

**This is an electronic reprint of the original article.
This reprint *may differ* from the original in pagination and typographic detail.**

Author(s): Finni Juutinen, Taija; Komi, Paavo; Lepola, Vesa

Title: In vivo triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump

Year: 2000

Version:

Please cite the original version:

Finni Juutinen, T., Komi, P., & Lepola, V. (2000). In vivo triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump. *European Journal of Applied Physiology*, 83(4/5), 416-426.
<https://doi.org/10.1007/s004210000289>

All material supplied via JYX is protected by copyright and other intellectual property rights, and duplication or sale of all or part of any of the repository collections is not permitted, except that material may be duplicated by you for your research use or educational purposes in electronic or print form. You must obtain permission for any other use. Electronic or print copies may not be offered, whether for sale or otherwise to anyone who is not an authorised user.

In vivo triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump

Taija Finni¹, Paavo V. Komi¹ and Vesa Lepola²

¹ Neuromuscular Research Center, Department of Biology of Physical Activity, University of Jyväskylä, 40100 Jyväskylä, Finland

² LIKES-Research Center for Sports and Health Sciences, 40100 Jyväskylä, Finland

Correspondence: Taija Finni

Department of Biology of Physical Activity
University of Jyväskylä
P.O. Box 35 (LL 227)
40351 JYVÄSKYLÄ
FINLAND

Tel: +358-14-260 2075

Fax: +358-14-260 2071

e-mail: finni@pallo.jyu.fi

Abstract

The optic fiber method was utilised to measure in vivo Achilles (ATF) and patellar tendon forces (PTF) during submaximal squat jumps (SJ) and counter movement jumps (CMJ). Normal two-legged jumps on a force plate and one-legged jumps on a sledge apparatus were performed by four volunteers. Kinetics, kinematics, and muscular activity from seven muscles were recorded. The loading patterns of the tendomuscular system differed across the jumping conditions, but were similar when the jumping height was varied. Peak PTFs were greater than ATFs in each condition. In contrast to earlier simulation studies it was observed that tendomuscular force could continue to increase during the shortening of muscle-tendon unit in CMJ. The concentric tendomuscular output was related to the force in the end of stretching phase while the enhancement of the output in CMJ as compared to SJ could not be explained with increases in muscular activity. The stretching phase in CMJ was characterised by little or no EMG activity. Therefore, the role of active stretch for creating beneficial conditions for utilisation of elastic energy within muscle was only minor in these rather submaximal performances. The modelling, as used in the present study showed, however, that tendon underwent a stretch-shortening cycle, thus having potential for elastic energy storage and utilisation. In general, the interaction between muscle and tendon components may be organised in a manner that takes advantage of the basic properties of muscle at given submaximal and variable activity levels of normal human locomotion.

Key words: muscle mechanics, tendon force, muscle-tendon interaction, power, optic fibre

Introduction

In human movement studies standing jumps with and without a counter movement have been widely explored both experimentally (e.g. Asmussen and Bonde-Petersen 1974; Komi and Bosco 1978; Gollhofer et al. 1992; Fukashiro et al. 1995) and with computer simulation (e.g. Pandy and Zajac 1991; Bobbert et al. 1996). The performance itself, with regard to jumping height and movement kinematics, as well as different aspects of neuromuscular function of the muscles involved in the tasks have been covered in the literature. For example, enhanced muscular performance during the counter movement jump (CMJ) as compared to the squat jump (SJ) has been attributed to increased myoelectrical activity (Bosco et al. 1982), recoil of elastic energy (Komi and Bosco 1978), the time available for force development (Bobbert et al. 1996) and a high force in the end of stretching phase (Zajac 1993). From these factors, recoil of elastic energy has been shown to be very important in economising the muscular performance both in animal (Alexander and Vernon 1975; Morgan et al. 1978) and human locomotion (Thys et al. 1975). In accordance with these observations Anderson and Pandy (1993) suggested that the elastic tissues improve the efficiency rather than jumping height of the CMJ as compared to SJ.

The complex multijoint nature of standing jumps raises difficulties when the function of individual muscles or muscle groups are of interest. Muscle modelling with inverse solutions has provided net joint moments (Fukashiro and Komi 1987) and estimated muscle forces (Bobbert et al. 1986). While forward dynamics models rely on measured neuromuscular signals (Zajac 1993) experimental muscle force data are scarce due to the methodological challenges involved. If muscle forces can be directly recorded, one can avoid often inaccurate estimation of changes in moment arms of a muscle during locomotion and its effect on calculated muscle forces. Furthermore, modelling studies usually simulate optimal performances while little is known about submaximal movements where one may

observe individually different movement strategies with variability in activation patterns and movement kinematics, which make calculation of forces produced by a muscle group or a single muscle even more complex.

In the course of developing a force transducer for human studies, animal experiments provided a basis for applying the buckle force transducer in recording Achilles tendon forces in men (Komi et al. 1987). Later, the method was used to measure human tendomuscular forces in natural movements such as walking, running, cycling and jumping (Fukashiro et al. 1993; Fukashiro et al. 1995; Gregor et al. 1991; Komi 1990; Komi et al. 1992). Recent development of the optic fiber technique (Komi et al. 1996) has opened up new possibilities for less invasive tendon force measurements in routine experiments (Finni et al. 1998). The fact that the insertion of the optic fibre is fast and virtually painless is a clear advantage over the more invasive buckle-type transducer technique. The optic fiber is inserted through the entire cross-section of the tendon and the tendon deformation during locomotion compresses the optic fiber inside the tendon. This compression can then be calibrated to represent the *in vivo* tendomuscular force (Komi *et al.* 1996).

In the present study the optic fiber method was utilised for studying m. triceps surae (TS) and m. quadriceps femoris (QF) loading during SJ and CMJ. Different levels of submaximal jumping performances were used to examine how the output of TS and QF muscles is modulated in order to achieve greater jumping height. Instantaneous force-length and force-velocity relationships together with muscular activation are presented in order to examine the nature of muscle action and the relationship between muscle input and output. As length changes in the entire muscle-tendon complex may not correspond to the changes in muscle fiber level, Achilles tendon and soleus muscle compartment interaction is also examined by means of modelling.

Methods

Experimental protocol. Four healthy subjects volunteered for this study (3 females: 63 ± 3 kg, 167 ± 4 cm, 1 male: 90 kg, 181 cm). The subjects were informed of all the risks associated with the study and gave their written consent to participate. The subjects were free to stop the experiment at will. The recommendations contained in the Declaration of Helsinki were followed and the ethical committee of the Central Hospital of Central Finland approved the study.

Prior to the day of measurement the subjects were introduced to the jumping activities to be performed on the sledge apparatus (Kaneko et al. 1984) and on the force plate. In the sledge condition the joint movements can be easily controlled and the contribution of hip joint is limited because subject's upper body is secured to back of the chair of the apparatus. The purpose of these measurements was also to evaluate possible alterations in jumping performance due to optic fiber insertion. Visual comparison of the EMG and ground reaction force patterns showed no disturbances by the presence of the optic fiber in the tendon.

Isometric maximal voluntary ankle plantarflexions (90° angle) were performed on the ankle ergometer (Kyröläinen and Komi 1994) where torque around the rotational axes of the pedal was measured by a piezoelectric crystal transducer (Kistler, Winterthur, Switzerland). The torque applied to the pedal was converted to force under the point of force application. Distance from one centimetre wide bar under the first metatarsal head to the rotational axis was measured as pedal lever arm length. Isometric maximal voluntary knee extensions (120° angle) were measured with the knee ergometer (Komi et al. 2000). In this machine the lever arm was equipped with strain gauge and the lever arm length could be read from inbuilt ruler. Same knee and ankle ergometer setting were used in calibration of the optic fibre force transducer.

On the measurement day, after the optic fibres were inserted, the subjects performed one-legged jumps on the sledge apparatus that had an inclination of 20° from the horizontal position. With one-legged performances the possible bilateral difference could be avoided. Furthermore, the hip joint angular displacement is reduced and easily controlled as compared to normal standing jumps. Two to five squat jumps and counter movement jumps were performed with increasing effort, thus jumping a little higher each time. The amplitude of the sledge movement in each jump was predetermined and the subjects were provided with visual feedback of their performances from a monitor in front of them. On the force plate, SJ and CMJ were repeated as normal two-legged jumps.

Optic fiber technique. The transmitter-receiver unit used in the optic fiber method contains a light emitting diode and a PIN photodiode receiver (Hewlett Packard, USA). The light signal (wavelength 820 nm) travels in the core of the plastic optic fiber and returns to the unit for conversion to the analogue signal which is sent telemetrically to the recording computer. Optic fiber as a transducer for tendomuscular forces is based on light intensity modulation. Tendomuscular loading develops tensile stress within the tendon fibres (Butler et al. 1978). This stress compresses the plastic optic fiber inside the tendon. A linear relationship has been reported between increasing loading of the tendon and the intensity of the light passing through the optic fiber (Arndt et al. 1998; Komi et al. 1996) even in maximal voluntary contractions (Finni et al. 1998).

The subjects arrived at the laboratory at least one hour before the insertion of the optic fiber took place. First, an anaesthesia cream pad containing lidocain-prilocain was placed over the skin of the calcaneal and patellar tendons of the right leg and maintained there at least for one hour. With the subject was lying in the prone position with the ankle angle secured at 90° , a hollow 19-gauge needle was passed through the right Achilles tendon two to three centimetres proximal to calcaneus.

The direction of the needle was perpendicular to the tendon. Aseptic conditions were ensured during the insertion procedure. The optic fiber, sterilised with ethylene oxide at 37°C, was then passed through the needle. By removing the needle the fiber remained in situ. After insertion, both ends of the fiber were carefully cleaned before being attached to the transmitter-receiver unit for the light intensity baseline correction. A similar procedure was used in the insertion of the optic fiber through the patellar tendon. During insertion the subject was lying in the supine position with the right knee at an angle of 120°.

Optic fiber calibration. The optic fiber was calibrated using an ankle ergometer (Achilles tendon) and the knee extension apparatus (patellar tendon) with approximately 10, 20, 30, and 40 % of maximum voluntary contraction (MVC). In each condition, the subjects maintained the predetermined force levels for few seconds. From the recorded data, the optic fiber output was related to the muscle force (F) that had been converted from the external force output (F') using the equation in Fig. 1. The optic fiber signal in volts was then converted into absolute force values according to the individual linear relationship. It is assumed in calibration with the simple model in Fig. 1 that the contribution of other plantarflexors than m. triceps surae can be neglected at low force levels. To test this assumption the calibration was repeated with percutaneous stimulation (Neuropack Four MINI) in two subjects. Stimulating electrodes were placed over the proximal end of medial and lateral gastrocnemius muscles (anodes) and over the distal end of soleus muscle (cathode). Stimulation at three to four low stimulus intensities was used in order to quantify the contribution of the 'other muscles' (flexor digitorum longus, flexor hallucis longus, peroneus, tibialis posterior) to the plantarflexor moment. The stimulation and voluntary calibration linearities were confirmed to be identical at low force levels (95 % confidence interval). In the present and earlier pilot studies the two different calibration protocols have produced identical coefficients for all subjects with forces under 20 % MVC and, in some cases, up to 40 % MVC (Fig. 1). In the pilot studies we observed the

possibility of not having identical regressions in calibration performed with voluntarily and percutaneous stimulation protocol (Fig. 2). This difference was taken as evidence that with voluntary effort the subject may activate 'other plantarflexors' depending on the used activation strategy. Therefore, in the present calibration conditions identical regression lines were taken as evidence that the 'other muscles' are not involved in plantarflexion at low force levels. Thus, the use of submaximal calibration minimises the overestimation of ATF caused by the other plantarflexor muscles. No stimulation protocol is necessary in the PTF calibration since all the force produced by the knee extensors is transmitted to the skeletal system through the patellar tendon. The calibration for both the Achilles and patellar tendon force produced a linear relationship ($0.864 < r < 0.986$).

One critical point in the calibration is the determination of the axis of rotation for ankle and knee joints. This was done both using external landmarks and rotating the distal segment of the corresponding joint on a marker table. Outlines of the leg around ankle and knee joints were first projected to the marker table at angle specific to calibration while subject contracted their leg muscles isometrically. The proximal segment was kept in place while the distal segment with attached markers was rotated. Lines drawn by the distal segment markers were used to determine a fixed centre of rotation. The moment arm distances (d , Fig. 1) measured utilising anatomical landmarks were then confirmed by measuring the distance also from the image in the marker table.

The optic fiber as a force transducer has been tested both in an animal model (Komi et al. 1996) and in human subject experiments (Arndt et al. 1998; Finni et al. 1998). During dynamic muscle work, the skin moves in relation to the tendon. To test a possible artefact of skin movement to the tendon force, the skin was pulled along the tendon near the insertion site of the optic fiber. The measured effect of skin movement was calculated to be less than 2 % of the recorded peak forces during locomotion. It is possible, however, that the effect of movement artefact becomes greater during full

ankle plantarflexion when the skin is folded around the Achilles tendon. This potential problem is not present in the patellar tendon.

Data recordings and processing. Reaction forces, sledge displacement and velocity were measured on the sledge apparatus. The sledge had an inclination of 20° and the seat on which subjects sat had a mass of 30 kg. Forces and EMG were collected with Cudas software (1 kHz) for further analysis. From the jumps on the force plate the vertical ground reaction forces (F_z) were measured and jumping height ($h = (NI / m)^2 / 2g$) was determined from the net impulse ($NI = \int [F_z(t) - mg]dt$).

Bipolar EMG electrodes (Beckmann miniature skin electrodes, Illinois, USA) were placed on the tibialis anterior, soleus (SOL), gastrocnemius medialis (GA), vastus lateralis (VL) and rectus femoris (RF) muscles of the right leg and on the soleus and vastus lateralis muscle of the left leg. An interelectrode distance of 20 mm used routinely in our laboratory conforms to the recent recommendation of SENIAM (Hermens and Freriks 1997). The positioning of the electrode was also done according to the recommendations of SENIAM (Hermens and Freriks 1997). The EMG signals were amplified (Glonner Biomes 2000, Glonner Electronic GmbH, Munich, Germany) and sent telemetrically to the recording computer. The signals were high-pass filtered (20 Hz, before sampling) and full-wave rectified. The EMG signal was integrated (iEMG) separately for the breaking and push-off phases and the iEMG was divided by the integration time to obtain average EMG (aEMG). Average EMG values were normalized to aEMG obtained in isometric calibration condition (40%) for each individual. Smooth activity patterns presented in figures resulted from time normalization and averaging of the jumping performances of the four subjects.

All performances were videotaped at 200 Hz in the sagittal plane from the subject's right side. Reflective markers were placed on the following points: on the neck at the level of the 5th cervical

vertebra, greater trochanter major, approximate centre of rotation of the knee, lateral malleolus, heel and 5th metatarsal head. These points were digitised from the video with Motus software (Peak Performance Technologies Inc., USA). The scaled co-ordinates were filtered with a 4th order Butterworth conditioner with a cut-off frequency of 8 Hz. The electrical pulse was used to synchronise the computer and video data. The length changes of the muscle-tendon complex (Δl) for the quadriceps femoris (QF) and triceps surae (TS) muscles were determined using the method of Hawkins and Hull (1990). No attempt was made to separate the loading of individual muscles contributing to ATF or PTF. A model where detailed anatomy of a muscle (pennation angles, fiber lengths) or tendon (areas corresponding to each muscle) is not taken into consideration provides a simple but, we feel, effective tool to examine muscle mechanics. Consequently, the QF muscle lengths correspond to averaged vasti muscles and the TS lengths represent the average of gastrocnemius and soleus muscles. Muscle-tendon velocities (v) were calculated by dividing infinitesimal change in muscle length with corresponding time (5 ms). Both instantaneous and average power outputs ($P = Fv$) were calculated over eccentric and concentric periods.

Changes in length of the Achilles tendon were calculated according to the method of Voigt et al. (1995) who utilised quadratic tendon force function $ATF = k(\Delta l)^2$, where Δl is change in tendon length and k is a constant which was derived from the following equation: $k = (YA_{atf}(\epsilon_{max} - \epsilon_t)10^2)/((\epsilon_{max}Lo)^2)$, where Y = Young's modulus (1.2 GPa used), A_{atf} = Achilles tendon cross-sectional area, ϵ_{max} = maximal tolerable strain (6% used), ϵ_t = length of nonlinear part of load-deformation curve (2% used) and Lo = tendon resting length. Tendon resting lengths were obtained from Yamaguchi *et al.* (1990) and Achilles tendon cross-sectional areas were determined individually by ultrasonography according to Kallinen and Suominen (1994). Tendon length changes ($\Delta l = (ATF/k)^{1/2}$) were subtracted from changes in length of the muscle-tendon complex for determination of muscle compartment length changes.

As the jumping heights varied individually only the performances with greatest jumping height were averaged across the subjects. The modulation of tendomuscular loading with increased jumping height is presented as individual recordings. Pearson's two-tailed correlation ($p < 0.05$) was employed to reveal the relationships between measured variables. T-test for paired samples was used to test differences in variable mean values between jumping conditions.

Results

Normal ground reaction force patterns (e.g. Gollhofer et al. 1992) were observed in each jumping condition. Increases in jumping height caused slight modulations in the tendomuscular force-time patterns but there was a tendency for greater peak forces that were reached later during the push-off of the higher jumping performance (Fig. 3). In one-legged jumps inclination of the sledge caused smaller reaction forces, although ATF and PTF did not differ significantly from those recorded during performances on the force plate (Table 1). In general, decreasing the hip joint angular movement (sledge condition) did not alter the TS and QF loading patterns from those observed in the force plate although the time course of the jumps was longer in the sledge condition (0.66 s vs. 0.42 s in SJ and 1.3 s vs. 0.66 s in CMJ, $p < 0.001$). Therefore the following results will focus on the normal two-legged performances.

During the CMJ, the tendomuscular force-length relationships of the QF muscle show that the force increase happened mainly during the lengthening but could continue also during shortening as seen for the subject in Fig. 4. The initial lengthening of the QF muscle occurred with very low EMG activity. This is illustrated in Fig. 5, where group mean EMG activities of the vastus lateralis muscle

are taken as a representative example. The activity of the VL muscle was greatest at long muscle lengths both during SJ and CMJ.

The force increase of the TS muscle occurred at relatively constant length both in SJ and CMJ, and coincided with EMG activity (Fig. 4 and 4). Soleus and gastrocnemius activity patterns were remarkably similar with only small inter- and intra-individual variation in the initiation of activation. The action of the TS muscle-tendon complex shown in Fig. 5 was further analysed by modelling the length changes of soleus muscle and tendon compartments. Figure 5 shows that during SJ the muscle itself shortened throughout the movement causing initial tendon lengthening. In the late push-off phase the tendon reached maximum shortening velocity before the muscle ($p > .05$). During the early downward movement in CMJ, the soleus muscle was passively lengthened. While the tendon length was constant the change in muscle length was responsible for lengthening of the muscle-tendon complex. When the muscle became active the muscle compartment started to resist lengthening simultaneously causing tendon elongation. During the push-off phase the length changes followed those seen during SJ.

The instantaneous force-velocity relationships (Fig. 7) illustrate how the concentric power output increased with increasing jumping height. Average concentric power was greater ($p < .05$) in CMJ than in SJ for the examined muscle groups although the range of muscular activity was similar in both jumping conditions (Fig. 8). An example of the TS and QF muscle power-time curves together with vertical ground reaction forces during SJ and CMJ are shown in Fig. 9. The timing of peak powers varied individually in these submaximal performances. The greatest peak powers produced by QF (1032 ± 310 W) were found during CMJ. Tendonmuscular forces in the end of the braking phase correlated positively with peak powers in the following concentric phase of CMJ (Fig. 10).

Discussion

In the present study where SJs and CMJs were performed with variable efforts the tendomuscular loading patterns changed in amplitude and duration when the jumping height was varied. The vertical ground reaction forces showed normally observed patterns and Achilles tendon loading was of similar magnitude as recorded earlier with the buckle type force transducer (Fukashiro et al. 1995). The interindividual variation found in peak forces (Table 1) was not unexpected (Finni et al. 1998), and followed observations from animal experiments (e.g. Gregor et al. 1988).

Some simulation studies have indicated that while the force development in the SJ occurs during shortening, the peak forces in CMJ are reached prior to the concentric phase. This observation has been made with the hip extensors (Bobbert et al. 1996) and with the vastus lateralis muscle (Anderson and Pandy 1993). The present results show, however, that the force development can continue during the shortening phase also in CMJ performance as shown in Fig. 4. Ingen Schenau et al. (1997) have concluded that the time for force development prior to shortening can explain the difference in muscular output in these two jumping conditions. While this conclusion can be quite valid in a maximal jumping condition, the present low level jumps may not necessarily follow the same mechanism. Naturally, the previous muscle history has an effect on the following concentric phase. For example, high force prior to shortening favours concentric muscle output (Fig. 10), which has been referred to in several studies (Bobbert et al 1996; Bosco et al. 1981; Takarada et al. 1997). Literature in human studies suggests that also increased myoelectrical activity (Bosco et al. 1982; Walshe et al. 1998) and elastic recoil (Fukashiro et al. 1995) may be operative. The present results showed that with a given EMG a higher power output was achieved in the CMJ (Fig. 8). This suggests that enhancement of muscular output cannot be solely explained with increased muscular activation. Thus, elastic recoil may have also contributed to the tendomuscular performance in the present low level jumps.

The direct comparison of muscular performances in SJ and CMJ requires that the concentric phases are identical. In the present experiment the knee joint angle amplitudes tended to increase with increasing effort in the CMJ. This is especially seen for the QF muscle in the group mean values of highest performances (Fig. 5). As the length changes in TS muscle were comparable (for some subjects also in QF) the comparison is feasible for that muscle group. Examination of Fig. 8 shows that for both TS and QF muscle groups concentric mean power output was greater in CMJ than in SJ although the range of vastus lateralis EMG activities is slightly different between jumping conditions, a fact that may have been caused by different knee angular movement.

For SSC to be effective, active lengthening of the muscle is required. An active muscle, when lengthened, may sustain high forces and stretch sufficiently the tendon that can store elastic energy to be utilised in the late concentric phase when the muscular activity starts to decay (Alexander and Bennet-Clark 1977; Lou et al. 1999). In the present study, the results from modelling suggest that at least part of the muscular lengthening occurred without or with low EMG activity. Therefore, the following possibility can be considered to describe events in the muscle-tendon complex. In CMJ, both ATF and PTF start to increase during the unweighting phase, first due to the passive stretch and then during the active braking phase (Fig. 4 and 5). According to Kawakami et al. (1998) and Narici et al. (1996) one may expect that the applied passive stretch takes up at least part of the slack in the tendon and that this passive stretch also lengthens the muscle fibres as shown for the soleus muscle (Fig. 6). Thus the muscle length, hypothetically, could be on the descending limb of the force-length curve prior to activation. Then, when activity starts the muscle shortens and moves towards the plateau of the force-length curve. Consequently, the force-producing capability of the tendomuscular complex is increased in the beginning of the activation phase. In other words, due to the counter movement of the body the muscles are being stretched, first passively and then actively

when breaking the downward movement. This could imply that the procedure takes advantage of the force-length properties of the muscle. The result that the tendon reached maximum shortening velocity prior to muscle compartment during the late push-off (modelling experiment), further suggests that interaction of the two components created beneficial conditions for muscle to act longer in the middle part of force-velocity relationship, thus having a greater force producing capacity. From the entire muscle-tendon unit point of view the tendon plays an important role for enhancing velocity and thus power in the push-off phase. As stated earlier, at least part of increase in the QF output in the push-off phase of CMJ (Fig. 7) may be achieved by an effective tendon action. It must be emphasized, however, that to refer more accurately to the contractile and tendon components in the force enhancement during the concentric phase may require more sophisticated measures of the length and velocity changes of both components.

Normally the peak joint and tendomuscular powers are reached in sequence of hip, knee and ankle (Bobbert and Ingen Schenau 1988; Fukashiro and Komi 1987; Pandy and Zajac 1991; Prilutsky and Zatsiorsky 1994). However, in the present study QF peak power was not always reached earlier than TS peak power (Fig. 9). Here the performances were very submaximal and possibly affected by bilateral loading difference, and thus the common patterns of segmental rotations were not always achieved. Furthermore, a submaximal performance can be executed with various techniques and interpretation of the present results may not be generalised to maximal performances.

When the jumping height increased the peak forces and shortening velocities tended to increase as well (Fig. 7). Thus, greater peak powers were reached in higher jumps. As expected, the greatest PTFs and peak QF powers were found during CMJ (Fig. 9). The present TS peak powers during SJ are similar to those reported by Fukashiro et al. (1995) (243 W) who measured *in vivo* ATFs with the buckle-type force transducer. Tendomuscular powers during maximal (or optimal) SJ have also

been estimated by means of muscle modelling. Prilutsky and Zatsiorsky (1994) reported muscular peak powers of 1000 W (knee extensors) and 700 W (ankle extensors) while Pandy and Zajac (1991) reported peak values of 1300 W (vasti muscles) and 600 W (gastrocnemius). Comparison of these values to the present values of 749 W (QF) and 294 W (TS) may give an idea about the degree of muscular effort in the present submaximal SJ performances.

Conclusions. The present study where SJs and CMJs were performed with variable efforts suggests that enhancement of tendomuscular output in the concentric phase of SSC cannot be solely explained by increases in muscular activity. The important role of active stretch for creating beneficial conditions for storage and utilisation of elastic energy within muscle may also be questioned in these submaximal performances where little or no activity was present during the stretching phase. However, results from modelling indicate that the tendon undergoes SSC both in SJ and CMJ. Therefore the submaximal activation of normal locomotion may not be effective for stretch-induced enhancement of muscle force to play a significant role but it is essential for creating conditions where the performance efficiency is improved by tendon action. In general, the interaction between muscle and tendon components in normal locomotion, where activity level is submaximal and varies considerably, must be taken into consideration before attributing differences in tendomuscular output to any single factor.

Figure legends

Fig. 1. Measured forces and moment arms for the calibration of Achilles tendon force and patellar tendon force. The optic fiber output was related to the muscle force (F) that had been converted from the external force output (F') using equation $Fd = F'd'$, where d = moment arm of tendon force and d' = moment arm of the foot or leg.

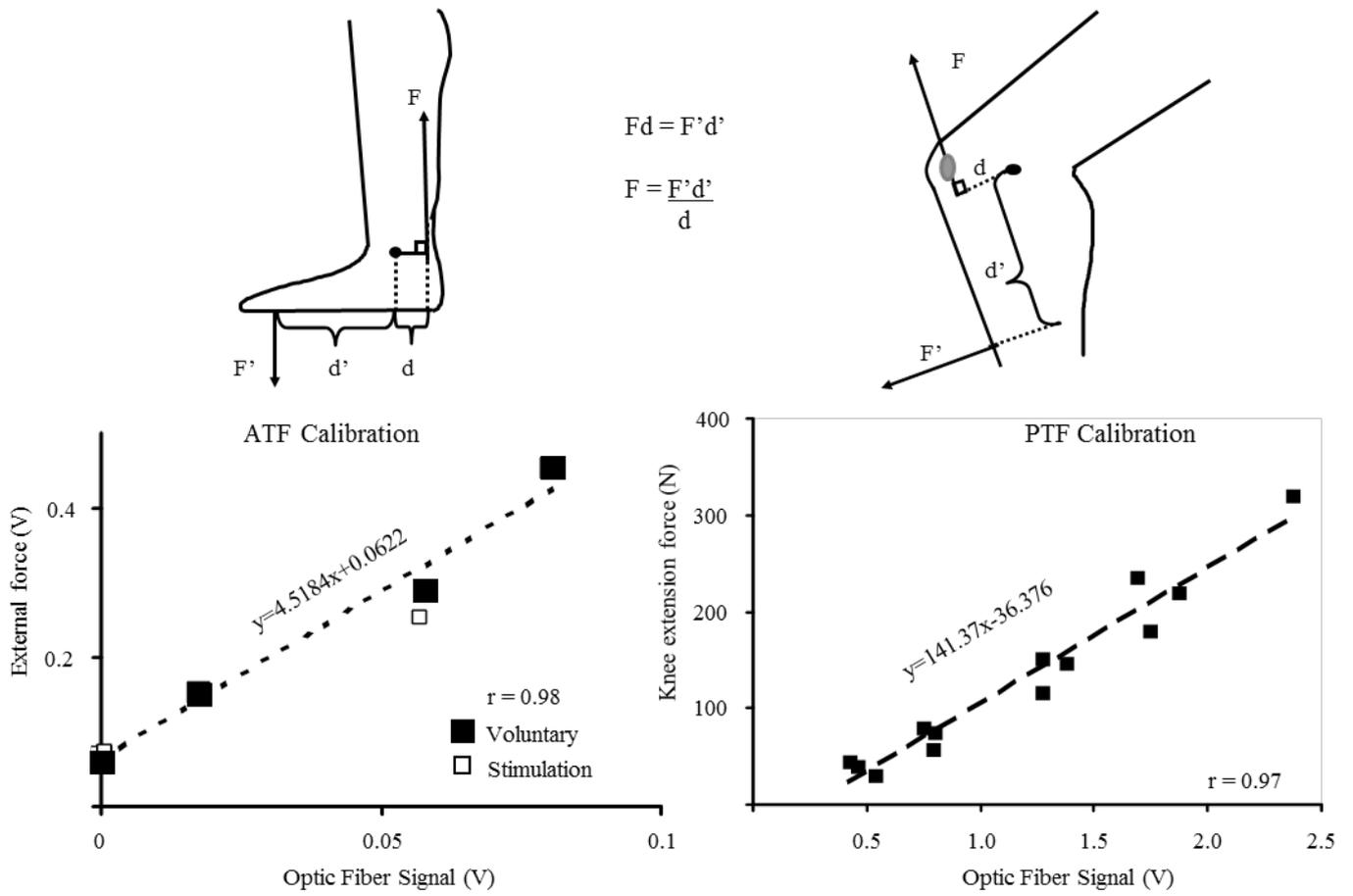


Fig. 2. Example of calibration from the pilot study. Depending on the strategy how the subjects activated their calf muscles during plantarflexion, it was possible that the use of voluntary calibration would lead to overestimation of the triceps surae muscle force. The difference between the inclinations of the regression lines can be attributed to force contribution from deeper plantarflexor muscles.

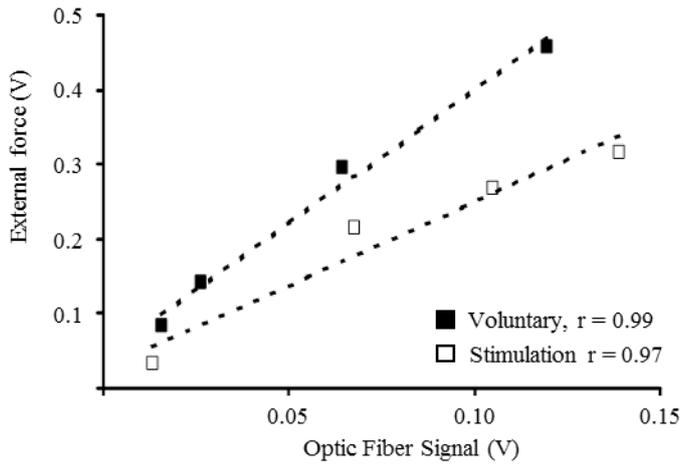


Fig. 3. Reaction forces (Fz), patellar tendon forces (PTF) and Achilles tendon forces (ATF) during squat jump (left) and counter movement jump (right). Group mean values are shown in the upper panel. In the lower panel: a representative example of loading patterns during low (thin line) and moderate intensity (thick line) performance. Arrows indicate time of take-off.

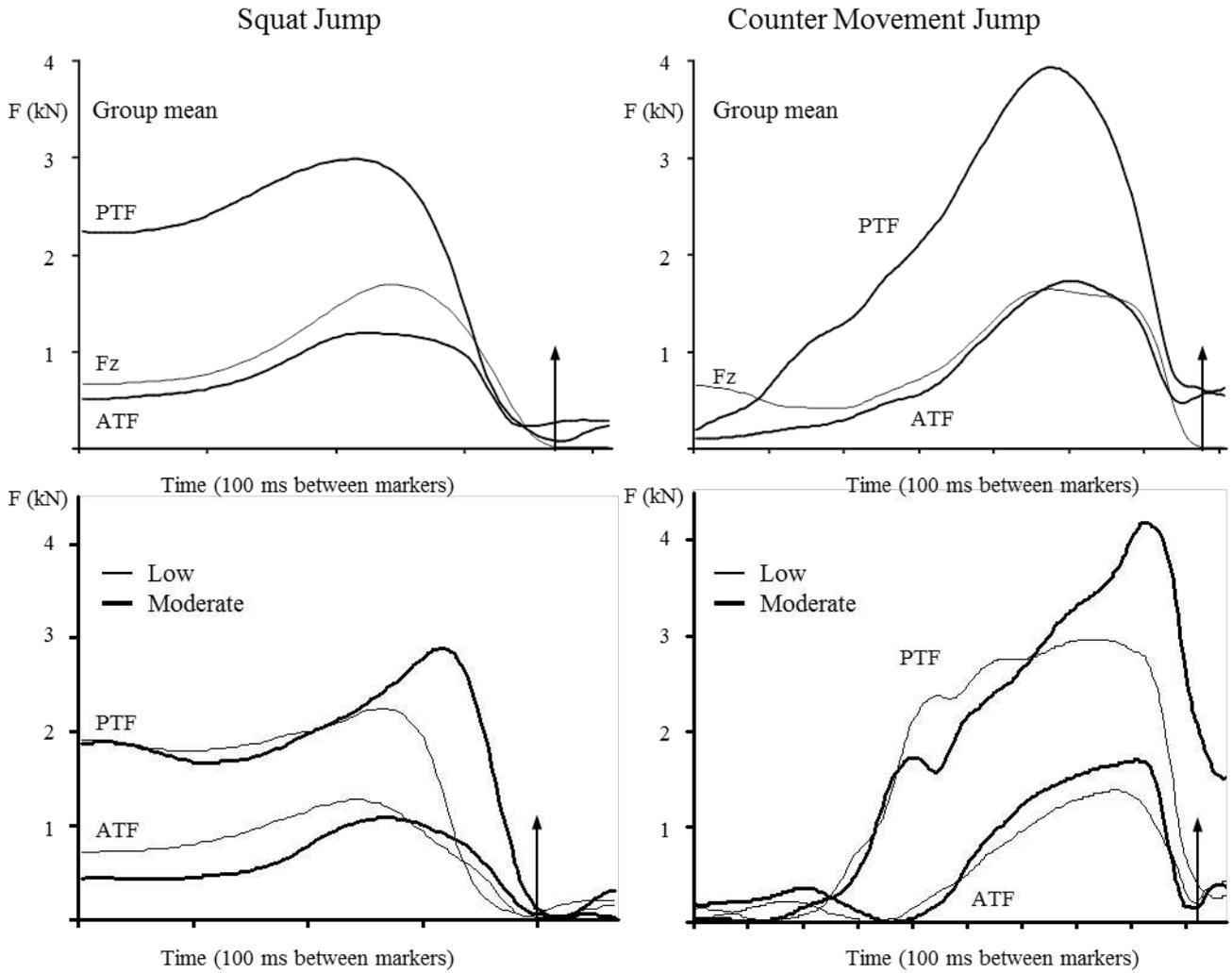


Fig. 4. Instantaneous force-length relationships for the quadriceps femoris (QF, left) and triceps surae muscles (TS, right) during SJ (up) and CMJ (bottom). Performances with low (thin line) and increased effort (thick line) are shown. Note: different force scales for ATF and PTF. On x-axis change in length is marked in intervals of 1 cm.

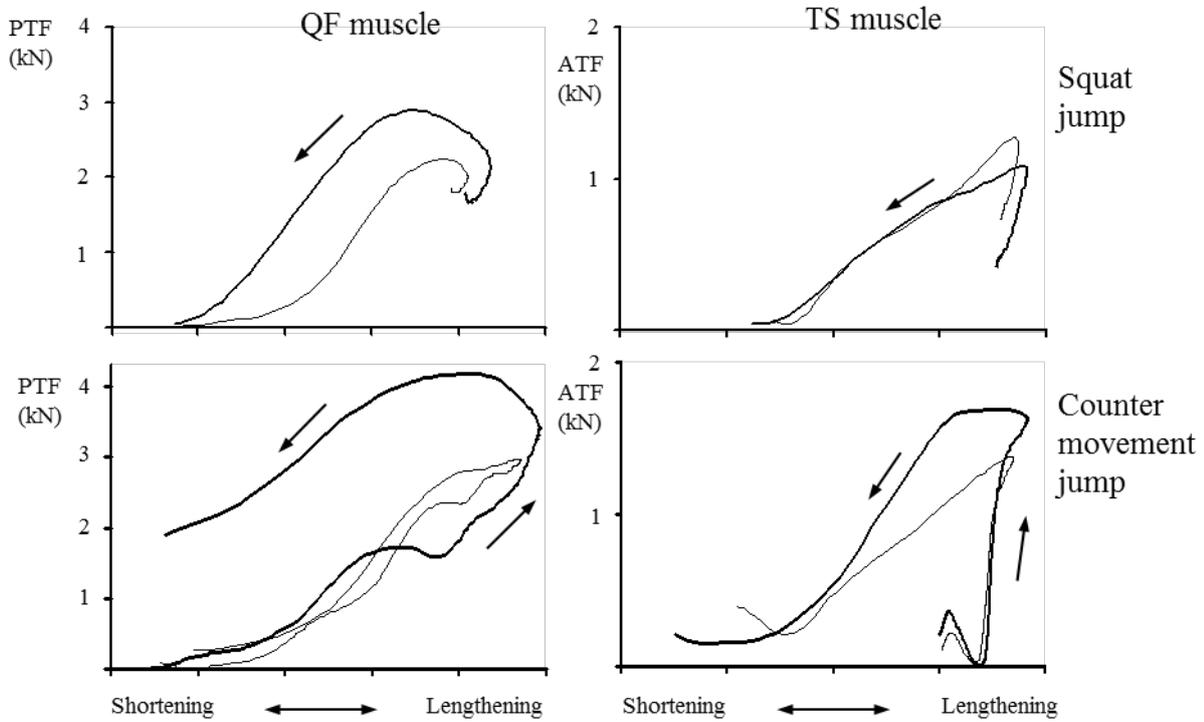


Fig. 5. The rectified and smoothed soleus (upper panel) and vastus lateralis EMG activities (lower panel) are plotted against relative length change of tendomuscular complex during SJ (dashed) and CMJ (solid). Mean values of highest performances.

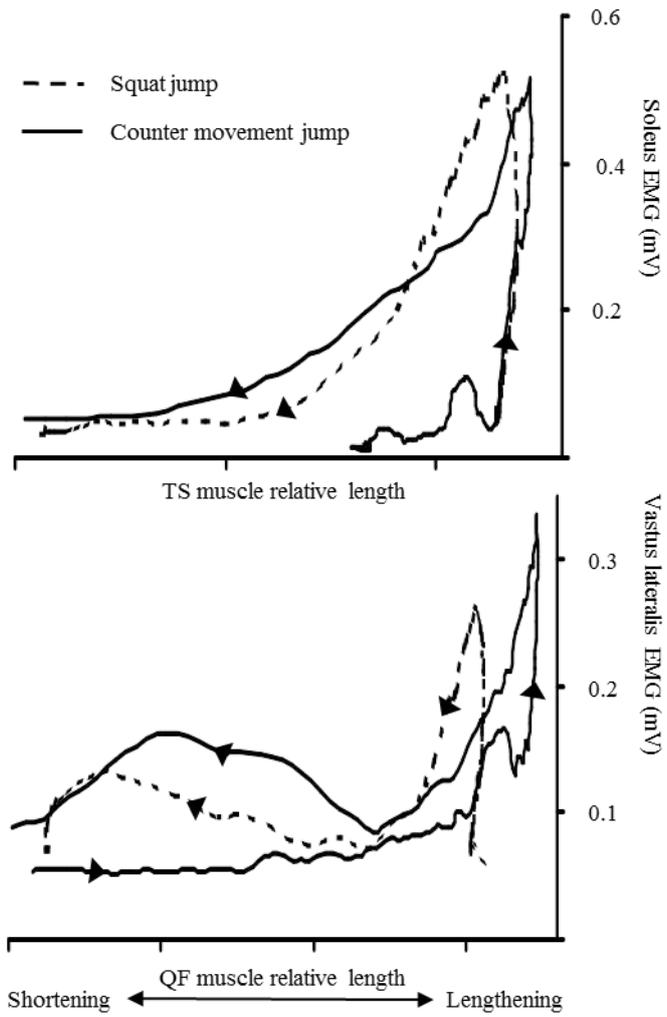


Fig. 6. Results of modelling changes in SOL muscle-tendon complex (thin), muscle (thick), and Achilles tendon compartment (circled) lengths during SJ (up) and CMJ (bottom). The rectified and smoothed muscular activities are shown as shaded area. Mean values of highest performances. Arrows indicate time of take-off.

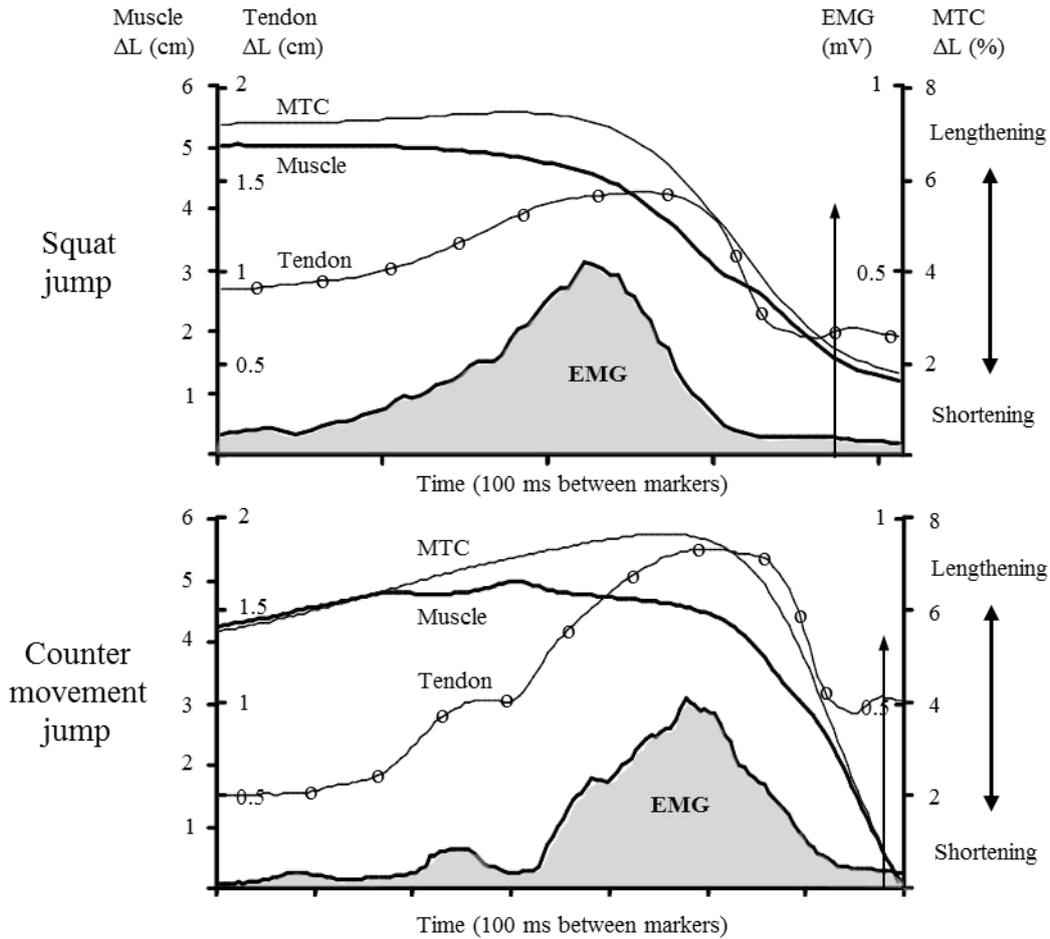


Fig. 7. Instantaneous force-velocity relationships for QF (left) and TS muscle groups (right) during low (thin line) and moderate intensity jumps (thick line). The rectified and smoothed m The rectified and smoothed muscular activity levels (dashed lines) are presented in the secondary y-axis.

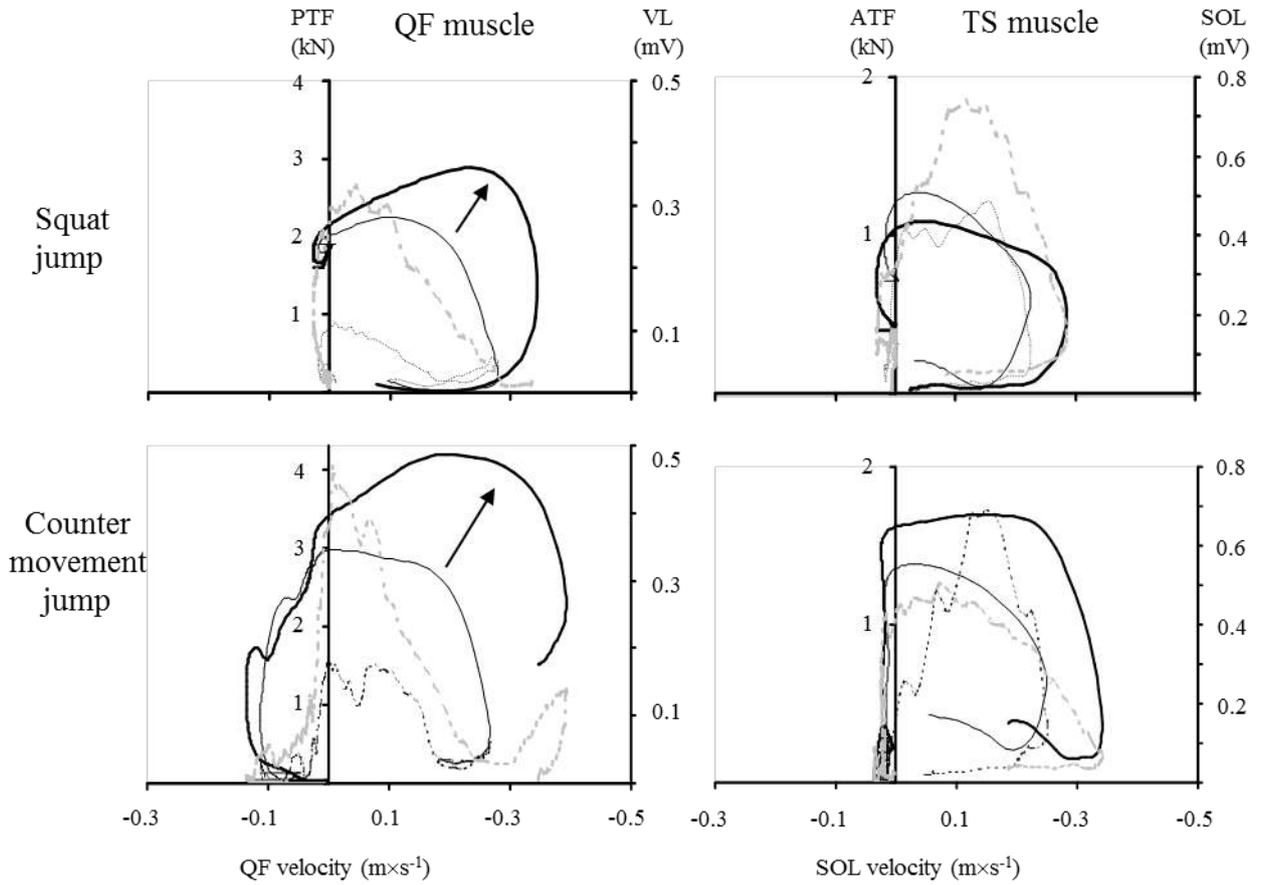


Fig. 8. Relationships between tendomuscular power output and aEMG in the concentric phase of SJ and CMJ. EMG was normalized to average EMG obtained in isometric calibration condition for each individual. Correlation values were not significant and the dashed lines merely illustrate that within the same range of EMG the concentric tendomuscular power output is higher in CMJ than in SJ.

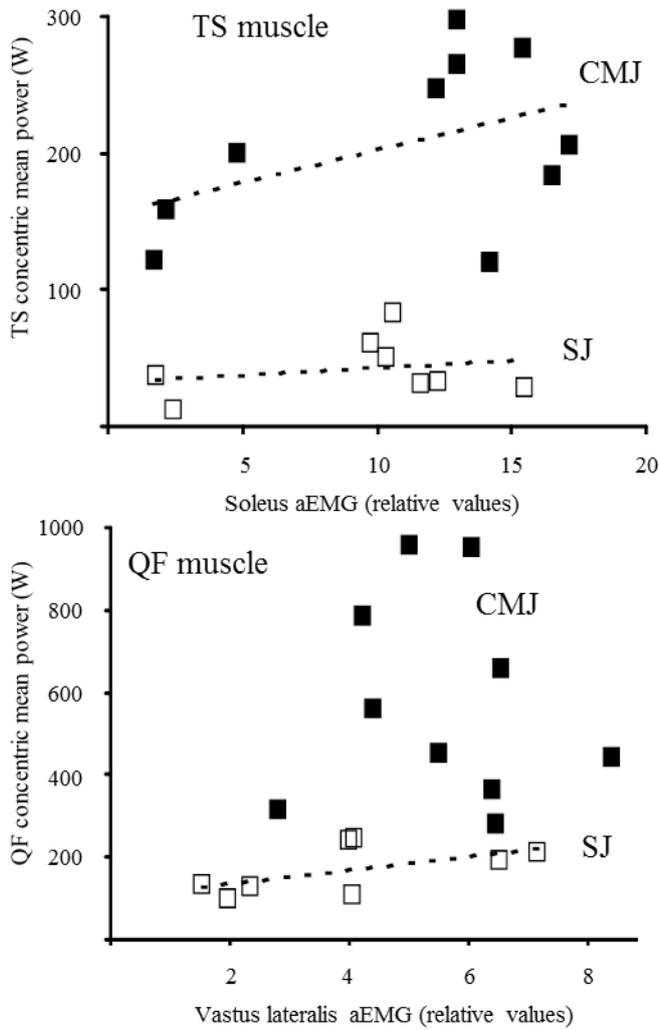


Fig. 9. Representative example of power-time curves during SJ (up) and CMJ (middle) together with average (\pm s.d.) peak powers (bottom).

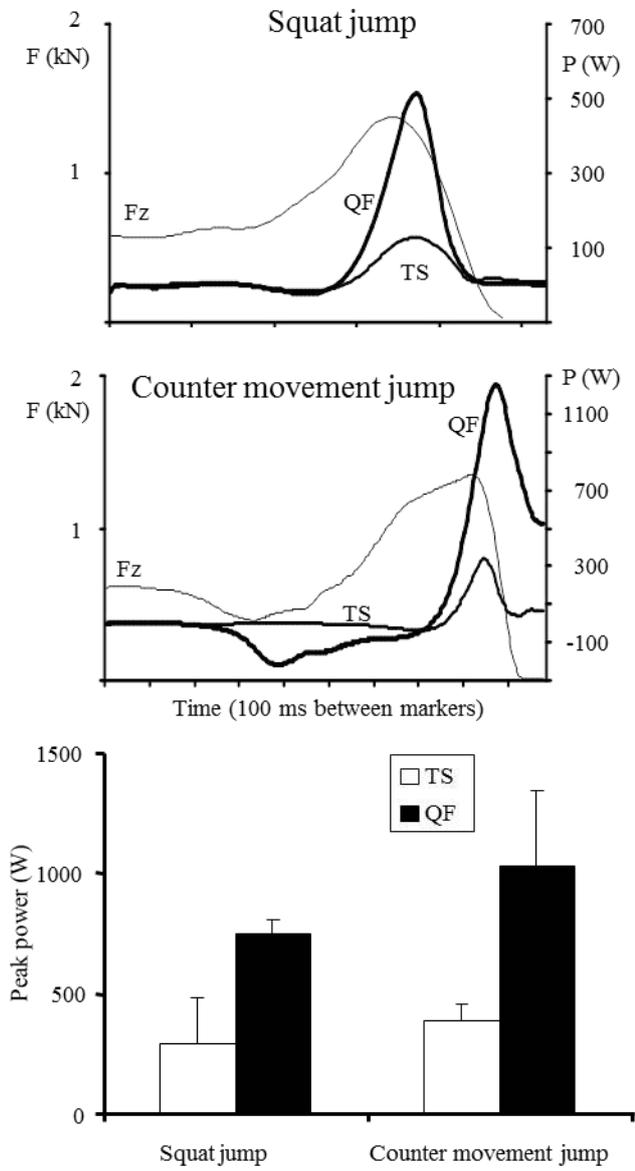
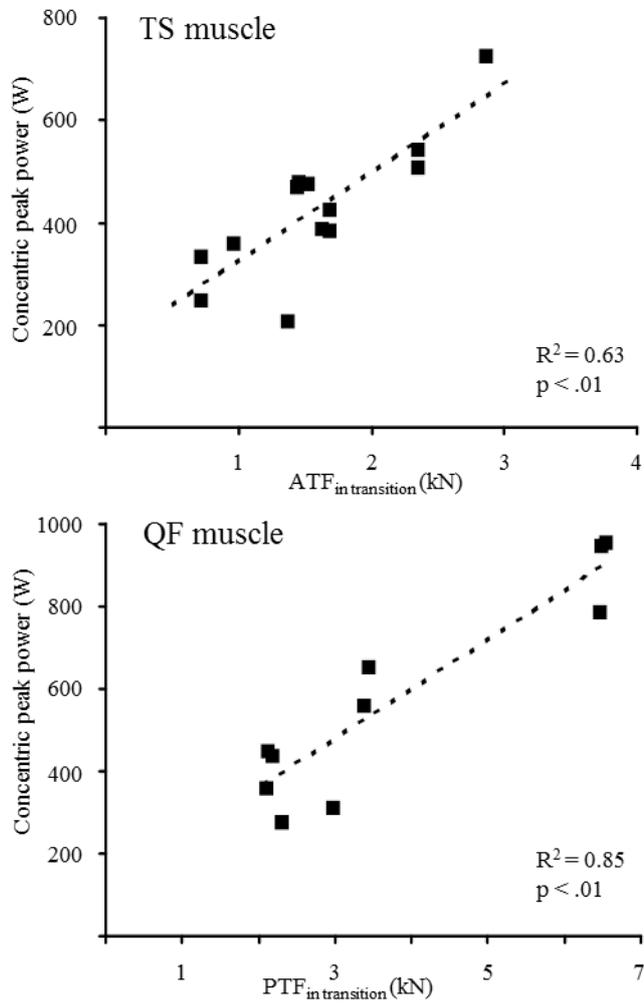


Fig. 10. Relationships between force in the end of stretching phase and tendomuscular power in the following concentric phase (n=4).



References

- Alexander RM, Vernon A (1975) The mechanics of hopping by kangaroos (Macropodidae). *J Zool Lond* 177: 265-303.
- Alexander RM, Bennet-Clark HC (1977) Storage of elastic strain energy in muscle and other tissues. *Nature* 265: 114-117.
- Anderson FC, Pandy MG (1993) Storage and utilization of elastic strain energy during jumping. *J Biomech* 26: 1413-1427.
- Arndt AN, Komi PV, Brüggemann G-P, Lukkariniemi J (1998) Individual muscle contributions to the in vivo achilles tendon force. *Clin Biomech* 13: 532-541.
- Asmussen E, Bonde-Petersen F (1974) Apparent efficiency and storage of elastic energy in human muscles during exercise. *Acta Physiol Scand* 92: 537-545.
- Bobbert MF, van Ingen Schenau GJ (1988) Coordination in vertical jumping. *J Biomech* 21: 249-262.
- Bobbert, M.F., Huijing, P.A., Ingen Schenau, G.J.van. (1986) A model of the human triceps surae muscle-tendon complex applied to jumping. *Journal of Biomechanics* 19, 11, 887-898.
- Bobbert MF, Gerritsen KGM, Litjens MCA, van Soest AJ (1996) Why is countermovement jump height greater than squat jump height? *Med Sci Sports Exerc* 28: 1402-1412.

- Bosco C, Komi PV, Ito A (1981) Prestretch potentiation of human skeletal muscle during ballistic movement. *Acta Physiol Scand* 111: 135-140.
- Bosco C, Viitasalo JT, Komi PV, Luhtanen P (1982) Combined effects of elastic energy and myoelectrical potentiation during stretch-shortening cycle exercise. *Acta Physiol Scand* 114: 557- 565.
- Butler DL, Grood ES, Noyes FR (1978) Biomechanics of ligaments and tendons. *Exerc Sport Sci Rew*, pp 125-181.
- Finni T, Komi PV, Lukkariniemi J. (1998) Achilles tendon force during walking: measurements with the optic fiber technique. *Eur J Appl Physiol* 77: 289-291.
- Fukashiro S, Komi PV (1987) Joint moment and mechanical power flow of the lower limb during vertical jump. *Int J Sports Med* 8, Suppl. 15-21.
- Fukashiro S, Komi PV, Järvinen M, Miyashita M (1993) Comparison between the directly measured achilles tendon force and the tendon force calculated from the ankle joint moment during vertical jumps. *Clin Biomech* 8: 25-30.
- Fukashiro S, Komi PV, Järvinen M, Miyashita M (1995) In vivo achilles tendon loading during jumping in humans. *Eur J Appl Physiol* 71: 453-458.

- Gollhofer A, Strojnik V, Rapp W, Schweizer L (1992) Behaviour of triceps surae muscle-tendon complex in different conditions. *Eur J Appl Physiol* 64: 283-291.
- Gregor RJ, Roy RR, Whiting WC, Lovely RC, Hodgson JA, Edgerton VR (1988) Mechanical output of the cat soleus during treadmill locomotion: in vivo vs. in situ characteristics. *J Biomech* 21: 721-732.
- Gregor RJ, Komi PV, Browning RC, Järvinen M (1991) A comparison of the triceps surae and residual muscle moments at the ankle during cycling. *J Biomech* 24: 287-297.
- Hawkins D, Hull ML (1990) A method for determining lower extremity muscle-tendon lengths during flexion/extension movements. *J Biomech* 23: 487-494.
- Hermens HJ, Freriks B (1997) The state of the art on sensors and sensor placement procedures for surface electromyography: A proposal for sensor placement procedures. Roessingh Research and Development b.v., The Netherlands, pp.125-146.
- van Ingen Schenau GJ, Bobbert MF, de Haan A (1997) Does elastic energy enhance work and efficiency in the stretch-shortening cycle? *J Appl Biomech* 13: 389-415.
- Kallinen M, Suominen H (1994) Ultrasonographic measurements of the Achilles tendon in elderly athletes and sedentary men. *Acta Radiol* 35: 560-563.
- Kaneko M, Komi PV, Aura O (1984) Mechanical efficiency of concentric and eccentric exercises performed with medium to fast contraction rates. *Scand J Sports Sci* 6: 15-20.

- Kawakami Y, Ichinose Y, Fukunaga T (1998) Architectural and functional features of human triceps surae muscles during contraction. *J Appl Physiol* 85: 398-404.
- Komi PV, Bosco C (1978) Utilization of stored elastic energy in leg extensor muscles by men and women. *Med Sci Sports Exerc* 10: 261-265.
- Komi PV (1990) Relevance of in vivo force measurements to human biomechanics. *J Biomech* 23, Suppl. 1: 23-34.
- Komi PV, Belli A, Huttunen V, Bonnefoy R, Geysant A, Lacour JR (1996) Optic fibre as a transducer of tendomuscular forces. *Eur J Appl Physiol* 72: 278-280.
- Komi PV, Fukashiro MS, Järvinen M (1992) Biomechanical loading of Achilles tendon during normal locomotion. *Clin Sport Med* 11: 521-531.
- Komi PV, Salonen M, Järvinen M, Kokko O (1987) In vivo registration of Achilles tendon forces in man. I. Methodological development. *Int J Sports Med* 8, Suppl. 3-8
- Komi PV, Linnamo V, Silventoinen P, Sillanpää M (2000) Force and EMG power spectrum during eccentric and concentric actions. *Med Sci Sports Exerc*. In print.
- Lou F, Curtin NA, Woledge RC (1999) Elastic energy storage and release in white muscle from dogfish *Scyliorhinus canicula*. *J Exp Biol* 202: 135-142.

- Morgan DL, Proske U, Warren D (1978) Measurements of muscle stiffness and the mechanism of elastic storage of energy in hopping kangaroos. *J Physiol* 282: 253-261.
- Narici MV, Binzoni T, Hiltbrand E, Fasel J, Terrier F, Cerretelli P (1996) In vivo human gastrocnemius architecture with changing joint angle at rest and during graded isometric contraction. *J Physiol* 496: 287-297.
- Pandy MG, Zajac FE (1991) Optimal muscular coordination strategies for jumping. *J Biomechanics* 24: 1-10.
- Prilutsky BI, Zatsiorsky VM (1994) Tendon action of two-joint muscles: transfer of mechanical energy between joint during jumping, landing, and running. *J Biomech* 27: 25-34.
- Takarada Y, Hirano Y, Ishige Y, Ishii N (1997) Stretch-induced enhancement of mechanical power output in human multijoint exercise with countermovement. *J Appl Physiol* 83: 1749-1755.
- Thys H, Gavagna GA, Margaria R (1975) The role played by elasticity in an exercise involving movements of small amplitude. *Pflüg Arch* 354: 281-286.
- Walshe AD, Wilson GJ, Ettema GJC (1998) Stretch-shorten cycle compared with isometric preload: contributions to enhanced muscular performance. *J Appl Physiol* 84: 97-106.
- Voigt M, Simonsen EB, Dyhre-Poulsen P, Klausen K (1995) The influence of tendon Young's modulus, dimensions and instantaneous moment arms on the efficiency of human movement. *J Biomech* 28: 281-291.

Yamaguchi GT, Sawa AGU, Morgan DW, Fessler MJ, Winters M (1990) A survey of human musculotendon actuator parameters. In Winters JM, Woo SL-Y (eds) Multiple muscle systems. Springer, Berlin Heidelberg New York Tokyo, pp 717-773.

Zajac FE (1993) Muscle coordination of movement: a perspective. J Biomech 26, Suppl. 1: 109-124.