Five male subjects 23.4 years of age (20-27), 91.6 kilograms (83-97), 195 cm (191-198) and with 9.6 years of competitive rowing experience (8-13) participated in the study.

Seventy-four markers 15 mm diameter markers were attached to the upper extremities, lower extremities, trunk and head to define seventeen segments. These marker placement and Cartesian coordinate systems defining segments have been described previously (Halliday et al. 2001). Kinematic data were collected with 12 MII cameras run by a Vicon 612 (Vicon Motion Systems, Oxford, UK) at 100 Hz. A Concept II Model C ergometer was instrumented with an AMTI MC36-6-250 force transducer (AMTI, Massachusetts, USA) beneath the right foot cradle. The rowing ergometer itself was made into a custom calibration object and the force transducer was rigidly attached to it. This allowed the integration of the kinematic and kinetic data to an accuracy of 3.05 mm.

The chain that joins the handle to the flywheel of the ergometer was cut and modified to insert a uniaxial transducer (DDE – 2500 Load Cell #7629) in series (Applied Measurements Ltd. Aldermaston, UK). Kinetic data from all six components of the foot cradle transducer and the tension transducer in series with the ergometer handle were sampled at 1000 Hz. EMG data were collected from ten muscles (tibialis anterior, gastrocnemius, soleus, rectus femoris, vastus medialis, semimembranosus, gluteus maximus, adductor magnus, erector spinae at the fifth lumbar vertebrae and rectus abdominus) using an MA 100 (Motion Lab Systems, Louisiana, USA) also sampled at 1000 Hz.

All kinematic and kinetic data were filtered with a Butterworth low-pass filter at 5 Hz to remove noise. The kinematic data were then processed using the global optimisation method of Lu and O’Connor (1999). Segment velocity and acceleration were obtained by numerically differentiating the marker positions. The point in time at which the anteroposterior velocity of the handle changed direction was used to divide the rowing cycle into two phases: drive and recovery.

Three-dimensional inverse dynamics model (Model 1).

Various in vivo loads, such as joint moments and muscle forces, were calculated from the measured kinematic and kinetic data by using a three-dimensional, whole-body musculoskeletal model (Figure 1) based on our previous work (Hase and Yamazaki 2002). The model comprised 13 rigid segments: feet, calves, thighs, pelvis, thorax, head, upper arms and forearms segments. Body segment parameters were estimated using regression formulae from measured body weight and segment lengths. The model joints at the ankle, knee, lumbar spine, neck, and elbow all had one degree of freedom, flexion-extension. The model hip and shoulder joints were taken to have three degrees of freedom: flexion-extension, abduction-adduction and internal-external rotation. In addition, 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 modelled as the compound joints having five degrees of freedom in total.

Joint forces and moments were calculated using three-dimensional inverse dynamics based on the Newton-Euler method. The equations of motion were solved sequentially from the terminal segment in each limb: foot, calf, and thigh for the lower limb, and forearm, upper arm, head, thorax and pelvis for the upper limb and trunk. The resultant force and moment at the lumbar joint were calculated from the equations of motion of the thorax segment.

The musculoskeletal model included 32 muscles in the right lower and upper extremities and trunk: 1-right rectus abdominis, 2-left rectus abdominis, 3-right erector spinae, 4-left erector spinae, 5-right neck extensor, 6-left neck extensor, 7-right neck flexor, 8-left neck flexor, 9-psosas major, 10-Iliacus, 11-gluteus maximus, 12-gluteus medius, 13-gluteus minimus, 14-adductor, 15-vastus, 16-biceps femoris short, 17-rectus femoris, 18-hamstrings, 19-tibialis anterior, 20-soleus, 21-gastrocnemius, 22-deltoides front,
Three-dimensional dynamics model (Model 1). Figure 1: Diagram of the three-dimensional model in control except for the shoulder joint. So, we call words, although the model was three-dimensional in one degree of freedom in the flexion and extension. In other joints, the shoulder joint had three rotational degrees of freedom, and the thighs, pelvis, thorax, upper arms, and forearms. The inertial properties of the human body were represented as EMG patterns. Such computer simulation model allows the system mentioned the blow without any measured data such as EMG patterns. The muscles included in the musculoskeletal model are as follows: 1-rectus abdominis, 2-erector spinae, 3-neck extensor, 4-neck flexor, 5-psoa major, 6-iliacus, 7-gluteus maximus, 8-gluteus medius, 9-gluteus minimus, 10-adductor, 11-vastus, 12-biceps femoris short, 13-rectus femoris, 14-hamstrings, 15-tibialis anterior, 16-soles, 17-gastrocnemius, 18-deltoideus front, 19-deltoideus middle, 20-deltoideus back, 21-triceps brachii long, 22-biceps brachii long, 23-triceps brachii med, 24-brachialis, 25-latissimus dorsi, 26-pectoralis major, 27- trapezius, 28-teres major. Each muscle was represented as a series of line segments connecting the origin, via points and insertion. The configuration of each muscle was modelled as a series of line segments through the origin, insertion, and via points (located on the edges of polyhedrons which were used to model the shapes of the bones). When a via point changed as a result of joint movement, its position was found by minimising muscle length between origin and insertion. The inter-segmental moments calculated from inverse dynamics were distributed amongst relevant muscles using the geometric constraints of the model and a static optimisation. The objective function that was minimised was defined as the cube of muscle stress, which was taken to represent muscular fatigue (Crowninshield and Brand 1981). The model joint contact forces were found using the inter-segmental forces found from inverse dynamics and the relevant muscle forces calculated as described above. Ligaments forces were not included in the model.

Quasi-three-dimensional forward dynamics model (Model 2).

The second musculoskeletal model proposed in this study was based on forward dynamics in which body movement was generated from joint moments or muscle forces without any motion data captured. Furthermore, the joint moments and muscle forces were synthesized based on the control system mentioned the blow without any measured data such as EMG patterns. Such computer simulation model allows us to conduct computational experiments that would be difficult to actually measure otherwise.

The musculoskeletal model similar to the Model 1 mentioned the above was employed as shown in Figure 2. The inertial properties of the human body were represented by a 12-rigid-link system consisting of the feet, calves, thighs, pelvis, thorax, head, upper arms, and forearms. The shoulder joint had three rotational degrees of freedom, and the other joints were represented by the hinge joints with one degree of freedom in the flexion and extension. In other words, although the model was three-dimensional in geometry, it was almost identical to a two-dimensional model in control except for the shoulder joint. So, we call this a quasi-three-dimensional model. The mechanical constraint between the seat and the body was equivalent to that in the other joints. That is, the pelvis segment was combined with the rail on the ergometer through a two-degree-of-freedom virtual joint with the translation along the rail and with the anterior-posterior tilt. No driving force was assumed to affect the virtual joint. In order to calculate the reaction forces of the handle and foot cradle, the visco-elastic elements were assumed to be between the handle and the upper extremities and between the foot cradle and the feet so that the forces were calculated from the displacement and velocity in the visco-elastic elements. This model assumed a rowing ergometer with a flywheel containing a fan that generated the resistance force. The resistance force of the ergometer was calculated from the wheel’s properties and air-damping resistance.

The musculoskeletal system of the model consisted of 56 muscle models of the entire body. The geometric configuration of each muscle was represented by a series of line segments connecting the origin, via points and insertion. The muscles included in the musculoskeletal model are as follows: 1-rectus abdominis, 2-erector spinae, 3-neck extensor, 4-neck flexor, 5-psoa major, 6-iliacus, 7-gluteus maximus, 8-gluteus medius, 9-gluteus minimus, 10-adductor, 11-vastus, 12-biceps femoris short, 13-rectus femoris, 14-hamstrings, 15-tibialis anterior, 16-soles, 17-gastrocnemius, 18-deltoideus front, 19-deltoideus middle, 20-deltoideus back, 21-triceps brachii long, 22-biceps brachii long, 23-triceps brachii med, 24-brachialis, 25-latissimus dorsi, 26-pectoralis major, 27- trapezius, 28-teres major. Each muscle exits on the right and left sides of the body, so the muscular models total 56.

The motion control system was based on the gait simulation model including the neural oscillators reported previously (Hase 1999). The control model consisted of the feedback system of the somatic senses from the musculoskeletal system, the neural oscillators representing the central pattern generator (CPG) in the nervous system, and the peripheral system to determine neuronal stimulus to each muscle. In the present study, the previous control model for walking was modified so as to generate rowing motion and to generalize the methodology of the motion control design as follows: (1) the sensory feedback system was represented by the combination of the PD (Proportional Derivative) feedback and the inverse dynamics of the musculoskeletal system, (2) simplifying the formulation of the neural oscillators, a first order lag element was employed as the neuronal model, (3) dividing the objective motion into several events, PD gains were adjusted for each motion event, (4) the output signal from the neural system was distributed to individual muscle tensions by using a static optimisation method in the peripheral system. There were more than 100 control parameters to be determined in the neuronal control system. They were determined using search computing with an optimisation method, such as genetic algorithms. The objective function for optimization was defined as the combination of the maximisation of the energy efficiency and the smoothness of the muscle force patterns.
Two-dimensional anatomical joint model (Model 3).
The third model was introduced to investigate problems of the Models 1 and 2 as mentioned the below. In the Model 3, only lower extremity was modelled two-dimensionally (Halliday 2003).

The above mentioned models were formulated based on three-dimensional equations of motion. However, each joint was modelled by either a hinge joint or a ball joint, so the model could not represent the precise mobility in the actual anatomical joint. In order to solve such problem with lower calculation cost, four-bar linkage models (O'Connor et al. 1989) were introduced as the alternative joint models with ligaments (Figure 3).

The mobility of the tibiofemoral joint was controlled by a four-bar linkage, formed by the two cruciates, ACL ad PCL, along with the femur and the tibia. The orientations of the ligaments and the contact force are determined from the four-bar linkage. Forces that are tangential to the articular surfaces have to be balanced by ligament forces and/or muscle action. The patella was represented as a point, the intersection of the quadriceps and patellar tendons. The model determines the changes in line of action of the patellar tendon, quadriceps tendon and patellofemoral contact forces, throughout the full range of knee flexion (Lu et al. 1998).

The mobility of the ankle joint was also controlled by a four-bar linkage (Leardini et al. 1999). The tibia/fibula and the talus/calcaneus were taken to be the two rigid bone segments of the model. The tibiofibular and tibiocalcaneal ligaments were assumed to be inextensible line segments. Furthermore, the metatarsophalangeal (MTP) joint was added to the foot segment to represent the flexible movement of the foot during rowing. The hip and MTP joints were modelled as hinge joints.

Another problem of the Models 1 and 2 is the validity computing the muscle forces from net joint moment or quasi-joint moment from the neuronal control system. In order to investigate muscle activities during rowing, the solution way for rowing motion should be also investigated. In other words, the associated indeterminate problem was formulated to distribute the joint moments to individual muscle forces. In the third model, a couple of computing methods were implemented to compare each other: the Dynamically Determine One-Sided Constrained (DDOSC) method (Collins 1995), and static optimisation methods.

The DDOSC method decomposes the original indeterminate problem into a series of dynamically determinate (DD) sub-problems established by considering at any one time only the number of the unknowns that makes the original problem determinate, setting the values of other forces to zero. In some solutions, the directions of the contact forces at the hip and ankle were calculated; in others, these were specified in advance to be along the long axis of the femur or tibia, respectively. Solutions corresponding to the DD sub-problems are referred to as DD solutions. Since muscles and ligaments transmit only tensile force and joint articular surfaces resist only compression, DD solutions that violate these one-sided constraints are rejected. The remaining DD solutions are referred to as DDOSC solutions.

As the static optimisation method in the Model 1 and Model 2, the minimisation of the cube of the muscle force was used to be taken to represent muscular fatigue. In the Model 3, several objective functions were proposed in order to compare each other: minimisation of muscle forces, minimisation of the cube of muscle forces, minimisation of ligament forces and minimisation of joint contact forces.

RESULTS AND DISCUSSION

Figure 4 indicates the muscle forces of 6 muscles of the 32 muscles modelled in the Model 1. In the figure, the EMG patterns collected from the six muscles are superimposed on the graphs of the muscle forces dealing with the muscles. The calculated muscle force patterns roughly agreed with the EMG patterns.
Figure 4: Muscle forces calculated from the Model 1. The bold and thin lines in the graphs show the average and the standard deviation of the muscle force, respectively. The grey zones indicate the widths of the +/- one standard deviation of the smoothed EMG data. The EMG data are scaled to equalize the amplitudes of the peaks with those of the muscle force patterns.

Figure 5: Comparison of the joint angles simulated by the Model 2 (bold lines) and obtained from the Model 1 (grey zones representing +/- one standard deviation). Figure 5 is a comparison of the joint angles simulated by the Model 2 and the ones measured based on the Model 1. The ‘(F)’ and ‘(E)’ signs in the figure denote the direction of the flexion and extension, respectively. As shown in the figure, the hip and knee joints were fully flexed, and the upper extremities were extended in front of the body at the beginning of the drive phase. At the end of the drive phase, the hip and knee joints were extended, while the shoulder and elbow joints were fully flexed. The pattern of joint motions generated by the Model 2 closely corresponded to those of actual rowers in the Model 1.

Figure 6: Comparison of the joint moments simulated by the Model 2 (bold lines) and obtained from the Model 1 (grey zones representing +/- one standard deviation).

Figure 6 is a comparison of the joint moments simulated by the Model 2 and those estimated from the experimental data using the Model 1. Large extension moments were generated in the lumbar, hip, knee, and ankle joints in the first drive phase. One remarkable difference in the simulation and measurement was observed in the large flexion moment of the ankle joint in the recovery phase. Another was that the extension moment of the knee joint did not occur at the end of the recovery phase in the simulation. The body of the model was returned to the fully flexed position by the flexion moment of the ankle joint in the recovery phase. This particular rowing style is often observed in a novice’s rowing. The simulated rowing by the Model 2 might be equivalent to that by an unskilled rower.

Figure 7 shows the mobility of the anatomical joint model of the knee in the Model 3 at 10, 90 and 140 degrees flexion. It is noted that the instant centre of the model knee moves relative to each bone and that the tibiofemoral contact point moves posteriorly during flexion. The contact point on the patella moves from the distal pole to the proximal pole as the patella rolls on the femur. The anatomical model could fully represent the mobility of the knee joint during rowing with the wide range of motion.

Figure 8 is the muscle force patterns based on the several computational methods in the Model 3. Depending on the computing methods, various muscle force patterns could be calculated from the identical net joint moment.

In the present study, we have proposed three types of musculoskeletal models to investigate from various biomechanical points of view. For example, if whole body motions are investigated based on the captured motions, the Model 1 should be used. If the biomechanical properties including the kinematic motion itself are predicted without any experiment, the Model 2 should be used as the prediction tool on biomechanics. Furthermore, if anatomical properties of the joint and ligaments and the computational method of muscle forces are investigated more precisely, the Model 3 should be employed.
In future work, these three models will be combined into one comprehensive musculoskeletal model to analyze ergometer rowing motion for competitive sport and fitness. Also, the validation of each model will be checked.

REFERENCES

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**Figure 7**: The model knee joint in the Model 3.

**Figure 8**: Muscle forces in the Model 3.