Non-invasive estimation of the joint torque of the vastus lateralis during pedaling based on tendon elongation

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Abstract

The purpose of this study was to estimate the knee extension torque generated by vastus lateralis (VL) during pedaling on the basis of VL tendon elongation (ΔLt). The fascicle length and pennation angle of VL during isometric knee extensions (0, 20, 60, 100 Nm; 30°, 45°, 60°, 75°, 90°, 105°) were measured using a B-mode ultrasonography (n=5). ΔLt was estimated from these measurements and the relations between ΔLt, knee joint angle and the knee extension torque generated by VL (TQ) were obtained. ΔLt of the same subjects during pedaling was taken from a previous study. TQ during pedaling was estimated on the basis of ΔLt during isometric knee extensions and pedaling (TQ synagogue). The peak value of TQ synagogue (19.8±11.2 Nm) during pedaling was significantly greater than that of TQ pedaling (9.3±4.0 Nm) (p<0.05), which might be related to the lower activation level and the different activation pattern of the rectus femoris compared with those of the VL. The pattern of TQ synagogue differed significantly from that of TQ pedaling, which might be related to the activation of the biceps femoris that increased in the latter half of the knee extension phase during pedaling. Since the methodology used in this study is non-invasive and measures the parameter that is closely correlated to the joint torque generated by a single muscle, it can be a useful tool for investigating the mechanics of human movements.

Key words: ultrasonography, muscle fascicle length, knee extension, inverse dynamics

1. Introduction

An accurate estimation of the joint torque generated by a single muscle is needed for the better understanding of the mechanics of human movements. Although the inverse dynamics method, by which net joint torque can be calculated from the combination of kinetic and kinematic data, has made an important contribution to the understanding of the mechanics of human movements, this method cannot evaluate the joint torque generated by synergist muscles and antagonist muscles separately. Therefore the inverse dynamics method can estimate the joint torque generated by synergist muscles properly only when agonist muscles do not generate torque, but synergist and agonist muscles generate torque simultaneously in most of human movements. To take an example, knee flexor and extensor muscles generated torque in the knee extension phase during pedaling, and consequently, the inverse dynamics method detected net knee flexor torque even in the knee extension phase (van Ingen Schenau et al. 1992).

Several methods for estimating the torque generated by a single muscle or synergist muscles have been proposed. Those methods used the surface electromyography (EMG) (Hof et al. 1987, Olney and Winter 1985), musculo-skeletal model (Bobbert et al. 1986, Crowninshield and Brand 1981, Hoy et al. 1990, Raikova and Prilutsky 2001) and intra-tendon optic fiber technique (Arndt et al. 1998, Finni et al. 1998). Although the EMG is useful to obtain the activation pattern of a single muscle, many factors affecting the relation between the EMG and the torque generated by a single muscle (for review see Dowling 1997) make it difficult to estimate the torque generated by a single muscle. A musculo-skeletal modeling method needs many parameters such as muscle fiber length, muscle fiber optimum length, pennation angle, tendon length, mechanical characteristics of muscle fibers and tendons, and
Fig. 1 A schematic illustration of the experimental setup. A synchroscope was used to develop the target knee extension torque. An electrical goniometer for measuring the knee joint angle was connected to a personal computer via an amplifier and an analog-digital (A/D) converter. A dynamometer for measuring the knee extension torque was connected to a synchroscope and to a personal computer via an A/D converter. An ultrasound apparatus and a personal computer were synchronized with a trigger.

Fig. 2 A longitudinal ultrasound image of the vastus lateralis muscle (VL). The directions of the superficial and deep aponeuroses and fascicle were extended by straight lines. Both parts around origin and end of the fascicle were extended separately because the fascicle was curved. The fascicle length (Lf) was measured from this estimated whole shape of the fascicle. The pennation angle (θ) was measured as the angle between the fascicle and deep aponeurosis.

hence difficulty resides in obtaining reliable values for these parameters. The optic fiber technique is attractive because high linearity was reported between optic fiber responses (light intensity modulation) and the force applied to tendon (Komi et al. 1996), though the optic fiber responses are highly influenced by skin movement (Erdemir et al. 2003). However, this technique is not completely non-invasive because an optic fiber must be passed through skin and tendon.

Human tendon elongation in vivo has been measured accurately and non-invasively using a real-time B-mode ultrasonography (Maganaris and Paul 2002, Muraoka et al. 2001, Muramatsu et al. 2001, Rosager et al. 2002). Moreover, tendon elongation is proportional to the force applied to tendon, though a history of tendon length influences the relation between tendon elongation and the force applied to tendon (Bennett et al. 1986, Ker 1981, Pollock and Shadwick 1994).

The purpose of this study was to estimate the joint torque generated by a single muscle during movements on the basis of tendon elongation. The muscle examined was vastus lateralis (VL). The relation between joint torque, tendon elongation and joint angle during isometric knee extensions was determined. On the basis of this relation and the reported values of the VL tendon elongation during pedaling (Muraoka et al. 2001), which was obtained for the same subjects in this study, the knee extension torque generated by VL (TQ) during pedaling was estimated. We estimated TQ on the basis of the net knee joint torque calculated by the inverse dynamics method. The inverse dynamics method has often been used to estimate joint torques during movements. The TQ estimated from those two methods were compared with each other.

2. Methods

2.1 Subjects

Five healthy males, the same subjects in the previous study (Muraoka et al. 2001), voluntarily participated in this study. The mean and standard deviation (±SD) for each of age, height, mass and thigh length were 24.6 (±0.5) yr, 171.4 (±5.5) cm, 66.7 (±6.3) kg and 40.0 (±1.6) cm, respectively. Thigh length was defined as the distance between the greater trochanter proximally and the lateral epicondyle of the femur distally. The subjects were fully informed about the procedures to be used in addition to the purpose of the study. Written informed consent was obtained from all subjects.

2.2 Isometric knee extension task

In the task of isometric knee extension, the sub-
jecot sat on the test bench of a dynamometer (MYORET RZ-450, Asics, Kobe, Japan) with hip joint angles of 80° (full extension=0°) (Fig. 1). Following a warm-up session with submaximal contractions, the subject developed submaximal knee extension torque at each of four levels (0, 20, 60, 100 Nm) isometrically for 2 s at each of six knee joint angles (30°, 45°, 60°, 75°, 90°, 105°; full extension=0°) with the help of visual feedback using a synchronoscope (model SS-5703, IWATSU, Tokyo, Japan). The order of testing was randomized for knee joint angles and knee extension torques. The torque was zeroed at each joint angle at rest. The knee joint angle was measured during contractions using an electrical goniometer (model 45313, San-ei, Tokyo, Japan) that was placed on the lateral aspect of the knee. The lever arm position of the dynamometer was changed so that the knee joint angle during contractions was equal to the target knee joint angle because knee joint angle changed even in an isometric condition due to the compliance of the dynamometer and human body. The test was repeated twice and the averaged value was used for the estimation of TQ.

2.3 Pedaling task

In the task of pedaling (Muraoka et al. 2001), the subject cycled at a pedaling rate of 40 rpm at a power output of 98 W without toe-straps on a cycle ergometer (model 814 E, MONARK, Varberg, Sweden). The saddle height was adjusted to the distance between the greater trochanter and the sole of the foot. The subject maintained a constant pedaling rate of 40 rpm with the help of a metronome (model SQ-77, SEIKO, Tokyo, Japan). This slow pedaling rate was adopted because of the limitation of the time resolution of a B-mode ultrasound apparatus (model SSD-2000, Aloka, Tokyo, Japan) used in this study. The ball of the foot was positioned at the center of the pedal. The pedaling task and the knee extension task were performed in a same day.

2.4 Ultrasonography

The longitudinal ultrasound images of VL were recorded in the isometric knee extension task using the same method in the pedaling task in the previous study (Muraoka et al. 2001). The ultrasound apparatus with an electronic linear array probe of 7.5 MHz was used to obtain longitudinal images of VL. The position of probe, which was firmly placed on
Non-invasive estimation of the joint torque of the vastus lateralis during pedaling based on tendon elongation

![Graph showing knee joint torque generated by VL (Nm) vs. Tendon elongation (mm)]

**Fig. 3** A typical example of the relation between the tendon elongation of vastus lateralis (ΔL\textsubscript{ISO}) and the knee joint torque generated by vastus lateralis (TQ), which was obtained during isometric knee extensions at each of four levels (net knee joint torque: 0, 20, 60, 100 Nm). Plots were curve fitted by TQ = a(\textsuperscript{\text{ΔL\textsubscript{ISO}} - b}) - a\textsuperscript{b}, where a and b are constants.

of determination (R\textsuperscript{2}) = 0.96 – 1.00]:

\[
L_f \times \cos \theta = A + B \times \delta + C \times \delta^2
\]

where \(\delta\) is knee joint angle (in degrees), and A, B and C are constants. The value of \(L_f \times \cos \theta\) over 105° (i.e. 105° ~ 114°) was extrapolated from this regression equation. At each knee joint angle where the VL tendon elongation during pedaling (ΔL\textsubscript{PEDAL}) was obtained in the pedaling task in the previous study (Muraoka et al. 2001), the VL tendon elongation during isometric knee extensions (ΔL\textsubscript{ISO}) at each of three contraction levels (20, 60, 100 Nm) was estimated by:

\[
\Delta L_{\text{ISO}} = L_f r - L_f c \times \cos \theta_c
\]

where \(L_f r\) and \(L_f c\) are the VL muscle fascicle lengths at rest and during contractions, and \(\theta_r\) and \(\theta_c\) are the pennation angles at rest and during contractions, respectively. Then the knee extension torque measured by the dynamometer in the isometric knee extension task was converted to TQ using the following equation (Ichinose et al. 1997, Narici et al. 1996):

\[
TQ = r\text{PCSA} \times TQ_{\text{quad}}
\]

where \(TQ_{\text{quad}}\) is the knee extension torque and \(r\text{PCSA}\) is the reported value of the relative contribution of VL to quadriceps femoris muscles in terms of physiological cross-sectional area (PCSA) (34% ; Akima et al. 1995). Then \(\Delta L_{\text{ISO}}\)-TQ relation was curve fitted at each knee joint angle where \(\Delta L_{\text{PEDAL}}\) was obtained [\(R^2=0.91–1.00\) ; Fig. 3]:

\[
TQ = a(\text{ΔL\textsubscript{ISO}} - b) - a^b
\]

where a and b are constants. The knee extension torque generated by VL during pedaling was estimated from the above equation by replacing \(\Delta L_{\text{ISO}}\) with \(\Delta L_{\text{PEDAL}}\). When \(\Delta L_{\text{PEDAL}}\) was a negative value, TQ was estimated to be zero because \(\Delta L_{\text{PEDAL}}\) would changed in an irregular way within the range of the estimation error of \(\Delta L_{\text{PEDAL}}\) when the true value of \(\Delta L_{\text{PEDAL}}\) was zero.

2.6 Estimation of net joint torque and the knee extension torque generated by VL

In the pedaling task (Muraoka et al. 2001), the subject was videotaped from the right side using a high-speed camera (MEMRECAM c 2 s, nac, Japan) at a sampling frequency of 200 Hz. The camera was...
located perpendicular to the sagittal plane of the subject at a distance of 3.5 m. Markers were applied to the skin overlying the greater trochanter, the approximate center of rotation of knee joint, the lateral malleolus, the fifth metatarsophalangeal, the center of the pedal and front and back edge of the pedal. Simultaneously, vertical and horizontal components of the pedal force were measured using the two extensometric gauges (strain gauges, KYOWA, Japan) attached on the pedal shaft (Alvarez and Vinyolas 1996). These kinetic data were stored on a personal computer through A/D converter (MacLab/8 s, ADInstruments, Australia) at a sampling frequency of 1 kHz. The video images were analyzed using a motion analyzer (Frame-DIAS, DKH, Japan). The coordinates of the markers were filtered with a Butterworth fourth-order low-pass filter at a cut-off frequency of 15 Hz, then the position for three body segments (feet, shank and thigh) and the pedal, knee and ankle joint angles were determined. On the basis of these kinematic and kinetic data, net knee joint torque was calculated using a link-segment model with the inverse solution (Winter 1990). Then the knee joint torque generated by VL was estimated from net knee joint torque by:

\[ TQ = rPCSAXnTQ \]

where \( nTQ \) is net knee joint torque and \( rPCSAX \) is the reported value of the relative contribution of VL to quadriceps femoris muscles in terms of PCSA (34\% ; Akima et al. 1995).

2.7 Reproducibility

The reproducibility of two trials of \( Lf \) and \( \theta \) measurement was evaluated on the basis of a coefficient of variation (CV). The difference between the peak value of the knee extension torque during pedaling estimated from VL tendon elongation and from the inverse dynamics was tested using paired t-test. Statistical difference was set at a level of \( p < 0.05 \).

3. Results

The CV values of \( Lf \) and \( \theta \) between two trials were 2.0\% and 3.9\% on average, respectively. These values were within the range of previously reported values (Fukunaga et al. 1997, Ichinose et al. 1997, Narici et al. 1996, Maganaris et al. 1998).

The constant \( a \) and \( b \) in the regression curve for the \( \Delta L_{ISO} - TQ \) relation were 1.2±0.1 and –15.8±17.6, respectively. The elongation of VL was mostly limited to the knee extension phase (Fig. 4 B). The \( \Delta L_{PEDAL} \) was less than \( \Delta L_{ISO} \) obtained during isometric knee extension at the level of 60 Nm (Fig. 4). Consequently, the TQ that was estimated from \( \Delta L_{PEDAL} \) (\( TQ_{a} \)) was less than 20.4 (60\times rPCSAX) Nm (Fig. 5). The \( \Delta L_{PEDAL} \) was almost zero in the knee flexion phase (Fig. 4 A). Both \( TQ_{a} \) and the TQ that was estimated from the inverse dynamics (\( TQ_{m} \)) reached each peak value at the middle of the knee extension phase (Fig. 5). The peak value of \( TQ_{a} \) (19.8±11.2 Nm) was significantly greater (\( p = 0.045 \)) than that of \( TQ_{m} \) (9.3±4.0 Nm). Then \( TQ_{a} \) dropped to negative values, though \( TQ_{a} \) gradually decreased to zero just after the end of the knee extension phase (Fig. 5).

4. Discussion

The knee extension torque generated by VL dur-
Non-invasive estimation of the joint torque of the vastus lateralis during pedaling based on tendon elongation

Fig. 5 The knee extension torque generated by VL estimated from VL tendon elongation (solid thick line) and the inverse dynamics (dashed thick line). The crank angle of 0° was the top dead center. Values are means for 5 subjects. Dark areas represent SD.

Fig. 6 A schematic illustration of the relation between the tendon elongation and the knee extension torque while tendon length is increasing (solid thick line), decreasing (solid thin line) and constant at a certain length (closed circles). When tendon elongation is ΔLt, the knee extension torque obtained while tendon length is increasing (TQm) is greater than that obtained while tendon length is maintained at ΔLt (TQm), and TQm is greater than the knee extension torque obtained while tendon length is decreasing (TQm).

Muramatsu et al. (2001) reported that significant elongation of the Achilles tendon and the aponeurosis of medial gastrocnemius could not be detected using ultrasonography above 30% and 60% of maximal isometric voluntary plantarflexion, respectively. The methodology presented in this study therefore could be applied only to low intensity movements.

Thirdly, the different activation pattern and level of rectus femoris compared with those of vasti muscles during pedaling (van Ingen Schenau et al. 1992, 1995) might affect the ΔLt-TQ relation because patellar tendon is the common tendon of quadriceps femoris muscles. However, the relative contribution of rectus femoris to quadriceps femoris muscles in terms of PCSA was 15% (Akima et al. 1995) and maximum patellar tendon elongation was 3.8 mm (Reeves et al. 2003), indicating that the differences in muscle force development level between rectus femoris and vasti muscles might change ΔLt by ~0.6 mm, which was practically negligible for the purpose of this study.

In the present study, the mean CV values of Lf and θ between two trials were 2.0% and 3.9%, respectively. The error of 2.0% for Lf and 3.9% for θ result in the overestimation of 1.8 mm and underesti-
mation of 1.9 mm for ΔLt, which means that the peak value of TQ could be overestimated by 55% and underestimated by 37%. These values were larger than the estimation error of the optic fiber method (14~48% ; Erdemir et al. 2003). It is suggested that Lf and θ should be measured more accurately to estimate the joint torque generated by a single muscle, to which repeated measurements of Lf and θ may contribute. The estimation method presented in this study might be applicable only when averaged values for subjects and/or trials were used (e.g. when investigating the mechanics of a human single muscle during movements).

The patterns and peak values of TQact and TQcon were different from each other (Fig. 5) because the inverse dynamics provided joint net torques generated by all muscles crossing a joint. The difference in peak value between TQact and TQcon during pedaling may be related to the lower activation level and the different activation pattern of rectus femoris compared with those of VL (Ericson et al. 1985, Marsh and Martin 1995, Neptune et al. 1997). The lower activation level and the different activation pattern of rectus femoris during pedaling was considered related to the different roles for a mono-articular muscle of VL and a bi-articular muscle of rectus femoris (van Ingen Schenau et al. 1992, 1995), and would result in the decrease of the peak value of net knee joint torque during pedaling. The difference in pattern between TQact and TQcon during pedaling may be related to the activation of biceps femoris. The activation level of biceps femoris increased in the latter half of the knee extension phase (Jorge and Hull 1986, Neptune et al. 1997), which resulted in an increase of knee flexion torque. Consequently, TQcon could not reflect TQ properly and the pattern of TQcon differed significantly from that of TQact (Fig. 5).

In summary, the present study shows that the joint torque generated by a single muscle in vivo during movements such as pedaling can be estimated non-invasively on the basis of tendon elongation measured using real-time B-mode ultrasonography, though the estimated torque is highly influenced by the error in the measurement of fascicle length and pennation angle. Since this methodology is non-invasive and measures the parameter that is closely correlated to the joint torque generated by a single muscle, it can be a useful tool for investigating the mechanics of human movements.

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