

Institute of Biomechanics and Orthopaedics

German Sport University, Cologne

**Sprint performance in relation to mechanical
properties of the muscle-tendon unit and run-
ning track compliance**

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by

Sawas Stafilidis

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Giannitsa / Greece

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First referee: PD Dr. Adamantios Arampatzis

Second referee: Prof. Dr. Gert-Peter Brüggemann

Chair of the doctorate committee: Prof. Dr. I Hartmann-Tews

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Sawas Stafilidis

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INTRODUCTION AND OUTLINE

From a mechanical point of view the following factors may affect the sprint performance of: a) the capacities of the human musculoskeletal system and b) the mechanical properties of the sprinting track. Therefore the objectives in this thesis are to investigate the effect of the mechanical and morphological properties of the leg extensor muscle tendon unit (MTU) of sprinters on the sprint performance and in addition the effect of different sprinting track configurations on sprint performance. For these purposes three studies have been concluded.

The scope of the first study was to examine the mechanical properties of the VL tendon, VL tendon plus aponeurosis, and VL aponeurosis *in vivo*. Although numerous researchers in the past examined the mechanical properties of the tendon and aponeurosis *in vitro* and *in vivo* (Bojsen-Møller et al. 2003; Maganaris and Paul, 1999/2000; Scott and Loeb, 1995; Zuurbier et al. 1994; Muramatsu et al. 2001; Trestik and Lieber, 1993; Lieber et al. 1991) the provided results are diverging and contrasting. The contrasting results presented in the *in vivo* studies could be explained due to the different methodological techniques used in order to identify the mechanical and morphological properties of the tendon and aponeuroses. For example the estimation of the load-elongation and load-strain relationship of the serial elastic elements during a maximal isometric knee extension or plantar flexion movement could be effected by the joint rotation, the joint axis displacement relative to the dynamometer axis or the separate examination of the mechanical properties of the series elastic elements (Magnusson et al. 2001,2003; Muramatsu et al. 2001; Bojsen-Møller et al.

2003; Arampatzis et al. 2004). If corrections of the joint rotation or joint axis displacement are not implemented in the estimation of the tendon and aponeurosis elongation an overestimation can occur and this could lead to erroneous results (Muraoka et al. 2002). But the most of the methodological problems could be avoided by measuring the tendon and aponeuroses elongation and strain in the same trial. Consequently it is highly interesting to examine in the same trial the strain and elongation of the vastus lateralis (VL) tendon, VL tendon plus aponeurosis and VL aponeurosis during graded isometric knee extensions *in vivo*. For this purpose twelve sprinters participated in this study and they performed isometric maximal voluntary knee extension contractions (MVC) on their left leg. Simultaneously two ultrasound probes monitored the tendon (myotendinous junction region) and aponeurosis (muscle belly) of the vastus lateralis. The results showed statistically significant differences ($p < 0.05$) between the measured and the corrected elongation as well for the tendon (myotendinous junction point) as for the tendon plus aponeurosis (muscle belly point). Due to the correction for joint rotation the maximal measured elongation decreased. Furthermore the elongation of the tendon and tendon plus aponeurosis showed significant different values. Although the maximal estimated elongation differs the strain of both structures did not vary. Therefore for a methodological point it is easier to follow the examined cross point at the VL muscle belly than at the myotendinous junction. For example in the present study from the 28 tested subjects only at 12 the myotendinous junction at VL tendon and aponeurosis could be visualised. Furthermore another important issue for the *in vivo* estimation of the mechanical prop-

erties of the VL tendinous tissues using ultrasonography is that due to the similar strain between the human vastus lateralis tendon and tendon plus aponeurosis during maximal isometric contractions the choice of the examined cross point on the ultrasound image does not influence the estimation of the strain when the initial resting length is known. Another important finding in this study is that although an isometric task was executed the knee joint was moved (rotated and translated). Due to rotation of the knee joint the estimation of the elongation of the tendon and aponeurosis can be overestimated. In this study the correction was made due to passive motion of the knee joint mounted between 0.41 and 0.51 mm/degree for the tendon and tendon plus aponeurosis respectively. Nevertheless the knee joint motion can also be influenced by the compliance of the dynamometer and the deformation of the soft tissues of the leg. We conclude that the strains of the human VL tendon, VL tendon plus aponeurosis and VL aponeurosis, as estimated *in vivo* by two-dimensional ultrasound during a maximal isometric contraction, do not differ from each other. The estimated displacement at the cross point in the VL muscle belly is significantly greater compared to that at the VL myotendinous junction. In consequence, to obtain the elongation of the whole tendon it is necessary to multiply the strain calculated for the examined part of the tendon by the total length of the examine tissue. The maximal extension effort causes a notable knee joint rotation and a correction of the estimated elongation is necessary.

The second objective of this thesis was to examine if the sprint performance is related to the mechanical and morphological properties of the

quadriceps femoris (QF) and triceps surae (TS) muscle-tendon units. Many studies that examined sprinters and untrained controls provided evidence that the sprint performance is related to the mechanical and morphological properties of the MTU of the lower extremities (Kumagai et al. 2000; Abe et al. 2000; Abe et al. 2001; Kubo et al. 2000). For example longer muscle fascicles can exhibit higher shortening velocities and mechanical powers compared to the shorter one. Moreover, the force-length relation of longer muscle fascicles have a wider range of force development compared to short muscle fascicles which can affect the muscles performance (de Haan et al. 2003). Additionally the tendon mechanical properties like elongation, strain or energy storage capacity can play an important role in the sprint performance (Cavagna et al. 1971; Kubo et al. 2000, Stafilidis et al. 2005). The contraction mechanics of the contractile component can be influenced by the non rigidity of the tendon and aponeurosis. Nevertheless, it exists not a clear position if the mechanical properties of the lower extremities MTUs (e.g. tendon compliance) affect the sprint performance by homogenous group of sprinters. Therefore based on the literature findings we can suggest that i) the sprint performance can be affected by the morphological and the mechanical properties of the lower extremities MTUs and ii) the morphological and the mechanical properties of the MTUs together could explain the sprint performance to a higher extent than separately.

For that purpose 28 sprinters participated in this study which were further divided in two groups (slow = 11; fast = 17). All subjects performed maximal isometric knee extension and plantarflexion contractions on a dyna-

momenter at 11 different MTU lengths. Parallel the activation of triceps surae (TS) and quadriceps femoris (QF) during the maximal voluntary contractions (MVC) was estimated by surface electromyography. The mechanical and morphological properties of the vastus lateralis (VL) and gastrocnemius medialis (GM)-MTUs were measured by means of ultrasonography. The results showed that the maximal joint moments at the ankle and knee joints and the morphological properties of the GM and VL showed no statistically significant differences between groups ($P > 0.05$). The fast group exhibited higher maximal elongation in the VL tendon and aponeurosis during the MVC ($P < 0.05$). Furthermore, the maximal elongation of the VL tendon and aponeurosis showed a significant negative correlation to the 100 m sprint time of all examined sprinters ($r = -0.567$, $P = 0.003$). Further the lack of difference in the maximal joint moment's at all joint angle configurations supports that there were no differences in the working ranges (widths) of the force length relationships of the triceps surae and quadriceps femoris muscles between groups. Furthermore it can be assumed that also the other synergistic muscles of the triceps surae and quadriceps femoris will have similar fascicle lengths at both groups. On the other hand the maximal elongation and strain of the VL tendon and aponeurosis showed significant greater ($P < 0.05$) values for the fast sprinter group than for the slow one. This higher maximal elongation of the VL tendon and aponeurosis could positive influence the sprint performance since at maximal velocities the extensor moment acting on the knee joint could reach greater values than in a isometric contractions due to the eccentric contraction of the of the quadriceps femoris muscles in the first

part of the ground contact phase (*Belli et al., 2002; Mann and Sprague 1980*). During the first part of the contact phase the muscle tendon unit lengthens and furthermore elastic energy is stored in the serial elastic elements (tendon and aponeurosis). Following that, at the second part of the contact phase the MTU shortens and the stored elastic energy of the tendon and aponeurosis can be reutilised (*Hof et al. 2002*). Moreover at the second part of the contact phase the muscle fibres shortening velocity is lower than of the whole MTU due to the additional shortening of the tendon and aponeurosis (*Hof et al. 2002*). Concluding it can be suggested that the higher elongation of the VL tendon and aponeurosis of the fast group benefits the energy storage and return as well as the shortening velocity of the MTU. Also the most mechanical and morphological properties of the lower extremities MTUs can not explain the performance difference of the two homogenous sprinter groups.

The aim of the third study was to examine if the sprint performance (time interval at given distance) could be affected by a track compliance-dependent manner. In literature it is well documented that in several sport events the interaction between the athlete and non rigid surfaces has a considerable effect on performance (*Bosco et al., 1997; Arampatzis and Brüggemann 1998; 1999; 2001; Kerdok et al., 2002; Arampatzis et al., 2004a*). In sprint events it is generally believed that very hard tracks improve the sprint performance (time at a given distance). On the other hand, model predictions (*McMahon and Greene 1979*) suggest that running speed is enhanced on tracks of an intermediate compliance compared to the hard ones and experimental findings propose that jumping

height and energy rates of the humans are greater on a soft sprung surface when compared to a hard one (Arampatzis et al., 2004b). Therefore it could be possible that sprint performance can be maximised on compliant tracks. However to our knowledge there is no experimental study examining the influence of track compliance on sprint performance. According to the literature (McMahon and Greene 1979; Bosco et al., 1997; Kerdok et al., 2002; Arampatzis et al., 2004a) we can expect: a) that by modifying the track compliance, sprinters would adjust their leg mechanics in order to achieve a more advantageous stretch-shortening cycle during the ground contact phase in a track compliance-dependent manner and b) to observe a compliance-dependency of the energy storage and return amplitude by the tracks during the ground contact phase. Basing on these expectations we predicted that sprint performance would be affected by track compliance with better values on the more compliant tracks. For that purpose 10 sprinters executed maximal sprints (60 m) on three different track configurations (hard, soft, spring). Also for that aim a single-lane running surface 60 m in length was constructed of 1.2 cm thick plywood boards. The boards were placed to a 60 × 0.6 m wooden chipboard frame which served as base surface. The kinematics of the body was recorded using the Vicon 624 system with 12 cameras operating at 250Hz. Four Kistler force plates (1250Hz) were used to record ground reaction forces. Double photocell sensors were positioned along the running lane to obtain the time intervals. The results showed that the time intervals from the examined sprinters had statistically no significant ($P=0.81$ to $P=0.943$) differences between the 3 track configurations. Similar sprint velocity, step

length, contact time, stride frequency and duty factor analysed at 60 m area has not been affected due to the different track configurations ($P=0.145$ to $P=0.908$). Furthermore the statistical comparisons of the discrete values every 10% of the contact phase of the three dimensional kinematics at the ankle and knee joint showed no statistically significant ($P=0.081$ to $P=0.992$) track effect. The ground reaction forces and the resultant ankle and knee joint moments have also not been affected by the track configurations. The ground reaction forces parameters, the horizontal and vertical impulses did not show any statistically significant ($P=0.15$ to $P=0.993$) differences between the three different tracks. Similar for the kinematic characteristics the statistical comparisons of the discrete values every 10 % of the contact phase showed no statistically significant ($P=0.083$ to $P=0.997$) track effect. Several possible explanations for the lack of influence of the track compliance on the sprint performance could be proposed. For example the kinematics and kinetic characteristic at the lower extremities during sprinting on the three tracks showed no statistically significant differences. These results indicate that the athletes are having the same strategy not making any specific adjustment in the leg mechanics at all examined tracks. Although the analysis was three dimensional, the values of this study in the sagittal plane are comparable to the data from two dimensional analyses (Stefanyshyn and Nigg 1998; Hunter et al 2004; 2005). Another point is that the similar kinematics at the ankle and knee joints suggests that the contraction velocity of the triceps surae and the monoarticular vasti muscle-tendon units would show similar values (since the length changes of the muscle tendon units depend on the

changes at the joint angles). Furthermore the maximal vertical ground reaction force while sprinting on the three different compliant tracks was about 2600 N. Therefore the expected maximal deformations of the examined tracks were 0.63 mm, 2.4 mm and 4.02 mm for the hard soft and spring surfaces respectively. These deformation values of the examined tracks compared to other studies are relatively low. For example, by walking ~35 mm (Marigold and Patla, 2005), running ~30 mm (Kerdok et al., 2002; Ferris et al., 1999) and hopping 26 to 146 mm (Farley et al., 1998; Ferris and Farley 1997; Moritz and Falrey 2005). Accordingly it is possible that the differences in track compliance in our study were not large enough to cause clearly diverging adjustment in leg mechanics. Additionally the similar moment values at the ankle and knee joint while running at the three different tracks indicates that the energy exchange in the series elastic element of the ankle and knee extensors as well as the work of their contractile element would be the same (Hof et al., 2002). Therefore it can be concluded that the function of the triceps surae and vasti muscle tendon units while sprinting on the examined tracks remained unaffected. Another reason for the lack of differences might be that the energy exchange between sprinters and tracks during the contact phase was too small to have a direct effect on sprint time. The expected maximal energy recovery for the hard, soft and spring tracks during the contact phase while sprinting was 0.6, 2.4 and 5.5 Joule (for the Hard, Soft and Spring surface respectively). On the other side the energy recovery values only from the Achilles tendon during the contact phase while submaximal running (~4.0 m/s, Hof et al., 2002) or hopping (Lichtwark and Wilson 2005) were between 33 and

38 Joule. These differences in the amount of energy storage and return between the surfaces and the series elastic elements suggest that the effect on the sprint performance might be marginal. Concluding the compliant surfaces (deformation by 2600 N from 0.63 mm to 4.02 mm) did not unambiguously affect the sprint performance (time intervals until 60m). These could be resulted a) due to the small amplitude of the track deformation b) the lack of benefit from the energy storage and return of the track. The expected magnitude of the energy storage in the track is small relative to the other connective tissues of the human system.

2 **FIRST STUDY** : STRAIN AND ELONGATION OF THE VASTUS LATERALIS

APONEUROSIS AND TENDON *IN VIVO* DURING MAXIMAL ISOMETRIC CONTRACTION

Savvas Stafilidis

Kiros Karamanidis

Gaspar Morey-Klapsing

Gianpiero DeMonte

Gert-Peter Brüggemann

Adamantios Arampatzis

German Sport University of Cologne, Institute of Biomechanics and Orthopaedics

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2.1 Abstract

The strain and elongation of the vastus lateralis (VL) tendon, tendon plus aponeurosis, and aponeurosis were examined during maximal voluntary contractions on a Biodex-dynamometer (knee angle: 115°, hip angle: 140°) in twelve sprinters. Following a warm up phase the subjects were instructed to perform a gradual maximal knee extension and hold it for about 3s. The kinematics of the leg were recorded using a Vicon 512 system with 8 cameras operating at 120 Hz. Ultrasonography images were taken simultaneously from the VL myotendinous junction and the mid lateral part of the VL muscle belly. During the maximal isometric knee extensions the knee joint rotated ($13.6 \pm 5.9^\circ$), leading to an overestimation of the elongation of the tendinous tissues. After correcting for this, the maximal elongation of the VL tendon examined at the myotendinous junction was lower ($p < 0.05$) than the maximal elongation of the VL tendon plus aponeurosis examined at the muscle belly (15 vs. 27 mm respectively). The maximal estimated strains of the tendon, tendon plus aponeurosis, and aponeurosis showed no statistical differences (8 ± 2 , 8 ± 1 , and $7 \pm 2\%$ respectively, $p > 0.05$). It is concluded that the strains of the human VL tendon, VL tendon plus aponeurosis and VL aponeurosis, as estimated *in vivo* by two dimensional ultrasound during maximal isometric contractions, do not differ from each other. The displacement measured at a cross point in the VL muscle belly is significantly greater compared to that measured at the VL myotendinous junction.

2.2 Introduction

Tendons transfer the force elicited by the contractile element to the bones. During the force transfer, tendons can store and release energy (Ker et al., 1988) and contribute to the efficiency of locomotion (Alexander & Bennet-Cark, 1977; Alexander, 1991). Many researchers have examined the mechanical properties of tendons and aponeuroses *in vitro* and *in vivo* (Bojsen-Møller et al., 2003; Maganaris & Paul, 1999/2000; Scott & Loeb, 1995; Zurbier et al., 1994). Research pertinent to the mechanical properties of tendons and aponeuroses has produced varying results. *In vivo* studies presented by Magnusson et al. (2003) have shown the strain of the tendon to be greater than the strain of the aponeurosis. On the other hand Maganaris & Paul (2000) reported a greater strain for the aponeurosis than for the tendon. Other researchers found similar strain values among tendons and aponeuroses *in vivo* (Muramatsu et al., 2001). Earlier *in vitro* studies reported similar strains along tendons and aponeuroses (Scott & Loeb, 1995; Trestik & Lieber, 1993) but also differing strain values have been reported (Lieber et al., 1991).

These contrasting findings, especially in the *in vivo* studies, may be explained by the different approaches governing the experiments. The examination of the mechanical properties is associated to several methodological problems concerning the estimation of the load-elongation, and load-strain relationship. During maximal isometric knee extension or ankle plantar flexion efforts, a substantial amount of joint rotation occurs (Magnusson et al., 2001,2003; Muramatsu et al., 2001; Bojsen-Møller et al.,

2003; Arampatzis et al., 2004). When this rotation is not considered, the elongation of the tendon and tendon plus aponeurosis can be overestimated (Muraoka et al. 2002) leading to erroneous conclusions concerning their mechanical properties. Another critical issue is the displacement of the joint axis relative to the dynamometer axis during isometric contractions and the consequent differences between the measured and the resultant joint moment (Arampatzis et al. 2004). Most *in vivo* experiments comparing tendons and aponeuroses have examined the respective strains during two separate trials at a given measured moment (Maganaris & Paul 2000; Magnusson et al. 2001; Kubo et al. 2000a). The resultant moments of two separate contractions having identical measured moments can be different because of differing axis misalignments. This would influence the comparison of strain and elongation between tendon and aponeurosis. Further, when separate contractions are needed to estimate the strain of the tendon and the aponeurosis, the creep effect (Wang & Ker, 1995; Schatzmann et al., 1998; Maganaris 2003) could influence the measured elongation of the examined tissue. Most of the methodological drawbacks can be overcome by measuring the elongation of the tendon and aponeurosis synchronously during the same trial. Therefore the aim of the present study was to examine the strain and elongation of the vastus lateralis (VL) tendon, VL tendon plus aponeurosis and VL aponeurosis during graded isometric knee extensions *in vivo*. To overcome the cited drawbacks the ultrasonography images of the tendon myotendinous junction and of the aponeurosis of the vastus lateralis were recorded simultaneously.

2.3 Methods

2.3.1 Calculation of the knee joint moment

Twelve sprinters participated in the study. The means and standard deviations of their age, body mass, and height were 20.2 ± 3.7 years, 77.4 ± 5.6 kg and 182 ± 7 cm, respectively. The subjects performed isometric maximal voluntary knee extension contractions (MVC) of their left leg. They were seated with hip and knee angles set at 140 and 115° respectively. The hip and knee joint angles were defined to be at 180° in the straight prone position, decreasing values indicating flexion. Knee extension moments were measured by a dynamometer (Biodex Medical Systems, Inc. USA). The tibia was secured to the lever arm of the dynamometer with inextensible Velcro straps. The trunk of the subjects was also secured (hip angle = 140°) to prevent excessive hip motion. Prior to the contraction the knee joint axis and the axis of the dynamometer were carefully aligned. Although efforts were made to limit any joint movement during the contraction, the knee joint changed its position. The angle of the knee joint increased and its axis did not stay aligned with the axis of the dynamometer. This relative motion between knee and dynamometer axis, significantly influenced the resultant joint moments (Arampatzis et al. 2004).

Therefore we monitored the knee and hip joint angles and position using a motion analysis system (Vicon 512, Vicon Motion Systems, United Kingdom) with 8 cameras operating at 120 Hz. For that purpose, reflective markers were placed on following anatomical landmarks: Lateral and me-

dial malleolus, most prominent points of the lateral and medial femoral condyles, trochanter major, Crista iliaca on the Tuberculum iliacum, axis of the dynamometer and the lever arm of the dynamometer. The axis of rotation of the knee joint was defined to be parallel to the axis of the dynamometer and passing through the midpoint of the line connecting the lateral and medial condyles of the femur. Prior to the measurement the subjects performed a warm up phase consisting of 3 slow gradual contractions up to maximum. After the warm up the subjects were instructed to gradually increase the exerted torque to reach the isometric maximal value in about 4 seconds and then hold it for about 3 seconds. The exerted moments measured by the dynamometer were registered synchronously to the kinematics by the Vicon-system at a sampling rate of 1080 Hz. The resultant moments around the knee joint were calculated according to Arampatzis et al. (2004).

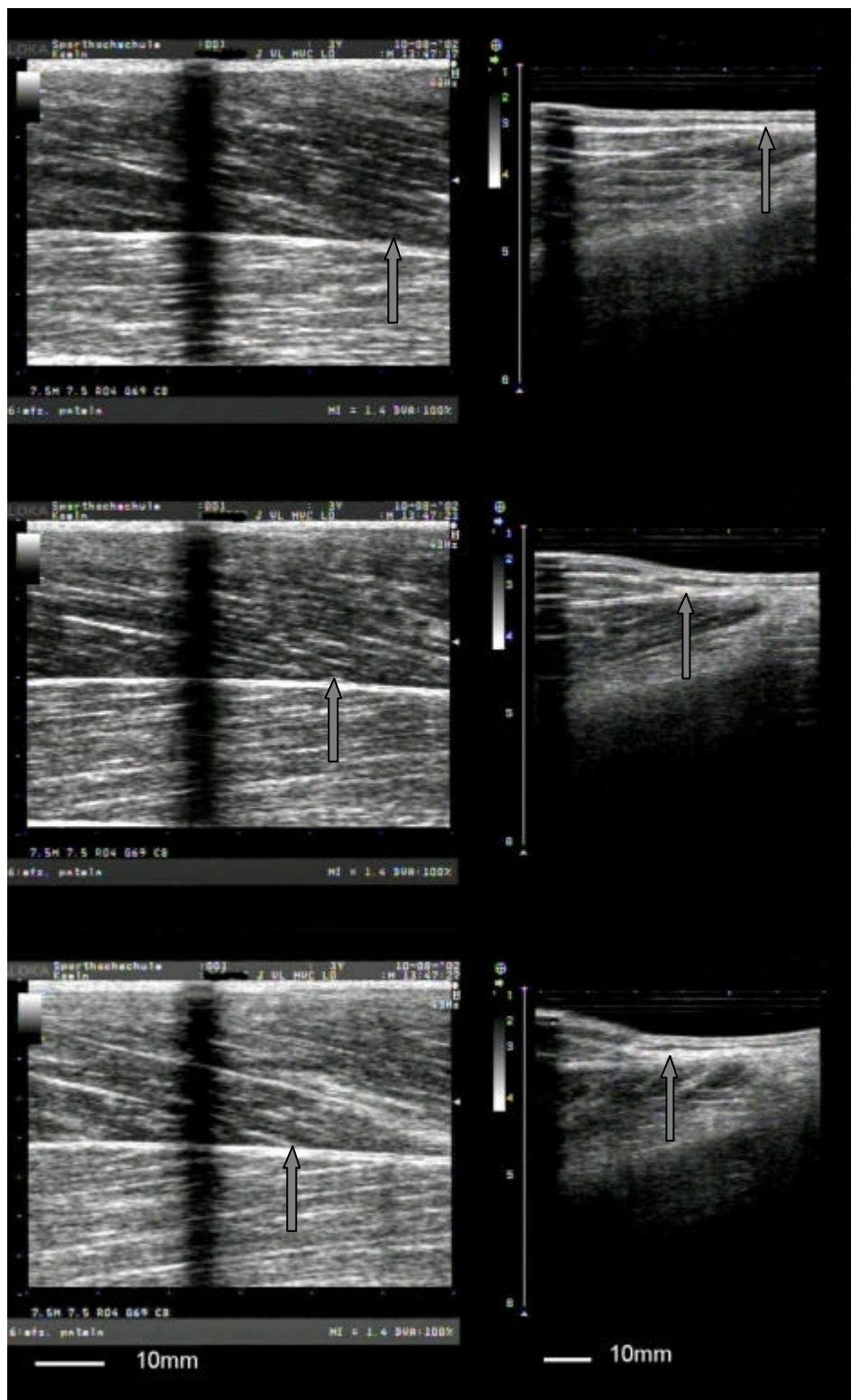


Figure 1. Ultrasound images at the VL myotendinous junction (left) and at the VL muscle belly (right) at 0, 50 and 100% of the resultant knee joint moment. A marker was set between the probe and the skin to prevent any motion of the probe in relation to the skin. Arrows are showing the analysed crosspoint.

2.3.2 Measurement of tendinous tissue elongation

Two 7.5 MHz linear array ultrasound probes (Aloka SSD 4000 and Shimadzu, JPN, SDU-350 XL) were used to visualise the tendon (myotendinous junction region) and aponeurosis (muscle belly) of the vastus lateralis (Fig. 1). To measure the elongation of the aponeurosis the ultrasound probe was placed over the muscle belly of vastus lateralis midway between trochanter major and epicondylus lateralis. The second ultrasound probe was placed over the most distal part (myotendinous junction) of the vastus lateralis. To overcome unevenness of the skin due to muscle contraction, a silicon pad (thickness 2 cm) was placed between probe and skin. The ultrasound probe was embedded in a rubber foam, which was anatomically designed specifically for this purpose and fixed to the leg using elastic stripes. Markers that could be identified on the ultrasound image were fixed to the skin to examine any motion of the probes relative to the skin during the contraction (Fig. 1). In the literature the *in vivo* elongation of the human myotendinous junction region has often been referred to as the elongation of the tendon (Maganaris and Paul, 2000; Muramatsu et al., 2001). Therefore to avoid any conflicts with the literature from now on in the text the displacement of the myotendinous junction region is referred to as elongation of the tendon and the displacement of the cross-point at the muscle belly (Fig. 1) as elongation of the tendon plus aponeurosis. The zone between the myotendinous junction point and the cross point at the muscle belly is referred to as aponeurosis.

A synchronising box (Peak performance technologies) was utilised to synchronise all systems. A manually triggered TTL signal was captured by the Vicon system. Simultaneously a dot was displayed on both ultrasound image sequences which were recorded on digital video tapes and stored for further analysis. It has been reported in the literature, that during maximal knee extensions a notable joint rotation occurs which can significantly influence the estimation of the elongation and strain of tendon and aponeurosis (Bojsen-Møller et al., 2003). Therefore knee kinematics were recorded during a passive rotation of the knee whilst the two ultrasound probes took images of both examined areas (myotendinous junction and muscle belly). This allowed to calculate the displacement at the myotendinous junction point and at the muscle belly point due to joint rotation during the maximal voluntary contractions. The measured displacement at these two points during the contraction of the knee extensors, was corrected by the above cited data. The difference in elongation between the values measured during the maximal voluntary contraction and the values due to joint rotation represent the elongation of the tendon and tendon plus aponeurosis due to the exerted force (corrected elongation). This means that the elongation of the tendon (myotendinous junction) or the elongation of the tendon plus aponeurosis (muscle belly) examined at the present study corresponds to the elongation of all structures distal to the analysed points. Precisely the elongation examined at the myotendinous junction includes the patella tendon, fibres of the quadriceps tendon over the anterior aspect of the patella (Staeubli et al. 1999b; Dye et al. 2003) and the VL tendon, whereas the elongation examined at the muscle belly

include the elongation of all above structures plus a portion of the distal aponeurosis of the vastus lateralis.

The analysis of the ultrasound images was done using a video analysis software (Simi Motion 5.0, SIMI Motion GmbH). The kinematic and ultrasound image data were interpolated using quintic splines (Engels-Müllers and Reutter, 1991) to achieve a common frequency (1080 Hz) with the moments. To estimate the strain of tendinous tissues, the resting length of these structures is required. In order to obtain this information, the subjects were seated on the dynamometer with hip and knee angles set at 140° and 115° respectively. The distance along the curved path along the skin from the tuberositas tibia (defined as origin of the patella tendon) to the markers on the skin was measured using flexible measuring tape. The distance between the skin markers and the analysed cross points was measured on the ultrasound images. The resting length of the aponeurosis corresponds to the difference between the resting lengths of the tendon and tendon plus aponeurosis.

The ultrasound images taken during the passive motion and during the maximal extension effort were digitised frame by frame until the maximal extension moment was achieved. The displacement at the myotendinous junction and the displacement at the muscle belly were corrected by considering the knee joint rotation observed during the knee extension effort. Then the data were analysed in steps of 10% of the maximal achieved moment up to 100%. All presented values correspond to the mean values at every step. So the values for 10% are the mean values from 0 to 20%, the values for 20% are the mean values from 10 to 30% and so on. At 0%

and 100% the values correspond to the exact values at the beginning of the contraction and at the maximal achieved resultant moment. A t-test for two dependent samples was used to identify differences between tendon and tendon plus aponeurosis in elongation and strain. The level of significance was set at $p < 0.05$. Since the reproducibility of this method was demonstrated in several studies in the past (Maganaris & Paul 1999, 2000; Bojsen-Møller et al. 2003) we examined the accuracy of the digitising procedure by having different examiners digitising several trials. The differences ranged from 1.0 to 1.2 mm (resolution of the ultrasound image 0.7 mm).

2.4 Results

We found statistically significant differences ($p < 0.05$) between the measured and the corrected elongation as well for the tendon (myotendinous junction point) as for the tendon plus aponeurosis (muscle belly point, Fig. 2). After correcting for joint rotation, the maximal measured elongation of the tendon and tendon plus aponeurosis decreased in average 6 ± 3 and 7 ± 3 mm respectively. Although special care was taken to prevent any changes in joint angle during contraction, the maximal rotation of the knee joint was $13.6 \pm 5.9^\circ$. These values are in accordance with the values reported in the literature during knee extension contractions (Kaufmann et al., 1995; Arampatzis et al., 2004). The correction factors obtained from the passive knee rotation for the displacement at the myotendinous junction and at the muscle belly points were in average 0.41 and 0.51 mm/degree respectively.

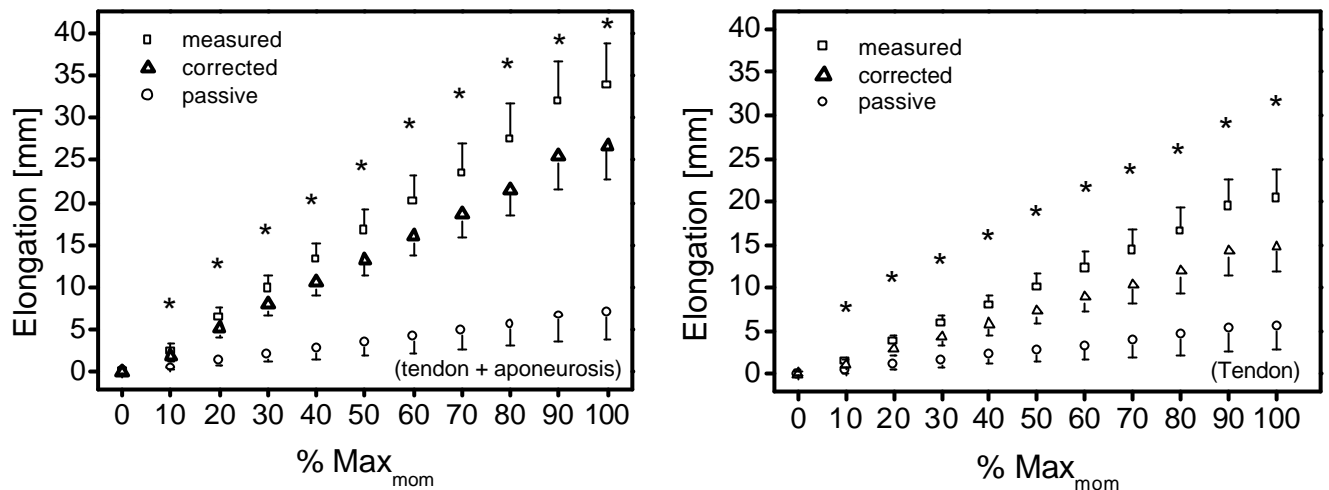


Figure 2: Measured elongation ($\Delta L_{\text{meas.}}$), passive elongation due to knee joint rotation ($\Delta L_{\text{pas.}}$), and corrected elongation ($\Delta L_{\text{corrected}}$) of the VL tendon and VL tendon plus aponeurosis during maximal knee extension efforts (mean \pm SD, $n=12$).

The elongation of the tendon and tendon plus aponeurosis showed statistically significant ($p < 0.05$) differing values. The maximal estimated displacement at the VL muscle belly point was almost double of that at the myotendinous junction point (27 ± 4 vs. 15 ± 3 mm respectively). Although the elongation of the tendon and of the tendon plus aponeurosis were significantly different, no statistical significant differences in strain were found (Fig. 3). Further the strain of the aponeurosis did not show any statistically significant differences to the strains of tendon and tendon plus aponeurosis either (Fig. 3). The maximal strains of the tendon, tendon plus aponeurosis and aponeurosis were 8 ± 2 , 8 ± 1 , $7 \pm 2\%$ ($p=0.394$ to $p=0.423$). This is due to the different resting lengths of the tendon (183 ± 12 mm), the tendon plus aponeurosis (345 ± 18 mm) and the aponeurosis (162 ± 25 mm).

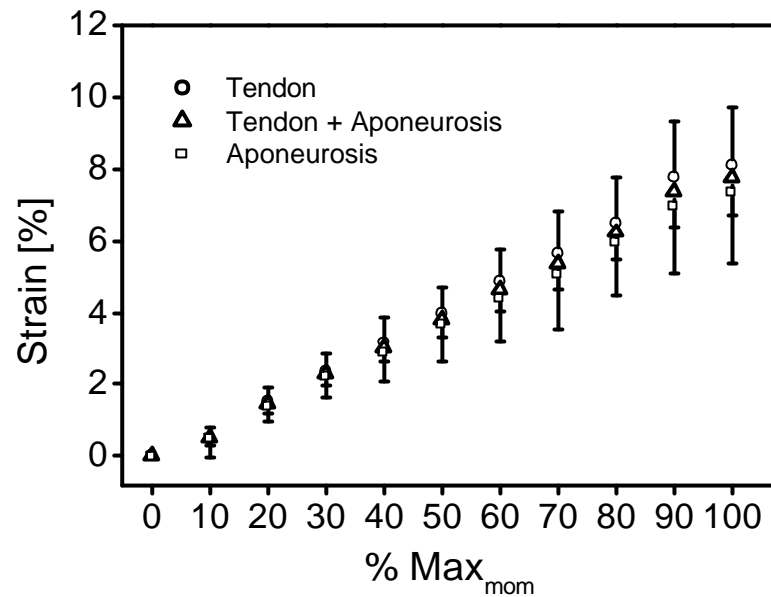


Figure 3: Strain values of the VL tendon, VL tendon plus aponeurosis and aponeurosis in relation to the exerted moment (mean \pm SD, $n=12$).

2.5 Discussion

The main results of this study can be summarised as follows: a) The elongation of the VL tendon plus aponeurosis was significantly greater than that of the VL tendon, b) the strain patterns of the tendon, tendon plus aponeurosis and aponeurosis were not different from each other and c) the knee joint angle changed during the maximal isometric contraction and significantly influenced the elongation and strain of the examined tendinous tissues.

The strain of the tendon and the strain of the tendon plus aponeurosis reported at the present study correspond to the strain of all structures distal to the analysed points: patella tendon, the fibres of the quadriceps tendon

over the anterior aspect of the patella (Staeubli et al. 1999b; Due et al 2003), VL tendon and a part of the VL aponeurosis. It is very difficult to examine the separate strains of all these structures *in vivo* using ultrasonography because of the accuracy of the method. However from a methodological point of view it is important to prove if the estimated strain is dependent on the analysed points (myotendinous junction vs. muscle belly) because it is much easier to follow the cross point at the VL muscle belly than at the myotendinous junction (exemplarily in this experiment we tested 28 subjects. Only 12 of them could be analysed at the myotendinous junction).

The uncorrected estimate of the elongation of the tendon plus aponeurosis was 33.8 ± 5.0 mm. This is partially in line with previously reported values (Kubo et al. 1999; 2000a,b). In our experiment we used a Biodex dynamometer (Biodex Medical Systems. Inc. USA) where the subjects were carefully fixed and the knee joint axis was aligned with the axis of the dynamometer. Although an isometric task was executed, the knee joint angle changed during the contraction of the knee extensors. The magnitude of the joint rotation reached values up to $13.6^\circ \pm 5.9$ at the maximum knee extension moment. This rotation plays an important role in the displacement of the tendinous tissues during maximal extension efforts (Fig. 2). The knee joint motion can be influenced by the compliance of the dynamometer and the deformation of the soft tissues of the leg. The deformation of the chair is also an important factor contributing to the knee joint rotation. The latter might be reduced by means of a more stable and stiffer chair. However during a maximal voluntary isometric knee extension the

average moment achieved was 205 ± 41 Nm. These are considerably high values and may cause soft tissue deformation at the thigh. It seems quite difficult to completely prevent this tissue deformation, since the mechanical properties of the soft tissues might vary a lot among subjects. In the present study we corrected the elongation of the tendon and tendon plus aponeurosis through a former passive motion of the knee. The measured displacements were corrected by the displacement happening due to joint rotation. The influence of passive elongation of the tendon and tendon plus aponeurosis was about 0.41 and 0.51 mm/degree respectively. These values are in accordance with those reported by Bojsen-Møller et al. (2003) who reported 0.41 mm/degree at the VL muscle belly.

Our major finding is that the strain values of the VL tendon, VL tendon plus aponeurosis and VL aponeurosis are not differing from each other (Fig. 3). The estimated strains at maximal load were 8 ± 2 , 8 ± 1 % and 7 ± 2 % ($p=0.394$ to $p=0.423$) for the VL tendon, VL tendon plus aponeurosis and VL aponeurosis. These values are within the failure values reported by Johnson et al. (1994) for patellar tendon strain *in situ* (15 ± 5 and 14 ± 6 % for the young and mature probes respectively) and by Schatzmann et al. (1998) who reported ultimate strain values for unconditioned patellar-ligament-bone and quadriceps-tendon-bone complex of 15.1 ± 4.4 % and 14.7 ± 3.7 % respectively. From a methodological point of view, a similar strain between the human vastus lateralis tendon and aponeurosis during maximal isometric contractions has relevant implications for the *in vivo* estimation or diagnostic of the mechanical properties of VL tendinous tissues using ultrasonography. The choice of the examined cross point on

the ultrasound image does not influence the estimation of the strain when the initial resting length is known.

As mentioned in the introduction there are studies reporting different results in the literature. Maganaris and Paul (2000) found higher strains at the aponeurosis compare to the tendon (9.2% vs. 2.5%) in the tibialis anterior muscle tendon unit. Oppositely, Magnusson et al. (2003) reported that the strain of the Achilles tendon was 5.7 times greater than that of the distal aponeurosis of the gastrocnemius medialis (8.0% vs. 1.4%). Muramatsu et al (2001) who examined the strain of the gastrocnemius medialis tendon and aponeurosis found no differences between the strain of these structures. The above differing and partially contrasting findings may be explained by two reasons: a) the behaviour of the extension of the tendon and aponeurosis might be different for different muscles and b) methodological differences between the studies. For example Maganaris and Paul (2000), and Muramatsu et al. (2001) examined the elongation of the tendon at the myotendinous junction. Magnusson et al. (2003) utilised an invasive marker to study the elongation of the free Achilles tendon. All these studies used two separate trials for their measurements. As mentioned in the introduction, all these can influence the findings. Another important issue is the accuracy of the ultrasound method. The resolution of the ultrasound images is (0.7 x 0.4 mm) this means that it is difficult to achieve an accuracy below 1 mm during the measurements. Therefore the accuracy for the calculation of the strain of short structures (maximal elongation: 3-5 mm) is much lower than for long structures (maximal elongation 15-26 mm).

The presented method is not free of limitations. We estimated the elongation of the tendinous tissues based on two dimensional images. A two-dimensional analysis of curved surfaces causes projection errors. Another critical issue is the definition of the 0% strain. The position chosen for the maximal isometric contraction (knee angle: 115°, hip angle: 140°) may exhibit a passive moment (Riener & Edrich 1999). This however is low (<4 Nm) and would correspond to < 2% of the maximal achieved moment. Therefore a pre-stretch of the tendon and aponeurosis seems to be too low to have a substantial influence on the estimated elongation and strain of both structures. However by the use of simultaneous ultrasound recordings an underestimation of the elongation would not effect the main statement of this study. The different lines of action of the VL tendon and aponeurosis and patellar tendon may influence the elongation and strain of both structures since different force components (Buff et al. 1988) may result into different elongation. The possible error induced by the different lines of action would be the same for both points and therefore it does not influence the results of this study.

We conclude that the strains of the human VL tendon, VL tendon plus aponeurosis and VL aponeurosis, as estimated *in vivo* by two dimensional ultrasound during a maximal isometric contraction, do not differ from each other. The estimated displacement at the cross point in the VL muscle belly is significantly greater compared to that at the VL myotendinous junction. In consequence, to obtain the elongation of the whole tendon it is necessary to multiply the strain calculated for the examined part of the tendon by the total length of the tendon. The maximal extension effort

causes a notable knee joint rotation and a correction of the estimated elongation is necessary.

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3 **SECOND STUDY: MUSCLE TENDON UNIT MECHANICAL AND MORPHOLOGICAL
PROPERTIES AND SPRINT PERFORMANCE**

Savvas Stafilidis

Adamantios Arampatzis

German Sport University of Cologne, Institute of Biomechanics and Orthopaedics

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in vivo

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3.1 Abstract

The objective of this study was to examine whether sprint performance is related to the mechanical and morphological properties of the quadriceps femoris (QF) and triceps surae (TS) muscle-tendon units (MTUs). Two sprinter groups (slow = 11; fast = 17) performed maximal isometric knee extension and plantarflexion contractions on a dynamometer at eleven different MTU lengths. The activation of TS and QF during the maximal voluntary contractions (MVC) was estimated by surface electromyography. The elongation of the tendon and aponeurosis of the gastrocnemius medialis (GM) and the vastus lateralis (VL) were measured by means of ultrasonography. The maximal joint moments at the ankle and knee joints and the morphological properties of the GM and VL showed no statistically significant differences between groups ($P > 0.05$). The fast group exhibited a higher maximal elongation in the VL tendon and aponeurosis during the MVC ($P < 0.05$). Furthermore, the maximal elongation of the VL tendon and aponeurosis showed a significant correlation with the 100 m sprint times ($r = -0.567$, $P = 0.003$). It is suggested that the higher elongation of the VL tendon and aponeurosis of the fast group benefits the energy storage and return as well as the shortening velocity of the MTU.

3.2 Introduction

The work producing ability of the muscle-tendon units (MTUs) can be influenced by the distribution of their muscle fiber type (fast vs. slow) as well as by their mechanical and morphological properties (Zuurbier and Huijing, 1992). Longer muscle fascicles can exhibit higher shortening velocities and mechanical powers than shorter fascicles. Furthermore, longer muscle fascicles have a wider length range of force development (force-length relationship) than shorter muscle fascicles, which can affect muscle's performance (De Haan *et al.*, 2003). In addition the non rigidity of the tendon and aponeurosis influences the contraction mechanics of the contractile component. The tendon compliance allow the muscle fibres to contract at lower shortening velocities than the whole MTU (Ettema *et al.*, 1990) and as a consequence of the force-velocity relationship their force generating potential will be higher (Hof *et al.*, 2002; Bobbert, 2001). Furthermore due to the non rigidity of the tendon and aponeurosis, when the MTU is elongated, strain energy can be stored. This way the whole energy delivered during the shortening of the muscle-tendon unit can be enhanced (Alexander and Bennet-Clark, 1977; Bobbert, 2001).

Based on the notion that the fascicle length might affect the sprint performance, several studies examined the relationship between the morphological (Abe *et al.*, 2000; 2001; Kumagai *et al.*, 2000) properties of the lower extremities and the 100 m sprint times. Kumagai *et al.* (2000) found a positive relationship between vastus lateralis (VL) fascicle length and sprint performance. Abe *et al.* (2000) revealed that a sprinter group had

significant longer VL and GM fascicle lengths than an endurance and an untrained control group. In addition the pennation angles of the sprinter group were significantly lower to those of the endurance group. The authors (Abe *et al.*, 2000) conclude that the greater fascicle length and the lower pennation angle observed at the sprinter group could favour the shortening velocity of the fascicles and in consequence the running speed. In a similar way Abe *et al.* (2001) examined the relationship between the VL and GL muscle fascicle lengths and sprint performance in female sprinters and untrained controls. Again a significant correlation between the fascicle lengths of the VL and GL and the sprint performance was found. However, whether the mechanical properties (e.g. tendon compliance) of the leg extensor MTUs affect sprint performance has not been clearly identified. Only Kubo *et al.* (2000) examined the mechanical properties of the tendon and aponeurosis of the GM and VL MTUs in sprinters and untrained controls. They showed that the compliance of the VL-tendon and aponeurosis influence sprint performance (Kubo *et al.* 2000). Furthermore the above mentioned studies examined separately either the morphological properties of the muscles (Abe *et al.*, 2000,2001; Kumagai *et al.*, 2000) or the mechanical properties of the tendon and aponeurosis (Kubo *et al.*, 2000) of the lower extremities. To our knowledge there is no study dealing with the influence of morphological and mechanical properties of the lower extremities MTUs on sprinting performance using the same group of sprinters.

Therefore, the objective of this study was to examine the relationship between the mechanical and morphological properties of the QF and TS

MTUs and sprint performance using the same subjects. We hypothesised that (a) sprint performance can be affected by the morphological and the mechanical properties of the lower extremities MTUs and (b) that the morphological and the mechanical properties of the MTUs together could explain the sprint performance to a higher extent than separately.

3.3 Materials and methods

3.3.1 Subjects

Twenty eight male sprinters (body mass: 78.3 ± 6.4 kg, height: 185 ± 6 cm, age: 21 ± 3.2 years) participated in this study. The subjects gave their informed written consent to the experimental procedure accomplishing with the rules of the local scientific board. All subjects performed sprint training at least 5 times per week over the last 5 years and participated regularly in sprint running competitions.

3.3.2 Measurement of maximal isometric ankle and knee joint moment

The subjects performed maximal isometric voluntary ankle plantarflexion and knee extension contractions with their left leg on a Biodex dynamometer (Biodex-System3, Biodex Medical Systems. Inc. USA) on two separate test days. The warm-up consisted of 2-3 minutes performing submaximal isometric contractions and three MVCs. Afterwards the subjects performed plantarflexion or knee extension MVCs at eleven different ankle-knee and knee-hip joint angle configurations respectively (Table 1). Different joint

angle configurations were chosen in order to examine the TS and QF muscle force potentials over the whole range of achievable MTU lengths. The different joint angle configurations were applied in random order. There were three minutes rest between contractions. The subjects were instructed and encouraged to produce a maximal isometric moment and to hold it for about 2-3 seconds.

The resultant moments at the ankle and knee joints were calculated through inverse dynamics (Arampatzis *et al.*, 2004; 2005b). Kinematic data were recorded using the Vicon 624 system (Vicon Motion Systems, United Kingdom) with 8 cameras operating at 120 Hz. To calculate the lever arm of the ankle joint during ankle plantarflexion the center of pressure under the foot was determined by means of a flexible pressure distribution insole (Pedar-System, Novel GmbH, Germany) operating at 99 Hz (Arampatzis *et al.*, 2005b). The compensation of moments due to gravitational forces was done for all subjects before each ankle plantarflexion or knee extension contraction. The method for calculating the resultant joint moments has been previously described (Arampatzis *et al.*, 2004; 2005b).

The moments arising from antagonistic coactivation during the ankle plantarflexion or the knee extension efforts were quantified by assuming a linear relationship between surface electromyography (EMG) amplitude of the ankle dorsiflexor or knee flexor muscles and moment (Baratta *et al.*, 1988). This was established by measuring EMG and moment whilst relaxed and during two submaximal ankle dorsiflexion or knee flexion contractions at each joint angle configuration (Mademli *et al.*, 2004). Consequently, in the following text maximal knee and ankle joint moments refer

to the maximal joint moment values considering the effect of gravitational forces, the effect of joint axis alignment relative to the dynamometer axis and the effect of the antagonistic moment on the moment measured at the dynamometer.

Table 1: Eleven ankle–knee and knee-hip joint angle configurations in degrees used for the isometric maximal voluntary plantar flexion and knee extension contractions respectively. Tibia perpendicular to the foot-sole was defined as 90°, ankle angle (plantarflexed position: ankle angle >90°; dorsiflexed position ankle angle: <90°). A straight trunk and knee were defined as 180° hip and knee joint angles respectively.

| | Plantar flexion | | Knee extension | |
|-----------------|--------------------|-------------------|-------------------|------------------|
| Position | Ankle joint | Knee joint | Knee joint | Hip joint |
| 1 | 120 | 75 | 170 | 95 |
| 2 | 120 | 110 | 170 | 110 |
| 3 | 110 | 100 | 170 | 140 |
| 4 | 120 | 140 | 160 | 125 |
| 5 | 100 | 110 | 160 | 140 |
| 6 | 110 | 150 | 140 | 125 |
| 7 | 90 | 130 | 140 | 140 |
| 8 | 100 | 170 | 110 | 95 |
| 9 | 80 | 10 | 110 | 125 |
| 10 | 90 | 170 | 80 | 95 |
| 11 | 80 | 170 | 80 | 140 |

3.3.3 Measurement of EMG-activity during isometric contractions

Bipolar pre-amplified EMG leadoffs (analogue RC-filter, 10-500 Hz bandwidth, Biovision, Wehrheim, Germany) and auto-adhesive surface electrodes (0.8 cm² pickup surface, 2 cm interelectrode distance, Blue sensor - Medicotest Denmark) were used to analyze muscle activity.

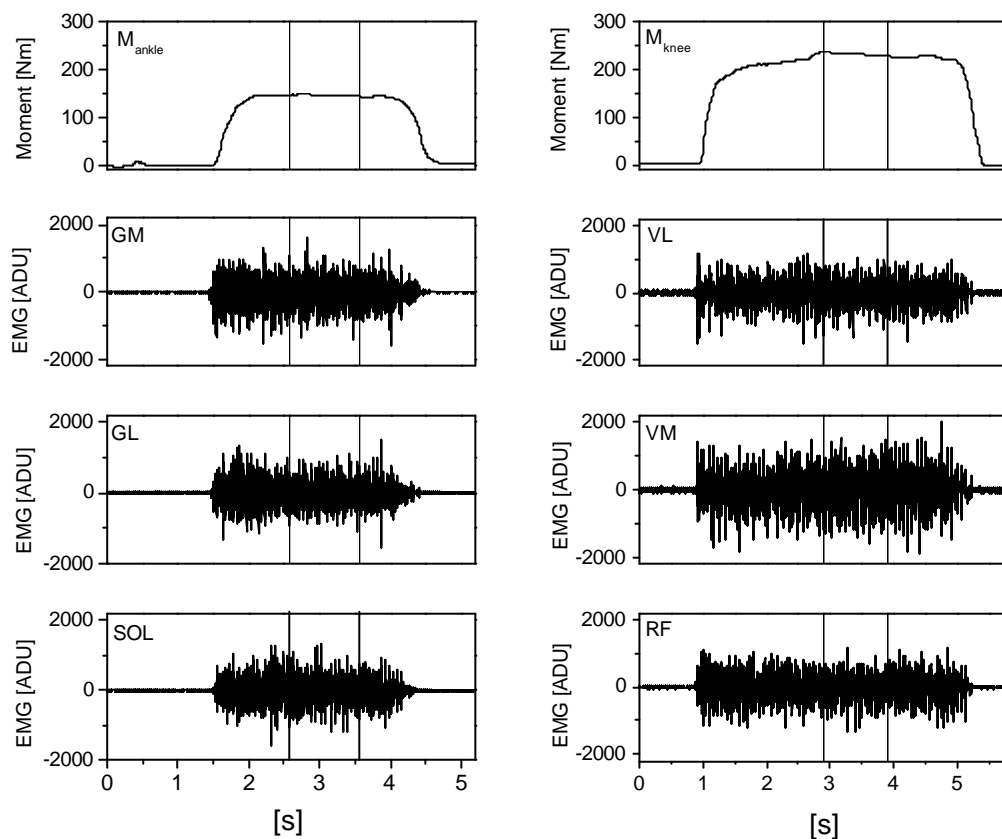


Figure 1: Moment- and EMG-time histories from one subject (left) during a maximal plantar flexion contraction at position 4 (ankle angle: 120, knee angle: 140) and (right) during a maximal knee extension contraction at position 6 (knee angle: 140, hip angle: 125). The ankle and knee joint moments, as well as the RMS of the EMG activity were determined as the average values from the steady state of the plantar flexion and knee extension moment. The two vertical lines indicate the considered time window (1 s).

M_{ankle} : ankle joint moment

M_{knee} : knee joint moment

Before placing the electrodes the skin was carefully prepared (shaved and cleaned with alcohol) to reduce skin impedance. The electrodes were positioned above the midpoint of the muscle belly as assessed by palpation and parallel to the presumed direction of the muscle fibres. The activation of the TS muscle was assessed from the EMGs of the GM, GL and soleus (SO). During knee extension the EMG-activities of the VL, vastus medialis (VM) and rectus femoris (RF) were analyzed. The EMG signals were recorded at 1080 Hz by the Vicon unit. After preparing the subjects the EMGs were checked during submaximal and maximal isometric contractions for each muscle group to determine whether an adequate signal was obtained from each muscle and to adjust the gains. Further, the EMG signal from each muscle was checked online for mechanical artefacts by passively shaking the leg. The preparation was renewed when such artefacts were observed. All isometric contractions at the knee or the ankle joint were performed within one testing session. No electrode replacement or re-adjusting of the EMG pre-amplification gain were done during the measurements.

To describe the EMG-activity the root mean square (RMS) of the raw signals during a 1000 ms time interval at peak joint moment were calculated (Fig. 1). The RMS from each muscle was normalized to the individual maximal RMS value of each muscle for each subject during the eleven isometric contractions. In order to determine the EMG-activity of the ankle plantarflexor and knee extensor muscles, the normalized RMS of the examined muscles were averaged and weighted by their physiological cross

sectional areas (PCSA): TS a 6:2:1 ratio for SO, GM, GL (Out *et al.*, 1996) and QF a 0.92:1.00:0.72 ratio for RF, VL and VM (Herzog *et al.*, 1990).

3.3.4 Measurement of tendon elongation

The tendon properties were determined on two additional test days. The subjects performed a MVC ankle plantarflexion (ankle joint angle 90°, knee joint angle 180°) and a knee extension (knee joint angle 115°, hip joint angle 140°) with their left leg on the dynamometer. A 7.5 MHz linear array ultrasound probe (Aloka SSD 4000, 43 Hz) was used to visualize the distal tendon and aponeurosis of the GM and VL respectively (Fig. 2). The exact protocol for the analysis of the tendinous tissue elongation during ankle plantarflexion and knee extension is described in detail elsewhere (Arampatzis *et al.*, 2005a; Stafilidis *et al.*, 2005).

The effect of inevitable joint angular rotation on the elongation of the tendon and aponeurosis during the loading phase was taken into account by capturing the motion of the tendons and

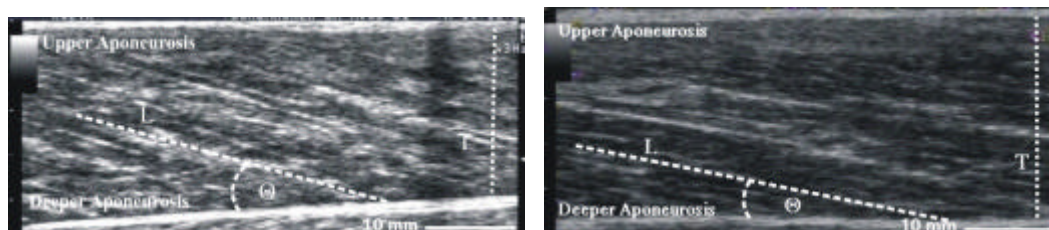


Figure 2: Muscle ultrasound images at rest. Left: gastrocnemius medialis; Right: vastus lateralis. L: fascicle length; T: muscle thickness; θ : pennation angle

aponeuroses from the GM and VL during a passive motion of the ankle or the knee joint (Muramatsu *et al.*, 2001; Bojsen-Møller *et al.*, 2003). This

allowed to subtract the elongation obtained for the tendon and aponeurosis due to joint rotation during each maximal ankle plantarflexion or knee extension trial (Fig. 3). The ultrasound images taken during the passive joint motion and during the MVCs were digitized frame by frame until the maximal moment was achieved.

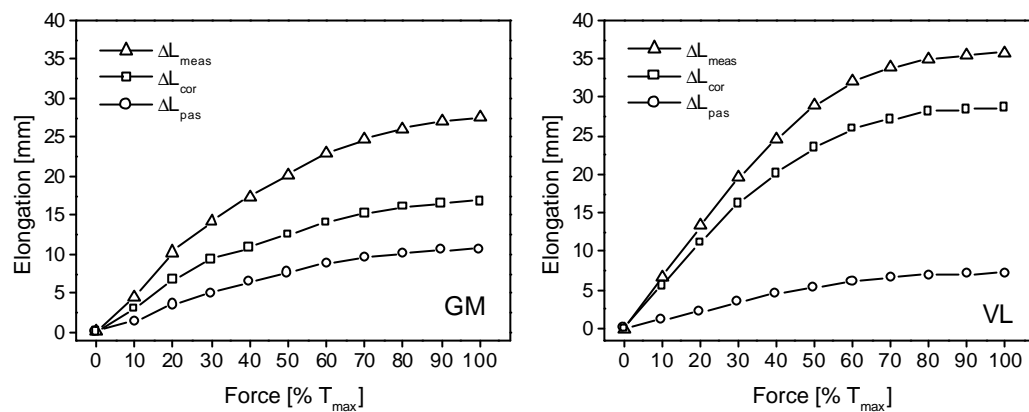


Figure 3: Average values of the elongation as measured from the ultrasound images (ΔL_{meas}), passive elongation due to joint angular rotation ($\Delta L_{passive}$) and corrected elongation ($\Delta L_{corrected}$) of the tendon and aponeurosis during plantar flexion (left) ($n=28$) and knee extension (right) effort ($n=25$, the ultrasound images of the VL tendon and aponeurosis from three subjects could not be analysed because of poor image quality).

The insertion of the fascicle into the deep aponeurosis (Fig. 2) was tracked during the contraction and during a passive trial to determine the elongation of the tendon and aponeurosis. The resting length of the GM (knee joint angle 180° , ankle joint angle 110°) and the resting length of the VL (knee joint angle 130° , hip joint angle 140°) tendon and aponeurosis were identified on the ultrasound images (Arampatzis *et al.*, 2005a; Stafilidis *et al.*, 2005). The specific joint angle configurations chosen for resting length

determination aimed to reduce passive joint moments as much as possible (Riener and Edrich, 1999).

3.3.5 Measurement of muscle architecture

The muscle architecture of the GM and VL (fascicle length, pennation angle and thickness) were determined by ultrasonography during the same test session as for the analysis of the tendon properties and using the same joint angle configurations (GM: ankle joint angle 90°, knee joint angle 180°; VL: knee joint angle 115°, hip joint angle 140°). All measurements were done on the relaxed muscles at the cited positions. The pennation angle of the GM and VL were measured as the angle of insertion of the muscle fascicles into the deep aponeurosis. Fascicle length was defined as the length of the fascicular path between the insertions of the fascicle into the upper and deeper aponeuroses. The ratios between fascicle length of the GM and tibia length, and between fascicle length of the VL and femur length were also analysed. Femur length was defined as the distance between the lateral femoral condyle and the major trochanter, and tibia length as the distance between the lateral malleolus and lateral femoral condyle. Muscle thickness was defined as the distance between the deeper and upper aponeuroses (Fig. 2).

3.3.6 Statistics

A cluster analysis, based on the best 100 m times of the subjects was used to group them. The cluster analysis revealed two reasonably homogenous groups (group 1: fast sprinters, $n = 17$; group 2: slow sprinters,

n = 11). The anthropometric data and the 100 m times for both groups are presented in Table 2. We used a two-factor (group \times position) analysis of variance (ANOVA) to detect differences in the exerted moment and EMG activity of the QF and TS MTUs during isometric contraction. A t-test for two independent samples was used to identify possible differences on the morphological properties (fascicle length, pennation angle, muscle thickness) and maximal elongation and strain of the tendon and aponeurosis between the two groups. The level of significance for all statistical comparisons was set at $\alpha = 0.05$. The Pearson's correlation coefficient (r) was used to test the correlation between sprint performance and mechanical and morphological properties of the MTU. In all figures the data are presented as mean \pm s_{ξ} (standard error of mean), whereas in the text and tables they are expressed as mean \pm 1 s (standard deviation).

Table 2: Means \pm s of the anthropometric data and the 100 m time of the examined sprinters

| | slow (n=11) | fast (n=17) |
|----------------|------------------|------------------|
| Age [years] | 19,5 \pm 2.5 | 21,2 \pm 3.6 |
| Height [m] | 1,83 \pm 0.06 | 1,86 \pm 0.06 |
| Body mass [kg] | 75,8 \pm 5.8 | 79,9 \pm 6.5 |
| 100m time [s] | 11,64 \pm 0.23 | 11,04 \pm 0.17 |

3.4 Results

3.4.1 Triceps surae muscle-tendon unit

There was no statistically significant difference ($P > 0.05$) between groups in the anthropometric characteristics (Table 2). The comparison of the resting length of the GM tendon and aponeurosis between slow (302 ± 23 mm) and fast sprinters (317 ± 22 mm) also failed to reveal statistically significant differences ($P = 0.112$). In the same manner neither the maximal elongation nor the maximal strain of the GM tendon and aponeurosis showed statistically significant differences ($P > 0.05$) between both performance groups (Table 3). Furthermore at the examined position (ankle-, knee angle: $90^\circ / 180^\circ$) there were no statistically significant differences ($P > 0.05$) between groups in fascicle length, ratio fascicle length to tibia length, angle of pennation, or thickness of the GM at rest (Table 4).

Table 3: Maximal elongation and maximal strain of the tendon and aponeurosis for both examined groups at the Triceps surae and Quadriceps femoris muscle-tendon unit

| | GM | | VL | |
|-------------------------|----------------|----------------|----------------|------------------|
| | slow (n=11) | fast (n=17) | slow (n=10) | fast (n=15) |
| Maximal elongation (mm) | 18.5 ± 2.5 | 19.4 ± 4.7 | 26.9 ± 5.7 | $32.9 \pm 7.0^*$ |
| Maximal strain (%) | 6.1 ± 1.0 | 6.2 ± 1.5 | 7.7 ± 1.6 | $9.0 \pm 1.5^*$ |

*: Statistical significant difference ($P < 0.05$) between both groups.

The maximal plantar flexion moment values as well as the normalized EMG-activity at the eleven examined positions were also similar for both groups (no group effect) (Fig. 4). As expected, the two-factor ANOVA provided a significant ($P < 0.05$) position effect on the maximal achieved ankle plantar flexion moment as well as on EMG-activity.

Table 4: Pennation angle, fascicle length, muscle thickness, tibia length and ratio fascicle length to tibia length of the gastrocnemius medialis muscle at rest for the examined sprinter groups (mean \pm s). The muscle architecture was determined at ankle angle: 90° and knee angle: 180°.

| GM | Slow group (n=11) | Fast group (n=17) |
|---------------------------------|----------------------|----------------------|
| Pennation angle [°] | 17.0 \pm 3.2 | 17.5 \pm 2.2 |
| Fascicle Length [cm] | 8.59 \pm 3.21 | 7.54 \pm 1.5 |
| Muscle Thickness [cm] | 1.97 \pm 0.12 | 1.99 \pm 0.22 |
| Tibia length [mm] | 440.8 \pm 20.9 | 453.1 \pm 19.4 |
| Ratio (fascicle / tibia length) | 0.196 \pm 0.077 | 0.167 \pm 0.032 |

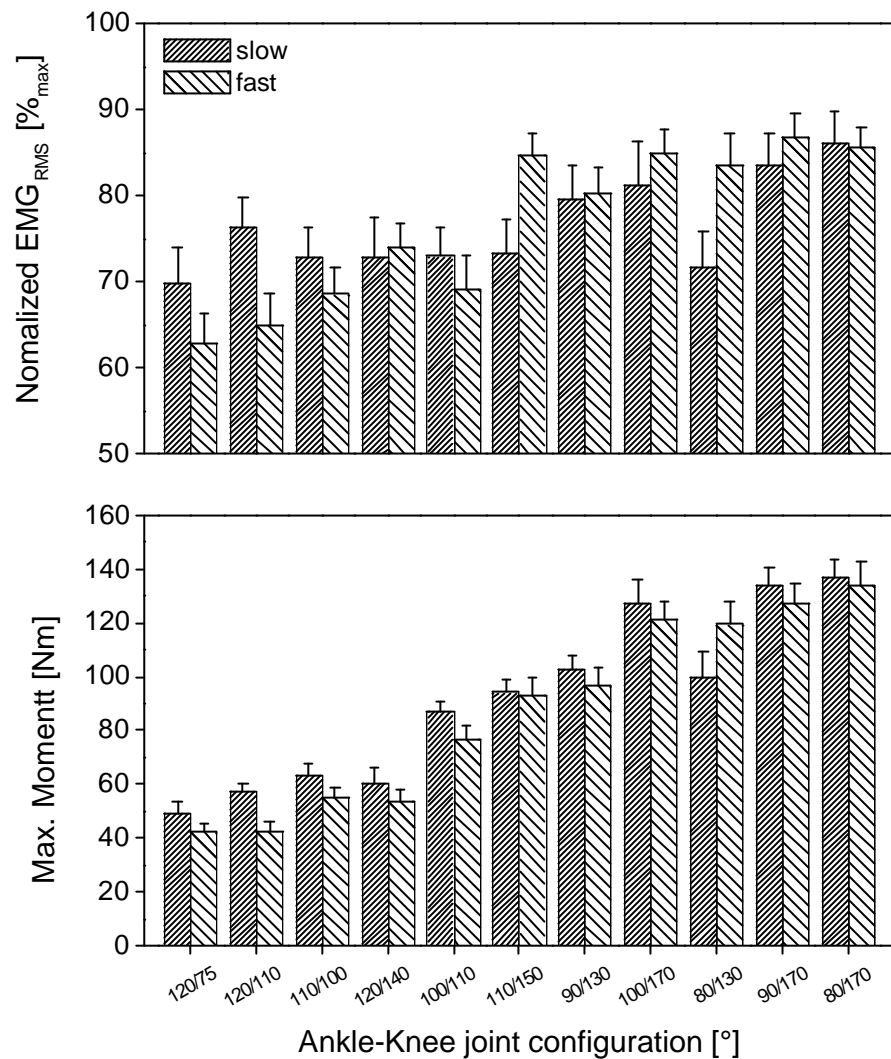


Figure 4: Normalized RMS values of the EMG signal (top) for the triceps surae muscle (gastrocnemius medialis, gastrocnemius lateralis and soleus) and ankle joint moment (bottom) during maximal isometric voluntary plantar flexion effort at 11 different joint angle configurations for the slow (slow: $n = 11$) and the fast (fast: $n = 17$) sprinter groups (mean $\pm s_e$). RMS values from each subject were normalized to the highest RMS value measured over all joint angle configurations. The two way ANOVA showed only a position effect ($P < 0.05$) on the maximal ankle plantar flexion moment as well as in the EMG values.

3.4.2 Quadriceps femoris muscle-tendon unit

The ultrasound images of the VL muscle from three subjects (one slow and two fast sprinters) could not be analyzed because of poor image quality. Therefore for the VL only data from 25 sprinters were available. The resting length of the VL tendon and aponeurosis revealed no statistically significant differences between the fast group (347 ± 15 mm) and the slow group (363 ± 24 mm) ($P > 0.05$). The fast group exhibited a significantly ($P < 0.05$) higher maximal elongation and strain of the VL tendon and aponeurosis than the slow group (Table 3). Furthermore when looking at all 25 sprinters together, the maximal elongation of the VL tendon and aponeurosis showed a significant negative correlation with the sprint 100 m time ($r = -0.567$, $P = 0.003$; Fig. 5).

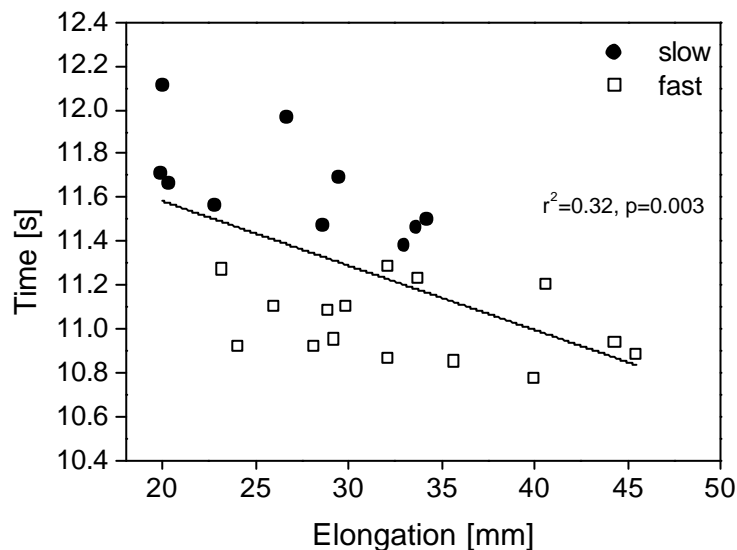


Figure 5: Relationship between the maximal elongation of the vastus lateralis tendon and aponeurosis and the 100 m sprint time ($n = 25$). Slow: slow group of sprinters ($n = 10$); fast: fast group of sprinters ($n = 15$)

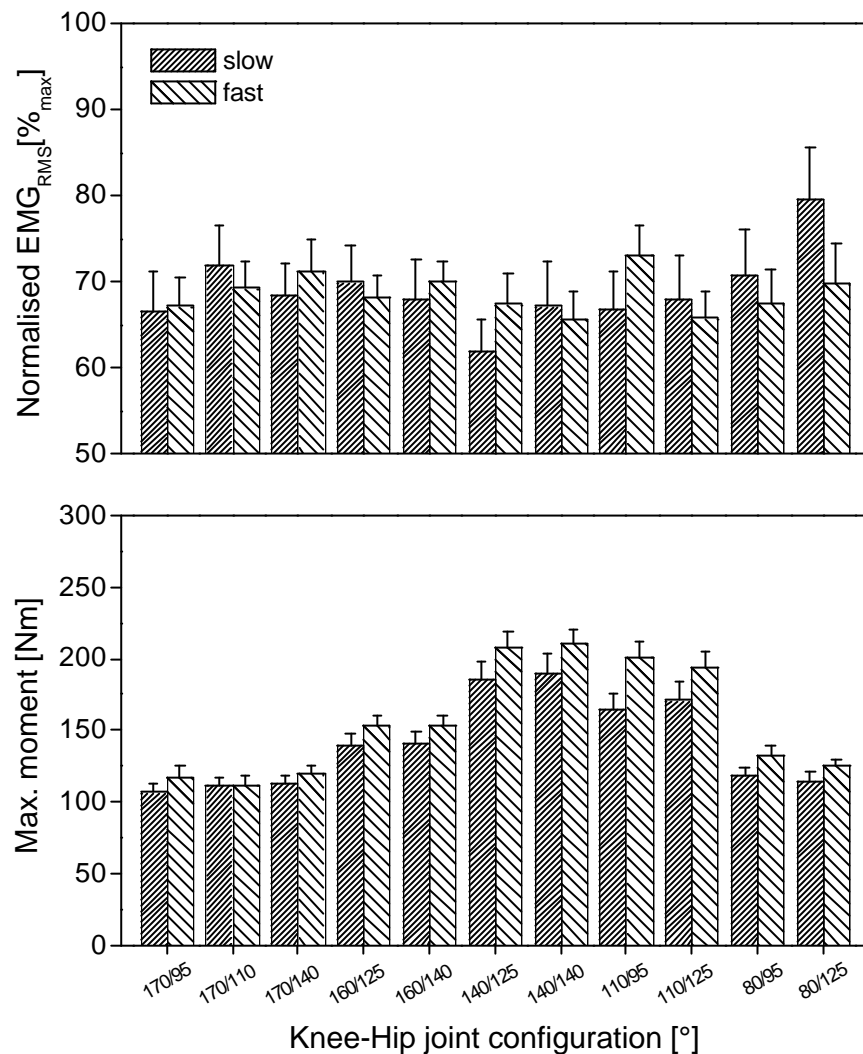


Figure 6: Normalized RMS values of the EMG signal (top) for the quadriceps femoris muscle (vastus medialis, vastus lateralis and rectus femoris) and knee joint moment (bottom) during maximal isometric voluntary knee extension effort at 11 different joint angle configurations for the slow (slow: $n = 11$) and the fast (fast: $n = 17$) sprinter groups ($\text{mean} \pm s_{\epsilon}$). RMS values from each subject were normalized to the highest RMS value measured over all joint angle configurations. The two way ANOVA showed only a position effect ($P < 0.05$) on the maximal knee extension moment.

At the VL muscle none of the examined morphological parameters at rest (fascicle length, pennation angle, ratio fascicle length to femur length and

muscle thickness) revealed statistically significant differences ($P > 0.05$) between groups (Table 5). The two-factor ANOVA showed no statistically significant differences ($P > 0.05$) in maximal knee extension moment between the two performance groups. There was only a position effect ($P < 0.05$) in the maximal knee extension moment (Fig. 6).

Table 5: Pennation angle, fascicle length, muscle thickness, femur length, and ratio fascicle length to femur length of the vastus lateralis muscle at rest for the examined sprinter groups (mean \pm s). The muscle architecture was determined at knee angle: 110° and hip angle: 140° .

| VL | Slow group (n=10) | Fast group (n=15) |
|---------------------------------|----------------------|----------------------|
| Pennation angle [°] | 9.5 \pm 2.4 | 10.5 \pm 1.7 |
| Fascicle Length [cm] | 11.77 \pm 1.74 | 12.06 \pm 2.25 |
| Muscle Thickness [cm] | 2.51 \pm 0.54 | 2.32 \pm 0.25 |
| Femur length [mm] | 451 \pm 19 | 453 \pm 18 |
| Ratio (fascicle / femur length) | 0.261 \pm 0.036 | 0.267 \pm 0.054 |

3.5 Discussion

The main findings of this study were that the maximal ankle and knee joint moments and the morphological properties (pennation angle, fascicle

length, thickness and ratio between fascicle length and segment length) of the GM and VL showed similar values for both sprint performance groups. However the fast group displayed a significantly higher maximal elongation and strain of the VL tendon and aponeurosis.

The results of this study indicate that the morphological properties of the triceps surae and quadriceps femoris MTUs do not relate to the sprint performance within homogenous groups of sprinters. Only the maximal elongation of the VL tendon and aponeurosis showed a significant correlation with the 100 m times of the examined sprinters ($r^2 = 0.32$, $P = 0.003$). Including the VL fascicle length into this correlation does not provide further explanation of the sprint performance ($r^2 = 0.28$). Furthermore the maximal isometric moments at the ankle and knee joints showed similar values for both groups at all examined positions. The lack of differences in the maximal joint moments between the two groups of sprinters within the examined joint angle configurations further supports that there were no differences in the working ranges (widths) of the force length relationships of the triceps surae and quadriceps femoris muscles between groups. Therefore it can be assumed that also the other heads of the triceps surae and quadriceps femoris will display similar fascicle lengths in both groups. From all the above we conclude that in our sample of sprinters the morphological characteristics of the triceps surae and quadriceps femoris were not responsible for the differences in sprint performance between the two groups.

Albeit no differences between groups were found in the muscles' morphology, the maximal elongation and strain of the VL tendon and aponeurosis

were higher for the fast group than for the slow one. A higher maximal elongation of the VL tendon and aponeurosis could favour the sprint performance since the extensor moment acting on the knee joint could reach greater values during the sprint race than during an isometric contraction because of the eccentric contraction of the quadriceps femoris muscles in the first part of the ground contact phase (Belli *et al.*, 2002; Mann and Sprague 1980). During this first part of the contact phase in running the MTU is lengthening and so does the tendon and aponeurosis due to the developed forces. Consequently strain energy is stored in the tendon and aponeurosis. In the subsequent second part of the contact phase the MTU shortens and the stored strain energy of the tendon and aponeurosis can be reutilised (Hof *et al.*, 2002). A higher elongation of the VL tendon and aponeurosis during the first part of the ground contact, could increase the energy storage and return from the tendon and aponeurosis. When running, at the second part of the contact phase the shortening velocity of the muscle fibres is lower than the shortening velocity of the MTU due to the additional shortening of the tendon and aponeurosis (Hof *et al.*, 2002). Thus an increase in the elongation of the VL tendon and aponeurosis during the first part of the ground contact phase can reduce the shortening velocity of the VL muscle fibres (contractile element) during the second part of the ground contact phase because of the increased contribution of the tendon and aponeurosis to the MTU shortening velocity. Exemplarily a simulation study carried out by Bobbert (2001) showed that increasing the maximal strain of the tendon from 4 to 10% would cause a 4 cm increase in jumping performance because of a higher work output of the contractile

elements of the muscles due to the higher contribution of the series elastic element to the muscle-tendon unit shortening velocity.

For the sprinters participating in this study the maximal elongation of the VL tendon and aponeurosis correlated significantly with their 100 m times. This indicates that the maximal elongation of the tendon and aponeurosis has an influence on the sprint performance. In accordance to the present study Kubo *et al.* (2000) comparing sprinters with non-sprinter controls having similar maximal isometric ankle- and knee joint moments, reported that sprinters showed a higher maximal elongation of the VL but no differences in maximal elongation of the GM tendon and aponeurosis. Therefore our findings and the reports from the literature (Kubo *et al.*, 2000) show that the maximal elongation of the quadriceps femoris tendon and aponeurosis affect sprint performance.

In the present study we calculated the tendon force using tendon moment arms taken from the literature (Herzog and Read, 1993; Maganaris *et al.*, 1998). It can not be excluded that individual differences in the anatomical moment arms between groups may have been present and could have influenced the calculated tendon forces. In order to estimate this potential source of error we analyzed the ratio between tendon and aponeurosis displacement and ankle / knee joint angular rotation during passive motion. This ratio is not an exact representation of the lever arm because the tendon and aponeurosis are not rigid bodies. However it can be used as an index to examine the possible influence of differences in the lever arms between the groups. The ratios were taken from the values for the correction of the tendon and aponeurosis elongation due to joint rotation. For the

ankle joint the average ratios were (slow group: 0.55 ± 0.1 and fast group: 0.58 ± 0.07 mm/°) and for the knee joint (slow group: 0.51 ± 0.11 and fast group: 0.47 ± 0.13 mm/°). There were no significant differences between groups ($P > 0.05$), which indicates that the moment arms were similar for both groups. Therefore it can be excluded that differences in the lever arms between groups influenced the results.

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4 THIRD STUDY: THE EFFECT OF TRACK COMPLIANCE ON SPRINT PERFORMANCE

Savvas Stafilidis

Adamantios Arampatzis

German Sport University of Cologne, Institute of Biomechanics and Orthopaedics

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4.1 Abstract

The purpose of this study was to examine the hypothesis that sprint performance (time interval for a given distance) would be affected by track compliance. Ten sprinters participated in the study with better values on the more compliant tracks. The athletes performed maximal sprints (60 m) onto three different track configurations (hard, soft, spring). A 60 m long single-lane running surface was constructed. Plywood boards (1.2 cm thick) were placed onto a 60 × 0.6 m wooden chipboard frame serving as base surface. All subjects run two times on each track configuration in randomised order. The athlete's kinematics were recorded using the Vicon 624 system with 12 cameras operating at 250Hz. Four Kistler force plates (1250Hz) were used to record ground reaction forces. Sprint performance (time interval of 60 m) remained unaffected throughout the different track compliances ($P=0.57$). In addition there was no track effect ($P>0.05$) on the sprinting kinematics and kinetics of the ankle or knee joint. The hypothesis that sprint performance is affected by track compliance is rejected because the sprinters showed similar performances (time intervals) whilst sprinting 60 m distance onto all three track configurations. We concluded that (a) the possible deformation of the track whilst sprinting is too small to cause a specific adjustment in the leg mechanics affecting the effectiveness of the stretch-shortening cycle of the sprinters and (b) the energy exchange between sprinters and tracks has only a marginal effect on sprint performance due to its small magnitude.

4.2 Introduction

In the literature it is well documented that in several sport events the interaction between the athlete and non rigid surfaces has a considerable effect on performance (Bosco et al., 1997; Arampatzis and Brüggemann 1998; 1999; 2001; Kerdok et al., 2002; Arampatzis et al., 2004a). Whereas the mechanical behaviour of the surface is exclusively determined by its material properties, humans are able to adjust their behaviour depending on the surface characteristics (Farley et al., 1998; Ferris and Farley 1997; Marigold and Patla 2005) and are able to use the surface to vary their own motor performance (Sanders and Allen 1993; Arampatzis et al., 2001; 2004a,b). In general non rigid (compliant) surfaces can substantially affect (a) the energy storage and return from the surface during a motor task (Bosco et al., 1997; Arampatzis et al., 2004a,b) and (b) the work producing capability of the muscles involved in that motor task (Sanders and Allen 1993; Arampatzis et al., 2004b; Moritz et al., 2004).

In sprint events it is generally believed that very hard tracks improve the sprint performance (time at a given distance). This opinion often rises up controversial debates between athletes of different field events (i.e. sprinters vs. endurance runners) and increases the cost of the track building. Oppositely to that, model predictions (McMahon and Greene 1979) suggest that running speed is enhanced onto tracks of an intermediate compliance as compared to the rigid ones. Additionally experimental data on drop jumping showed a statistically significant increase of ~7% in jumping height for a compliant surface compared to a rigid one (Arampatzis et al.,

2004b). The reason for the higher jumping height was a higher ratio of positive to negative mechanical work done by the subjects during the ground contact on the compliant surface as compared to the rigid one. Therefore it could be possible that sprint performance is not maximised onto hard tracks. However to our knowledge there is no experimental study examining the influence of track compliance on sprint performance.

Basing on the literature (Bosco et al., 1997; Kerdok et al., 2002; Arampatzis et al., 2004a) we expected: a) that by modifying the track compliance, sprinters would adjust their leg mechanics in order to achieve a more advantageous stretch-shortening cycle during the ground contact phase in a track compliance-dependent manner and b) to observe a compliance-dependency of the energy storage and return amplitude by the tracks during the ground contact phase. Basing on these expectations we predicted that sprint performance would be affected by track compliance with better values on the more compliant tracks. Therefore the purpose of this study was to investigate experimentally the effect of track compliance on sprint performance.

4.3 Method

4.3.1 Measurement of running kinematics and kinetics

The investigation was conducted on 10 experienced male sprinters (age 20.2 ± 2.6 years, height 1.82 ± 0.06 m, body mass 76.1 ± 5.5 kg). All examined subjects trained sprint running at least 5 times per week over the last 5 years and participated regularly in national and regional sprint run-

ning competitions. The average 100 m sprint performance was 11.49 ± 0.3 s (range: 10.95 to 11.96 s). None of the subjects had a history of neuromuscular or musculoskeletal impairments. The athletes were asked to performed maximal sprints over 60 m with varying track configurations. All sprinters wore their own sprint running shoes (spikes) with 6 mm spikes. Kinematic data were recorded using the Vicon 624 system (Vicon Motion Systems, Oxford, United Kingdom) with 12 cameras operating at 250 Hz. The ground reaction forces (GRFs) were measured using four force plates (Kistler AG, Winterthur, Switzerland, two force plates $90 \text{ cm} \times 60 \text{ cm}$ and two force plates $60 \text{ cm} \times 40 \text{ cm}$) in order to increase the probability of getting valid trials (whole foot placed on the plate). The force plates were arranged in series (Fig. 1) and embedded in the runway flush with the ground. The ground reaction forces were sampled at 1250 Hz.

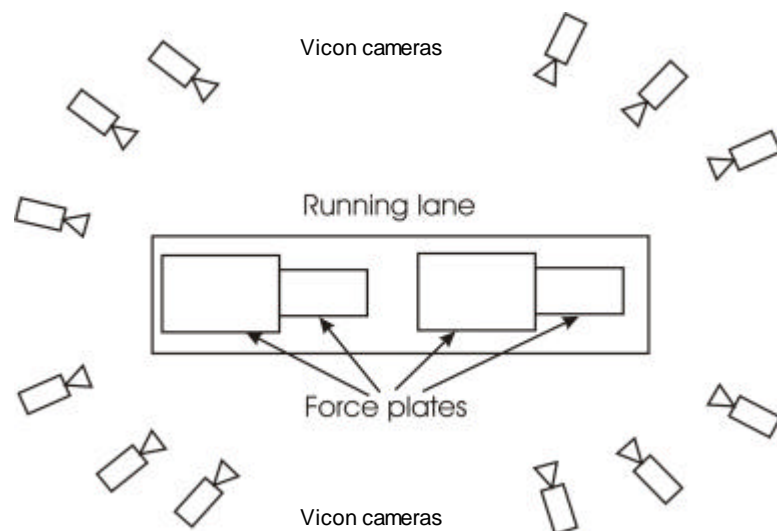


Figure 1: Top view of the experimental set-up: Four force plates in series were embedded in the floor. Twelve Vicon cameras arranged around of the force plates captured the athlete's motion.

The calculation of the leg kinematics and kinetics was done by means of a three segment rigid bodies model of the lower extremity (leg model, Fig. 2) Twenty eight reflective markers (\varnothing 15 mm) were attached to the left and right femur, shank and foot on predefined locations using double sided adhesive tape. Eight markers fixed on predefined anatomical landmarks were used to define the joint co-ordinate system. The anatomical landmarks used in this study were: malleolus medialis, malleolus lateralis, medial femoral condyle (most medial point) and lateral condyle (most lateral point). Further markers were placed at the caput metatarsale V (most lateral point, the markers were placed on the shoe), calcaneus (most posterior aspect of the shoe), calcaneus medial (most medial aspect of the shoe), calcaneus lateral (most lateral aspect of the shoe), shank (anterior aspect of the shank), medial tibial condyle, lateral tibial condyle, thigh proximal, thigh distal and trochanter major. Thus during the experiment 5 markers for each segment (thigh, shank and foot) were recorded. Sprinting leg motion (thigh, shank and foot) was described with reference to a neutral position (Fig.2). To define the neutral position, each leg of the subjects was placed in a global reference co-ordinate system (RCS). The RCS had axis 1 pointing forward, axis 2 in medio-lateral direction and axis 3 being vertical pointing towards the ground. The plane defined by axis 1 and 2 was parallel to the ground. The neutral position of the leg was defined as follows: the foot was flat on the ground its longitudinal axis parallel to the 1st axis of the RCS. The shank was orthogonal to the ground and the thigh vertical to the shank with no medio-lateral inclination. In this neutral position all markers were filmed to calculate their 3D-coordinates..

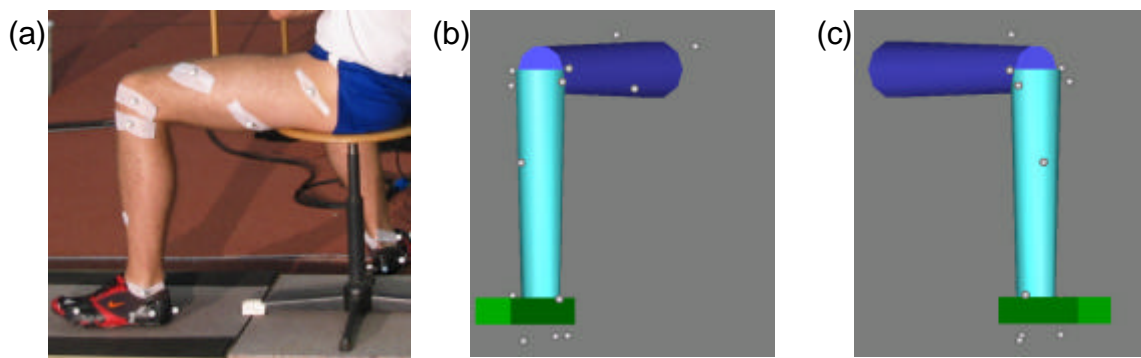


Figure 2: (a) Lateral view of the reference measurement (neutral position) to define the leg model. (b) lateral and (c) medial views of the model representing the three rigid bodies and the corresponding markers.

The leg model was realised using the simulation software “alaska” (advanced lagrangian solver in kinetic analysis, Chemnitz). The model included the ankle and knee joint (both modelled as ball-socket joints, 3 degrees of freedom). The connection between thigh and space was modelled as a free joint (6 degrees of freedom). Each joint was defined by two joint co-ordinate systems (JCS1 and JCS2) attached to each of the connected segments (JCS1 to the proximal and JCS2 to distal segment). The JCSs were defined in the neutral position by using the anatomical landmarks. For the ankle joint, the JCS1 and JCS2 in the neutral position results from moving the RCS to the midpoint of the line connecting both malleoli (origin of the ankle joint). Similarly for the knee joint the JCS1 and JCS2 in the neutral position results from moving the RCS to the midpoint of the line connecting both epicondyli (origin of the knee joint) but with a rotation of 90° of the JCS1 around its axis 2. Thus axis 3 of the JCSs cor-

responds to the longitudinal, axis 2 to the medio-lateral and axis 1 to the antero-posterior axis of the shank and thigh.

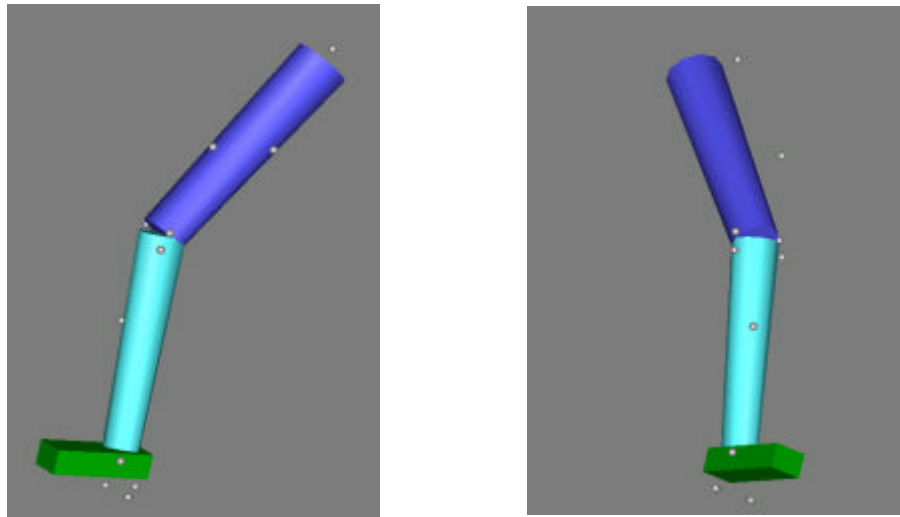


Figure 3: Lateral (left) and medial (right) views of the leg model with the markers used for tracking the motion whilst sprinting. The markers are attached to the model by means of 3D-spring damping elements.

All markers were fixed to the corresponding segments of the leg model, using their 3D-coordinates in the reference measurement (neutral position Fig. 2). The markers were attached to the model using 3D linear spring damping elements to account for the relative motion of the markers caused by movements and deformation of the soft tissues of the leg. Basing on earlier experience (Arampatzis et al., 2002; 2003) the spring and the damping constants were set to $k=10^6$ N/m and $\beta=10^4$ Ns/m respectively. The motion of the model during sprinting was obtained by tracking the 3D-coordinates of the markers (Fig. 3). Detailed information about this method can be found in Arampatzis et al., (2002; 2003). The kinematics of the ankle and knee joints are described by the orientation of JCS2 with regard to JCS1 using the Bryant angles (Arampatzis et al., 2002; 2003).

The resultant (external) joint moments at the ankle and knee joints were calculated in the JCS1. All calculations were done for the ground contact phase of sprinting.

4.3.2 Track configuration

In order to analyse the effect of different track compliances on sprint performance we used 3 different track configurations. A 60 m long single-lane running surface was constructed. Plywood boards (1.2 cm thick, 250 cm long, 60 cm wide) were placed onto a 60 × 0.6 m wooden chipboard frame serving as base support (Fig. 4).

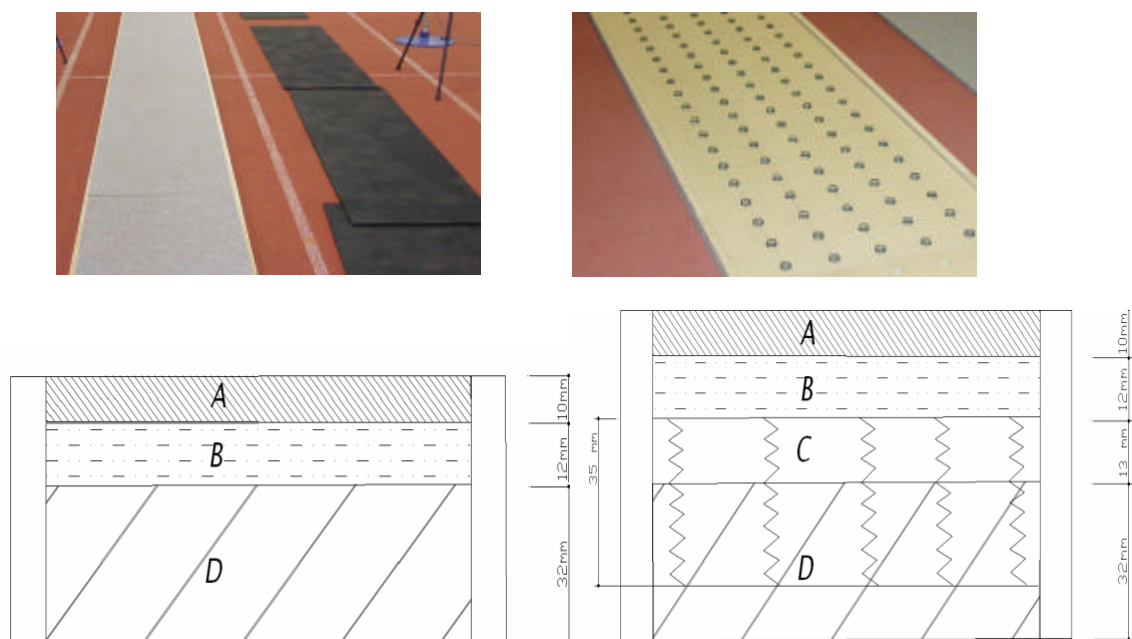


Figure 4: Photographs (top) and schematic drawings (bottom) of the track configurations, (left rubber, right spring)

A) rubber surface=10mm

B: plywood plate=12mm

C) springs=35mm

D) chipboard base plate=32mm

A frame with the same configuration as the running tracks was fixed on top of the chipboard, which was firmly fixed on top of the 4 force plates by means of screws. The stiffness of the track could be altered by changing either the rubber surface material (soft or hard) or by setting metal springs under the plywood surface boards in holes on the wooden chipboard frame explicitly prepared for that purpose.

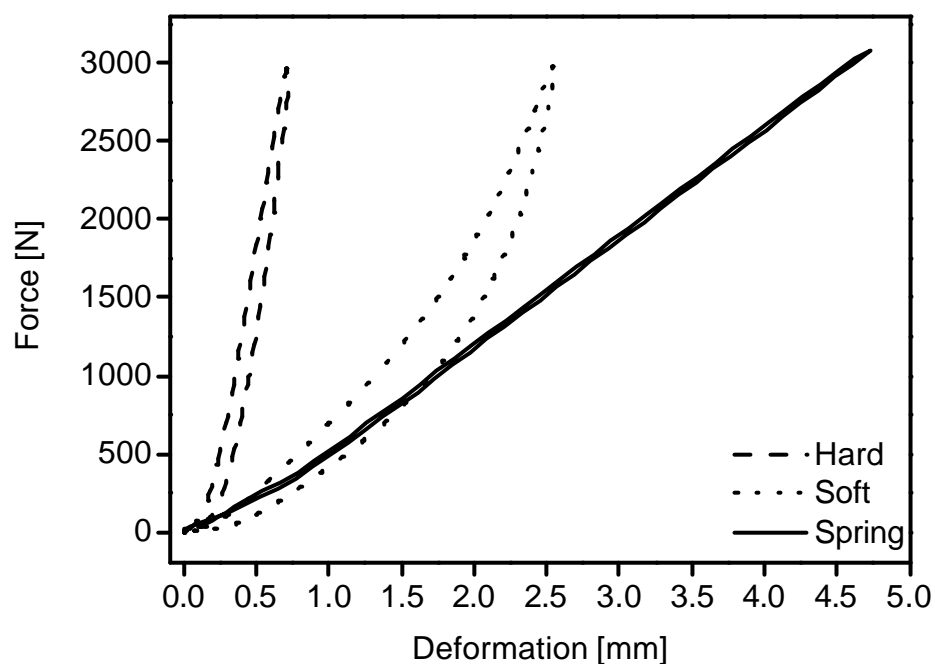


Figure 5: Force deformation plots of the two examined rubber surfaces (hard and soft) and the springs. The applied maximum load was 3kN using a 10cm diameter circular stamp. Hard: hard surface, Soft: soft surface, Spring: spring surface.

The spring surface consisted of linear springs combined with the hard rubber surface. The springs were 35 mm long with a diameter of 24 mm. The springs were arranged at every 10 cm to the anterior-posterior and medio-lateral direction in the wood chipboard frame. The force deformation relationship of the rubber surfaces and the springs were obtained by applying a load up to 3 kN using a Zwick compression device (Zwick-GmbH. Ulm,

Germany 100 Hz). The stiffness of the rubber surfaces and the springs were calculated by the slope of its force versus displacement relation between 2 kN and 3 kN (Fig. 5). The stiffness of the two rubber surfaces and the springs were 408.2, 191.5 and 236 kN/m for the hard, the soft and the spring surfaces respectively. The different track configurations were presented in a randomised order. The subjects were allowed to perform several submaximal speed trials to get used to each track. In order to avoid fatigue the subjects had only 2 maximal 60 m sprint trials on each track configuration. Furthermore, all subjects had a minimum of 10 min break interval between the runs. Double photocell sensors were positioned along the running lane to obtain the time intervals. The light barriers were placed at the start (1.5 m from the start line), at 30 and 60m. Near to the 60m, at the force plate region, two additional double photocell barriers were placed at 10m from each other to determine the velocity of the sprinters at 60 m.

4.3.3 Statistics

A one-way ANOVA for repeated measures was used to test the influence of the different tracks on the time intervals until 60m, the kinematic and kinetic characteristics. The ground reaction forces, the kinematic and kinetic characteristics were compared in intervals of 10 % of ground contact duration (average values every 10 % intervals). When a significant surface condition effect ($P < 0.05$) was present a post hoc test (Bonferroni) pairwise comparison was conducted in order to determine where these differences occurred. In the tables the results are presented as means and

standard deviation (SD), whereas at the figures the results are presented as means and standard error of mean (SEM).

4.4 Results

The time intervals from the examined sprinters showed no significant differences between the 3 track configurations ($P=0.81$ to $P=0.943$) (Table 1).

Table 1: 60m sprint time intervals onto the different track configurations (mean \pm SD, $n=10$).

| Distance | Hard | Soft | Spring |
|----------|-----------------|-----------------|-----------------|
| 0-30 m | 4.14 \pm 0.14 | 4.15 \pm 0.12 | 4.18 \pm 0.14 |
| 30-60m | 3.29 \pm 0.12 | 3.27 \pm 0.10 | 3.28 \pm 0.09 |
| 0-60 m | 7.42 \pm 0.24 | 7.42 \pm 0.21 | 7.46 \pm 0.22 |

[s]

Similarly, sprint velocity, step length, contact time, stride frequency and duty factor at the 60 m region were not affected by the different track configurations ($P=0.145$ to $P=0.908$) (Table 2). The three dimensional kinematics of the ankle and knee joints are presented in Fig. 6. Again the sprinters showed similar kinematic characteristics at 60m region onto all three track configurations (Fig. 6). The statistical comparisons of the discrete values every 10% of the contact phase showed no statistically significant ($P=0.081$ to $P=0.992$) track effect.

The kinetic analysis revealed that the ground reaction forces and the resultant ankle and knee joint moments were not affected by the track configurations.

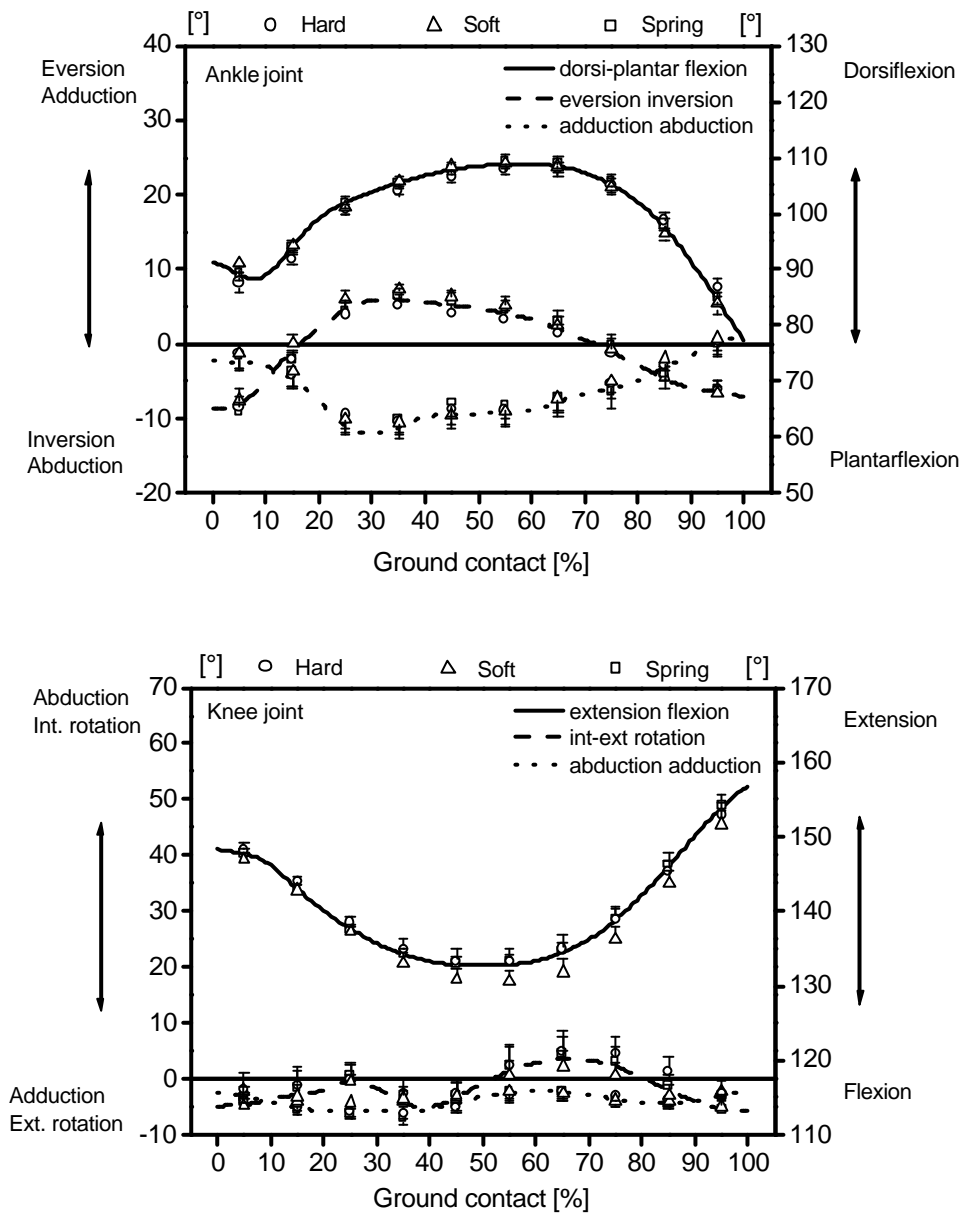


Figure 6: Foot eversion-inversion, dorsiflexion-plantarflexion and foot adduction-abduction angles of the ankle joint (top) and extension-flexion, shank internal-external rotation and shank abduction-adduction at the knee joint (bottom). The x-axis was normalised from 0 to 100 % of the ground contact phase (mean \pm SEM). Hard: hard surface, Soft: soft surface, Spring: spring surface.

For the ground reaction force parameters the horizontal and vertical impulses as well as the values every 10 % of the contact phase did not show any statistically significant differences among the different tracks ($P=0.15$ to $P=0.993$) (Table 3; Fig. 7).

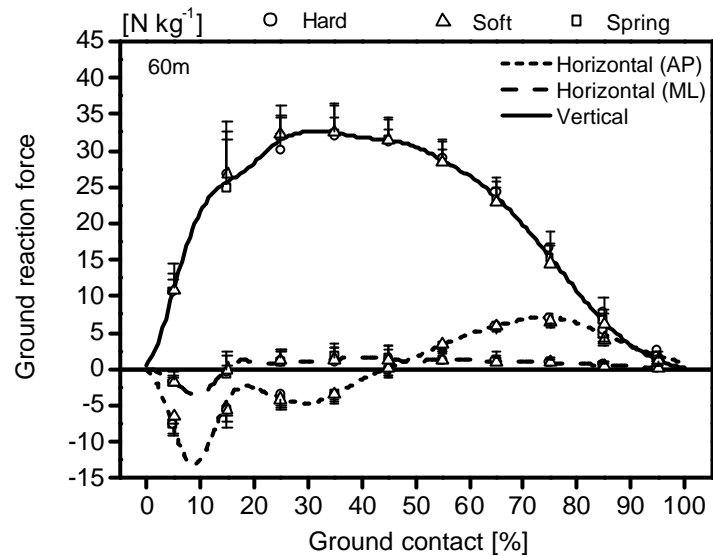


Figure 7: Average values of the ground reaction forces (AP: antero-posterior; ML: medio-lateral; vertical) whilst sprinting at 9.38 ± 0.45 m/s velocity. The x-axis was normalised from 0% to 100% of the ground contact phase (mean \pm SEM). Hard: hard surface, Soft: soft surface, Spring: spring surface.

Figure 8 illustrates the external resultant ankle and knee joint moments in all three axes during the contact phase whilst sprinting. Same as for the kinematic characteristics the statistical comparisons of the discrete values every 10 % of the contact phase showed no significant track effect ($P=0.083$ to $P=0.997$).

Table 2: Ground contact time, flight time, step length, stride frequency, duty factor, and velocity at the 60 m region whilst sprinting on the different track configurations (mean \pm SD)

| Track | Contact time [ms] | Flight time [ms] | Step length [m] | Stride frequency [Hz] | Duty factor [%] | Velocity [m/s] |
|--------|-------------------|------------------|-----------------|-----------------------|-----------------|-----------------|
| Hard | 115 \pm 6 | 126 \pm 15 | 2.17 \pm 0.11 | 2.1 \pm 0.1 | 23.9 \pm 1.1 | 9.23 \pm 0.65 |
| Soft | 117 \pm 6 | 123 \pm 7 | 2.18 \pm 0.16 | 2.1 \pm 0.1 | 24.5 \pm 2 | 9.54 \pm 0.34 |
| Spring | 112 \pm 9 | 131 \pm 11 | 2.23 \pm 0.19 | 2.1 \pm 0.1 | 23.1 \pm 1.8 | 9.37 \pm 0.37 |

4.5 Discussion

We hypothesised a track compliance-dependent behaviour of sprint performance (time interval of 60m) whilst sprinting, with better values on the more compliant of the examined tracks (i.e. spring track). This hypothesis was based on literature reports stating that running speed is enhanced on tracks having an intermediate compliance as compared to running on a rigid surface (McMahon and Greene 1979) and that human jumping height and energy rates are greater on a soft sprung surface when compared to a hard one (Arampatzis et al., 2004b). However sprint performance was unaffected by the tested track compliances since the time interval for the 60m sprinting onto the different tracks showed similar values (no statistically significant track effect, $P=0.57$).

The results of the present work are difficult to compare to other studies because we didn't find any investigation experimentally examining the influence of track compliance on sprint performance. Earlier studies (Kerdok

et al., 2002) examining the influence of surfaces with different compliance on submaximal running (i.e. 3.7 m/s) reported that a more compliant surface increases the energy conversion by the surface and cause a reduction in metabolic cost of running.

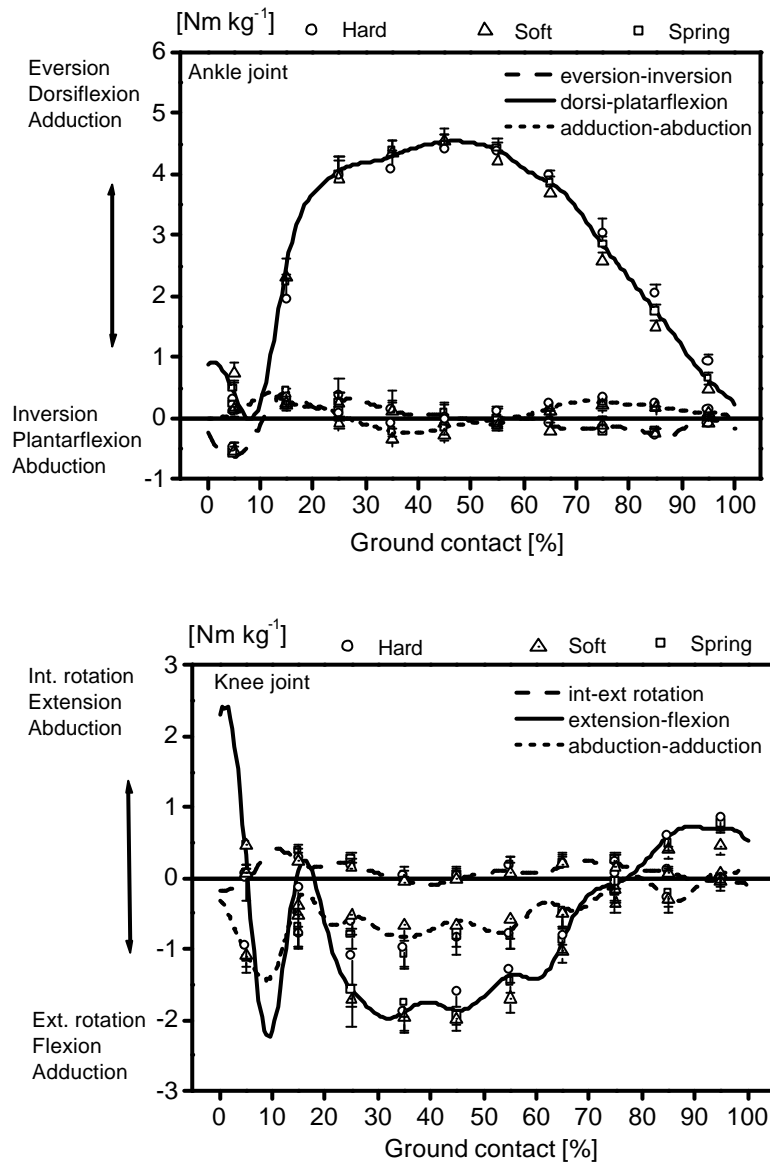


Figure 8: External eversion-inversion, dorsi-plantar flexion and adduction-abduction ankle joint moments (top) and internal-external, extension-flexion and abduction-adduction knee joint moments (bottom). The x-axis was normalised from 0% to 100% of the ground contact phase (mean \pm SEM). Hard: hard surface, Soft: soft surface, Spring: spring surface.

Furthermore it is well known that during submaximal walking, running and hopping on different compliant surfaces the human system adjusts its leg mechanics to maintain a stable centre of mass dynamics (Ferris et al., 1997; 1998; 1999; Moritz et al., 2004, Moritz and Farley 2005; Marigold and Patla, 2005; MacLellan and Patla 2006). Such adjustments in leg mechanics may happen proactively in a feedforward manner (Farley et al., 1998; Hardin et al., 2004; Marigold and Patla 2005; Moriz and Farley 2005) as well as reactively in a feed back loop (Ferris et al., 1999; MacLellan and Patla 2006). Therefore we expected an improvement in sprint performance in a track compliance-dependent manner (a) due to a higher energy storage and return from the track during the contact phase and (b) due to a suitable adjustment of the sprinters leg mechanics leading to a more advantageous stretch-shortening cycle of the lower extremity muscle-tendon units. An improvement in sprint performance could not be confirmed.

Table 3: Normalised impulses in antero-posterior and vertical direction at the 60m region whilst sprinting onto the different track configurations (mean \pm SD).

| | Hard | Soft | Spring |
|-------------------------|------------------|------------------|------------------|
| 60m impulse (posterior) | -0.25 \pm 0.05 | -0.23 \pm 0.04 | -0.22 \pm 0.07 |
| impulse (anterior) | 0.26 \pm 0.03 | 0.25 \pm 0.03 | 0.25 \pm 0.06 |
| impulse (vertical) | 2.38 \pm 0.11 | 2.42 \pm 0.11 | 2.34 \pm 0.16 |

[Ns/kg]

There are several possible reasons explaining the lack of influence of track compliance on sprint performance in the present study. We did not find any statistically significant differences in the kinematic and kinetic characteristics of the lower extremities during sprinting on the three tracks. These findings show that the sprinters run using the same strategy without any specific adjustment in the leg mechanics onto all examined tracks. Although to our knowledge, there is no study in the literature reporting three dimensional kinetic and kinematic characteristics during sprint, the values in the sagittal plane are comparable to the data from two dimensional analyses (Stefanyshyn and Nigg 1998; Hunter et al 2004; 2005). The similar kinematics at the ankle and knee joints whilst sprinting onto the three differently compliant tracks suggest that also the contraction velocity of the triceps surae and the monoarticular vasti muscle-tendon units would be similar (the length changes of the triceps surae and the vasti muscle-tendon units depend on the ankle and knee joint angles). Furthermore the maximal vertical ground reaction force whilst sprinting on the three differently compliant tracks was about 2600 N (Fig. 7). This means that the maximal expected deformation of the examined tracks was 0.63 mm, 2.4 mm and 4.02 mm for the hard soft and spring surfaces respectively (Fig. 5). These deformation values are relatively low in comparison to the studies done on walking (~35 mm, Marigold and Patla, 2005), running (~30 mm, Kerdok et al., 2002; Ferris et al., 1999) and hopping (26 to 146 mm, Farley et al., 1998; Ferris and Farley 1997; Mritz and Farley 2005). Therefore it is possible that in the present study, the differences in track compliance were not large enough to cause clearly diverging ad-

justments in leg mechanics. Nevertheless the differences in track deformation at a given force between the hard surface and the spring surface were more than the 6 fold from the hard surface (Fig. 5) and we feel that this difference is high enough to have a practical relevance for sprinting tracks. Additionally the similar values of the ankle and knee joint moments whilst running onto the three different tracks suggest that the energy exchange in the series elastic elements of the ankle and knee extensors as well as the work of their corresponding contractile elements would be also similar (Hof et al., 2002). Therefore it can be concluded that the function of the triceps surae and the vasti muscle tendon units remained unaffected when sprinting onto the examined tracks.

Another cause for the lack of effect of surface compliance on sprint performance might be that the energy exchange between sprinters and tracks during the contact phase was too small to have a clear effect on sprint time. The maximal expected energy recovery by 2600 N maximal ground reaction force (see Fig. 5) for the hard, soft and spring tracks during the contact phase whilst sprinting is about 0.6, 2.4 and 5.5 Joules respectively. The reported energy recovery values for the Achilles tendon during the contact phase of submaximal running (~4.0 m/s, Hof et al., 2002) or hopping (Lichtwark and Wilson 2005) are between 33 and 38 Joule. This means that the magnitude of the energy storage and return in the tracks are very small compared to the energy exchange within the tendon and aponeurosis of the human musculoskeletal system, suggesting that its effect on the sprint performance would be only marginal.

In conclusion surfaces of different compliances (deformation at 2600 N ranging from 0.63 mm to 4.02 mm) did not have any clear effect on sprint performance (time intervals until 60m). The possible explanations for this finding are: (a) Due to the short track deformation the athletes did not show any specific adjustment in their leg mechanics (similar kinematic and kinetic characteristics at the ankle and knee joints) in a track compliance-dependent manner and (b) the benefit of the sprinters due to a higher energy storage and return of the more compliant tracks was marginal because the expected magnitude of the energy storage in the track is too small relative to that of the involved connective tissues.

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5 MAIN FINDINGS AND CONCLUSION

The influence of the mechanical and morphological properties of the muscle tendon unit (MTU) of the lower extremities and of the running track compliance on the sprint performance has been the subject of research in this thesis. On the basis of the available literature the determination of the mechanical properties of the tendon and aponeurosis of the lower extremities provided contrasting results. Therefore it was first attempted to clarify a methodological problem by the estimation of the mechanical properties of the VL tendon and VL tendon and aponeurosis. The findings showed that the strains of the human VL tendon, VL tendon plus aponeurosis and VL aponeurosis, as estimated *in vivo* by two dimensional ultrasound during a maximal isometric contraction, do not differ from each other. Moreover the estimated displacement at the cross point in the VL muscle belly is significantly greater compared to that at the VL myotendinous junction. In consequence, to obtain the elongation of the whole tendon it is necessary to multiply the strain calculated for the examined part of the tendon by the total length of the tendon.

Furthermore it was attempted to clarify the influence of the mechanical properties of the leg extensors MTU on the sprint performance. The main result shows a significant negative ($p < 0.05$) correlation of the VL tendon and aponeurosis elongation on the sprint time (100 m). This has a consequence that due to the elongation of the tendon and aponeurosis energy can be saved and returned during the entire contact phase and furthermore as a cause of the muscle interaction, the mechanical properties of

the contractile element can be altered and thus leading to an higher force production capability due to the force velocity relationship. On the other hand the muscle mechanical and morphological properties (ankle and knee joint moments, pennation angle, fascicle length, thickness and ratio between fascicle length and segment length) did not clarify the sprint performance (time at 100 m) differences between the fast and the slow sprinter group. Therefore it can be suggested that both groups had similar working ranges of the muscles and furthermore it can be assumed that the other synergistic muscles have similar fascicle lengths and that the morphological properties of the triceps surae and quadriceps muscles cannot be responsible for the difference between a homogenous sprinter group.

In the completion of this study the influence of the tracks with different compliance in the sprint performance (time interval at 60 m) was examined. The results suggest that the track with small compliance at given force (2600 N) can not affect the sprint performance. This is caused by the small deformation of the tracks (0.63 to 4.02mm; from hard to most compliant track), accordingly the kinematics of the athletes did not show any difference between the different track configurations which means that the human system is not provoked enough to change his leg mechanics. Furthermore the maximum energy return of the most compliant track used (at 2600 N) was ~5-5 J whereas at the human system the energy exchanges of the Achilles tendon ranges from 33 to 38 J (walking and running respectively). This means, that the energy contribution of the different tracks to the human system is small to have a substantial influence to the sprint performance.

In summary it can be concluded that the sprinting performance is related to the mechanical properties of the lower extremities muscle tendon units in a higher proportion than to the mechanical properties of running tracks. The reason is that the deformation of the series elastic elements is greater than the tracks. As a consequence the energy storage and return can benefit to a higher degree as well as the effectiveness of the stretch shortening cycle of the sprinters.

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7 CURICULUM VITAE

Sawas Stafilidis was born in Thessaloniki, Greece, in 1971. From 1977 to 1989 he finished basic school while in 1989/1990 he began studying sport sciences at the Aristoteleion University of Thessaloniki in the Department of Physical Education. Stafilidis then graduated in 1994 with a diploma in physical education. After this he moved to Germany where in 1999 he had the opportunity to start his doctoral thesis at the Institute for Biomechanics in Cologne. In 2003 he became a scientific co-worker at the Institute of Biomechanics and Orthopaedics (Cologne). Stafilidis's basic research interests are in the fields of the human performance and muscle mechanics.

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