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Publisher: Routledge
Informa Ltd Registered in England and Wales Registered Number: 1072954
Registered office: Mortimer House, 37-41 Mortimer Street, London W1T 3JH, UK

Journal of Sports Sciences

Publication details, including instructions for authors and subscription information:

<http://www.informaworld.com/smpp/title~content=t713721847>

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Online Publication Date: 01 December 1996

To cite this Article: Sørensen, Henrik, Zacho, Morten, Simonsen, Erik B., Dyhre-Poulsen, Poul and Klausen, Klaus (1996) 'Dynamics of the martial arts high front kick', Journal of Sports Sciences, 14:6, 483 — 495

To link to this article: DOI: 10.1080/02640419608727735

URL: <http://dx.doi.org/10.1080/02640419608727735>

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Dynamics of the martial arts high front kick

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Accepted 19 April 1996

Fast unloaded movements (i.e. striking, throwing and kicking) are typically performed in a proximo-distal sequence, where initially high proximal segments accelerate while distal segments lag behind, after which proximal segments decelerate while distal segments accelerate. The aims of this study were to examine whether proximal segment deceleration is performed actively by antagonist muscles or is a passive consequence of distal segment movement, and whether distal segment acceleration is enhanced by proximal segment deceleration. Seventeen skilled taekwon-do practitioners were filmed using a high-speed camera while performing a high front kick. During kicking, EMG recordings were obtained from five major lower extremity muscles. Based on the kinematic data, inverse dynamics computations were performed yielding muscle moments and motion-dependent moments. The results indicated that thigh deceleration was caused by motion-dependent moments arising from lower leg motion and not by active deceleration. This was supported by the EMG recordings. Lower leg acceleration was caused partly by a knee extensor muscle moment and partly by a motion-dependent moment arising from thigh angular velocity. Thus, lower leg acceleration was not enhanced by thigh deceleration. On the contrary, thigh deceleration, although not desirable, is unavoidable because of lower leg acceleration.

Keywords: Electromyography, inverse dynamics, kicking, martial arts, proximo-distal.

Introduction

It is generally accepted that fast unloaded movements (i.e. striking, throwing and kicking), in which the aim is maximal linear speed of the involved segmental chain's most distal end (i.e. hand or foot), are performed in a so-called 'proximo-distal sequential' order (e.g. Müller, 1982; Jöris *et al.*, 1985; Herring and Chapman, 1992; Putnam, 1993). The term 'proximo-distal sequential' refers to the order in which the different segments or joints start their movement towards the target. In kicking, for instance, it is observed that the total movement starts with forward angular acceleration of the thigh while the lower leg lags behind. Then the thigh decelerates while simultaneously the lower leg accelerates (Putnam, 1983, 1993). This raises two questions. First, is the thigh actively decelerated by the glutei or ham-

string muscles, or passively decelerated by joint reaction forces from the accelerating lower leg? Secondly, is acceleration of the lower leg enhanced by the thigh's deceleration? These questions can be combined into one: Does kicking performance benefit from active thigh deceleration?

Simultaneous thigh deceleration and lower leg acceleration often lead to a situation in which the thigh is completely decelerated when the lower leg reaches its maximal angular velocity. Considering that the resultant linear velocity of the foot relative to the ground equals the vector sum of the resultant linear velocities of the knee relative to the ground and the foot relative to the knee (an application of the optimization principle of Hochmuth, 1967), it would appear disadvantageous from a kinematic point of view to decelerate the knee to zero velocity.

However, from a kinetic point of view, it can be argued that thigh deceleration enhances lower leg acceleration to a degree where loss of knee velocity is

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more than accounted for by the gain in foot velocity. This was indirectly supported by a simulation study of the overarm throw (Herring and Chapman, 1992), in which it was found that maximum throwing distance was enhanced if the upper arm and the forearm were actively decelerated in a proximo-distal sequence late in the throw. As the active deceleration was simulated by changing the joint torques in the shoulder and elbow from extension to flexion, Herring and Chapman (1992) termed this phenomenon 'torque reversal'. Jöris *et al.* (1985) used the term 'a whiplash-like action', which pinpoints the kinetics supposed to take place: when the proximal end of a distal segment is accelerated in one (non-axial) direction, the resulting angular movement causes the distal end to accelerate in the opposite direction. This is similar to the powerful jerk performed to make a whip crack.

In conflict with the whiplash principle is the theory that states that thigh deceleration is caused by lower leg motion. Jöris *et al.* (1985) found it most likely that proximal segment deceleration could be explained by Newton's third law, which states that a more proximal segment's action on a more distal segment will cause an equal but opposite directed reaction on the more proximal segment.

In an attempt to clarify these contrasting theories, we examined the high front kick used in most oriental martial arts. This is a highly proximo-distal sequential movement in which knee resultant linear velocity is almost decelerated to zero at the time of maximal foot linear resultant velocity (Fig. 1). To our knowledge, this type of kick has not been analysed hitherto. Previous work has dealt with soccer or punt-style kicking (e.g. Putnam, 1983; Robertson and Mosher, 1985; Dunn and Putnam, 1988). In these types of kicks, the target is typically situated at ground level or slightly above, whereas the high front kick is aimed at chin level. Possible differences also include kinematic peak values. In the present study, the averaged peak knee joint angular velocity was about 26 rad s^{-1} , while Robertson and Mosher (1985) reported approximately 19 rad s^{-1} for soccer kicking.

Similar to previous studies of kicking, the present study used kinematic measurements to derive the kinetics through inverse dynamics. Furthermore, in addition to methods used in previous kicking studies, we recorded the electrical activity of selected muscles in order to assess their temporal activation during the kick. Active deceleration of the thigh is likely to be performed by the gluteus maximus or the hamstring muscles. It was therefore of particular interest to determine the exact temporal activity of these muscles during the movement.

In summary, the aim of this study was to establish the kinetics of a typically proximo-distal sequential

Figure 1 Linear resultant velocity of the foot and knee. When the knee decelerates, the foot accelerates rapidly. Data from a pilot investigation preceding the present study.

movement with special emphasis on the possible use of active deceleration of proximal segments to enhance acceleration of distal segments.

Materials and methods

Subjects

Seventeen skilled taekwon-do practitioners served as subjects for this study (see Table 1). Their skills ranged from club level to top European level. All the subjects provided informed consent to participate and the study was approved by the local ethics committee.

The kick

Execution of the basic front kick analysed in the present study starts from a normal standing position with both feet on the floor. The kick is initiated by hip flexion of the kicking leg. While the thigh is moving upward, the knee flexes. When the thigh is oriented approximately horizontal but still moving upwards, the knee is flexed approximately 110° . From here the knee starts to extend. While the angular velocity of the lower leg is increasing, the angular velocity of the thigh decreases. When the lower leg angular velocity is maximal, the thigh is oriented approximately 40° above horizontal and its angular velocity is close to zero (a

Table 1 The subjects' physical characteristics and training experience ($n = 17$). Values shown are the mean and range

	<i>n</i>	Age (years)	Body mass (kg)	Height (cm)	Training	
					Years	Sessions per week
Males	13	24.1 (18–34)	70.4 (59.0–79.4)	178 (167–189)	6.3 (4–10)	2.8 (2–4)
Females	4	21.0 (18–26)	52.9 (46.4–64.0)	160 (155–164)	4.0 (3–6)	2.3 (2–3)
Total	17	23.4 (18–34)	66.3 (46.4–79.4)	173 (155–189)	5.8 (3–10)	2.6 (2–4)

stick figure sequence of the kicking leg is shown in Fig. 3). The ankle joint of the kicking leg is kept fully plantarflexed and the toes fully dorsiflexed in order to hit the target with the ball of the foot. During the kick, the supporting leg remains almost stationary on the floor. The kick is performed without body rotation, that is in a single, sagittal plane (contrary to most other taekwon-do kicking techniques), which makes it suited for two-dimensional analysis. Each subject performed three high front kicks aimed at a tennis ball suspended from the ceiling and adjusted to chin level. The fastest kick from each subject was selected for further analysis.

Cinematography

The subjects were filmed from their right side while they kicked in a sagittal plane with a 16 mm high-speed camera (Teledyne DBM 45) operating at a picture rate of 200 frames per second (Hz). A high-speed camera was used because the kicking motion is so fast that a detailed registration cannot be obtained by conventional video operating at 50 Hz (PAL). To check the filming rate, the camera was customized with an electronic outlet emitting a square pulse signal for each exposure. This signal was sampled by a DT-2801-A analog-to-digital converter (Data Translation, Inc., Marlboro, MA) operating at 1000 Hz and fed to an IBM PC DX2 clone. Subsequent inspection of this signal verified the accuracy of the high-speed camera to be exactly 200 Hz.

The camera was placed approximately 7 m from the subject. Markers were placed on selected anatomical landmarks in order to identify body segments and joints. These landmarks were the highest point on the iliac crest, the greater trochanter, the lateral epicondyle of the femur, the lateral malleolus and the fifth metatarsal joint. Special attention was given to the hip marker (greater trochanter) because some rotation at the hip was unavoidable. A broad marker was used so the visual midpoint of the marker seen from the camera position would always be as close as possible to the hip joint centre.

Position coordinates of the optical centres of the markers were automatically digitized (Peak Perform-

ance Technologies). As this equipment is video-based, it was necessary to transfer the developed 16 mm film to video. This was accomplished by a film projector with a built-in video recorder (Elmo TRV-16G| CCD). The Elmo projects exactly 25 film frames per second. The built-in video recorder transfers each film frame to an odd and an even field. Accordingly, we set the Peak Performance software to skip every other field.

Displacement data were filtered using a digital fourth-order Butterworth low-pass filter with zero degree phase lag (Winter, 1990). Optimal cut-off frequencies (6–10 Hz) were determined by use of residual analysis (Winter, 1990) and the Jackson Knee method (Jackson, 1979). From the filtered displacement data, angular velocities and accelerations of the thigh and lower leg, as well as linear velocities and accelerations of the hip, knee and ankle joints, were derived by finite difference computation.

Electromyography

During kicking, electromyographic (EMG) recordings were obtained from five lower extremity muscles: gluteus maximus, vastus lateralis, rectus femoris, biceps femoris caput longum and gastrocnemius caput mediale. The recordings were performed with Ag/AgCl surface electrodes (pick-up area 25 mm²) placed on the middle of the muscle belly approximately 2 cm apart (bipolar recording). Before mounting the electrodes, the skin was shaved, lightly abraded with emery-cloth and cleaned with pure alcohol. The electrodes were mounted with lightweight connectors and the signals were conducted directly to small custom-built pre-amplifiers taped to the skin. The EMG signals were led through 5 m long shielded wires to custom-built amplifiers with a frequency response of 20 Hz to 10 kHz. The wire set-up permitted the subjects to kick unrestricted. The pre-amplifiers lowered the impedance, which effectively reduced movement artefacts. Neither passive movement of the leg nor tapping it produced any visible artefacts.

The signals were sampled by a DT-2801-A analog-to-digital converter (Data Translation, Inc., Marlboro, MA) operating at 1000 Hz and fed to an IBM PC DX2 clone. Treatment of the digital data comprised several

steps. First, any remaining DC-components were removed by digital detrending. The signals were then high-pass filtered at 20 Hz (Butterworth fourth-order zero-lag digital filter) to remove any remaining movement artefacts. The next step, a digital full-wave rectification, prepared the signals for the final step, consisting of a 10 Hz low-pass filtering (Butterworth fourth-order zero-lag digital filter) yielding a linear envelope signal (Winter, 1990).

The EMG data were not normalized, as they were merely intended to provide information about the temporal aspects of muscle activation. However, after mounting the electrodes, the subjects were asked to forcefully contract the different muscles, thus allowing visual confirmation that specific muscle activation resulted in a recordable signal. This allowed us, during the following analysis, to assume that a 'silent' signal indicated an inactive muscle. No cross-talk was observed between EMG channels. This was verified by simultaneously monitoring the five EMG channels while the subject contracted each muscle in turn. Synchronization between the film and EMG data was provided by simultaneously triggering the analog-to-digital converter and flashing a light in the field of view.

Modelling

To facilitate inverse dynamics computations, the kicking leg was modelled as a two linked-segment model with the lower leg and foot treated as one segment. The assumptions typically associated with link-segment models (e.g. rigid segments and hinge joints) previously defined by, for instance, Zajac and Gordon (1989), Winter (1990) and Nigg and Herzog (1994), were not considered to impose serious limitations on the model. Furthermore, the combination of the lower leg and the foot as a single segment, simplifying the modelling and computations, has previously been applied to models used for kicking studies (Philips *et al.*, 1983; Dunn and Putnam, 1988; Putnam, 1993). In studies where three-segment models have been used (i.e. lower leg and foot treated as two individual segments), ankle moment of force was found to be negligible with respect to knee and hip moments (Zernicke and Roberts, 1976; Robertson and Mosher, 1985). In a simulation study of punt kicking using a three-segment model, Marshall *et al.* (1985) found that changes in ankle moment of force influenced movement of the foot segment, but not the movement of the other two proximal segments. This can be explained by the foot's smaller moment of inertia about the ankle joint when compared with the lower leg's moment of inertia about the ankle joint. As the high front kick movement is

determined by hip and knee joint kinematics, and since these have been reported not to be influenced by the ankle moment (Marshall *et al.*, 1985), we considered our two-segment model adequate.

To explain the kinetics of the kicking motion, we used the equations developed by Putnam (1983). These equations are particularly suited for the analysis of fast unloaded movements, as described in the following.

The motion of a segment rotating about its proximal end can be described by the Newton/Euler equation:

$$I\alpha = \Sigma M_i \quad (1)$$

where I is the segment's moment of inertia about its proximal end, α is its angular acceleration and ΣM_i is the summation of all moments of force acting upon the segment. Each moment is generated by a force (F) and a moment arm (s):

$$M = Fs \quad (2)$$

The forces include muscle forces, gravity and reaction forces from the environment. For an individual segment in a segmental chain, some or all of this environment consists of the adjacent segments, in which case the forces are referred to as joint reaction forces. The magnitude of these forces is in part determined by the kinematics of the involved segments. Accordingly, joint reaction forces increase with increasing speed of movement. The moment arms vary with the segment's absolute orientation in space and mutual position. This is apparent from the free body diagram shown in Fig. 2, where all the forces and moments causing acceleration of the lower leg are illustrated. The following nomenclature applies to Fig. 2 as well as to the remainder of this paper:

Nomenclature

θ	segment or joint angle
ω	segment angular velocity
α	segment angular acceleration
a	joint linear acceleration
g	gravitational acceleration
I	moment of inertia about segment centre of mass
m	segment mass
M	muscular moment
r	length between segment proximal end and centre of mass
s	segment length
b	moment arm of centripetal force
c	moment arm of tangential force

Indices

H	hip
K	knee

Figure 2 Free body diagram of the thigh and lower leg with forces and moments acting on the lower leg. Vector terms are accelerations which should be multiplied by the mass of the lower leg to yield forces. When the thigh rotates, the knee follows the dashed trajectory. Knee flexion angle θ_K equals $\theta_L - \theta_T$. Moment arm b equals $r_L \sin \theta_K$. Moment arm c equals $r_L \cos \theta_K$. See text for nomenclature and explanation.

L lower leg
T thigh

Based on Fig.2, equation (1) can now be thoroughly detailed (equation 3). To simplify subsequent reference to equation (3), each term in the equation is associated with a shorthand notation shown in parentheses to the left of the equation.

$$\begin{aligned}
 (I\alpha) & (I_L + m_L r_L^2) \alpha_L = \\
 (M_K) & M_K \\
 (g) & -m_L \cdot g \cdot r_L \cos \theta_L \\
 (\omega_T) & -m_L \cdot \omega_T^2 s_T \cdot r_L \sin \theta_K \\
 (\alpha_T) & -m_L \cdot \alpha_T s_T \cdot r_L \cos \theta_K \\
 (a_{Hx}) & +m_L \cdot a_{Hx} \cdot r_L \sin \theta_L \\
 (a_{Hy}) & -m_L \cdot a_{Hy} \cdot r_L \cos \theta_L
 \end{aligned} \quad (3)$$

The left-hand side of the equation is the moment of inertia about the proximal end ($I_L + m_L r_L^2$) multiplied by the observed angular acceleration of the lower leg (α_L). The moment of inertia about the proximal end is calculated using the parallel-axis theorem as the moment of inertia about the centre of mass (I_L) plus segment mass (m_L) multiplied by the square of the distance from the proximal end to the centre of mass (r_L^2). All values necessary to calculate this left-hand side of the equation are known. Values for I_L , m_L and r_L were obtained from the anthropometric data of Winter (1990), while α_L was derived from the cinematographic measurements. The left-hand side of the equation expresses the result of all moments acting on the lower

leg and is, accordingly, referred to as the resultant moment. The shorthand notation $I\alpha$ indicates that the result of the summed moments' action is that the lower leg with moment of inertia I is given the angular acceleration α .

The right-hand side of the equation is the summation of moments (i.e. ΣM_i in equation 1) acting on the lower leg, hence causing the observed angular acceleration. The resulting moment is composed of six components. These components are positive or negative depending on whether they cause counterclockwise or clockwise angular acceleration (counterclockwise direction is by definition positive).

The first component (M_K) is the muscular knee joint moment, or knee moment for short. This can be either a flexor or extensor moment, depending on whether the knee flexor or extensor muscles are dominating. For example, even though the extensor muscles are dominating, the knee might be flexing, in which case the extensor muscles perform eccentric contraction. The knee moment is the only unknown in equation (1). The approach of inverse dynamics is to measure the kinematics and then solve the equation of motion (equation 1) for this remaining unknown.

The remaining five moment components (g , ω_T , α_T , a_{Hx} and a_{Hy}) are all expressed according to equation (2) as a force (F) multiplied by a moment arm (s), and the force itself as a mass (m) multiplied by an acceleration (a):

$$M = F \cdot s = m \cdot a \cdot s \quad (4)$$

We have tried to emphasize this in the layout of equation (3) by using multiplication dots to separate each expression into parts of mass, acceleration and moment arm. In the expression of the second moment component ($m_L \cdot g \cdot r_L \cos \theta_L$), for instance, the mass is that of the lower leg (m_L), the acceleration is due to gravity (g) and the moment arm is the horizontal distance from the lower leg's centre of mass to a vertical line through the knee joint ($r_L \cos \theta_L$).

The elegance of Putnam's (1983) equations is based on her choice of generalized coordinates describing the movement and in the decomposition of the resulting moment. Using segment angles as coordinates instead of joint angles, for example, makes it easier to interpret the results, considering that the purpose is to describe the forces and moments which accelerate/decelerate the thigh and lower leg. The decomposition splits the resulting moment into six components (equation 3). Of these, one is the knee moment (M_K) and one is due to gravity (g), as previously described. The remaining four components are due to the movement of adjacent segments. Each of these four motion-dependent moments is a function of only one kinematic parameter; for example, the angular velocity of the thigh (ω_T) or the

linear horizontal acceleration of the hip (a_{Hx}). Thus, for example, this decomposition enables isolation of the influence of thigh angular velocity on lower leg angular acceleration. Each shorthand notation to the left of equation (3) focuses on the key parameter of the respective component.

The first motion-dependent moment is caused by the angular velocity of the thigh, as its shorthand notation (ω_T) implies. This angular velocity causes the proximal end of the lower leg to follow a circular path (dashed curve in Fig. 2). The centripetal acceleration of this circular movement is $\omega_T^2 s_T$ and, consequently, the centripetal force acting on the proximal end of the lower leg is $m_L \cdot \omega_T^2 s_T$. This force acts with a moment arm $r_L \sin \theta_K$ (dashed line b in Fig. 2) to create a moment $m_L \cdot \omega_T^2 s_T \cdot r_L \sin \theta_K$ causing a positive angular acceleration of the lower leg. When the knee is fully extended ($\theta_K = 0$), the moment arm b and thus the moment is zero, while the longest moment arm is obtained when the knee is flexed 90° . In the motion equation for the lower leg, this moment component is negative, although it causes counterclockwise acceleration. The reason is that the knee flexion angle (θ_K) is computed as $\theta_L - \theta_T$. Consequently, the term $\sin \theta_K$ and thereby the full moment component is negative (unless the knee is hyperextended).

The second motion-dependent moment is caused by the angular acceleration of the thigh. The linear acceleration of the proximal end of the lower leg due to the thigh's angular acceleration is $\alpha_T s_T$ and, consequently, a force $m_L \cdot \alpha_T s_T$ acts here tangentially to the circular path mentioned above. Contrary to the centripetal force, which always acts in a direction towards the centre of the circular movement, the direction of this tangential force depends on the direction of the thigh's angular acceleration. The force acts with a moment arm $r_L \cos \theta_K$ (dashed line c in Fig. 2) to create a moment $m_L \cdot \alpha_T s_T \cdot r_L \cos \theta_K$, which causes angular acceleration of the lower leg. The sign of the moment arm expression depends on whether the knee is more or less than 90° flexed. Thus, whether this moment causes positive or negative angular acceleration of the lower leg depends on the direction of the thigh's angular acceleration as well as on the knee joint angle. A possible whiplash action requires that the thigh is angularly accelerating in a negative direction (i.e. actively decelerated) and that the knee is less than 90° flexed. As the length of the moment arm c decreases when the knee angle approaches 90° , the accelerating effect from thigh deceleration on the leg would be most pronounced when the knee is close to full extension.

The remaining two motion-dependent moments are caused by linear acceleration of the hip. They turned out to be negligible in the high front kick in which the hip is kept almost stationary.

The equation of motion for the thigh is:

$$\begin{aligned} (I\alpha) & (I_T + m_T r_T^2) \alpha_T = \\ (M_H) & M_H \\ (g) & -m_T g r_T \cos \theta_T + m_L g s_T \cos \theta_T + m_L g r_L \cos \theta_L \\ (\omega_T) & -m_L \omega_T^2 s_T r_L \sin \theta_K \\ (\alpha_T) & -m_L \alpha_T s_T r_L \cos \theta_K + m_L \alpha_T s_T^2 \\ (\omega_L) & + m_L \omega_L^2 s_T r_L \sin \theta_K \\ (\alpha_L) & + m_L \alpha_L s_T r_L \cos \theta_K + m_L \alpha_L r_L^2 + I_L \alpha_L \\ (\alpha_{Hx}) & + m_T a_{Hx} r_T \sin \theta_T + m_L a_{Hx} s_T \sin \theta_T + \\ & m_L a_{Hx} r_L \sin \theta_L \\ (\alpha_{Hy}) & - m_T a_{Hy} r_T \cos \theta_T + m_L a_{Hy} s_T \cos \theta_T + \\ & m_L a_{Hy} r_L \cos \theta_L \end{aligned} \quad (5)$$

Components need to be added for each new segment in the segmental chain, but the overall composition (and nomenclature) is similar for the thigh and lower leg equations. Detailed elaboration of these equations can be found in Putnam (1991). This was extended from two to three segments by Hoy and Zernicke (1986) and to three-dimensional motion for two segments by Feltner and Dapena (1989).

Applying equation (3) (lower leg) or equation (5) (thigh) to the measured kinematic data yields data for each moment component as a function of time. The impulse-momentum equation

$$\int_{t_1}^{t_2} M dt = I \omega_{t_2} - I \omega_{t_1} \quad (6)$$

states that integration of a moment over a period of time yields a change in angular momentum ($I\omega$). Assuming constant moment of inertia (I), this is similar to the change in angular velocity. Accordingly, time integration of each moment component reveals the individual component's influence on the segments' change in angular velocity.

Results

Kinetics

In Fig. 3, the time-dependent moments acting on the thigh and lower leg during the kicking motion from one representative subject are shown together with a stick diagram of the kicking leg and the rectified EMG recordings. The resulting moment as well as each right-hand component from the equations of motion are represented by a separate curve labelled according to the shorthand notation listed earlier. Horizontal and vertical linear acceleration of the hip (a_{Hx} and a_{Hy}) turned out to play a negligible role and were therefore combined into a single component, a_H . Positive moments accelerate the respective segments in a counterclockwise direction, whereas negative moments accelerate the segments in a clockwise direction (i.e. decelerate the segments).

In the following discussion, we focus on the time period between the start of movement and the instant of maximal lower leg angular velocity. 'Start of movement' was defined as the instant when either of the resulting moment curves ($I\alpha$) deviated from zero, indicated by the leftmost dashed line labelled t_1 in Fig. 3. The instant of maximal lower leg angular velocity was defined as the instant when the lower leg resulting moment curve ($I\alpha$) crossed the zero-line after its positive phase, indicated by the rightmost dashed line labelled t_2 . The time period t_1 to t_2 is hereafter referred to as the analysed time interval.

The thigh moment curves show that the resulting moment ($I\alpha$) was positive (the thigh was accelerating) during the first approximately 80% of the analysed time interval and thereafter negative (the thigh was decelerating). The component primarily responsible for the positive resulting moment was the hip moment (M_H), which during the last third of the analysed time interval in particular reached a very high magnitude. During the second half of the analysed time interval, the hip moment was counteracted by the moment component from lower leg angular acceleration (α_L), and during the last third also by the moment component from lower leg angular velocity (ω_L). These negative moments, and especially the latter, were of such magnitude that they forced the resulting moment negative despite the large positive hip moment.

The lower leg moment curves show that the resulting moment ($I\alpha$) had a negative phase between 25% and 50% of the analysed time interval, when the lower leg was accelerated backwards into knee flexion. During the last 50% of the analysed time interval, the backward swing was stopped and the lower leg was forcefully accelerated forwards, as indicated by the large positive resulting moment. The components primarily responsible for the positive resulting moment were the knee moment (M_K) and the moment from thigh angular velocity (ω_T). The decline of both these components caused the decline of the resulting moment and eventually its shift from positive to negative. The instant of this shift marks the end of lower leg acceleration and thus the instant of maximal lower leg angular velocity (t_2).

Computed moments from the very first part of the movement before the foot left the ground should be treated with caution. This is because the biomechanical model used in this study did not include external forces and as long as the foot is in contact with the ground, ground reaction forces can influence the movements of the whole system. We do not consider this a major problem because the foot left the ground well before the phase of the movement during which thigh deceleration and lower leg acceleration took place.

To quantify the contribution from each moment component to the velocity changes of the segments

during the analysed time interval, the moment curves in Fig. 3 were numerically integrated yielding angular momentum in accordance with equation (6). Averaged integrals from all 17 subjects are shown in Fig. 4. Before averaging, contributors to thigh and lower leg angular momentum of each subject were normalized relative to the subject's resulting angular momentum, which was set to 100%.

For the thigh, the primary accelerating factor was the hip moment (M_H), while the primary decelerating factor was lower leg angular velocity (ω_L). The percentage values from the thigh frame in Fig. 4 (mean and S.E.M.) are $I\alpha$ 100 (normalized), M_H 3560 ± 985 , ω_T 1310 ± 352 , α_T -479 ± 132 , ω_L -2370 ± 693 , α_L -1150 ± 328 , g -698 ± 210 and a_H -69 ± 17 . For the lower leg, the primary accelerating factors were knee moment (M_K) and thigh angular velocity (ω_T). The percentage values from the lower leg frame in Fig. 4 are $I\alpha$ 100 (normalized), M_K 41.2 ± 1.3 , ω_T 73.9 ± 2.5 , α_T -26.0 ± 2.2 , g -11.4 ± 0.80 and a_H -0.6 ± 0.1 .

Electromyography

The full-wave rectified EMG signals in Fig. 3 show that the gluteus maximus was not activated. The vastus lateralis and rectus femoris were activated about 35% and 55% into the analysed time interval and then remained active for the rest of the time interval. The biceps femoris and gastrocnemius were active in the first 30% and 40% of the analysed time interval and then showed some activity again during the last third of the interval. Immediately after the analysed time interval, the biceps femoris and gastrocnemius were activated.

Temporal muscle activation was determined from visual inspection of the linear envelope of the EMG. In Fig. 5, EMG 'on/off' is shown together with the resulting moment and the muscle moment acting on the thigh and lower leg. In Fig. 5, the analysed time interval for each subject was normalized to 100%. Absolute values of the analysed time interval ranged from 250 to 400 ms, with smaller subjects tending to have a shorter interval and vice versa. Electromechanical delay was not taken into account in Fig. 5.

The temporal EMG activity of all 17 subjects was in general similar to that of the single subject depicted in Fig. 3 and described above. Note that only four subjects showed gluteus maximus activity.

Discussion

In the present study, the general points of interest were the cause of thigh deceleration and whether this thigh

deceleration has any beneficial effect on lower leg acceleration.

Thigh deceleration

First, it was clear that the thigh was almost completely decelerated to zero velocity at the time of maximal

Figure 4 Time integrals of moments acting on the thigh and lower leg from start of movement until time of maximal lower leg angular velocity. Nomenclature follows the shorthand notation explained in the text. Values for each subject were normalized relative to the resulting angular momentum, which was set to 100% before averaging. Values are means $\pm$ S.E.M. ($n = 17$).

lower leg angular velocity. This can be seen in Fig. 4, where the thigh angular momentum bar ($I\alpha$) is barely visible. Of the moments making up the resulting momentum, the hip flexor moment (M_H) was dominant among the accelerating factors. Hip flexion was counteracted by two factors: lower leg movement (α_L and ω_L) and gravity.

No support was found for active thigh deceleration by a hip extensor moment (whiplash action). This was further supported by the EMG recordings. The large hip flexor moment originated from the rectus femoris (Fig. 5G) and presumably also from the iliopsoas, which we did not record. The rectus femoris was in general first activated when about 40% of the analysed time interval had elapsed, most likely so as to avoid a knee extensor moment in the first part of the movement. The initial thigh acceleration exerted by the hip flexor muscles is maximized when the radius of gyration of the leg is minimized, which occurs when the knee joint is allowed to flex.

Only 4 of the 17 subjects showed activity in the gluteus maximus (Fig. 5E), which further emphasizes the absence of active thigh deceleration. Hamstring activity could, however, generate a hip extensor moment. In soccer kicking, Robertson and Mosher (1985) and Zernicke and Roberts (1976) found that the hip extensors became dominant just prior to ball contact. In none of these studies were EMG recordings performed, but Robertson and Mosher (1985) suggested that the hip extensor moment was created by the hamstrings. In the present study, the biceps femoris activity shown by 13 of the 17 subjects at the end of the analysed time interval (Fig. 5H) did not seem to influence the hip moment, which continued to exhibit pronounced flexor dominance (Figs 3 and 5B). The hamstring muscles were presumably activated to prevent hyperextension of the knee joint.

There are two reasons why the hip flexor muscles were unable to maintain a resulting hip flexor moment. First, the lower leg angular velocity was building up and gave rise to a motion-dependent hip extensor moment (ω_L in Figs 3 and 4). Second, the iliopsoas and rectus femoris force and thus the hip flexor moment itself decreased, partly because both muscles shortened (force-length relationship) and partly

Figure 3 Moments acting on the thigh and lower leg during a full kicking movement. Nomenclature follows the shorthand notation explained in the text. Positive direction is defined as counterclockwise (i.e. positive moments accelerate thigh and lower leg in the kicking direction). Vertical dashed lines indicate start of movement (t_1) and time of maximal lower leg angular velocity (t_2). Simultaneous EMG recordings are shown without the final linear envelope low-pass filtering. The uppermost end of the hip segment (iliac crest marker) determines the temporal alignment of the stick figures. For the thigh, note that between t_1 and t_2 , the large accelerating component (M_H) is counteracted by the components from lower leg angular velocity and acceleration (α_L and ω_L). For the lower leg, note that between t_1 and t_2 , the resulting knee extensor moment ($I\alpha$) is almost exclusively composed of the knee muscle component (M_K) and the component from thigh angular velocity (ω_T).

A Thigh
Resulting moment ($I\alpha$)
Positive moment
Negative moment

B Thigh (Hip)
Hip moment (M_H)
Positive (flexor) moment
Negative (extensor) moment

C Lower leg
Resulting moment ($I\alpha$)
Positive moment
Negative moment

D Lower leg (Knee)
Knee moment (M_K)
Positive (extensor) moment
Negative (flexor) moment

E
gluteus maximus

F
vastus lateralis

G
rectus femoris

H
biceps femoris

gastrocnemius

because the shortening velocity of the rectus femoris increased with knee extension velocity (force-velocity relationship). However, muscle force appeared to be of minor importance, as the time course of the moments revealed that a large hip flexor moment was maintained throughout the analysed time interval (M_H in Fig. 3).

Our conclusion concerning the observed thigh deceleration is that active deceleration does not take place. On the contrary, all subjects maintained a hip flexor moment throughout the kicking movement. Therefore, the thigh is decelerated by motion-dependent moments caused by lower leg motion, primarily lower leg angular velocity. This is in agreement with the results of Putnam (1983, 1991) and Dunn and Putnam (1988).

Although Robertson and Mosher (1985) and Zernicke and Roberts (1976) found a hip extensor moment just prior to ball contact in soccer kicking, we disagree with Robertson and Mosher (1985) that the hip extensor muscles caused the thigh deceleration. A hip extensor moment will decelerate the thigh provided nothing else causes the resulting hip moment to be directed towards flexion, but to decelerate the thigh fully, an extensor moment must be of sufficient magnitude for a sufficient period of time to counteract the build-up of angular momentum in the thigh during the period of thigh acceleration. We do not think that this was the case in the studies mentioned above, where hip moments were slightly negative (extensor moment) over a short period of time. Furthermore, in the study by Robertson and Mosher (1985), thigh angular velocity seemed to decrease before the hip moment shifted from flexor to extensor dominance. Although the hip extensor moment observed in these studies could have contributed in a small way to thigh deceleration, we believe the major contributor to have been motion-dependent moments caused by lower leg movement, primarily lower leg angular velocity, as was the case in the present study.

Lower leg acceleration

Figure 4 shows that lower leg angular momentum was built up by the knee extensor moment (M_K) and by the motion-dependent moment from thigh angular velocity (ω_T). It is readily accepted that the knee extensor muscles play an important role in the kicking motion, but the importance of thigh angular velocity is seldom recognized.

In Fig. 3, the time course of these moments is displayed. As mentioned above, the resulting moment ($I\alpha$) primarily consisted of the knee moment (M_K) and the motion-dependent moment from thigh angular velocity (ω_T). Therefore, when M_K and ω_T decreased towards zero, $I\alpha$ followed until it became negative and the lower leg stopped accelerating and started to decelerate.

The simplest reason for the decrease in the knee extensor moment (M_K) is that the knee extensor muscles stopped their activity. This occurred in about 50% of the subjects (Figs 5F and 5G). However, when relaxation time of up to approximately 100 ms is accounted for (Vos *et al.*, 1990, 1991), we consider these muscles to have been active at the end of the analysed time interval. Despite being active, the knee muscles' potential for force production is presumably limited, partly because of the increasing knee extension velocity (force-velocity relationship) and partly because of the decreasing knee flexion angle (force-length relationship). Furthermore, antagonist activity in the biceps femoris and gastrocnemius towards the end of the analysed time interval (Figs 5H and 5I) caused the knee extensor moment to decrease and ultimately to shift from extensor to flexor dominance (Figs 3 and 5D). This was in contrast to the hip joint, where antagonist activity in the biceps femoris towards the end of the analysed time interval did not seem to influence the hip joint moment. It has previously been demonstrated that an activated biceps femoris can influence the knee joint moment without influencing the hip joint moment (Hoy *et al.*, 1990). These authors demonstrated how hamstring action on the hip and knee joints depends on joint angles. When the hip joint is flexed approximately 100° and the knee joint is flexed approximately 80° , which was the case at the end of the analysed time interval in the present study, the hamstrings are able to exert a considerably larger moment at the knee joint than at the hip joint. Furthermore, at the knee joint, the biceps femoris was assisted by the gastrocnemius in about 50% of the subjects (Fig. 5I), while at the hip joint there was very little assistance from the gluteus maximus (Fig. 5E), as mentioned above. At an 80° knee joint angle, the gastrocnemius is able to enhance the knee flexor moment from the hamstrings by about 25% (Hoy *et al.*, 1990).

The decrease in the motion-dependent moment from thigh angular velocity (ω_T) can in part be explained by the decrease in thigh angular velocity and

Figure 5 Temporal moment and muscle activation data. Moments (A–D) can be positive (counterclockwise), negative (clockwise) or zero. Muscle activation (E–I) can be on or off. Time from start of movement until maximal lower leg angular velocity was normalized to 100% for each subject. The uppermost end of the hip segment (iliac crest marker) determines the temporal alignment of the stick figures.

in part by the decrease in the moment arm (dashed line *b* in Fig. 2), due to a decrease in knee flexion angle below 90°. As maximal lower leg angular velocity occurred when the knee was flexed approximately 80° (i.e. close to 90°), the decrease in thigh angular velocity was probably the most important factor.

The time course of the motion-dependent moment component from thigh acceleration (α_T) was seen to shift between positive and negative phases during the analysed time interval (Fig. 3); that is, it shifted between accelerating and decelerating the lower leg. When the thigh is accelerating, this moment component will accelerate the lower leg if the knee joint is flexed more than 90°; otherwise, it will decelerate the lower leg. When the thigh is decelerating, this moment component will accelerate the lower leg if the knee joint is flexed less than 90°; otherwise, it will decelerate the lower leg. During the analysed time interval, the thigh first accelerated then decelerated, while the knee joint was first flexed less than 90°, then more than 90° and then again less than 90°. Accordingly, these dynamic changes in direction of thigh acceleration and knee joint angle caused the phase shifts and thereby the shifts in action on lower leg acceleration. Taken over the full analysed time interval, thigh acceleration (positive and negative) was seen to reduce the build-up of lower leg angular momentum (Fig. 4).

The relative magnitude of the positive angular momentum components in the present study was approximately 64% for ω_T and 36% for M_K (Fig. 4). In a study of punt kicks, which can be characterized as a 'lower' version of the high front kick, Putnam and Dunn (1987) found the relative distribution of positive angular momentum components to be approximately 50% (ω_T), 25% (M_K) and 25% (a_H), the latter being the linear acceleration of the hip. Although not stated explicitly by these authors, we speculate that the contribution from hip linear acceleration could have been due to their subjects performing the kick after a short preliminary run-up. This is in contrast to the present study, where the kick was performed without any preliminary run-up, resulting in a negligible contribution from a_H (Fig. 4). If the 25% contribution of the a_H component is arithmetically removed from the results of Putnam and Dunn (1987) and this is assumed not to influence the other components, the relative distribution of positive angular momentum components was similar to the present study.

Our conclusion is that thigh deceleration has no beneficial effect on lower leg acceleration. Any discussion of this question may be seen to be rather academic after having shown that thigh deceleration is not performed actively. Nevertheless, including it in this descriptive kinetic analysis enables us to speculate about active thigh deceleration. When the thigh is

swung forward by the hip flexor muscles, the whole leg can be regarded as a double pendulum. Angular acceleration of the lower leg is caused by maintaining a high thigh angular velocity. But this is a negative feedback system: when the lower leg angular velocity increases, it causes the thigh angular velocity to decrease, thereby removing the cause of its own acceleration. In this light, it would appear reasonable to try to maintain a high thigh angular velocity by maintaining a large hip flexor moment and refraining from active deceleration, as in the present study. Because the kicking leg has to obey the laws of physics, it is questionable to speak out about a certain kicking strategy. After the hip flexors have initiated the movement, the intersegmental dynamics are responsible for the gross result, that is the proximal-to-distal sequential movement. This does not mean that the knee extensor muscles cannot contribute to the lower leg angular velocity. We wonder, however, whether the function of the knee muscles is coordinative instead of, or in addition to, just lower leg acceleration. If the knee angle is close to 90° when thigh angular velocity is maximal, the lower leg accelerating moment component (ω_T) will be maximized. Coordination of knee angle with respect to thigh angular velocity could be the most important task of the knee muscles.

Conclusion

The results of this study suggest that thigh deceleration was caused by motion-dependent moments from lower leg motion, primarily lower leg angular velocity. No support for active deceleration was found, neither in the kinetic computations nor in the EMG recordings. Furthermore, our results suggest that lower leg acceleration was caused by a knee extensor moment contributing about one-third of the total lower leg angular momentum, and by a motion-dependent moment from thigh angular velocity contributing the remaining two-thirds. Thigh angular deceleration was not seen to enhance lower leg acceleration.

We would like to point out that, to assess the contribution of different joint moments, joint or segment angular velocities, or joint linear velocities to the outcome (final velocity) of a certain movement, it is necessary to study the time course of the respective joint and motion-dependent moments, for instance by time integration of these moments as in the present study. Regarding final velocity as an instantaneous phenomenon can lead to erroneous conclusions. For instance, in the high front kick in the present study, the thigh was almost stationary at the instant of maximal lower leg velocity, which could lead to the naive conclusion that lower leg angular velocity or foot linear velocity is gen-

erated solely by the knee extensor muscles and that hip flexor muscles merely serve to position the thigh for the high target. One has to acknowledge instantaneous velocity as the result of an acceleration history and thus study the forces and moments responsible for the accelerations.

Finally, a causal relationship between the deceleration of proximal segments and acceleration of distal segments has previously been proposed (Müller, 1982), and the data from the present study support this. However, the causality must be properly oriented: distal segment acceleration and velocity cause proximal segment deceleration, not vice versa. Thus, the intersegmental dynamics of a kicking leg should presumably not be described as whiplash-like, but rather as flail-like.

Acknowledgements

This study was supported by grants from the Danish Elite Sport Association (Team Danmark), the Danish Taekwon-do Federation (Dansk Taekwon-do Forbund) and the Danish Research Council for Sports (Idrættens Forskningsråd).

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