Steady-state force–velocity relation in human multi-joint movement determined with force clamp analysis

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Abstract

To study the force–velocity characteristics of human knee–hip extension movement, a dynamometer, in which force was controlled by a servo system, was developed. Seated subjects pressed either bilaterally or unilaterally a force plate, a horizontal position of which was servo-controlled so as to equalize the measured force and a force command generated by a computer at a time resolution of 2 ms (force clamp). The force command was based on the relation between maximum isometric force and foot position within the range between 70% and 90% of “leg length” (LL: longitudinal distance between the sole of the foot and the hip joint), so that the same force relative to the maximum isometric force was consistently applied regardless of the foot position. By regulating the force according to this function, the force–velocity relation was determined. The force–velocity relation obtained was described by a linear function (n = 17, \( r = -0.986 \) for 80% LL, \( r = -0.968 \) for 85% LL) within a range of force between 0.1 and 0.8 \( F_0 \) (maximum isometric force). The maximum force extrapolated from the linear regression (\( F_{\text{max}} \)) coincided with \( F_0 \) (n = 17, \( F_0/F_{\text{max}} = 1.00 \pm 0.09 \) for 80% LL and 1.00 \( \pm 0.20 \) for 85% LL). Also, the velocity at zero force (\( V_{\text{max}} \)) was obtained from the extrapolation. When compared to the bilateral movements, unilateral movements gave rise to a smaller \( F_{\text{max}} \) but the same \( V_{\text{max}} \), suggesting that \( V_{\text{max}} \) is independent of force and therefore represents the proper unloaded velocity. It is suggested that some neural mechanisms may be involved in the force–velocity relation of the knee–hip extension movement, and make it exhibit a linear appearance rather than a hyperbola.

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1. Introduction

The relation between force and shortening velocity in contracting muscle (the force–velocity relation) has been extensively studied from molecular (Ishii et al., 1997) to human in vivo movements (Wilkie, 1950). It provides us with information about not only mechano-chemical characteristics of the actin–myosin interaction (Huxley and Simmons, 1971) but also the contractile properties of human movements. At all levels, the velocity decreases with the increase in force and becomes zero when a force eventually reaches the maximum isometric force. The force–velocity relation of skeletal muscle of the frog has entirely been described with either an exponential (Fenn and Marsh, 1935) or hyperbolic function (Hill, 1938), the latter of which has been more widely accepted. Wilkie (1950) has conducted a detailed study on the force–velocity relation in human single
joint movement under the isotonic condition. He confirmed the hyperbolic nature of the force–velocity relation postulated by Hill (1938) after he corrected it for inertia. Therefore, it can be assumed that the hyperbolic nature of the force–velocity relation appears to be kept essentially unchanged up to the level of single-joint movements around a small joint. However, it remains unclear whether multi-joint movements in vivo are also characterized in a similar manner.

The muscle function during multi-joint movements has not been extensively studied because of their complexity. So far, studies on the force–velocity relation have been reported for some multi-joint movements, such as half-squat (Bosco et al., 1995; Rahmani et al., 2001), leg press (Macaluso and De Vito, 2003; Pearson et al., 2004) and leg or arm cranking (Driss et al., 1998; Vandewalle et al., 1987) exercises. However, interpreting the shape of the force–velocity relation determined in these studies needs caution, because the forces were not controlled throughout the movements. The force may change with time due mainly to the effect of inertial mass. It has been shown that the force–velocity relation is deviated to a large extent from its original hyperbolic curve when the force is continuously changed (Ishii et al., 1997; Iwamoto et al., 1990). The problem would be solved if the “force clamp” method would be used to control rapidly the force by means of a feedback system. Although this method has been successfully used in studies with single muscle fibers (Ford et al., 1977), its application to human movements is challenging because of the need to control a large force (∼10^3 N) in addition to the effect of a large inertial mass.

The aim of the present study was to determine the force–velocity characteristics of human multi-joint movements in a condition close to steady state. For this purpose, we developed a dynamometer, in which force was controlled by the high time-resolution servo system for knee–hip extension movement.

2. Methods

2.1. Experimental procedure

The methods and all procedures used in these experiments accorded with current local guidelines and the Declaration of Helsinki, and were approved by the Ethical Committee for Human Experiments, The University of Tokyo. All of the subjects were informed about the experimental procedure and the purpose of the study prior to study onset. Written consent was obtained from all participants.

The study was conducted in two separate experimental phases. In the first part of experiments (experiment 1), the relation between percentage of “leg length (LL)” and the force-generating capacity during the isometric knee–hip extension movements was determined and used as a simplified theoretical function of force during the dynamic movements. The longitudinal distance between the sole of the foot and the hip joint (greater trochanter of the femur) was determined here as LL. Accordingly, in the major part of experiments (experiment 2), the force–velocity relation of knee–hip extension movements was determined by regulating the force with a servomotor so as to keep the relative force constant during the dynamic movements.

2.2. Subjects

A total of 49 recreationally active men and women volunteered to participate. In experiment 1, a few subjects (n = 10; age, 36.5 ± 11.9 yr; height, 164.9 ± 10.0 cm; body mass, 56.5 ± 10.1 kg; mean ± SD) participated because it was required to set parameters for the servo system that had to work over a wide range of movement. Also, it was required for fine modifications in mechanism and electric parts in the apparatus. In experiment 2, subjects of the remainder (n = 39; age, 27.8 ± 13.3 yr; height, 161.2 ± 7.5 cm; body mass, 54.7 ± 7.6 kg; mean ± SD) were used to determine the force–velocity relations. In each experiment, subjects wore the same short training pants and flat sole shoes to minimize the external effects on performance. Before each experiment, subjects warmed-up with few minutes walking and stretched mainly the quadriceps, hamstrings and triceps surae muscles.

2.3. Dynamometer

A dynamometer was developed in collaboration with Matsushita Electric Works Ltd. (Osaka, Japan) for precisely controlling force and displacement during the knee–hip extension movement. The outline of the system is illustrated in Fig. 1. The system consisted of a vertically placed tri-axial force plate (MC3A-X-500, AMTI Inc., USA), a servomotor (MSM352A1G, Matsushita Inc., Japan) and a computer-assisted control unit to produce a servo loop. The force plate was connected to a servomotor, the position of which was detected with a rotary encoder placed co-axially with the servomotor. A recline angle of the seat was set at 55°, which was previously determined by EMG measurements showing the most balanced activation of knee and hip extensor muscles without large stress on back muscles during the knee–hip extension movement (data not shown).

The force clamp, which is the method to keep the force constant during the shortening of muscles (isometric contraction), was established by using the servo feedback control of displacement. In the present study, displacement of the force plate was controlled with a function of force (“force command”) so as to keep the
force relative to the maximum isometric force constant at any position. The force command was generated by a personal computer (SL2-A50K, Epson Inc., Japan) and sent to a servo amplifier (MSD353A1V, Matsushita Inc., Japan). An output from a force plate was introduced to a computer and fed back to a servo amplifier, which then generated a signal for controlling a servomotor so as to cancel the difference between the force command and actually measured force. For a servo control, only vertical component of force with respect to the surface of force plate ($f_z$) was used, because contributions of both frontal ($f_x$) and transverse ($f_y$) components were much smaller than that of $f_z$ (Fig. 2). Oscillation frequency of the whole mechanical system was about 5 kHz, the time resolution of a servo-controlled system was 2 ms, and the controlled range of force was 150–3000 N.

2.4. Experiment 1: the “LL”–force relation

Maximum isometric force at different LL was measured. After the range of motion was determined by measuring the distance between a stopper of the machine and a fully extended leg position, maximum isometric force was measured at seven positions ranging between ~55 and <100% LL. Subjects first performed a few submaximum efforts to familiarize themselves with the measurements, and then exerted maximum voluntary isometric force as explosively as possible and then tried to keep the force plateau. During the measurements, subjects placed their arms in front of their chest. They were secured at pelvis and trunk by straps and at shoulders by adjustable pads, and instructed to perform the task without raising their hips from the seat. At each position, measurement was repeated three times with at

![Fig. 1. Schematic design of the servo-controlled dynamometer. The force command was generated by a personal computer and sent to a servo amplifier. The output from the force plate was introduced to the computer and fed back to the servo amplifier, which generated a signal for controlling the servomotor so as to cancel the difference between force command and actually measured force. The information about the position of force plate was obtained with a rotary encoder.](image1)

![Fig. 2. Vector components of force during the knee-hip extension. The $f_x$, $f_y$, and $f_z$ were a transverse, frontal and sagittal components of force, respectively. Force was servo-controlled to be 700 N ($f_z$) at 85% LL.](image2)
least 2-minutes rest period between bouts, and the largest value among the trials was chosen for the analysis.

2.5. Experiment 2: the force–velocity relation

During measurements of the dynamic knee–hip extension movements, subjects sat with their feet or foot on a force plate and their legs or leg flexed at 70% LL as a starting position. After a signal, subjects extended their legs or leg as fast as possible. Based on the LL–force relation determined in experiment 1, the force command for the range between 70% and 90% LL was applied so as to obtain the same force relative to isometric force regardless of LL (isotonic in terms of relative force). The relative force applied was varied from 10% to 80% of the maximum isometric force within 10% increments. Prior to measurements, the subjects performed 3–5 trials at a force level close to the body mass, which was about ~50% of maximum isometric force. The measurements were then made in a low-to-high force order, and finally, the measurements at low forces were repeated. The subjects performed three trials at each relative force and were allowed to rest for at least 2-minutes between bouts. The best trial of three measurements was chosen for the analysis.

Maximum isometric force ($F_0$), force ($F$) and velocity ($V$) were measured at 80% and 85% LL ($n = 17$). Within a range of force clamp, the most stable control of force was seen around 80% and 85% LL (see Section 3). Therefore, we measured values of $F_0$, $F$ and $V$ at this relative LL to determine the force–velocity relation. The angles of hip, knee, and ankle joints at 80 (80.0 ± 0.01)% and 85 (86.0 ± 0.01)% LL were also measured with a goniometer, and were $100.2 ± 4.38^\circ$, $106.3 ± 6.29^\circ$ and $82.5 ± 2.75^\circ$ at 80% LL, and $105.7 ± 3.63^\circ$, $114.0 ± 7.16^\circ$ and $88.2 ± 5.28^\circ$ at 85% LL, respectively.

In addition, the measurements of bilateral and unilateral movements were made and compared within the same individuals ($n = 10$). The reproducibility of the measurement was also examined by means of a test–retest procedure separated by a week ($n = 12$).

2.6. Data analysis

Data are presented as means ± SD. Least-squares regression analyses were made for the relations between force and velocity. Paired $t$-test was used to compare the extrapolated maximum isometric force ($F_{\text{max}}$) and $F_0$, and also compare either $F_{\text{max}}$ or unloaded velocity ($V_{\text{max}}$) obtained from bilateral and unilateral measurements. Reproducibility of $F_{\text{max}}$ and $V_{\text{max}}$ was evaluated using intraclass correlation coefficients (ICCs), and paired $t$-test was used to examine differences between test–retest data. The level of significant was set at $p < 0.05$.

3. Results

3.1. Relation between isometric force and “LL”

The relative LL–force relation in the knee–hip extension movement is shown in Fig. 3. The isometric force exhibited a peak when the foot position was at 80–90% LL. The isometric force progressively increased.

![Fig. 3. The relation between maximum isometric force and relative LL ($n = 10$). 100% LL means a full extension of knee joint in the seated position. Values of isometric force at each measured position were normalized in each subject with respect to the maximum value measured at an optimal leg length (usually 80–90% LL). Data represent means and SD. Thick straight lines show a theoretical function determined by a least-squares regression, which was then used for servo-controlling force (force command) in the dynamic measurements.](https://example.com/fig3.png)
with LL from the most flexed position (i.e., about 55% LL) up to this optimal position, and then declined rapidly. Based on this result, the maximum targeted force \((F_{t_{\text{max}}})\) was set at 85% LL where maximum isometric force was reached on an average, and the minimum targeted force \((F_{t_{\text{min}}})\) was started from 70% LL where mean \((F_{70\%}/F_{\text{max}})\) was 0.69. The following linear function (solid line in Fig. 3) was made for the force command so that the same relative force was consistently generated throughout the movement (range, 70–90% LL):

\[ F = F_{t_{\text{max}}} \left[ 0.69 + 0.31(\text{LL} - 70)/15 \right] (70\% \leq \text{LL} \leq 85), \]

\[ F = F_{t_{\text{max}}} (85 < \text{LL} \leq 90), \]

where \(F\) is applied force and LL is “leg length” (%).

3.2. The force and velocity records during force clamp

Typical records of force and velocity are shown in Fig. 4(a) as functions of relative LL. Actually measured force appeared to match with the force command (straight line) except for the early phase of servo control. Deflections by such an “impact artifact” or “overshoot” were larger at lower controlled force. The initial, oscillatory force response was likely due to both the large inertia and the time lag of servo control, and could not be completely eliminated. However, in most cases, both force and velocity approached to almost steady level near the end of movement (80–90% LL) even at the lowest level of controlled force.

The force–velocity relation appeared to be described well by a linear function, so that the \(F_{\text{max}}\) and \(V_{\text{max}}\) could be estimated by extrapolating the linear regressions onto the force and velocity axes, respectively as shown in Fig. 4 (the upper panel in b). The force–power relation was then determined by using a simple, quadratic function as shown in Fig. 4 (the bottom panel in b). The relation between force and velocity was described by the following equation:

\[ V = -aF + b, \]

where \(V\) is velocity and \(F\) is force, and \(a\) and \(b\) are constants. Power \((P)\) is obtained as the product of force applied and velocity, hence

\[ P = FV = -aF^2 + bF. \]

From this function, the maximum power \((P_{\text{max}})\) is directly obtained:

\[ P_{\text{max}} = b^2/4a. \]

The force at \(P_{\text{max}}\) is 0.5\(F_{\text{max}}\), where the velocity is also 0.5\(V_{\text{max}}\).
3.3. The force–velocity relation

The force–velocity relation determined at 80 (80.4±0.7) % and 85 (84.7±1.2) % LL are shown in Fig. 5(a) and (b), respectively. The relations between force and velocity were described well with linear functions and the correlation coefficients between force and velocity in each subject were \( r = -0.906 \) to \(-0.998\) for 80% LL, and \( r = -0.946\) to 0.998 for 85% LL. When data from all subjects were normalized with their respective \( F_{\text{max}} \) and \( V_{\text{max}} \), the force–velocity relation showed a better fit to a linear function \( (r = -0.986 \) for 80% LL, \( r = -0.968 \) for 85% LL) than to an exponential function \( (r = -0.950\) for 85% LL, \( r = -0.962\) for 85% LL) within a range of force between 0.1 and 0.8\( F_{\text{max}} \), although the differences were small. Also, \( F_{\text{max}} \) coincided with the maximum isometric force \( (F_{\text{0}}) \) at both 80% and 85% LL and the ratio between \( F_{\text{0}} \) and \( F_{\text{max}} \) (\( F_{\text{0}}/F_{\text{max}} \)) was \( 1.00 ± 0.09 \) at 80% LL and 1.00±0.20 at 85% LL.

When \( F_{\text{max}} \) and \( V_{\text{max}} \) obtained from the bilateral and unilateral measurements were compared within the same individuals as shown in Fig. 6, \( F_{\text{max}} \) in the unilateral measurements were significantly smaller than in the bilateral measurements (bilateral = 1686.97 ± 367.40 N vs. right unilateral = 1048.80 ± 206.58 N, left unilateral = 1028.02 ± 183.26 N), whereas \( V_{\text{max}} \) was substantially the same \( (p > 0.05) \) in both measurements (bilateral = 2.19 ± 0.33 m s\(^{-1}\) vs. right unilateral = 2.12 ± 0.27 m s\(^{-1}\), left unilateral = 2.09 ± 0.26 m s\(^{-1}\)). Also, right and left unilateral measurements resulted in the similar values for \( F_{\text{max}} \) and \( V_{\text{max}} \ (p > 0.05) \). When \( F_{\text{max}} \) in the bilateral measurements and in the sum of right and left unilateral measurements (2076.82 ± 383.80 N) was compared, \( F_{\text{max}} \) in the bilateral measurements was significantly smaller than in the sum of right and left unilateral measurements and the bilateral deficit was observed with \(-18.8\% \) of the mean magnitudes of the bilateral index (BI), which was calculated as: \( BI \% = 100\left[\frac{\text{bilateral}}{\text{right unilateral} + \text{left unilateral}}\right] - 100 \) (Howard and Enoka, 1991).

ICCs values between first and second measurements for \( F_{\text{max}} \) and \( V_{\text{max}} \) were 0.87 and 0.91, respectively, and no significant differences between test-retest data were seen.

4. Discussion

The newly developed servo-controlled dynamometer was able to evaluate muscular function during the knee–hip extension movement by using a force clamp method. The major finding in the present study was that, under the servo-controlled feedback system, the relation between force and velocity during the knee–hip extension movements showed a linear-like appearance rather than hyperbolic one typical to an isolated muscle (Hill, 1938) and a single-joint movement (Wilkie, 1950). In addition, the force–power relation was obtained as a simple, quadratic function and thus the peak of mechanical power appeared at 50% of the maximum isometric force, although it appears at 30–35% of the maximum isometric force in the hyperbolic force–velocity relation of the elbow flexion (Kaneko et al., 1983; Toji et al., 1997).

A computer-assisted servo system could effectively control force in human knee–hip extension movements. To attain relative isotonic condition, the LL–force relation during the knee–hip extension movements was initially determined and used to construct the force command in the servo feedback system. By keeping the relative force constant (force clamp) with this function, the effect of inertial mass was minimized and did not affect the later phase of muscle contraction. The servo
system thus enabled to obtain force under the steady state ranging from 0.1 to 0.8\(F_0\). This wide range of force makes it possible to evaluate appropriately human muscular function in vivo when compared with other existing methods, since it has been reported that isokinetic measurements are difficult to attain a high velocity close to \(V_{\text{max}}\) (Froese and Houston, 1985; James et al., 1994; Prietto and Catozzo, 1989; Perrine and Edgerton, 1978; Wickiewicz et al., 1984). Also, other studies in the knee–hip extension movements (Bosco et al., 1995; Macaluso and De Vito, 2003; Pearson et al., 2004; Rahmani et al., 2001) have measured force and velocity over limited range of velocity, so that the shape of force–velocity curve at high or low velocity regions has not been well understood. It has been reported that, in single fibers, maximum velocity estimated by extrapolation will be much closer to its true value when measurements will be made for force closer to zero (Julian et al., 1986). However, the force–velocity properties at the highest velocity (<0.1\(F_0\)) is still unclear, due to a limitation of the measurements in order to reach maximal velocity under the steady state in this study.

The linear appearance of the force–velocity relation of the knee–hip extension movement was consistent with other studies on half squat (Bosco et al., 1995; Rahmani et al., 2001), leg press (Macaluso and De Vito, 2003; Pearson et al., 2004), leg cycling (Driss et al., 1998, 2002; Vandewalle et al., 1987; Yamauchi et al., 2005), and arm cranking (Driss et al., 1998) exercises. Therefore, the maximum isometric force (\(F_{\text{max}}\)) and the maximum unloaded velocity (\(V_{\text{max}}\)) could be estimated by extrapolating the linear regressions. It showed that \(F_{\text{max}}\) coincided with the measured \(F_0\) and values of \(F_{\text{max}}\) and \(V_{\text{max}}\) for second measurements were well correlated with those for first measurement. The lower correlation for \(F_{\text{max}}\) than that for \(V_{\text{max}}\) is due possibly to day-to-day variations in strength (Driss et al., 2002). From physiological point of view, the intersections on the force and velocity axes in the force–velocity relation are related to the performance of contractile machinery, thus of special importance (Gülch, 1994). The intersection on the force axis corresponds to the measured maximum isometric force (\(F_0\)), and the intersection on the velocity axis may represent the maximum unloaded velocity. \(F_{\text{max}}\) and \(V_{\text{max}}\) were not related to each other so that the magnitude of \(F_{\text{max}}\) did not affect \(V_{\text{max}}\), indicating that \(V_{\text{max}}\) is independent of force and can be regarded as proper unloaded velocity in terms of how fast the feet or foot can move away from the body at the zero loads.

The linear force–velocity relation also affected the force–power relation so as to determine the optimum force and velocity of the maximum power output. The power is important in physical performance, such as sprinting and jumping. It has been reported that a maximum elbow flexion power is produced when the load is about one-third of the maximum isometric force and increased after the training with about one-third of the maximum isometric force (Kaneko et al., 1983; Toji et al., 1997). However, for knee–hip extension movements, our study and others (Macaluso and De Vito, 2003; Pearson et al., 2004) demonstrated that the maximum mechanical power was generated under \(~50\%\) of the maximum isometric force. Therefore, different exercise intensities may have to be used in single-joint and multi-joint movements for optimizing the development of power.

The mechanism underlying the linear appearance of the force–velocity relation in the multi-joint movements has been remained unclear. The force–velocity relation described by the Hill (1938) equation has been mathematically explained in terms of the sliding filament theory (Huxley, 1957; Huxley and Simmons, 1971). However, in single fibers, the measured maximum isometric force is lower than maximum isometric force extrapolated by the Hill equation (Edman et al., 1976), suggesting changes in the kinetics of cross-bridge cycle when load exceeds about 78% of the measured isometric force (Edman, 1988). It has been thought that the force–
velocity relations of human movements are influenced by a number of factors, including muscle fiber type and anatomical factors such as moment arm length, muscle length, and pennation angle of muscle fibers (Vandewalle et al., 1987; Wickiewicz et al., 1984). For multi-joint movements, contractions of many muscles are required at the same time. For example, during the knee–hip extension movement, many muscles are involved in either knee or hip extension simultaneously. Therefore, the coordination of contraction between each muscle is required to gain the quality of movement. Also, co-contraction of antagonist muscles cannot be ignored. It is known that antagonist co-contraction influences the voluntary activation of agonist muscle and that it may be important for the joint stabilization and for the prevention of over-acceleration (Aagaard et al., 2000; Isear et al., 1997; Marsden et al., 1983; Osternig et al., 1986; Sale, 1988). Furthermore, it has been suggested that, under conditions of high loading, the inhibitory mechanism may act as a protector of the musculoskeletal system from an injury when all available muscles become maximally activated (Westing et al., 1991).

An elongation of the series elasticity may influence the linear force–velocity relation, although it has been suggested that during dynamic movements, changes in the joint angle are proportional to changes in the muscle length and to changes in the length of muscle fibers (Perrine and Edgerton, 1978; Wickiewicz et al., 1984). It has been demonstrated that during isokinetic knee extension, the length of muscle fibers changes with angular velocity even at the same joint angle, suggesting that the length of the elastic component is changed due to the force on the muscle–tendon complex (Ichinose et al., 2000). During knee–hip extension, the strain of series elastic component may result in a shorter muscle length in knee extensor muscles under larger force, and thus may have an effect of moving the force–velocity relation on downward as the force is large.

The lower velocities at high velocity region may be caused by either short duration of the movement or the heterogeneity of muscle fiber type. In most measurements, the velocity under small force region reached a steady level near the end of the movement, so that the movement might be finished before the maximum activation level was attained. It has been suggested that the shortening velocity under a given load is affected by the activation level of whole muscle, thus the maximum unloaded velocity extrapolated from the Hill equation may be affected by the activation level (Chow and Darling, 1999; de Haan, 1998), although maximum unloaded velocity is independent of activation level in single fibers (Edman, 1979). In a whole muscle with a heterogeneous fiber composition, the maximum unloaded velocity measured by the slack test ($V_0$) is determined by the fastest fibers recruited, whereas the maximum unloaded velocity determined by hyperbolic regression ($V_{max}$) is affected by the slower fibers giving rise to increases in an internal load, so that $V_{max}$ is lower than $V_0$ (Claflin and Faulkner, 1985, 1989; Josephson and Edman, 1988). This is further supported by the dependence of maximum velocity on muscle fiber composition in humans. Individuals with a high percentage of slow twitch muscle fibers in vastus lateralis muscle have lower velocities in knee extension movement under isokinetic (MacIntosh et al., 1993; Suter et al., 1993; Thorstensson et al., 1976) and isotonic afterloaded (Tihanyi et al., 1982) conditions. This suggests that slow twitch fiber may interfere with the shortening of fast twitch fibers, although velocity of muscle contraction should be primarily determined by fast fibers because most muscle fibers are arranged mechanically in parallel.

It was found that muscular forces produced during maximum voluntary contractions of knee–hip extension ($F_{max}$) were smaller by ~20% in bilateral movement when compared to the sum of forces in unilateral movements. The bilateral deficit may be prominent in movements requiring coordination of a large number of muscles such as multi-joint movements (Häkkinen et al., 1995, 1997; Schantz et al., 1989), although others have shown that it also occurs in single joint movement (Koh et al., 1993). The mechanism responsible for bilateral deficit has been mostly studied in the area of neural motor control, and it has been thought that full motor unit activation is more difficult to achieve in bilateral contraction than in the unilateral contraction (Ohtsuki, 1983). Although the neural mechanisms of the bilateral deficit are beyond the scope of the present study, it remains unclear how the two major mechanical parameters, force and velocity are affected by bilateral and unilateral conditions. The same $V_{max}$ was obtained from bilateral and unilateral movements, although a significant bilateral deficit was shown in $F_{max}$. This indicates that steady-state velocity under light load is not so strongly affected by the neural activation level, as is the generation of large force.

In conclusion, the force–velocity relation of the knee–hip extension movement under a condition close to steady state was successfully determined by using the servo-controlled dynamometer, and showed a linear appearance. The present method is effective to study muscle function and neuromuscular control of human multi-joint movements without generation of large force to accelerate the inertial mass. Further investigations are required to understand the mechanism underlying the linear force–velocity relation of the knee–hip extension movements. In particular, it is interesting to see whether relative contributions of knee extensor muscles and hip extensor muscles are changed with the levels of force and velocity.
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