



MECHANICAL OUTPUT FROM INDIVIDUAL MUSCLES DURING EXPLOSIVE LEG EXTENSIONS: THE ROLE OF BIARTICULAR MUSCLES

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Abstract—The main result of this study is that biarticular leg muscles contribute significantly to the work done at joints, due to transfer of power during explosive leg extensions. In particular, a net power transfer was shown from hip to knee joint during jumping and sprinting.

Seven elite athletes performed explosive one legged jump and sprint push-offs. Kinematics, ground reaction forces and electromyography (EMG) of leg muscles were recorded. The mechanical output of six individual muscle groups was estimated by using Hill-based muscle models. The EMG and kinematics served as input to these models.

For jumping as well as for sprinting, the model estimated similar results for the relative work contribution done about a joint due to transfer of power by the biarticular muscles. Rectus femoris showed a power transfer from hip to knee joint, while in contrast hamstrings showed a power transfer from knee to hip joint. Regardless of these opposite directions of power transfer, a net transfer occurred from the hip to the knee joint.

The relative work contribution of hamstrings done in hip extension was 7% in jumping and 11% in sprinting. For rectus femoris, the relative work contribution done in knee extension was 21% in jumping and 31% in sprinting. Power transferring actions by gastrocnemius from knee to ankle contributed 25% in jumping and 28% in sprinting to the work done in plantar flexion.

These results support the hypothesis that the action of biarticular muscles contributes to a net transfer of power from proximal to distal joints during explosive leg extensions. This action of the biarticular muscles causes an efficient conversion of body segment rotations into the desired translation of the body centre of gravity.

INTRODUCTION

The existence of biarticular muscles has fascinated many researchers (see Ingen Schenau *et al.*, 1990). In previous studies on jumping and sprinting, the hypothesis was forwarded that biarticular muscles play an important role in transferring power from proximal to distal joints (Bobbert and Ingen Schenau, 1988; Jacobs and Ingen Schenau, 1992a). Specifically, a proximal to distal power transfer is believed to cause an efficient conversion of successive rotational motions of body segments into translation of the body centre of gravity (BCG). The mechanism by which this is accomplished is thought to be the timely activation of rectus femoris and gastrocnemius before the end of push-off. Activation of rectus femoris was proposed to decrease the angular acceleration of the trunk, which, combined with the onset of knee extension, ensued in power being transferred from hip to knee joint. Similarly, activation of gastrocnemius prior to the end of the push-off transferred power generated by the knee extensors from knee to ankle joint.

The transfer action of gastrocnemius from knee to ankle joint was demonstrated for jumping (Bobbert *et al.*, 1986). Based on simulation experiments of plantar flexors

in jumping, using only kinematics as input to Hill-based muscle models to estimate soleus and gastrocnemius forces, the relative contribution to the total amount of work done at the ankle joint due to the transfer action of gastrocnemius was quantified (Bobbert *et al.*, 1986). As a result, they computed that 25% of the total amount of work done about the ankle is due to a transfer action by gastrocnemius from knee to ankle joint.

The question then arose whether a net flow of power from hip to knee joint could also be demonstrated. In jumping, electromyographic (EMG) activity of rectus femoris was recorded during simultaneous hip and knee extension; however, no estimates of rectus femoris forces were available. Moreover, hamstrings were also found to be active. To estimate the flow of power from hip to knee joint, individual forces of hip and knee muscles have to be estimated. However, the indeterminacy problem impedes a straightforward calculation. An approach for obtaining these estimates can be the use of kinematics and EMG as input to muscle models. In the present study, this is done for one-legged jumping and sprinting. Analogous approaches were applied to estimate the mechanical output of the planar flexors during walking, stepping (Hof *et al.*, 1983, 1987) and running (Hof, 1990; Jacobs *et al.*, 1993) and of the jaw muscles during mastication (Ruijven and Weijs, 1990).

It is important to test the role of biarticular muscles in jumping as well as in sprinting, since both tasks start at similar joint angles and are based on a proximal to distal

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sequence of muscle activation but the translation of BCG is different (Bobbert and Ingen Schenau, 1988; Jacobs and Ingen Schenau, 1992a). In the sprint, a more reciprocal activation between hamstrings and rectus femoris muscles was found compared to the activation patterns in the jump (see Jacobs and Ingen Schenau, 1992a).

In the present study, we quantified the contribution of the biarticular hamstrings, rectus femoris and gastrocnemius to the work done about joints due to the phenomenon of power transfer between adjacent joints in one-legged jumping and sprinting.

METHODS

Subjects and experimental procedure

Seven elite male athletes (age 23 ± 2 yr, body mass 77.8 ± 4.3 kg, upper leg length 46.8 ± 1.5 cm, lower leg length 43.0 ± 1.6 cm) participated in this study. After warming-up, subjects performed standard isometric contractions (SICs) to obtain reference EMG levels. Since the subjects were primarily runners and not familiar with performing one-legged jumping, they were allowed to practice jumping before the experiment session. Subsequently, each athlete performed three one-legged maximal executed vertical jumps as well as three maximal executed sprint push-offs. During the execution of both tasks, the movement pattern, ground reaction forces, and EMG of eight leg muscles were recorded. The analysis of the sprint push-offs focused on the second stance phase after leaving the starting blocks, during which the body is mainly accelerated in the horizontal direction. The analysis of the jump push-offs focused on the phase from the initial squat position to foot-off. Methods used for the analysis of both tasks will be described briefly since they have been set out in detail previously (Bobbert and Ingen Schenau, 1988; Jacobs and Ingen Schenau, 1992a).

Kinematics and kinetics

Landmark coordinates (Fig. 1) were extracted from cine film with the aid of a motion analyzer (Dynamic Frame, NAC), and low-pass-filtered (zero phase lag by bidirectional application of a fifth approximation, second-order Butterworth filter) with a net cut-off frequency of 16 Hz. These coordinates were used to calculate the orientation of four body segments in a sagittal plane, and joint angles were calculated as the angles between adjacent body segments (Fig. 1). The ground reaction force vector was recorded by means of a force platform (Kistler, type 9281B). Instantaneous net moments about hip, knee and ankle joints were obtained by means of linked-segment modelling. Hip and knee extension moments and plantar flexion moments were defined as positive. Net joint power was calculated by multiplication of net joint moment and joint angular velocity.

Electromyography

Surface electrodes were applied to the skin over gluteus maximus, semitendinosus and biceps femoris (caput

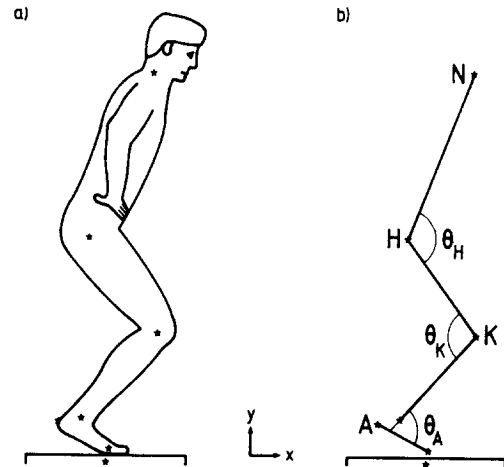


Fig. 1. (a) Positions of landmarks applied to the skin of the subject and to the force platform. (b) Definition of angles in joints. N, H, K and A indicate neck, hip, knee and ankle, respectively. θ_H , θ_K and θ_A indicate angles in hip, knee and ankle joints, respectively.

longum), vastus medialis and lateralis, rectus femoris, soleus and gastrocnemius medialis. The subjects wore a skin suit to reduce movement artefacts due to swinging of cables. Landmarks remained visible through openings in the suit. The EMG signals were transmitted telemetrically (BIOMES 80), filtered to further reduce movement artefacts (bandpass 25–200 Hz), and sampled at 500 Hz. Subsequently, these signals were rectified and low-pass-filtered (zero phase lag by bidirectional application of a fifth approximation, second-order Butterworth filter) with a net cut-off frequency of 75 Hz.

Muscle model

Six muscle groups were modelled: three monoarticular muscles, gluteus maximus (GU), vasti (VA, as an average of vasti medialis and lateralis) and soleus (SO), and three biarticular muscles, hamstrings (HA, as an average of semitendinosus and biceps femoris CL), rectus femoris (RF) and gastrocnemius (GA) (Fig. 2). Forces of these individual muscle groups were estimated by means of Hill-based muscle models. The experimentally measured kinematics and EMGs served as input to the model (Fig. 3).

The muscle model comprises three parts: (1) EMG which serves as estimator of muscle stimulation. (2) Activation dynamics corresponding to the transformation of muscle stimulation to active state of the contractile apparatus, and (3) muscle dynamics corresponding to the transformation of active state to muscle force. A description of these parts follows together with the computational procedure for calculating individual muscle force.

Muscle stimulation. To obtain estimates of muscle stimulation, surface EMG records were used. Although it would have been more ideal to use individual excitation of all motor units, it was not possible at this time to record unitary activity from single motor units during explosive movements. The only accessible measure was

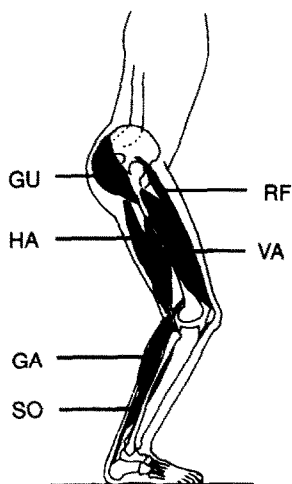


Fig. 2. An illustration of the position of the leg muscles. GU is gluteus maximus, HA is hamstrings (semitendinosus and biceps femoris CL), VA is vasti (vastus medialis and lateralis), RF is rectus femoris, SO is soleus and GA is gastrocnemius.

the surface EMG reflecting the spatial and temporal activity of many motor units (Hermens *et al.*, 1986; Ruijven and Weijis, 1990). In surface EMG, firing rate and recruitment control are indistinguishable (Zajac, 1989). Thus, increases in EMG coincide with either active motor units discharging faster and/or additional motor units being recruited. Given the fact that both firing rate and motor unit recruitment increase with excitatory drive to the muscle, EMG can meaningfully be interpreted as the net neural control signal to the muscle (Zajac, 1989). Therefore, it was acceptable to use the rectified surface EMG as a reliable estimator of muscle stimulation. Maximal EMG was estimated by multiplying the EMG measured during SIC by the ratio of maximum joint moment to joint moment during SIC. In doing so, it was assumed that during SIC the muscles involved in delivering the moment were producing the same relative force (Table 1). Subsequently, all EMG values were expressed in percentages of this maximum (Fig. 3).

Activation dynamics. Active state, defined by Ebashi and Endo (1968) as the relative amount of calcium bound to troponine, is related to the level of stimulation of the muscle (estimated from EMG). This process can be modelled by a first-order process according to Hatze (1981). The concentration of free calcium (algebraically related to active state) and length of contractile element of the muscle (L_{CE}) served as state variables. Instantaneous time derivatives of calcium concentration were calculated from instantaneous values of calcium concentration, L_{CE} , and muscle stimulation. The transformation of stimulation to active state was scaled to the relative amount of slow twitch fibres according to Hatze (1981). Estimations of these percentages were depicted on the basis of data reported by Yamaguchi *et al.* (1990).

Muscle dynamics. The use of Hill-based muscle models yields an adequate description of muscle dynamics in the

context of modelling of multi-joint behaviour (e.g. Winters, 1990). Soest *et al.* (1992) described a general-purpose Hill-type muscle model comprised of a series elastic element (SEE), a contractile element (CE), and an elastic element (PEE) parallel to the CE (Fig. 4). The force of SEE and PEE depends in a non-linear way on length of these elements (L_{SEE} and L_{PEE} , respectively). The force of CE depends on active state, length of CE (L_{CE}) and velocity of CE (V_{CE}).

Where possible, parameter values were derived from morphological data reported in the literature [see Soest *et al.* (1995) for a definition of the parameters]. A number of parameters were given identical values for all muscles: stretch of SEE at maximum isometric force (F_{MAX}) was set to 4% of SEE slack length (E_{SEEMAX} ; Morgan *et al.*, 1978); L_{CE} at which active force was zero was set to 0.44 and 1.56 times $L_{CE(OPT)}$. Hill's force velocity parameters a/F_{MAX} and $b/L_{CE(OPT)}$ were scaled to the percent of slow twitch fibers according to Winters (1990). Values of the muscle specific parameters are listed in Table 1 (see also Soest *et al.*, 1995). CE optimum lengths were derived from estimated sarcomere numbers (Huijing, pers. commun.). Relative values of maximal isometric force of agonists were based on muscle cross-sectional areas, whereas absolute values of the total maximal isometric force were chosen in such a way that realistic maximal isometric moments were obtained. SEE slack lengths were chosen in such a way that the joint angle at which the isometric joint moment is maximal corresponds with experimentally obtained isometric moment-angle relations. The muscle moment arm data used in this study were derived from experimentally reported data by Grieve *et al.* (1978) and Visser *et al.* (1991), as well as from Huijing *et al.* (in preparation). Moment arm values as well as $L_{CE(OPT)}$ and L_{SLACK} for each subject were scaled to the ratio of upper leg length (for GU, HA, VA and RF) or lower leg length (for SO and GA) obtained experimentally to the upper and lower segment lengths at which these values were determined (see Table 1).

Computational procedure. Since muscle mass was neglected, the instantaneous muscle force (F_{SEE}) was calculated from L_{SEE} ($L_{SEE} = L_{OI} - L_{CE}$). From F_{CE} , V_{CE} was computed according to the force-velocity relationship. To complete the computational circle (Fig. 5), new L_{CE} and Ca values were obtained by integrating V_{CE} and rate of change of calcium concentration with respect to time.

Treatment of data

The jump push-off phase was defined as the phase starting at the instant that the body's centre of mass started moving in the vertical direction and ending at toe-off. The sprint push-off phase was defined from touchdown to toe-off. Simulations were performed starting 75 ms before the push-off phase and ending at toe-off ($t = 0$). Previous research showed that subjects performed both explosive tasks in a rather stereotyped manner as expressed by the small variability (Bobbett and

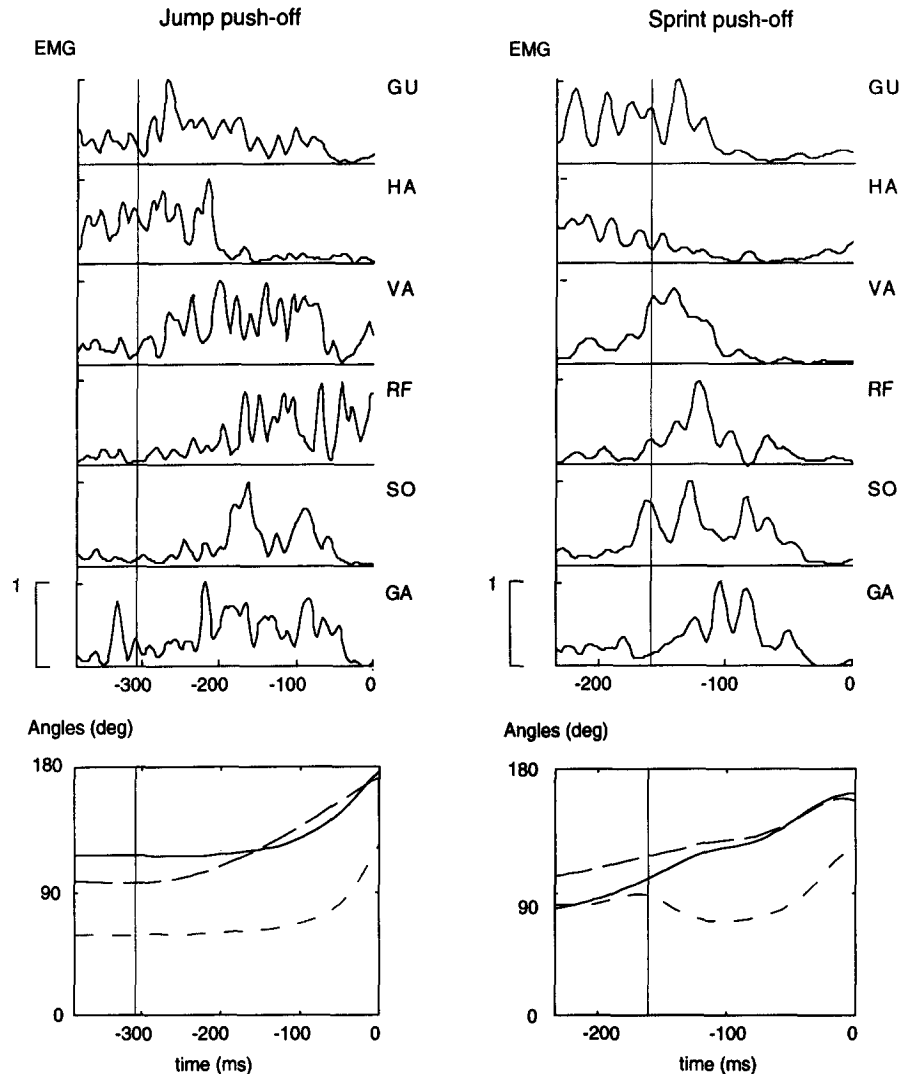


Fig. 3. Typical EMG and joint angle data that have been used as input to the muscle models recorded from the same subject in jumping and sprinting. The EMG of six muscle groups is shown: GU is gluteus maximus, HA is hamstrings (semitendinosus and biceps femoris CL), VA is vasti (vastus medialis and lateralis), RF is rectus femoris, SO is soleus and GA is gastrocnemius. Maximal EMG was estimated by multiplying the EMG measured during SIC by the ratio of maximum joint moment to joint moment during SIC. Subsequently, all EMG values were expressed in percentages of this maximum. The joint angles of hip (solid), knee (long dashes) and ankle (small dashes) are shown. The jump push-off phase was defined as the phase starting at the instant that the body's centre of mass started moving in the vertical direction (about -310 ms) and ending at toe-off (0 ms). The sprint push-off phase was defined from touchdown (about -160 ms) to toe-off (0 ms). The start of the push-offs is indicated by a thin vertical line. Simulations were performed starting 75 ms before the push-off phase and ending at toe-off.

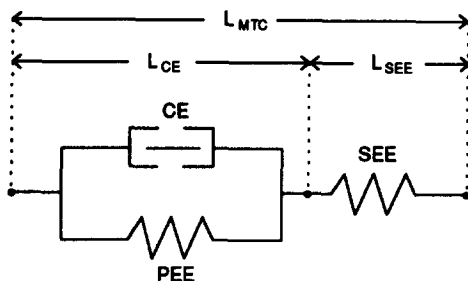


Fig. 4. Structure of the model to simulate the individual muscle forces. The model consists of a series elastic element (SEE), a contractile element (CE) and an elastic element (PEE) parallel to CE.

Ingen Schenau, 1988; Jacobs and Ingen Schenau, 1992a). Therefore, in this study only mean time histories of curves and values are given. Individual time histories of all variables were normalized to 100% of the mean stance time (Winter, 1983). From these normalized curves, mean curves for each subject and the total group were obtained.

Individual muscle moments were obtained by multiplication of the calculated muscle-tendon force (F_{MTC}) and the moment arm. The power produced by the muscular part of the muscle models (P_{CE}) was obtained by multiplication of V_{CE} and F_{CE} and the power of the tendinous part in series of the muscle models (P_{SEE}) was calculated

Table 1. Values of the muscle specific parameters. D is the average moment arm; $L_{CE(OPT)}$ means optimum CE length; L_{SLACK} is SEE slack length; F_{MAX} is maximum isometric force given for one leg; ST is the percentage of slow twitch fibres. Length values (D , $L_{CE(OPT)}$ and L_{SLACK}) were scaled by the percentage of subject specific segment lengths and the segment lengths at which these values were determined; upper leg length of 0.44 m and a lower length of 0.40 m

	D_{HIP} (m)	D_{KNEE} (m)	D_{ANKLE} (m)	$L_{CE(OPT)}$ (m)	L_{SLACK} (m)	F_{MAX} (N)	ST (%)
GU	0.062			0.200	0.150	2650	52
HA	0.077	0.026		0.104	0.370	6000	58
VA		0.042		0.093	0.160	5400	47
RF	0.035	0.042		0.081	0.360	930	40
SO			0.046	0.045	0.246	4235	81
GA		0.017	0.046	0.049	0.382	2370	50

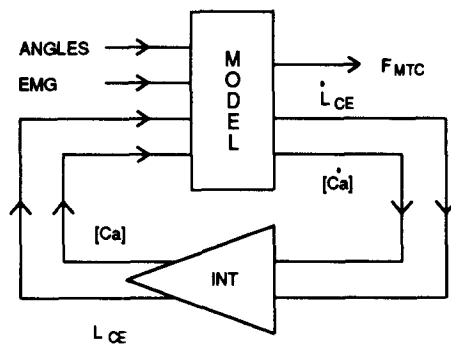


Fig. 5. Flow diagram for simulation used for studying the individual muscle behaviour. Joint angles and EMG served as inputs to the muscle model. The derivative of L_{CE} (\dot{L}_{CE}) Ca (\dot{Ca}) are calculated by the model. Integrating (INT) these values with respect to time, new L_{CE} and Ca are obtained. The calculated muscle force (F_{MTC}) is a function of L_{CE} and active state (Q).

by multiplication of V_{SEE} and F_{MTC} . In order to determine the power transferred (P_{TRANS}), e.g. from hip to the knee joint by RF, the F_{MTC} produced by RF was multiplied first by the moment arm of RF at the hip, resulting in the moment of RF about the hip, and second by the hip angular velocity, resulting P_{TRANS} for RF. P_{TRANS} was calculated from knee to hip joint for HA, from hip to knee joint for RF and from knee to ankle joint for GA.

To decide whether the model gave an accurate representation of the real system, the model outcome was evaluated. To evaluate the model, for jumping and sprinting, similarity was checked between experimentally acquired net joint moments (by means of inverse dynamics) and the sum of the estimated moments (by means of muscle model). The net hip moments were compared with the sum of GU, HA and RF moments, net knee moments were compared with the sum of VA, RF, HA and GA moments, and net ankle moments were compared with the sum of SO and GA moments. In addition, net joint powers were related to the total summation of individual muscle power curves (including P_{CE} and P_{SEE}) corresponding to the particular joint. Work values were calculated by integration of the power curves with respect to time.

Since the model parameters were based on an 'initial guess' from the literature, the question arises as to what extent model predictions will be affected by changing these parameters. To determine the influence of relative changes of the parameters choices on model predictions, for a range of parameters a sensitivity analysis was performed for the plantar flexors, according to the method proposed by Winters and Stark (1988). It was found that an increase in F_{MAX} and moment arm values positively influenced the amplitude of the plantar flexion moment, as could be expected. Changes in the time constant of the activation dynamics and E_{SEEMAX} affected the rise and decline of the plantar flexion moment curve.

RESULTS AND DISCUSSION

In the present study, the contribution of biarticular muscles to the work done about joints due to power transfer was estimated in jumping and sprinting. To estimate the individual contribution of muscles, Hill-based muscle models were used in which measured EMG and kinematics served as input.

This section will focus first on the comparison of the muscle model output and the results from inverse dynamics. Subsequently, the power output of the individual muscles will be presented. The main hypothesis addressing the net transfer of power from hip to knee joint will be discussed. Finally, the importance of the biarticular muscles in the execution of multi-joint movements will be discussed.

Evaluation of muscle model output with results from inverse dynamics

Figure 6 shows the comparison between the individual estimated muscle moments crossing a particular joint and the corresponding net joint moments obtained from inverse dynamics (see the Methods section). As can be expected, there was no perfect match. However, based on this comparison the conclusion was drawn that the muscle model can be used for the intended quantification.

There are a number of different sources that can account for the discrepancies. First, the comparison was

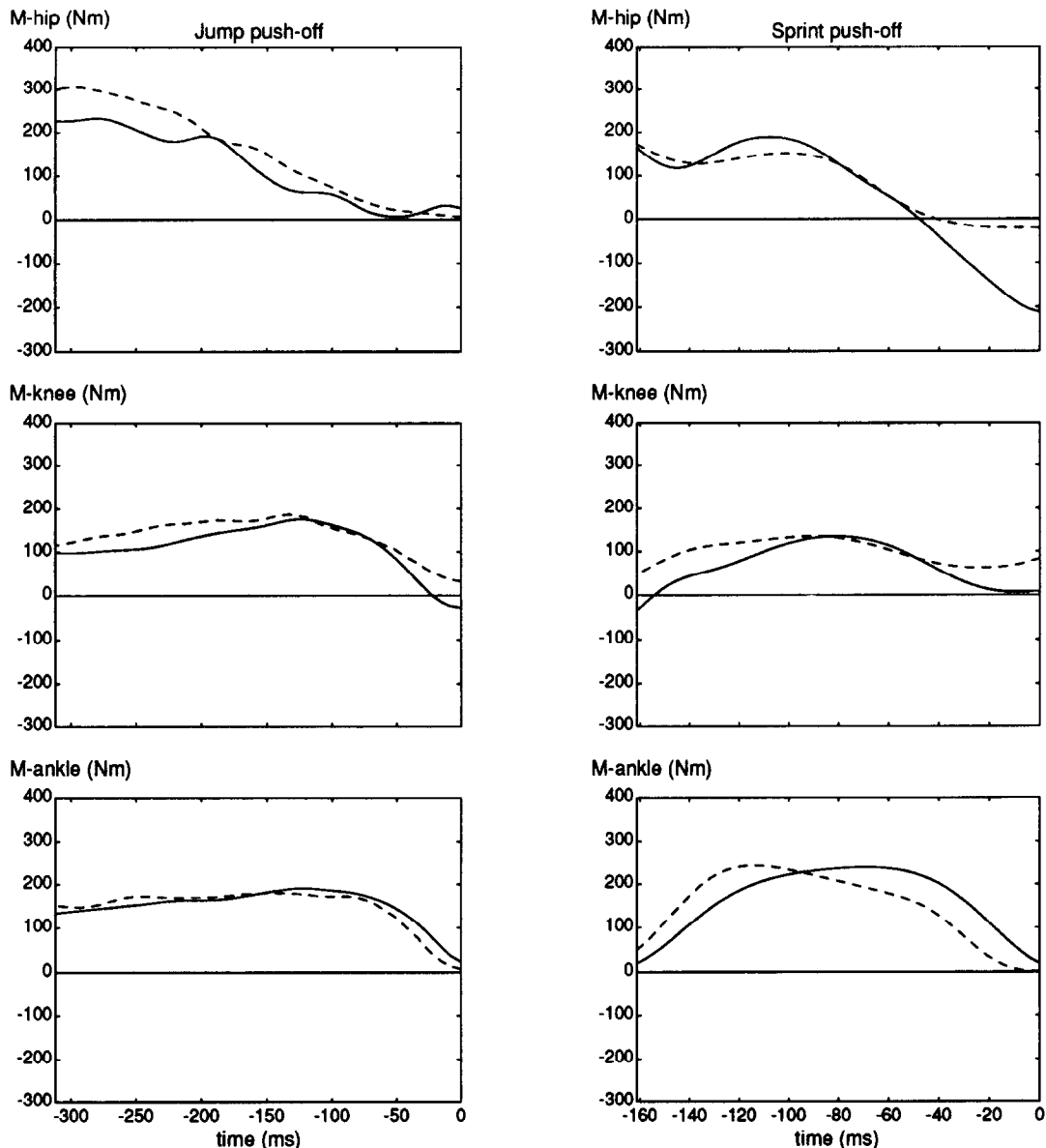


Fig. 6. Mean time histories ($n = 7$) of the *net* hip, knee and ankle joint moments obtained by inverse dynamics (solid line) and the sum of the estimated individual muscle moments corresponding to that particular joint (dotted line) for jumping and sprinting. The following comparisons are made: for the hip between the net moment and the sum of moments of HA, RF and GU; for the knee between the net knee moment and the sum of moments of HA, GA, RF and VA; for the ankle between the net ankle moment and the sum of moments of GA and SO.

made between a net joint moment and a summed moment of only a few muscles (2 to 4). Some of the deviations could have been minimized by modelling other muscles which were also activated; e.g. tibialis anterior, the deep polyarticular plantar flexors and iliopsoas (see Jacobs and Ingen Schenau, 1992a; Nilsson *et al.*, 1992). Second, based on the sensitivity analysis of the model, F_{MAX} , moment arm and E_{SEEMAX} influence the shape of the individual muscle moment curves. For instance, F_{MAX} was based on the literature and assumed to be constant despite the interindividual differences in maximal

isometric force. Finally, surface EMG could only be used as an estimator of stimulation (see the Methods section).

The contribution of a muscle to external work depends on the amount of power a muscle can generate. Figure 7 shows the net joint powers based on the inverse dynamic calculations, together with the sum of individual muscle powers estimated by the muscle model from the corresponding joint (see the Methods section). Based on this comparison, the muscle model was again considered appropriate for the power transfer quantification, despite

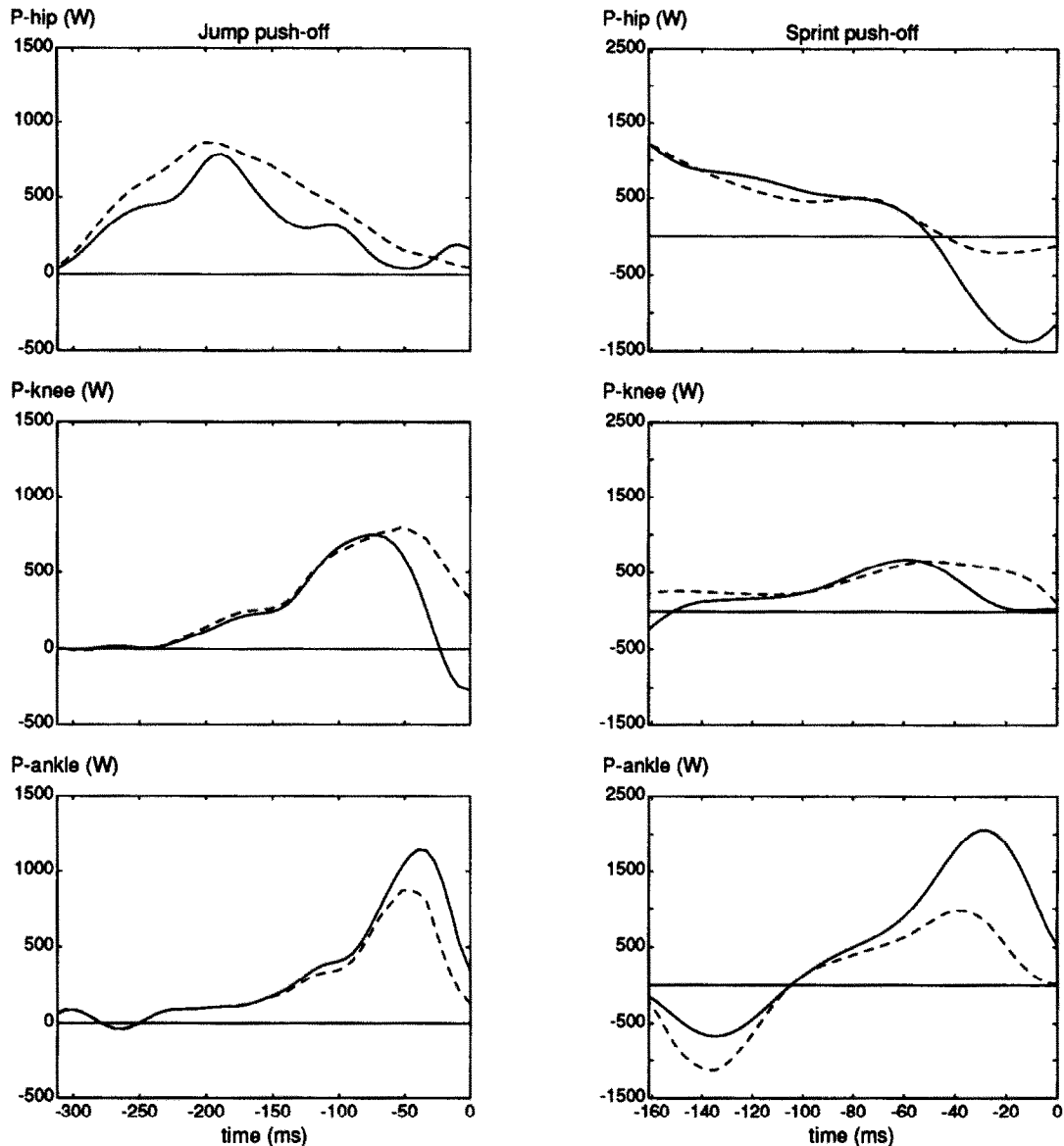


Fig. 7. Mean time histories ($n = 7$) of the *net* hip, knee and ankle joint powers obtained by inverse dynamics (solid line) and the sum of the estimated individual muscle powers corresponding to that particular joint (dotted line) for jumping and sprinting. With respect to the comparison of both power curves see the information in Fig. 6.

discrepancies being amplified by the angular velocity of each joint.

Power output of individual muscles

Figure 8 shows the power curves of contractile and series elastic elements of monoarticular and biarticular muscles as well as the curves of the transferred power by biarticular muscles. The estimated power curves of the plantar flexors in one-legged jumping are quite similar to previous estimations of these power curves (Bobbett *et al.*, 1986). The fact that the subjects in the present study were not considered elite jumpers, as in the study by Bobbett *et al.* (1986), was demonstrated by somewhat

lower power values for the plantar flexors when compared to this previous study.

Note that in comparing the present results of jumping and sprinting, the total duration of the push-off was longer for jumping than for sprinting: 310 and 160 ms, respectively. In addition, some different patterns of the estimated power curves exist, which will be described in general terms only. First, in sprinting, HA as well as VA muscles liberated a considerable amount of positive power in the first part of the push-off. In jumping, HA also liberated most of the positive power in the first part of the push-off, while relatively less positive power was produced by VA in this phase. Second, in jumping all the

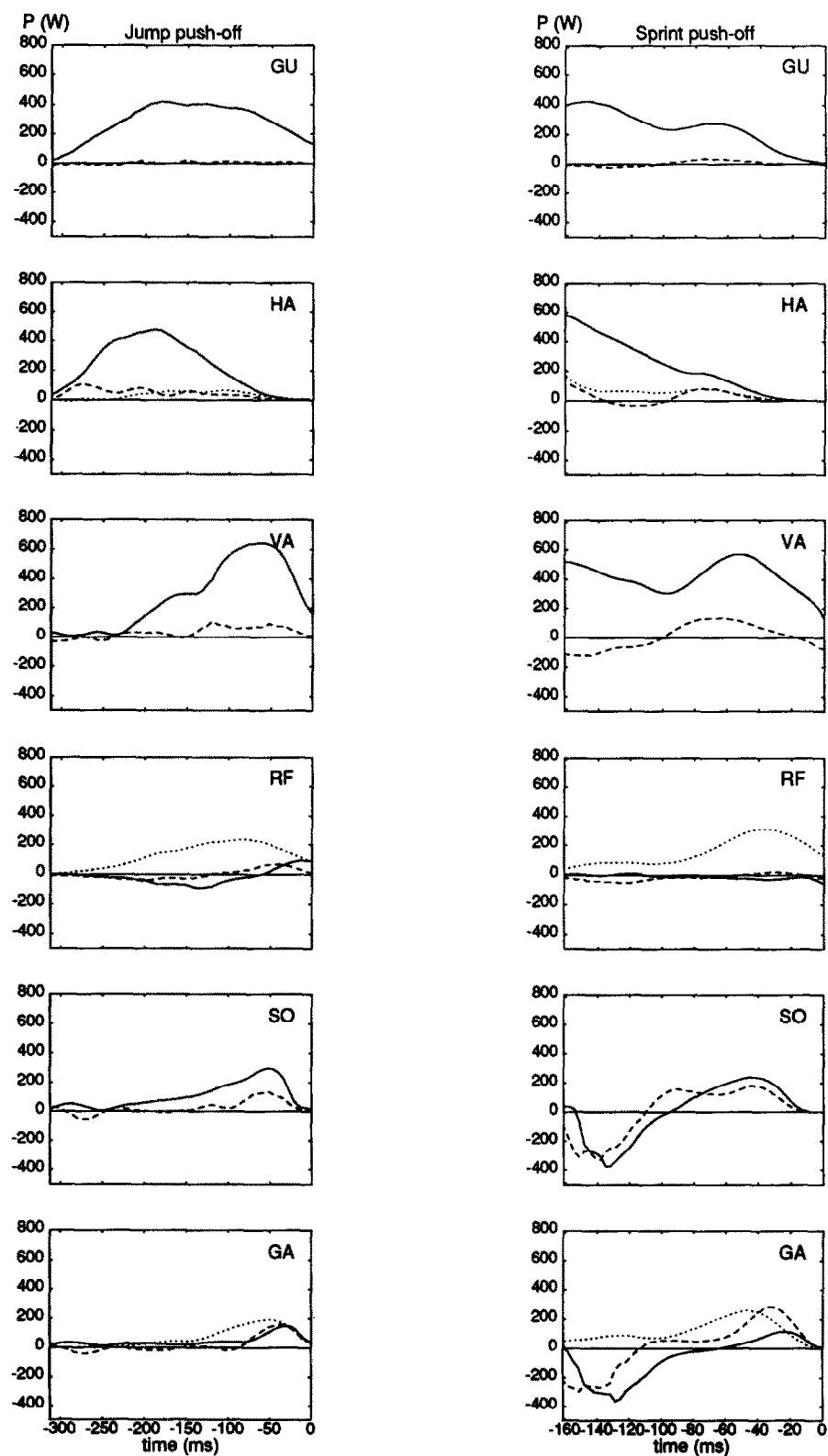


Fig. 8. Mean time histories ($n = 7$) of the power output of the muscle fibers (solid line) and series elastic element (dashed line) for the six muscle groups as well as the transferred power for the biarticular HA, RF and GA muscles (dotted line) for jumping and sprinting.

muscles produce positive power, whereas, in sprinting, the plantar flexors produced negative power starting at touchdown and ending at -90 ms. During the explosive part of the push-off phase (the last 90 ms), the total amount of positive muscle power liberated by GU was still high in jumping, while in sprinting the total amount of positive power liberated by GU decreased to zero during the explosive phase.

To make a realistic comparison between jumping and sprinting the work values were calculated for the last 90 ms (respectively Tables 2 and 3). In this period of time, angular velocities of the hip, knee and ankle joint increased considerably in sprinting as well as in jumping (see Bobbert and Ingen Schenau, 1988; Jacobs and Ingen Schenau, 1992a), and during this period all muscles contributed to positive work in the sprint. The difference in work output of all the six muscles as estimated by the model was 122 J in jumping and 103 J in sprinting.

The role of biarticular muscles in power transfer

As stated in the Introduction, the phenomenon of power transfer by biarticular muscles requires that these muscles distribute power from one joint to another. The total amount of power liberated by all muscles is not affected, but the distribution is directed where power could be applied more effectively. It is known for GA, from a previous study, that this muscle plays a unique role in power transfer from knee to ankle joint in one-legged jumping (Bobbert *et al.*, 1986). In the present study, a similar conclusion could also be drawn for GA in sprinting.

One of the questions to be answered in the present study was whether a net power transfer action occurred from hip to knee joint. The absolute work contribution due to power transfer is shown in Tables 2 and 3 for, respectively, jumping and sprinting. In Fig. 9, the relative work contribution due to the transfer action of HA, RF and GA is shown as well as the direction of transfer. This relative work contribution of biarticular muscles was calculated by taking the ratio of work due to power

Table 3. Model calculations of work done about the hip, knee and ankle joint of contributions of muscle fibres (W_{CE}) and series elastic tendon (W_{SEE}) and of the contribution of transportation (W_{TRANS}) by HA, RF and GA for the last 90 ms of the push-off in one-legged jumping

Joint	Muscle	W_{CE}	W_{SEE}	W_{TRANS}	Total
Hip	GU	14	1		15
	HA	7	3	3	13
	RF			-19	-19
Knee	VA	37	6		43
	RF	-1	0	19	18
	HA			-3	-3
	GA			-14	-14
Ankle	SO	12	10		22
	GA	3	11	14	28
Total		72	31	0	103

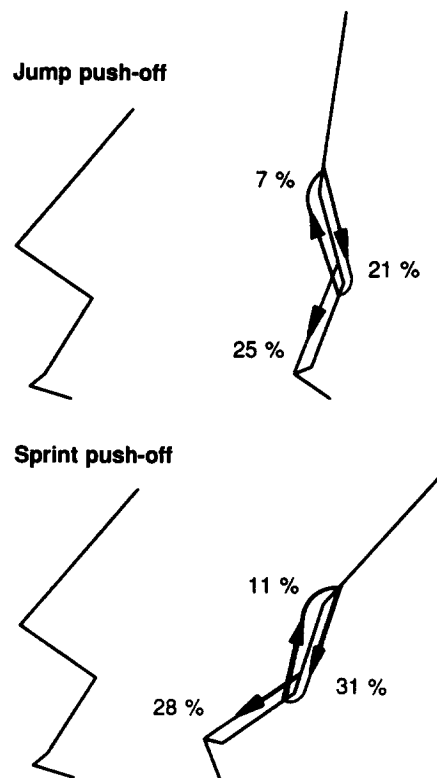


Fig. 9. The relative work contribution of the biarticular muscles (HA, RF and GA) to the work done about a joint due to its unique action in transferring power is illustrated as well as the direction of the transfer action. These values were calculated over the last 90 ms. The work values for jumping and sprinting during the last 90 ms are, respectively: for the hip 29 and 28 J, for the knee 70 and 61 J and for the ankle 53 and 50 J (see Tables 2 and 3). During both tasks, there is a net transfer of power from the hip to the knee and finally to the ankle joint.

transfer to the total amount of work produced by the muscles around that particular joint (see Tables 2 and 3).

In both tasks, jumping and sprinting, similar results were found for power transfer. For RF a transfer action occurred from the hip to the knee joint. However, HA

Table 2. Model calculations of work done about the hip, knee and ankle joint of contributions of muscle fibres (W_{CE}) and series elastic tendon (W_{SEE}) and of the contribution of transportation (W_{TRANS}) by HA, RF and GA for the last 90 ms of the push-off in one-legged jumping

Joint	Muscle	W_{CE}	W_{SEE}	W_{TRANS}	Total
Hip	GU	22	1		23
	HA	3	1	2	6
	RF			-15	-15
Knee	VA	44	5		49
	RF	2	4	15	21
	HA			-2	-2
	GA			-13	-13
Ankle	SO	17	7		24
	GA	8	8	13	29
Total		96	26	0	122

contributed to a transfer action from the knee to the hip joint. The amount of work due to transfer from knee to hip joint by HA was considerably lower than from hip to knee joint by RF (Tables 2 and 3). For the last 90 ms in jumping and sprinting, it was calculated that the absolute work for hip extension due to transfer of HA was, respectively, 2 and 3 J, while the absolute work for knee extension due to transfer of RF was, respectively, 15 and 19 J. Collectively, a net power transfer action occurred from hip to knee joint.

Also, the biarticular muscles contributed significantly to the work done at joints. This is shown by the relative work contribution due to power transfer as illustrated for jumping and sprinting in Fig. 9. For HA, the relative work contribution to the total amount of work in hip extension was, respectively, 7 and 11%. For RF, the relative contribution to the total amount of work in knee extension attained values of, respectively, 21 and 31%. Power transfer actions by gastrocnemius contributed, respectively, 25 and 28% to the total amount of work in plantar flexion. For gastrocnemius in jumping similar relative values were obtained as in the previous study on jumping (Bobbert *et al.*, 1986).

These results support the hypothesis that actions of biarticular muscles contribute to power transfer actions from proximal to distal joints during explosive leg extensions. The work done by large monoarticular muscles spanning the proximal joints is efficiently transferred to distal joints by the action of biarticular muscles. This power transfer allows high power output at distal joints despite the smaller distal muscles.

The present results are complementary to those of a previous study in which we examined the role of HA and RF in a static experiment (Jacobs and Ingen Schenau, 1992b). Subjects were instructed to generate constant forces with their foot in different directions on the ground. A very high correlation of 0.97 was found between the difference of net hip and knee moment and the difference in EMG activity of HA and RF. Hence, it was concluded that HA and RF distribute the net knee and hip moments in order to meet the specific demands of a static task. For the dynamic situation in the present study, it is further concluded that the biarticular muscles contribute to a distribution of net power during dynamic tasks.

Importance of biarticular muscles for performance

Based on the results of the present study, additional evidence was obtained for the hypothesis that biarticular leg muscles play an effective role in power transfer from proximal joints to distal joints. The question can be asked whether power transfer by biarticular muscles enhances the performance of a task. To study the influence of biarticular muscles on the performance of a task, an intervention must take place to the human musculoskeletal system. Unfortunately, this cannot be done experimentally in humans. What can be done is to make a model of the human musculoskeletal system and to optimize it to a certain objective function (see e.g. Pandey and Zajac, 1991; Soest *et al.*, 1992).

In their optimal control model for jumping, Pandey and Zajac (1991) replaced the biarticular GA with a monoarticular plantar flexor. Surprisingly, this adjustment resulted in a minor difference in performance, with the conclusion that the action of GA is not different from that of any other monoarticular plantar flexor. However, Leeuwen and Spoor (1992) argued that the moment arm of GA in the knee joint was unrealistically small in the Pandey and Zajac model. Using a more realistic moment arm of GA in the knee joint, Soest *et al.* (1992) showed that the biarticularity of GA did affect jumping performance. However, neither Pandey and Zajac (1990) nor Soest *et al.* (1992a) included an anatomical constraint (see e.g. Bobbert and Ingen Schenau, 1988) in their optimization routines. Because of this, their models allow peak knee extension velocities at full extension [see Fig. 6 of Pandey *et al.* (1990) and Fig. 4 of Soest *et al.* (1992)] which in a real situation will damage joint structures. Since no anatomical constraint has been included in optimal control models, the question of to what extent the biarticularity of muscles influences performance is still an open one.

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